

## ESMAC Abstract 2025

### The relationship between walking speed and lower limb kinematics in typically developing children across age groups in the MENA region

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#### Introduction

Walking speed significantly influences lower limb kinematics is a key determinant of kinematic changes in typically developing children's (TDC) walking gait (Stansfield et al., 2001). While many studies have investigated the relationship between walking speed and lower limb kinematics during walking gait, there remains room for improvement. For example, previous studies often combined wide age ranges into a single group or focused on only one plane of motion (Fukuchi et al., 2019; Schwartz et al., 2008). In addition, the influence of age on this relationship during walking has not yet been investigated specifically among children under 14 years of age. This is especially relevant in the Middle East and North Africa (MENA) region, where TDC-specific gait datasets remain limited.

#### Research Question

What is the relationship between normalized walking speed and key lower limb kinematics during walking gait in TDC, and how does this relationship vary across age groups?

#### Methods

A cross-sectional experimental study was conducted involving 18 TDC (12 males, 6 females), aged 5–13 years, who were equally divided into three age groups (mean  $\pm$  SD): Group A (6.50  $\pm$  0.55 years), Group B (9.17  $\pm$  0.75 years), and Group C (12.11  $\pm$  0.98 years). Participants underwent three-dimensional (3D) gait analysis using a 14-camera Vicon motion capture system (Oxford, UK) with

full-body marker placement following the Conventional Gait Model 2.3. Each child walked barefoot at three self-selected speeds: slow, normal, and fast. Eleven lower limb kinematic 3D parameters across the sagittal, frontal, and transverse planes were averaged over three trials per condition. Pearson correlation coefficients and coefficient of determination ( $R^2$ ) were used to quantify the relationship between normalized walking speed and 3D kinematic variables ( $p < .05$ ). Walking speed was normalized using Hof's (1996) equation, which accounts for each child's leg length.

#### Results

Significant correlations were found between normalized walking speed and hip flexion at initial heel contact for Group A ( $r = 0.742$ ,  $p < .001$ ) and Group B ( $r = 0.895$ ,  $p < .001$ ). For Group C, normalized walking speed significantly correlated with ankle plantarflexion at toe-off ( $r = -0.672$ ,  $p = .002$ ). Across all participants combined, hip flexion at initial contact was consistently correlated with normalized walking speed ( $r = 0.641$ ,  $p < .001$ ).

#### Discussion

Walking speed significantly influences 3D lower limb kinematics in TDC during walking, particularly hip flexion at initial heel contact. This study identified significant correlations between walking speed and 3D lower limb kinematics across three planes of motion and under different walking speeds. The findings suggest a nearly linear relationship between specific kinematic variables and walking speed, which could be beneficial for clinical gait analysis by emphasizing the importance of considering both walking speed and age-related effects in assessment and intervention planning.

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### Three-dimensional gait analysis of typically developing children in the MENA region

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### Introduction

Normative gait data for typically developing children (TDC) is essential for identifying gait abnormalities and understanding age-related changes. Region-specific datasets are needed due to reported variations in lower limb gait parameters across countries (Ito et al., 2022; Lythgo et al., 2011; Smith et al., 2016). Currently, no normative gait database exists for TDC in the Middle East and North Africa (MENA) region. The Gait Deviation Index (GDI), which is a measure of gait quality (where 100 indicates no gait deviation), varies with age and body mass index in children (Schwartz & Rozumalski, 2008). However, the potential influence of walking speed on the GDI score has not yet been investigated.

### Research Question

Do gait parameters in TDC from the MENA region differ from other regions, and does walking speed affect the GDI scores?

### Methods

A cross-sectional experimental study was conducted in Doha, Qatar, involving 18 TDC (12 males, 6 females) aged 5–13 years, who were equally divided into three age groups (mean  $\pm$  SD): Group A (6.50  $\pm$  0.55 years), Group B (9.17  $\pm$  0.75 years), and Group C (12.11  $\pm$  0.98 years). Three-dimensional (3D) gait analysis was conducted using a 14-camera Vicon motion capture system and the Conventional Gait Model 2.3. Participants completed barefoot walking trials at three self-selected speeds: slow, normal, and fast. Spatio-temporal parameters, 3D lower limb kinematics, and GDI scores were averaged across three trials per speed. One-way ANOVA and effect size ( $\eta^2$ ) assessed differences across age groups and speeds and the effect of speed on the GDI ( $p < .05$ ).

### Results

Non-normalized spatio-temporal parameters differed significantly across age groups in cadence, stride length, step length, and

step width ( $p < .05$ ), particularly between Group A and the older groups. Cadence and step width remained significantly different after normalization ( $p < .05$ ). Walking speeds varied significantly across conditions (slow: 0.91  $\pm$  0.14 m/s; normal: 1.18  $\pm$  0.15 m/s; fast: 1.56  $\pm$  0.17 m/s;  $p < .001$ ). Speed significantly affected cadence, stride length, and step length ( $p < .001$ ), but not step width ( $p = 1.000$ ). 3D kinematic differences between Group A and the older groups were observed in hip rotation, hip abduction/adduction, and foot progression angle (significance:  $p < .05$ ). Speed significantly influenced the GDI scores ( $\eta^2 = 0.35$ ,  $p = .003$ ), with lowest scores at the slow walking speed (slow: 89.59  $\pm$  1.79; normal: 93.08  $\pm$  1.87; fast: 95.09  $\pm$  1.93;  $p = .001$ ).

### Discussion

Normalized and non-normalized spatio-temporal parameters and 3D lower limb kinematics were significantly influenced by age and speed. Several findings aligned with data from other countries. However, Group A (6.50  $\pm$  0.55 years) consistently differed from the older groups, suggesting distinct early-childhood gait characteristics. Walking speed significantly impacted the GDI scores, emphasizing its importance in clinical gait assessments of TDC.

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### The role of high-demand tasks in 3D gait analysis to enhance the functional assessment of Total Knee Arthroplasty designs

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## Introduction to clinical case

Total Knee Arthroplasty (TKA) is a well-established treatment for end-stage knee osteoarthritis. The bicruciate-retaining (BCR) and bicruciate-stabilized (BCS) designs represent two approaches to restore knee function [1,2]. The BCR design preserves both the anterior (ACL) and posterior cruciate ligament, with the goal of improving joint stability, proprioception, and natural knee kinematics. The ACL, and its proprioceptive function, is sacrificed in the BCS design. So far, it remains unclear whether different types of designs results in different functional outcomes. This case study aims to determine whether high-demand functional are more sensitive to postoperative biomechanical differences than overground walking.

## Methods

Two male patients undergoing TKA, one receiving a BCR prosthesis and the other a BCS prosthesis, were assessed using three-dimensional gait analysis (Vicon®) both preoperatively and one year postoperatively. Kinematic and kinetic waveforms were collected. In addition to standard overground walking, participants performed a series of high-demand functional tasks: walking with a side step, walking with a crossover turn, stair ascent, stair descent, uphill walking and downhill walking on a treadmill (Motek®) [3].

## Results

Both patients showed similar sagittal knee kinematics pre- and postoperative in all tasks. The BCR patient tended more towards extension and less loading response postoperative. In the frontal plane the BCR patient showed a more neutral knee alignment postoperative over all overground tasks. A similar trend was seen during the treadmill tasks. No changes were observed in the BCS patient, with the exception of stair ascent, where a more neutral alignment was noted postoperative. In the transverse plane (Fig.1), the BCS patient demonstrated a slight decrease in range of motion (ROM) during overground walking postoperative, with a more prominent reduction during side stepping, crossover turning, and stair descent. A decrease in internal rotation during the whole gait cycle was noted during stair ascent. The BCR patient showed improved rotational stability (less ROM) during the crossover task. Less internal rotation was observed during stair ascent and descent postoperative. During uphill walking, the BCR patient showed a slight improvement in transverse plane ROM. These pattern was similarly observed during downhill walking.

Fig. 1

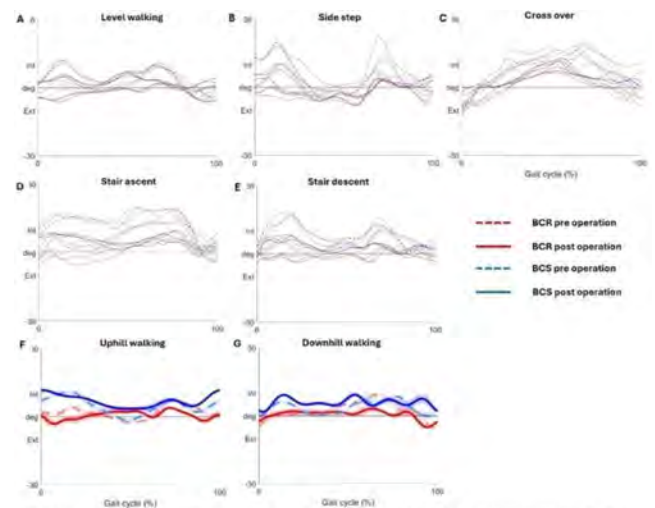


**FIGURE 1** All graphs present the knee kinematics in the transverse plane of the BCR and BCS patient pre and one year postoperative. Graph A, B, C, D and E show three trials of each patient for the overground tasks. Graphs F and G show the average and  $\pm 1$  one standard deviation of eight consecutive steps of uphill and downhill walking on a treadmill. Abbreviations: BCR, bicruciate-retaining; BCS, bicruciate-stabilized.

## Conclusion and clinical relevance

This case study highlights the clinical value of incorporating high-demand functional tasks into 3DGA to assess TKA designs. These activities, such as walking with side step, crossover turn and stair descent, revealed more prominent changes postoperative in rotational stability than during level walking. Such findings suggest that high-demand tasks provide deeper insights into joint stability and prosthetic behavior during real-life movements. Including these tasks in postoperative 3DGA may enhance clinical decision-making and better reflect the functional capabilities of different TKA designs.

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## Understanding Normative Gait Spatio-Temporal variability: A Multi-Factor Analysis of Age, Gender, and BMI

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### Introduction

Assessing joint range of motion and related parameters reveals insights into gait dynamics influenced by age, gender, and BMI. Despite global research, Indonesia lacks normative gait data. This study captures gait patterns of healthy individuals from Indonesian to establish baseline spatio-temporal and kinematic data, supporting population-specific analysis and future clinical applications in assessing gait variability across demographic factors.

### Research Question

This study aims to establish baseline gait data from healthy individuals and assess how age, gender, and BMI influence gait variability to support population-specific analysis and clinical use in Indonesia.

### Methods

This cross-sectional study was conducted from June to September 2024 at Telkom University Surabaya's Motion Laboratory from 158 adult participants. Using Vicon motion capture and 16 reflective markers, the research recorded spatio-temporal and kinematic gait data. Anthropometric data were collected, and motion was analyzed using Plug-in Gait protocols. Data were processed in Nexus where joint kinematics (hip, knee, ankle) and spatiotemporal variables were analyzed using Python and ANOVA to evaluate demographic effects across the gait cycle.

### Results

The ANOVA results revealed that walking speed had the most significant effect on joint kinematics across the gait cycle, influencing hip, knee, and ankle movements throughout all phases. Gender and BMI demonstrated localized effects, with gender significantly affecting knee flexion-extension ( $p = 0.043$ ) and pelvic tilt ( $p = 0.008$ ) at specific time intervals during mid-stance and terminal swing phase, while BMI influenced ankle dorsiflexion and knee angles which showed significant effects on ankle dorsiflexion ( $p = 0.024$ ) and knee movement ( $p = 0.041$ ), with angular deviations reaching  $5-8^\circ$  in higher BMI groups. Age showed minimal influence on joint range of motion ( $p > 0.1$ ), suggesting limited variability across age groups within the healthy population. The hip joint kinematics show statistically insignificant effects for gender, age, or BMI (all  $p$ -values  $> 0.05$ ) across the gait cycle. While minor variations were observed in hip flexion-extension patterns, particularly during terminal stance and swing, they did not reach

significance thresholds to indicate demographic influence. These findings emphasize the importance of considering walking speed as a dominant factor in gait variability and highlight the nuanced contributions of gender and BMI in shaping joint kinematics for population-tailored clinical gait assessments.

### Discussion

This study found that walking speed strongly influences lower limb kinematics across the gait cycle, especially affecting the pelvis, hip, knee, and ankle. It was the most dominant factor in gait variability, while gender and BMI had more localized effects, mainly in the knee, ankle, and pelvis. Age showed minimal impact, likely due to the participants' limited and healthy age range. These results highlight the need to normalize walking speed and consider demographic factors in gait analysis, supporting the development of population-specific data for better clinical and rehabilitation outcomes in Indonesia.

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## Introduction

A strict stretching methodology, which was defined earlier in the literature, was previously applied to reduce the maximum isometric voluntary muscle force (MIVMF) for different lower extremity muscles, such as hip adductors, flexors, and plantar flexors in healthy individuals (1-4). This approach is well understood as both safe and reliable in a healthy population. Yet, there is a notable absence of research examining the use of this protocol in children with spastic cerebral palsy (SCP).

## Research Question

Is the artificial muscle weakening methodology effective in developing a temporary MIVMF reduction for children with SCP?

## Methods

13 children with SCP (GMFCS Level I-II) were included in the study. 4 children (2 females, 2 bilateral, 12.20 ± 4.25 years) were included for the plantar flexor (PF), 4 children (3 females, 2 bilateral, 12.75 ± 4.65 years) for the rectus femoris (RF) primed knee extensor, and 3 children (1 female, 3 bilateral, 17.3 ± 3.29 years) for bilateral hip adductor (HAdd) muscles stretching concept. The stretching protocol is performed for the most affected extremity's PF and RF muscles, lasting 135 seconds, with 13 repetitions and 5-second resting intervals, at the level of 7/10 discomfort. For the hip adductors, the stretching was applied bilaterally. 3D movement analysis system (VICON, Vantage 5 cameras, 5MP@420FPS) and 2 force plates (AMTI) were used before (BS) and after (AS) stretching to determine the altered gait biomechanics. Targeted muscle's MIVMF was measured with a hand-held dynamometer (Lafayette, US) in BS and AS (3,5,6). Paired t-test was used for statistical comparison.

## Results

The MIVMF decreased by 15.59% (from 133.91N ± 59.89N to 113.04N ± 46.35N) for the knee extensors, 38.51% (from 89.72N ± 18.90N to 55.17N ± 17.55N) for the PFs, and 49.30% (from 143.5N ± 1.41N to 72.75N ± 45.04N) for the HAdds. No significant difference was observed between walking speeds (p=0.353 in the RFs, p=0.790 in the PFs, p=0.071 in HAdds). The relevant parameters were given in Table 1.

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## Artificially induced temporary muscle weakening methodology to determine the related gait alterations for cerebral palsy children: a long-term concept

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Table 1: Statistical comparison of selected gait parameters before and after stretching of the plantar flexors, rectus femoris, and bilateral hip adductors.

|                                                       | BS     |       | p       | AS     |       |
|-------------------------------------------------------|--------|-------|---------|--------|-------|
|                                                       | mean   | SD    |         | mean   | SD    |
| PF Muscle Stretching Methodology                      |        |       |         |        |       |
| P1 - Peak knee flexion (°)                            | 58.01  | 10.16 | 0.007*  | 53.4   | 11.43 |
| P2 - Range of knee flexion from toe-off to peak (°)   | 34.65  | 9.77  | 0.178   | 31.48  | 11.45 |
| P3 - Total knee flexion-extension range (°)           | 51.95  | 10.43 | 0.072   | 48.46  | 12.68 |
| P4 - Time difference peak to toe-off (%)              | 7.6    | 1.8   | 0.225   | 8.04   | 1.87  |
| Peak hip flexion (°)                                  | 43.78  | 13.09 | 0.005*  | 40.42  | 12.33 |
| Hip flexion-extension range (°)                       | 46.23  | 6.15  | 0.005*  | 40.79  | 8.09  |
| Peak ankle DF (°)                                     | 13.3   | 7.6   | 0.003*  | 18.59  | 4.48  |
| Peak ankle PF (°)                                     | -16.43 | 11.31 | 0.011*  | -13.61 | 10.34 |
| Mean ankle DF-PF (°)                                  | 0.45   | 9.33  | 0.015*  | 3.02   | 4.28  |
| Ankle DF-PF range (°)                                 | 29.73  | 12.41 | 0.115   | 32.3   | 11.09 |
| Peak ankle PF moment (N°mm/kg)                        | 1.05   | 0.19  | 0.018*  | 0.83   | 0.39  |
| Mean ankle DF-PF moment (N°mm/kg)                     | 0.36   | 0.14  | 0.016*  | 0.3    | 0.19  |
| Peak ankle power (Watt/kg)                            | 2.21   | 0.9   | 0.119   | 1.23   | 0.5   |
| Mean ankle power (Watt/kg)                            | 0.06   | 0.06  | 0.011*  | -0.04  | 0.06  |
| Rectus Femoris Muscle Stretching Methodology          |        |       |         |        |       |
| P1 - Peak knee flexion (°)                            | 40.03  | 11.5  | 0.002*  | 45.94  | 9.1   |
| P2 - Range of knee flexion from toe-off to peak (°)   | 20.59  | 6.3   | 0.33    | 24.23  | 10.9  |
| P3 - Total knee flexion-extension range (°)           | 39.05  | 8.9   | 0.048*  | 42.08  | 8     |
| P4 - Time difference peak to toe-off (%)              | 12.48  | 5.7   | 0.063   | 15.63  | 6.4   |
| Peak anterior pelvic tilt (°)                         | 27.02  | 5.7   | 0.051   | 26.67  | 4.9   |
| Peak posterior pelvic tilt (°)                        | 18.04  | 3.6   | 0.222   | 18.16  | 4.3   |
| Mean pelvic tilt (°)                                  | 23.52  | 4.9   | 0.279   | 23.03  | 4.6   |
| Mean knee flexion-extension (°)                       | 16.97  | 6.2   | <0.001* | 21.13  | 4.7   |
| Bilateral Hip Adductor Muscles Stretching Methodology |        |       |         |        |       |
| P1 - Peak knee flexion (°)                            | 50.96  | 6.34  | 0.189   | 48.41  | 11.53 |
| P2 - Range of knee flexion from toe-off to peak (°)   | 7.96   | 7.31  | 0.84    | 7.56   | 3.44  |
| P3 - Total knee flexion-extension range (°)           | 29.17  | 10.67 | 0.154   | 24.32  | 8.57  |
| P4 - Time difference peak to toe-off (%)              | 9.42   | 1.79  | 0.395   | 8.97   | 6.73  |
| Peak hip adduction (°)                                | 2.89   | 2.38  | 0.012*  | -0.59  | 1.65  |
| Peak hip abduction (°)                                | -5.21  | 2.62  | 0.021*  | -9.82  | 2.32  |
| Hip abduction-adduction range (°)                     | 8.1    | 2.56  | 0.315   | 9.22   | 1.65  |
| Mean hip abduction-adduction (°)                      | -0.89  | 2.45  | 0.014*  | -5.02  | 2.31  |
| Mean hip flexion-extension (°)                        | 19.44  | 10.77 | 0.011*  | 14.61  | 8.06  |

\*p<0.05, BS: before stretching, AS: after stretching, SD: standard deviation, DF: dorsiflexion, PF: plantar flexion, P1-2-3-4: Goldberg et. al.'s (11) stiff-knee gait parameters

## Discussion

The stretching methodology effectively reduced MVIMF in all three targeted muscle groups and allowed observation of muscle weakness-related gait alterations. For the RF, MVIMF reduction was associated with increased peak knee flexion and sagittal plane knee range. In the case of PF weakness, reduced PF moment and power were observed, contributing to stiff-knee gait in children with cerebral palsy (7–10). Additionally, decreased mean hip flexion and increased peak hip abduction suggest that adductor muscles may play a more significant role in hip flexion during walking than hip adduction. This non-invasive muscle weakening method allows assessment of temporary force loss and may be enhanced with EMG or ultrasonography to evaluate muscle activity. This protocol may offer early insight into potential gait changes following clinical interventions, such as botulinum toxin A injections or orthopedic surgeries.

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## Investigating the biomechanical importance of gracilis muscle weakness on gait kinematics for healthy individuals

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## Introduction

The gracilis muscle plays an important role in medial hip stabilization during gait and is often targeted surgically in children with cerebral palsy (CP) to prevent hip subluxation (1). Hip adductor weakness, especially in conditions like CP, has been associated with gait deviations and hip displacement (2). However, the specific biomechanical role of the gracilis during gait remains unclear, even in healthy individuals. This study therefore investigates the biomechanical effects of isolated gracilis weakness on gait in healthy individuals, providing baseline data on its contribution relative to other hip adductors.

## Research Question

What is the importance of the gracilis among the other hip adductor muscles during walking?

## Methods

Eight healthy individuals were divided into two groups: Group 1 (hip flexion stretch; N=3, 22.0±1.0y, 81.3±8.1kg, 1.76±0.01m); Group 2 (hip extension stretch targeting gracilis; N=5, 19.4±1.9y, 61.4±6.5kg, 1.62±0.06m). A passive stretching protocol (13 rounds: 135s stretch, 5s rest) was applied to Group 1 (hip flexion) and Group 2 (hip extension targeting gracilis)(3). Verbal confirmation was obtained every 15 seconds to prevent exceeding the individual discomfort threshold. Hip adduction strength (MIVMF) and hip abduction ROM were measured (hand-held dynamometer, goniometer). Pre/post-stretching 3D gait kinetics/kinematics were assessed (VICON motion analysis, 6 cameras, 2 force platforms, Plug-in Gait model)(4). Paired t-test was used for comparison (p<0.05). Significant differences in gait parameter changes between stretching protocols were considered to indicate gracilis weakness.

## Results

Following the protocol, significant hip adductor strength decreases (acute MIVMF reduction) were observed (Avg Group1: 249N to 185N; Group2: 355N to 280N). There was no significant difference in walking speed between the two groups (p=0.30). Analysis showed significant decreases for both groups in knee flexion/extension and ankle dorsi/plantarflexion range of motion (ROM). However, pelvic tilt, hip abduction/adduction ROM, and hip flexion/extension ROM showed significant increases only in Group 1 (Table 1).

Table 1: Summary of the kinematic data before (PRE) and after (POST) stretching intervention.

| PARAMETERS (°)   |                                | Knee Extension<br>PRE-Stretch<br>Mean ± SD | Knee Extension<br>POST-Stretch<br>Mean ± SD | Knee Flexion<br>PRE-Stretch<br>Mean ± SD | Knee Flexion<br>POST-Stretch<br>Mean ± SD |
|------------------|--------------------------------|--------------------------------------------|---------------------------------------------|------------------------------------------|-------------------------------------------|
| Pelvic obliquity | Mean                           | 0.08 ± 1.25                                | 0.05 ± 0.58                                 | 0.11 ± 0.81                              | 0.13 ± 0.59                               |
| Pelvic tilt      | Mean                           | <b>10.13 ± 1.31</b>                        | <b>12.94 ± 3.57</b>                         | 8.89 ± 1.74                              | 8.91 ± 1.11                               |
|                  | ROM                            | <b>2.95 ± 1.18</b>                         | <b>4.05 ± 1.58</b>                          | 2.99 ± 1.16                              | 2.83 ± 1.36                               |
| Pelvic rotation  | Mean                           | -0.01 ± 1.31                               | 0.26 ± 2.44                                 | 0.19 ± 1.87                              | 0.32 ± 0.64                               |
|                  | ROM                            | 15.47 ± 5.78                               | 17.53 ± 7.25                                | 9.35 ± 2.65                              | 9.65 ± 1.22                               |
| Hip abd-add      | Mean                           | -2.15 ± 2.89                               | -3.59 ± 2.30                                | -1.23 ± 1.05                             | -1.66 ± 0.94                              |
|                  | Max ADD                        | <b>6.63 ± 3.69</b>                         | <b>4.64 ± 2.83</b>                          | 3.83 ± 1.52                              | 3.76 ± 1.61                               |
|                  | ROM                            | <b>10.08 ± 2.03</b>                        | <b>17 ± 1.90</b>                            | 11.39 ± 2.51                             | 11.53 ± 2.27                              |
| Hip rotation     | Mean                           | -20.29 ± 9.38                              | -21.68 ± 8.52                               | 2.33 ± 7.86                              | 0.62 ± 7.52                               |
|                  | Maximum IR                     | <b>-12.10 ± 10.06</b>                      | <b>-14.45 ± 9.42</b>                        | 9 ± 7.74                                 | 7.43 ± 7.27                               |
|                  | ROM                            | <b>21.27 ± 8.27</b>                        | <b>18.38 ± 7.04</b>                         | 15.63 ± 1.62                             | 16.69 ± 4.58                              |
| Hip flex-ext     | Mean                           | 10.60 ± 3.21                               | 11.01 ± 3                                   | 13.29 ± 1.16                             | 13.10 ± 1.35                              |
|                  | Max hip flex (terminal stance) | 30.30 ± 3.25                               | 32.67 ± 8.82                                | <b>30.82 ± 3.16</b>                      | <b>29.53 ± 2.48</b>                       |
|                  | Max hip ext                    | -13.96 ± 3.35                              | -13.39 ± 2.25                               | <b>-9.99 ± 2.26</b>                      | <b>-8.37 ± 2.37</b>                       |
|                  | ROM                            | 44.27 ± 1.97                               | 46.06 ± 4.88                                | <b>40.81 ± 4.12</b>                      | <b>37.90 ± 3.36</b>                       |
| Knee Flex-Ext    | Mean                           | 17.16 ± 1.78                               | 15.53 ± 3.89                                | 19.36 ± 1.59                             | 18.84 ± 2.47                              |
|                  | Max flex                       | <b>59.85 ± 3.22</b>                        | <b>55.92 ± 3.64</b>                         | <b>61.91 ± 2.69</b>                      | <b>60.09 ± 2.95</b>                       |
|                  | ROM                            | <b>60.91 ± 3.95</b>                        | <b>58.52 ± 5.47</b>                         | 61.68 ± 3.01                             | 59.55 ± 4.82                              |
| Ankle DF-PF      | Mean                           | 4.26 ± 2.26                                | 2 ± 4.76                                    | 3.84 ± 2.65                              | 3.09 ± 1.98                               |
|                  | Max DF                         | 18.18 ± 2.21                               | 12.53 ± 6.36                                | 17.60 ± 1.78                             | 16.74 ± 1.79                              |
|                  | ROM                            | <b>34.99 ± 5.47</b>                        | <b>28.62 ± 5.81</b>                         | 31.22 ± 5.89                             | 20.30 ± 5.94                              |

Statistically significant parameters were given in bold (p<0.05). DF: dorsiflexion, ROM: range of motion, PF: plantar flexion, ABD: abduction, ADD: adduction, FLEX: flexion, EXT: extension, MAX: maximum, IR: internal rotation

## Discussion

The study showed that, different from contributing to hip abduction and hip rotational range, gracilis force is important in decreasing anterior pelvic tilt, tilt range, and in increasing knee and ankle range during walking for healthy individuals. As it is seen in hip flexor weakness, the gracilis may have the potential to contribute to reduced anterior pelvic tilt and sagittal plane pelvic range (5). However, different from other adductors, it may have little or no effect on contributing to hip flexion during walking. Although the findings might be different for children with CP. Therefore, as well

as the other hip adductor weakness, gracilis muscle weakness should be investigated not only for healthy individuals but also for children with cerebral palsy by a larger number of participants with kinetic and EMG assessments in the future.

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## Influence of hip adductor weakness on lower limb biomechanics during walking in adolescents with cerebral palsy: a pilot study

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## Introduction

Hip adductors are essential for medial hip stability during gait (1). In adolescents with spastic cerebral palsy (CP), spastic adductor tightness contributes to pathological gait and hip displacement risk (2). While orthopedic treatments often target these muscles (3),

resulting weakness is a concern. However, its biomechanical effects remain underexplored. This pilot study aimed to examine how hip adductor weakness affects gait parameters in adolescents with CP.

## Research Question

How does a temporary reduction in maximal isometric voluntary contraction force (MIVCF) of the bilateral hip adductor muscles affect gait biomechanics in adolescents with cerebral palsy?

## Methods

Three adolescents with CP (GMFCS Level I-III, 2 males, 3 diplegia,  $17.3 \pm 3.29$  years) and six healthy adolescents (HG) (1 male,  $18.8 \pm 1.77$  years) participated in the study. A stretching protocol ( $135 \text{ seconds} \times 13$  repetitions with 5-second rest intervals) was applied bilaterally to the hip adductor muscles to temporarily reduce MIVCF (4). The stretching intensity was symmetrically adjusted to 7/10 on the Visual Analog Scale. Three-dimensional gait analysis (VICON, UK; Plug-in Gait Model) and a hand-held dynamometer (Lafayette, USA) were used before and after the stretching protocol to assess gait parameters and MIVMCF, respectively (5). Statistical comparisons were performed using a paired t-test.

## Results

MIVCF of hip adductors decreased after stretching intervention (For CP: from  $127.14 \pm 23.14$  to  $67.22 \pm 27.18$ , 57.62%, HG: from  $235.62 \pm 60.27$  to  $179.73 \pm 43.08$ , 26.24%). There was no significant gait velocity difference between before stretching and after stretching conditions (For CP: BS:  $0.63 \pm 0.15$  AS:  $0.56 \pm 0.08$  p: 0.09, HG BS:  $1.35 \pm 0.11$  AS:  $1.34 \pm 0.1$  P: 0.68). All relevant lower extremity gait parameters were compared in Table 1.

**Table-1:** Statistical comparison of the interested gait parameters before and after the stretching protocol for both groups

| Joint Ang. (°)                            | Bilateral CP<br>BS (1)<br>Crt ± SE | Bilateral CP<br>AS (2)<br>Crt ± SE | P      | Bilateral HG<br>BS (3)<br>Crt ± SE | Bilateral HG<br>AS (4)<br>Crt ± SE | P      |
|-------------------------------------------|------------------------------------|------------------------------------|--------|------------------------------------|------------------------------------|--------|
| Mean Pelvic oblique                       | -0.57 ± 2.51                       | -0.36 ± 2.87                       | 0.665  | 0.13 ± 0.96                        | -0.08 ± 0.58                       | 0.471  |
| Pelvic Abduction - Adduction Range        | 9.04 ± 9.56                        | 9.38 ± 5.41                        | 0.927  | 14.79 ± 2.65                       | 14.47 ± 1.45                       | 0.653  |
| Pelvic Tilt Range                         | 8.85 ± 3.59                        | 8.52 ± 3.87                        | 0.522  | 7.48 ± 0.62                        | 4.42 ± 1.48                        | 0.005* |
| Pelvic Rotation Range                     | 23.74 ± 14.98                      | 24.33 ± 12.47                      | 0.720  | 16.69 ± 5.5                        | 14.65 ± 8.1                        | 0.523  |
| Mean Hip Abduction - Adduction            | -0.68 ± 2.45                       | 9.09 ± 2.58                        | 0.014* | -1.56 ± 1.91                       | -2.84 ± 0.14                       | 0.064* |
| Maximum Hip Abduction                     | 2.88 ± 2.37                        | -0.59 ± 1.68                       | 0.01*  | 7.40 ± 2.49                        | 4.62 ± 1.3                         | 0.43   |
| Maximum Hip Abduction                     | -3.2 ± 2.81                        | -9.82 ± 2.32                       | 0.14   | -10.71 ± 1.74                      | -11.35 ± 2.04                      | 0.05*  |
| Maximum Hip Internal Rotation             | 8 ± 15.35                          | -0.09 ± 20.39                      | 0.00*  | 13.89 ± 17.64                      | 12.48 ± 19.59                      | 0.007  |
| Hip Rotation Range                        | 25.76 ± 8.15                       | 20.20 ± 9.44                       | 0.013* | 26.8 ± 5.55                        | 23.29 ± 6.86                       | 0.050* |
| Mean Hip Flexion - Extension              | 19.44 ± 10.77                      | 14.61 ± 8.05                       | 0.011* | 14.41 ± 3.52                       | 16.58 ± 2.98                       | 0.799  |
| Maximum Hip Extension                     | 2.21 ± 12.71                       | -0.63 ± 9.62                       | 0.146  | -9.32 ± 2.42                       | -8.69 ± 2.51                       | 0.054  |
| Hip Flexion - Extension Range             | 32.63 ± 7.78                       | 31.02 ± 4.73                       | 0.354  | 47.77 ± 4.28                       | 48.8 ± 4.9                         | 0.811  |
| Mean Knee Flexion - Extension             | 34.38 ± 13.81                      | 16.16 ± 4.57                       | 0.081  | 34.39 ± 13.58                      | 32.72 ± 14.82                      | 0.496  |
| Maximum Knee Flexion                      | 62.48 ± 2.53                       | 59.7 ± 5.81                        | 0.169  | 50.98 ± 6.33                       | 48.41 ± 11.52                      | 0.047* |
| Maximum Knee Extension                    | -0.89 ± 2.91                       | -3.84 ± 4.89                       | 0.509  | 21.81 ± 16.14                      | 22.09 ± 15.44                      | 0.572  |
| Knee Flexion - Extension Range            | 83.37 ± 4.3                        | 63.54 ± 5.98                       | 0.154  | 29.19 ± 10.66                      | 24.32 ± 8.57                       | 0.182  |
| Ankle Dorsiflexion - Plantarflexion Range | 5.4 ± 5.31                         | 2.37 ± 8.45                        | 0.084  | 15.01 ± 6.27                       | 15.98 ± 6.43                       | 0.048* |

\*p<0.05, int: internal, rot: rotation, fl: flexion, ext: extension, abd: abduction, add: adduction max: maximum min: minimum

## Discussion

Adductor muscle weakness reduced hip rotation range and peak internal rotation in both adolescents with CP and healthy peers. However, in contrast to HG, adductor weakness in adolescents with CP also reduced mean hip flexion and increased peak hip abduction. These findings suggest that adductor muscles may play a more significant role in hip flexion during walking in CP patients than in healthy individuals. Adductor weakness is commonly seen with iliopsoas muscle weakness in CP patients, which may result in reduced hip flexion in the CP group. Inducing muscle weakness through the present stretching methodology caused a greater force reduction in CP than in healthy individuals, indicating that adductors in CP may respond more easily to stretching. (6) Additionally, the stretching protocol appeared to be effective and reliable in

reducing MIVMCF, serving as a potential model for evaluating botulinum toxin efficacy prior to injection in CP (7). The higher reduction in MIVMCF in CP compared to healthy peers may have influenced the kinematic results, warranting further investigation with a larger sample size.

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## Acute effects of insoles on trunk-pelvis and hip kinematics during jumping in athletes with flexible pes planus: A pilot study

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## Introduction

Vertical jumping is frequently used to monitor lower extremity strength by trainers (1,2). Flexible Pes Planus (FPP) is a common structural foot condition that is shown to reduce the jump performance and alter lower extremity kinematics by reducing sagittal plane pelvis and hip range in athletes (2,3). In addition to physical therapy, individuals with FPP often use insoles to improve foot posture. Although inlay insoles are frequently used in clinical practice, current literature lacks sufficient evidence regarding their effects on jump performance and upper body kinematics in athletes with FPP (4). The aim of this study was to examine the effect of inlay insoles on trunk-pelvis and hip kinematics during arms-free vertical jump for athletes with FPP.

## Research Question

Do insoles acutely affect trunk-pelvis and hip kinematics during jumping in athletes with flexible pes planus?

## Methods

Six male with FPP (age:  $17.50 \pm 1.25$  y.o., height:  $181 \pm 5.94$  cm., weight:  $79.66 \pm 10.20$  kg.) athletes participated in the study. The Foot Posture Index score (FPI-6) was measured barefoot for all participants to determine Pes Planus on standing. Jack's Test was also used to determine FPP (4,5,6). Upper and lower body kinematics were assessed before ([Pre Insole Vertical Jump], Prel-VJ) and after ([Post Insole Vertical Jump], Postl-VJ) the application of insoles. Data was collected using a 3D motion capture system (VICON Vantage, 6 cameras, 5MP@420FPS) synchronized with two AMTI force plates (7). For each participant, three trials were analyzed in two phases, take-off and landing. Paired sample t-tests were used for comparisons.

## Results

For all players, after the inlay insole application the foot posture was significantly improved ( $p < 0.05$ ). During jumping, thoracic angles measured relative to the laboratory coordinate system, as well as trunk angles calculated relative to the pelvis, did not show any significant differences between groups in any of the three anatomical planes. Similarly, no significant differences were found in pelvic or hip joint kinematics across the groups ( $p > 0.05$ ) (Table 1).

Table 1: The statistical comparison of all parameters of interest.

| Kinematic Variables (°)         | Take-Off Phase             |                             |                       | Landing phase              |                             |                       |
|---------------------------------|----------------------------|-----------------------------|-----------------------|----------------------------|-----------------------------|-----------------------|
|                                 | Pre-VJ<br>(mean $\pm$ std) | Post-VJ<br>(mean $\pm$ std) | p value<br>(Pre-post) | Pre-VJ<br>(mean $\pm$ std) | Post-VJ<br>(mean $\pm$ std) | P value<br>(Pre-post) |
| Max. Anterior Thoracic Tilt     | 55.22 $\pm$ 15.82          | 56.19 $\pm$ 14.58           | 0.780                 | 35.37 $\pm$ 14.88          | 34.74 $\pm$ 16.24           | 0.857                 |
| Max. Thoracic Elevation         | 2.67 $\pm$ 2.89            | 2.75 $\pm$ 2.88             | 0.897                 | 2.99 $\pm$ 3.75            | 2.52 $\pm$ 2.41             | 0.544                 |
| Max. Thoracic Internal Rotation | 3.20 $\pm$ 2.01            | 3.69 $\pm$ 2.83             | 0.343                 | 3.57 $\pm$ 3.07            | 3.55 $\pm$ 3.57             | 0.924                 |
| Max. Trunk Flexion              | 21.71 $\pm$ 10.99          | 21.55 $\pm$ 10.56           | 0.960                 | 9.99 $\pm$ 11.37           | 10.61 $\pm$ 12.61           | 0.881                 |
| Max. Trunk Lateral Flexion      | 3.50 $\pm$ 1.91            | 3.49 $\pm$ 1.47             | 0.987                 | 3.37 $\pm$ 5.11            | 2.43 $\pm$ 2.79             | 0.374                 |
| Max. Trunk Internal Rotation    | 3.27 $\pm$ 2.57            | 3.52 $\pm$ 2.65             | 0.662                 | 3.01 $\pm$ 2.29            | 3.01 $\pm$ 2.14             | 0.940                 |
| Max. Anterior Pelvic Tilt       | 37.69 $\pm$ 5.65           | 38.07 $\pm$ 5.67            | 0.794                 | 25.78 $\pm$ 4.87           | 26.67 $\pm$ 6.78            | 0.563                 |
| Max. Pelvic Elevation           | 2.60 $\pm$ 1.82            | 2.82 $\pm$ 2.28             | 0.545                 | 3.12 $\pm$ 2.70            | 2.84 $\pm$ 2.02             | 0.518                 |
| Max. Pelvic Internal Rotation   | 3.85 $\pm$ 2.17            | 2.84 $\pm$ 2.20             | 0.306                 | 5.89 $\pm$ 5.38            | 7.84 $\pm$ 7.65             | 0.069                 |
| Max. Hip Flexion                | 90.56 $\pm$ 9.41           | 90.78 $\pm$ 7.72            | 0.914                 | 76.40 $\pm$ 16.50          | 76.13 $\pm$ 15.03           | 0.943                 |
| Max. Hip Adduction              | -2.51 $\pm$ 2.95           | -2.79 $\pm$ 3.08            | 0.690                 | -3.08 $\pm$ 4.12           | -3.71 $\pm$ 3.12            | 0.463                 |
| Max. Hip Internal Rotation      | 7.31 $\pm$ 6.30            | 6.94 $\pm$ 6.33             | 0.806                 | 8.50 $\pm$ 9.17            | 6.52 $\pm$ 7.65             | 0.323                 |

Max.: Maximum.

## Discussion

The present study demonstrated that acute application of inlay insoles which improves foot position on standing may not significantly change trunk pelvis or hip kinematics during arm-free jumping. Athletes with flexible pes planus often present stiffer pelvis and hip during take off and impaired shock absorption, which may affect jump mechanics and increase the risk of overuse injuries (2,3). Inefficient energy transfer through the kinetic chain can potentially compromise both performance and neuromuscular control during dynamic tasks such as jumping (8). Future studies with larger sample sizes with controls are needed to better understand the kinematic adaptations associated with foot posture correction for acute and long time insole use in athletes with pes planus.

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## The effects of different shoulder angles on high plank exercise on upper extremity biomechanics

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## Introduction

The plank exercise is widely used to enhance core stabilization, aerobic capacity, and posture. Variations such as high plank, side plank, and forearm plank activate different muscle groups (1-4). However, improper upper extremity alignment may reduce the effectiveness of the exercise or increase injury risk by placing abnormal loads on joints. Arm position during planks can alter overall body mechanics, yet the effect of shoulder angle on biomechanical safety and efficiency has not been investigated in the literature (4-6). This study aims to examine how different shoulder angles in the high plank affect upper limb biomechanics and load distribution.

## Research Question

How does performing the plank exercise at different shoulder angles affect upper limb biomechanics?

## Methods

Ten healthy young adults (5 females, 5 males; age:  $22.8 \pm 1.22$  years) participated. Each performed four high plank variations: neutral (N) at  $90^\circ$  shoulder flexion,  $90^\circ$  flexion with  $30^\circ$  abduction (ABD), internal rotation (IR), and external rotation (ER), maintaining their position for a minimum of 10 seconds for a total of 3 times. Alterations in upper extremity biomechanics during plank were captured using a 3D motion capture system (VICON, Vantage 5, UK) with both hands on 2 different force plates (AMTI). 2 successful trials out of 3 were statistically analyzed with One-way ANOVA.

## Results

The statistical comparison of the relevant biomechanical upper extremity parameters is presented in Table-1.

Table 1: Statistical comparison of related upper extremity parameters

|                                              | N<br>mean $\pm$ SD          | ABD<br>mean $\pm$ SD       | ER<br>mean $\pm$ SD  | IR<br>mean $\pm$ SD | ANOVA   |
|----------------------------------------------|-----------------------------|----------------------------|----------------------|---------------------|---------|
| <b>Kinematics (Angle<math>^\circ</math>)</b> |                             |                            |                      |                     |         |
| <b>Shoulder Sagittal Plane</b>               |                             |                            |                      |                     |         |
| Minimum                                      | 57.69 $\pm$ 8.6 $\phi$      | 51.04 $\pm$ 5.8 $\phi$     | 59.60 $\pm$ 22.1     | 50.35 $\pm$ 41.0    | 0.000*  |
| Maximum                                      | 62.54 $\pm$ 8.8 $\phi$      | 56.06 $\pm$ 5.9 $\phi$     | 66.03 $\pm$ 21.9     | 56.55 $\pm$ 40.2    | 0.000*  |
| Range                                        | 4.84 $\pm$ 3.1              | 5.02 $\pm$ 4.0             | 6.34 $\pm$ 4.4       | 6.67 $\pm$ 3.3      | 0.039*  |
| Mean                                         | 59.8 $\pm$ 8.4 $\phi$       | 53.4 $\pm$ 5.5 $\phi$      | 62.8 $\pm$ 21.9      | 53.1 $\pm$ 40.8     | 0.000*  |
| <b>Shoulder Frontal Plane</b>                |                             |                            |                      |                     |         |
| Minimum                                      | 36.13 $\pm$ 22.7 $\phi$ +   | 61.49 $\pm$ 17.4 $\phi$    | 32.79 $\pm$ 33.2 +   | 55.32 $\pm$ 22.7    | 0.002*  |
| Maximum                                      | 44.83 $\pm$ 24.4 $\phi$ +   | 75.38 $\pm$ 19.2 $\phi$    | 45.85 $\pm$ 35.6 +   | 68.63 $\pm$ 22.4    | 0.000*  |
| Range                                        | 8.50 $\pm$ 6.3 +            | 13.50 $\pm$ 17.7           | 12.54 $\pm$ 12.3     | 13.81 $\pm$ 7.6     | 0.003*  |
| Mean                                         | 39.70 $\pm$ 19.7 $\phi$ +   | 65.77 $\pm$ 18.0 $\phi$    | 37.97 $\pm$ 38.0 +   | 62.76 $\pm$ 62.6    | 0.005*  |
| <b>Shoulder Transvers Plane</b>              |                             |                            |                      |                     |         |
| Minimum                                      | -19.97 $\pm$ 18.4 $\phi$    | -51.24 $\pm$ 18.3 $\phi$ ^ | -25.57 $\pm$ 37.1    | -30.31 $\pm$ 43.9   | 0.000*  |
| Maximum                                      | -11.70 $\pm$ 15.2 $\phi$    | -35.97 $\pm$ 16.8 $\phi$ ^ | -11.50 $\pm$ 34.9    | -22.22 $\pm$ 44.9   | 0.000*  |
| Range                                        | 8.27 $\pm$ 6.1 $\phi$ +     | 15.27 $\pm$ 17.7           | 15.07 $\pm$ 10.9     | 18.08 $\pm$ 12.7    | 0.004*  |
| Mean                                         | -14.86 $\pm$ 15.9 $\phi$    | -40.64 $\pm$ 16.4 $\phi$ ^ | -17.69 $\pm$ 35.6    | -9.77 $\pm$ 44.0    | 0.000*  |
| <b>Elbow Sagittal Plane</b>                  |                             |                            |                      |                     |         |
| Minimum                                      | 30.31 $\pm$ 8.6 +           | 30.09 $\pm$ 7.4 $\phi$ ^   | 25.33 $\pm$ 9.3      | 24.80 $\pm$ 6.7     | 0.001*  |
| Maximum                                      | 32.93 $\pm$ 9.2 +           | 34.62 $\pm$ 8.5 $\phi$ ^   | 28.07 $\pm$ 10.4     | 28.33 $\pm$ 6.8     | 0.000*  |
| Range                                        | 2.62 $\pm$ 2.2 +            | 3.62 $\pm$ 3.0             | 3.02 $\pm$ 3.0       | 3.93 $\pm$ 2.6      | 0.786   |
| Mean                                         | 31.41 $\pm$ 8.9 +           | 32.16 $\pm$ 8.0 $\phi$ ^   | 26.40 $\pm$ 9.6      | 26.48 $\pm$ 6.4     | 0.000*  |
| <b>Kinetics (N/men/kg)</b>                   |                             |                            |                      |                     |         |
| <b>Shoulder Sagittal Plane</b>               |                             |                            |                      |                     |         |
| Minimum                                      | 12.72 $\pm$ 13.6 $\phi$ *   | 19.61 $\pm$ 19.7 $\phi$ ^  | 0.83 $\pm$ 13.7 +    | 9.20 $\pm$ 19.9     | 0.046*  |
| Maximum                                      | 21.51 $\pm$ 12.6 $\phi$ +   | 30.79 $\pm$ 19.7 $\phi$    | 13.80 $\pm$ 21.3 +   | 41.40 $\pm$ 59.0    | <0.001* |
| Range                                        | 8.79 $\pm$ 5.5 +            | 11.18 $\pm$ 7.2 +          | 13.17 $\pm$ 17.4 +   | 32.20 $\pm$ 62.2    | <0.001* |
| Mean                                         | 17.41 $\pm$ 12.5 $\phi$ *   | 25.06 $\pm$ 19.1 $\phi$ ^  | 6.80 $\pm$ 14.1 +    | 19.21 $\pm$ 16.8    | 0.000*  |
| <b>Shoulder Frontal Plane</b>                |                             |                            |                      |                     |         |
| Minimum                                      | 16.89 $\pm$ 11.9 $\phi$     | 45.38 $\pm$ 9.3 $\phi$ ^   | 19.66 $\pm$ 10.2     | 19.64 $\pm$ 15.8    | 0.030*  |
| Maximum                                      | 28.59 $\pm$ 10.1 +          | 56.75 $\pm$ 10.4 $\phi$ ^  | 31.54 $\pm$ 14.7     | 39.67 $\pm$ 22.3    | <0.001* |
| Range                                        | 9.61 $\pm$ 8.2 $\phi$ +     | 11.34 $\pm$ 9.3            | 11.78 $\pm$ 16.3     | 19.89 $\pm$ 26.1    | <0.001* |
| Mean                                         | 23.86 $\pm$ 9.9 $\phi$ +    | 50.29 $\pm$ 8.6 $\phi$ ^   | 47.25 $\pm$ 37.8     | 28.51 $\pm$ 11.5    | <0.001* |
| <b>Shoulder Transvers Plane</b>              |                             |                            |                      |                     |         |
| Minimum                                      | -4.14 $\pm$ 2.0 $\phi$      | -8.56 $\pm$ 2.9 $\phi$ ^   | -4.34 $\pm$ 5.3      | -4.63 $\pm$ 2.3     | <0.001* |
| Maximum                                      | -3.71 $\pm$ 2.0 $\phi$ *    | -7.44 $\pm$ 2.6 $\phi$ ^   | -2.60 $\pm$ 1.3      | -1.97 $\pm$ 5.2     | <0.001* |
| Range                                        | 0.83 $\pm$ 0.7              | 1.07 $\pm$ 0.7             | 1.74 $\pm$ 6.4       | 2.71 $\pm$ 6.7      | <0.001* |
| Mean                                         | -3.73 $\pm$ 2.0 $\phi$ *    | -7.94 $\pm$ 2.7 $\phi$ ^   | -3.65 $\pm$ 1.3 +    | -3.62 $\pm$ 1.4     | <0.001* |
| <b>Elbow Sagittal Plane</b>                  |                             |                            |                      |                     |         |
| Minimum                                      |                             |                            |                      |                     |         |
| Maximum                                      | 9.64 $\pm$ 4.6 $\phi$ +     | -7.42 $\pm$ 4.4 $\phi$     | -10.49 $\pm$ 4.5 +   | -7.97 $\pm$ 4.7     | 0.020*  |
| Range                                        | 7.85 $\pm$ 4.4 $\phi$ +     | -5.59 $\pm$ 4.2            | -7.55 $\pm$ 7.4 +    | -2.80 $\pm$ 26.6    | <0.001* |
| Mean                                         | 1.86 $\pm$ 1.4 $\phi$       | 1.75 $\pm$ 0.8             | 3.16 $\pm$ 6.6       | 10.82 $\pm$ 27.6    | <0.001* |
| Minimum                                      | -8.65 $\pm$ 4.3 $\phi$ +    | -6.45 $\pm$ 4.2 $\phi$     | -9.19 $\pm$ 4.3 +    | -6.67 $\pm$ 4.3     | 0.015*  |
| <b>Kinetics (Watt/kg)</b>                    |                             |                            |                      |                     |         |
| <b>Shoulder</b>                              |                             |                            |                      |                     |         |
| Minimum                                      | 0.00345 $\pm$ 0.0 $\phi$ +  | -0.00333 $\pm$ 0.0         | -0.00591 $\pm$ 0.0   | -0.00758 $\pm$ 0.0  | 0.000*  |
| Maximum                                      | 0.003483 $\pm$ 0.0 $\phi$ + | 0.010270 $\pm$ 0.0 $\phi$  | 0.003932 $\pm$ 0.0 + | 0.007393 $\pm$ 0.0  | 0.000*  |
| Range                                        | 0.006035 $\pm$ 0.0 $\phi$ + | 0.016606 $\pm$ 0.0         | 0.009637 $\pm$ 0.0   | 0.014873 $\pm$ 0.0  | 0.000*  |
| Mean                                         | 5.81E-05 $\pm$ 0.0          | 0.006132 $\pm$ 0.0 $\phi$  | 2.12E-05 $\pm$ 0.0   | 5.45E-05 $\pm$ 0.0  | 0.000*  |
| <b>Elbow</b>                                 |                             |                            |                      |                     |         |
| Minimum                                      | -0.00135 $\pm$ 0.0          | -0.00097 $\pm$ 0.0 $\phi$  | -0.00201 $\pm$ 0.0   | -0.00523 $\pm$ 0.0  | 0.000*  |
| Maximum                                      | 0.001106 $\pm$ 0.0          | 0.000735 $\pm$ 0.0         | 0.001621 $\pm$ 0.0   | 0.003712 $\pm$ 0.0  | 0.000*  |
| Range                                        | 0.002454 $\pm$ 0.0          | 0.001703 $\pm$ 0.0 $\phi$  | 0.003626 $\pm$ 0.0   | 0.008933 $\pm$ 0.0  | 0.000*  |
| Mean                                         | 2.81E-06 $\pm$ 0.0          | 8.2E-07 $\pm$ 0.0          | 4.81E-06 $\pm$ 0.0   | 3.01E-06 $\pm$ 0.0  | 0.000*  |

Significant difference ( $p < 0.05$ ) between groups illustrated as  $\phi$  for N and ABD, \* for N and ER, + for N and IR, ^ for ABD and ER, ^ for ABD and IR, + for ER and IR. ( $p < 0.05$ ). SD, standard deviation; mm, millimeter; kg, kilogram; W, watt.

## Discussion

The findings indicated that performing planks in internal rotation and abduction significantly increased shoulder extension, adduction, and internal rotation moments, suggesting higher neuromuscular demand and potential injury risk. In contrast, external rotation was associated with improved biomechanical efficiency, reduced elbow flexion moment, and better postural alignment. Internal rotation also led to greater joint moment variability, potentially reducing upper extremity efficiency. Overall, frontal plane dynamics had a stronger impact on shoulder mechanics, while transverse plane changes more significantly affected the elbow. Future studies can incorporate electromyography or ultrasonography to assess muscle activation and joint loading more comprehensively. In conclusion, performing planks in the shoulder in neutral or external rotation appears more biomechanically advantageous for shoulder and elbow health, and these positions should be prioritized and further examined in future research.

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### More than just weight: the biomechanical impact of carrying sensitive objects during gait

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### Introduction

Biomechanical analysis is proposed as a tool to investigate physical and mental health during carrying loads in the workplace (1). Carrying load is a prevalent activity not only in workplace settings but also in daily living across all age groups. While much attention has been given to the physical implications of load-bearing, less is understood about how different types of weight impact gait biomechanics (2-4). Specifically, the difference in biomechanical responses when carrying sensitive weights, such as fragile or valuable items, compared to carrying typical loads, remains underexplored.

### Research Question

How does carrying sensitive weight affect gait biomechanics differently than carrying normal weight during walking?

### Methods

Fifteen healthy young adults (8 males, 7 females;  $21.33 \pm 0.48$  years) participated in the study. Gait was analyzed using a 3D motion capture system (VICON, UK; Plug-in Gait Full-Body Model) under three conditions: self-selected speed without weight (N), non-sensitive weight carrying (NSWC), and sensitive weight carrying (SWC). Equal weights of eggs (SWC) and flour (NSWC) were carried unilaterally during trials. Heart rate and oxygen saturation were measured via basic pulse-oximetry before and after each walk to assess autonomic responses, allowing normalization between trials. Upper extremity, lower extremity and trunk kinematic values, and temporospatial data of walking were used for analysis. Paired t-test was used for statistical comparison among conditions.

### Results

The mean ankle range of motion (ROM) was significantly higher in the SWC condition compared to NSWC ( $p=0.023$ ). Peak spine rotation was significantly reduced during NSWC ( $p=0.013$ ) compared to self-selected walking, with an even further decrease observed in the SWC condition ( $p<0.001$ ). Similarly, peak hip moment was lower in SWC compared to NSWC ( $p=0.026$ ). Additionally, both elbow and wrist ROM decreased during NSWC ( $p<0.001$ ), with more reductions in the SWC ( $p<0.001$ ) condition. Cadence ( $p=0.016$ ), step width ( $p=0.026$ ), double step length ( $p=0.012$ ), and walking speed ( $p<0.001$ ) were all significantly reduced in the SWC condition compared to NSWC. All interested parameters were given in Table-1.

Table 1: Statistical comparison of the interested gait parameters

| Interested gait parameters                        | N<br>Mean $\pm$ SD  | NSWC<br>Mean $\pm$ SD | SWC<br>Mean $\pm$ SD |
|---------------------------------------------------|---------------------|-----------------------|----------------------|
| Mean hip flexion-extension (°)                    | 12.12 $\pm$ 5.29    | 12.50 $\pm$ 5.28      | 11.85 $\pm$ 5.85     |
| Peak knee flexion (°)                             | 63.13 $\pm$ 4.0 *   | 64.04 $\pm$ 3.76 *    | 64.34 $\pm$ 4.52     |
| Mean knee flexion-extension (°)                   | 21.78 $\pm$ 2.48    | 21.98 $\pm$ 2.50      | 21.85 $\pm$ 2.84     |
| Mean ankle dorsi-plantar flexion (°)              | 3.13 $\pm$ 3.98     | 2.91 $\pm$ 4.08       | 3.21 $\pm$ 4.03 *    |
| Peak spine lateral flexion (°)                    | 6.50 $\pm$ 2.80 *   | 4.93 $\pm$ 2.81 *     | 4.87 $\pm$ 2.61      |
| Mean spine lateral flexion (°)                    | -0.89 $\pm$ 2.26 *  | -1.72 $\pm$ 2.20 *    | -1.72 $\pm$ 2.27     |
| Peak spine rotation (°)                           | 5.21 $\pm$ 2.77 *   | 4.48 $\pm$ 3.06 *     | 4.23 $\pm$ 3.08      |
| Mean elbow flexion-extension (°)                  | 38.21 $\pm$ 6.02 *  | 29.83 $\pm$ 5.58 *    | 29.34 $\pm$ 5.11     |
| Peak elbow flexion (°)                            | 53.14 $\pm$ 5.82 *  | 33.56 $\pm$ 5.06 *    | 32.80 $\pm$ 5.14     |
| Elbow flexion-extension range (°)                 | 26.07 $\pm$ 11.24 * | 6.50 $\pm$ 2.49 *     | 7.80 $\pm$ 7.69      |
| Wrist flexion-extension range (°)                 | 10.00 $\pm$ 6.28 *  | 6.04 $\pm$ 1.52 *     | 5.42 $\pm$ 2.15      |
| Peak hip flexion-extension moment (N*mm/kg)       | 0.88 $\pm$ 0.24     | 0.85 $\pm$ 0.18       | 0.80 $\pm$ 0.17 *    |
| Peak knee flexion-extension moment (N*mm/kg)      | 0.63 $\pm$ 0.23     | 0.68 $\pm$ 0.28       | 0.71 $\pm$ 0.33      |
| Peak ankle dorsi-plantar flexion moment (N*mm/kg) | 1.38 $\pm$ 0.40 *   | 1.44 $\pm$ 0.41       | 1.38 $\pm$ 0.42      |
| Peak hip power (W/kg)                             | 1.78 $\pm$ 0.34     | 1.81 $\pm$ 0.44       | 1.80 $\pm$ 0.43      |
| Peak knee power (W/kg)                            | 0.81 $\pm$ 0.44     | 0.80 $\pm$ 0.43       | 0.91 $\pm$ 0.65      |
| Peak ankle power (W/kg)                           | 4.40 $\pm$ 1.51 *   | 4.05 $\pm$ 1.50       | 4.53 $\pm$ 1.61      |

Significant difference between groups illustrated as \* for NSWC and SWC, \* for N and NSWC, and \* for N and SWC ( $p<0.05$ ). N: self-selected non-weight carrying gait. NSWC: non-sensitive weight carrying gait. SWC: sensitive weight-carrying gait. SD: standard deviation, mm: millimeter, kg: kilogram, W: watt

### Discussion

This study reveals significant biomechanical and physiological differences when carrying SWC versus NSWC. Increased ankle range of motion and reduced peak spine rotation in SWC suggest greater lower body strain and more cautious movement to stabilize the load. Similarly, lower peak hip moment and reduced upper body motion during SWC indicate altered body mechanics to prevent damage to the sensitive load. Additionally, walking parameters like cadence, step width, and speed were significantly reduced in the SWC condition, suggesting a slower, more deliberate gait. Overall, carrying sensitive weight appears to require more physical effort and

attention than carrying normal weight, with implications for load-carrying tasks involving fragile items.

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## Muscle synergies alterations during irregular ground walking in Parkinson's Disease with and without freezing of gait

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## Introduction

Walking requires coordinated muscle activity, often analyzed through muscle synergies(1). Parkinson's disease (PD) affects gait patterns and intermuscular coordination, especially in patients with freezing of gait (FOG)(2). Limited research exists on how muscle synergies adapt across different ground conditions. This study compared muscle synergies during walking on flat and irregular ground between healthy individuals and PD patients with and without FOG.

## Research Question

Do lower limb muscle synergies differ between flat and irregular ground walking in PD patients with and without FOG?

## Methods

Five PD patients without FOG (1 female; age  $62.6 \pm 8.68$  yrs.; UPDRS-III score  $11.0 \pm 4.06$ ; FOG-Q score  $0.8 \pm 0.84$ ;  $37.6 \pm 37.2$  months since diagnosis), five PD patients with FOG (3 females; age  $70.8 \pm 8.44$  yrs.; UPDRS-III score  $19.4 \pm 3.71$ ; FOG-Q score  $13.6 \pm 3.65$ ;  $85 \pm 48.6$  months since diagnosis) and five healthy subjects (HS) (1 female; age  $67.6 \pm 8.96$  yrs.) were recruited. Following ethical approval (ENM095/25260220251782025), participants underwent 3D kinematics and surface electromyography (sEMG) recordings while walking at self-selected speed on both flat and irregular ground with  $10^\circ$  ascending/descending slopes. sEMG was collected from eight lower limb muscles (GIMa:Gluteus maximus, GIMe:Gluteus medius, BiFe:Biceps Femoris, ReFe:Rectus Femoris, VaMe:Vastus Medialis, Sol:Soleus, GaMe:Gastrocnemius medialis, TiAn:Tibialis Anterior) of the most affected side in patients and dominant side in HS using the TrignoTM system, following SENIAM guidelines (3). Muscle synergies were extracted using the Synergy-LAB toolbox in Matlab2023b(4). Average envelopes were analyzed using Statistical Parameter Mapping (SPM), with one-way ANOVA followed by post hoc comparisons with significance threshold of  $p < 0.05$ .

## Results

Four synergies were required to reconstruct sEMG envelopes in all groups. ReFe muscle activity differed significantly between 30–40% of the gait cycle in the FOG group compared with HS ( $p < 0.001$ ). In early stance phase on irregular ground, the FOG group showed higher GIMa( $p = 0.004$ ) and GIMe( $p = 0.004$ ) activations than the PD group. During terminal stance–early swing on irregular ground, BiFe activity was higher in the FOG group compared with HS( $p = 0.028$ ), while ReFe activity was higher in the PD group compared with FOG group( $p = 0.001$ ) (Figure 1).

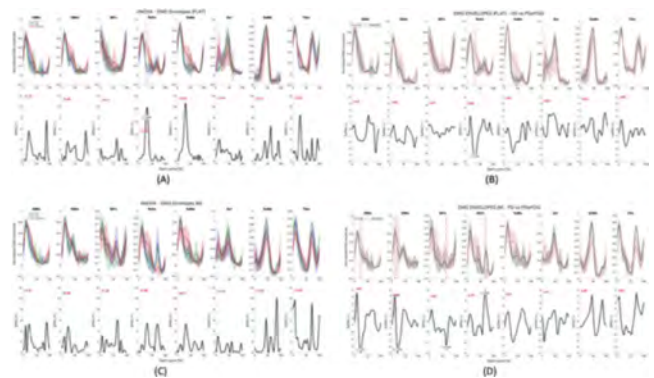


Figure 1. SPM results for EMG envelopes of muscle synergies during walking.

## Discussion

Previous research showed lower-limb oscillations in FOG progress from distal to proximal segments(5). When distal control diminishes, proximal muscles compensate to maintain balance, explaining the ReFe over-activation on flat ground. The heightened activation of GIMa, GIMe, and BiFe muscles in FOG patients on irregular ground suggests both adaptations to terrain changes and compensations for impaired distal control due to freezing. Additionally, the distinct lines of the irregular surface may have functioned as

visual cues, potentially affecting gait in the FOG group, consistent with literature showing transverse floor lines can alleviate freezing in PD(6). Findings are preliminary due to limited sample size.

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## Influence of surface type on postural sway and tibialis anterior muscle tone in quiet bipodal stance

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## Introduction

Postural control is essential for maintaining balance while standing (1), with surface properties playing a significant role in postural adjustments (2). While previous studies have mainly focused on postural sway (quantified by center of pressure (CoP) parameters) (3), the contribution of muscle tone to postural control remains less understood. This study investigates how surface type (flat versus soft) influences postural sway and tibialis anterior muscle tone during a bipodal stance.

## Research Question

How do flat and soft surfaces affect postural sway (ellipse area and CoP mean velocity), and how do tibialis anterior muscle tone parameters (frequency, stiffness, and elasticity) change in the initial and final phases of a 30-second stance in healthy individuals?

## Methods

The study included 22 healthy right-dominant participants (mean age:  $19.6 \pm 1.52$  years, 36.36% male). Tibialis anterior muscle tone parameters (frequency, stiffness, and elasticity) were measured using Myoton (Tallinn, Estonia) during the first and last 10 seconds of a 30-second bipodal stance task on two surfaces: flat and soft. Postural sway was quantified via ellipse area and CoP mean velocity using force plates (Kinvent Inc., Montpellier, France). Non-parametric (Wilcoxon signed-rank test) and parametric (t-test) tests were applied depending on data normality.

## Results

Statistical tests revealed a significant increase in ellipse area on soft ground compared to flat ground ( $p = 0.0077$ ). CoP mean velocity was significantly greater on soft ground ( $p < 0.0001$ ). Tibialis anterior frequency and stiffness were significantly higher on soft ground during both the first (frequency:  $p = 0.0406$ , stiffness:  $p = 0.0001$ ) and last 10 seconds (frequency:  $p = 0.0012$ , stiffness:  $p = 0.0003$ ). Elasticity showed a significant difference in the first 10 seconds ( $p = 0.0257$ ), but no difference in the last 10 seconds ( $p = 0.7752$ ).

## Discussion

Our findings confirm that postural sway is more pronounced on compliant soft surfaces, as indicated by increased ellipse area and CoP mean velocity. This suggests a greater challenge to the postural control system, requiring continuous adjustments to maintain stability (4). The concurrent increase in tibialis anterior frequency and stiffness likely reflects an upregulation of muscle tone to cope with this increased demand. These results support the idea that postural sway is not merely a mechanical outcome but closely linked to neuromuscular strategies that adapt to environmental conditions (5). Understanding these adaptations may inform balance training or rehabilitation approaches in populations with impaired postural control.

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### Balancing accuracy and wearability: IMU configuration effects on human activity recognition in patients undergoing total hip arthroplasty

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### Introduction

Total hip arthroplasty (THA) is a common and successful intervention performed to alleviate pain and restore function in individuals with end-stage hip osteoarthritis [1]. Despite significant functional improvement, patients' function does not reach asymptomatic control levels [1] and clinicians need objective means to better understand patient function during their daily living activities (ADL). Inertial measurement units (IMUs) combined with machine learning methods enable human activity recognition (HAR) for patients during daily life [2]. To enhance patient compliance, the minimal IMU set-up for accurate task detection should be identified and used. Identifying the best set-up relies on the preference of assessing either activity quality or quantity.

### Research Question

What is the impact of IMU configuration on the accuracy of HAR in patients undergoing THA?

### Methods

Participants included 20 patients measured pre-surgery, 3-months post-surgery (n = 11) and 1-year post-surgery (n = 6), as well as ten healthy subjects measured once. Participants were asked to perform a 10-min semi-standardized tour in the hospital vicinity (gait on flat ground, up/down ramp, turns and stairs ascent/descent). They were equipped with 8 IMUs (Physilog 6, MindMaze)

and video-recorded providing gold-standard for identifying the activities performed. The IMU positions are presented in Figure 1. The three-dimensional acceleration and angular velocity signals (128 Hz) were fed to a bi-directional Gated Recurrent Unit (GRU) model to classify six activities: gait, walking up and down ramps, stairs ascent and descent and static sitting. The signals were filtered using a butterworth filter (zero-phase, cut-off:3Hz), scaled (StandardScaler, Python) and segmented into 300 ms windows before being input to the GRU. Model performance was evaluated through a 10-fold group cross validation on the training data (split ratio 70:30), and F1-score, precision and recall were used as evaluation metrics.

### Results

The F1-scores for tested configurations were: 1)8-IMUs:0.86, 2)5-IMUs:0.84, 3)3-IMUs:0.79, 4)2-IMUs:0.69, 5)Sacrum IMU:0.63, and 6)Trunk IMU:0.49.

### Discussion

Our results using the 8-IMU and 5-IMU configurations are consistent with findings in the literature [2]. Overall, the decrease in F1-score from 8 IMUs to 1 IMU showed a significant negative impact in activity classification performance (Figure 1). Precision remained relatively stable (>0.8) as the number of IMUs decreased, indicating that 80% of the predicted activities were correctly identified. However, recall dropped notably with less IMUs, suggesting that the model failed to detect a substantial percentage of activities. This implies that the model is suitable for assessing activity quality (e.g., gait speed) due to high precision in correct activity identification, however it is less reliable in assessing activity quantity (e.g., total gait time), since it missed a notable amount of activity occurrences. Among the tested combinations, the 3-IMU configuration offers a good balance between classification accuracy and wearability for HAR. Further model performance improvement is possible through fine-tuning, potentially enabling HAR in THA patients during ADL.

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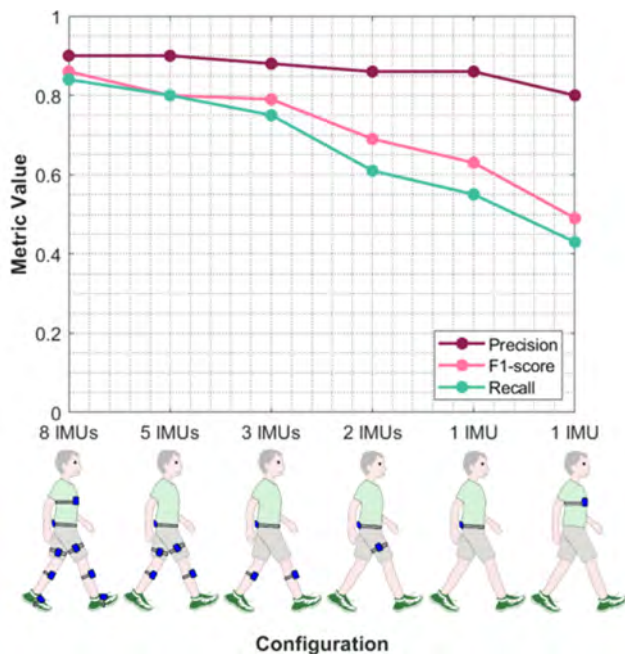


Figure 1 - Precision, F1-score and recall values vs. IMU configuration

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## Joint Kinematic and Kinetic Variability in Children with Periventricular and Intraventricular Brain Lesions

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## Introduction

Cerebral palsy (CP) is an umbrella term for injury to the developing brain occurring pre-, peri-, or early postnatally, and is the leading cause of motor developmental disability in childhood, with the highest incidence in infants born before 28 weeks of gestation [1]. CP in preterm infants may originate from distinct neuropathological processes, most commonly periventricular leukomalacia (PVL) or intraventricular hemorrhage (IVH), including its severe form, periventricular hemorrhagic infarction (PVHI) [2]. It was shown that the extent of brain lesions strongly correlate with gait pathology [3], however, further research is needed to better understand the precise relationship between brain injury and gait patterns.

## Research Question

The aim of the current study is to assess differences in gait pattern in children with PVL and those with IVH.

## Methods

We included 20 children in this retrospective analysis (PVL: n=10, IVH: n=10). Kinematic (Vicon, 200 Hz) and kinetic data (AMTI, 1000 Hz) during gait were recorded and lower extremity kinematics and kinetics were analyzed. Statistical parameter mapping [4] t-test was used to compare kinematic and kinetic waveforms ( $\alpha=0.05$ ). In addition, variability in kinematic and parameters between groups was compared using F-tests across initial contact, terminal stance, and swing phases (swing phase not considered for joint kinetics).

## Results

At the time of the gait patients in the PVL and IVH group were on average  $7.7 \pm 2.1$  years and  $7.8 \pm 1.2$  years old ( $p > 0.05$ ), respectively. No differences in gait waveforms for lower limb kinematic or kinetics were found. Significant differences in variance were observed between groups for hip adduction during initial contact ( $p=0.007$ ,  $F=0.14$ ) and for the foot progression angle during both initial contact ( $p=0.005$ ,  $F=7.95$ ) and terminal stance ( $p<0.001$ ,  $F=31.54$ ), with PVL showing higher variance in foot progression angles and IVH exhibiting higher variance in hip adduction at initial contact. Significant group differences in variance were found for the vertical ground reaction force during initial contact ( $p=0.017$ ,  $F=5.60$ ), sagittal knee moment during initial contact ( $p=0.025$ ,  $F=4.99$ ), knee power during initial contact ( $p<0.001$ ,  $F=16.70$ ), and ankle power during initial contact ( $p<0.001$ ,  $F=15.61$ ).

## Discussion

Although not all differences reached statistical significance, PVL participants consistently exhibited greater movement variability across gait parameters. Specifically, higher variance was observed in 23 kinematic and 19 kinetic measures for the PVL group, compared to 7 and 5, respectively, for the IVH group. These findings suggest that individuals with PVL exhibit less consistent motor control during gait, likely due to the more diffuse brain injury characteristic of PVL, in contrast to the more localised lesions typical of IVH. While the sample size in this study is relatively small, the observed trends offer valuable insights into how distinct types of brain injury may differentially impact gait stability and motor coordination.

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## Exploring Muscle Coordination in response to Irregular Terrains: Implications for Gait Adaptation in Healthy Older Adults

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### Introduction

Gait relies on the coordination between the muscular and the Central Nervous System (CNS) (1). The muscle synergy theory proposes that the CNS activates coordinated muscle groups (motor modules) instead of sending individual signals to each muscle (2). Motor control complexity stems from the CNS's ability to efficiently group muscles, optimizing communication. During walking on flat terrain, the neuromuscular system organizes into four modules: weight-acceptance, propulsion, foot-clearance and swing (3,4).

### Research Question

How is muscle coordination related to changes in joint angles when comparing gait of healthy older adults (HOA) over flat and irregular terrain?

### Methods

Twenty HOA (aged  $67.6 \pm 6.85$ ) were recruited from the LAMBE-COM group (Universidad Rey Juan Carlos, Madrid, Spain) and provided informed consent. All procedures followed the Declaration of Helsinki and were approved by the university's ethics committee (record:ENM095/25260220251782025).

Participants completed four trials of 6-meter walking on flat and irregular terrain with consecutive ascending and descending slopes (10°). Data were recorded using 17 Inertial Measurement Units (Xsens Technologies BV) and sEMG with Wireless Trigno™ System (Delsys Inc.) from eight muscles on the dominant side, following SENIAM guidelines(5): Gluteus-Maximus (GlMa), Gluteus-Medius (GlMe), Biceps-Femoris (BiFe), Rectus-Femoris (ReFe), Vastus-Medialis (VaMe), Soleus (Sol), Gastrocnemius-Medialis (GaMe) and Tibialis-Anterior (TiAn). sEMG signals were analyzed with the SynergyLAB toolbox in Matlab2023b (4), using Non-Negative-Matrix-Factorization (NNMF) to obtain muscle synergies. Statistical-Parameter-Mapping (SPM)(6) was then applied to compare activation coefficients (C), average envelopes and joint angles between both tasks ( $p < 0.05$  for statistical significance).

### Results

Although the Variability-Accounted-For (VAF) was lower on irregular terrain, four motor modules were still effective (Fig.1).

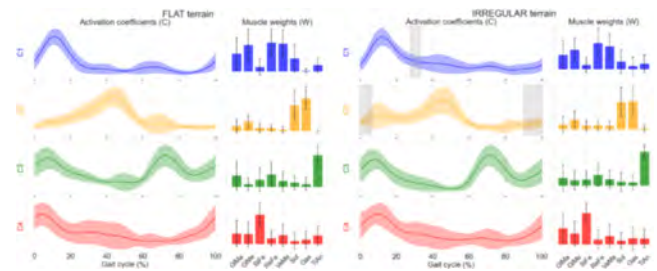


Figure 1. Muscle synergies for flat (left) and irregular terrain (right), displaying activation coefficients (C) and muscle weights (W). Grey areas indicate significant SPM differences.

Muscle weights remained similar to flat reference. Temporal activation increased in early mid-stance (C1). In C2 the rise occurred from terminal-swing to initial-contact, where ReFe and VaMe also showed lower activation. BiFe decreased around ~90% of gait cycle (GC). TiAn increased at heel-strike. Joint angles confirmed greater knee and hip flexion (70-40%GC) and more dorsiflexion in pre-swing.

### Discussion

According to the literature (7), irregular terrain with slopes elicits the same four modules and muscle weights as flat, but with increased temporal activation. Specifically, in weight-acceptance (C1), likely for postural control to maintain balance—achieved through a more "crouched" and stable posture—via the decrease in ReFe and VaMe (higher knee flexion), and BiFe (higher hip flexion); and propulsion (C2, for allowing obstacle clearance) synergies. Related to C2, we found greater activity in pre-swing TiAn with increased dorsiflexion, potentially facilitating ankle adaptation to slopes and enabling improved foot clearance to overcome terrain irregularities.

Our findings suggest the CNS tends to manage significant kinematic changes by reusing established muscle patterns with temporal adjustments (8), leading to greater variability, complexity and lower VAF.

### Acknowledgements

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### Comparative case studies of lower limb and trunk kinematics in AIS patients following spinal fusion at different vertebral levels

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### Introduction to clinical case

Adolescent idiopathic scoliosis (AIS) is a three-dimensional deformity of the spine in which patients typically maintain normal functionality [1]. However, it is still unclear if this functionality is preserved after spinal fusion, thus affecting their quality of life [2]. In this report, three case studies investigated the impact of spinal fusion, performed at various vertebral levels, on full-body kinematics during gait and trunk anatomical movements.

### Methods

Three AIS patients (Pt-1: thoracic scoliosis apex at T7, 16Y; Pt-2: thoracolumbar scoliosis apex at T12, 30Y and Pt-3: double thoracic and thoracolumbar scoliosis apex at T8 and L1, 15Y, with Cobb angle = 48°, 31°, 58° respectively) had T4-T12, T10-L3 and T4-L4 spinal fusion respectively. They all underwent full body biplanar X-ray, at baseline and 2-year post-operatively to calculate pelvic incidence (PI), lumbar lordosis (L1S1), thoracic kyphosis (TK: T1T12), sagittal vertical axis (SVA), in the sagittal plane, and the Cobb angle in the frontal plane. 3D kinematics of the trunk, pelvis, and lower limbs were obtained during gait and anatomical trunk movements (flexion, extension, lateral bending and rotation) [3].

### Results

The Cobb angle was reduced post-operatively in all three patients (Pt-1: 48 vs 30°, Pt-2: 31 vs 17°, Pt3: 58 vs 28°). Pt-1 had a decreased

TK (33 vs 24°), Pt-2 had an increased TK (19 vs 39°), while Pt-3 maintained the same TK (38°) (Figures 1a, 1b). At baseline, lower limb kinematics were within normative range for the three patients. However, Pt-3 presented limited trunk extension (maximum = -18 vs -45° in controls). Post-operatively, the three patients preserved normal lower limb kinematics during gait. Pt-3 had an improved trunk extension (maximum = -22°). Pt-1 and 2 kept normative trunk kinematics (Figure 1c).

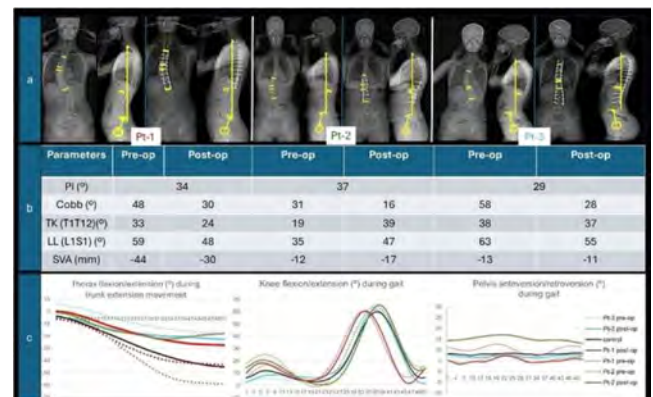
### Conclusion and clinical relevance

Despite their different clinical background (age, Cobb angle, type of scoliotic deformity, sagittal alignment), and the different surgical interventions they underwent (thoracic, thoraco-lumbar, and lumbar arthrodesis), the 3 patients preserved walking kinematics, and improved ROM in their trunk anatomical movements, specifically for Pt-3, despite the severity of the case and the extent of the arthrodesis. In all three cases, the spinal fusion did not include the distal lumbar segment (L4S1). This might explain the conservation of trunk movement in these patients who maintained normal recruitment of the lumbo-pelvic joint.

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Figure 1: Presentation of the 3 patients. a) pre, post-op, frontal, sagittal radiographs; b) radiographic parameters; c) kinematics wave forms during trunk extension movement and gait.



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### Combined effects of increased femoral anteversion and hypermobility on pelvis and hip biomechanics

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## Introduction

Increased femoral anteversion (IFA) is highly prevalent among individuals with joint hypermobility. IFA refers to the forward rotation of the femoral head relative to the transcondylar axis of the knee(1). Joint hypermobility is characterized by excessive joint compliance, which can reduce muscle strength and alter movement mechanics(2). Adolescents with IFA demonstrate significant differences in pelvis, hip, and foot biomechanics during normal walking compared to their peers(3). Altered frontal plane pelvis and hip biomechanics have been associated with various musculoskeletal injuries(4). However, studies investigating the effects of IFA on frontal plane pelvis and hip biomechanics during running are limited. Therefore, the aim of this study is to examine the effects of IFA combined with joint hypermobility (IFAwH) on frontal plane pelvis and hip biomechanics.

## Research Question

Does IFAwH contribute to greater pelvic drop, hip adduction, and knee valgus during running biomechanics?

## Methods

A total of 31 adolescents participated in the study, including 13 with IFAwH ( $12.42 \pm 2.43$  years) and 18 controls ( $14.45 \pm 2.78$  years). Femoral anteversion angles were measured using Craig's test. Joint laxity was assessed with the Lower Limb Assessment Scale (LLAS)(5,6). Participants with a Craig's test value greater than 30 degrees and an LLAS score of 7 or higher were classified into the IFAwH group. All participants underwent 3D running analysis while jogging overground. Independent t-tests were performed for statistical analysis.

## Results

The Craig's angles were significantly greater ( $p < 0.05$ ) in the IFAwH group ( $36.34 \pm 7.12$ ) compared to the control group ( $11.7 \pm 7.25$ ). There was no significant difference in running speeds between the IFAwH ( $2.52 \pm 0.69$  m/s) and control ( $2.56 \pm 0.40$  m/s) groups. In the sagittal plane, the IFAwH group demonstrated higher anterior pelvic tilt and hip flexion peak angles during the stance phase (Table-1). In the transverse plane, peak hip internal rotation was greater in the IFAwH group, while peak pelvic internal rotation, knee external rotation, and foot progression angles were lower. In the frontal plane, aside from peak ankle valgus, there were no significant differences in peak kinematic and kinetic values of the pelvis, hip, and knee between groups.

Table 1: Lower limb kinematics and kinetics

| Stance Phase Kinematics and Kinetics | IFAwH       | Control    | p       |
|--------------------------------------|-------------|------------|---------|
| Peak Pelvic Anterior Tilt (°)        | 16.55±5.40  | 19.85±4.71 | 0.0001* |
| Peak Hip Flexion (°)                 | 35±8.10     | 27.96±7.33 | 0.001   |
| Peak Knee Flexion (°)                | 45.89±11.29 | 46.82±7.87 | 0.42    |
| Peak Pelvic Drop (°)                 | 4.06±2.47   | 4.13±3.06  | 0.828   |
| Peak Hip Adduction (°)               | 10.01±4.22  | 10.15±5.36 | 0.743   |
| Peak Knee Valgus (°)                 | -1.8±3.42   | -1.22±4.52 | 0.219   |
| Peak Ankle Valgus (°)                | -0.70±2.47  | -1.64±2.78 | 0.002   |
| Peak Pelvic Internal Rotation (°)    | 4.06±4.12   | 5.14±4.40  | 0.03    |
| Peak Hip Internal Rotation (°)       | 14.73±8.88  | 10.48±8.17 | 0.0001* |
| Peak Knee External Rotation (°)      | -1.58±8.71  | -6.76±9.33 | 0.01    |
| Peak Foot External Progression (°)   | -0.83±7.63  | -5.43±6.24 | 0.0001* |
| Peak Adduction Hip Moment (Nm/kg)    | 1.26±0.34   | 1.36±0.46  | 0.082   |
| Peak Knee Valgus Moment (Nm/kg)      | -0.09±0.10  | -0.06±0.06 | 0.075   |
| Peak Ankle Valgus Moment (Nm/kg)     | -0.09±0.09  | -0.10±0.03 | 0.847   |

## Discussion

During jogging, IFAwH results in greater hip internal rotation during the stance phase, while reducing external rotation at the knee and foot. In contrast to walking, IFA-induced hip internal rotation leads to a reduction in pelvic rotation(3). Increased hip adduction during the stance phase is associated with common musculoskeletal injuries in the literature(4,7). Additionally, reduced hip abductor strength is known to increase hip adduction during stance, contributing to sports-related injuries. However, the effect of IFA on hip abductor strength remains debated. In this study, adolescents with IFAwH didn't demonstrate increased hip adduction and contralateral pelvic drop during the stance phase compared to controls. These findings suggest that IFAwH may not be a predisposing factor for running-related overuse musculoskeletal injuries.

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## Biomechanical Analysis of Single-Leg Drop Landing in Individuals with Joint Hypermobility

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## Introduction

The non-contact Anterior Cruciate Ligament rupture is a very common knee injury during sport activities. Generalized joint hypermobility is considered one of the risk factors for ACL injuries (1). Biomechanical epidemiologic studies indicate that high knee abduction moments and knee abduction angles and ground reaction forces during a jump landing task are important predictors for ACL injury risk (2,3). The single-leg drop jump (SLDJ) is a dynamic task frequently used to evaluate lower limb biomechanics, neuromuscular coordination, and injury risk. The aim of this study was to investigate the effects of lower extremity joint hypermobility on knee biomechanics in single leg drop landing activity.

## Research Question

Does joint hypermobility lead to increased frontal plane knee valgus in landing activity?

## Methods

Four participants with lower limb joint hypermobility (mean age: 17.25 ± 1.25 years) and six control participants (mean age: 15.66 ± 2.58 years) were included in the study. Joint hypermobility was assessed using the Lower Limb Assessment Scale (LLAS)(4). SLDL activity was analyzed using the VICON motion analysis system, with five repetitions performed on each side on a platform positioned 30 cm from the force platform and elevated 40 cm. Knee and hip biomechanics were evaluated in frontal and sagittal planes. Independent t-test was used for statistical analysis.

## Results

There was no difference between the age and Body Mass Index's (BMI) of the adolescent participants. The hypermobility group demonstrated a significantly higher LLAS score (18.75 ± 5.12) compared to the control group (8.5 ± 2.34) ( $p=0.02$ ). Participants with lower extremity joint hypermobility exhibited significantly greater hip adduction angle, knee valgus angle, knee valgus moment, and foot valgus angles in the frontal plane compared to the control group (Table-1). In the sagittal plane, participants in the hypermobility group demonstrated a lower peak knee flexion angle than control group.

|                                   | Controls<br>(n=6)<br>(Mean±SD) | Hypermobiles<br>(n=4)<br>(Mean±SD) | P       |
|-----------------------------------|--------------------------------|------------------------------------|---------|
| Age                               | 15.66 ± 2.58                   | 17.25 ± 1.25                       | 0.235   |
| BMI                               | 22.1 ± 2.62                    | 20.79 ± 1.08                       | 0.311   |
| Peak Pelvic Down (°)              | 2.85 ± 6.89                    | 1.37 ± 5.66                        | 0.382   |
| Peak Hip Adduction Angle (°)      | 5.95 ± 1.79                    | 8.98 ± 2.32                        | 0.03    |
| Peak Hip Adduction Moment (Nm/kg) | 1.53 ± 0.12                    | 1.47 ± 0.13                        | 0.619   |
| Peak Knee Flexion Angle (°)       | 66.7 ± 3.72                    | 54.74 ± 4.74                       | 0.006   |
| Peak Knee Flex. Moment (Nm/kg)    | 2.19 ± 0.36                    | 2.17 ± 0.62                        | 0.360   |
| Peak Knee Valgus Angle (°)        | 0.53 ± 2.87                    | -8.07 ± 2.65                       | 0.0001* |
| Peak Knee Valgus Moment (Nm/kg)   | 0.13 ± 0.09                    | -0.09 ± 0.11                       | 0.0001* |
| Peak Ankle Valgus (°)             | 1.22 ± 0.78                    | -1.64 ± 1.68                       | 0.006   |

BMI: Body Mass Index, Nm: Newton meter

## Discussion

The findings indicate that individuals with lower extremity joint hypermobility display altered movement patterns compared to those without hypermobility. Specifically, greater hip adduction, knee valgus angle and moment, and foot valgus angles were observed in the frontal plane among the hypermobile group. These deviations suggest a tendency toward dynamic instability and poor alignment during movement, which may increase mechanical stress on the lower limb joints. Additionally, the reduced peak knee flexion angle observed in the sagittal plane among hypermobile participants suggests limited shock absorption capability during tasks involving knee loading.

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## Walking speed reserve and associated clinical parameters in children with cerebral palsy

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## Introduction

The ability to increase walking speed on demand is essential for participating in daily life activities (e.g., playing football) and responding to environmental demands (e.g., crossing a street safely). In children with cerebral palsy (CP), gait impairments may limit this adaptability [1-3]. Walking Speed Reserve (WSR)—defined as the difference between maximal and spontaneous walking speeds—is a potentially valuable indicator of functional mobility and motor adaptability [4], yet it remains underexplored in the CP population.

## Research Question

What gait adaptation strategies do children with CP use to increase their walking speed, and how are these strategies and WSR influenced by specific clinical parameters?

## Methods

This retrospective study analyzed 102 gait analysis sessions from children with CP (GMFCS I and II) classified as unilateral ( $n = 66$ , mean age  $12.8 \pm 3.9$  years) and bilateral ( $n = 36$ , mean age  $12.4 \pm 3.1$  years). Each session included walking trials at both habitual (spontaneous) and maximal speeds. Spatiotemporal gait parameters were computed from 3D gait analysis, and a standardized physical examination assessed joint range of motion (ROM), muscle strength, and spasticity. Statistical analyses (paired t-tests, and univariate regressions) were performed to evaluate differences between conditions and associations with clinical measures.

## Results

All participants were able to increase their walking speed during the fast walking trial, with a mean WSR of  $0.51 \pm 0.17$  m/s. The majority (93.2%) increased both cadence and stride length, whereas 4.9% increased cadence only and 2% increased stride length only. In 73.5% of cases, the walk ratio (step length divided by cadence) decreased. Maximal walking speed was significantly associated with knee flexion ROM, while WSR was negatively associated with spasticity in the hip flexors and knee extensors.

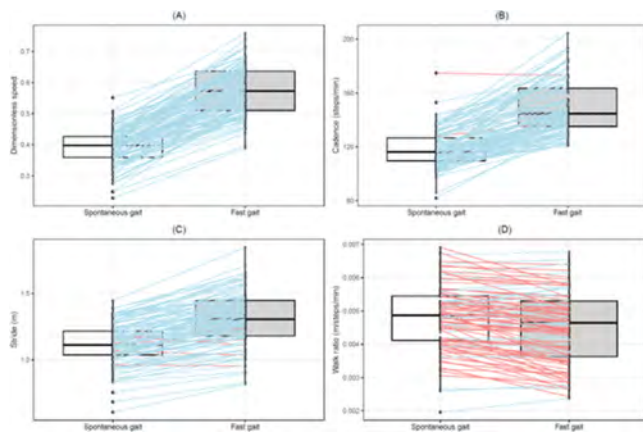


Figure 1: Paired box plots showing changes in dimensionless speed, cadence, stride length, and walk ratio (WR) between spontaneous and fast walking across sessions, with improvements above MCID in blue and non-improvements (or WR decreases) in red.

## Discussion

This study demonstrates that children with CP, even those with mild to moderate motor impairments (GMFCS I–II), retain a significant capacity to increase walking speed. Most children relied on a combined increase in cadence and stride length, reflecting preserved but variable motor adaptability. WSR appears to capture meaningful inter-individual differences in this capacity and is influenced by

neuromuscular impairments such as spasticity and ROM limitations. These findings support the integration of fast walking tests and WSR measurement into clinical gait assessments to better characterize functional potential and inform individualized rehabilitation strategies aimed at enhancing mobility and participation.

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## Kinematic adaptations during trunk movements in adolescent idiopathic scoliosis with different types of curvature

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## Introduction

Adolescent Idiopathic Scoliosis (AIS) is a three-dimensional deformity of the spine that can affect one or more spinal segments (thoracic, thoracolumbar, lumbar), described by the Lenke type. Scoliosis is known to limit kinematics during walking and affect patient's quality of life [1]. The spinal deformity can also affect trunk movements such as flexion, extension, lateral bending and axial rotation that are highly used in daily life activities. While surgical correction using spinal fusion increases limitation in trunk movements, it is essential to describe these limitations in AIS pre-operatively.

## Research Question

What are the kinematic strategies and compensatory mechanisms used by AIS patients with different Lenke types during trunk movements?

## Methods

43 AIS patients: 30 Lenke-1 with right thoracic convexity, 13 Lenke-5 with left thoraco-lumbar/lumbar convexity (average Cobb:  $40^\circ$  &  $23^\circ$  respectively) and 24 controls underwent low-dose biplanar X-rays, with the calculation of 3D spinopelvic & scoliosis parameters, and filled the SRS22r questionnaire. 3D movement analysis was performed during maximal trunk flexion, extension, lateral bending and axial rotation with the calculation of the kinematics of the head, trunk, shoulders, spinal segments and pelvis [2,3]. Kinematics parameters (mean, ROM) and waveforms (statistical parametric mapping [4]) were compared between groups and their relationship with radiographic parameters and QOL scores were investigated.

## Results

Lenke-1 patients had a reduced radiographic thoracic kyphosis (T1T12: 32 vs 42° in controls) and altered self-image score (3 vs 4 in controls; both  $p < 0.05$ ). Lenke-5 patients achieved maximal trunk movements similarly to controls while recruiting different kinematic strategies. Lenke-1 patients reached maximal trunk rotation to the right (65°) despite a reduced shoulder rotation (20 vs 27°,  $p < 0.01$ ). During trunk flexion, Lenke-1 patients had a reduced overall movement (100 vs 115° in controls) with a reduced thoracic flexion (T6T9-T9T12: 10 vs 15°). Throughout trunk flexion, they maintained a left lateral bending of their trunk (-3 vs 2°) and an increased shoulder rotation to the left (5 vs 0°; all  $p < 0.05$ ). The reduced trunk flexion was correlated to the severity of the Cobb angle ( $r = -0.4$ ), the reduced radiographic thoracic kyphosis ( $r = 0.5$ ) and the altered self-image ( $r = 0.3$ ; all  $p < 0.05$ ).

## Discussion

The rigidity of the scoliotic thoracic segment might have limited Lenke-1 patients' ability to flex their trunk and led them to recruit compensatory mechanisms in frontal and horizontal planes. Limitations in the trunk's sagittal movement seem to be related to the severity of the frontal and sagittal scoliotic deformity and the alteration of patient's self-image. Thus, daily life activities requiring extreme trunk flexion might be limited in Lenke-1 patients. Future studies will explore trunk movement limitation following selective and non-selective spinal fusion in AIS.

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Figure 1- Trunk flexion and correlation with radiographic parameters and QOL scores.



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## Muscle fat infiltration in the lumbar spine is related to functional impairment in patients with adult spinal deformity

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## Introduction

Adult Spinal Deformity is known to cause postural malalignment and gait limitations in patients, associated with a deterioration of their quality of life [1]. The spinal deformity can be due to musculoskeletal disorders affecting the spinopelvic alignment or tissue degeneration. Fat infiltration of paraspinal muscles, mainly assessed quantitatively on MRI, has been associated with global malalignment and increased postoperative complications in ASD [2]. However, its relationship with functional limitations is not yet explored. While quantitative measurement of muscle fat infiltration requires specific MRI acquisition protocols, the qualitative Goutallier classification remains underexplored in functional assessment.

## Research Question

Is the qualitative Goutallier classification for muscle fat infiltration on lumbar MRI related to severe malalignment and functional impairment in ASD?

## Methods

64 ASD patients underwent full body biplanar X-ray to assess 3D spinopelvic and global alignment parameters: sagittal vertical axis (SVA: sagittal plumbline from C7 to the sacrum), pelvic retroversion (measured through pelvic tilt PT), pelvic incidence (PI), lumbar lordosis L1S1, and their mismatch (PI-LL), and the frontal Cobb angle. They all filled the Oswestry Disability Index (ODI) questionnaire. Patients performed 3D gait analysis to calculate joint kinematics of the thorax, spinal segments, pelvis and lower limbs, in addition to the spatio-temporal parameters and the Gait Deviation Index [3, 4, 5]. Lumbar MRIs were collected to evaluate fat infiltration in the paraspinal muscles using the Goutallier classification. Patients were grouped into high (High-Infiltration: Goutallier grades 3-4) or low-to-moderate infiltration (Low-Mod-Infiltration: Goutallier grades 0-2). Radiographic and kinematic parameters as well as ODI scores were compared between groups and correlations with Goutallier grade were computed.

## Results

Patients were grouped as 33 with High-Infiltration and 31 with Low-Mod-Infiltration. On standing radiographs, the High-Infiltration group had increased SVA (76 vs 22mm), PT (28 vs 21°), PI-LL (19 vs 3°) and Cobb angle (23 vs 14°; all  $p < 0.05$ ). During gait, they had a decreased GDI (78 vs 92), knee sagittal ROM (48 vs 57°) and walking speed (0.7 vs 0.9m/s) associated with an increased thorax flexion (13 vs 4°; all  $p < 0.05$ ). They also had increased ODI scores (48 vs 36,  $p = 0.02$ ). Goutallier grades positively correlated with SVA ( $\rho = 0.43$ ),

PI-LL mismatch ( $\rho=0.44$ ), thorax flexion during gait ( $\rho=0.34$ ), and negatively correlated to knee sagittal ROM ( $\rho=-0.39$ ) and walking speed ( $\rho=-0.43$ ; all  $p<0.05$ ; Fig.1).

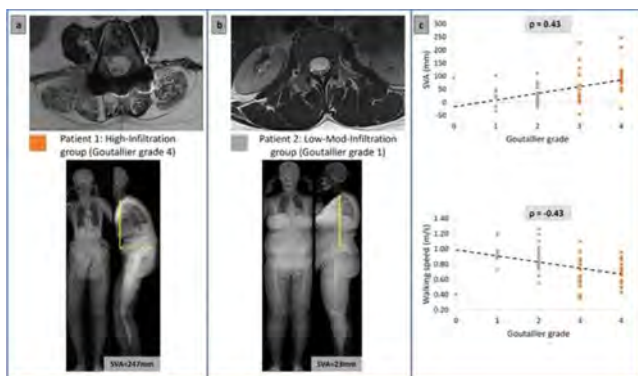
## Discussion

The Goutallier classification, an easy and accessible method using routine MRI, can indicate malalignment severity and functional impairment in ASD patients, making it a relevant clinical indicator of disease severity. Thus, a high Goutallier grade should alert clinicians to the need for targeted interventions, such as muscle reinforcement, to mitigate fat infiltration and optimize outcomes in both operative and non-operative patients.

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Fig. 1- Example of patients with (a) high and (b) low-moderate muscle fat infiltration; correlations between Goutallier grade, SVA and walking speed (c).



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## Influence of anthropometric variation in femur and tibia length on muscle-tendon length during walking

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## Introduction

Muscle-tendon lengths (MTL) are critical for understanding movement disorders such as cerebral palsy and developing effective treatment strategies [1]. Although musculoskeletal modelling enables MTL estimation, clinical adoption is often limited by the complexity of scaling and inverse kinematics. Hence, workflows

have been developed using joint angles calculated with the conventional clinical gait model and a generic (unscaled) musculoskeletal model to estimate MTL [2]. However, this approach neglects anthropometric variations in segment lengths, which might affect MTL.

## Research Question

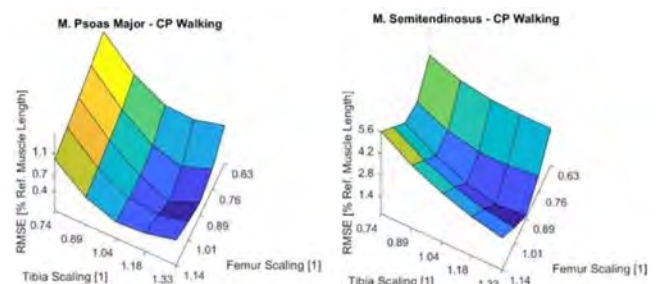
How does anthropometric variations in femur and tibia lengths affect the calculation of MTL?

## Methods

Previously collected three-dimensional gait data and joint angles from a healthy participant and one with cerebral palsy were used for the sensitivity analysis. Gait data from both participants, together with the Rajagopal model [3], were used to calculate MTL in OpenSim. MTL from the unscaled model served as reference for all further analysis. Twenty-five additional models were created using varying scale factors. To assess the effect of anthropometric variation, femur and tibia scaling factors were varied across the 95% confidence interval from a normative dataset of 7,970 individuals aged 2–92 years (mean:  $16.9 \pm 13.1$  years). Scaling intervals were defined as [0.63, 1.14] for the femur and [0.74, 1.33] for the tibia. Within each interval, five equidistant scaling factors were selected, resulting in 25 femur–tibia scaling combinations and corresponding OpenSim models. MTL were calculated using all models and both gait kinematics as input. MTL continuity was ensured using the Muscle Moment Arms Checker [4]. Four muscles were analyzed: M. semitendinosus, M. rectus femoris, M. psoas major, M. gluteus maximus. For each motion, muscle, and model, the root mean square error (RMSE) in normalized muscle lengths between scaled and reference models was computed to assess the impact of anthropometric variations on MTL.

## Results

Variations in femur and tibia lengths in the simulations based on healthy gait kinematics led to maximum MTL changes in M. semitendinosus, M. rectus femoris, M. psoas major, and M. gluteus maximus of 8.7, 2.1, 0.6, and 4.1%, respectively. In the CP gait simulations, anthropometric variations caused maximum changes in M. semitendinosus, M. rectus femoris, M. psoas major, M. gluteus maximus MTL of 7.9, 2.6, 0.9, and 5.2 %, respectively.



RMSE of normalized muscle-tendon lengths between scaled and unscaled models across 25 femur–tibia scaling combinations. Kinematics from a child with cerebral palsy. Left: Musculus psoas major. Right: Musculus semitendinosus.

## Discussion

Our study showed that anthropometric variations affect MTL. However, only combinations of extreme scaling factors (e.g., femur: 0.63; tibia: 0.74), which are unlikely in practice, caused large discrepancies from the reference MTL. Nevertheless, subject-specific segment ratios should be included in musculoskeletal simulations to ensure accuracy. We aim to develop easy-to-use workflows to estimate MTL and other musculoskeletal parameters in clinical settings.

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## Sonomyography of the gastrocnemius medialis muscle during walking in persons post-stroke

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## Introduction

Sonomyography (SMG) is an emerging tool for the in-vivo study of muscle-tendon behaviour during movement. When combined with 3D gait data, SMG can be used to quantify the behaviour of the gastrocnemius medialis muscle (GM) during gait, a muscle often impaired in subjects with neurological disorders, such as cerebral palsy and stroke. GM SMG in these populations can help identify

pathology specific changes in this muscle and how they influence gait recovery. While GM SMG has been well described in children with cerebral palsy [1-3], its feasibility in subjects post-stroke remains unexplored.

## Research Question

What are the effects of GM SMG on gait kinematics in persons post-stroke compared to healthy controls and can it capture differences in GM behaviour between these populations?

## Methods

3D lower-limb kinematics (Vicon, Oxford Metrics) were collected from 12 persons (6 subjects  $7 \pm 7$  m post-stroke and 6 age- and sex-matched controls, mean age  $54 \pm 8$  y, 5 males) while they walked at a self-selected walking speed on a treadmill (GRAIL, Motek Medical) with and without SMG (Teleded SmartUS, 59mm linear probe). SMG was collected from the paretic (or corresponding) leg with the ultrasound-probe positioned on the GM muscle-tendon junction (MTJ) using a probe-holder (Probedex DynamicT, USONO -Figure1C), together weighing 330g. Data were averaged over three time-normalised gait cycles per condition, per subject. Spatiotemporal and sagittal plane kinematic hip, knee and ankle data were compared between conditions using paired t-tests and SPM one-way repeated measures ANOVA ( $\alpha = 0.05$ ). Semi-automatic MTJ-tracking software was used to quantify muscle, tendon and muscle-tendon-unit length change relative to length at initial contact [2].

## Results

SMG had no significant effects on the spatiotemporal or kinematic data in persons post-stroke and healthy controls (Figure1A). Walking speed differed between post-stroke and control subjects ( $0.6 \pm 0.3$  vs.  $1.1 \pm 0.3$  m/s,  $p < 0.004$ ). Corresponding with a smaller ankle range of motion and increased knee flexion, the relative MTU length change during the stance phase (Figure1B) was, on average, lower post-stroke (+17mm) compared to control (+25mm). In both groups, the tendon aided in lengthening the MTU during midstance. The muscle length changed minimally over the entire gait cycle in stroke (4mm) compared to control (16mm), indicative of post-stroke GM paresis.

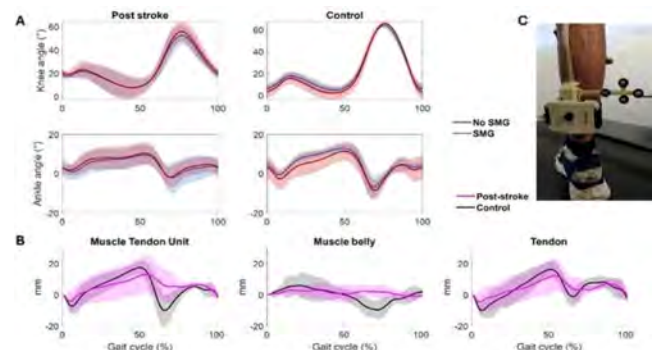


**Figure 1.** A) Sagittal-plane kinematics with and without sonomyography (SMG). B) Relative gastrocnemius medialis length changes. C) Ultrasound as used during SMG.

## Discussion

Confirming previous findings in cerebral palsy [1,4], GM SMG had minimal effects on the gait pattern of persons post-stroke walking at a self-selected speed on a treadmill. Ankle and knee kinematic impairments post-stroke were accompanied primarily by reduced GM muscle-belly length changes, reflective of GM paresis. SMG data collection in persons post-stroke is feasible and can provide insights on dynamic GM behaviour during walking, possibly leading to the finetuning of rehabilitation strategies to improve gait performance in this population.

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## Separation of patients with reverse total shoulder arthroplasty from controls based on their Movement Deviation Profiles derived from shoulder kinematics

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## Introduction

Reversed Total Shoulder Arthroplasty (RSA) is commonly performed to treat patients with degenerative deficiencies. However, one of the main challenges remains the impact on functional movement, particularly in placing tasks. This study is the first to evaluate and compare shoulder and elbow kinematics during a functional placing task among unimpaired elderly volunteers and patients with RSA.

## Research Question

*Can an AI-based multidimensional approach distinguish between patients with RSA and healthy controls during a placing task?*

## Methods

A convenience sample of 10 healthy individuals (mean 71.2y  $\pm$  9.19y, 8 ♂) and 10 RSA patients (mean 66.0y  $\pm$  8.36y, 7 ♀) were invited to participate. Sex, age, dexterity, affected side, BMI, date of surgery, Shoulder Pain and Disability Index (SPADI) functional scores, active Range of Motion (ROM), and forward flexion strength were recorded for the RSA patients. All participants placed a 1kg object on a 120 cm shelf. Movement was captured by six Noraxon IMUs placed on C7, T12, S2, upper arm, lower arm and hand. Shoulder flexion and internal rotation, elbow flexion and forearm supination angles were normalised to 101 points and exported from Visual3D. The self-organizing neural network underlying the Movement Deviation (MDP) program [1] was trained with control kinematics, and then tested with the same control kinematics and angles of the RSA patients. Pearson correlations were calculated between average MDP and age, BMI, days post-surgery, active ROM and forward flexion strength for the RSA patients.

## Results

The MDP profiles of RSA patients significantly differed from controls (figure 1). Thanks to its dimensionality reduction of a multivariate problem, the MDP clearly separated RSA patients from healthy controls during a placing task with increasing separation between groups towards the second half of the task. No significant correlations were found. An estimated sample size of 29 patients was calculated to be required to find significant positive correlation between MDP and SPADI score and to establish if movement deviations from controls reduces over time, as these two variables showed highest correlation coefficients ( $r = 0.509$  and  $r = -0.501$  respectively).

Figure 1: Movement Deviation Profiles of RSA compared to Controls.

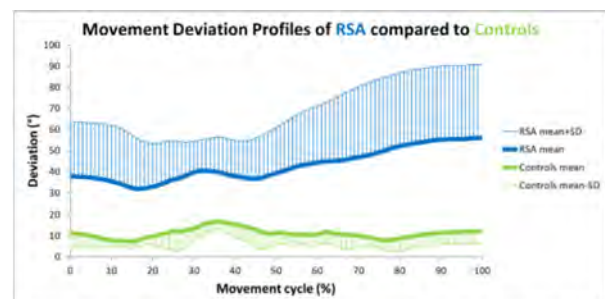


Figure 1: Movement Deviation Profiles of RSA compared to Controls. Only half of the SD bands are shown for clarity of the image. RSA: Reverse Shoulder Arthroplasty; SD: Standard Deviation, °: degrees.

## Discussion

The MDP approach successfully highlighted differences between RSA patients and healthy controls, capturing deviations in movement performance that are not apparent through traditional clinical assessments. Larger differences between groups in the second half of the task might be explained by the lever arm increasing when the loaded hand (1kg) approaches the shelf. Interestingly, some patients' MDP closely resembles that of healthy controls. Future studies might explore if the MDP is lower with better functionality and more time after surgery. The MDP could be a valuable tool for assessing rehabilitation progress, guiding clinical decision-making, and identifying subtle motor deficits that impact post-surgical recovery.

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## Retrospective analysis of hamstring lengthening in children with cerebral palsy: Muscle-tendon-length as base for identifying surgical indications and assessing outcomes

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## Introduction

Hamstring shortening is a common contributor to crouch gait in children with cerebral palsy (CP) and is frequently addressed via surgical lengthening. The primary surgical goal is to reduce excessive knee flexion at initial contact without causing compensatory increases in anterior pelvic tilt. Recent studies suggest that muscle-tendon length (MTL), estimated through musculoskeletal modeling, could serve as an objective parameter to support surgical decision-making. This study retrospectively evaluates outcomes in patients who underwent hamstring lengthening without MTL-based planning analyzing the relative to preoperative MTL and postoperative changes.

## Research Question

What are the functional and biomechanical outcomes of hamstring lengthening in children with CP when performed without considering preoperative muscle-tendon length, based on a retrospective analysis of 3D gait data?

## Methods

Gait and clinical data of 33 children with CP (mean age:  $10.2 \pm 3.1$  years) were retrospectively analyzed. Inclusion required hamstring lengthening and pre- and postoperative 3D gait analysis. MTL of semimembranosus, semitendinosus, and biceps femoris long head were calculated using OpenSim (modified Rajagopal model) and normalized to upright standing. Patients were grouped by surgical procedure (percutaneous myofascial or open hamstring lengthening). MTL was classified as "short" when it fell more than 2 standard deviations (SD) below the normative mean and as "normal" or "long" otherwise. Statistical analyses included descriptive statistics, statistical parametric mapping, one-sample t-tests, repeated measures general linear model (GLM), and correlation analysis.

## Results

Not all patients with previously "short" MTL showed an increased anterior pelvic tilt (figure 1). Postoperative MTL significantly increased across all muscles ( $p < .001$ ). GLM showed significant between-subject effects between OP-category and outcomes for both knee flexion and pelvic tilt ( $p < .01$ ). No significant interaction was found between surgical methods and outcomes for MTL. For all variables there was significant interaction between outcomes and MTL-category ("short" or "normal"). In patients with preoperatively shortened hamstrings changes in knee extension, pelvic tilt and MTL were more pronounced, compared to those with normal-length hamstrings. One-sample t-tests indicate significantly increased MTL with larger effects in the "short" group, while in the "normal" group only small effects of significant postoperative change was observed.

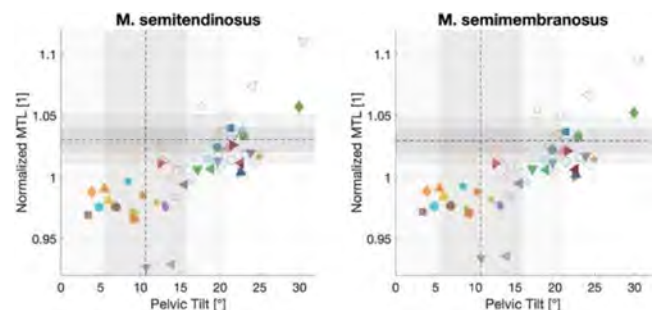


Figure 1: Pre- (filled) and post-operative (empty) values of pelvic tilt and normalized MTL for each patient undergoing the Ulzibat lateral-medial procedure. Grey bands indicate deviation of norm:  $\pm 1SD$  dark,  $\pm 2SD$  light.

## Discussion

Surgical indications were determined clinically by integrating physical examination, 3D gait analysis, patient history and intraoperative assessment. In contrast, this study relied exclusively on quantitative gait analysis. MTL quantification appears to be a valuable, objective predictor for surgical indication and outcome evaluation in hamstring lengthening. Including MTL into preoperative decision-making may help improving the outcome and anticipating biomechanical consequences better. These findings support the clinical integration of musculoskeletal modeling in surgical planning for children with CP.

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## Stepping forward in recognising activities of daily living: Sharing labelled datasets between individuals for identifying unsupervised clustering

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### Introduction

The progression of musculoskeletal diseases such as osteoarthritis [1] are influenced by activities of daily living (ADLs). Traditionally, ADLs are recorded with supervised machine learning methods trained on body worn accelerometer data, but these are limited by the need for individualised labelled samples. This may be solved by using unsupervised learning [2,3]. The self-organising map (SOM) [4] has been used to recognise ADLs [5], but it is yet unknown if this solves the requirement of individualised labelled data.

### Research Question

Can labelled walking data be shared between individuals on a self-organising neural network trained with activities of daily living?

### Methods

One triaxial accelerometer (ActiGraph GTX9) was worn on the dominant wrist of two unimpaired participants (P1: male, 1.84m, 92kg and P2: male, 1.79m, 80kg) to capture 12 seconds of normal walking each. P1 continued unlabelled data collection of ADLs for a further 11 days. All data were resampled to 10Hz and formatted to one-second windows. A SOM was trained on the 11 days of ADLs using the SOM Toolbox 2.1 [6]. P1 and P2 walking data were projected onto the SOM and the nodes that these mapped to were inspected along with the number of times they were mapped to.

### Results

The trained SOM consisted of 15,540 nodes on a 140 x 111 matrix which mapped all ADLs over 11 days. Walking data from P1 and P2 mapped to the same region of the SOM, covering twelve nodes all along the right-most edge of the map. Within this region, the walking data formed two (sub)clusters, see figure 1. The first cluster contained 100% of P1 walking data and 50% from P2, where 45.45% of all P1 walking data mapped to the exact same node as 38.18% of all P2 walking data. The remaining 50% of P2 walking made the second cluster.

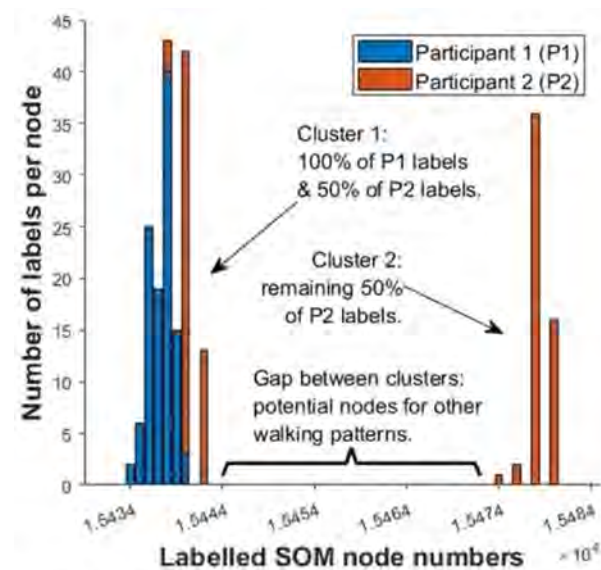


**Figure 1.** The nodes of the SOM trained on 11 days of ADLs that were mapped to (and the frequency at which they were mapped to) by the labelled walking data from P1 (blue) and P2 (orange). Twelve of the 15,540 nodes were labelled to one region, forming two clusters.

### Discussion

Known walking data from two individuals were successfully mapped to the same region of a SOM trained on 11 days of unsupervised ADLs. The overlap of SOM nodes that the walking data from two individuals mapped to suggests that labelled data may be shared between individuals. The second cluster of walking data from one individual is likely due to natural variations in walking patterns [7]. Furthermore, the unlabelled gap between the two clusters likely contains variants of walking patterns from the 11 days of ADLs but these were not mapped onto by the in-lab labelled data. A larger labelled dataset with more variability from multiple people may connect the two clusters and could be used as a single reference for identifying ADLs in future SOMs without the need for individual-specific labelled data.

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### Duration of muscle contraction as a key variable in FXS children classification from typically developing peers: an Unsupervised approach

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### Introduction

Fragile X Syndrome (FXS) is a rare genetic disorder caused by CGG triplet repeats in the *FMR1* gene, leading to reduced or absent Fragile X Mental Retardation Protein (FMRP) [1]. Individuals with over 200 repeats have a full mutation (FXSFull), while 55–200 indicates a premutation. Somatic mosaicism (FXSMos) can influence FXS phenotypes [2]. FXS is a leading cause of intellectual disability and autism spectrum disorder (ASD), often accompanied by flat feet, joint laxity, hypotonia, that can lead to an altered gait [3]. Gait analysis (GA) can be used to quantify movement impairments and guide the assignment of personalized physical therapy [3]. To assist in physical therapy assignment, grouping children with similar gait patterns can be a valuable support [4], particularly through unsupervised machine learning. In this context, selecting the relevant biomarkers is crucial, to enhance model performance and prevent overfitting.

### Research Question

The aim is to identify which biomarkers from gait analysis (kinematic and EMG-derived features) may best aid unsupervised machine learning algorithms to distinguish children with FXS from typically developing children.

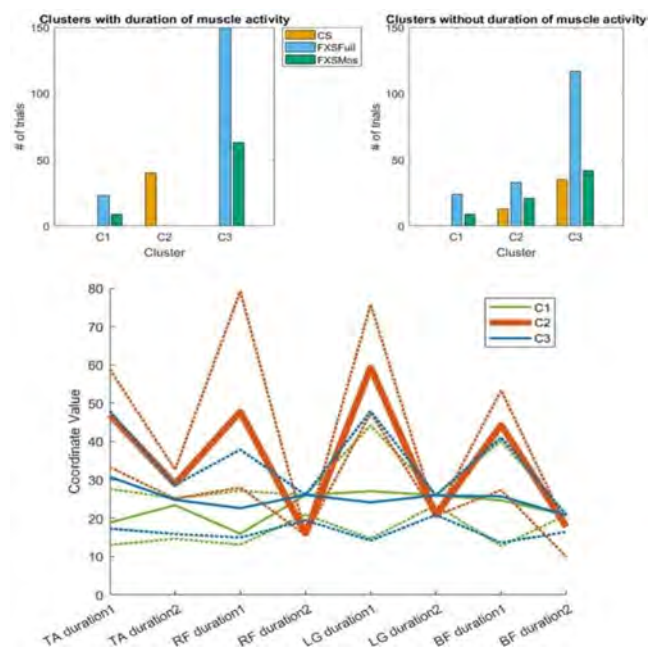
### Methods

41 children with FXS (age:  $10.04 \pm 3.55$ ; BMI:  $18.02 \pm 6.66$ ) and 10 age and BMI-matched controls (CS) were enrolled (of whom the data of two subjects were discarded). Within the FXS group, 30 had FXSFull and 11 had FXSMos. Gait data were collected through four synchronized GoPro Hero 7 cameras (60 fps) and a surface EMG

system (Cometa, 2000 Hz), recording the activity of Tibialis Anterior (TA), Gastrocnemius Lateralis (GL), Rectus Femoris (RF), and Biceps Femoris (BF). Space-time parameters, joint angles and EMG-derived features were extracted as described in [3]. Clustering algorithms were applied to the dataset and to different subset of feature sets.

### Results

The best performance was achieved using hierarchical clustering on stride length, velocity, step width, double support time, stride time and duration of muscle activity as features. The model separated CS (cluster C2) from FXS group (clusters C1 and C3), as shown in the top-left plot of Figure 1. When using only spatio-temporal features, CS group was splitted in cluster C2 and C3 (top-right plot). The parallel plot of the first cluster analysis (bottom of Figure 1) shows that C2 (CS group), orange line, had longer muscle activity duration across all muscles, except for the duration of the second activation of RF and GL.



### Discussion

Cluster analysis showed that muscle activity duration can efficiently distinguish the CS group from children with FXS, whereas spatio-temporal parameters alone were insufficient. As expected, children with FXS had shorter activity duration, likely due to increased number of muscle activations within a gait trial and the presence of co-contractions. This muscle activation strategy suggests the presence of compensatory mechanisms, possibly related to anaerobic metabolism and/or a prevalence of a recruitment of fast-twitch muscle fibers [5].

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## Pedobarographic Normative Data for Typically Developing Children aged 4-17 years: Age-Related Changes in Foot Posture and Load Distribution

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## Introduction

Pedobarography is frequently employed to identify and characterise foot pathologies in children. Pes planus and pes cavus can be difficult to distinguish when an incipient flatfoot is pronated but the medial midfoot lacks ground contact. In adults, the centre of pressure excursion represented by the gaitline has been shown to distinguish between pes cavus and pes planus, while the area between the gaitline and foot axis describes foot posture (pronated/supinated) [1].

## Research Question

As these parameters could also be applicable in paediatric diagnostics, the objective of this cross-sectional study was to provide pedobarographic data of typically developing children aged 4-17y, including the aforementioned parameters measuring foot posture.

## Methods

101 typically developing children, aged 4-17 years and divided into five age groups: 4-6, 7-8, 9-11, 12-14 and 15-17 years were included. Dynamic foot pressure was measured at self-selected speed using the mid-gait protocol. Force and pressure variables measuring peak and total load as well as the start of ground contact were calculated for seven foot regions: lateral and medial heel, midfoot, lateral and medial forefoot, hallux and toes 2-5. The centre of pressure excursion index (CPEI) [2] and the area between the gait line and the axis of the foot (AGL) were then calculated. The latter was also normalised by the foot contact area (AGL.norm) to take account of the growth in length.

## Results

The most pronounced changes were observed between 7 and 14y including an increase in load on the foot with advancing age and a load shift to the forefoot. The onset of hallux contact was significantly earlier in the stance phase in the 12-17y olds. The evaluation

of foot posture during the stance phase showed that AGL and AGL.norm increased with age (see Table 1).

|                        | Age Group (years) |             |             |              |              | p-value |
|------------------------|-------------------|-------------|-------------|--------------|--------------|---------|
|                        | 4-6               | 7-8         | 9-11        | 12-14        | 15-17        |         |
| AgL [cm <sup>2</sup> ] | 9.9 ± 3.7         | 11.6 ± 5.3  | 14.4 ± 3.7* | 16.2 ± 4.6** | 17.3 ± 6.4** | <0.000  |
| AgL.norm [%]           | 0.11 ± 0.02       | 0.11 ± 0.03 | 0.13 ± 0.03 | 0.13 ± 0.04  | 0.14 ± 0.04  | 0.020   |
| CPEI                   | 18.4 ± 7.0        | 17.0 ± 8.9  | 18.3 ± 5.8  | 16.8 ± 6.8   | 11.4 ± 8.1   | 0.041   |

Table 1: AGL, AGL.norm and CPEI are expressed as mean ± standard deviation with ANOVA results. \*significant difference compared to Group 4-6, \*\*significant difference compared to Group 7-8.

## Discussion

Since a low AGL is associated with a pronated foot position, this supports previous findings that the foot develops from a pronated to a more neutral position [3,4]. Conversely, CPEI decreased with age. This result seems to contradict the development of a neutral foot posture as a low CPEI is associated with a pronated foot position [5]. However, CPEI primarily describes the concavity of the gait line in the forefoot. It has been shown that in children, a faster weight transfer occurs from the lateral to the medial forefoot [6]. Together with the earlier ground contact of the hallux in adolescents, this may indicate that the concavity of the gait line is reduced due to a more controlled medial weight transfer. These findings should be considered in the diagnostic differentiation between pes planus and pes valgus in children.

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## Sport-specific biomechanical and neuromuscular profiles during sidcutting: A comparative study of young female handball and football players

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## Introduction

Female athletes in sports such as handball and football are at high risk of anterior cruciate ligament (ACL) injuries, particularly during landings and sidcutting maneuvers, and the incidence during match has been shown to be higher for female handball players compared to football players (1-2). While biomechanical and neuromuscular factors have been implicated as risk factors (3), few studies have directly compared these parameters between sports with differing ACL injury rates.

## Research Question

This study aimed to investigate differences in isometric muscle strength, hip and knee joint biomechanics, and neuromuscular coordination during sidecutting between young female elite handball and football players. It was hypothesized that handball players would display biomechanical patterns more associated with ACL injury risk than football players.

## Methods

Ninety female elite athletes (54 handball, 36 football), all selected for youth national teams or national-level training, participated. Each performed five sport-specific sidecutting maneuvers on their non-dominant leg, simulating a game situation. Kinematic and kinetic data were collected using an 8-camera Vicon motion analysis system synchronized with force plate measurements. Surface electromyography (EMG) recorded activity from the quadriceps and hamstring muscles. Maximal voluntary isometric strength of the hamstrings, hip abductors, and external rotators was measured using a hand-held dynamometer. Group differences were analyzed using unpaired t-tests ( $\alpha = 0.05$ ), with effect sizes estimated by Cohen's  $d$ .

## Results

Handball players were significantly taller and heavier than football players, but groups were age-matched. Absolute isometric strength in knee flexion and hip abduction was higher in handball players, but these differences disappeared after normalization to body mass; football players showed greater normalized hip external rotation and hip extension strength. Football players approached the cut with higher velocity, while handball players executed a greater change of direction and higher lateral take-off velocity. Kinematically, football players demonstrated more knee and hip flexion at initial contact, with no difference in foot strike pattern, whereas handball players exhibited greater pelvic rotation towards the direction of movement. Kinetic analysis revealed that handball players had larger knee extension and hip extension moments during early ground contact, with no significant differences in frontal or transverse plane moments. EMG analysis showed higher pre-activation of the hamstrings in football players, with no significant differences in quadriceps activity.

## Discussion

The findings indicate that young female elite handball and football players differ in anthropometrics, muscle strength profiles, and sidecutting biomechanics. Handball players' movement patterns—characterized by less hip external rotation strength, greater knee and hip extension moments, and lower hamstring pre-activation—may contribute to their higher ACL injury risk compared to football players. These sport-specific biomechanical and neuromuscular differences highlight the need for tailored injury prevention strategies in female athletes.

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## Moving towards personalization of split belt treadmill-based treatment of gait asymmetry

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## Introduction

Split-Belt Treadmills (SBTM) are used to address gait asymmetry (GA) in patients with conditions like Parkinson's disease (PD) [1]. The SBTM includes two belts, one for each leg. The belts can operate at equal speeds, i.e., Tied-Belt (TB) mode, or different speeds, i.e., Split-Belt (SB) mode. When exposed to SB mode for 5-15 min, participants experience an 'adaptation period' (AP), adjusting their gait to the uneven conditions. This 'imprints' a new gait pattern in the central nervous system. Returning to TB mode or walking over ground immediately after adaptation shows improved GA and coordination [2]. This effect rapidly subsides; more intense SBTM training is required to achieve lasting effects [3]. Research indicates optimal results when reducing belt speed on the side with initial longer step length [2, 4]. Early in the adaption period (EA), GA is increased (augmented) due to the SB manipulation. As adaptation proceeds augmentation is reduced [4].

## Research Question

Which periods of TB-SB-TB training can be used to predict correction outcomes?

## Methods

### Participants:

7 lower-limb amputees (age:  $35.4 \pm 11.1$ , 2 females), 5 persons with PD ( $65.4 \pm 7.9$ , 0 f), 4 old adults ( $64.0 \pm 3.7$ , 2 f).

### Apparatus:

SBTM (R-Mill, ForceLink, Netherlands). Vicon motion capture system (Oxford, UK) captured kinematics from passive markers.

### Procedure:

(i) TB baseline (BL; 1.5 min); (ii) SB training with 50% speed reduction on the side with the longer step length at BL ( $SL_{BL-Long}$ , 4.5 – 12 min - based on cohort); and (iii) TB mode (Post-Adaptation, PA).

### Analysis:

GA was calculated as:

$$GA[\%] = (SL_{BL-Long} - SL_{BL-short}) / (SL_{BL-Long} + SL_{BL-short}) * 100$$

To assess the efficacy of the trial in affecting GA we defined a correction factor (CF):

$$CF = - (GA_{PA} - GA_{BL}) / GA_{BL}$$

Augmentation factors (AF) were calculated for the entire SB period, EA (first 30 sec of SB) and late augmentation (LA, last 30 sec of SB):

$$AF_{SB/EA/LA} = (GA_{SB/EA/LA}) / GA_{BL}$$

## Results

GA was reversed in thirteen participants, reduced in two, and increased in one. Mean  $GA_{BL}$  ( $\pm$ SD) values were  $3.87 \pm 3.02$ ,  $GA_{PA}$ :  $-2.8 \pm 4.16$  (Fig. 1A). After excluding one outlier ( $CF=19.51$ ;  $GA_{BL}=0.57$ ),  $AF_{LA}$  best predicted the resulting correction ( $R^2 = 0.55$ ,  $p = 0.02$ ), with higher  $AF_{LA}$  corresponding to lower CF (Fig. 1B).

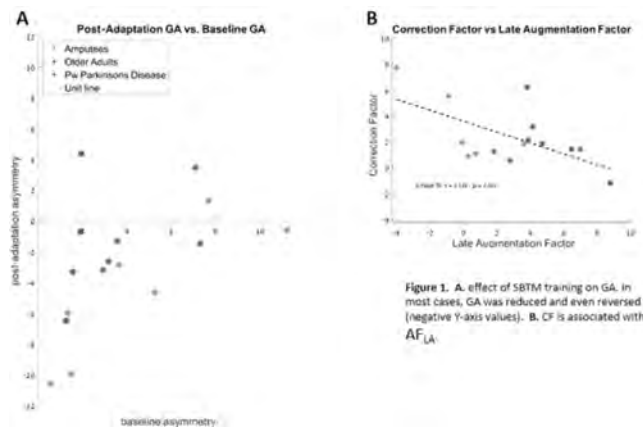


Figure 1. A. effect of SBTM training on GA. In most cases, GA was reduced and even reversed (negative Y-axis values). B. CF is associated with  $AF_{LA}$ .

## Discussion

Our study confirms that SBTM training can reduce gait asymmetry. In this single-dose paradigm we discerned that the level of gait asymmetry correction is inversely correlated with the level of GA augmentation during the adaptation period, in particular during late adaptation. Meaning, the more a participant is able to return to more symmetrical gait after initial augmentation, the more successful the correction. We posit that our study can facilitate individualization of SBTM training protocols to optimize outcomes.

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## VAF-CI: Quantifying low-dimensional global movement complexity during a novel lower limb motor learning task

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## Introduction

Gait rehabilitation typically uses key spatio-temporal metrics to assess progress, however they fail to capture the coordinated joint dynamics that define normal gait [1]. To address this, we explore the use of the Variance-Accounted-For based Complexity Index (VAF-CI), adapted from methods used to analyze muscle synergies in upper limbs [2]. This index uses synergy-based analysis to quantify the overall complexity of multi-joint movement pattern coordination in healthy individuals.

## Research Question

What is the relationship between the Complexity Index and task performance scores across different stages of learning a novel motor task?

## Methods

Sixteen healthy adults (mean age:  $26.7 \pm 2.9$  years) participated in a two-day walking protocol on a split-belt treadmill at 0.7 m/s. Participants performed one 6-minute Normal Walk (NW) and six 5-minute Target Walk (TW) bouts where they received continuous real-time visual feedback. The target to reach was matching their right ankle dorsiflexion to an inverted swing trajectory. Performance scores, rescaled to a unit range, reflected continuous target tracking accuracy. Full-body kinematics were recorded using a Vicon motion capture system, and joint angles were computed in real time using the Human Body Model 2 and Motek D-Flow. The VAF-CI was computed offline in MATLAB and takes into account the cumulative variance-accounted-for across synergies over the total number of time-stamped observations [2].

## Results

We identified significant correlations between VAF-CI values and task scores. Specifically, we found a correlation on day 2 ( $p = 0.028$ ) that strengthened when considering both days combined ( $p = 0.0012$ ). A two-way ANOVA examining the interaction between days and bouts showed that VAF-CI values were significantly lower during the NW phase compared to the first TW bout ( $p = 0.0085$ ), with this difference not persisting throughout subsequent TW bouts. Scores increased significantly during bouts 5 and 6 of TW compared to the initial bout ( $p = 0.0026$ ). Moreover, task performance improved between day 1 and day 2 ( $p < 0.001$ ).

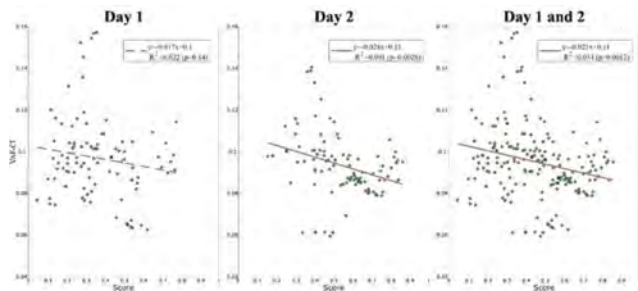


Figure: VAF-CI vs. task scores over two days. Solid lines indicate significant correlations ( $p < 0.01$ ); dashed lines indicate non-significant.

## Discussion

Performance in a novel lower limb motor task is related to the VAF-CI, a measure of complexity. Higher scores reflect better performance and lower complexity, while lower scores are associated with greater complexity. This may be caused by motor exploration as participants try numerous strategies to increase their score. High task performance on the other hand, typically achieved through significant training only, is related to low VAF-CI values. This may be linked to motor patterns becoming ingrained and tuned over the training period [3]. These results suggest that VAF-CI shows potential as a global indicator of movement complexity in motor learning studies for both upper and lower limb tasks.

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## Sitting radiographs may assist surgeons in selecting fusion levels for posterior spinal fusion in AIS

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## Introduction

Adolescent idiopathic scoliosis (AIS), is a 3D spinal deformity that affects one or more spinal segments (thoracic, thoracolumbar, lumbar), described by its Lenke type [1]. In moderate to severe

cases, where spinal alignment, function, or aesthetics are altered, surgical correction, most commonly posterior spinal fusion, is required to stop curve progression and restore alignment [2]. Fusion level selection in AIS typically relies on classical standing radiographs, where surgeons determine the upper (UIV) and lower instrumented vertebra (LIV) based on key vertebrae such as the last-touched (LTV), last-substantially touched (LSTV), neutral (NV) and stable vertebra (SV) [3]. While the standing radiograph represents a functional position, subjects may adopt other functional postures in daily life, such as sitting.

## Research Question

How can sitting radiographs alter key parameters used in the selection of the UIV and LIV?

## Methods

33 Lenke-1 (major thoracic) and 15 Lenke-5 (major thoracolumbar/lumbar) patients (average Cobb = 45 and 30° resp.) underwent low-dose biplanar X-rays in both standing and sitting positions, followed by 3D reconstruction of the spine and calculation of spinopelvic and scoliosis parameters. Key vertebrae for fusion level selection were identified: LTV, LSTV, NV, and SV, along with their tilt in the sagittal plane. Sagittal disc angles were also measured. Parameters were compared between the standing and sitting positions using the Wilcoxon paired t-test.

## Results

In Lenke-1 patients, 66% of LTV, 75% of LSTV, 54% of NV and 66% of SV changed when transitioning from standing to sitting (Figure 1). Moreover, 45% of patients changed their lumbar modifier, i.e. 77% of Lenke-1C in standing changed to Lenke-1B in sitting. The following parameters decreased with the change in position: sagittal tilt of NV (11 to 6°), sagittal tilt of SV (9 to 2°), and the sagittal disc angles: L3-L4 (9 to 5°), L4-L5 (16 to 6°, all  $p < 0.001$ ). In addition, apical vertebral rotation (AVR) of the thoracolumbar/lumbar (TL/L) and proximal thoracic (proxT) curves also decreased (AVR TL/L: 7 to 3° and AVR proxT = 10 to 6°, both  $p < 0.05$ ). In Lenke 5 patients, sagittal disc angles decreased at the levels of L2-L3 (9 to 5°), L3-L4 (12 to 5°), L4-L5 (12 to 7°), and sagittal distances between L3 and L4 to C7PL (5 to 0° both, all  $p < 0.05$ ).

## Discussion

Sitting is a key posture in daily life, yet postoperative evaluations rely only on standing radiographs, hiding sitting-induced biomechanical constraints that may contribute to long-term complications. This study highlights the need to integrate sitting radiographs into surgical planning, as key parameters used for UIV and LIV selection vary between positions. Therefore, relying only on standing radiographs may be insufficient, proving the necessity for further research into the clinical benefits of sitting radiographs for optimizing surgical outcomes.

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Figure 1 - Example of an AIS Lenke-1 patient (major thoracic curve) for whom key vertebrae changed between the standing and sitting radiographs.



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## Evaluation of Inter-Limb Jump Symmetry Performance in Adolescents with Idiopathic Scoliosis

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## Introduction

Adolescent idiopathic scoliosis (AIS) is a three-dimensional spinal deformity, defined by a lateral curvature exceeding 10 degrees on frontal plane radiographs(1). Depending on the type and severity of the curvature, the body's center of gravity may shift within the frontal plane, potentially affecting the use of the extremities and muscle performance. Individuals with AIS have higher gait asymmetry than controls(2). Inter-limb asymmetries in jump performance have been linked to an increased risk of sports injuries(3,4). However, there is limited information in the literature about jump performance and asymmetry in individuals with AIS.

## Research Question

Does trunk asymmetry influence jump performance and inter-limb jump symmetry in individuals with AIS.

## Methods

Thirty adolescents were included: 15 with AIS and 15 controls (mean age: AIS  $14.4 \pm 2.6$ ; control  $13.8 \pm 3$  years). Physical activity

was assessed using the Physical Activity Questionnaire for Older Children(PAQ-C; ages 11–14) and for Adolescents(PAQ-A; ages 15–18). Trunk asymmetry was measured from C7 to vertical lines drawn from the midpoint of both spina iliaca posterior superior(SIPS) on photographs. It was considered absent when the vertical line from C7 equally bisected the distance between the two SIPS. Jump tests included Double-Leg Vertical Jump(DLVJ), Single-Leg Vertical Jump(SLVJ), Single-Leg Horizontal Jump(SLHJ), and Triple Hop Distance (THD), with vertical jump flight time assessed via video. Flight time was analyzed using Kinovea software(version 2023.1.2). Inter-limb asymmetry was calculated using the bilateral asymmetry index(5), and all performance data were normalized to limb length. Statistical analysis used independent t-tests and Pearson's correlation.

## Results

The mean Cobb angle in the AIS group was  $23.6 \pm 11.3$  degrees. There were no significant differences between groups in age ( $p=0.53$ ), body mass index( $p=0.89$ ), or physical activity scores. The AIS group showed significantly lower performance in double-leg vertical jump and horizontal jump tests compared to controls ( $p<0.001$ ). Performances in DLVJ, SLVJ, SLHJ, and THD were also reduced in the AIS group(Table-1). No significant differences were observed in SLVJ or SLHJ asymmetry indexes; however, the THD asymmetry index was significantly higher in the AIS group ( $p=0.024$ ). Frontal plane trunk asymmetry was strongly correlated with DLVJ( $r=-0.698$ ), strongly correlated with the SLVJ asymmetry index( $r=0.602$ ), and moderately correlated with the THD asymmetry index( $r=0.452$ ).

Table 1: Vertical and Horizontal Jump Performance results and interlimb asymmetry indexes.

|                                       | AIS<br>(n=15) | Control<br>(n=15) | p     | Δ<br>(%) |
|---------------------------------------|---------------|-------------------|-------|----------|
| Double Leg Vertical Jump (sec.)       | 0,005±0,001   | 0,0056±0,001      | 0,001 | 12       |
| Single Leg Vertical Jump-Right (sec.) | 0,0035±0,001  | 0,0040±0,001      | 0,001 | 33.3     |
| Single Leg Vertical Jump-Left (sec.)  | 0,0033±0,001  | 0,0040±0,001      | 0,001 | 33.3     |
| Single Leg Vertical Jump (ILA%)       | 6,55±2,47     | 5,56±2,90         | 0,330 | 15.1     |
| Double Leg Horizontal Jump (cm)       | 1,67±0,41     | 2,21±0,39         | 0,001 | 32.3     |
| Single Leg Horizontal Jump-Right (cm) | 1,45±0,26     | 1,85±0,45         | 0,003 | 27.5     |
| Single Leg Horizontal Jump-Left (cm)  | 1,38±0,2      | 1,83±0,4          | 0,018 | 40.7     |
| Single Leg Horizontal Jump (ILA%)     | 6,71±3,44     | 5,49±4,22         | 0,382 | 18.1     |
| Triple Hop Distance-Right (cm)        | 4,19±0,78     | 5,26±1,42         | 0,003 | 25.5     |
| Triple Hop Distance-Left (cm)         | 4±0,90        | 5,23±1,33         | 0,027 | 30.7     |
| Triple Hop Distance (ILA%)            | 6,82±3,14     | 4,17±2,83         | 0,024 | 38.8     |

AIS: Adolescent Idiopathic Scoliosis, ILA: Interlimb Asymmetry Index, sec.: Second.

## Discussion

Despite similar physical activity levels, individuals with AIS exhibited significantly reduced performance in both vertical and horizontal jump tests compared to controls, while inter-limb asymmetry in both groups remained below injury risk thresholds reported in the literature(3). In individuals with AIS, inter-limb asymmetry indices were 15.1% higher for SLVJ, 18.1% higher for SLHJ, and 38.8% higher for THD compared to controls. This is the first study to report jump performance asymmetry in individuals with AIS,

highlighting the need for further research on its potential impact on sports injury risk.

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## Test-Retest Reliability of Gait Event Detection Methods Using Optical Motion Capture and a Single Sacral Inertial Measurement Unit

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## Introduction

Gait events, such as heel strikes and toe-offs, are critical for deriving spatiotemporal parameters and segmenting biological signals. Although force platforms and optical motion capture (OMC) systems are gold standards for detecting these events, inertial measurement units (IMUs) offer a cost-effective and portable alternative, making them particularly valuable for use beyond traditional laboratory environments [1]. However, there is a lack of studies evaluating the test-retest reliability of gait event detection methods, especially when using IMUs.

## Research Question

What is the test-retest reliability of gait event detection methods using optical motion capture or a single sacral IMU?

## Methods

Nine adults participated in two walking sessions, spaced at least seven days apart, during which they walked along a walkway and stepped on two embedded force plates. Foot kinematics were captured using an inertial measurement unit (IMU) placed on the sacrum and retro-reflective markers on the calcaneus and fifth metatarsus. Gait events were identified using two different methods: one based on optical motion capture (OMC) and one based on IMU data (IMU). OMC detected gait events via shifts in vertical marker velocity [2], while IMU employed wavelet-based signal differentiation [3]. Heel strikes and toe-offs identified by each method were compared to force plate data, which served as the reference standard. Intraclass correlation coefficients (ICCs) were calculated between the two testing sessions. Moreover, we calculated the absolute difference and mean absolute error (MAE) between sessions. The difference between each method on MAE and the absolute difference was assessed using one-way ANOVA (significance level at  $p < 0.05$ ).

## Results

The OMC method showed significantly lower error in detecting heel strikes ( $12.5 \pm 6.6$  ms) and toe-offs ( $42.5 \pm 64.0$  ms) compared to the IMU method (heel strikes:  $147.5 \pm 4.4$  ms;  $p < 0.001$ ; toe-offs:  $185.0 \pm 22.7$  ms;  $p < 0.001$ ), Figure 1. No significant differences were found between methods for absolute differences in heel strike ( $p = 0.342$ ) or toe-off detection ( $p = 0.317$ ), Figure 1. Test-retest reliability was high for OMC heel strikes (ICC = 0.80) but low for toe-offs (ICC = 0.22). IMU reliability was low for both events (heel strikes: ICC = 0.28; toe-offs: ICC = 0.48).

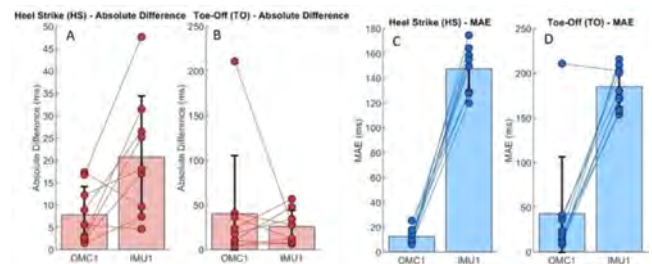


Figure 1. Absolute differences of heel strikes (A) and toe-offs (B) and mean absolute error between heel strikes (C) and toe-offs (D) between sessions.

## Discussion

The results showed that the OMC method had lower errors and greater reliability for heel strike detection than the IMU, consistent with its status as a more precise method. In contrast, both methods demonstrated limited reliability for toe-off detection, with particularly low consistency for OMC. This variability may stem from the more subtle nature of toe-off and inconsistent foot trajectories across sessions. While IMUs are advantageous for their portability and practicality, their reduced temporal accuracy, especially for heel strikes, suggests data analyses and/or acquisition setup improvements. Enhancing IMU-based methods could support their broader use in real-world gait assessments.

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## Mind in motion: lived experiences, systematic review and meta-analysis on how anxiety and/or depression changes the way we move

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## Introduction

Mental ill-health is a global issue<sup>1</sup>, with rising health costs and loss of quality of life<sup>2</sup>. Clinical diagnosis pathways for mental health are over-stretched, and current diagnostic tools rely on subjective and lengthy questionnaires. Therefore, there is a need to find novel diagnostic and monitoring tools. Movement may provide this tool. Embodiment theory proposes a feedback loop between motor processes and our emotions<sup>3</sup>, yet there is no synthesis on the link between mental health and our movements.

## Research Question

To assess the differences in biomechanical outcomes between mental well- and ill-health in anxiety and/or depression. To determine the involvement of service users in mental health and movement research.

## Methods

Following Cochrane and PRISMA guidelines, databases PubMed, EBSCOhost, Web of Science and CENTRAL were searched from inception until 26th April 2024. Searches were completed separately

for anxiety and depression. During the anxiety full-text sifting, only studies involving trait anxiety were included. Meta-analyses were run in R using random-effects model on movement outcomes ( $\geq 3$  articles).

## Results

Searches resulted in 4,392 and 6,249 anxiety and depression articles respectively. After 2 phases of sifting, 26 and 68 articles for anxiety and depression respectively were synthesised. A range of questionnaires were used to assess mental health state (anxiety = 9; depression = 14); the majority reported using a binary approach for non- vs anxious (77.8%) or depressed (88.2%) group. Most study populations included a co-morbidity (i.e. Parkinson's, knee osteoarthritis). Meta-analyses were performed on 21 outcome measures. Depressed populations had significantly poorer walking speed, stride length and stride time ( $p < 0.001$ ;  $p = 0.01$ ;  $p = 0.007$  respectively; Fig.1) compared to non-depressed. No other differences in spatio-temporal parameters were found for either anxiety or depression. Six-minute-walk, timed-up-and-go, and five-sit-to-stand tests were significantly poorer in depressed populations ( $p < 0.001$ ;  $p < 0.001$ ;  $p = 0.038$  respectively). No studies incorporated service user involvement.

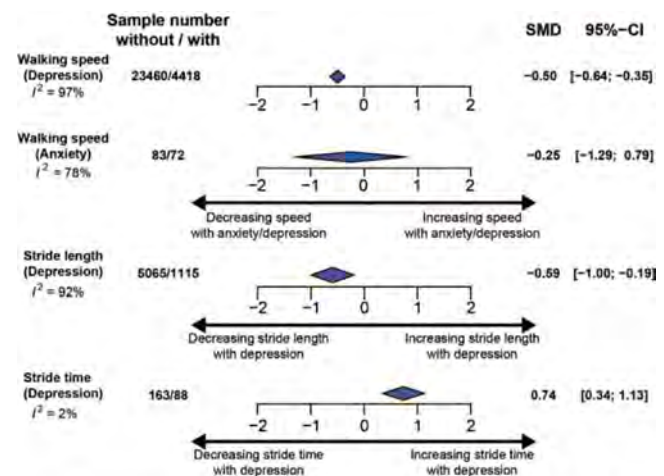


Figure 1: Overall effects from meta-analyses for movement parameters for anxiety or depression. Sample represents the total numbers for those without depression or anxiety and then the number for those with anxiety or depression. For depression meta-analyses showed consistent poorer performance in movement outcomes compared to those without depression.

## Discussion

We show that depression impacts our movement by decreasing walk speed through stride length reduction and stride time increase. Reducing walk speed is likely a protective response, as seen in older adults and neurodegenerative conditions e.g.4,5. There is a scarcity of data to determine any other specific changes in biomechanics due to mental ill-health. To date, the focus within anxiety has been on situational and temporal anxiety (state). Trait anxiety studies tended to include co-morbidities and exhibited dichotomised findings from limited papers. Future work should focus on understanding how trait anxiety impacts movement, as state and trait anxiety alter biological processing differently<sup>6</sup>. To use biomechanics as a novel diagnosis and monitoring tool, we should ensure consistent use of mental health questionnaires and biomechanical outcomes to clarify the impact of mental ill-health. Moreover, further work needs to incorporate service users to ensure relevancy for those with lived experience of mental health.

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## Effects of functional electrical stimulation and time on gait in short- and long-term users with neurological impairments

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## Introduction

Functional Electrical Stimulation (FES) is widely used to improve gait function in subjects with neurological impairments and foot drop <sup>1</sup>. FES stimulates intact peripheral nerves and muscles to restore motor function <sup>2</sup>. While benefits have been reported <sup>3</sup> the initial effects may not always be beneficial for overall foot position <sup>4</sup>.

## Research Question

Standard healthcare evaluation periods may not fully capture the potential of FES, particularly for adults with congenital neuropathies. Therefore, this study evaluates the effects of FES and Time on gait function in short-term users during the insurance covered test phase and long-term users.

## Methods

This within-subject design included two groups: G1, short-term users starting their treatment and G2, long-term users of at least 12 months of treatment time. Both groups underwent 3D gait analysis at baseline (T1) and after 4 to 12 weeks (T2). The outcomes assessed included ankle kinematics, kinetics, and spatio-temporal parameters. Statistical analyses used linear mixed models and t-tests, with significance set at  $p < 0.05$ .

## Results

In G1, there was a significant time-related shift toward a more plantarflexed foot position during stance ( $p = 0.017$ ) and swing ( $p = 0.002$ ). Walking speed significantly improved ( $p = 0.019$ ), and a trend toward reduced step width was observed. No significant FES-related changes in ankle kinematics were found ( $p > 0.527$ ).

**Table 1.** P-values of fixed effects of linear mixed model for gait variables of short (G1) and long-term users (G2) with TIME and FES as fixed effects and subjects as random effects.

| n=11 (kinetics n=10)              | G1    |       |          |
|-----------------------------------|-------|-------|----------|
|                                   | TIME  | FES   | TIME*FES |
| <i>Kinematics and kinetics</i>    |       |       |          |
| GPS [°]                           | 0.516 | 0.616 | 0.886    |
| Max. dorsiflexion stance [°]      | 0.017 | 0.661 | 0.922    |
| Dorsiflexion at heel-strike [°]   | 0.052 | 0.375 | 0.933    |
| Max. dorsiflexion swing [°]       | 0.002 | 0.527 | 0.800    |
| Push off energy [J/kg]            | 0.101 | 0.167 | 0.244    |
| <i>Spatio-temporal parameters</i> |       |       |          |
| Speed [m/s]                       | 0.019 | 0.569 | 0.251    |
| Step length [m]                   | 0.072 | 0.518 | 0.532    |
| Step width [cm]                   | 0.071 | 0.428 | 0.656    |
| n=9                               | G2    |       |          |
|                                   | TIME  | FES   | TIME*FES |
| <i>Kinematics and kinetics</i>    |       |       |          |
| GPS [°]                           | 0.827 | 0.817 | 0.985    |
| Max. dorsiflexion stance [°]      | 0.001 | 0.084 | 0.698    |
| Dorsiflexion at heel-strike [°]   | 0.593 | 0.015 | 0.386    |
| Max. dorsiflexion swing [°]       | 0.356 | 0.038 | 0.251    |
| Push off energy [J/kg]            | 0.006 | 0.071 | 0.364    |
| <i>Spatio-temporal parameters</i> |       |       |          |
| Speed [m/s]                       | 0.082 | 0.013 | 0.295    |
| Step length [m]                   | 0.273 | 0.004 | 0.198    |
| Step width [cm]                   | 0.122 | 0.223 | 0.143    |

FES: functional electrical stimulation, GPS: gait profile score

Long-term users (G2) demonstrated FES-related improvements in ankle kinematics and gait function, particularly in dorsiflexion at heel strike and during swing as well as in walking speed and step length. While G1 showed time-related changes including worsening of foot position alongside improved spatio-temporal parameters, G2 mainly showed FES-related improvements in both kinematics and spatio-temporal parameters (Table 1).

## Discussion

The previously observed differences between G1 and G2 indicate that the effects of FES differ based on treatment stage. In short-term users, foot position worsened initially while walking speed improved, suggesting that the full therapeutic benefit of FES may not be realised in early stages of treatment. This is particularly relevant for adults with congenital neurological disorders, who may require extended adaptation periods. Long-term users demonstrated more consistent FES-related improvements in ankle kinematics,

especially dorsiflexion, which is essential for stable gait and toe clearance. Improvements in spatio-temporal parameters further support enhanced overall gait function with continued FES use.

These findings suggest that short-term users may not experience the full benefits of FES within standard evaluation windows commonly used by insurance companies of the healthcare system. The initial changes, particularly worsening foot positioning, require further analysis. Extending treatment periods based on individual neurological conditions and severity may be necessary to fully evaluate the therapeutic potential of FES for newly treated subjects with foot drop due to neurological conditions.

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## Impact of mild leg length discrepancy on pelvic alignment and gait compensation in children

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## Introduction

Leg length discrepancy (LLD) is common in children and often leads to pelvic lateral tilt as a primary gait deviation. While children rarely report low back pain, prolonged LLD can induce lasting changes in lumbar spine biomechanics according to the Hueter-Volkman Law, potentially resulting in future back issues [1]. To mitigate the effects of LLD, children often employ various compensatory mechanisms. However, the degree of LLD that significantly impacts gait remains unclear.

## Research Question

This study aims to identify gait deviations in children and adolescents with mild LLD and examine the relationship between LLD severity and lower limb compensatory strategies. We

hypothesized that pelvic lateral tilt would be significantly correlated with the degree of LLD, and that ankle dorsiflexion of the long leg would be the primary compensatory mechanism to shorten the longer limb, also significantly correlated with the LLD.

## Methods

Fifty-one children and adolescents (mean age 12.6 years, SD = 2.5) with mild LLD ranging from 0.0 to 3.0 cm were enrolled. Exclusion criteria included neurological involvement, scoliosis, syndromes. Patients were also excluded if they had joint contractures exceeding 2 standard deviations (SD) from typically developing controls or muscle strength below 4 on a five-point scale. These exclusions were based on clinical examination of range of motion and strength parameters. Pelvic motion and gait compensations were evaluated across three groups: LLD 0-1 cm, LLD 1-2 cm, LLD 2-3 cm. Correlations between gait parameters and LLD were assessed.

## Results

Pelvic lateral tilt showed significant correlations with LLD severity. Kinematic analyses revealed significant joint adaptations on the long leg side, leading to functional shortening, particularly in the 2-3 cm group, which displayed multiple and clear compensations. Notably, only ankle dorsiflexion on the long side significantly correlated with LLD.

## Discussion

This study confirms both initial hypotheses: pelvic lateral tilt and ankle dorsiflexion of the longer leg are significantly correlated with LLD. Even in mild cases, children and adolescents exhibited clear compensatory patterns, primarily through functionally shortening the longer leg. This was achieved via increased ankle dorsiflexion, greater knee flexion, and reduced hip extension—adaptations that help reduce vertical height during stance. Conversely, toe-walking to lengthen the shorter leg was not observed, likely due to its higher energetic demands and the exclusion of participants with ankle contractures. Despite these compensations, pelvic obliquity remained evident, suggesting these mechanisms are insufficient for full correction. Additional adaptations in pelvic rotation and tilt further indicate complex multi-planar responses to LLD. These biomechanical changes increase asymmetrical loading on the spine and joints, raising the risk of overuse injuries. Clinically, the findings underscore the need to assess even mild LLD ( $\geq 1-2$  cm), as targeted interventions like shoe lifts can alleviate compensatory strain and reduce related symptoms.

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## The Biomechanical Impact of Joint Axis Misalignment in Hinged Ankle-Foot Orthoses

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### Introduction

Ankle-foot orthoses (AFOs) are essential for supporting gait and joint stability, especially in individuals with neuromuscular impairments. In hinged AFOs, the alignment of the orthotic joint axis is usually based on manufacturer guidelines rather than patient-specific biomechanics. Misalignment may alter joint loading, restrict ankle range of motion, and cause pistoning—shaft-to-leg movement that can reduce comfort and effectiveness [1]. This study examines how deviations in joint axis positioning affect the mechanical performance and user interaction with AFOs.

### Research Question

This study investigates the biomechanical effects of anterior-posterior and proximal-distal shifts in the joint axis of bilateral hinged AFOs. It explores whether anterior-posterior shifts more strongly influence ankle plantarflexion moments than proximal-distal shifts, and whether all deviations induce pistoning and restrict ankle range of motion relative to a shod condition.

### Methods

Six healthy adult males ( $35 \pm 12$  years) participated in gait trials under six conditions: five bilateral individually fitted AFO configurations with joint axis positions varied (anterior, posterior, proximal, distal, and neutral) and one shod condition. Kinematic and kinetic data were captured using a 3D motion analysis system. Outcomes included ankle joint moments, ankle range of motion, and pistoning between the orthotic shaft and leg. Additionally, musculoskeletal simulations were performed in OpenSim using a modified, individually scaled Gait2392 model. In this setup, the foot was rigidly attached to the orthosis while the shaft was allowed to move, enabling estimation of internal ankle joint moments across different alignment conditions

### Results

Anterior-posterior axis shifts significantly influenced ankle joint mechanics. Anterior alignment generated the smallest dorsiflexion moments and reduced pistoning during dorsiflexion. Proximal-distal shifts had minimal effect on ankle joint moments but resulted in increased pistoning. All orthotic conditions demonstrated reduced ankle range of motion compared to the shod condition. The neutral axis position, as recommended by the manufacturer, did not consistently minimize pistoning or preserve range of motion. Simulated internal joint moments showed no significant differences across the various axis alignments.

### Discussion

This study demonstrates that anterior-posterior alignment of AFO joint axes has a more pronounced effect on ankle joint function than proximal-distal positioning. Anterior axis alignment reduced pistoning and enhanced joint kinetics, suggesting clinical benefits in terms of comfort and gait efficiency. The neutral alignment, while commonly recommended, did not provide superior mechanical performance. Proximal-distal variations had limited biomechanical relevance but did increase shaft-to-leg movement. Musculoskeletal simulations, though useful, were limited by the fixed-foot modeling assumption, which may have underestimated push-off dynamics, and indicates a necessity of precise interaction modelling between foot and orthosis. Overall, personalized joint axis alignment that closely matches anatomical motion may improve AFO outcomes by preserving joint mechanics, reducing pistoning, and enhancing user comfort.

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### Investigating fast walking speed and walking speed reserve in patients before total knee arthroplasty

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### Introduction

Patients with knee osteoarthritis (KOA) typically walk slower than age-matched healthy individuals [1]. The ability to increase walking speed is crucial for daily functioning and safety [2]. Fast Walking Speed (FWS) and Walking Speed Reserve (WSR)—the difference between self-selected preferred walking speed (PWS) and FWS—are important indicators of functional mobility and adaptability in real-life situations, such as crossing streets, catching a bus, or climbing stairs safely [3].

## Research Question

What are the FWS and WSR values in patients with KOA awaiting total knee arthroplasty (TKA), and how do these metrics relate to clinical factors such as pain, patient-reported outcomes (PROMs), and muscle strength?

## Methods

Thirty-nine patients with end-stage KOA scheduled for primary TKA (20 women, 51%; mean age:  $69.3 \pm 8.2$  years; BMI:  $30.6 \pm 6.8$  kg/m<sup>2</sup>) underwent full-body barefoot gait analysis on a 10 m walkway before surgery. Each participant completed 5–10 trials at PWS and 2 trials at FWS ("as fast as possible without running"). WSR was calculated as the difference between FWS and PWS. PROMs included pain (10 cm VAS), KOOS subscales, and normalized strength of the quadriceps, hamstrings, and hip adductors (N/kg). Paired t-tests assessed differences between PWS and FWS, and Pearson's correlations explored relationships between walking speeds (PWS, FWS and WSR), pain, PROMs, and muscle strength ( $p < 0.05$ ).

## Results

Patients demonstrated a significant increase in walking speed from PWS ( $0.9 \pm 0.2$  m/s) to FWS ( $1.3 \pm 0.3$  m/s;  $p < 0.05$ ) (Figure 1), with a mean WSR of  $0.39 \pm 0.19$  m/s. Spatiotemporal parameters were significantly higher at FWS compared to PWS, primarily due to increased cadence ( $126.5 \pm 16.1$  steps/min) and, to a lesser extent, stride length ( $1.2 \pm 0.2$  m). PWS and FWS showed moderate negative correlations with pain (VAS) ( $r = -0.38$  and  $r = -0.32$ , respectively;  $p < 0.05$ ) and moderate to strong positive correlations with normalized muscle strength ( $r = 0.37$ – $0.56$ ;  $p < 0.05$ ). WSR showed a moderate positive correlation with PWS ( $r = 0.32$ ) and a strong positive correlation with FWS ( $r = 0.79$ ) but also a moderate correlation with hamstrings and adductor normalized strength ( $r = 0.43$  and  $r = 0.40$ , respectively).

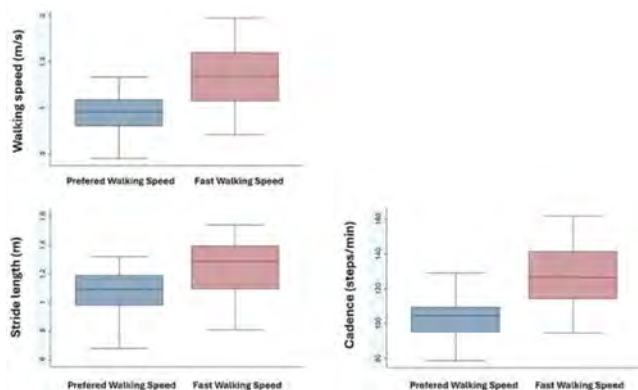


Figure 1: Illustration of walking speed, stride length and cadence for Preferred Walking Speed and Fast Walking Speed

## Discussion

Patients with KOA demonstrated the capacity to increase their walking speed. The correlations showed the impact of the pain on the capacity to increase walking speed and the importance of muscle strength in functional mobility. The results support the clinical value of assessing FWS and WSR to better capture functional mobility and physical reserve beyond habitual walking performance. Incorporating fast walking assessments in routine evaluations may enhance the understanding of mobility impairments and inform pre- and post-surgical rehabilitation strategies.

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## ☆Gait patterns according to the CPAK classification in patients with end-stage knee osteoarthritis

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## Introduction

The Coronal Plane Alignment of the Knee (CPAK) classification provides a framework for categorizing functional knee phenotypes [1]. Knee alignment is increasingly studied in primary total knee arthroplasty (pTKA), since phenotypes influence knee kinematics and overall movement patterns [2, 3].

## Research Question

How do different CPAK knee phenotypes influence gait kinematics and compensatory movement patterns in patients with end-stage knee osteoarthritis?

## Methods

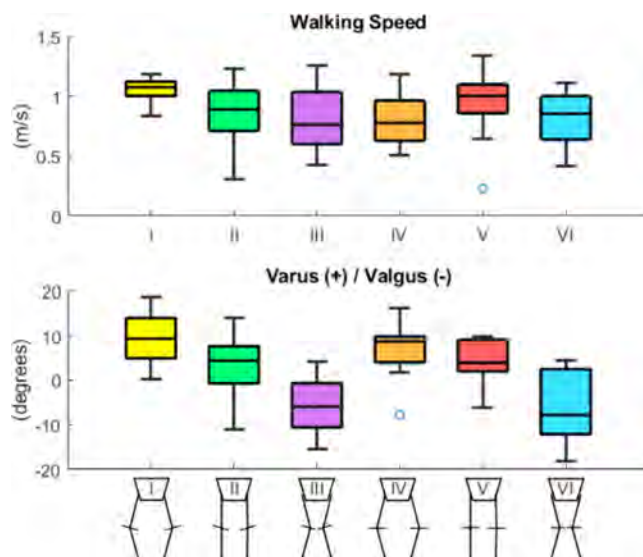
We included 91 patients with end-stage knee OA planned for pTKA (53 women (58%), mean (SD) age:  $69.3$  ( $8.2$ ) years, BMI:  $31.2$  ( $6.3$ ) kg/m<sup>2</sup>, WOMAC pain  $44.1$  ( $16.7$ ) WOMAC function  $48.5$  ( $16.8$ ), SF12 mental  $45.3$  ( $11.6$ ), SF12 physical  $35.6$  ( $6.7$ )). Patients

underwent full-body clinical gait analysis (CGA) before surgery. From the CGA, spatiotemporal and kinematic parameters were analyzed, including the three degrees of freedom of thorax and pelvis, hip flexion-extension and adduction-abduction angles, knee flexion and varus-valgus during the stance phase, as well as ankle flexion-extension and foot progression angles. CPAK phenotypes were determined from long-leg axis X-rays taken preoperatively. Group differences based on CPAK phenotypes were assessed using ANOVA for continuous variables with post-hoc Bonferroni tests applied when significant ( $p < 0.05$ ). Chi-squared tests were used to evaluate categorical variables.

## Results

The CPAK types analyzed were I ( $n = 17$ , 4 women), II ( $n = 26$ , 13 women), III ( $n = 10$ , 9 women), IV ( $n = 9$ , 6 women), V ( $n = 16$ , 9 women), and VI ( $n = 12$ , 12 women). Knee varus-valgus angles during stance phase closely aligned with the CPAK classification, showing significant differences between valgus and varus knee groups (type I:  $5.1^\circ \pm 5.9$  vs. type III:  $-8.3^\circ \pm 5.1$ ;  $p < 0.01$  and type I vs. type VI:  $-13.7^\circ \pm 7.5$ ;  $p < 0.01$ , and type II:  $1.1^\circ \pm 5.3$  vs. type III;  $p < 0.01$  and type II vs. type VI;  $p < 0.01$ ) (Figure 1). Type IV exhibited full-body compensations, including increased trunk inclination, greater pelvic obliquity, and restricted hip adduction and ankle flexion-suggesting antalgic gait or weakness of hip abductors. Types I had the greatest step length (significant differences with types II and VI) and the highest gait velocity (significant differences with types II and III). No significant differences were observed in spatio-temporal parameters and in pre-surgery PROMs.

Figure 1: Illustration of walking speed and knee varus/valgus angle during gait for each CPAK type.



## Discussion

The most notable differences were in knee varus-valgus kinematics, with type IV exhibiting the most pronounced full-body compensations, suggesting greater functional impairments. These findings underscore the relationship between CPAK type and function, which could aid surgical planning and post-operative

expectations. Larger studies are needed to confirm these results. Future research should assess gait one year post-TKA using updated CPAK types to clarify kinematic changes and the impact of alignment strategies. In particular, it is questionable whether restoring pre-arthritis phenotypes improves gait and within which boundaries.

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## Can pelvic protraction contribute ipsilateral knee flexion in stance during walking in children with increased femoral anteversion ?

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## Introduction

Increased femoral anteversion, as a sub-pelvic lever arm dysfunction, can alter lower extremity alignment, leading to increased internal hip rotation, pelvic rotation, and knee flexion during mid and terminal stance phases[1–3]. Pelvic rotation occurs in all three planes during typical gait and significantly influences lower extremity biomechanics[4,5]. In the extremity where the pelvis is protracted during walking, hip and knee biomechanics may change, though this is not yet clearly defined in the literature. This effect may be more pronounced in children with increased femoral anteversion, potentially increasing knee flexion on the protracted side compared to the retracted side. Comparing knee flexion between extremities may enhance our understanding of lower limb biomechanics in this population.

## Research Question

Does forward rotation of the pelvis, due to a sub-pelvic reason, increase knee flexion in stance during walking?: Increased Femoral Anteversion.

## Methods

The study included 15 children with Increased Femoral Anteversion (IFA) (male/female:5/10,  $10.80 \pm 1.74$  y.o., TPAT:  $40.29^\circ \pm 8.66^\circ$ ) and 15 typically developing controls (male/female:9/6,  $10.53 \pm 2.32$  y.o., TPAT:  $16.53^\circ \pm 5.39^\circ$ ), totaling 60 lower extremities. Each extremity was analyzed separately and categorized by pelvic position during stance: Protracted vs. Retracted. Gait

kinematics were assessed using 3D motion analysis (VICON Plug-in Gait Model) [6]. Participants with neuro-orthopedic conditions (e.g., tibial torsion, scoliosis) were excluded. Kinematic parameters were compared between groups using paired samples t-tests.

## Results

Gait velocities were similar across all groups ( $p > 0.05$ ). In both groups, protracted limbs showed significantly greater pelvic internal rotation during stance compared to retracted limbs. No significant inter-limb differences were found in sagittal or frontal pelvic kinematics ( $p > 0.05$ ). However, only the IFA group showed increased knee flexion at toe-off ( $p < 0.05$ ) (Table 1).

**Table 1:** The statistical comparison of all interested walking parameters.

| Kinematic Parameters (°)            | CONTROLS                        |                                  | IFA GROUP                       |                                  |
|-------------------------------------|---------------------------------|----------------------------------|---------------------------------|----------------------------------|
|                                     | Retracted Pelvis Side (Mean±SD) | Protracted Pelvis Side (Mean±SD) | Retracted Pelvis Side (Mean±SD) | Protracted Pelvis Side (Mean±SD) |
| Pelvic Internal Rotation @ Ic       | 4.05 ± 3.59                     | 9.71 ± 3.51*                     | 5.02 ± 3.28                     | 10.01 ± 3.31*                    |
| Mean Pelvic Internal Rotation @ Lr  | 2.29 ± 3.40                     | 7.89 ± 3.39*                     | 3.36 ± 2.64                     | 5.58 ± 1.90*                     |
| Mean Pelvic Internal Rotation @ Mst | -0.49 ± 2.95                    | -4.86 ± 2.88*                    | 0.73 ± 2.71                     | 5.32 ± 2.59*                     |
| Mean Pelvic Internal Rotation @ St  | -3.12 ± 2.62                    | 2.02 ± 2.55*                     | -2.62 ± 2.46                    | 1.97 ± 2.11*                     |
| Max. Pelvic Internal Rotation @ St  | 4.30 ± 3.46                     | 9.88 ± 3.49*                     | 5.62 ± 2.59                     | 10.45 ± 3.16*                    |
| Max. Anterior Pelvic Tilt @ St      | 12.44 ± 4.41                    | 11.79 ± 4.46                     | 11.73 ± 4.01                    | 11.49 ± 4.00                     |
| Max. Pelvic Elevation @ St          | 5.32 ± 2.15                     | 4.87 ± 1.88                      | 5.63 ± 2.10                     | 6.42 ± 2.30                      |
| Knee Flexion @ Ic                   | 1.70 ± 4.78                     | 2.37 ± 4.56                      | 5.91 ± 5.74                     | 4.07 ± 4.40                      |
| Mean Knee Flexion @ Lr              | 10.38 ± 7.03                    | 12.15 ± 5.75                     | 15.80 ± 7.29                    | 14.58 ± 4.13                     |
| Max. Knee Flexion @ Lr              | 14.61 ± 8.41                    | 16.83 ± 6.63                     | 20.59 ± 8.35                    | 19.01 ± 4.64                     |
| Mean Knee Flexion @ Mst             | 8.28 ± 5.87                     | 9.61 ± 4.62                      | 12.62 ± 6.02                    | 11.42 ± 4.32                     |
| Max. Knee Flexion @ Mst             | 14.73 ± 8.37                    | 16.89 ± 6.54                     | 21.27 ± 7.39                    | 19.74 ± 4.55                     |
| Mean Knee Flexion @ St              | 10.06 ± 5.66                    | 10.97 ± 4.09                     | 14.20 ± 4.98                    | 14.03 ± 4.20                     |
| Max. Knee Flexion @ Toe-off         | 32.89 ± 10.94                   | 33.25 ± 7.05                     | 36.01 ± 6.22                    | 42.05 ± 6.70*                    |
| Max. Knee Extension @ St            | 0.17 ± 4.51                     | 0.69 ± 3.88                      | 0.55 ± 4.98                     | 0.88 ± 4.72                      |

Significantly different parameters: \* $p < 0.05$  (IFA groups: Retracted Pelvis Side vs. Protracted Pelvis Side), \* $p < 0.05$  (Controls: Retracted Pelvis Side vs. Protracted Pelvis Side), Max.: Maximum, Ic: Initial contact, Lr: Loading Response, Mst: Midstance, St: Stance.

## Discussion

Increased pelvic protraction may affect the kinematics of the knee joint, leading to an increase in the knee flexion angle during the loading response phase[7]. This increased knee flexion can alter the loading behavior on the knee joint and change the joint moments, potentially increasing the risk of osteoarthritis and anterior knee pain[8]. This study found that children with increased femoral anteversion in the limb with the pelvis in the protraction side exhibited higher knee flexion angles compared to the limb with the pelvis in retraction during walking. Studies on pelvis and knee biomechanics suggest that limited range of motion in the pelvis could lead to restricted sagittal plane movement in the knee as well [9,10]. Although these findings are agreed with the literature which made a connection with increased femoral anteversion, asymmetrically increased knee flexion may be more related with the asymmetric pelvic rotation during walking. Therefore, the effects of pelvic rotational alterations should be distinguished from the alterations related to increased femoral anteversion in order to improve the treatment plan especially for the individuals with femoral anteversion and supra-pelvic originated transverse plane pelvic asymmetries such as scoliosis.

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## Investigation of knee flexion effect of ipsilateral pelvic drop at stance for individuals with lumbar scoliosis during walking.

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## Introduction

Scoliosis is a three-dimensional deformity defined by a lateral spinal curvature over 10° in the frontal plane. Idiopathic scoliosis, with unknown cause, can disrupt pelvic alignment, leading to abnormal pressure and torsion on the pelvic girdle. In closed-chain kinetics, spinal deformities may alter lower extremity dynamics via pelvic displacement. Supra-pelvic misalignment may elevate knee injury risk in sports due to altered biomechanics [1–3]. However, no study has explored the link between coronal pelvic motion and sagittal knee kinematics during walking. This study aimed to investigate that relationship.

## Research Question

How does coronal pelvic motion alteration affect the sagittal knee flexion during the stance phase of walking in patients with lumbar scoliosis?

## Methods

This study included two children (male/female:1/1, 10.66 ± 0.94y.o., Cobb angle ≥ 40°) diagnosed with Adolescent Idiopathic Scoliosis (AIS) characterized by lumbar S-type curves and dominant coronal pelvic alterations. A total of four lower extremities were evaluated independently based on pelvic position in all three anatomical planes during the stance phase. Pelvic sides were classified as Anterior/Posterior (sagittal), Upper/Lower (frontal), and Protracted/Retracted (transverse). Gait kinematics were assessed using a 3D motion analysis system (VICON Plug-in Gait Model) [4]. Participants with additional neuro-orthopedic conditions (e.g.,

tibial torsion) were excluded. Paired samples t-tests were used to compare kinematic parameters across pelvic positions.

## Results

Gait velocities were similar across participants ( $p > 0.05$ ). In the frontal plane, pelvic elevation was significantly higher on the upper side ( $p < 0.05$ ), where both loading response and peak stance knee flexion were also increased. No significant differences were found in sagittal or transverse pelvic kinematics between anterior and posterior sides ( $p > 0.05$ ) (Table 1).

**Table 1: The statistical comparison of all interested walking parameters.**

| Kinematic Parameters (°)           | AIS GROUP                   |                              |
|------------------------------------|-----------------------------|------------------------------|
|                                    | Protracted Pelvic Side      | Retracted Pelvic Side        |
|                                    | Mean ± Sd                   | Mean ± Sd                    |
| <b>Transvers Plan</b>              |                             |                              |
| Mean Pelvic Internal Rotation @ Gc | -0.01 ± 2.15                | -0.34 ± 1.09                 |
| Mean Pelvic Internal Rotation @ St | 0.81 ± 2.50                 | 1.04 ± 2.44                  |
| <b>Frontal Plan</b>                | <b>Upper Pelvic Side</b>    | <b>Lower Pelvic Side</b>     |
|                                    | Mean ± Sd                   | Mean ± Sd                    |
| Pelvic Elevation @ Ic              | 3.92 ± 1.28                 | 0.79 ± 0.97                  |
| Mean Pelvic Elevation @ Gc         | 1.81 ± 0.57*                | -2.12 ± 0.36                 |
| Max. Pelvic Elevation @ Gc         | 6.62 ± 1.49                 | 2.53 ± 0.16                  |
| Mean Pelvic Elevation @ St         | 2.14 ± 0.88                 | -1.47 ± 0.47                 |
| Max. Pelvic Elevation @ St         | 6.62 ± 1.49                 | 2.53 ± 0.16                  |
| Mean Pelvic Elevation @ Lr         | 5.56 ± 1.45                 | 1.78 ± 0.18                  |
| Mean Pelvic Elevation @ Mst        | 3.58 ± 1.46                 | -0.15 ± 0.19                 |
| Mean Knee Flexion @ Gc             | 59.10 ± 1.46                | 51.75 ± 7.22                 |
| Max. Knee Flexion @ Gc             | 18.82 ± 1.14                | 17.73 ± 2.14                 |
| Mean Knee Flexion @ St             | 8.32 ± 1.02                 | 10.08 ± 0.77                 |
| Max. Knee Flexion @ St             | 30.53 ± 6.37                | 34.87 ± 10.42                |
| Max. Knee Extension @ St           | -1.68 ± 0.33*               | 1.52 ± 0.10                  |
| Knee Flexion @ Ic                  | 9.02 ± 0.29                 | 3.93 ± 3.50                  |
| Max. Knee Flexion @ Mst            | 14.84 ± 0.59                | 13.52 ± 0.92                 |
| Mean Knee Flexion @ Mst            | 7.64 ± 0.12                 | 8.38 ± 1.25                  |
| Max. Knee Flexion @ Lr             | 14.84 ± 0.59                | 13.52 ± 0.92                 |
| Mean Knee Flexion @ Lr             | 12.39 ± 0.38*               | 10.47 ± 0.25                 |
| <b>Sagittal Plan</b>               | <b>Anterior Pelvic Side</b> | <b>Posterior Pelvic Side</b> |
|                                    | Mean ± Sd                   | Mean ± Sd                    |
| Mean Anterior Pelvic Tilt @ Gc     | 6.89 ± 0.83                 | 6.25 ± 1.74                  |
| Mean Anterior Pelvic Tilt @ St     | 7.36 ± 0.24                 | 5.97 ± 1.74                  |

Significantly different parameters: \* $p < 0.05$ , Max.: Maximum, Ic: Initial contact, Lr: Loading Response, Mst: Midstance, St: Stance, Gc: Gait cycle.

## Discussion

Scoliosis is a three-dimensional deformity of the spine that can directly affect pelvic alignment and lower extremity movements. Idiopathic scoliosis can cause alterations in muscles near the spine, particularly decreasing the lumbar muscle strength, which can lead to spinal and lumbopelvic instability [5]. This study demonstrated that selectively supra-pelvic originated pelvic drop may contribute higher knee flexion in stance phase which may put extra load on knee extensor mechanism [6]. On the other hand in the loading response phase, knee flexion at the up side is more than the dropped

side which may load the knee extensor mechanism eccentricity more on the upper pelvis side. In the clinic, SIAS level (Spina-Iliaca-Anterior-Superior) which represents static pelvic drop and knee flexion in stance participation should be considered. Supra-pelvic originated ipsilateral pelvic drop may contribute anterior knee pain in time and may increase the knee injury risk on sportive activities which should be investigated in detail by a higher number of participants and by including full body kinematics and kinematics. Establishing a plane-specific connection between the spine, pelvis and lower extremity biomechanics can help to understand corrected-plane and improved-joint kinematics relations during walking in treatment plan.

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## Can a treadmill tell us more than the eye? Quantitative gait assessment using C-Mill in clinical practice

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## Introduction

Quantitative gait analysis is a valuable tool for detecting gait asymmetries in children with Cerebral Palsy (CP), yet its clinical use is often limited by the need for specialized equipment and facilities<sup>1,2</sup>. In recent years, technological advancements have opened the possibility for new tools capable of offering meaningful biomechanical insights within more accessible clinical environments. This case study investigates the use of the C-Mill treadmill, a rehabilitation device equipped with force platform and virtual reality features, to conduct a spatiotemporal gait assessment in a child with unilateral CP.

## Research Question

Can the C-Mill treadmill provide clinically meaningful gait asymmetry data that can be used to inform and personalize rehabilitation strategies?

## Methods

We assessed an 8-year-old boy with left unilateral CP (Gross Motor Function Classification System – GMFCS level I; Manual Ability Classification System – MACS level II) using the C-Mill VR + (Motek Medical, Culemborg, the Netherlands). After 10 minutes of familiarization, two walking trials of 2 minutes each were performed, resulting in approximately 150 gait cycles per trial. A custom Python script extracted key spatiotemporal gait parameters based on gait events automatically detected by the treadmill. After checking data distribution, a Wilcoxon-sign test ( $\alpha = 0.05$ ) was performed for each parameter, comparing right and left sides.

## Results

Significant asymmetries were found in multiple gait parameters, most notably in double support, stance and swing time. These differences were consistent across both trials. Table 1 reports average values for the more affected (left) and less affected (right) limbs and p-values. Notably, the more affected side exhibited a shorter stance phase (62.4% vs 73.8%,  $p < 0.001$ ) and longer swing phase (37.6% vs 26.1%,  $p < 0.001$ ), suggesting instability or compensatory strategies during weight acceptance. The reduction in double support time on the affected side (15.4% vs 20.8%,  $p < 0.001$ ) may indicate reduced confidence or impaired control in bilateral stance phases. In contrast, step length and stride duration showed no significant differences between sides ( $p = 0.788$  and  $p = 0.812$  respectively), suggesting relatively preserved spatial symmetry despite temporal alterations.

| Parameter           | Right (less affected) | Left (more affected) | p-value |
|---------------------|-----------------------|----------------------|---------|
| Step length (m)     | 0.33 (0.04)           | 0.33 (0.04)          | 0.788   |
| Stance duration (s) | 1.03 (0.15)           | 0.87 (0.09)          | < 0.001 |
| Stance duration (%) | 73.85 (3.99)          | 62.43 (4.88)         | < 0.001 |
| Swing duration (s)  | 0.37 (0.08)           | 0.53 (0.10)          | < 0.001 |
| Swing duration (%)  | 26.15 (3.99)          | 37.57 (4.88)         | < 0.001 |
| Double support (s)  | 0.29 (0.05)           | 0.21 (0.05)          | < 0.001 |
| Double support (%)  | 20.82 (3.74)          | 15.43 (3.14)         | < 0.001 |
| Stride duration (s) | 1.40 (0.17)           | 1.39 (0.14)          | 0.812   |

Table 1: Gait spatiotemporal parameters. Mean (SD).

## Discussion

This case study highlights the potential of the C-Mill treadmill as a practical and accessible tool for objective gait analysis in a clinical setting. The automated detection of gait events and compatibility with custom scripts make it possible to quantify gait asymmetry without the need for a full motion analysis lab. Early identification of asymmetry can support the development of individualized rehabilitation strategies aimed at pattern normalization. This approach may help bridge the gap between research-grade analysis and everyday clinical decision-making.

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## Speaking the Same Gait Language? Inter-Professional Agreement Between Physiotherapists and Pediatric Orthopedic Surgeons on Edinburgh Visual Gait Score Assessments

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## Introduction

Accurate gait assessment in cerebral palsy (CP) is essential for determining treatment strategies. The Edinburgh Visual Gait Score (EVGS) provides a standardized framework for assessing gait deviations visually. However, inter-observer variability between surgeons and physiotherapists may impact the agreement in assessment and decision-making in CP. This study aimed to address this point, exploring whether these groups share a consistent evaluative "language" when using the EVGS to assess gait in children with cerebral palsy.

## Research Question

Can pediatric orthopedists and physiotherapists working with CP have the same language while doing visual gait assessment with using EVGS?

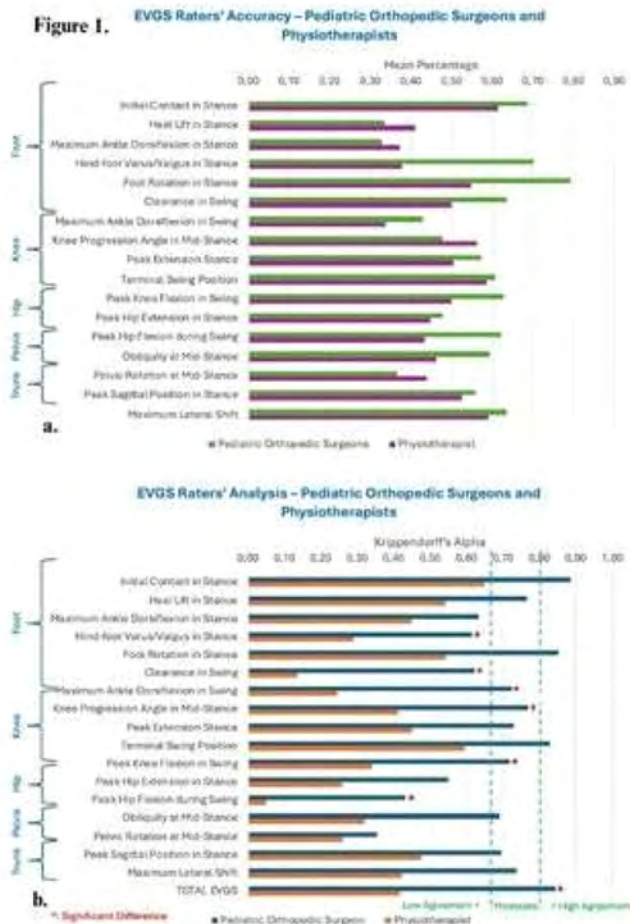
## Methods

This study was conducted among physiotherapists and pediatric orthopedic surgeons who had the same gait assessment training. The gait data of 10 children and the standardized EVGS scoring form were shared with professionals via email. The collected data were analyzed to evaluate both intergroup agreement between the two professional groups and intragroup consistency within each group and the accuracy level for each group.

## Results

21 physiotherapists and 15 pediatric orthopedic surgeons participated in this study. The overall accuracy of the total EVGS was 55.5% for pediatric orthopedic surgeons, and 48.3% for physiotherapists (Figure 1a.). Pediatric orthopedic surgeons exhibited high-moderate agreement (Krippendorff's Alpha coefficients over 0.70) for parameters of distal segments (foot and knee). However, IRR for pelvic and hip parameters showed lower agreement with

Krippendorff's Alpha coefficients below 0.70. Physiotherapists demonstrated "low agreement" for all items and the total score, with IRR values below 0.67 (Figure 1b.).



## Discussion

This study highlights significant differences in EVGS scoring accuracy and reliability between pediatric orthopedic surgeons and physiotherapists, emphasizing the influence of professional expertise, training and experience. While variability persists between professional groups, the EVGS remains a valid and valuable tool for evaluating gait deviations in children with cerebral palsy, supported by its good concurrent validity with 3D gait analysis and sensitivity to detect clinical changes. Clinicians who deal with gait abnormalities can use the score early in their practice knowing that scoring accuracy improves with training. Current study findings may suggest the importance of continued education and practice in maximizing the clinical and research potential of the EVGS.

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## Evaluating ChatGPT-4o's Performance in Gait Pattern Classification Using Sagittal Kinematic Data and Video-Based Inputs in Children with Cerebral Palsy

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## Introduction

Sagittal gait classification plays a vital role in understanding and managing gait abnormalities in children with spastic diplegia. With recent advancements in artificial intelligence, using tools like ChatGPT are increased for decision-making and diagnosis in the medical area (1). This study aimed to evaluate the classification performance of ChatGPT-4o in identifying gait patterns in children with spastic diplegia according to Rodda's sagittal gait classification system, using both kinematic gait graphs and 3D gait videos as input sources (2).

## Research Question

Can ChatGPT-4o accurately classify spastic diplegic gait types according to Rodda's sagittal gait classification system when provided with either kinematic gait graphs or 3D gait videos?

## Methods

Fifteen patients representing all five Rodda gait types (three patients per type) were included in the study. Three physiotherapists and 3 orthopedists established ground truth classifications based on the sagittal plane kinematic data and the 3D videos by consensus. ChatGPT-4o was asked to classify each patient twice: once just using gait graphs and once just using 3D gait model videos. In terms of patient privacy, normal gait videos were not shared. The patient data is given in a random order. The AI-generated classifications were compared to the ground truth using accuracy, Cohen's kappa, and confusion matrices. The confusion matrices are given in Figure 1.

## Results

Graph-based predictions achieved an accuracy of 66.7% with a Cohen's kappa of 0.58, indicating moderate-to-good agreement with experts' classification. In contrast, video-based predictions showed a lower accuracy of 40% and a kappa of 0.25, reflecting lower agreement. Per-class analysis revealed that Type 4 was correctly

predicted across all settings, while Type 25 was frequently misclassified in the video-based condition. Comparison between ChatGPT-4o's own predictions from graphs and videos showed an internal consistency of 46.7% (kappa = 0.32).

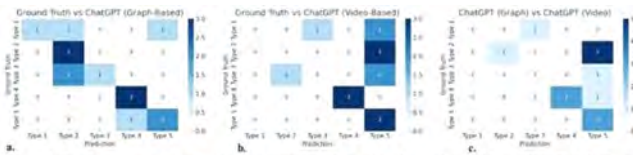


Figure 1. The confusion matrices for a. Ground Truth vs ChatGPT (Graph-Based Classification), b. Ground Truth vs ChatGPT (Video-Based Classification), c. ChatGPT (Graph) vs ChatGPT (Video).

## Discussion

ChatGPT-4o demonstrates a promising ability to classify gait patterns when provided with structured kinematic graphs, whereas its performance significantly decreases when interpreting 3D video data. This highlights the importance of data format in the effectiveness of AI-based clinical tools. It is also the first study in the literature that evaluates the using an AI tool in the classification of gait.

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## Understanding the Gait Kinematics and Functionality in an 8-Year-Old Children with Freeman-Sheldon Syndrome

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## Introduction to clinical case

Freeman-Sheldon syndrome (FSS) is an exceedingly rare congenital disorder with a reported prevalence of less than 1 per 1 million. It is also known as Arthrogryposis Multiplex Type IIA which is a rare type of the disease. This syndrome is characterized by contractures on different body parts, which limits the functionality and ambulation level of the child, but the literature on this issue is limited (1). There is only one study in the literature reporting that these children have a stiff knee gait. However, this study didn't comprehensively evaluate gait parameters (2). The resulting musculoskeletal deformities may require multiple orthopedic treatments, and having more information about the gait characteristics and functionality might help increase the success of the treatment methods. This study aims to highlight the gait characteristics and the functionality of a case

with FSS. This study will be the first study showing the gait kinematics and the functionality of a patient with FSS.

## Methods

We report a case of an 8-year-old girl who was diagnosed with FSS in infancy due to contractures of the extremities and a peculiar facial appearance. She underwent instrumental gait analysis for the lower extremity and the functional gait parameters were evaluated by using the Gillette Functional Assessment Questionnaire 22-item Walking Scale (GFAQ-WL), Gross Motor Function Measurement (GMFM) D (standing) and E (walking) parts, the 1 Minute Walk Test (1MWT) and Time Up & Go (TUG) Test. She had the measurement with Statik Ankle foot Orthosis because she cannot walk independently without it.

## Results

Gait kinematics results in all planes are given in Figure 1. The cadence was 115 steps/min and the walking speed was 0.57 m/s. The patient got 2,25 points from the GFAQ-WL and showed moderate walking function. The score for TUG was 8.3 seconds and for 1MWT was 42,6 meters. While the GMFM D score was 51,3%, the GMFM E score was 48,6%. All the parameters show that the patient has moderate walking function.

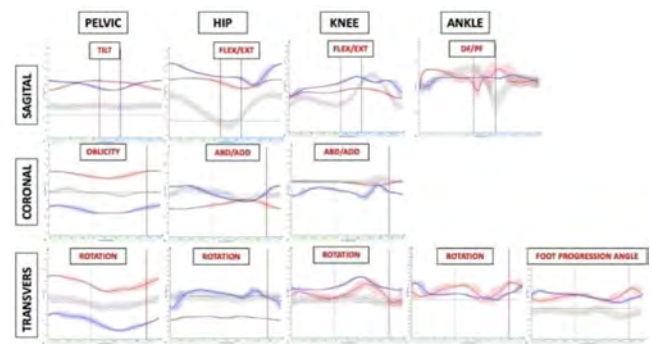


Figure 1. Gait kinematics of lower extremity in three planes.

## Conclusion and clinical relevance

It is the first study to show the gait results of a case with FSS. It may help to understand the gait characteristics and the functional level of the patients with this rare disease. The gait analysis results show that the patient has lots of asymmetries and compensations. That's why the instrumented gait analysis might be essential for decision-making and the management of musculoskeletal disorders in this patient group.

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## Assessing Forward Propulsion and Propulsion Efficiency and Their Association with Ankle Dorsiflexion Range of Motion in Idiopathic Toe Walking

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### Introduction

Idiopathic toe walking (ITW) is a gait disorder where children persistently walk on their toes without identifiable causes [1]. ITW children often exhibit a “bouncy” gait [2] with excessive vertical motion. This study aims to evaluate whether forward propulsion and propulsion efficiency are impaired in an ITW population, and how this relates to the degree of plantarflexor contracture.

### Research Question

1. Do children with ITW exhibit compromised forward propulsion and altered propulsion strategies compared to typically developing (TD) peers? 2. Are inefficient propulsion patterns associated with a restricted range of ankle dorsiflexion

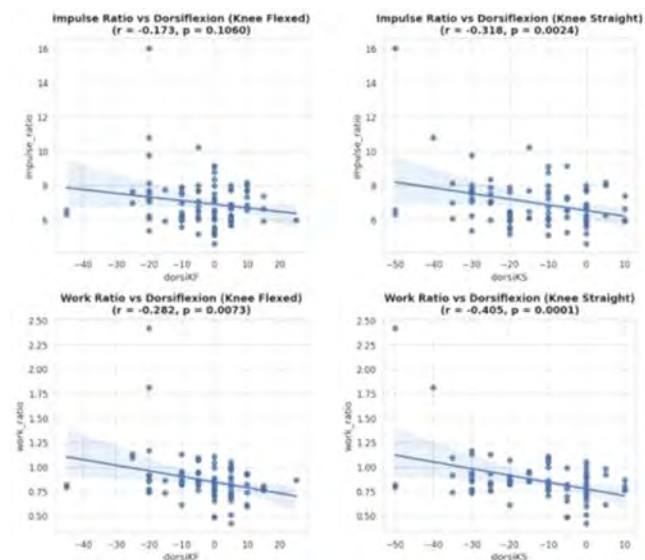
### Methods

The study retrospectively analysed gait data from 45 children with ITW (mean age =  $10.11 \pm 3.32$ ) and 45 age-matched TD peers (mean age =  $12.24 \pm 3.57$ ). Overall gait performance was assessed through normalised gait speed, step length, and relative double support duration. COM dynamics, including normalised peak-to-peak vertical displacements, and mechanical work [3] in the anterior-posterior (AP) and vertical directions during the propulsive phase [4] were assessed to characterise propulsion pattern. Propulsion efficiency was evaluated using the vertical/AP work ratio. Propulsion strategy was examined by analysing AP and vertical ground reaction force (GRF) impulses [4], and their ratio used to identify directional bias. Group differences were evaluated using independent t-tests ( $\alpha = 0.05$ ). Passive ankle dorsiflexion measured in the Silfverskiöld test was correlated with propulsion efficiency metrics using Pearson coefficients in the ITW group to explore the impact of contracture on propulsion.

### Results

Children with ITW had reduced normalised gait velocity and step length ( $p < 0.001$ ), as well as increased relative double support phase ( $p = 0.0021$ ). The ITW group had higher peak-to-peak vertical displacement and vertical mechanical work ( $p < 0.001$ ). AP work did not differ significantly between groups, and the vertical/AP work ratio was higher in ITW ( $p < 0.001$ ) in the ITW children. GRF analysis revealed higher AP and vertical impulses ( $p < 0.001$ ), while

the AP/vertical impulse ratio was not significantly different ( $p = 0.08$ ). Ankle dorsiflexion ROM showed significant negative correlations with work ratio ( $p < 0.001$ ) and impulse ratio ( $p = 0.002$ ) with the knee extended. With the knee flexed, work ratio remained significant ( $p = 0.007$ ), but impulse ratio did not ( $p = 0.10$ ) (Fig 1).



### Discussion

Children with ITW walked more slowly, spent longer in double-support, and exhibited propulsion pattern biased toward vertical rather than forward motion. The association between these inefficiencies and reduced passive ankle dorsiflexion indicates the role of plantarflexor contracture in limiting forward progression. The resulting increase in mechanical demand may lead to reduced endurance and pain, supporting the necessity for intervention in symptomatic cases. Propulsion metrics offer objective indicators of functional impairment, with potential to guide treatment decisions, and their use in post-intervention assessment could offer insight into the treatment success.

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## Analysis of Toe Clearance Mechanisms in Idiopathic and Voluntary Toe Walking

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### Introduction

Idiopathic toe walking (ITW) is defined as persistent toe walking beyond the age of three without identifiable musculoskeletal or neurological causes [1]. ITW is associated with increased tripping risk [2] and altered toe clearance [3]. The toe clearance strategies adopted by toe walkers remain poorly understood. This study investigates the toe clearance mechanism in an ITW population and compares them with typically developing (TD) children during both normal walking (TDW) and voluntary toe walking (TDTW).

### Research Question

1. Do children with ITW achieve comparable toe clearance to typically developing peers? 2. What biomechanical strategies do children with ITW use to achieve toe clearance, and do these differ compared to TDW and TDTW?

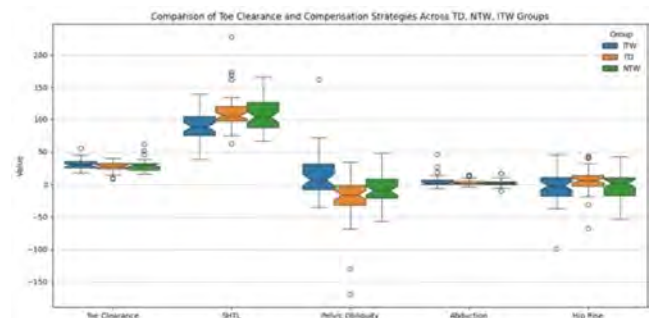
### Methods

The study included 45 children with ITW (mean age = 10.11 ± 3.32) and 45 age-matched TD peers (mean age = 12.24 ± 3.57). TD participants walked both normally (TDW) and on toes (TDTW), while ITW participants only walked on toes. To assess the amount of toe clearance, the vertical displacement of the toe from mid-stance to mid-swing was calculated. Four biomechanical contributors to clearance, based on the framework proposed by Matsuda et al. (2016) [4], including functional limb shortening (SHTL), pelvic obliquity, hip abduction, and contralateral hip elevation were also computed. Group comparisons were performed using one-way ANOVA and post-hoc Tukey HSD tests where significant effects were found.

### Results

No significant group differences were found in the amount of toe clearance ( $p = 0.1087$ ); however, differences in clearance components were identified (Fig. 1). Children with ITW showed reduced functional limb shortening and increased pelvic obliquity compared to both TDW ( $p = 0.0001$ ;  $p < 0.0001$ ) and TDTW groups

( $p = 0.0015$ ;  $p = 0.0050$ ), with no significant difference between TDW and TDTW groups. Contralateral hip rise was lower in ITW compared to TDW ( $p = 0.0117$ ), but it did not differ significantly from TDTW ( $p = 0.6536$ ). No significant group differences were found in hip abduction ( $p = 0.2318$ ).



### Discussion

The increased pelvic obliquity observed in the ITW group, which was absent in the TDTW group, suggests that is not simply a biomechanical consequence of absent heel contact, warranting exploration into other potential contributors to this altered pelvic strategy. Similarly, the reduced SHTL observed may be associated with restricted ankle dorsiflexion, which is commonly reported in ITW, although this relationship remains to be clarified. Although all three groups achieved similar toe clearance levels, different strategies were adopted. While such adaptation may effectively preserve toe clearance in controlled lab settings, increased reliance on proximal compensation may reduce endurance, and excessive pelvic tilt could contribute to discomfort or low back pain over time, which may hinder overall gait function and increase risk of falling during prolonged walking in real-world environments.

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## Relationship between thoracolumbar/lumbar scoliotic deformity and hip rotational mobility in adolescent idiopathic scoliosis: a kinematic evaluation

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### Introduction

Adolescent idiopathic scoliosis (AIS) is a three-dimensional deformity of the spine defined as a multifactorial condition involving anatomical, biomechanical and neuromuscular factors [1]. The Nottingham concept is one of many principles that had reported the interaction between the pelvis and hip in the pathogenesis and compensation of the scoliosis, especially in the presence of a thoracolumbar/lumbar deformity (Lenke-5 type of scoliosis) [2]. This concept suggests that scoliosis is associated with an abnormal pelvic and hip rotation during gait, leading to an increased femoral anteversion, coupled with thorax and vertebral rotations. However, this concept has never been investigated quantitatively.

### Research Question

What is the relationship between the range of motion (ROM) of hip rotation and the thoracolumbar/lumbar scoliotic deformity?

### Methods

16 AIS Lenke-5 patients (14F, Cobb =  $29 \pm 12^\circ$  with left convexity) and 19 control subjects (12F) underwent low-dose biplanar X-rays and physical examination to measure hip internal (IR) and external (ER) rotation bilaterally. In addition, 3D movement analysis was performed during gait, trunk anatomical exercises (flexion, extension, lateral bending and axial rotation) and sit-to-stand. Kinematic parameters of the head, trunk, spinal segments, pelvis and lower limbs were extracted [3,4]. Parameters were compared between the 2 groups.

### Results

On the physical exam, Lenke-5 patients presented a reduced external rotation in both hips compared to controls (left hip:  $27 \pm 12^\circ$  vs.  $36 \pm 10^\circ$ ; right hip:  $26 \pm 12^\circ$  vs.  $37 \pm 13^\circ$ , both  $p < 0.05$ ). During the trunk left rotation exercise, an increased lateral bending towards the concavity (right side) of the scoliosis was noted in the T12L3-L3L5 segment ( $7 \pm 7^\circ$  vs.  $-1 \pm 7^\circ$ ;  $p = 0.001$ ). During the trunk left bending exercise, the lateral bending of the T12L3-L3L5 segment towards the convexity (left side) was reduced ( $-10 \pm 6^\circ$  vs.  $-17 \pm 6^\circ$ ). IR/ER ratio of the left hip was negatively correlated to the bending of T12L3-L3L5 segment in both trunk left rotation and lateral bending exercises (both  $r = -0.5$ ,  $p < 0.05$ ).

### Discussion

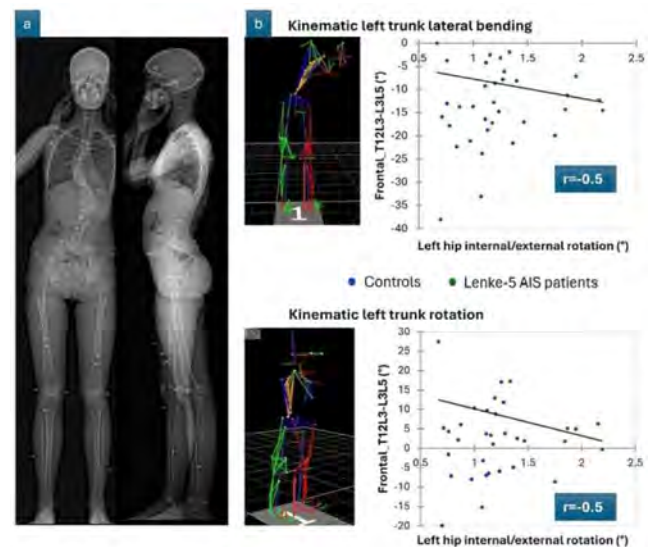
This is the first study to provide quantitative data, based on 3D spinal kinematic analysis supporting the Nottingham concept in Lenke-5 AIS patients. The current findings revealed a relationship between decreased external rotation of the hip with a reduced

mobility of the scoliotic segment towards the concave side during trunk rotation and towards the convexity during trunk lateral bending. This might underline an association between the hip asymmetrical ROM and the spinal kinematic alterations in Lenke-5 patients. Strategies such as postural education and rehabilitation to gain hip external rotation at an early age may reduce functional impairment in AIS patients.

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Figure 1: a) Example of AIS Lenke-5 patient. b) Correlation between IR/ER ratio and T12L3-L3L5 segment kinematics during left trunk lateral bending and rotation.



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### Smartphone accelerometry as a tool for assessing bradykinesia in Parkinson's disease

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## Introduction

Bradykinesia is one of the core motor symptoms of Parkinson's disease and is a critical component in diagnosis. The Unified Parkinson's Disease Rating Scale (UPDRS) is the gold standard for bradykinesia assessment. Studies have explored the use of accelerometers for objective quantification of bradykinesia, often employing standalone devices (1). Given their widespread availability, smartphones offer a promising alternative by leveraging built-in accelerometers for bradykinesia assessment.

## Research Question

Does smartphone-based accelerometer data correlate with bradykinesia severity in patients with Parkinson's disease, and can it effectively differentiate patients from healthy controls?

## Methods

Leg movements of 39 idiopathic patients with Parkinson's disease and six healthy controls were measured with a smartphone accelerometer. For the measurement, the phone was placed proximal to the knee joint and fixed with a belt; then, participants were asked to walk for 10 seconds. The Phyphox (RWTH Aachen University, Germany) application obtained the accelerometer data (2). Velocity and displacement are calculated using the trapezoid rule with integral calculations. Centralization was performed by subtracting the average from the raw acceleration data. The data passed through a 1-4 Hz band-pass filter, and the velocity and displacement were calculated from the filtered and centralized data. Microsoft Excel (Microsoft Corporation) and QTIPlot (ver. 1.2.4.) programs were used to perform mathematical calculations. The accelerometer used is the ICM-20608-G model of the InvenSense brand. UPDRS and Hoehn and Yahr scale assessed disease and symptom severity. Correlations between variables were calculated with the appropriate Pearson/Spearman analysis. The ROC curve was used to distinguish participants with and without bradykinesia.

## Results

Significant correlations were obtained between leg acceleration and Hoehn-Yahr stage, UPDRS total score, section 3 and bradykinesia items. The correlation between walking bradykinesia and leg acceleration was  $r = -0.641$  ( $p < 0.001$ ). The correlation analysis results are given in Table 1. Leg acceleration data provided feasible results in evaluating bradykinesia standing up from a chair. The area under the ROC curve in the prediction model of bradykinesia severity in standing up from a chair was calculated as 0.890 for the amount of displacement obtained from both the raw acceleration and the filtered data. While the displacement calculated from the filtered data was predicted with a sensitivity of 0.759 and a specificity of 0.937, the amount of displacement obtained from the raw acceleration was predicted with a sensitivity of 0.897 and a specificity of 0.750.

Picture 1. "Correlation analysis results"

|                             | Raw Acceleration Average | Raw Acceleration Peak | Raw Velocity | Raw Displacement | Velocity (Centralized) | Displacement (Centralized) | Velocity (Filtered) | Displacement (Filtered) | Velocity (Filtered Centralized) | Displacement (Filtered Centralized) |
|-----------------------------|--------------------------|-----------------------|--------------|------------------|------------------------|----------------------------|---------------------|-------------------------|---------------------------------|-------------------------------------|
| Distance                    | -0.000                   | -0.005                | 0.002        | 0.021            | -0.119*                | 0.105                      | 0.002               | 0.021                   | -0.107                          | -0.108                              |
| Duration                    | -0.700                   | -0.111                | -0.250       | -0.213           | -0.614*                | -0.145                     | -0.240              | -0.219                  | -0.610*                         | -0.029                              |
| H&Y Stage                   | -0.412*                  | -0.494*               | -0.420*      | -0.362*          | -0.312                 | -0.113                     | -0.420*             | -0.107*                 | -0.601*                         | -0.290                              |
| UPDRS Total                 | -0.434*                  | -0.467*               | -0.419*      | -0.382*          | -0.607*                | -0.095                     | -0.419*             | -0.300*                 | -0.500*                         | -0.220                              |
| UPDRS Section 3             | -0.425*                  | -0.433*               | -0.404*      | -0.385*          | -0.605*                | -0.108                     | -0.404*             | -0.303*                 | -0.643*                         | -0.303                              |
| UPDRS 3.4 - 3.11 Free Turn  | -0.400*                  | -0.400*               | -0.391*      | -0.362*          | -0.215                 | -0.107                     | -0.391*             | -0.334*                 | -0.101                          | -0.147                              |
| Head Rest Tremor            | -0.360*                  | -0.407*               | -0.314       | -0.293           | -0.312                 | -0.084                     | -0.314              | -0.283                  | -0.316                          | -0.295                              |
| Action Tremor               | -0.371*                  | -0.377*               | -0.311       | -0.305           | -0.276                 | -0.171                     | -0.311              | -0.288                  | -0.303*                         | -0.267                              |
| Head Rigidity               | -0.100                   | -0.117                | -0.104       | -0.110           | -0.200*                | -0.241                     | -0.104              | -0.110                  | -0.200*                         | -0.074                              |
| Foot Rigidity               | -0.240                   | -0.299                | -0.218       | -0.254           | -0.080*                | -0.008                     | -0.218              | -0.214                  | -0.080*                         | -0.130                              |
| Finger Tap Bradykinesia     | -0.409*                  | -0.370*               | -0.363*      | -0.347*          | -0.502*                | -0.101                     | -0.363*             | -0.329*                 | -0.514*                         | -0.221                              |
| Head Movement Bradykinesia  | -0.444*                  | -0.520*               | -0.408*      | -0.401*          | -0.440*                | -0.276                     | -0.408*             | -0.402*                 | -0.440*                         | -0.385*                             |
| Proximal Bradykinesia       | -0.370*                  | -0.400*               | -0.433*      | -0.419*          | -0.281                 | -0.233                     | -0.433*             | -0.440*                 | -0.500*                         | -0.177                              |
| Foot Movement Bradykinesia  | -0.250                   | -0.283                | -0.250       | -0.240           | -0.467*                | -0.110                     | -0.250              | -0.228                  | -0.511*                         | -0.210                              |
| Stand to Stand Bradykinesia | -0.321*                  | -0.303*               | -0.301*      | -0.301*          | -0.189                 | -0.425*                    | -0.301*             | -0.303*                 | -0.221                          | -0.110                              |
| Walking Bradykinesia        | -0.401*                  | -0.433*               | -0.404*      | -0.391*          | -0.510*                | 0.021                      | -0.404*             | -0.300*                 | -0.641*                         | -0.300*                             |
| Arm Swing Bradykinesia      | -0.202                   | -0.434*               | -0.282       | -0.293           | -0.470*                | -0.111                     | -0.282              | -0.277                  | -0.390*                         | -0.205                              |

\* Statistical significance, H&Y Hoehn and Yahr staging, UPDRS Unified Parkinson's Disease Rating Scale

## Discussion

When subjected to signal processing and mathematical modeling, smartphone-based accelerometer data are significantly correlated with clinical measures of bradykinesia severity. These data can accurately distinguish between individuals with and without bradykinesia, especially during walking and functional tasks such as standing up from a chair. The findings support the potential use of smartphones as accessible tools for objective bradykinesia assessment in clinical and remote settings.

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## Using smartphone accelerometer data to predict lower limb bradykinesia, tremor, and rigidity in Parkinson's disease

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## Introduction

Smartphones are increasingly utilised in the assessment of movement disorders due to their accessibility, affordability, and versatility for both patients and healthcare providers [1]. In this study, we investigated whether smartphone-based accelerometer data could be used to predict the presence of bradykinesia, tremor, and rigidity in patients with Parkinson's disease.

## Research Question

Can lower limb bradykinesia, tremor, and rigidity be predicted using accelerometer-derived movement data in patients with Parkinson's disease?

## Methods

Accelerometer data were collected from 23 patients with idiopathic Parkinson's disease during routine outpatient visits. A smartphone equipped with an Invensense ICM-20608-G accelerometer was secured at the hypogastric region using a belt. Participants were instructed to perform a sit-to-stand task. Data were recorded using the Phyphox app (RWTH Aachen University, Germany) [2]. Raw acceleration data were processed by centralising (subtracting the mean) and applying a 1–4 Hz band-pass filter. Velocity and displacement were calculated using trapezoidal integration of raw, centralised, and filtered signals. Data processing and analysis were conducted using Microsoft Excel and QTIPlot (v1.2.4). Disease severity was assessed using the Modified Unified Parkinson's Disease Rating Scale (UPDRS), and the disease stage was determined via the Hoehn and Yahr scale. ROC curve analysis was employed to assess the discriminative ability of acceleration-derived metrics and to determine cut-off values for predicting clinical symptoms.

## Results

Acceleration metrics did not reliably predict walking or postural bradykinesia. However, lower extremity tremors and rigidity were effectively identified. The strongest result was for lower extremity rigidity, predicted using velocity derived from centralised acceleration data, yielding 87.5% sensitivity and 86.7% specificity (AUC = 0.850, 95% CI: 0.656–1.000,  $p = 0.007$ ). Detailed results are provided in table.

Picture 1- "Significant results obtained from the ROC curve analysis"

| Model                                        | UPDRS               | Cut-off | AUC   | 95% CI        | Sensitivity | Specificity | p     |
|----------------------------------------------|---------------------|---------|-------|---------------|-------------|-------------|-------|
| Velocity (From centralized raw acceleration) | Lower limb rigidity | 0.0026  | 0.850 | 0.656 – 1.000 | 0.875       | 0.867       | 0.007 |
| Raw acceleration average                     | Lower limb rigidity | 0.8457  | 0.775 | 0.538 – 1.000 | 0.875       | 0.667       | 0.033 |
| Velocity (From centralized raw acceleration) | Lower limb tremor   | 0.0026  | 0.841 | 0.673 – 1.000 | 0.727       | 0.917       | 0.006 |
| Raw acceleration peak                        | Lower limb tremor   | 4.1061  | 0.833 | 0.646 – 1.000 | 0.818       | 0.833       | 0.007 |
| Velocity (From raw acceleration)             | Lower limb tremor   | 4.0892  | 0.818 | 0.632 – 1.000 | 0.727       | 0.833       | 0.010 |
| Velocity (From filtered raw acceleration)    | Lower limb tremor   | 4.0879  | 0.818 | 0.632 – 1.000 | 0.727       | 0.833       | 0.010 |
| Raw acceleration average                     | Lower limb tremor   | 0.9713  | 0.788 | 0.587 – 0.989 | 0.636       | 0.917       | 0.019 |

AUC area under curve, CI confidence interval, UPDRS Unified Parkinson's Disease Rating Scale.

## Discussion

Our findings suggest that smartphone accelerometry during a sit-to-stand task can accurately detect lower extremity tremors and rigidity in Parkinson's disease. However, contrary to our hypothesis, this method was ineffective in identifying bradykinesia. Enhancing such accessible and low-cost measurement tools may contribute to early diagnosis and improved monitoring of Parkinsonian symptoms [3].

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## The co-design and validation of a lab-based, multi-task protocol to evaluate fatigue across daily living activities: pilot results

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## Introduction

Fatigue is a prevalent and functionally limiting issue in older adults and individuals post-stroke. Within the WHO's Healthy Ageing framework, fatigue is recognized as a key attribute of vitality capacity—an essential component of intrinsic capacity [1]. Among stroke survivors, fatigue is common and significantly reduces quality of life [2]. Fatigue may be general or activity-based [3], and its multidimensional nature complicates objective assessment. Current fatigue protocols often rely on isolated, repetitive tasks that lack ecological validity and fail to reflect the complex, challenging nature

of everyday life. A new approach is needed to assess real-world fatigue in a standardized yet meaningful way.

## Research Question

This study aims to iteratively develop and validate a lab-based protocol that reliably induces fatigue through performance of lifelike tasks. Success criteria include: (1) progressive fatigue is experienced by participants, and (2) severe fatigue is reached within one hour.

## Methods

The protocol is developed using a user-centred, iterative design approach, incorporating input from healthy older adults, older adults with mobility limitations, and stroke survivors. Initial task selection was guided by focus group discussions, which identified challenging and fatiguing activities [4]. These insights informed the protocol's design, which integrates five key elements known to influence real-world walking: speed, slopes/steps, surface, directional path change, and cognitive demand [5]. Participants perform a continuous sequence of demanding tasks at a self-selected pace, ending when they (i) reach a predefined fatigue threshold of  $>7$  on the modified Borg scale ("very hard exertion" [6]), (ii) stop voluntarily, or (iii) reach the one-hour mark. The protocol allows individual adaptations (e.g., use of mobility aids) to accommodate varying physical abilities. Participant feedback on fatigue, realism, and safety is used to iteratively refine the protocol between sessions.

## Results

Eleven participants contributed to the iterative refinement of the protocol: three healthy older adults (1M/2F; age range:65–89), three older adults with mobility limitations (1M/2F; age range:80–89), and five chronic stroke survivors (4M/1F; age range:35–69; Functional Ambulation Categories: 3–5). One healthy participant did not experience steady fatigue build-up, as their exertion scores plateaued (Fig.1A). As a result, subject-driven adjustments were introduced, enabling the other two healthy subjects to experience progressive fatigue. All other participant groups consistently reported increasing exertion over time (Fig.1B–1C), with task durations ranging from 10:50 to 51:41 minutes (Fig.1D). Feedback confirmed the protocol's realism and safety.

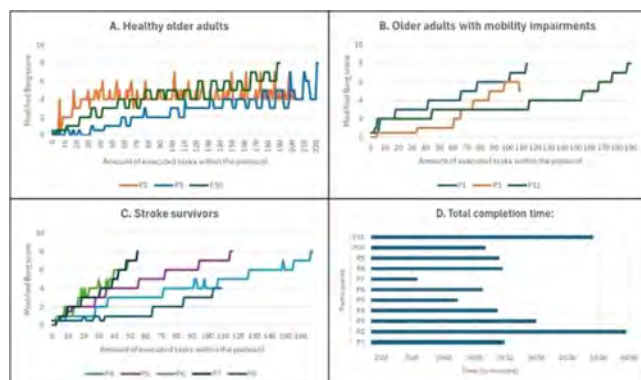


Fig. 1. Participant-rated perceived exertion (modified Borg score) during protocol execution (A–C) and total completion time (D). Note: Protocol versions potentially varied between participants (see Methods).

## Discussion

Preliminary findings support the protocol's potential to elicit progressive, real-life fatigue using ecologically valid tasks. Future steps include validation with larger cohorts and collection of normative data. Ultimately, this protocol may standardize fatigue assessment and enable evaluation of interventions—such as lower-limb exoskeletons—designed to mitigate fatigue in daily life [7].

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## Identifying knee extensor thrust in stroke patients using shank kinematics

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## Introduction

Kinematic analysis makes it possible to define specific gait abnormalities, which are common in stroke patients [1]. Focusing on the knee, the most well-known is stiff-knee gait, which primarily affects the swing phase. During stance, the focus is often on hyperextension, quantified by the maximum knee extension angle. Another typical gait abnormality in this population is the extensor thrust [2]. This is usually observed visually and indirectly through the slope of knee flexion during stance, measurable via the minimum angular velocity of the knee. We propose a new indicator based on the angular velocity of the shank. From heel strike onwards, the shank is expected to move continuously forward during stance. In patients with extensor thrust, the shank briefly moves backward, resulting in a negative angular velocity that distinguishes them from other patients.

## Research Question

Is the minimum angular velocity of the shank a good indicator to detect extensor thrust? Are shank segment kinematics more effective than knee joint kinematics during stance?

## Methods

We recruited 22 post-stroke patients, divided into two groups (with and without extensor thrust). Each performed 8 trials at self-selected speed and 8 at fast speed along a 12-meter walkway. A 3D motion capture system (Qualisys, Miquis 3) recorded the trajectories of 26 anatomical markers, placed according to the Institute of Rizzoli (IOR) marker set, natively implemented in the Qualisys suite. Knee and shank angular velocities were computed using Visual3D (C-Motion). Minimum angular velocity was calculated between 0–40% of the gait cycle. A univariate logistic regression evaluated the ability of shank and knee indicators to detect the gait abnormality at both speeds. We used Area Under the Curve (AUC) as a performance metric, and also extracted sensitivity and specificity. Statistical significance was tested with a Student's t-test, and effect size with Cohen's d.

## Results

Trials at self-selected speed better identified patients with extensor thrust (AUC > 0.85, Sens > 0.92, Spec > 0.78,  $p < 0.01$ ,  $d > 1.47$ ). Shank angular velocity slightly outperformed knee angular velocity (AUC = 0.87, Sens = 0.92, Spec = 0.89,  $p = 0.007$ ,  $d = 1.54$ ). Shank angular velocity was positive without extensor thrust (11.5°/s) and negative with it (–22.5°/s), reflecting backward movement.

## Discussion

Our findings show that extensor thrust can be clearly identified using shank angular velocity. Only one false negative and one false positive were observed. The false positive may reflect poor knee control, occasionally causing sudden extension. This pattern is often more pronounced at higher walking speeds. Shank behavior at high speed may point to spasticity or contracture in lower limb muscles [3]. Larger cohorts could help identify subtypes of this gait, as done for stiff-knee gait [4]. Finally, this indicator could be easily monitored using inertial measurement units [5].

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## Virtual Reality and Motion Capture in Orthopedic Rehabilitation: A Preliminary Study Using the eMotion System

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## Introduction

Modern technologies, like motion tracking systems and VR games, offer promising support for rehabilitation by motivating patients and enabling precise progress monitoring (1–3). This study evaluated the effectiveness of the eMotion system—a VR game with integrated motion tracking—in orthopedic rehabilitation, aiming to encourage patients to perform therapeutic exercises.

## Research Question

Can the eMotion VR system effectively support the rehabilitation of orthopedic patients? What are its practical advantages and limitations in clinical settings?

## Methods

The study was conducted among 15 orthopedic patients who were at least six months post-trauma and knee surgery (Mean participant age:  $42 \pm 8$  years; height:  $172 \pm 6$  cm; weight:  $73 \pm 9$  kg). Each participant completed a single rehabilitation session using the eMotion virtual reality system. During the session, they performed a series of exercises—such as squats, jumps, and kicks—designed to promote functional movement. The task involved earning points by avoiding or hitting virtual objects. The eMotion application allows full customization of gameplay parameters, including the speed and placement of virtual elements, and can be tailored to individual characteristics such as height and arm span. Throughout the session, motion data were collected using a dedicated eMotion module with seven HTC Vive Tracker sensors on the lower limbs, controllers, and HTC Vive Pro goggles, all synchronized via Lighthouse 2.0 base stations to form a cost-effective motion capture system. The usability of the eMotion system in orthopedic rehabilitation was analyzed, along with the potential to use the collected data to assess the quality of performed exercises by examining sensor trajectories and changes in lower limb joint angles during individual movements.

## Results

The eMotion system enabled continuous data acquisition and provided a comprehensive description of lower limb kinematics. The game mechanics made it possible to design a physiotherapy session tailored to the needs of a specific group of patients. Figure 1 presents

an example of recorded joint angle changes in the right lower limb during a squat performed by one of the orthopedic patients.

## Discussion

The eMotion rehabilitation game offers intuitive mechanics adapted to virtual reality, engaging users in physical tasks. A scenario editor allows users to personalize exercises for orthopedic patients and adjust the difficulty of the game by modifying the placement, speed and configuration of virtual objects. An integrated motion analysis module enables objective evaluation of movement quality and rehabilitation progress. Functional tests have confirmed the system's usefulness in orthopedic rehabilitation, providing continuous data for detailed lower limb kinematics. High interactivity and immersion positively influence patient motivation and engagement. Importantly, its design and low-cost components enable adaptation for remote rehabilitation, opening opportunities for use beyond clinical centers, including in home settings.

## Acknowledgments

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## Toward a synergy-based ecological framework for application of musculoskeletal modelling in clinics

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## Introduction

Computational neuromusculoskeletal (NMS) modelling is a non-invasive approach to estimate the activity of muscles in dynamic conditions [1]. The adoption of electromyography (EMG)-informed approaches (i.e. Moco [2]), enable the personalization of the model

and allow obtaining more reliable estimation of muscle forces. Despite the advantages, their translation in clinics is still hampered due to the complexity of the acquisition setup, as the EMG activity of several muscles is required [3].

## Research Question

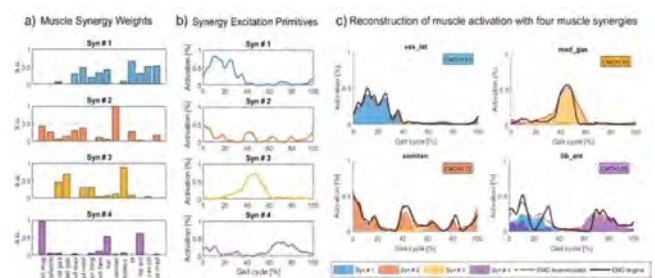
This study aims to investigate the possibility of overcoming the need of acquiring several EMG signals by including the analysis of muscle synergies in a NMS pipeline developed in Moco.

## Methods

Fifteen walking trials of four healthy subjects (mean age =  $60 \pm 2.16$ , mean BMI =  $26.66 \pm 4.13$ ) were acquired through a motion capture system (BTS, 60 Hz), synchronized with 2 force plates (Bertec, 960 Hz) and a 15 channels surface EMG system (FreePocketEMG, 1000 Hz). Markers were placed according to the IOR-gait protocol [4], while the EMG activity of the following muscles was acquired: Gluteus Maximus, Gluteus Medius, Tensor Fasciae Latae, Sartorius, Adductor Longus, Rectus Femoris, Biceps Femoris, Semitendinosus, Vastus Lateralis, Vastus Medialis, Tibialis Anterior, Peroneus Longus, Gastrocnemius Lateralis, Gastrocnemius Medialis and Soleus. After converting the data to an OpenSim-compatible format (through MOtoNMS [5]), scaling, inverse kinematics, and dynamics procedures were performed. Moco EMG-informed was first applied to estimate muscle activations and forces by informing the NMS model with a complete experimental EMG setup. The results were processed in Python for the extraction of four muscle synergies by applying the ‘Non-negative Matrix Factorization’ method [6]. The Coefficient of Multiple Correlation (CMC) [7] was used to identify the EMG signal best representing each synergy within each muscle group (Ankle Plantar-flexors, Ankle Dorsi-flexors, Knee Extensors, Knee Flexors). Afterwards, the original NMS model was informed with a reduced setup of the EMG signals of the four muscles that best matched with the muscle synergies. Comparison was done between the different setups for each subject.

## Results

Figure 1 shows: a) muscle synergy weights, b) identified synergy excitation primitives, c) excitation time series for a representative subject (mean of 6 trials). Black represents experimentally measured excitations, red the reconstructed ones via muscle synergy analysis. CMC between primitives and the four reconstructed EMGs are reported (a.u. = arbitrary unit).



## Discussion

The implemented methodology allowed muscle forces analysis during walking using reduced EMG setup, with a sensitivity analysis on NMS models with limited inputs. Primitives' extrapolations reveal that each individual recruits different muscles to approximate each

synergy, highlighting the subject-specific and distinctive nature of neural activation in human movement. This finding opens avenues for future reflection and analysis. Although further validation on a larger cohort, including individuals with pathological conditions, is necessary, the preliminary results seem promising.

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<https://doi.org/10.1016/j.gaitpost.2025.07.061>

## A 6DOF knee musculoskeletal model for estimating neuromuscular control patterns in drop landing: an EMG-informed approach

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## Introduction

The anterior cruciate ligament (ACL) is one of the most frequent knee injury that occurs in presence of sudden knee torsional misalignment, concurrent knee valgus and deep squat position, which are generally associated with poor landing and change of direction manoeuvres [1]. Different ACL injury screening tests have been proposed over the years including the drop landing (DL) test [2], as it allows assessing the ability to cope with sudden impacts [1]. Knee musculoskeletal models (MSMs) by allowing estimating muscle-tendon forces (MF) and ligaments' tension during task such as DL in a simulation framework (i.e. OpenSim [3]), provide a deeper understanding of ACL injury biomechanics. In this scenario,

MSMs informed by surface electromyography signal (sEMG), can provide a higher degree of personalisation by accounting for the subject specific muscle activity (MA). In this regard, Computed Muscle Control [3] and Moco [4], by incorporating the excitation-contraction dynamics of the muscle, allow estimating MAs that are closer to the experimental sEMG patterns.

## Research Question

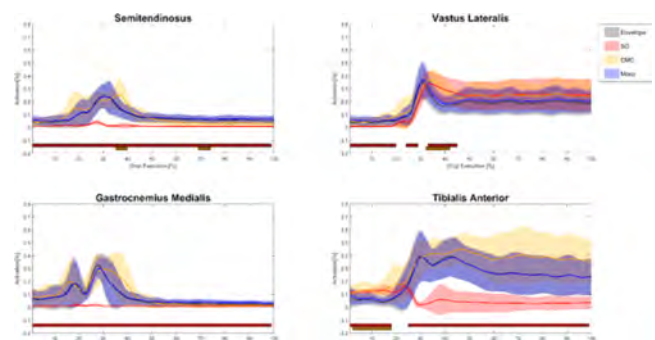
This study aims to assess the reliability of an EMG-informed 6DOF knee model in estimating MAs and MFs in a healthy population of subjects while performing DL tasks.

## Methods

After signing the informed consent, ten healthy subjects (mean age and BMI:  $26.1 \pm 1.5$  years,  $22.6 \pm 2.0$  kg/m<sup>2</sup>) were acquired through six BTS cameras (60 Hz), synchronized with two Bertec force platforms (960 Hz) and an 8-channel sEMG system (1000 Hz) during three consecutive double legs DLs. The sEMG activity of 4 muscles (Vastus Lateralis, Semitendinosus, Tibialis Anterior and Gastrocnemius Medialis) was recorded bilaterally. Markers were placed according to a modified IOR-gait protocol including 4 clusters at both leg and shank [5]. Data were processed in OpenSim (Inverse Kinematics, Inverse Dynamic, Static Optimization) using a 6DOF knee MSM to generate input files for Moco EMG-informed analysis and constrained muscle excitations for CMC analysis. Moco EMG-informed was then applied to estimate MAs and MFs. sEMG envelope peaks were calculated via bandpass filtering (10–450 Hz) with a fifth-order Butterworth filter. To assess the impact of incorporating sEMG data into the knee MSM, both root mean square error (RMSE) and coefficient of multiple correlation [6] were computed between MAs estimated through the different approaches and the experimental sEMG; Paired t-tests using SPM [7] were also conducted to compare MAs and MFs across all methods.

## Results

In **Figure 1** the comparison between MAs computed through Static Optimization (SO), CMC and Moco can be observed. Results of the present study showed that Moco EMG-informed estimates are closer to the gold standard (i.e. experimental sEMG, correlation > 0.98 and RMSE < 0.02).



## Discussion

Given that MFs cannot be experimentally measured, our results suggest that MFs estimated through Moco EMG-driven should be preferred in the context of a 6DOF knee model.

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## ☆Patients suffering from femoroacetabular impingement syndrome show altered external hip joint moments

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## Introduction

Femoroacetabular impingement syndrome (FAIS) increases the risk of early hip osteoarthritis [1] with higher hip joint loads, assessed by external joint moments, as a suggested risk factor [2]. However, it remains inconclusive whether FAIS patients exhibit altered hip joint moments compared to healthy individuals for low-intensity activities such as walking [3], while more demanding tasks have rarely been considered. Also, most of the previous studies focused on peak rather than phase-related loading and suffered from limited sample sizes. This study therefore aims to enhance the understanding of hip joint loading in FAIS patients by analyzing sport-related tasks and phase-specific loading.

## Research Question

Do external hip joint moments differ between FAIS patients and healthy controls?

## Methods

Fifty FAIS patients and 50 matched healthy controls were evaluated during walking, running, stepping up a two-step stair, standing up/sitting down, lateral hop, drop jump, pivot turn and single-leg squat. Motion capture relied on 13 infrared cameras (Vicon, Oxford Metrics Ltd, UK), a full-body marker set [4] and four decoupled force plates (Kistler Instruments, Switzerland). External hip joint moments were calculated using an inverse quasi-static approach, integrating ground reaction forces, functional joint centers and sex-specific segment masses [5]. Time-series data from patients and controls, were compared using vector-field statistical parametric mapping (SPM) with Hotelling's T2 statistic, followed by post hoc scalar-field SPM t-tests in case of significance [6].

## Results

External hip joint moments were significantly different in FAIS patients compared to healthy controls during parts (reported in percentage) of respective movement cycles for the following tasks: stepping up (lower adduction 72-85%  $p < .001$ ), standing up (higher adduction 75-89% and 95-100%  $p = .005$ ), sitting down (higher adduction 0-25%  $p < .001$ ), lateral hop (lower flexion 31-57%, lower internal rotation 41-78%, higher adduction 71-83%,  $p < .001$ , Figure 1), drop jump (lower flexion 31-37%  $p = .002$ , lower internal rotation 29-34%  $p = .007$ ), pivot turn (higher adduction 9-13%  $p = .009$ ).

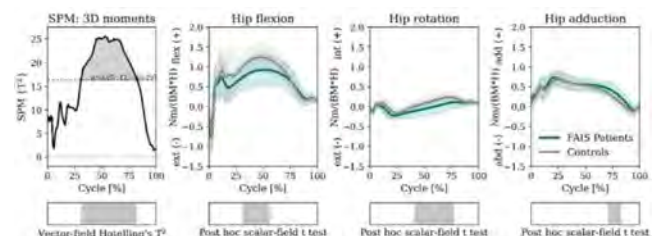


Figure 1 – External hip joint moments (mean  $\pm$  SD) of FAIS patients and healthy controls during a lateral hop, normalized by body mass (BM) and height (H).

## Discussion

The largest differences in external hip joint moments between FAIS patients and healthy controls were observed during the lateral hop, with smaller yet similar differences during the drop jump. The lower flexion and internal rotation moments of patients may be attributed to a shifted center of mass, substantiated by a more upright trunk posture, and a lower push off force. These compensatory strategies potentially reflect pain avoidance and weakness of hip abductor [7] and hip extensor [8] muscles, that were previously reported in FAIS patients. To clarify the role of joint loading in osteoarthritis development, longitudinal assessments of joint moments are needed in FAIS patients - from early disease stages and across conservative and surgical treatments, as surgery may delay osteoarthritis progression [9].

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## How small can your sample size get for reliable EMG reference data considering age and gender variations?

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## Introduction

Surface electromyography (EMG) is a primary tool in clinical gait analysis, providing objective information about muscle activation during walking [1]. Reliable interpretation of EMG data requires normative reference values; however, most existing datasets inadequately consider age- and gender-related differences. A recent Delphi survey suggested about 20 participants could establish EMG norms, recommending inclusion of children and adults due to developmental EMG variations [2]. Previous studies either excluded gender differences or analyzed isolated gait phases rather than continuous data across the gait cycle, leaving a gap in comprehensive reference datasets [3,4].

## Research Question

We aimed to systematically determine how age and gender affect lower-limb muscle activation across the full gait cycle and identify the minimum sample size required to reliably detect these interactions.

## Methods

Surface EMG data were retrospectively analyzed from 97 typically developing individuals (35 children <13 years; 62 adults >18 years; balanced gender distribution). Participants walked barefoot at self-selected speeds. EMG signals were collected from seven muscles (tibialis anterior, soleus, gastrocnemius lateralis, rectus femoris, vastus lateralis, biceps femoris, semimembranosus), processed, and normalized. Statistical Parametric Mapping (SPM) and repeated-measures MANOVA (age × gender) evaluated continuous EMG patterns. Post-hoc power analysis (G\*Power) determined minimal subgroup sample sizes.

## Results

Significant age- and gender-specific EMG differences emerged. Children displayed higher rectus femoris activity in initial swing and increased hamstring activation during midstance, whereas adults showed elevated semimembranosus at initial contact, greater soleus during push-off, and higher rectus femoris activation in late swing. Females exhibited greater tibialis anterior activity during terminal stance and increased vastus lateralis activity during swing, while males had greater vastus lateralis and biceps femoris activation in terminal swing. Significant age-gender interactions were evident in hamstring muscles, with pronounced gender differences emerging in adulthood. Power analysis indicated at least 47 participants (≈12 per subgroup) were necessary to reliably detect these interaction effects.

## Discussion

This study uniquely demonstrates significant interactions between age and gender on EMG patterns across the entire gait cycle, reinforcing the need to stratify EMG reference datasets accordingly. Continuous analysis using SPM effectively captures subtle neuromuscular variations previously overlooked [5]. Clinically, gait laboratories (e.g. those within ESMAC) are advised to consider establishing age- and gender-specific normative EMG datasets, ideally including approximately 12 participants per subgroup (male children, female children, male adults, and female adults) and a total sample size around 47 individuals. Adopting these practical guidelines will substantially enhance the reliability and clinical utility of EMG-based gait assessments.

This research was funded by German Research Foundation (DFG) (no: WO 1624/ 8-1).

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## Exploration of early muscle growth and its interaction with spontaneous movement patterns in typically developing 3-month-old infants.

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## Introduction

Preterm-born infants with brain injuries are at increased risk of developing cerebral palsy (CP)<sup>1</sup>. This neuromotor disability is associated with impaired lower-limb muscle growth, observable early in life<sup>2</sup>. These infants commonly display atypical spontaneous motor patterns during the first months, reflecting impairments of the central nervous system<sup>3</sup>. Although altered muscle development is a key characteristic of CP, the underlying mechanisms remain poorly understood. An interactive feedback loop is hypothesized between disrupted muscle innervation and impaired muscle growth, which may, in turn, affect the development of movements<sup>4</sup>. This study aimed to explore the relationship between lower limb muscle morphology and motor output during typical development, providing a reference framework for understanding early altered muscle development.

## Research Question

Is medial gastrocnemius muscle morphology associated with lower limb movement patterns at 3 months of age in typically developing infants?

## Methods

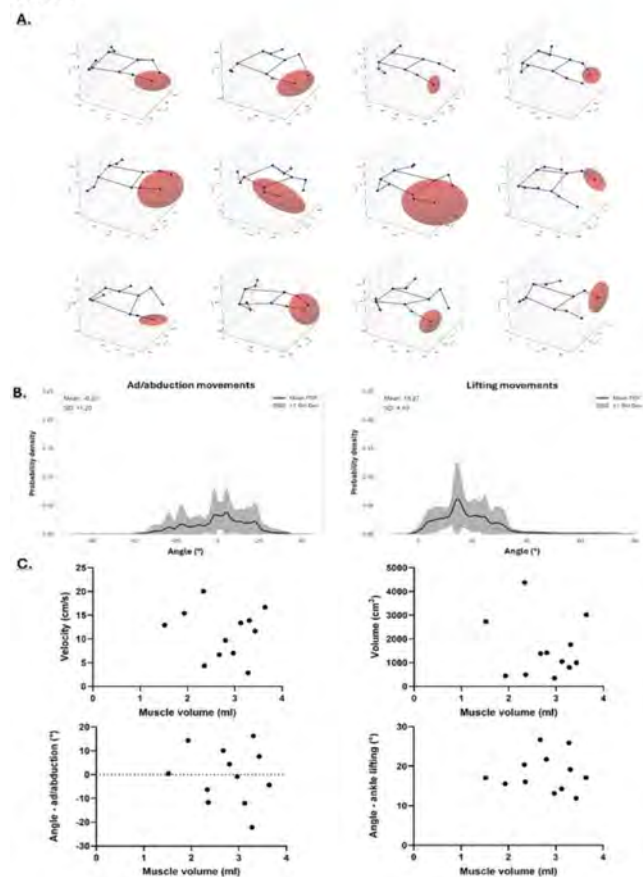
This exploratory study included typically developing (TD) infants (gestational age > 37 weeks) approximately 3 months old. In each infant, one leg was randomly selected for analysis. Medial gastrocnemius muscle morphology was assessed using 3D ultrasound<sup>5</sup>. Muscle volume (mL) and muscle belly length (mm) were quantified using Slicer software<sup>6</sup>. Three-dimensional motion capture system (Qualisys) with 18 optoelectronic cameras recorded the spontaneous lower limb movements in supine position. A common window of 55 seconds was selected for each child, based on the time segment with the largest ankle velocity movements. The movements were quantified by calculating the ankle marker velocity and the volumetric envelope of the movement trajectory, approximated as a 3D

ellipsoid. Movement patterns were analyzed (a) in the frontal plane: 0° represents alignment of the ankle with the pelvic; negative/positive angles correspond with ad-/abduction, respectively; (b) in sagittal plane: 0° represents full leg extension relative to the trunk; positive angles indicate lifting of the ankle. Muscle-movement associations were examined using Pearson correlation coefficients.

## Results

Twelve TD infants were included (age  $3.0 \pm 0.2$ mo, body weight  $6.1 \pm 0.8$ kg and height  $61.0 \pm 2.3$ cm). Muscle volume of  $2.78 \pm 0.64$  ml and muscle length of  $49.38 \pm 5.02$  mm were observed. Ankle trajectory quantification demonstrated a mean velocity of  $11.37 \pm 5.22$  cm/s and volume of  $1545 \pm 1109$  cm<sup>3</sup> (Figure 1.a, representing the 3D ellipsoid per child). Descriptive results of the movement patterns are presented in Figure 1.b. No significant associations were found between the muscle morphological and movement parameters (Figure 1.c).

Figure 1



## Discussion

Ankle movements were generally aligned with the hip joint but showed substantial inter-individual variability. Ankle lifting corresponds with age-adequate motor patterns of anti-gravity movements<sup>7</sup>. These preliminary findings were not sufficient to link muscle morphology with movement patterns in TD infants, though more data are needed. Future work should advance the movement parameters, including other patterns (e.g., kicking, midline movements) and electromyography assessing muscle activation.

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## Insights into early gait development: variability and harmony during supported and independent walking of typically developing 12-month-olds

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## Introduction

Independent walking typically emerges within the first years of life and is initially characterized by high variability, reflecting infant's motor learning through exploration<sup>1</sup>. At the onset of acquired walking, typically developing (TD) infants show greater variability in step length than in step width<sup>2</sup>. In parallel, gait harmony, defined by the ratio of gait cycle to the stance phase, gradually evolves toward the golden ratio of 1.6, indicating maturation of rhythmic movement patterns<sup>3</sup>. In contrast, infants with neurodevelopmental disorders often show delayed or absent independent walking. Early motor milestones, such as supported walking, may offer early insight into these developmental differences. Assessing variability and harmony in supported and independent walking in TD infants is essential for establishing normative data and may offer information for early intervention strategies.

## Research Question

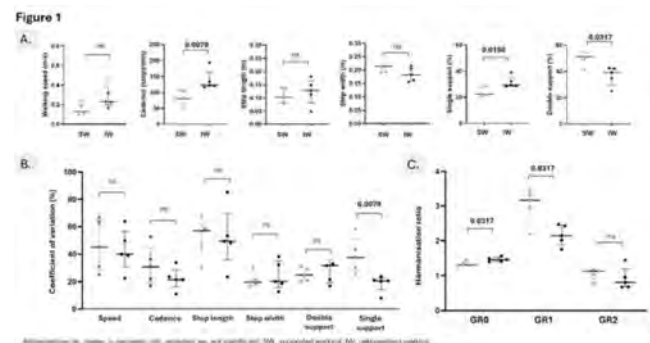
What is the difference in spatiotemporal variability and walking harmony between supported and independent walking in 12-month-old TD children?

## Methods

Twelve-month-old TD children were grouped based on walking ability: supported walking (SW) or independent walking (IW). SW was standardized with anterior arm support at approximately 90 degrees of shoulder ante flexion. Spatiotemporal parameters from the right leg were recorded via 3D motion capture system (Qualisys) with markers placed according to the full-body Conventional Gait Model<sup>4</sup>. Gait harmony was calculated based on temporal ratios between different stride phases: GR0 = stride/stance, GR1 = stance/swing, and GR2 = swing/double support. Intra-subject variability was expressed as coefficient of variation (CV). Between-group comparisons were performed using the non-parametric Mann-Whitney U test ( $p < 0.05$ ).

## Results

Five children walked with support on average 20.8 cycles (age: 11.9 mo (inter-quartile range (IQR) 11.8–12.6); leg length: 31.5 cm (30.8–34.6)) and five children (12.0 mo (IQR 11.9–12.6), 30.0 cm (IQR 29.0–32.4)) were walking independently (average: 54 cycles). Compared to the SW-group, the IW-group demonstrated a significantly higher cadence, longer single support, and shorter double support duration (Figure 1.A). Only the CV of single support duration was significantly lower in IW-group (Figure 1.B). The median GR0 was significantly lower in the SW-group while the GR1 was significantly higher compared to the IW-group (Figure 1.C).



## Discussion

This explorative study focusing on spatio-temporal parameters found that, during SW, children walked slower and with shorter single support durations combined with higher variability compared to IW. Consistent with previous findings, both conditions showed more variability in step length than in step width. Notably, the GR0 ratio did not yet approximate the golden ratio during both walking experiences, however, values were closer to it during IW, suggesting a more mature temporal organization. Expanding these normative datasets are promising to understand the locomotion in infants with (high-risk for) neuromotor disabilities. However, the small sample size and cross-sectional design are limitations. Additional muscle synergy analysis could also offer deeper insights into gait development across early milestones.

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<https://doi.org/10.1016/j.gaitpost.2025.07.066>

## The effects of Soleus and Gastrocnemius spasticity on gait: A comparison between predictive simulations and expert consensus

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## Introduction

Spasticity of the calf muscles can have a detrimental impact on gait [1][2], but it can be challenging to assess if soleus or gastrocnemius spasticity is most prominent. One approach to evaluate the impact of spasticity is to use predictive gait simulations, which allow a systematic assessment of the biomechanical cause-effect relationship between spasticity and gait. Another approach is to rely on expert consensus regarding how muscle-specific impairments affect gait [3]. Currently, it is unclear how these two approaches align.

## Research Question

Do the kinematic effects resulting from soleus and gastrocnemius spasticity align between predictive gait simulations and kinematic gait features expected according to expert consensus?

## Methods

We employed a predictive simulation framework [4], using a planar neuromusculoskeletal model [5], featuring nine musculotendon units per leg. Muscles were activated by a reflex-based control model. Gait simulations were optimized by predominantly

minimizing cost of transport, head movements, and large foot-ground impacts [6].

Spasticity was modelled by adding a velocity-dependent reflex with a constant value to either the soleus or gastrocnemius muscle throughout the gait cycle. Five spasticity levels were simulated as percentages (20–100%) of a predefined maximum spasticity gain value (3.6 for Gastrocnemius, 4.0 for Soleus), which was based on convergence failure (< 4/6 optimizations converged) or negligible effects on muscle activation and kinematics with increasing gain. Kinematic outcomes were compared with a healthy simulation and evaluated against gait features based on expert consensus as previously obtained using a Delphi procedure [3].

## Results

Soleus spasticity resulted in direct kinematic changes, rather independent of the level of spasticity, with increased plantarflexion across the whole gait cycle, increased knee extension during stance and increased knee flexion during late swing (Fig 1A). These changes aligned with all gait features based on expert consensus, except for knee extension movement in loading response.

Incrementing gastrocnemius spasticity resulted in more progressive kinematic changes, with increased plantarflexion throughout the gait cycle (Fig 1B). Knee flexion increased during the loading response and late swing. These changes aligned with expert consensus for plantarflexion, toe walking and knee flexion during stance.

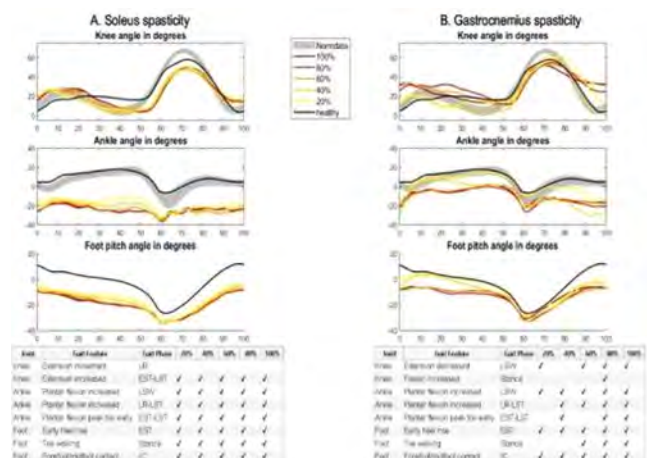


Figure 1: Effect of A.Soleus and B.Gastrocnemius spasticity on gait kinematics and alignment with expert consensus [3] (✓ indicates agreement)

## Discussion

The simulations captured most of the kinematic gait features expected by expert consensus, reflecting a large agreement between the two. Interestingly, soleus spasticity led to an instant change in kinematics, whereas the effects of gastrocnemius spasticity emerged more gradually, possibly due to its bi-articular nature.

Future research should further validate the model against experimental data to elucidate if current mismatches are due to modelling approaches or incorrect expert expectations. Ultimately, predictive simulations may help to further refine clinical understanding by contributing to a deeper insight in the mechanisms underlying spasticity and its impact on gait.

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### Long-Term Effects of Botulinum Toxin on Medial Gastrocnemius Muscle Morphology Parameters in Children with Cerebral Palsy: A One-Year Follow-Up Study

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## Introduction

Botulinum NeuroToxin type A (BoNT-A) is a frequently applied treatment to reduce spasticity in children with cerebral palsy (CP). While it facilitates functional gains, adverse effects on the muscular level have been observed. Previous literature reported a reduced muscle volume (MV) of the medial gastrocnemius (MG) three months after injection [1,2]. At six months, MV was still decreased, and the anatomical cross-sectional area (CSA) and muscle quality were also reduced [2,3]. Additionally, a trend toward reduced muscle growth-rate (increase in MV per month) was noted in treated children compared to an untreated CP group [3]. Yet, the long-term effects of BoNT-A on MG muscle parameters, particularly up to one year post-treatment, remains unknown.

## Research Question

What is the recovery trajectory of the MG muscle (morphology) parameters over one year following BoNT-A treatment? Are the muscle parameters at one year comparable to these of an age- and severity-matched untreated CP group?

## Methods

Twenty-six children with spastic CP (mean age:  $5.19 \pm 3.26$  years) were included. Three-dimensional freehand ultrasound measurements of the MG were performed at baseline, and at three, six and 12 months post-BoNT-A. Muscle morphology parameters included MV, muscle length (ML) and CSA. Anthropometric-related percentile curves of typically developing children ( $N=145$ ) were used to convert patient outcomes into z-scores, excluding natural muscle growth and allowing standardized comparisons across groups and outcomes [4]. Echo-intensity (EI) estimated muscle quality. The BoNT-A effect over time was analyzed using linear mixed models. Additionally, muscle parameters at one year post-BoNT-A were compared to one-year follow-up data of an age- and severity-matched untreated CP group ( $n=26$ , mean age:  $4.98 \pm 2.15$  years) using repeated measures ANOVA. Bonferroni correction was applied to account for multiple outcomes and pairwise comparisons.

## Results

There was a trend for decreased MV at three ( $p=0.0049$ ), six ( $p=0.0034$ ) and 12 months ( $p=0.0076$ ) post-BoNT-A compared to baseline, while ML remained unchanged throughout the follow-up period. CSA was reduced only at six months ( $p=0.0005$ ), and EI increased only at three months post-BoNT-A ( $p<0.0001$ ) (Figure 1). At one-year follow-up, MV tended to be lower compared to baseline in the intervention group ( $p=0.0076$ ), but not in the untreated CP group. ML, CSA and EI did not change significantly in both groups. There was a trend for reduced growth-rate in the BoNT-A treated group ( $-30\%$ ,  $p=0.035$ ), whereas the untreated group exhibited increased muscle growth ( $+32\%$ ,  $p=0.029$ ).

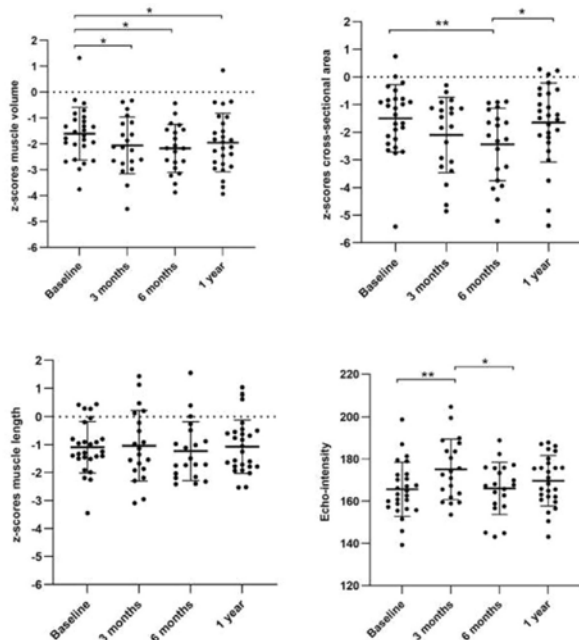


Figure 1. Scatterplot of individual datapoints at baseline, 3 months, 6 months and 1 year post botulinum neurotoxin type A (BoNT-A) injection, with mean and standard deviation displayed as black stripes. Significant results at the 0.013 alpha-level are indicated with a single asterisk (\*), to correct for 4 muscle parameters. A double asterisk (\*\*) indicates significant results at the 0.002 corrected alpha-level for 6 pairwise comparisons

## Discussion

These findings suggest distinct recovery timelines for CSA and ML, which may explain the incomplete MV recovery and highlight substantial interindividual variation in recovery processes. This was the first study to incorporate a case-control matched group, demonstrating a different growth trajectory over a one-year period between the treated and untreated control group. These results highlight the importance of personalized long-term monitoring of muscle (morphology) parameters post-BoNT-A and targeted muscle-specific interventions.

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<https://doi.org/10.1016/j.gaitpost.2025.07.068>

## The effect of orthoses in children with cerebral palsy with limited rehabilitation services in Suriname: a preliminary evaluation

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## Introduction

Ankle-foot orthoses (AFOs) are frequently integrated into rehabilitation programs to improve gait in children with cerebral palsy [1]. However, in low- and middle-income countries (LMICs), such as Suriname, children with cerebral palsy often have limited access to rehabilitation services and, consequently, to orthotic devices such as AFOs[2]. This study investigates the effect of AFOs in children with spastic cerebral palsy (SCP) who have had limited access to rehabilitation services both prior to receiving and during the use of the AFOs.

## Research Question

Are gait deviations decreased and quality of gait improved by the use of AFOs in children with SCP with limited access to integrated rehabilitation services?

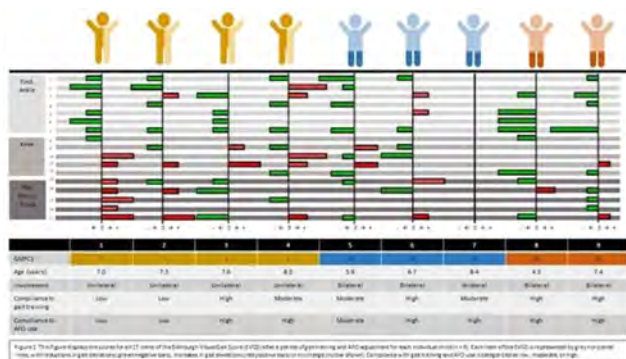
## Methods

A prospective observational study was conducted in 9 children with SCP, GMFCS level I (n=4), II (n=3) and III (n=2) with uni- and bilateral involvement. Children with GMFCS level I-II never received surgery, botulinum toxin, casting, orthoses and had limited physiotherapy ranging from a few up to 10 sessions per year. Children with GMFCS level III never received surgery, botulinum toxin, but had casting prior to receiving the orthoses and visited the physiotherapist once a month prior receiving the orthoses. All participants received custom-made orthoses uni- or bilaterally. After receiving the orthoses, 5 out of the 9 children received weekly gait training by a physiotherapist for 1-2 months. The other children

received less or no gait training due to difficulties in transport and financial constraints. Gait deviations were evaluated using the Edinburgh Visual Gait Score (EVGS) for both legs[3]. Evaluations were performed prior to receiving the orthoses and after a period of gait training and adjustment period of 6-10 months. One child was assessed after a shorter adaptation period of 2.5 months. The Wilcoxon signed rank test was used to compare the overall score between baseline and after a period of gait training and adjustment of AFO use.

## Results

Gait deviations significantly decreased following the AFO use ( $p = .042$ ). Figure 1 presents the scores for all 17 EVGS items (grey horizontal lines), with reductions in gait deviations (green bars) and increases (red bars) for each individual child. One child, assessed after only 2.5 months, showed no change in all items. Improvements were observed across all levels (foot and ankle, knee, hip, pelvis, and torso), although the extent of improvement varied between individuals. No clear trends were found regarding age, GMFCS level, or participation in gait training in relation to the decrease of gait deviations.



## Discussion

These preliminary findings suggest that AFOs can reduce gait deviations and improve gait quality in children with SCP in Suriname, despite limited access to integrated rehabilitation services. To further enhance outcomes for all children, targeted strategies for the implementation of AFOs in LMICs should be developed to optimize gait improvements in this population.

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## A deep convolutional autoencoder to predict the position of the full-body centre of mass from lower limb kinematics

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## Introduction

Clinical gait analysis provides an objective and quantitative assessment of the walking function in persons with a range of impairments. Walking safely requires the capacity to maintain balance, which may be evaluated using the margin of stability during gait analysis [1] if we know the position of the full body centre of mass (CoM). However, clinical gait analysis often involves only lower limb kinematics, and the position of the full body CoM is unknown.

## Research Question

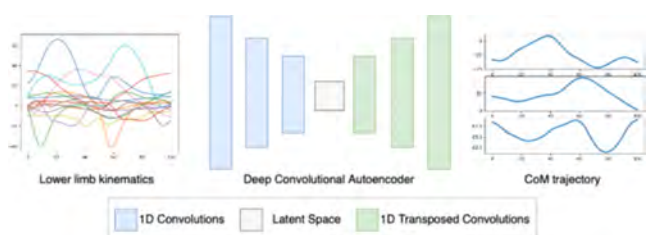
Can we predict the position of the full body CoM from lower limb kinematics only?

## Methods

Full body gait analysis was collected in 21 typically developing children (age range: 7-17 years). Full body kinematics was calculated with the CGM2.1 and the reference position of the CoM was derived (CoM\_Dumas) [2,3]. For comparison, the root mean square error (RMSE) between CoM\_Dumas and a simple model, rigidly attached to the pelvis, (CoM\_simple) was also calculated [4].

A deep Convolutional AutoEncoder [5], Deep\_CoM, was trained to predict the time varying relative position, expressed as a percentage of clinical leg length, of the CoM within the pelvis segment during gait (CoM\_pred). The model used lower limb kinematics only. The accuracy of the prediction was evaluated as the leave-one-out-cross-validation (LOOCV) RMSE between CoM\_pred and CoM\_CGM.

Figure 1: Architecture of Deep\_CoM



## Results

The average (SD) RMSE between CoM\_Dumas and CoM\_simple was 20(5.3) mm and 12(4.4) mm between CoM\_Dumas and CoM\_pred.

## Discussion

Deep\_CoM provided a 40% improvement over a simple model of the position of the CoM rigidly attached to the pelvis segment. Deep\_CoM accuracy (12mm) is similar to the difference between full-body models or between full-body models and CoM trajectories derived from forceplate data [4].

Deep\_CoM predicts the position of the CoM in the pelvis as a percentage of clinical leg length to generalise well across persons with different size, as was the case in our dataset. Deep\_CoM will be further trained and evaluated in a large dataset of typically developing persons and in persons with disability.

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<https://doi.org/10.1016/j.gaitpost.2025.07.070>

## Limits of stability in men with obstructive sleep apnea and effects of concurrent chronic neck pain: A three-group comparison study

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## Introduction

Individuals with obstructive sleep apnea (OSA) have been reported to have altered neck posture, and OSA has been associated with chronic spinal pain complaints (1,2). Furthermore, individuals with OSA have decreased balance performance, a finding that has also been observed in those with chronic neck pain (CNP) (3,4). Therefore, data comparing balance among individuals with OSA, those with both OSA and CNP, and asymptomatic controls may provide valuable insights for the clinical management of OSA.

## Research Question

Does balance performance differ among individuals with OSA, those who have both OSA and concurrent CNP, and asymptomatic controls?

## Methods

The study included 92 male participants aged between 40 and 65 years: 31 with OSA, 31 with both OSA and CNP, and 30 asymptomatic controls. All participants with OSA had been diagnosed by polysomnography. In participants with OSA, daytime sleepiness and the severity of obstructive sleep apnea were evaluated using the Epworth Sleepiness Scale (ESS) and the Apnea-Hypopnea Index (AHI), respectively. In participants with CNP, neck pain severity and related disability were assessed using the Visual Analog Scale and the Neck Disability Index, respectively. Limits of stability (LoS) were assessed with the computer-aided balance measurement system (NeuroCom Balance Master, NeuroCom® International Inc., USA). The system software calculated composite scores for reaction time (RT), movement velocity (MV), endpoint excursion (EPE), maximum excursion (MXE), and directional control (DC). As the p-value for the group differences in BMI was close to the significance level ( $p=0.051$ ), analysis of covariance was performed to compare groups, with the LoS variables as the dependent variable, group as the independent variable, and body mass index as a covariate. Pearson correlation coefficients were calculated to determine the relationships of ESS, AHI, NDI, and pain intensity scores with LoS scores.

## Results

RT, MV, EPE, and MXE differed between groups with large effect sizes ( $p<0.05$ ,  $\eta^2p=0.161-0.214$ ). In the post hoc comparisons, participants with only OSA and those with both OSA and CNP had lower LoS scores compared to controls ( $p<0.05$ ), while no significant difference was observed between the two OSA groups ( $p>0.05$ ). DC did not differ between groups ( $p>0.05$ ). In individuals with only OSA, ESS score was negatively correlated with EPE and MXE ( $r=-0.381$  and  $-0.433$ ,  $p<0.05$ ). In those with both OSA and CNP, AHI score was correlated with RT and current neck pain intensity was correlated with MV ( $r=-0.451$  and  $-0.419$ ,  $p<0.05$ ).

## Discussion

The study suggests that LoS is impaired in men with OSA, regardless of the presence of CNP. Specific associations between sleep apnea-related and pain-related measures and LoS parameters indicated that both physiological and symptomatic factors may contribute to balance deficits in this population. Incorporating balance assessment and targeted interventions into clinical management may support improved functional stability and enhance overall care strategies.

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<https://doi.org/10.1016/j.gaitpost.2025.07.071>

## Adapting machine learning-based gait event detection models from walking to running: Evaluating transfer learning vs. training from scratch

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## Introduction

Movement tasks like stair climbing or running introduce higher biomechanical loads than walking, offering valuable insights into foot deformities and musculoskeletal conditions [1,2]. However,

analyzing these movements requires accurate event detection. While machine learning (ML) models show strong performance in detecting gait events during walking [3,4], there's limited research on applying ML to other movement tasks. Accurate and robust (<5.5 ms) [3,5] gait event models across different movement tasks are needed in clinical practice and research. A promising approach is transfer learning (i.e., fine-tuning), which leverages knowledge from a source domain (walking) to improve performance in a target domain (running). This study investigates the effectiveness of fine-tuning for gait event detection in overground running.

## Research Question

Is there a difference in model performance between fine-tuning and training from scratch? How much data is needed to achieve accurate results for each approach?

## Methods

The balanced dataset consisted of 3D overground running data from 474 children, grouped into three categories: orthopaedic disorders (OD; n = 158, female = 71, male = 87), neurological disorders (ND; n = 158, female = 65, male = 93), and cerebral palsy (CP; n = 158, female = 74, male = 84). All data was captured at 200Hz. The dataset was split into 20% test and 80% development data. Model development used 10-fold cross-validation (90% training, 10% validation), with training sets progressively reduced by 10% down to a single participant per condition, simulating varying data availability. The original IntellEvent [3] model was evaluated on the test set as a baseline. For fine-tuning, the two top layers of IntellEvent were fine-tuned. Mean absolute errors (MAE) were computed between predicted and ground truth initial contact events.

## Results

Figure 1 shows that fine-tuning allowed better performance with small datasets, while training from scratch benefited from more data and outperforms fine-tuning in large-scale scenarios.

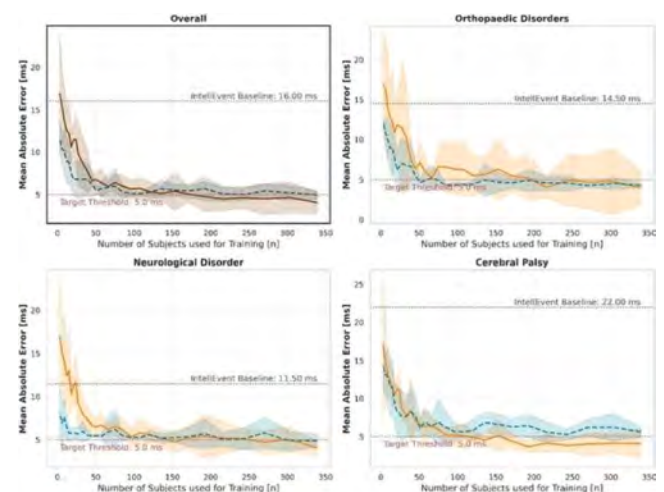


Figure 1. MAE overall and for each patient group, with different amounts of training data for fine-tuning (dotted-blue) and training from scratch (orange). The dotted-grey line represents the MAE of the original IntellEvent model.

## Discussion

The original IntellEvent model's results (16.0 ms) highlight the need for adaptation. Overall, fine-tuning shows lower MAEs with up to 120 subjects, while training from scratch performs better with larger datasets. This is especially evident for CP, where training from scratch surpasses fine-tuning with 75 subjects. For OD and ND, both methods perform similarly, though training from scratch shows greater variability, suggesting a need for more data. Training from scratch demands significant computational resources, time, and ML expertise, whereas, fine-tuning offers high performance with fewer resources and no model architecture changes, making it practical when data or resources are limited. Future research should extend to diverse movement tasks, e.g., stair climbing, treadmill running, and functional activities.

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<https://doi.org/10.1016/j.gaitpost.2025.07.072>

## Within- and Between-Assessor Reliability of Lower-Limb Inter-Joint Coordination During Gait in Individuals With and Without Cerebral Palsy

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## Introduction

Inter-joint coordination plays a key role in walking, particularly in people with cerebral palsy (CP), who experience altered movement patterns (1). The Continuous Relative Phase (CRP) method quantifies lower-limb coordination by assessing the phase relationships between joints (2-3). However, the reliability of CRP measurements during walking in individuals with CP remains unexplored, and may be affected by measurement variability due to marker placement errors, soft tissue artifacts, and natural movement fluctuations. It is important to quantify this reliability to ensure the appropriate use of this metric for clinical comparisons.

## Research Question

What is the within- and between-assessor reliability of lower-limb CRP measurements in individuals with CP and their healthy peers?

## Methods

Reliability was determined in CP (n=19, age=18.437 (7.30) years, GMFCS I–III) and healthy (n=19, age=18.32 (11.20) years)) individuals. Participants completed two gait assessment sessions, each including 3D motion capture of at least 10 walking trials. Two trained assessors independently placed reflective markers according to the conventional gait model (4) and conducted gait analyses. Standard error of measurement (SEM) and minimal detectable change (MDC) were computed for knee-hip and ankle-knee coordination across gait subphases.

## Results

The SEM and MDC were lower for KH knee-hip than AK ankle-knee coordination, suggesting higher measurement reliability for proximal joint coupling. For knee-hip coordination, MDC reached  $15.1 \pm 0.7^\circ$  (CP) and  $9.3 \pm 0.6^\circ$  (NI) between assessors, and  $23.8 \pm 3.0^\circ$  (CP) and  $9.1 \pm 1.6^\circ$  (NI) within assessors. For ankle-knee coordination, MDC reached  $29.0 \pm 2.6^\circ$  (CP) and  $25.0 \pm 3.5^\circ$  (NI) between assessors, peaking at  $47.3 \pm 10.9^\circ$  (CP) and  $28.6 \pm 1.5^\circ$  (NI) in mid-swing within assessors. CP individuals exhibited greater SEM in coordination patterns than NI individuals, particularly during double support and mid-swing phases.

## Discussion

This study provides the first metrological reference for reliability of CRP-based inter-joint coordination during gait in CP. High SEM may be related to the sensitivity of the signal at points of bifurcation (or phase transitions), such as during the switch from flexion to extension, where angular velocity is low. The results underscore the importance of standardized protocols and assessor training in minimizing kinematics variability, which may result in high SEM of the resulting CRP curves during specific subphases of the gait cycle (i.e., during the first double support and early single stance for the knee-hip coordination and during the mid-swing for the ankle-knee coordination).

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<https://doi.org/10.1016/j.gaitpost.2025.07.073>

## Experimental investigation of muscle-tendon unit geometry in lower-limb muscles during gait analysis: A Scoping Review

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### Introduction

Three-dimensional gait analysis (3DGA) is a widely used clinical tool to assess gait abnormalities in populations with neuromotor impairments, such as cerebral palsy (1). However, 3DGA does not provide direct insights into muscle-tendon unit (MTU) dynamics, which are critical for tailoring effective interventions (2). To fill this gap, advanced techniques such as musculoskeletal (MSK) modeling and 3D ultrasound imaging (3DUSI) have been integrated with 3DGA to estimate MTU geometry and function during gait.

### Research Question

How are MSK modeling and 3DUSI employed during 3DGA to assess MTU geometry in individuals with neuromotor impairments, and how do these methods contribute to clinical decision-making?

### Methods

A scoping review was conducted following PRISMA-ScR guidelines. Literature was retrieved from PubMed, Scopus, and Clinical-Trials.gov up to February 2025. Inclusion criteria focused on studies assessing MTU geometry during walking in populations with neuromotor impairments using non-invasive techniques (MSK modeling or 3DUSI).

### Results

Out of 984 studies, 50 met the inclusion criteria: 42 used MSK modeling, 5 used 3DUSI, and 3 combined both techniques. MSK modeling studies (n=45) primarily focused on children with cerebral palsy (n=27/45), using models like Gait2392 (n=8/45) or Delp et al.'s lower-limb models (n=23/45) during treadmill (n=2/45), or overground walking (n=43/45). The semimembranosus (n=22/45) and the gastrocnemius medialis (n=20/45) were the most investigated muscles. Concerning 3DUSI studies (n=8), all focused on gastrocnemius medialis during treadmill (n=5/8) or overground gait (n=3/8). The majority of (n=6/8) focused on children with cerebral palsy. Overall, MTU parameters assessed included length, lengthening velocity, fascicle length, and pennation angle.

### Discussion

MSK modeling allows detailed estimations of MTU dynamics but is limited by assumptions and lack of real-time capability. 3DUSI is constrained by its labor-intensive nature and technical challenges in data acquisition and processing. These technologies have enabled the assessment of muscle length and lengthening velocity across gait phases, the identification of abnormal MTU behavior in pathological gait, the evaluation of treatment effects such as surgery or botulinum toxin injections. They have also been used to investigate muscle-tendon interactions, classify gait patterns, and explore predictors of clinical outcomes. Together, results enhance understanding of neuromuscular impairments and support personalized clinical decision-

making. However, research on other neuromotor conditions (e.g., stroke, multiple sclerosis) remains limited. Future work should aim for broader inclusion and explore automation (e.g., deep learning for 3DUSI) to enhance clinical integration.

Methodological studies have addressed the practical and technical aspects of using these technologies during gait. Some evaluated whether wearing ultrasound equipment affected gait patterns, generally concluding minimal to no impact, thereby supporting the feasibility of in-motion data collection. Others focused on improving the reliability and accuracy of muscle length estimations in MSK modeling, highlighting the importance of consistent workflows, normalization techniques, and model selection, including anatomical assumptions and scaling methods.

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### Usability of iGait@Healthcore: Gait analysis from a smartphone for use in clinical routine

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### Introduction

Gait analysis is an essential method to unmask movement deficits for clinical diagnosis and rehabilitation. Traditional optical motion capture systems are, however, costly and impractical for routine use. Advances in computer vision and mobile computing enable affordable mobile gait assessment. These emerging technologies are still confronted with fragmented adoption, unvalidated metrics, and clinical usability and interpretation challenges. While the pose estimation and modeling aspects are arguably solved, there is limited progress in understanding the contribution of altered movement patterns to the overall health state.

### Research Question

To approach this question, we introduce two components: (1) iGait, a modular computer vision solution for gait analysis that can scale from monocular, free-moving RGB camera to fixed, multi-

camera installations. (2) HealthCore, an embedded learning health system, which aggregates detailed information across health domains such as activity, upper limb function, neuropsychology, and others. This approach facilitates aggregation of movement kinematics with contextual data at scale, across clinical sites, in quasi real-time.

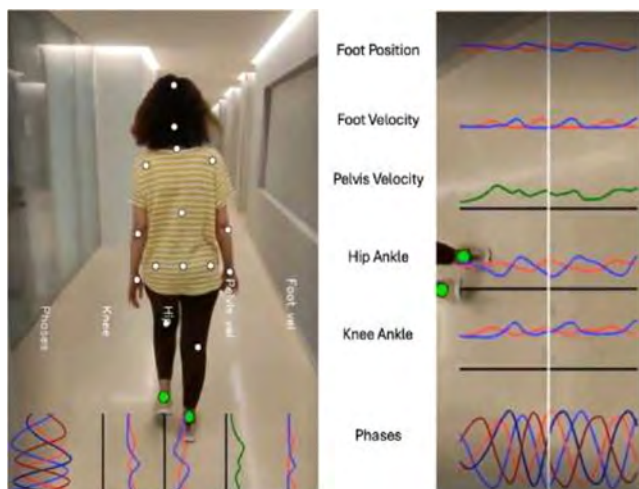
## Methods

HealthCore Tools are a modular family of mobile-based digital data collection tools. iGait is an Android application, designed in an iterative, user-centered design process, to record standardized 15-meter walk tests with the mobile device camera. Next to instruction and guidance for collecting structured gait data, we also provide an option for voice-based commentary and meta-data aggregation (walking aids, stick heights, etc). Captured videos are encrypted and processed on our secure computational health backend, HealthCore. A pose estimation microservice allows the calculation of spatio-temporal parameters and joint angles on a frame-by-frame basis. There are rendered, together with traditional gait parameters on a clinician-facing dashboard. Pose estimation is performed using a previously validated algorithmic suite that is robust to changes in perspective and calibration procedures (1–3). Recording application usability was investigated using a mixture of structured and unstructured feedback through interviews, as well as technical fail logs.

## Results

iGait has been successfully deployed and tested in a clinical setting for neurological rehabilitation. Initial results are promising: Low fail rates of <10% demonstrate the system's ability to reliably capture and process gait data in challenging environments. End-user reports are positive, with clinicians highlighting the ease of use and the additional kinematic information obtained over time-distance measures or IMU-based spatio-temporal data. End-users also state a need for more easily interpretable outcome metrics.

Example video



## Discussion

To bridge gait metrics with broader health insights, HealthCore is designed to ingest multimodal clinical data beyond locomotion—such as neuropsychological assessments, upper limb function tests, and patient-reported outcomes. By linking gait patterns with these

additional data streams, our system enables longitudinal and cross-domain analyses of patient health. While this integration is still being expanded, the modular architecture already supports future incorporation of diverse health domains, facilitating a more holistic understanding of mobility impairments in clinical populations.

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<https://doi.org/10.1016/j.gaitpost.2025.07.075>

## Usability of capturing upper limb movement kinematics in clinical routine: Do wearable sensors beat markerless motion capture?

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## Introduction

During neurorehabilitation, improving movement quality is a central therapeutic goal (1). Yet, movement kinematics are rarely directly and objectively assessed due to complexities in data acquisition and interpretation. Recent advancements in wearable sensors and markerless motion capture technologies offer promising tools for routinely capturing movement quality in clinical practice (2) (3). While these IMU-based (Inertial Measurement Unit) and camera-based systems each present unique technical capabilities and challenges, their integration into clinical routines depends crucially on usability and time efficiency. Understanding which aspects of the different technologies influence their usability in clinical routine can lead to systems with a higher probability of clinical adoption.

## Research Question

This study investigates the usability and end-user preference of IMU-based and markerless motion capture systems when applied within clinical routine.

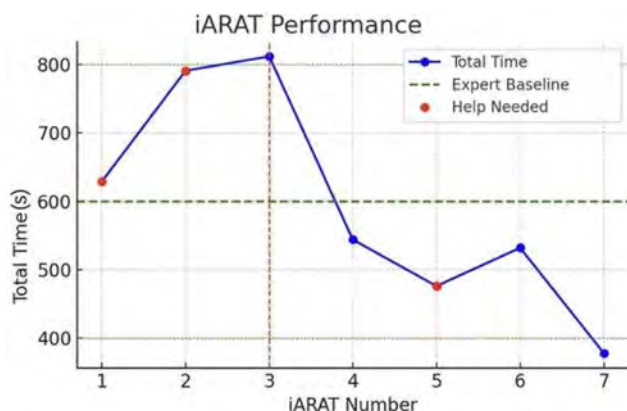
## Methods

We recruited five movement therapists from cereneo clinic for a two-stage usability evaluation. In the first stage, therapists used the iARAT (4), an instrumented version of the Action Research Arm Test (ARAT) employing body-worn IMU sensors connected to a tablet application. We measured the learning curve by recording the time taken to prepare and mount the technology, the necessity for support, and usability questionnaire responses. Each therapist completed up to seven sessions or until they achieved expert status—defined as performing the assessment independently, within a time-frame comparable to experienced users, without external assistance, three times in a row. In the second stage, therapists conducted an instrumented drinking task (5)—a shorter assessment compared to the complete ARAT—using both an IMU-based system (4) and our custom camera-based, markerless motion capture system (3). We recorded the duration of each technology preparation and collected therapists' preferences through questionnaires.

## Results

In the first stage, therapists successfully learned to independently operate the iARAT application within several sessions. However, using the IMU-based system added approximately 12 minutes compared to non-instrumented assessments. In contrast, the second stage revealed that camera-based assessments added only around 2 minutes compared to non-instrumented assessments. Consequently, most therapists expressed a clear preference for the camera-based solution.

Fig. 1 Learning curve of one therapist conducting seven iARAT sessions. Green horizontal lines represent time expenditure of an expert group (mean and two standard deviations). Red dots indicate that support was required during the session, red vertical lines indicate taking longer than expected expert time.



## Discussion

Camera-based assessments demonstrated superior usability and significantly lower time requirements compared to IMU-based systems, complementing their known advantages in accuracy. Future developments should prioritize automation of assessment workflows and enhancement of user experience, employing user-centered design approaches that actively involve therapists as end-users. This approach can be conducive to broader adoption and more seamless integration of instrumented assessments into daily clinical routines.

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## Investigation of the Efficiency of Home Exercise Program with Sensor System in Individuals with Scoliosis

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## Introduction

Scoliosis has many subcategories, the most common type is Adolescent Idiopathic Scoliosis (COBB angle >10°, vertebral rotation >4°). It is a species that can be seen in the 10-18 age group and has various types. (1) Conservative treatment is exercise and/or corset. There are seven different concept exercise types recommended by the SOSORT Association (3). In our study, we used the 'International Scoliosis Three-Dimensional Scoliosis Therapy' Schroth exercise configuration, which includes exercises specific to three-dimensional scoliosis. After receiving the diagnosis from the doctor, a daily program treatment exercise program was recommended, the exercises were not only analyzed clinically but also planned as an exercise program (1).

## Research Question

What is the difference in efficiency between the exercises specific to scoliosis performed in a clinical setting and the exercises performed in a home exercise program for individuals with scoliosis?

## Methods

The study included 4 volunteer girls diagnosed with scoliosis. Three exercises specific to scoliosis were selected that were suitable for the home program. The exercises called Sitting on a Ball/Chair, Kneeling on one knee, Between Two Poles. These exercises were stated as 2 sets, 8 breaths/repetitions. The Xsens sensor system was used in the evaluation. The locations where the sensors were positioned were determined in 4 areas (C7 vertebra, right and left acromion, T8-T10 level and L2-L3). Individuals were asked to perform the selected exercise without physiotherapist supervision and stimulation. (2 sets, 8 breaths) After resting, the individuals performed the following sets of exercises, including verbal and tactile stimulation, under the supervision of a physiotherapist. In comparing the angular data of exercises performed with physiotherapist commands and warnings and exercises without physiotherapist accompaniment, Student T-Test paired version was used.

## Results

As a result of the statistical data performed in the Student T-Test paired version, the p value was obtained below 0.05 in the angles obtained in the X and Z axes of the movements performed with and without physiotherapist accompaniment. Therefore, it was determined that there was a significant difference in the exercises performed with and without physiotherapist accompaniment obtained from the p value (with values of 0.0047 for the X axis and 0.028 for the Z axis). As a conclusion, the usability of the IMU sensor system in exercise evaluation with scoliosis has been demonstrated.

| X                | Unsupervised | Supervised |
|------------------|--------------|------------|
| Subject 1        | 52.842       | 9.574      |
| Subject 2        | 72.928       | 16.216     |
| Subject 3        | 46.188       | 17.594     |
| Subject 4        | 53.356       | 23.82      |
| P(T<=t) one-tail | 0.00473511   |            |

Table 1. Statistical Result of X-Axis Data of Exercises Performed with and without Physiotherapist Supervision. (p<0.05)

| Z                | Unsupervised | Supervised |
|------------------|--------------|------------|
| Subject 1        | 66.872       | 10.072     |
| Subject 2        | 86.218       | 26.126     |
| Subject 3        | 47.444       | 26.114     |
| Subject 4        | 58.23        | 47.178     |
| P(T<=t) one-tail | 0.02859002   |            |

Table 2. Statistical Result of Z-Axis Data of Exercises Performed with and without Physiotherapist Supervision. (p<0.05)

## Discussion

Postural wearable devices have the potential to assist healthcare professionals (doctors, physiotherapists, etc.). When the data of the evaluations were examined, it was observed that the angular data were clearly provided in the 3-dimensional scoliosis-specific exercise application of the IMU sensor system. It was determined that there were fewer angular changes in the exercises performed with the commands and warnings of the physiotherapist compared to the

exercises performed by the individual alone without receiving warnings and commands, and that there was a statistically significant difference.(4)

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## The influence of clothing on the accuracy of markerless 3D gait analysis using DeepLabCut

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## Introduction

Marker-based motion capture systems are the gold standard for three-dimensional (3D) gait analysis but require minimal clothing, limiting their use in real-life and clinical settings. While Van Hooren et al. showed that markerless systems offer a more practical solution by using video-based tracking, their accuracy under typical clothing conditions is not well established [1]. Keller et al. showed that when using Theia3D for 3D markerless gait analysis, the type of clothing (sport and street) affects the accuracy slightly, but does not affect the results for clinical interpretations [2]. However, it is unclear whether DeepLabCut, an open-source and low-cost approach, would present comparable results. Therefore, this study compares a marker-based and a markerless system under two clothing conditions, minimal and normal, to assess the markerless system's ability to capture 3D spatiotemporal and kinematic gait parameters.

## Research Question

Does 3D markerless motion capture using DeepLabCut provide gait parameters consistent with marker-based systems, and is this consistency affected by clothing?

## Methods

Twenty healthy adult participants (aged 20-55yrs, equally gender distributed) walked at a constant speed of 3.5km/h on a treadmill in three conditions: 1) minimal clothing and retroreflective markers, 2) minimal clothing, and 3) normal clothing (Figure-1). Marker-based gait analysis was performed using a Vicon motion capture system with the Conventional Gait Model 2.4 (CGM 2.4). Simultaneously, two Vicon Flir cameras, positioned on the left-side and at a 45°

posterior angle, recorded two-dimensional RGB videos for markerless analysis. Markerless gait analysis was conducted using DeepLabCut (v2.2rc3), an open-source computer vision platform.



Figure 1- Conditions: A- Minimal clothing and retroreflective markers B- Minimal Clothing C- Normal Clothing

Gait events will be identified and data will be time-normalized according to the method described by Zeni et al [3]. Differences between marker-based and markerless systems will be assessed with a paired approach for the spatiotemporal and kinematic parameters. The effect of clothing will be assessed in an unpaired manner in the markerless analysis. A (un)paired samples T-test will be performed for the step length, width, and height. The Root Mean Square Error will be calculated and Statistical Parametric Mapping will be used for the hip, knee, and ankle joint angles.

## Results

As data collection is still in progress, outcome measures will be made available at the time of presentation.

## Discussion

Directly comparing a marker-based and markerless system in conditions that do not involve markers is impossible, thus an indirect comparison has to be made. This comparison consists of the effect of clothing in the markerless analysis and the difference between marker-based and markerless system in the minimal clothing condition. We expect that the type of clothing has a larger effect than the difference between the marker-based and markerless system.

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## Dynamic postural stability and compensation strategy for balance deficit during lateral jump landing in individuals with chronic ankle instability

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## Introduction

Lateral ankle sprains are among the most common injuries in athletes, with up to 40% of first-time injured individuals developing chronic ankle instability (CAI). Most studies on dynamic balance in CAI individuals have focused on forward jump landing or drop landing and have reported a decrease in dynamic postural stability and changes in lower limb kinematics, such as a decrease in ankle dorsiflexion, knee flexion, and hip flexion [1, 2]. However, considering that ankle sprains occur most frequently in the frontal plane, dynamic postural balance in the lateral direction may provide important insights for injury prevention.

## Research Question

Do the CAI individuals show a decrease in dynamic postural stability and changes in lower limb and trunk kinematics during lateral jump landing compared with non-CAI individuals?

## Methods

Twenty-four individuals (32 feet in total; 13 males, 16 feet; 11 females, 16 feet) with CAI (CAI group [average age:  $21.0 \pm 1.3$ ]) and 32 (16 males and 16 females) controls without ankle sprains (non-CAI group [average age:  $20.9 \pm 1.0$ ]) participated in the study. The participants were instructed to stand at a distance equal to 33% of their body height away from the force plate (AccuGait, AMTI, USA), jump laterally over a 6-inch hurdle to the force plate, land on their test leg, stabilize as quickly as possible, and balance for 10 s with their hands on their hips [3]. The dynamic postural stability index (DPSI) for anterior–posterior, medial–lateral, and vertical directions and a composite index were calculated from the ground reaction forces collected at 200 Hz [4]. The lower limb and trunk kinematics during lateral jump landing were recorded using 3 smartphones (iPhone SE; Apple Inc., USA) and an OpenCap (Stanford University) application at 240 Hz. Ankle, knee, hip, and trunk angles were analyzed in the interval from 200 ms before to after the landing point. DPSI and joint angle data were compared between the CAI and non-CAI groups at a 5% significance level using Independent-samples t-tests.

## Results

There were no significant differences in DPSI in any direction or composite index between the CAI and non-CAI groups. Compared with the non-CAI group, the CAI group had significantly greater hip

flexion angles in the range of 29–61% (mean difference: 4.1°,  $p < 0.01$ ) and 77–99% (mean difference: 6.9°,  $p < 0.05$ ) and no significant differences in other lower limb and trunk angles.

## Discussion

Although many studies have reported a decrease in dynamic balance in individuals with CAI, our results showed no significant difference in dynamic postural stability between the two groups. This may be explained by the finding that the CAI group compensated for balance deficits by flexing the hip joints more deeply. Our results suggest the need to instruct different movement strategies in rehabilitation for balance impairments in CAI than in healthy individuals.

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## The influence of the ankle-foot orthoses on the gross motor function of children with neurological disorders

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## Introduction

Children with cerebral palsy (CP) or acquired brain injury (ABI) often experience motor impairments affecting gross motor function (GMF). Ankle-foot orthoses (AFOs) are frequently prescribed to enhance gait, improve postural stability, and prevent muscle contractures, but their influence on GMF capacity and movement quality outside of gait-related tasks remains understudied[1].

## Research Question

How do AFOs affect both the capacity and quality of GMF performance in ambulatory children with CP or ABI?

## Methods

This prospective within-subject study included 29 ambulatory children with CP or ABI (16 males;  $10.4 \pm 3.6$  years; GMFCS levels I–III), all of whom used AFOs daily. The majority (93%) wore rigid AFOs. Participants completed dimensions D (Standing) and E (Walking, Running, Jumping) of the GMFM-88 under two conditions: barefoot and while wearing AFOs. Movement quality was assessed using the Quality Function Measure (QFM), which evaluates five key attributes: alignment, coordination, dissociated movement, stability, and weight-shift[2]. GMFM and QFM scores were statistically compared between barefoot and AFO conditions using paired t-tests or Wilcoxon signed-rank tests, depending on data distribution. Subgroup analyses examined (1) dynamic tasks requiring ankle mobility and (2) stability-oriented standing and walking tasks.

## Results

No significant change was observed in total GMFM scores ( $p = 0.831$ ), though standing scores (Dimension D) declined slightly with AFOs ( $p = 0.031$ ). Among QFM attributes, alignment improved ( $p = 0.023$ ), while weight-shift decreased ( $p = 0.017$ ). For dynamic tasks, both GMFM and QFM scores declined significantly with AFOs ( $p < 0.05$ ), particularly in coordination, stability, and weight-shift. In contrast, performance in stability-oriented tasks improved with AFOs ( $p < 0.05$ ), notably in alignment and dissociated movement. All results are summarized in Table 1.

Table 1 | Statistics of comparison in GMFM and QFM attributes scores between barefoot and AFO condition

|                                                      | BF (mean $\pm$ SD)                 | AFO (mean $\pm$ SD)                | p-value | Difference (mean AFO-BF) |
|------------------------------------------------------|------------------------------------|------------------------------------|---------|--------------------------|
| <b>All items</b>                                     |                                    |                                    |         |                          |
| GMFM                                                 | 69.61 ( $\pm 39.41$ ) <sup>a</sup> | 66.67 ( $\pm 41.60$ ) <sup>a</sup> | 0.831   | -2.94                    |
| GMFM dimension D                                     | 77.78 ( $\pm 30.56$ ) <sup>a</sup> | 74.36 ( $\pm 29.91$ ) <sup>a</sup> | 0.031*  | -3.42                    |
| GMFM dimension E                                     | 60.61 ( $\pm 47.54$ ) <sup>a</sup> | 63.64 ( $\pm 45.56$ ) <sup>a</sup> | 0.422   | 3.03                     |
| Alignment                                            | 38.96 ( $\pm 22.21$ )              | 43.08 ( $\pm 21.61$ )              | 0.023*  | 4.12                     |
| Coordination                                         | 47.38 ( $\pm 24.90$ )              | 45.20 ( $\pm 23.72$ )              | 0.057   | -2.18                    |
| Dissociated movement                                 | 41.25 ( $\pm 24.16$ )              | 40.24 ( $\pm 22.45$ )              | 0.508   | -1.01                    |
| Stability                                            | 40.99 ( $\pm 23.98$ )              | 40.96 ( $\pm 22.71$ )              | 0.976   | -0.03                    |
| Weightshift                                          | 45.40 ( $\pm 22.75$ )              | 43.15 ( $\pm 20.25$ )              | 0.017*  | -2.25                    |
| <b>Dynamic tasks requiring high ankle mobility</b>   |                                    |                                    |         |                          |
| GMFM                                                 | 47.62 ( $\pm 33.81$ ) <sup>a</sup> | 43.56 ( $\pm 36.06$ ) <sup>a</sup> | 0.041*  | -4.06                    |
| Coordination                                         | 41.77 ( $\pm 26.13$ )              | 36.30 ( $\pm 23.67$ )              | <0.001* | -5.47                    |
| Dissociated movement                                 | 44.02 ( $\pm 26.19$ )              | 39.03 ( $\pm 24.73$ )              | 0.011*  | -4.99                    |
| Stability                                            | 32.57 ( $\pm 26.40$ )              | 27.85 ( $\pm 23.57$ )              | <0.001* | -4.72                    |
| Weightshift                                          | 41.87 ( $\pm 25.08$ )              | 36.60 ( $\pm 22.00$ )              | <0.001* | -5.27                    |
| <b>Stability-oriented standing and walking tasks</b> |                                    |                                    |         |                          |
| GMFM                                                 | 84.62 ( $\pm 28.04$ ) <sup>a</sup> | 87.18 ( $\pm 21.79$ ) <sup>a</sup> | 0.023*  | 2.56                     |
| Alignment                                            | 38.96 ( $\pm 22.21$ )              | 43.08 ( $\pm 21.61$ )              | 0.023*  | 4.12                     |
| Coordination                                         | 52.25 ( $\pm 24.30$ )              | 54.68 ( $\pm 25.50$ )              | 0.058   | 2.43                     |
| Dissociated movement                                 | 50.48 ( $\pm 28.83$ )              | 55.84 ( $\pm 26.66$ )              | 0.037*  | 5.36                     |
| Stability                                            | 45.85 ( $\pm 21.60$ )              | 49.26 ( $\pm 21.66$ )              | 0.011*  | 3.41                     |
| Weightshift                                          | 47.22 ( $\pm 21.63$ )              | 47.84 ( $\pm 19.80$ )              | 0.621   | 0.62                     |

Statistics for (top) all items of dimension D&E of GMFM; (middle) a subgroup of items (i.e. 52, 60, 61, 62, 63, 64, 77, 80, 81, 82, 83, 85, 87) that are dynamic tasks requiring high ankle mobility; and (last) a subgroup of items (i.e. 53, 54, 55, 56, 57, 58, 65, 66, 67, 68, 69, 70, 71, 72, 73, 74) that are stability-oriented standing and walking tasks. No differences exceeded the MDC threshold

Symbols: <sup>a</sup>=median-IQR \* = significant difference between barefoot and AFO condition with  $\alpha = 0.05$

Abbreviations: SD=standard deviation; IQR=inter-quartile range; BF=barefoot; AFO=ankle-foot orthosis; GMFM=gross motor function measure; MDC=minimal detectable change

## Discussion

AFOs influence GMF in a task-specific manner. They improved alignment and postural control during static or predictable tasks,

which is consistent with previous findings suggesting that AFOs enhance standing stability and support upright posture by promoting proper limb alignment and reducing compensatory movements[3]. In contrast, performance during dynamic activities that require ankle mobility, such as stair descent or jumping, was negatively affected. This finding is in line with studies by Buckon et al., who observed reduced coordination and movement quality in children wearing more restrictive orthotic designs[3,4].

While earlier studies have reported overall improvements in GMFM-scores with AFO use[5], our results showed no significant change in total GMFM scores and a slight decline in standing performance. This may be attributed to limited ankle mobility during tasks such as floor-to-stand transitions, particularly when rigid AFOs are used[6]. These findings highlight the importance of individualized orthotic decisions. AFOs offer clear benefits for stability and postural control in children with greater motor involvement but may hinder performance in more dynamic tasks, particularly in higher-functioning or older children. Clinicians should carefully balance the mechanical support provided by AFOs with the child's functional goals and need for mobility during everyday activities.

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## The immediate effect of myofascial release of the iliotibial band on knee kinematics in dynamic knee valgus

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## Introduction

Dynamic knee valgus (DKV) is a biomechanical deviation that occurs in three planes of motion in the lower extremity (1). The iliotibial band (ITB), an extension of the tensor fascia latae (TFL) and gluteus maximus muscles, contributes to the stabilization of the knee joint in the frontal plane. The literature, it has been reported that ITB flexibility alters DDV by affecting lower extremity alignment (1,2). Myofascial release may reduce tension in fascial layers, improve the sliding between tissues. Recently, self-myofascial release (SMFR) has become a commonly used intervention. Although SMFR applied to various muscles has been shown to affect lower limb kinematics, the evidence regarding its effect on the ITB remains limited (3). It is important to examine this movement mechanics in terms of prevention of knee injuries and treatment planning. The aim of this study was to investigate the acute effect of SMFR on knee kinematics in individuals with DKV.

## Research Question

Does self-myofascial release have an acute effect on knee kinematics in individuals with dynamic knee valgus?

## Methods

The study included 20 healthy women aged 18-25 years with symptomatic DDV (mean age  $21.6 \pm 1.79$  years). Frontal plane projection angle (FPPA) was analyzed using Kinovea® 2D video analysis software during the single-leg drop landing (SLL) and single-leg squat (SLS) tests. Reflective markers were placed on anatomical landmarks, and the mean of three trials was recorded. Participants with FPPA  $> 10^\circ$  during the SLS test were considered to have DKV (4). ITB flexibility was measured with the Ober test using an inclinometer (5). Participants were randomized into two groups. Participants in the SMFR group underwent SMFR along the ITB on the dominant side with foam roller. It was repeated five times with 1 minute of intervention and 30 seconds rest intervals (6). Measurements were repeated immediately after the intervention. The control group only participated in the evaluations. The results were analyzed with Mann-Whitney U and Wilcoxon tests.

## Results

The mean SLL, SLS and Ober test scores of the groups were similar before the intervention ( $p > 0.05$ ). The mean SLL test scores of the SMFR group decreased significantly after the intervention ( $p = 0.022$ ), whereas the mean Ober test scores increased

significantly ( $p=0.005$ ). In the control group, there was no statistical change in the mean scores of both tests ( $p>0.05$ ). In the intergroup comparison, the SMFR group was significantly superior in the changes in SLL and Ober test scores ( $p=0.003$ ,  $p=0.001$ , respectively).

## Discussion

The findings suggest that a single session of SMFR applied to the ITB acutely increases ITB flexibility and reduces DKV. Similar acute effects have been reported with SMFR applied to the gastrocnemius muscle (7). The results support the potential of SMFR as a preventive and therapeutic strategy for knee injuries, further research is needed to evaluate its long-term effects.

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## Stronger toe strength is related to enhanced subtalar joint function during counter movement jump in adolescents

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## Introduction

The countermovement jump (CMJ) is widely used to assess lower extremity performance, and toe flexor strength is a contributing factor to jump performance (1). Individuals with flatfoot demonstrate reduced toe strength and altered biomechanics (2,3). The biomechanical mechanisms underlying jump performance associated with toe muscles remain unclear. Multi-segment foot models offer detailed kinematic data to better understand foot–ankle function during dynamic tasks (4). This study investigated the relationship between toe flexor strength, subtalar joint function, and ankle kinetics during CMJ in adolescents.

## Research Question

Is toe flexor strength associated with hindfoot motion and ankle kinetics during a jump in adolescents?

## Methods

Eight adolescents (4 with flatfoot, mean Foot Posture Index-6 (FPI-6) score:  $8.5 \pm 1.73$ , and 4 typically developed, mean FPI-6 score:  $3.8 \pm 0.5$ ) were included in this study. Three countermovement jumps (CMJ) were recorded for each participant using a 3D motion analysis system (Vicon, UK) and force plates (AMTI, USA) with Plug-in Gait and Oxford Foot Model marker sets. The frame count between peak hindfoot-to-tibia eversion (HF-EVR) at take-off phase and followed first peak hindfoot-to-tibia inversion (HF-INV) angles during CMJ (Figure-1), ankle kinetics and toe strength values were the interested parameters. Toe and lesser toe strength were assessed in a seated position using a hand-held dynamometer (5) and adjusted by body weight (N/kg). Spearman's rank correlation analyses were performed to assess relationships between variables. The strength of correlations was interpreted according to Cohen's classification: weak ( $0.10 \leq r < 0.30$ ), moderate ( $0.30 \leq r < 0.50$ ), and strong ( $0.50 \leq r < 1.00$ ) (6).

## Results

A strong negative correlation was found between toe and lesser toe strength and the frame count from HF-EVR to the HF-INV while toe strength values demonstrated a strong positive correlation with ankle kinetics (Table-1).

Figure-1: Hindfoot-to-tibia frontal plane motion during CMJ

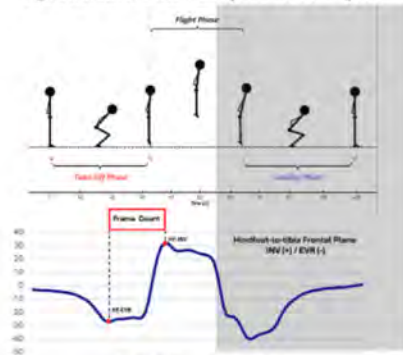


Table-1: Correlation analysis

| Parameters                            |                            | Frame Count<br>(from Peak HF-EVR to First Peak HF-INV) | Peak Ankle Power<br>(W/kg) | Peak Ankle Moment<br>(Nm/kg) |
|---------------------------------------|----------------------------|--------------------------------------------------------|----------------------------|------------------------------|
| Toe Strength<br>(N/kg)<br>n:16        | Correlation<br>Coefficient | -.854**                                                | .686**                     | .594*                        |
|                                       | Sig.<br>(2-tailed)         | <0.01                                                  | <0.01                      | 0.015                        |
| Lesser Toe Strength<br>(N/kg)<br>n:16 | Correlation<br>Coefficient | -.664**                                                | .819**                     | .703**                       |
|                                       | Sig.<br>(2-tailed)         | <0.01                                                  | <0.01                      | <0.01                        |

\* Correlation is significant at the 0.05 level (2-tailed).

\*\* Correlation is significant at the 0.01 level (2-tailed).

## Discussion

There are limited studies investigating the relationship between multi-segmental foot motion and intrinsic foot muscle strength in the context of jumping performance. The findings of this study suggest that greater toe strength contributes to more efficient subtalar joint function, as evidenced by a faster transition from hindfoot eversion to inversion and increased ankle power during CMJ. Also, better toe strength related with higher ankle power which contribute to enhancing jump performance. As contributing the gap in the literature, this study highlights the importance of assessing both foot muscle strength and multi-segmental foot kinematics to better understand their combined effects on lower limb function during CMJ. Future research with larger sample sizes and intervention-based studies targeting plantar muscle strength is warranted to further explore these biomechanical improvements and their potential clinical applications in adolescent populations.

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## Indirect improvement of multi-segmental foot kinematics via lower limb strengthening in an adolescent with flatfoot: A case report

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## Introduction to clinical case

Flatfoot is characterized by the collapse of the medial longitudinal arch, resulting in altered lower extremity alignment within the closed kinetic chain (CKC) (1,2). According to CKC theory, reactivating the gluteal muscles, knee flexors, and extensors through CKC exercises may improve lower limb alignment, enhance muscle recruitment patterns, and indirectly support foot function in individuals with flexible flatfoot (1,3). However, effects of CKC on foot biomechanics and performance in individuals with flatfoot remain underexplored. This case report aims to investigate the impact of lower extremity-focused exercises on foot-specific outcomes and jump performance in a flatfoot case.

## Methods

A 15-year-old adolescent boy with flexible flatfoot (BMI: 21.12 kg/m<sup>2</sup>, Foot Posture Index-6: 8) participated in this study. The participant completed an 8-week exercise program, consisting of one face-to-face and two online supervised sessions per week. Functional CKC exercises targeting the lower extremities including squats, lunges, heel raises, single-leg standing, were performed with an emphasis on foot core activation. Exercise intensity and complexity were progressively increased each week under the supervision of a physiotherapist. Three-dimensional gait analysis (Vicon, UK) using the Plug-in Gait and Oxford Foot Model marker protocols was conducted to assess multi-segmental foot motion, with one representative trial selected for comparison. Navicular drop, single heel rise test, and standing long jump test were also examined. Toe and lesser toe flexor strength were measured in a seated position with a hand-held dynamometer(N/kg)(4). Two measurements were taken with an 8-week interval. Graphical representations of the results were created using Microsoft Excel.

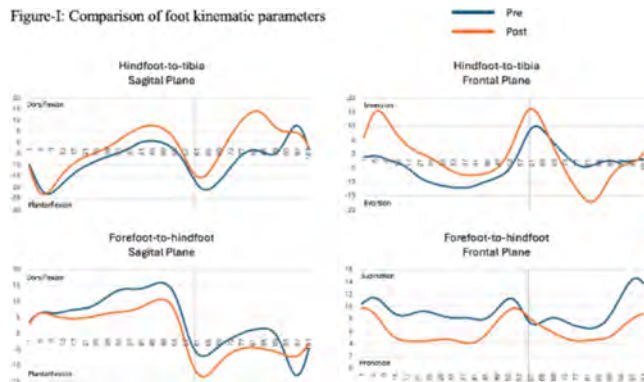
## Results

The navicular drop value decreased from 16 mm to 13 mm and the single heel rise test score increased from 32 to 35 repetitions. Foot kinematics showed a decrease in hindfoot-to-tibia plantarflexion and eversion, as well as in forefoot-to-hindfoot dorsiflexion and supination angles, while mean arch height increased (9.15 mm to 10.29 mm)(Figure-I). Gait velocity improved from 1.27 to 1.37 m/s, and ankle power increased from 4.98 to 5.26 W/kg. Additionally, improvements were observed in hallux and lesser toe flexor strength, as well as standing long jump distance, with gains of 7.2%, 10.5%, and 13%, respectively.

## Conclusion and clinical relevance

The findings suggest that lower extremity-focused CKC exercises contribute to enhanced foot kinematics which are altered due to flat foot (2), foot strength, and alignment by activating proximal muscle groups. This approach may serve as an effective complementary strategy for adolescents with flatfoot, improving both lower limb strength and foot-specific function while promoting program adherence. Further studies with control groups and larger samples are recommended to better determine its clinical relevance.

Figure-I: Comparison of foot kinematic parameters



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## Assessment of muscle activity in individuals with temporomandibular disorder using the Movement Deviation Profile

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## Introduction

Temporomandibular disorders (TMD) refer to a group of musculoskeletal conditions affecting the temporomandibular joints, masticatory muscles, and associated structures, leading to pain and dysfunction in the orofacial region [1,2]. For functional analyses, surface electromyography (sEMG) has been used to understand the masticatory muscle activity in patients with TMD [3]. However, up to this moment, no description of an EMG signal analysis method has been found that demonstrates clear differences between these patients and healthy individuals. In this context, the Movement Deviation Profile (MDP) emerges as a promising approach [4]. It is an innovative method that condenses and simplifies multivariate data through the use of a self-organizing artificial neural network, potentially offering a new avenue for the analysis and interpretation of muscle function in TMD.

## Research Question

Can the Movement Deviation Profile (MDP) differentiate individuals with and without TMD based on a set of sEMG variables?

## Methods

This was a cross-sectional study including 36 women with TMD (age:  $27.83 \pm 6.00$  years; height:  $1.70 \pm 8.56$  m; mass:  $67.23 \pm 7.78$  kg) and 24 healthy controls (age:  $26.32 \pm 4.72$  years; height:  $1.67 \pm 4.44$  m; mass:  $65.24 \pm 16.12$  kg). The sEMG signal was recorded from the right (RM) and left (LM) masseter muscles under resting (REST), mastication (MAST) and maximum habitual intercuspation (teeth clenching, MHI). The amplitude of the sEMG signal was determined using the root mean square (RMS) with a 200 ms moving window. The mean RMS values from the three mastication trials were normalized by the highest RMS value obtained across three maximum voluntary contraction (MVC) trials. The Movement Deviation Profile (MDP) [4] was utilized to reduce the dimensionality of the EMG signal through a self-organizing artificial neural network, and the Euclidean distance of the TMD group was measured from the control group. Subsequently, the MDP curve was averaged to create the MDPmean.

## Results

The Shapiro-Wilk normality test indicated that the data were not normally distributed. The TMD group presented significantly higher MDPmean values ( $p < 0.05$ ; Mann-Whitney test) in all comparisons,

with effect sizes ranging from moderate (0.30–0.50) to large (above 0.50).

Table 1. Median and interquartile range [25–75%] of the MDPmean recorded from the right(RM) and left(LM) masseter muscles during maximum habitual intercuspation (MHI), mastication (MAST) and rest (REST).

|             | TMD               | Healthy             | p value   | Effect Size |
|-------------|-------------------|---------------------|-----------|-------------|
| <b>MHI</b>  |                   |                     |           |             |
| LM          | 0.48 [0.31–0.74]  | 0.30 [0.25–0.44]    | 0.001*    | 0.40        |
| RM          | 0.35 [0.29–0.58]  | 0.24 [0.20–0.31]    | < 0.001*  | 0.46        |
| <b>MAST</b> |                   |                     |           |             |
| LM          | 0.38 [0.33–0.42]  | 0.32 [0.30–0.40]    | 0.02*     | 0.32        |
| RM          | 0.33 [0.29–0.39]  | 0.29 [0.25–0.34]    | 0.008*    | 0.34        |
| <b>REST</b> |                   |                     |           |             |
| LM          | 0.02 [0.007–0.10] | 0.004 [0.003–0.005] | < 0.0001* | 0.62        |
| RM          | 0.01 [0.007–0.03] | 0.004 [0.002–0.007] | 0.001*    | 0.60        |

\* statistically significant difference

## Discussion

These results demonstrate that the MDPmean can differentiate neuromuscular activity patterns of TMD patients from those of healthy individuals, whether assessed at rest, during MHI, or MAST. The elevated MDPmean values in the TMD group suggest a disruption in neuromuscular control and altered activation of the masticatory muscles, which may serve as an objective marker for diagnosing and monitoring TMD. This approach provides a comprehensive profile of muscle activity, and it may help guide more specific therapeutic interventions. Future studies should further investigate the MDP's diagnostic accuracy and clinical applicability in managing TMD.

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## Continuous Margin of Stability: a stride-to-stride interpretation of dynamic stability

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## Introduction

An individual's stability is vital to their health, indicating independence and fall risk in aging populations [1]. The Margin of Stability (MoS) has been adapted from static posture to walking but is limited to specific timed metrics [2]. Previous work suggests that quantifying changes in stability from one step to the next could insightfully assess an individual's stability by classifying their current step as stable, recovering, unstable, or destabilizing [3]. Building on this work, we propose extending the medial-lateral (ML) MoS reporting across the entire gait cycle as a stability waveform.

## Research Question

What is a useful method of visualizing the MoS during walking to assess an individual's dynamic stability?

## Methods

We acquired treadmill walking data for 18 aging adults (10 male, 8 female, age  $63 \pm 8$  years) from a public dataset [4]. Participants walked at 70% and 130% of self-selected walking speed (SWSS), and 4 participants used hand rail support. We computed the MoS using a Base of Support (BoS) defined by Heel, M1, and M5 markers [5]. Moments when the extrapolated Center of Mass (xCoM) was inside the BoS were noted as 0, while xCoM extending beyond the BoS to the right (left) was noted as positive (negative) MoS. Modeling and processing were done in Visual3D and Sift (HAS-Motion Inc.). We used a Python autocorrelation function on MoS signals, normalized to 101 points per stride, to find the appropriate lag value of an autoregressive model [6]. The MoS was plotted against its autoregressed version for visualization.

## Results

Our analysis found that the MoS autocorrelation between a current stride and the preceding one was strongest at the gait cycle's full length (lag of 100 points). Plotting this relationship showed a recurring pattern between quadrants 1 and 3. Higher walking speeds resulted in larger magnitude MoS scores and participants who held the railing had smaller magnitude MoS scores than participants who did not. Additionally, the MoS pattern of individual strides can be compared against those of a group, providing context for individual assessment.

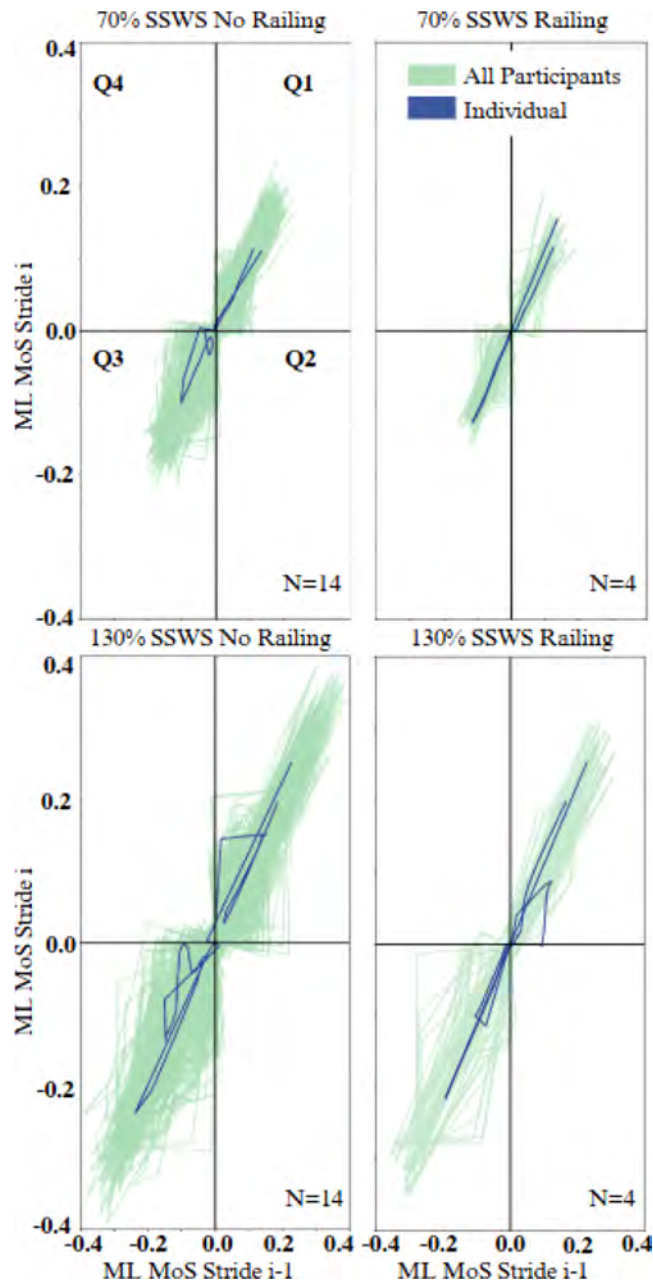


**Figure 1:** ML MoS of the current right stride (i) against the previous stride (i-1) for participants walking with and without railing support at 70% SSWS and 130% SSWS speeds.

## Discussion

Our method builds on previous work comparing MoS scores between consecutive heel strikes by visualizing these scores across the entire gait cycle. Although “unstable” walking is indicated by a negative or positive MoS, our method suggests the recurring pattern between quadrants 1 and 3 is a hallmark of stable walking. Maintaining this pattern may be more critical to stability during gait than by assessing a discrete MoS value. We should assess instability as the degree to which an individual deviates from this established pattern and shifts into quadrants 2 or 4.

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## Validation of Mobilize-D algorithm for estimating of cadence, stride length and walking speed in patients undergoing total hip arthroplasty

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## Introduction

Total hip arthroplasty (THA) is a common, successful and cost-effective surgery indicated for end-stage osteoarthritis to alleviate pain and restore function during activities of daily living [1]. The evaluation of patient function during ADL has been recommended as outcome for randomized control trials [2]. However, the function during daily life has been mainly studied with subjective measures either reported by patients or clinicians [3] despite significant disparity between objective and subjective measures of function [4]. Recently, the Mobilize-D initiative led to multiple recommendations for cadence (CAD), stride length (SL) and walking speed (WS) estimation with one inertial measurement unit (IMU) attached to the sacrum [5, 6] in multiple populations excluding patients undergoing THA.

## Research Question

Are the algorithms recommended by the Mobilize-D initiative validated for assessing CAD, SL and WL in patients undergoing THA?

## Methods

The study included 18 patients measured pre-surgery (n=18), 3 months post-surgery (n=9) and 12 months post-surgery (n=9). Measurements were performed concurrently with one IMU on the sacrum (Physilog6, MindMaze, Switzerland), and a twelve-camera OMC system (Oqus7+, Qualisys, Sweden) with a marker set following the Conventional Gait Model [7]. Participants were asked to walk barefoot at self-selected and slow speed along a 10m walkway. CAD, SL and WS were estimated with IMU based on the Mobilize-D algorithms [8] available in matlab (Matlab2021a, The MathWorks Inc., USA). For marker-based method, gait events were computed automatically [9] and manually checked. For each session of each patient, CAD, SL and WS were averaged for IMU and marker-based

and then compared with Bland-Altman methods (correlation, bias, limits of agreements [LoA]) and mean absolute error (MAE) expressed in measurement units and percentage of the mean of the marker-based method.

## Results

Strong to very strong correlations with limited bias (below 3%) were found for all parameters (Figure 1) with MAE reaching 3.4%, 8.4% and 10.8% respectively for CAD, SL and WS. However, despite LoA for cadence between 8% and 10%, the LoAs of SL and WS were between 21.6% and 31.5% of the mean marker-based values.

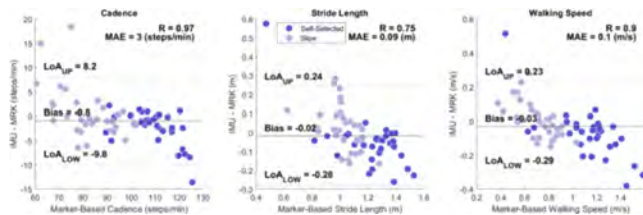


Figure 1: Bland-Altman plots of IMU versus Marker-Based method for estimating cadence, stride length and walking speed. LoA stand for limits of agreements, MAE stands for mean absolute error.

## Discussion

The correlations and MAE were consistent with previous validation studies [5, 6]. The high relative limits of agreements may limit the applicability of the tested methods for the analysis of gait speed with a single IMU in daily-living conditions.

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## Characterizing movement patterns in individuals with knee osteoarthritis using inertial measurement units

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## Introduction

Improved knowledge of movement-related changes in individuals with knee osteoarthritis (KOA) may enhance functional assessment and provide a deeper understanding of disease-related impairments. This study aimed to characterize movement patterns in individuals with KOA using inertial measurement units (IMUs) during a series of functional tests, and to explore associations between movement characteristics and molecular markers identified through targeted proteomics.

## Research Question

What characterizes movement patterns in individuals with KOA performing a series of functional tests?

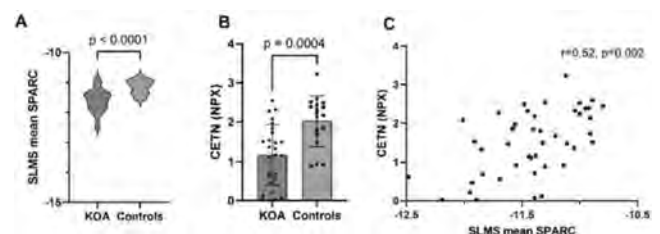
## Methods

Thirty-six individuals (mean age 63 years, 71% women) with KOA and nineteen healthy controls (mean age 62 years, 63% women) were evaluated during single sessions. Each session included a set of four performance-based tests to objectively assess joint function using IMUs: 10-meter fast-paced walk test (10MWT), 30-second Sit-to-Stand (30STS), Single Leg Mini Squat test (SLMS) and Jump for Distance (JFD) (1). Plasma samples were analyzed with Olink panels. This study focused on sagittal plane knee kinematics during the 30STS and SLMS. Smoothness of kinematic movement trajectories was assessed using Spectral Arc Length (SPARC) analysis.

## Results

The KOA group performed significantly worse compared to controls in all four functional tests. Smoothness metrics quantified using SPARC indicated better movement quality in the control group, with significant group differences in both mean SPARC (Figure A) and SPARC variance. Expression of the proteins CETN2 (Figure B) and PRTFCD1 differed between KOA and controls. Mean SPARC values during the 30STS and SLMS were positively correlated with the expression of CETN2 (Figure C) and PRTFCD1. In contrast, SPARC variance during SLMS and 30STS showed negative correlations with CETN2.

Figure A-C.



## Discussion

This study highlights clear differences in functional performance and movement quality between individuals with KOA and healthy

controls. IMU-based assessments quantified using SPARC revealed reduced movement smoothness and consistency in the KOA group. Interestingly, the mean SPARC values, reflecting overall movement quality, were positively associated with the expression of CETN2 and PRTFDC1, both downregulated in KOA. This suggests that lower levels of these proteins, which are involved in cellular regulation and metabolic processes, may be associated with movement-related impairments in individuals with KOA. Additionally, SPARC variance, which indicates within-subject variability in movement smoothness, showed negative correlations with the same proteins. Higher variance values, observed in the KOA group, indicate reduced movement repeatability. These findings support the potential of combining objective movement analysis with molecular profiling to deepen our understanding of movement impairments and underlying biological mechanisms in KOA.

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## A 3D gait analysis in a girl with right hemipelvectomy after Ewing's sarcoma. A case study.

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## Introduction to clinical case

Bone sarcomas are rare and occurs only 5% of malignancies in children and adolescents [1]. A 3 years old girl after a fall was diagnosed with Ewing's sarcoma in right side of her pelvis. At 4 years old she received a right hemipelvectomy (March 2021) and now she has 5cm leg length discrepancy (right limb shorter).

Picture 1. R hemipelvectomy



## Methods

The girl did a 3D gait analysis (December 2024, 7 years old) in ELEPAP Athens Gait & Motion Lab. The data which were collected includes clinical assessment and muscle strength test for lower limbs, kinematics, kinetics, spatiotemporal parameters and EMG data. Additionally, there was a comparison between gait deviation and asymmetry indexes while walking barefoot and with her shoes (the right has an external sole elevation of 5cm). The Gait Lab is equipped with 10 three dimensional Vicon Cameras (6 MX T10 & 4 Vero at 100Hz), 2 digital video cameras (Vicon Vue 720p at 100Hz), 2 force platforms (AMTI at 1000Hz) and 1 EMG Noraxon system with 8 sensors (2000Hz). The biomechanical model was the Plug in Gait with 16 markers.

## Results

In the clinical assessment she had a normal range of motion, however she had important muscle strength asymmetries in her right side. When she was walking barefoot she showed an excessive plantar flexion on her right ankle joint and asymmetries in most of her kinematics while when she was walking with her shoes most of the joints moved symmetrically.

## Conclusion and clinical relevance

In her case the external sole elevation seems beneficial for lower limbs kinematics, however she has acquired a great scoliosis. Her medical team has a dilemma between a limb lengthening at this stage or to wait for seven to eight years to adjust a prosthetic pelvic device [2]. This case study could launch a constructive discussion between the clinicians in order to propose what option would be preferable for the girl.

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## A musculoskeletal modeling approach in studying the biomechanics of clubfoot recurrence.

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## Introduction

Idiopathic clubfoot recurs in 20-40% of children, after initially successful treatment, at the age of six years [1,2]. Observational, imaging and molecular studies indicate that this recurrence is commonly linked to imbalances in the inverting and evertor muscle and ligament forces surrounding the foot, disrupting ankle-foot joint alignment during gait [3,4,5]. These studies are informative but insufficient to mechanistically link muscle weakness and deforming forces to recurrence risk. This study explored the validity of musculoskeletal modelling of the foot with personalized muscle-tendon properties to explore the biomechanics of clubfoot recurrence.

## Research Question

How do reductions in evertor muscle strength, increases in joint stiffness, and gradually recurring clubfoot deformities impact ankle-foot joint and muscle moment balances during gait?

## Methods

Optimal muscle fiber lengths and tendon slack lengths of a 4 degree of freedom musculoskeletal model of the ankle-foot complex were optimized in OpenSim-Moco by solving an optimal control problem with electromyographic tracking and evaluating normalized muscle-tendon lengths. Evertor muscle strength was systematically reduced by decreasing isometric muscle tension from 100% to 5% of the original value. Joint stiffness was simulated by decreasing the tibialis' posterior optimal muscle fiber length from 100% to 50% of its original value. Gradually recurring clubfoot deformities were simulated through systematic increases in ankle plantarflexion, subtalar inversion and midfoot supination from 0 to 18 degrees. For each degree of deformity, muscle moment arm balances and moment producing capacities were computed. Experimental data, including 3D segment positions, lower limb muscle activity, and ground reaction forces, from a typically developing child were used to run the simulations.

## Results

Muscle-tendon parameter optimization in OpenSim-Moco was feasible and improved physiological plausibility of inverse dynamic simulations. In the absence of deforming inversion forces, maintaining 30% of typical evertor muscle strength was sufficient to follow normal ankle-foot kinematics. However, when inversion joint stiffness was simultaneously increased by 20%, evertor strength demands increased twofold. Increasing inversion and plantarflexion joint stiffness leads to systematic increases in required evertor forces. At 30% inversion joint stiffness, the model was unable to follow the prescribed kinematics. Surprisingly, increasing clubfoot deformities did not disadvantage evertor moment arms compared to invertor moment arms.

## Discussion

Only in the presence of deforming inversion forces, evertor muscle weakness is a main contributor to the risk of clubfoot recurrence. Our findings emphasize the feasibility of using advanced musculoskeletal modelling to uncover mechanical factors that contribute to clubfoot recurrence. Incorporating muscle-tendon parameter optimization and manipulations of deep ankle-foot muscles increases the model's ecological validity, and its biomechanical and

clinical relevance of the results. This study establishes a foundation for future research incorporating patient-specific data to validate the clinical relevance of this approach, potentially leading to more personalized and preventive treatment strategies, improving long-term outcomes for children with clubfoot pathologies.

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## Assessing dynamic instability in bilateral vestibulopathy patients during daily living tasks using inertial measurement units

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## Introduction

Bilateral vestibulopathy (BV) has a severe impact on quality of life [1]. Symptoms, particularly imbalance, during the performance of daily living tasks (DLT) remains poorly understood. To our knowledge, existing research has focused mainly on the analysis of patient movements in controlled laboratory settings, leaving movements associated with DLT largely unexplored [2]. Inertial measurement units (IMUs) offer promising potential to fill this gap by enabling objective assessment of patient movements in daily life settings. They could facilitate screening and clinical follow-up on a larger scale, without the need for a specialized laboratory environment.

## Research Question

This study investigates the potential of IMU-derived parameters to highlight dynamic instability in BV patients during DLT.

## Methods

A cohort of 60 participants was examined, comprising 20 BV patients (11 females;  $61.0 \pm 11.6$  years), 20 unilateral vestibulopathy patients (UV) (10 females;  $59.5 \pm 5.3$  years) and 20 healthy subjects (HS) (10 females;  $57.9 \pm 5.2$  years). Nine IMUs were placed on anatomical segments (Figure 1) and participants performed 15 DLT in a semi-standardized environment. Parameters indicative of dynamic stability such as jerk [3], root mean square (RMS) of angular velocity [4], and mean of quantity of activity (QoA) [5] were calculated. Non-parametric ANOVA tests were used to evaluate the discriminating capacity of these parameters across groups and tasks.

## Results

Preliminary analyses identified 3 tasks as the most discriminative when analyzing trunk IMU data (walking on an inclined plane, walking on a beam and walking in darkness; Figure 1). The BV group showed significantly lower jerk values for the walking in darkness (BV/HS  $p < 0.01$ , BV/UV  $p < 0.01$ ) and for the walking on an inclined plane (BV/HS  $p < 0.01$ , BV/UV  $p < 0.001$ ) tasks. Conversely, jerk values were significantly higher for the walking on a beam task for the BV group (BV/HS  $p < 0.05$ ). Finally, significant differences were observed during the inclined plane task for RMS of angular velocity and mean of QoA (BV/HS  $p < 0.01$ , UV/HS  $p < 0.05$ ; BV/HS  $p < 0.05$ , UV/HS  $p < 0.01$  respectively).

## Discussion

While low jerk values typically indicate smoother movements, in BV patients they may reflect motor compensatory strategies aimed at minimizing abrupt movements due to vestibular deficits [6]. Conversely, healthy subjects can perform more rapid motor adjustments, resulting in increased jerking without loss of stability. This study reveals restricted movement patterns in BV patients, suggesting that rehabilitation efforts should focus on restoring movement fluidity and range of motion to mirror those of healthy individuals. Identifying and analyzing dynamic stability holds promise for advancing our understanding of vestibulopathy and monitoring the effectiveness of rehabilitation therapies.

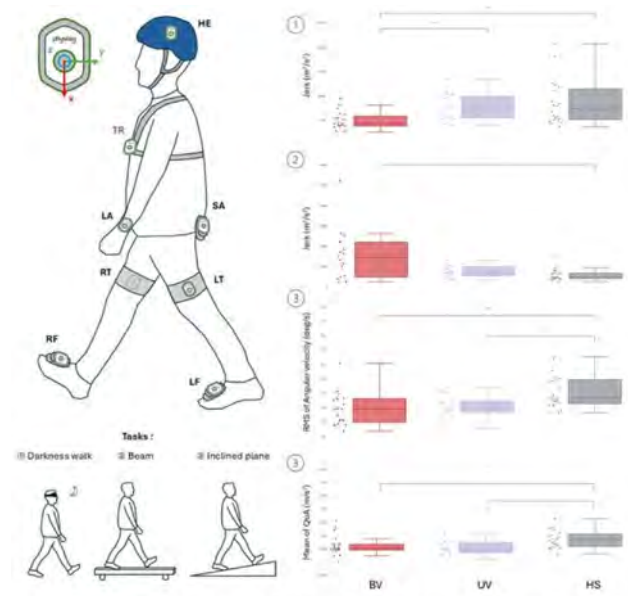


Figure 1. Inertial measurement unit setup; task descriptions; jerk, Root Mean Square of Angular Velocity, Mean of Quantity of Activity results for the Trunk (T9) IMU for the darkness walk, beam, and inclined plane tasks. A point corresponds to the individual results of one subject. Statistical analyses performed with a non-parametric ANOVA: \*  $p < 0.05$ , \*\*  $p < 0.01$ . HE: Head, TR: Trunk, SA: Sacrum, LRA: Left/Right arms, LRF: Left/Right legs, LRF: Left/Right feet, BV: Bilateral vestibulopathy group, UV: Unilateral vestibulopathy group, HS: Healthy subject group, QoA: Quantity of activity, RMS: Root mean square.

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## Investigation of the effects of hamstring stretching exercises on pelvic posture and biomechanics

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## Introduction

In normal posture, the pelvis has an anterior inclination angle of approximately 15°, which contributes to a lumbar lordotic curvature ranging from 30° to 50°(1). Short or tight hamstring muscles can pull the pelvis into a posterior pelvic tilt, which reduces the angle of pelvic inclination(2). Hamstring stretching has been widely utilized to increase pelvic inclination angle; however, evidence supporting its effectiveness remains limited. The aim of this study was to investigate the effect of hamstring stretching on sagittal-plane pelvic posture and gait mechanics.

## Research Question

Does unilateral hamstring stretching increase the pelvic anterior tilt angle (APTA) during standing and walking?

## Methods

The study included 21 healthy individuals (mean age:23,9 ± 1,83 years; 10 males, 11 females). Popliteal angles for hamstring muscle-tendon length were measured bilaterally, and the side with the larger angle was selected for stretching(3). Participants underwent passive static hamstring stretching (PSHS) protocol in a supine position, consisting of 30-second stretches with 15-second rest intervals, repeated six times(4). Stretch intensity was maintained at a level of 7–8 on a subjective discomfort scale. Anterior pelvic tilt angle (APTA) was assessed in the standing position with an inclinometer aligned unilaterally between the anterior and posterior superior iliac spines(5). Three-dimensional gait analysis was conducted both prior to the stretching protocol and within the first 10 minutes following its completion. Statistical analysis was performed using independent t-tests and Pearson's correlation coefficient.

## Results

Following the PSHS protocol, the popliteal angle showed a significant immediate decrease, dropping from 60.76 ± 7.50° to 48.42 ± 8.45° ( $p < 0.001$ ). Posture analysis showed a significant increase in APTA, measured by inclinometer (9.95 ± 4.54° to 13.66 ± 5.18°) and 3D motion analysis (9.4 ± 3.81° to 11.99 ± 4.01°;  $p < 0.001$ ). The APTA increase from stretching remained consistent during the gait cycle ( $p < 0.001$ ). There was no significant difference in walking speed ( $p = 0.256$ ) or hip extensor moments ( $p = 0.36$ ) before and after stretching(Table-1). A moderate negative correlation was found between the popliteal angle and APTA ( $r = -0.328$  with inclinometer,  $r = -0.415$  with 3D analysis).

**Table 1: Biomechanics of the Pelvis and Hip**

|                                   | Before Hamstring Stretching | After Hamstring Stretching | p             |
|-----------------------------------|-----------------------------|----------------------------|---------------|
| Peak Anterior Pelvic Tilt@%GC (°) | 12,27±5,03                  | 14,98 ± 4,89               | $p < 0,001^*$ |
| Peak Pelvic Tilt@ST (°)           | 12,03±5,03                  | 14,72 ± 4,85               | $p < 0,001^*$ |
| Peak Hip Flexion@ST (°)           | 31,17 ± 6,47                | 34,81 ± 5,91               | $p < 0,001^*$ |
| Peak Hip Extension@ST (°)         | -11,55 ± 5,61               | -9,10 ± 5,77               | $p < 0,001^*$ |
| Peak Hip Extensor Moment@ST       | -1,29±0,41                  | -1,30±0,37                 | 0,367         |
| Cadence (step/min)                | 118,36 ± 8,40               | 118,85 ± 8,47              | 0,102         |
| Walking Speed (m/s)               | 1,37 ± 0,10                 | 1,39 ± 0,12                | 0,256         |
| Step Length (m)                   | 0,69 ± 0,05                 | 0,70 ± 0,05                | 0,077         |

GC: Gait cycle, ST: Stance Phase of Gait Cycle, m/s: Meter per second, m: Meter.

## Discussion

In this study, the PSHS protocol increased hamstring length, consistent with findings reported in the existing literature(6). With

PSHS, the increase in APTA in pelvic posture is maintained throughout the gait cycle. Due to the lumbopelvic rhythm, the lumbar lordosis angle is expected to increase as the APTA increases(7). In conditions associated with decreased sagittal plane lumbar lordosis, such as scoliosis and hyperkyphosis, hamstring stretching should be considered as part of a corrective exercise program for improving sagittal plane spinal posture. Further studies are needed to evaluate the long-term effects of a repetitive and bilateral hamstring stretching exercise program on pelvic posture and mechanics.

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## Soft ankle exoskeleton for biplanar assistance of dropfoot with a human-in-the-loop optimization approach

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## Introduction

Exoskeletons are frequently lauded for their potential efficacy in rehabilitation and assistance in people with motor disorders [1,2] yet relatively few have convincing evidence for use. Dropfoot, or inability to dorsiflex the foot during swing and position the foot for contact, is a common secondary gait disorder after e.g., stroke. It is often accompanied by excessive subtalar inversion [3] due to spasticity of the tibialis posterior, which can affect stability in stance and risk of falling.

Using our custom-developed soft powered cable-driven ankle exoskeleton [4] designed to improve dropfoot gait by simultaneously assisting dorsiflexion and eversion, and a newly proposed multi-objective human-in-the-loop (HIL) that seeks improve gait symmetry and quality metrics, this study aimed to individualize exoskeleton support for individuals with dropfoot gait and to identify assistive profiles that enhance multiple gait metrics.

## Research Question

Does multi-objective HIL optimization lead to improvements in gait symmetry and quality metrics for people with dropfoot gait?

What is the self-reported satisfaction with the exoskeleton's usability?

## Methods

The exoskeleton we developed is comprised of motors worn on a backpack that control cables that span to the lateral and medial forefoot to dorsiflex and evert the foot. Five persons with chronic stroke donned the exoskeleton during 3D motion capture of over-ground walking. Three generations of exoskeleton control parameters were tested in the HIL optimization that sought to minimize 2 objectives - step length asymmetry and foot kinematics deviation - using each participant's response to guide the next generation. Participants also responded to a questionnaire QUEST 2.0 about their subjective experience with exoskeleton.

## Results

Gait symmetry and quality metrics improved with the exoskeleton (Fig. 1). A family of optimal solutions, converged towards low asymmetry and kinematics deviation, were obtained in the final generation for each participant; these represent a balanced trade-off that improve both gait quality metrics (optimization criteria) to varying degrees. Participants expressed positive satisfaction in terms of weight, stability and effectiveness (scores > 4/5), while comfort (3.6/5) and dimension (3.2/5) needs to be improved.

## Discussion

The exoskeleton successfully improved the specified gait quality metrics in persons with chronic stroke. To the best of our knowledge, this is the first individualized approach that seeks optimality of two objectives simultaneously. The generated family of optimal solutions can be considered as a group of “good” solutions and represent possible alternatives for individual goals and persons; one solution may be optimal when step symmetry is more important than foot orientation, and vice versa in another situation. Inherent in the HIL approach, optimal solutions are identified for each person and the specified goals; solutions will be different if different goals are specified, for instance maximum speed or minimum muscular effort.

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## ☆ Dystonic posture detection of the hand from common video recordings through MediaPipe and Convolutional Neural Networks

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## Introduction

Dystonia is a movement disorder characterized by involuntary movements and abnormal postures leading to severe limitations in mobility and manual function. In children, it is most commonly associated with dyskinetic Cerebral Palsy (CP) [1]. Accurate assessment of dystonia is crucial for selecting appropriate treatments and monitoring their effects over time. However, measuring dystonia is challenging as symptoms vary during the day and exacerbate with factors such as stress, emotions or noise. Current methods rely on expert visual scoring of videos, making them hugely time-consuming [2,3]. Current advances in computer vision offer potential to develop new movement analysis techniques for this group [4]. Here we explored a video-based deep learning approach to develop a methodology for automated dystonic posture detection.

## Research Question

Is it possible to train a deep-learning model to automatically detect dystonic hand posture from images extracted from standard video recordings?

## Methods

A total of 199 videos recorded during rest (each approximately 30 seconds long) collected at different timepoints were available from 94 participants with dyskinetic CP aged between 5-24 years, spanning all functional levels. In each video the section with the right hand was automatically detected using MediaPipe [5], by identifying the landmarks of wrist, index, thumb and pink. One or two images per video were cut depending on the variety of hand postures within the video, leading to 269 hand images. Additionally five hand images during the same resting task from five typically developing children (TDs) were added to the dataset, to allow the model to learn from the whole functional spectrum. The dataset of 274 images was then manually annotated by a trained assessor for "Dystonia YES" or "Dystonia NO", and split in a training- and test set (80%/20%), by subject-wise random split, with the images of TDs being forced into the training set. From the training set we took another 10% for internal validation. A binary image classification

model was trained using MobileNetV2 architecture [6]. Model performance was evaluated on the test set consisting of 55 images of 19 participants with dyskinetic CP. Precision, sensitivity, F1 score and specificity were calculated.

## Results

In the test set a precision of 0.77, a sensitivity of 0.91, a F1 score of 0.84 and a specificity of 0.47 was found. The confusion matrix of the results is shown in Fig.1.



Fig. 1. Confusion matrix of results testing image classification for automated dystonia detection. Examples of correctly identified hand images are framed in green and of incorrectly identified images in red.

## Discussion

The trained model performed very well in correctly identifying true positive instances (i.e. dystonia), with only some instances being incorrectly identified as dystonia without actually having dystonia. For clinical use the model will be further improved with a larger dataset and an optimal classification threshold will be defined. Then this approach can be applied to videos recorded at home by caregivers to enable an increase in frequency, ease and validity of dystonia assessments.

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## Impact of self-reported fatigue on walking capacity in adults with cerebral palsy

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## Introduction

Fatigue is a common complaint among individuals with cerebral palsy (CP), with a significant negative impact on quality of life (QoL). While interest in the adult CP population has grown in recent literature, this age group remains markedly under-studied. Only few investigations have explored the relationships between fatigue and walking capacity in adults with CP.

## Research Question

What is the impact of self-reported fatigue on walking capacity in adults with CP (aged 22–45), and how is fatigue associated with QoL?

## Methods

This cross-sectional study collected data on fatigue using the Fatigue Severity Scale (FSS) and walking capacity using the modified Gait Profile Score (mGPS), normalized walking speed, and 6-minute-walk distance (6MWD). Additional measures included self-perceived physical function (effort and pain during the 6-minute walk test - 6MWT) and health-related QoL assessed by the Short Form-36 (SF-36). Participants were categorized into two fatigue groups: Group A (No fatigue, FSS < 4) and Group B (Fatigue, FSS ≥ 4). Between-group differences were analysed using t-tests, and associations between FSS and other outcomes were examined using Pearson's correlation coefficients ( $p < 0.05$ ).

## Results

The study included 45 adults with CP (mean age  $27.6 \pm 5.1$  years), with 23 in Group A and 22 in Group B. Significant differences between the two groups were observed for: FSS ( $2.7 \pm 0.7$  in Group A vs.  $5.1 \pm 1.0$  in Group B), 6MWD ( $531.8 \pm 115.6$  m vs.  $372.0 \pm 260.3$  m), normalized walking speed ( $1.3 [1.2-1.4]$  (m/s)/LL vs.  $1.0$

[0.6–1.3] (m/s)/LL), mGPS ( $8.1^\circ \pm 2.1$  vs.  $10.6^\circ \pm 4.9$ ), Physical Component Summary (PCS) of the SF-36 ( $56.0 \pm 7.0$  vs.  $47.6 \pm 11.7$ ). Significant correlations were found between fatigue and 6MWD ( $r = -0.49$ ), PCS ( $r = -0.65$ ), modified GPS ( $r = 0.34$ ), and walking speed ( $r = -0.44$ ).

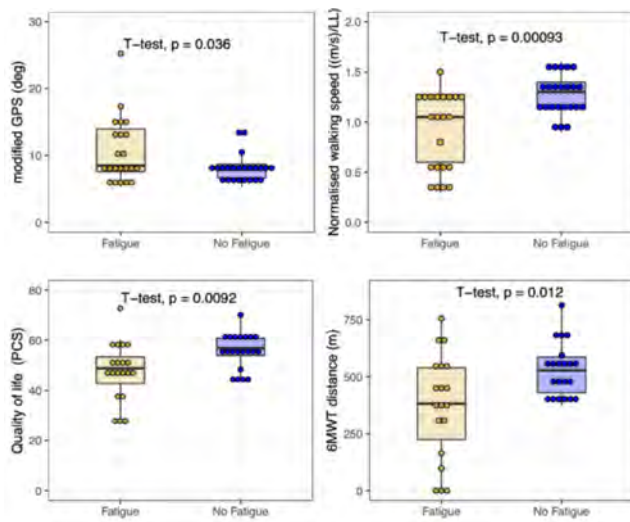


**Figure 1.** Comparison box plots between the group A (No fatigue, FSS < 4) and group B (Fatigue, FSS ≥ 4) for the modified GPS, normalized walking speed, PCS of the SF-36 and 6MWD.

Abbreviations: 6MWT, 6-minute-walk test; 6MWD, 6-minute-walk distance; FSS, fatigue severity scale; GPS, gait profile score; PCS, physical component score.

## Discussion

These results support associations between fatigue and different components of walking including 6MWD, speed, and modified GPS. The strong correlation between fatigue and the PCS of the SF-36 further underscores its impact on perceived physical health. These results suggest that fatigue negatively affects both walking capacities and QoL. Importantly, fatigue did not correlate significantly with pain or perceived exertion during the 6MWT, which may indicate an altered perception of physical exertion or a central origin of fatigue in this population.

Fatigue in adults with CP is strongly associated with reduced walking capacity and lower perceived physical health. These findings highlight the need to consider fatigue levels when designing personalized interventions aimed at improving mobility and quality of life in this population.

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## Acute effects of gait training watching virtual reality video of faster speed on comfortable walking

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## Introduction

Virtual reality (VR) has gained significant attention in rehabilitation, offering immersive environments for motor learning. In gait training, VR can simulate scenarios that are difficult to replicate in clinical settings. While some studies report positive effects of VR-based gait training, others show conflicting results, indicating the need for optimized settings [1]. This study examined the acute effects of VR-based gait training by focusing on the optical flow speed of VR videos. It investigated whether viewing VR videos faster than one's actual walking speed affects lower limb kinematics, spatiotemporal parameters, and subjective walking experiences.

## Research Question

Do gait training with VR video at a faster walking speed than the actual gait training speed lead to change gait parameters?

## Methods

Twenty-eight healthy male university students (mean age:  $21.6 \pm 0.7$  years) participated in this study. All participants performed 10 minute treadmill walking sessions three times using a Meta Quest 2 head-mounted display (Meta Platforms Inc., USA) with different VR visual flow speed conditions (corresponding to walking speeds of 2.5, 4.5, and 6.0 km/h, respectively). The treadmill speed was fixed at 2.5 km/h. To examine the immediate effects of gait training, spatiotemporal parameters and lower limb kinematics during comfortable walking were measured before and after gait training. The spatiotemporal gait parameters, including walking speed, stride length, stride time, stance time, and cadence, were assessed using ORPHE CORE 3.0 sensors and ORPHE ANALYTICS applications (ORPHE Inc., Japan) at 200Hz. The lower limb kinematics were recorded using three smartphones (iPhone SE, Apple Inc., USA) and an OpenCap application at 120Hz [2]. The peak ankle, knee and hip angle were calculated for a 5-step gait cycles. Subjective experiences, including VR sickness, fear of falling, stumbling, and fatigue, were evaluated using a Visual Analog Scale. Two-way repeated-measures ANOVA was conducted, followed by Holm's post-hoc tests [3]. The significance level was set at 5%.

## Results

The 2.5 km/h visual flow condition significantly increased walking speed, stride time, and stance time while reducing cadence

after 10 minutes gait training ( $p < 0.05$ ). No significant changes in gait parameters were observed in the 4.5 km/h and 6.0 km/h conditions. Lower limb kinematics showed no significant differences across conditions. Subjective experiences, including VR sickness and fatigue, were also unchanged.

## Discussion

The results suggest that when the visual flow speed matches the treadmill speed, participants exhibit a more cautious gait, characterized by longer stride time and reduced cadence. This may reflect an increased sense of physical stability. However, faster visual flow speeds did not further influence gait parameters, possibly indicating a lack of perceptual adaptation to incongruent visual input. These findings highlight the importance of aligning visual and motor stimuli in VR-based gait training [1]. Further studies should investigate long-term effects and clinical applications.

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## Patterns of gait stability in patients with Adult Spinal Deformity

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## Introduction

Patients with Adult Spinal Deformity (ASD) are known to present with postural malalignment in the frontal and sagittal planes. Subjects can maintain an upright posture with minimal energy expenditure inside the cone of economy described by Dubousset [1], where the center of mass (COM) can oscillate [2]. Due to their spinal malalignment, ASD patients require increased energy expenditure to maintain stability in this cone. While ASD patients often present with functional impairment [3], it is important to investigate factors influencing their stability during walking.

## Research Question

What are the factors contributing to COM deviation throughout the gait cycle in ASD patients?

## Methods

174 ASD and 44 controls underwent low dose biplanar X-rays with calculation of spinopelvic parameters (sagittal vertical axis SVA, pelvic incidence-lumbar lordosis mismatch PI-LL). They all filled the Oswestry Disability Index (ODI) questionnaire and performed 3D gait analysis to calculate kinematics of the head, trunk, pelvis, and lower limbs [4,5]. The COM was extracted during the gait cycle in the 3 directions and normalized to subject's BMI. Corridors of normality were established in the sagittal and horizontal planes and used to classify ASD patients into 4 groups: Inside corridors in both planes (Inside), outside sagittal corridor (Outside-Sag), outside horizontal corridor (Outside-Hor), outside corridors in both planes (Outside-Sag-Hor). Radiographic and kinematic parameters were compared between groups.

## Results

ASD patients were classified as 55 Inside (mean age:54y), 46 Outside-Hor (62y), 35 Outside-Sag (62y), and 38 Outside-Sag-Hor (66y). Abnormal sagittal COM trajectory was always below normative corridor. Outside-Sag-Hor patients had an increased SVA (67 vs 22mm), PI-LL (16 vs -3°; all  $p < 0.05$ ) and an increased ODI (47 vs 33) compared to the Inside group. During gait, Outside-Sag-Hor patients had an increased odontoid-hip axis angle both in sagittal (ODHA:10 vs 5°) and frontal planes (5 vs 2°) and an increased trunk flexion (13 vs 6°; all  $p < 0.05$ ). They presented a decreased ROM of the hip (34 vs 42°), knee (47 vs 59°), ankle (24 vs 29°) and foot progression angle (-16 vs -11°). They also had a decreased walking speed (0.64 vs 0.98m/s) and step length (0.44 vs 0.56m; all  $p < 0.05$ ). Spino-pelvic and kinematics of Outside-Sag and Outside-Hor groups ranged between the Outside-Sag-Hor and Inside groups.

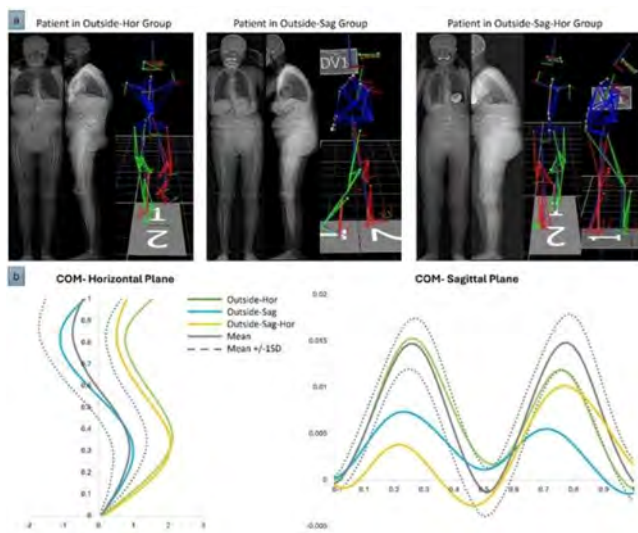
## Discussion

32% of ASD were able to walk within an economic zone, most probably due to a conserved muscle strength and their younger age. ASD with moderate sagittal malalignment tend to walk outside the COM corridor in one plane, underlying a more pronounced muscle weakness and skeletal abnormalities, causing insufficient compensatory mechanisms. Older ASD with severe sagittal malalignment present gait instability in both planes (Fig.1). Thus, gait stability seems to deteriorate with age and sagittal malalignment severity. Future studies should investigate whether early rehabilitation programs may help avoid migration from one group to another.

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Fig.1- Examples of patients from each gait instability group (a) and their COM trajectory in the horizontal and sagittal planes (b).



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## Effects of ballet training on kinematics and subjective parameters in children with idiopathic coxa antetortia: a pre-post intervention study

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## Introduction

Increased femoral anteversion (IFA) is rarely considered in healthy children and adolescents as it usually normalizes spontaneously. However, IFA can lead to several gait deviations, including in-toeing and increased knee flexion during mid stance (MSt). The available evidence indicates that IFA during childhood increases the risk of patella malalignment, femoropatellar complaints, and hip/knee osteoarthritis in adulthood. Therefore, IFA is not merely a cosmetic issue. Thus far, the impact of targeted training has not been investigated.

## Research Question

What are the effects of a six-week ballet training on kinematics and subjective functional mobility in children with idiopathic coxa antetortia (ICA)? Is the integration of a ballet training program into the everyday life of schoolchildren feasible?

## Methods

A pre(T1)/post(T2), intervention study without control was conducted, in which children with ICA participated in 1-h group ballet classes twice a week for six weeks. Additionally, the participants were provided with a diary containing instructions for home ballet exercises. A two-dimensional gait analysis was conducted on a C-Mill treadmill by Motek to measure knee flexion during the MSt and to record spatiotemporal gait parameters. The parent-reported Pediatric Outcomes Data Collection Instrument (PODCI) evaluated the functional mobility. The clinical examination entailed the measurement of hip muscle strength with a dynamometer DynaMo Plus by VALD, the assessment of hip joint mobility with a goniometer, and the Craig's test. The feasibility of the intervention was evaluated by analyzing six feasibility outcomes (Table 1).

## Results

5 healthy children (80% males;  $9.6 \pm 1.5$  years old) with femoral neck anteversion  $30^\circ < \text{FNA} < 40^\circ$  were included. There was a trend towards reducing excessive knee flexion during the MSt (median value  $17.7^\circ$  ( $15.7$ - $18.9^\circ$ ) at T1 and  $13.8^\circ$  ( $12.4$ - $17.2^\circ$ ) at T2;  $p=0.06$ ). Regarding the PODCI normative global functioning score (mean value  $49.2 \pm 6.3$  at T1 and  $54.4 \pm 3.5$  at T2;  $p=0.05$ ), subjective functional mobility improved. Self-selected gait speed increased significantly ( $1.05 \pm 0.13$  m/s at T1 and  $1.11 \pm 0.10$  m/s at T2;  $p=0.02$ ). The strength of the hip abductors left, the hip external rotation left and the external rotation right increased significantly. Table 1 summarizes the feasibility outcomes and criteria for success.

| Feasibility outcome                           | Criteria for success                                     | Actual level achieved | Criteria for success reached |
|-----------------------------------------------|----------------------------------------------------------|-----------------------|------------------------------|
| Recruitment and enrolment rates               | 25% of all eligible and contactable persons are enrolled | 45.5%                 | yes                          |
| Retention rates at each point                 | Drop-outs < 15%                                          | 100%                  | yes                          |
| Adherence to ballet training                  | group mean attendance rate is $\geq 80\%$                | 96.7%                 | yes                          |
| Adherence to home workout                     | group mean attendance rate is $\geq 80\%$                | 64.7%                 | no                           |
| Number of moderate or serious adverse event   | 0                                                        |                       | yes                          |
| Completed diaries returned after intervention | $\geq 80\%$                                              | 100%                  | yes                          |

Table 1; Summary of feasibility outcomes and criteria for success

## Discussion

The present pioneer study indicate that ballet training may be a useful intervention to reduce excessive knee flexion during MSt in children with ICA and improve functional outcomes, as well as subjective functional mobility. The integration of ballet training program into the children's everyday lives, is feasible and safe with some modifications. Ballet training is known to require high hip external rotation mobility and strength. Further studies including randomized controlled trials with larger sample sizes and three-dimensional gait analysis are required to gain understanding of the influence of ballet training on the reduction of FNA and compensatory gait mechanisms in children with ICA.

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## A delphi-consensus study about macroscopic muscle morphology in the clinical decision-making process of children with cerebral palsy

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## Introduction

Children with cerebral palsy (CP) present with altered macroscopic muscle morphology (MMM), such as decreases in muscle size and length, increases in tendon length, and changes in muscle architecture [1,2]. Although research regarding MMM properties in CP has increased over the last two decades, large heterogeneity exists in assessment methods and outcome parameters [3]. To bridge the gap between research and clinical practice, consensus on key MMM properties is required, informed by both evidence and expert opinion.

## Research Question

Can expert consensus be reached on clinically relevant MMM outcomes, their assessment methods, and interpretation in children with CP?

What are the most important research priorities to support clinical implementation of MMM in decision-making?

## Methods

A two-round Delphi study was conducted with 23 international experts in CP and MMM (clinicians and researchers). Survey development was based on an update of the review of Williams et al. [3], including 121 peer-reviewed studies [4]. The Delphi survey addressed eight subgoals (SG, Figure 1). Round 1 featured multiple-choice and open-ended questions; Round 2 included primarily statement-based items with some questions that remained or emerged after Round 1. Consensus was defined as  $\geq 75\%$  agreement.

## Results

Consensus was reached on 89% of the statements in Round 2 (Figure 1 for results per SG). The clinical relevance of MMM was confirmed, with agreement on their value across patient subgroups and functional levels (SG1, SG7). Muscle size and length were prioritized, and the importance of normalization and of assessing multiple muscles and tendon properties was emphasized (SG1). However, variation remained in preferred assessment technique, with a noted discrepancy between ideal modalities (MRI) and clinical feasibility (ultrasound) (SG2). Reliability was rated high for volume and length, but not for architectural parameters, whilst the discriminative validity was the strongest for muscle volume (SG3). Standardization of data acquisition, processing, and reporting was strongly supported (SG4). No consensus emerged on hardware and software choices, but advice for future users was formulated (SG5). MMM's relevance to clinical symptoms, function, and treatment history was affirmed (SG6). For clinical implementation, the importance of trained assessors and knowledge transfer was acknowledged; however, no consensus was reached on the timing and frequency of assessments (SG7). The value of normative and clinical datasets was stressed (SG8). Further research priorities identified included the creation of these large datasets (e.g., by data pooling) (SG8), validation studies (SG3) and studies on specific subpopulations like non-ambulatory children (SG7).

### Subgoals

- SG1: Define the most relevant macroscopic morphological muscle properties  
 SG2: Seek agreement on assessment techniques  
 SG3: Seek agreement on the level of evidence regarding reliability and validity of methods and outcomes  
 SG4: Define standardized operating procedures for the definition and reporting of outcomes  
 SG5: Define recommendations of hardware and software  
 SG6: Seek agreement on the importance of macroscopic morphological muscle properties in relation to the different ICF levels  
 SG7: Seek agreement on the concrete clinical implementation of macroscopic morphological muscle properties  
 SG8: Seek agreement on the need and size of normative and clinical databases



**Figure 1**  
 Top: overview of the eight subgoals. Bottom: number of statements (left) or questions (right) that reached or did not reach agreement per subgoal. Statements involved the following answer options: agree, disagree or no opinion. Questions involved open questions (no agreement can be obtained here) or multiple choice questions with the number of answer options ranging from 3 to 5.  
 Abbreviations: SG = subgoal; ICF = International Classification of Functioning, Disability and Health.

## Discussion

The results of this Delphi study contribute to the integration of MMM properties into the clinical decision-making process for children with CP. The findings underscore the need for and define standardization of outcomes, assessments and interpretation, and further research to enable consistent and evidence-based application of MMM in clinical settings.

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## ☆ Self-reported leisure time activities, sports participation and standing frame use in adults with cerebral palsy

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## Introduction

Both children and young adults with cerebral palsy (CP) show lower levels of participation and exhibit higher rates of sedentary behaviour (SB) [1,2]. Yet, participation in leisure time activities (LTA) is important for overall well-being. While transitioning into adulthood, participation is further complicated by a shift in autonomy from parents to the person with CP [3]. For adults with lower functional abilities (gross motor function classification system [GMFCS] levels III–V), the use of a standing frame can be used to decrease SB, improve fitness and participation [4]. However, the influence of GMFCS-level and age on LTA, sport and standing frame use in adults with CP has not been examined yet.

## Research Question

What are the self-reported LTA, sports participation and standing frame use in adults with CP? Is there an influence of GMFCS-level and age?

## Methods

This retrospective study used data from consultations at the CP reference centre. The population consisted of 500 adults with CP

(≥17y), equally distributed across the GMFCS-levels. Data extracted from clinical files included age, GMFCS-level, LTA including sports, and use of a standing-frame. Participants were divided into two age-groups (≤25y and >25y). A Chi-Square test was performed to compare age-group proportions between GMFCS-levels, and to compare participation in LTA, sports and the use of a standing frame between GMFCS-levels and age-groups. Post-hoc analysis consisted of the z-test of two proportions with a Bonferroni correction.

## Results

The dataset consisted of 268 males and 232 females with an average age of  $28.4 \pm 8.2$ y. Age-group proportions did not differ between GMFCS-levels. Participation in at least one LTA was reported by 57% of all participants, and 50% reported participating in sport(s). LTA and sports showed significant differences between GMFCS-levels, with reported participation in LTA being 80% and 28% and in sports 79% and 21% in GMFCS I and V, respectively ( $p < 0.001$ , Figure 1A–B). There was no significant difference in reported LTA and sports participation between adults ≤25y (55% and 47%) and adults >25y (58% and 52%, Figure 1C–D). For standing frame use, there was a significant difference between GMFCS-levels ( $p = 0.018$ ) and age-groups ( $p < 0.001$ ), with higher use in less functional adults (GMFCS III–IV–V: 43%–60%–59%) and in the younger age-group (≤25y - >25y: 71%–43%).

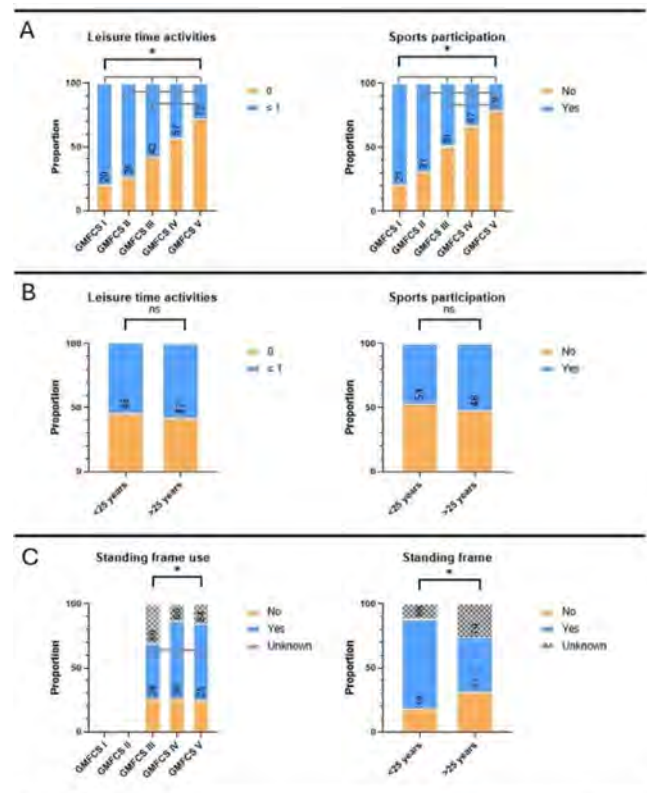


**Figure 1** Results of Chi-squared and post-hoc tests on comparisons of leisure time activities and sports participation between functional levels (A) and age groups (B) and standing frame use between functional levels and age groups (C). **Black brackets** indicate results of overall Chi-square test ( $p < 0.05$ ). **Gray brackets** indicate significant differences of post-hoc analysis with z-test ( $p < 0.05$  with Bonferroni-correction). Numbers inside the bars present the proportion that does not participate in leisure time activities or sports (part A and B) or the proportion that does not use a standing frame and the combined proportion of those that do not and do use a standing frame.

Abbreviations: GMFCS = Gross Motor Function Classification System; ns = not significant; \* = significant.

## Discussion

Only 57% of adults reported participation in LTA and only 50% in sports, with a significant influence of GMFCS-level. Standing frame use appears an effective intervention to influence SB in adults with higher GMFCS-levels, but the decrease in use with increasing age is alarming. Strategies to improve participation and address barriers are needed to facilitate engagement in recreation and physical activity in adults with CP across all GMFCS-levels. Further research should include frequency, duration and type of LTA and standing frame use, and more structured assessments like questionnaires and physical activity trackers.

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## A markerless gait analysis in a child with achondroplasia. A case study

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## Introduction

Achondroplasia (ACH), the most common form of dwarfism (approximately 1 in 25,000 births), is a genetic condition characterized by short stature, disproportionately short limbs, and joint misalignments, which can all significantly affect gait and mobility (Pauli, 2019; Broström et al., 2022). Gait analysis in individuals with ACH is often performed using marker-based motion capture systems. However, these systems face limitations such as soft tissue artifacts and challenges in accurately identifying anatomical landmarks, particularly in individuals with atypical skeletal structures. Markerless motion capture systems like Theia3D (Theia Markerless Inc., 2025), which use artificial intelligence to estimate joint centers and segment positions, present a promising alternative. However, research on their accuracy in cohorts with atypical body structures remains limited.

## Research Question

Therefore this study aims to evaluate the feasibility of using the markerless system Theia3D to conduct gait analysis in a child with ACH.

## Methods

One child with ACH (female, age 10 years, height 115.8 cm, weight 34.04 kg) participated in this study. Gait trials were recorded using a Vicon motion capture system consisting of 13 infrared cameras (200 Hz) and 6 video cameras (100 Hz). A modified Plug-in Gait model (Stief et al., 2013) was used for marker-based analysis. The participant walked a 10-meter aisle at a self-selected speed and completed ten trials with one gait cycle being extracted per trial. The lower body joint angles over the whole gait cycle in all three planes were processed using Vicon Nexus, and the same trials were analyzed with Theia3D (v2023.1.0.3160) to obtain markerless motion capture data. The agreement between the two systems was assessed using intraclass correlation coefficients (ICC), root mean square error (RMSE), and standard error of measurement (SEM).

## Results

Theia3D failed to achieve strong agreement with the marker-based system across most gait parameters. Only knee movement in the sagittal plane showed moderate ICC values (see Table 1). RMSE values were particularly high for sagittal and transverse plane measurements, with the largest errors observed in pelvic movement. SEM values also indicated significant measurement variability, especially in pelvic parameters within the sagittal plane.

## Discussion

The results suggest that markerless motion capture, as implemented by Theia3D, is not currently suitable for analyzing gait in a child with ACH. The system struggled to accurately calculate pelvic motion and did not adequately account for the participant's disproportionate limb lengths and joint misalignments. These limitations indicate that Theia3D's algorithms require further refinement to improve accuracy and reliability for individuals with rare conditions like ACH. Future research should focus on adapting markerless motion capture systems to better accommodate atypical body structures.

Table 1 Results of the lower body joint angles over the gait cycle for SEM=Standard Error of Measurement, RMSE=Root Mean Square Error, ICC=Intraclass Correlation Coefficient (poor: ICC < 0.5; moderate = 0.5 < ICC < 0.75; good = 0.75 < ICC < 0.9; excellent = ICC > 0.9) in all three planes.

| Parameter          | SEM  | RMSE  | ICC   |
|--------------------|------|-------|-------|
| <i>Sagittal</i>    |      |       |       |
| Pelvic             | 1.83 | 29.70 | -0.01 |
| Hip                | 3.25 | 24.35 | 0.04  |
| Knee               | 1.50 | 13.77 | 0.63  |
| Ankle              | 1.27 | 7.71  | 0.46  |
| <i>Frontal</i>     |      |       |       |
| Pelvic             | 0.96 | 0.98  | -0.22 |
| Hip                | 0.93 | 6.24  | 0.03  |
| Knee               | 2.79 | 4.18  | 0.15  |
| Ankle              | 4.96 | 3.99  | -0.68 |
| <i>Transversal</i> |      |       |       |
| Pelvic             | 9.43 | 11.29 | 0.31  |
| Hip                | 0.96 | 0.98  | -0.22 |
| Knee               | 3.27 | 13.14 | -0.02 |
| Ankle              | 1.52 | 10.75 | 0.14  |

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## Variabilities in both Modified Ashworth Scale scores and biomechanical parameters among different physiotherapists when assessing lower limb spasticity

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## Introduction

Spasticity, common in central neurological disorders, is characterized by an increased resistance during rapid mobilization due to an increased stretch reflex. This resistance is assessed using the Modified Ashworth Scale (MAS score) [1]. However, the inter-operator repeatability of this score is moderate for the lower limb [2]. This study investigates the relationship between quantitative biomechanical parameters (including intersegmental moments) and the MAS score of one patient assessed by several physiotherapists.

## Research Question

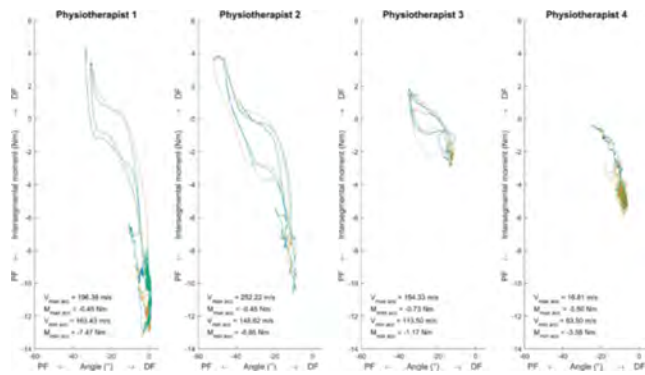
Do physiotherapists produce similar biomechanical parameters when quantitatively assessing spasticity?

## Methods

Four physiotherapists (PTs) assessed the spasticity of the right lower limb of one male patient (mass: 58.8kg, height: 1.75m, pathology: serious head injury). Five muscle groups (tibialis anterior, triceps, soleus, quadriceps and hamstring) were successively assessed using the MAS score. Electromyography was used to record muscle activity. A 6-axis dynamometer was used to measure forces and moments applied by the physiotherapists, and motion analysis was performed. The EMG onset of the involuntarily contraction of the agonist muscle was determined visually [3]. Patient's joint angles and velocities were obtained by inverse kinematics and intersegmental moments by inverse dynamics. Intersegmental moments and joint angles were plotted against each other, and the angular velocities and the intersegmental moments were quantified at maximum acceleration and at maximum deceleration [4].

## Results

For the soleus muscle, there were variabilities in both MAS scores and biomechanical parameters between PTs (Fig. 1). The MAS scores were 1+, 1+, 0 and 2 respectively for PT1, PT2, PT3, and PT4. Clinical experiences of physiotherapists in spasticity assessment were 15, 7, 7 and 3,5 years, respectively. PT1 and PT2 showed greater intersegmental moments and greater range of motion than PT3 and PT4. Furthermore, velocities at maximum acceleration (or deceleration) were greater in PT1 and PT2 vs PT3 and PT4. Intersegmental moments of PT1 and PT2 were greater at the maximum deceleration but smaller at the maximum acceleration. Biomechanical parameters from physiotherapist 4 stood out, considering the low velocities reached at maximum acceleration and deceleration and the high intersegmental moment reached at maximum acceleration. EMG onsets were only obtained for the first physiotherapist which agreed MAS scores. There were no EMG onsets for the other PTs but only PT3 showed consistent MAS score.



Intersegmental moment as a function of the ankle angle for the four physiotherapists

## Discussion

This study highlights variabilities for both MAS scores and biomechanical parameters for several physiotherapist assessing spasticity of a single patient. These results could be influenced by the experience of the physiotherapists, the use of the dynamometer, as well as a progressive change in muscle tone due to repetitive assessments. Future studies are welcome to further analyse and attempt to identify additional biomechanical parameters that may explain lack of consistency between MAS scores, biomechanical parameters and EMG onsets.

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## A gait lab in your pocket? Accuracy and reliability of monocular smartphone-based markerless 3D gait analysis in pathological gait

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## Introduction

Recent advances in 3D markerless motion capture are sparking hopes of revolutionizing gait analysis by offering a cost-effective, fast, and accessible alternative to traditional marker-based systems. Tools like OpenCap [1] or Pose2Sim [2] reconstruct 3D motion from calibrated smartphone videos, though they still require at least two recording devices. However, progress in computer vision, especially in 3D skinned multi-person linear models (SMPL) that parametrically represent human shape and pose [3,4], enables full-body 3D pose estimation from a single camera. If shown to be accurate and reliable, this technology could significantly broaden access to gait analysis across diverse settings by bringing gait analysis to a single smartphone.

## Research Question

How accurate and reliable is a monocular markerless approach to estimate 3D joint kinematics?

## Methods

We re-used videos and marker-based motion capture data from two studies assessing the validity and test-retest reliability of OpenCap [5,6]. The validity dataset included 18 healthy participants walking with four patterns: physiological, crouch, circumduction, and equinus. In this dataset 3D kinematics were simultaneously recorded using OpenCap and a state-of-the-art marker-based system. For reliability, 19 participants were instructed to walk physiologically on two separated days. In both studies OpenCap (v0.3) was used in its standard two-iPhone setup. We re-processed all videos from one phone using cameraHMR [4] to estimate the 3D pose from each participant (Fig. 1E). Virtual bony landmarks were defined on the SMPL model to extract marker coordinates per frame, which were then input into OpenSim's inverse kinematic tool using the same OpenSim model as OpenCap to obtain 3D gait kinematics. Validity outcomes included root-mean-square-error (RMSE) of the kinematic waveforms compared to the marker-based system; reliability outcomes included between-session RMSE of the kinematic waveforms and the standard-error-of-measurement (SeM) of discrete kinematic parameters. All metrics from cameraHMR and OpenCap were compared for left and right pooled time-normalized data.

## Results

CameraHMR's validity and reliability was in general similar to OpenCap, with slightly worse validity results for crouch gait ( $p = 0.031$ ,  $+0.7^\circ$  RMSE) and some differences in discrete parameters for reliability (Figure 1A-D).

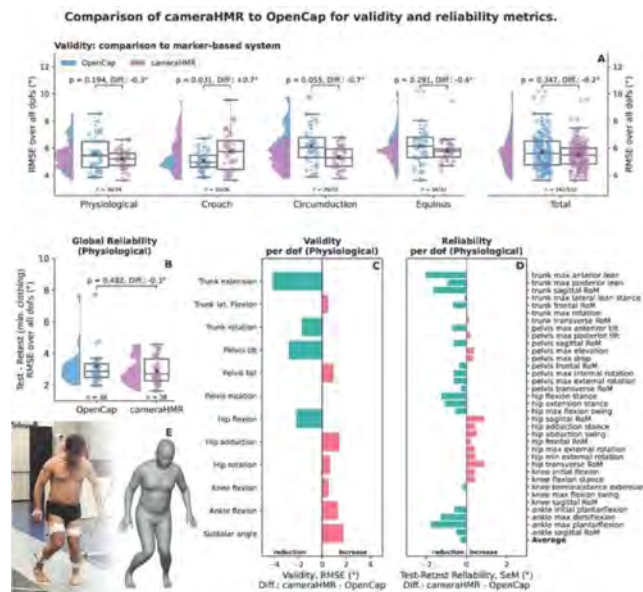


Figure 1. Validity and reliability between OpenCap and cameraHMR. Bottom-Left: example of an SMPL-based monocular 3D pose and shape estimation.

## Discussion

Results suggest that a single-camera setup can perform nearly on par with OpenCap's two-device system in terms of validity and reliability, enabling 3D gait analysis virtually anywhere. In some cases, cameraHMR even outperformed OpenCap in tracking joint kinematics but struggled, e.g. with accurate ankle joint tracking (Figure 1C). Single-camera 3D motion capture could greatly expand access to clinical gait assessment and enable home-based monitoring. The bad news: while reliability seems a strength of markerless methods, and validity looks promising, overall accuracy still requires improvements (preferably below  $2.5^\circ$ ) before it can be used in clinical applications.

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## Frontal plane X-ray-based regression equation for more accurate hip joint center localization in gait analysis of children with Achondroplasia

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## Introduction

In Achondroplasia (ACH), the most frequent form of dwarfism, along with a disproportional short stature, knee malalignments are common. Typically, radiographs are used them but 3D gait analysis (3DGA) is getting more popular [1-3]. Yet, to locate the hip joint centers (HJC), conventional gait models use equations derived from typically developed [e.g. 4-6] but skeletal disproportions may contribute to a misplacement of HJC in ACH.

## Research Question

What is the agreement between frontal knee angles from radiography and from 3DGA in ACH using different HJC estimation techniques?

## Methods

We retrospectively included 46 measurements of 37 patients with ACH (6-18 years) and 23 pediatric patients with idiopathic knee malalignments who had a 3DGA (Vicon Nexus) and a frontal plane long leg x-ray within  $\pm 4$  months. First, data was analyzed with a mod. Plug-in-Gait model using the Davis [4] and Hara [5] HJC regressions. In addition, we created an ACH-specific regression (Achreg) using the radiologically measured distance between femoral heads [7] ( $\Delta$ HJC, Fig. 1A). Thus, a multiple linear regression was performed to predict the HJC location based on thorax length (TL), leg length (LL), pelvic width (PW) and pelvic depth (PD). Differences between mechanical femoro-tibial angle (mFTA) from radiography and knee angle from 3DGA were compared between the different methods.

## Results

Leg length was  $39 \pm 2\%$  body height in ACH and  $54 \pm 1\%$  in ID ( $P < 0.001$ ). HJC distance in relation to pelvic width in ACH was 10.8% larger (Fig. 1C,  $p < 0.001$ ), showing that the HJC are located more laterally. The HJC regression equation with TL and PW had the highest  $R^2$  (Fig. 1D) and particularly the equation by Hara [5] showed misfits in ACH. Repeated measures ANOVA revealed different knee angles between all regressions ( $p < 0.001$ ), with an absolute difference between mFTA and knee angle from 3DGA of  $3.7 \pm 3.1^\circ$  for Davis,  $6.4 \pm 3.1^\circ$  for Hara and  $3.0 \pm 2.2^\circ$  for the new equation in ACH. In ACH the joint line convergence angle (JLCA) was a confounder, as an opening of the joint surfaces was correlated to the discrepancy ( $r = 0.33$ ,  $P = 0.001$ ).

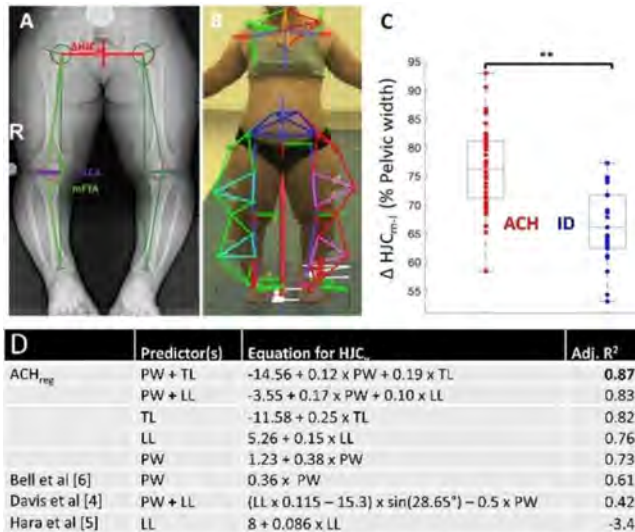


**Fig. 1** (A) X-ray with mechanical femuro-tibial angle in a patient with Achondroplasia (mFTA) (green), distance between hip joint centers ( $\Delta$ HJC) (red), joint line convergence angle (JLCA) in purple (B) 3DGA standing trial (C)  $\Delta$ HJC in children with Achondroplasia (ACH) and idiopathic deformity (ID) (D) Comparison of regression equations for medio-lateral HJC location.

## Discussion

Due to the distinct pelvic morphology in ACH, the m-l distance between the HJC is greater compared to typically grown individuals, highlighting the need for ACH-specific regression methods. Although the method proposed by Hara [5] has been shown to be beneficial in individuals of average stature, it performed poor in ACH patients, likely due to their disproportionately short lower limbs. The new ACH-specific regression equation results in a significantly smaller discrepancy between x-ray and 3DGA and may hence offer improved support for monitoring knee malalignments in ACH. The remaining discrepancy observed in ACH was partially explained by the JLCA, an indicator of frontal plane knee instability. In future, also functional calibration of the hip and knee should be further tested in ACH.

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## Impact of Spasticity Markers during Gait on Daily Mobility in Children with Cerebral Palsy: Insights from EMG and Musculoskeletal Modeling

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## Introduction

Spasticity is caused by stretch hyperreflexia [1] and is still primarily assessed clinically through manual evaluation of resistance to passive stretch. However, these assessments are criticized for being subjective and may not fully capture the extent or functional impact of the impairment. This is particularly relevant for bi-articular muscles, which play a key role in motor control but pose diagnostic challenges due to their complex behavior. To improve objectivity, instrumented methods such as EMG-based assessments have been proposed [2-4]. These approaches may offer more reliable insights, especially if they reflect relevant impairments that affect daily mobility, thereby enhancing their clinical validity and ecological relevance.

## Research Question

Are dynamic gait-derived markers of spasticity linked to daily mobility in children with BSCP?

## Methods

59 children with bilateral spastic cerebral palsy (BSCP;  $8.1 \pm 4.5$  years; GMFCS I: n = 12, II: n = 28, III: n = 19) and 30 age-matched typically developing (TD) underwent 3D gait analysis (3DGA) using Vicon Nexus with integrated surface EMG and musculoskeletal modeling. Muscle-tendon unit (MTU) lengths of the rectus femoris, semitendinosus, and gastrocnemius muscles were normalized to segment length, and stretch velocities were obtained by temporal

differentiation. EMG envelopes (RMS-EMG) were computed, and statistical parametric mapping (SPM1D) was applied to compare RMS-EMG and MTU velocity profiles. EMG/MTU velocity ratios were calculated during swing phase and compared between groups using ANCOVA with walking speed as Co-Variate. In BSCP, associations with caregiver-reported mobility (MobQues47-Questionnaire) were examined via bi-variate correlations.

## Results

Children with BSCP walked significantly slower than TD and displayed significantly reduced daily life mobility. Peak MTU stretch velocities were lower in children with BSCP (Fig 1a) than in TD for gastrocnemius, semitendinosus and rectus femoris while EMG was larger (Fig 1b) in late swing for gastrocnemius, semitendinosus and in midwing for rectus femoris.

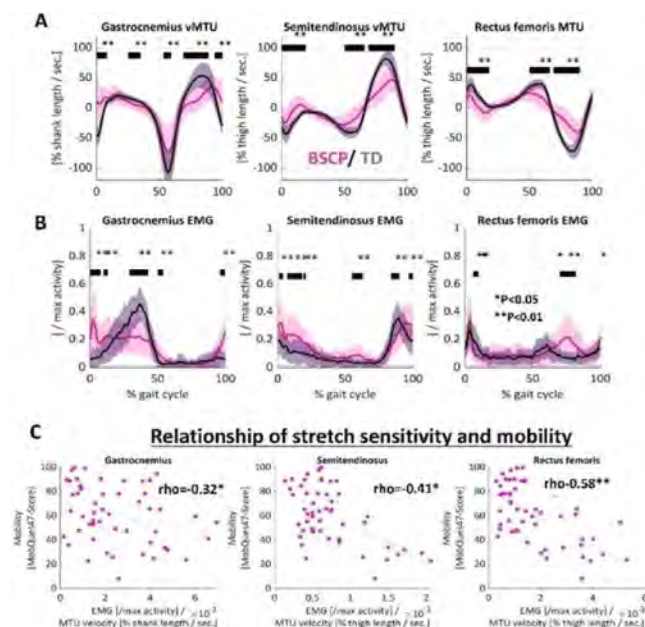


Fig. 1 A) Muscle-Tendon Unit velocities B) Surface EMG of children with BSCP and TD C) Relationship of stretch sensitivity and mobility for children with BSCP.

Also Co-Variate adjusted for slower walking speed, MTU stretch velocities of rectus femoris and semitendinosus remained slower than in TD ( $P < 0.01$ ). A significantly larger EMG/MTU velocity ratio in children with BSCP was noted and this marker negatively correlated with MobQues47-Scores for gastrocnemius ( $\rho = -.32$ ), semitendinosus ( $r = -.41$ ) and rectus-femoris ( $-r = -0.58$ , all  $p < 0.05$ ), which showed stronger correlations to mobility than modified Asworth-Scores for all muscles ( $\rho = 0.21$ ,  $-0.20$ ,  $-0.40$ ).

## Discussion

This study extends previous work [1–4] by demonstrating that the coupling between muscle stretch velocity and muscle activity may serve as relevant marker for spasticity-related mobility impairments. The findings also support earlier evidence that a reduced stretch reflex threshold may limit peak muscle lengthening velocity during gait [5], and suggest that elevated EMG-to-MTU velocity

ratios may reflect increased reflex sensitivity contributing to mobility impairments. These markers may also be considered complementary to standard manual tone assessments by capturing different neural and non-neural components of the impairment. Overall, this approach holds value as a diagnostic tool to inform decision-making.

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## Gait is minimally affected in adults with soft tissue sarcoma after limb-salvage surgery

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## Introduction

Sarcomas are a rare type of cancer and represent a heterogeneous group of tumors that affects 2-4 per 100 000 population. In soft tissue sarcomas (STS), limb-salvage surgery (LSS) is the preferred treatment, aiming to preserve maximal tissue and limb function<sup>1,2</sup>.

LSS has been associated with improved quality of life and favorable functional and psychological outcomes<sup>2</sup>. However it can be hypothesized that the loss of muscle strength will also affect gait<sup>6</sup>. So far, studies reporting gait abnormalities in STS patients are scarce<sup>1,3-5</sup>.

## Research Question

What is the impact of muscle resection in limb salvage surgery for deep soft tissue sarcomas of the thigh on gait?

## Methods

Twenty adults (10/10 male/female, median-age=58.5y (IQR: 52-70) treated with LSS for sarcoma in quadriceps (N=11), hamstrings (N=7) and adductors (N=2) underwent 3D gait analysis (Vicon, Oxford).

Spatiotemporal parameters, the Gait Profile score (GPS) and kinematic and kinetic waveforms from the affected side were extracted. The total patient group, as well as two muscle specific subgroups (i.e. quadriceps and hamstrings resection) were compared

to a healthy control group ( $n=20$ ), matched for age, gender and bodyweight (median-age=60.3y (IQR: 52-73)).

Spatiotemporal parameters and GPS were compared using the Kruskal-Wallis test with post-hoc t-test and Bonferroni correction for multiple testing. Statistical non-parametric mapping (SnPM) defined significant changes in the kinematic and kinetic waveforms: where Hotelling's test indicated significance on vector level, post-hoc t-tests on joint levels were conducted with Bonferroni correction.

## Results

Compared to healthy controls, cadence and walking velocity were decreased in the quadriceps group ( $p<0.05$ ), while GPS and sagittal plane GPS were significantly higher in the total group ( $p<0.01$ ,  $p<0.05$ ) and the quadriceps group ( $p<0.05$ ). No significant alterations were found in the hamstrings group.

SnPM highlighted kinematic alterations in all motion planes in early stance and swing. Especially in the quadriceps group, reduced hip and knee flexion in loading response was preceded by reduced hip flexion in terminal swing (fig.1a).

Kinetics showed increased hip abduction moment in stance in both total and quadriceps group (fig.1b). Kinetics of the knee only significantly differed in the quadriceps group (fig.1c).

Figure 1 represents the kinematic and kinetic waveforms of the quadriceps group compared to the control group.

Fig. 1a: hip and knee flexion - extension

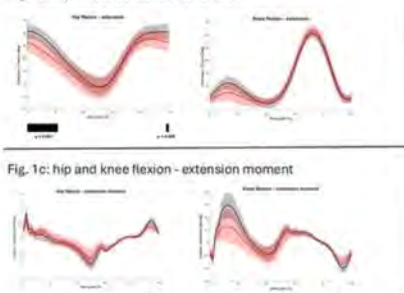


Fig. 1b: hip abduction moment

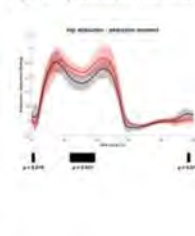


Fig. 1c: hip and knee flexion - extension moment

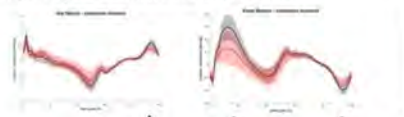


Fig. 1a,b,c represent the post-hoc t-test for kinematic and kinetic waveforms. Red = Quadriceps group (N=11); black = control group (N=20)

## Discussion

Interestingly, only small gait alterations were found in STS patients following LSS, particularly in the quadriceps group, with negligible changes in the hamstring group. The observed within-group variability might be related to the variability in amount of muscle resection.

These findings align with Tanaka et al who reported good post-operative outcomes when at least two quadriceps muscles were preserved<sup>6</sup>, and with Gu et al who noted good function after single quadriceps resection<sup>4</sup>.

The clinical relevance lies in setting realistic expectations for patients pre-operatively. Future studies should also include functional tasks to understand the LSS-impact on overall performance and more high-demanding tasks.

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## Crouch gait in children with cerebral palsy: a cross-sectional study

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## Introduction

Crouch gait, characterised by excessive knee flexion during stance phase, is the most common gait impairment in ambulant children with Cerebral Palsy (CP) and can lead to an increased energy expenditure and chronic knee pain [1]. While hamstring lengthening surgery is a common intervention, evidence suggests that knee extensor weakness and impaired Selective Motor Control (SMC) play a significant role [2,3]. However, the interplay between SMC and knee extensor strength on crouch gait remains unclear. This study examines the impact of SMC and knee extensor strength on crouch gait during an everyday walking circuit

## Research Question

What is the Impact of Selective Motor Control and Knee Extensor Strength on Crouch Gait in Children with Cerebral Palsy?

## Methods

Ambulant children with CP (CwCP) (GMFCS level I-III) and typically developing children (TDC) were recruited from local clinical departments. Participants were assessed using the Selective

Control Assessment of the Lower Extremity (SCALE). Isometric Knee extensor strength was measured at 20° (inner range) and 90° (outer range) via dynamometry (Biodex System 4, UK). Knee extension at Initial Contact (IC) was recorded across an everyday walking circuit (level walking, ramp ascent/descent, and stair descent) using electro-goniometers and accelerometers (DataLITE Biometrics, UK). A paired t-test was used to compare groups and associations between walking and impairment variables were analysed using linear regression

## Results

Twenty-one CwCP (mean age 13.4 years) and ten TDC (mean age 12 years) were included. CwCP demonstrated significantly lower SCALE scores ( $p < 0.001$  CwCP  $5.4 \pm 2.4$  TDC  $10 \pm 0$ ) and reduced inner-range (20°) knee extensor strength ( $p = 0.006$  CwCP  $28.4 \pm 20.2$  TDC  $66.4 \pm 49.2$  Nm/Kg) compared to TDC. A significant difference was found in the knee extensor maximum voluntary contraction (MVC) ratio measured at 20°/90° between CwCP and TDC ( $p = 0.003$  CwCP  $0.42 \pm 0.15$  TDC  $0.60 \pm 0.15$ ). However, outer-range (90°) knee extensor strength did not differ significantly ( $p = 0.233$  CwCP  $65.5 \pm 39.7$  TDC  $122 \pm 94.2$  Nm/Kg). No significant relationship was found between average knee extension at IC and SMC ( $p > 0.05$ ) or outer ( $p > 0.05$ ) or inner ( $p > 0.05$ ) range knee extensor strength across the everyday walking tasks. A negative trend was observed between knee extension at IC and knee extensor MVC ratio ( $R^2 = 0.18$   $p = 0.059$ )

## Discussion

This study highlights that inner range knee extensor strength is more affected in CwCP than outer range strength. The inner/outer range strength ratio was related to the degree of crouch gait when measured across a variety of walking tasks. SMC did not significantly predict knee extension variability, suggesting that other neuromuscular factors may also contribute to crouch gait. Limitations include a small sample size and the cross-sectional design. Future research should explore longitudinal interventions targeting inner range knee extensor strength to optimise gait outcomes in CwCP.

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## A scoping review exploring variables associated with crouch gait in ambulatory children with cerebral palsy

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## Introduction

Crouch gait is a prevalent gait abnormality affecting 72–76% of ambulant children with bilateral Cerebral Palsy (CwBCP). It is characterized by excessive knee, hip, and ankle flexion during stance, leading to knee pain and progressive mobility loss [1]. Factors such as muscle weakness, spasticity, contractures, and skeletal deformities have been implicated in its development, though exact causes remain unclear. The multifactorial nature of crouch gait complicates treatment, resulting in varied management strategies across disciplines [2]. This scoping review aimed to identify and map variables associated with crouch gait to guide future research and clinical decision making.

## Research Question

What variables are associated with crouch gait in ambulant CwBCP?

## Methods

A systematic search was performed across four electronic databases: CINAHL, Medline, Cochrane, and Web of Science. Studies were eligible if they included CwBCP, aged 18 years or younger, classified within Gross Motor Function Classification System (GMFCS) levels I–III and exhibited bilateral knee flexion of  $\geq 15^\circ$  during stance phase, objectively measured using gait kinematics. Two independent reviewers screened titles and abstracts for relevance, with full-text review conducted on shortlisted articles. Potential variables associated with crouch gait were extracted from included studies and classified into one of three groups: Group A (Good association): study findings supported an association with crouch gait. Group B (Uncertain association): Mixed or inconclusive findings. Group C (No association): Findings from included studies found no association.

## Results

The search identified 531 articles, with 33 meeting the inclusion criteria. These comprised 22 interventional studies, 4 computer modelling studies, and 7 cross-sectional studies. Across these studies, 12 variables were identified as significantly associated with crouch gait, Group A -Good association, (1. Ankle/foot position, 2. Gastrocnemius length, 3. Plantar flexor weakness and activation timing, 4. Hamstring length, 5. Hamstring spasticity, 6. Patella alta, 7. Knee extensor activation and fatigability, 8. Psoas length, 9. Gluteal activation and weakness, 10. Walking speed, 11. Selective motor control 12. Visual feedback. One variable was categorised as Group B -Uncertain association (1. Hip extensor weakness). Five variables showed no association with crouch gait, Group C - No association (1. Hamstring weakness, 2. Knee extensor weakness, 3. Iliopsoas weakness, 4. Daily activity levels and 5. Functional health status). Shortened hamstring length and the presence of patella alta accounted for 39% of all evidence with most of these studies (60%) being surgical interventions.

## Discussion

This scoping review identifies key variables associated with crouch gait but did not account for study quality. Findings support the complex multifactorial nature of crouch gait, where causal relationships remain unclear. This suggests treatment should remain

tailored to individual needs. This scoping review provides a platform from which future research in crouch gait can be launched.

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## Spinal fusion limits 3D trunk movement, but improves gait in patients with adult spinal deformity

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## Introduction

Adult Spinal Deformity (ASD) causes significant alterations in posture and gait, leading to functional limitations and reduced mobility [1-3]. Coronal deformities (scoliosis) and sagittal imbalance impact gait patterns differently [4]. Spinal fusion may improve pain, physical function and quality of life, yet little is known about the impact of spinal fusion on gait in ASD patients with either coronal or sagittal imbalance [5-7].

## Research Question

Does spinal fusion in ASD patients with coronal (ASD-C) or predominantly sagittal (ASD-S) spinal imbalance improve spatiotemporal (SPT) parameters and joint kinematics, comparable to levels of healthy controls?

## Methods

Gait of ASD patients scheduled for posterior spinal fusion of four segments or more, and of age-, gender-, leg length- and BMI-matched healthy controls was measured while walking at comfortable speed at Computer Assisted Rehabilitation ENvironment (CAREN; Motek-force Link, Amsterdam, the Netherlands). Measurements were performed before and 12 months (median; range 4-30 months) after spinal fusion. SPT, 3D trunk and pelvic and sagittal hip, knee and ankle angles were determined using custom-made algorithms programmed in Matlab (Mathworks V2016)[8]. Joint angles were normalized for time (gait cycle (GC) 0-100%). SPT parameters and range of motion (ROM) in joint angles before and after spinal fusion were compared using independent t-test or Mann-Whitney U test. Statistical parametric mapping (SPM) was used to compare kinematic waveforms between groups.

## Results

29 ASD-C patients with average (sd) age of 22 (5) years and 29 ASD-S patients aged 53 (16) years were included.

Before spinal fusion, ASD-C patients walked significantly slower than controls with smaller strides. After spinal fusion, ASD-C reached similar SPT as controls. Spinal fusion resulted in a shift of the trunk from the thoracic concave side during most of the gait cycle, towards the body's midline and a decreased 3D trunk ROM (Figure 1). Decreased hip extension persisted postoperatively, but only during toe-off.

ASD-S patients walked significantly slower with decreased cadence, stride length, stance time and wider steps compared to matched controls before spinal fusion. After spinal fusion, step width matched control group levels. Increased anterior trunk tilt in the ASD-S patients resolved almost completely after spinal fusion, but 3D trunk ROM was decreased (Figure 1). The limited knee extension during terminal stance in ASD-S patients resolved after spinal fusion.

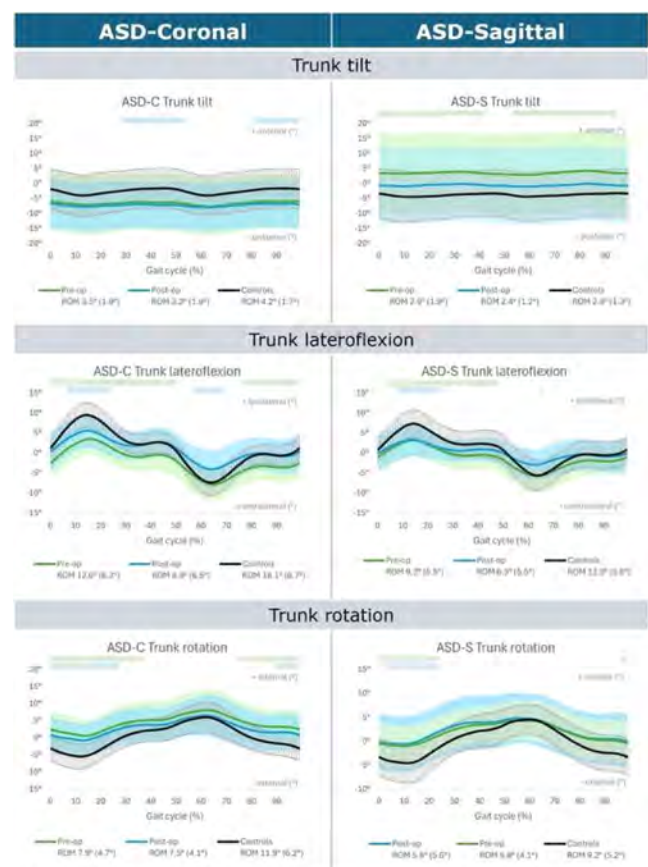


**Figure 1:** 3D joint angles (°) of the trunk reported as group average with standard deviations during the complete gait cycle (0-100%). ROM presented as median (interquartile range). Pre-operative patients in green, post-operative patients in blue and controls in grey. Bars above the graph indicate the part of the gait cycle where the kinematic waveforms significantly differ between pre-op patient group (green) and post-op patient group (blue) compared to controls tested with SPM two-tailed two-sample t-test and  $p < 0.005$ . ASD-C: adult spinal deformity patients with coronal spinal imbalance; ASD-S: adult spinal deformity patients with sagittal spinal imbalance; ROM: Range of Motion; Pre-op: pre-operative; Post-op: post-operative; SPM: statistical parametric mapping.

## Discussion

Spinal deformity and spinal fusion both affect SPT parameters and joint kinematics in ASD patients. Before surgery, ASD-C patients walk with their trunk leaning towards the opposite side of the thoracic convexity throughout the entire gait cycle. After surgery, the trunk crosses the body's midline at initial contact of either leg, similar to controls. The excessive anterior trunk tilt in preoperative ASD-S patients is resolved completely following spinal fusion.

However, in both patient groups, 3D trunk ROM remains significantly limited due to spinal fixation.

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## Can IMU-based systems match clinician Edinburgh Visual Gait Score ratings? A gait assessment study in cerebral palsy patients

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## Introduction

Gait analysis is a crucial clinical method for evaluating individuals' walking patterns and identifying gait abnormalities (Lizama et al., 2016). Standardized systems such as the Edinburgh Visual Gait Score (EVGS) are commonly employed in clinical settings to support these assessments (Rathinam et al., 2014). In recent years, inertial measurement unit (IMU)-based wearable sensors have emerged as a promising alternative due to their portability, low cost, and ease of use (Picerno et al., 2021).

In this study, the reliability of custom-designed IMU sensors in EVGS assessments was investigated. The results obtained from these sensors were compared with expert clinician evaluations, and the validity of the findings was further assessed using a 3D gait analysis system.

## Research Question

Can custom-made IMU sensors provide more reliable and consistent results than clinician-based evaluations in gait assessment using the EVGS?

## Methods

In this study, gait analysis was conducted on eight children with Cerebral Palsy (mean age: 12.8 years). IMU sensors, a 3D motion analysis system (Qualisys, Sweden), and video cameras (Basler, Germany) were used simultaneously to assess the seven parameters of the EVGS. Video recordings were evaluated independently by six clinicians for EVGS scoring. IMU data was transmitted in real-time to the IIoT-based server proposed by Yapakçı in 2022. The data was processed with a Kalman filter to remove noise, and joint angles (hip, knee, ankle) were determined through rotation matrix calculations. All data were cleaned and synchronized. The consistency between clinical scores, the IMU-based system, and 3D gait analysis was evaluated using Scikit-learn.

## Results

The custom-made IMU-based system demonstrated a high level of agreement with the gold standard, providing consistent and accurate results across all EVGS parameters. The highest accuracy was observed in ankle dorsiflexion and hip flexion, both at 93.8%, while the lowest was recorded in knee flexion at 68.8%. Although the clinician evaluations showed high accuracy in certain parameters, their results were generally more variable and, on average, less accurate than those obtained from the IMU-based system.

## Discussion

In conclusion, the specialized IMU-based system generally provided more reliable and objective measurements compared to clinician evaluations, particularly in parameters such as ankle dorsiflexion and hip flexion. However, clinicians demonstrated more consistent results in assessing knee flexion during the swing phase, indicating that certain joint movements may still be better interpreted through expert visual assessment. IMU sensors offer significant advantages in terms of repeatability and objectivity, making them a promising tool for gait assessment in clinical settings. Nevertheless, the findings suggest that further optimization—especially in the calibration and placement of sensors—is necessary to improve the accuracy of knee joint assessments. Future studies should focus on evaluating the performance of IMU-based systems in larger and more diverse patient populations. Such efforts will contribute to the development of clinically validated, sensor-based gait assessment tools that can complement or potentially replace traditional observational methods.

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## Exercise-specific shoulder muscle activation across kettlebell variations: A pilot EMG study

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## Introduction

Effective shoulder training and rehabilitation require selective activation of target muscles to enhance dynamic stability and motor control [1]. Kettlebells provide distinct loading patterns through their shape and grip configuration. However, the effects of different kettlebell exercises on shoulder muscle activation remain under-explored. This study focused on muscles critical for scapular stability and glenohumeral control—specifically the upper trapezius (UT), serratus anterior (SA), and infraspinatus (IS)—due to their clinical relevance and consistent EMG accessibility.

## Research Question

How do four common kettlebell exercises affect activation levels of selected shoulder muscles?

## Methods

Ten healthy, right-hand dominant participants (aged 18–35) with no shoulder pathology performed ten repetitions of four kettlebell exercises: Kettlebell Press, Kettlebell Bottoms-Up Press, Kettlebell Swing, and Kettlebell Single-Arm Swing. Surface EMG was recorded for UT, SA, and IS muscles. Signals were normalized to %MVIC and log-transformed due to non-normal distribution. Five technically correct repetitions—verified via video—were averaged for analysis. Repeated-measures ANOVAs with Greenhouse-Geisser correction and Bonferroni-adjusted post-hoc tests were performed separately for each muscle. Muscle selection was based on clinical relevance and signal reliability; deeper muscles like the lower trapezius were excluded due to difficulty acquiring clean signals during dynamic tasks.

## Results

Significant differences in log-transformed EMG activity were observed across exercises for all three muscles (UT:  $F(1.406, 12.650) = 90.20$ ,  $p < .001$ ; SA:  $F(1.393, 12.534) = 109.42$ ,  $p < .001$ ; IS:  $F(3, 27) = 55.29$ ,  $p < .001$ ). Greenhouse-Geisser correction was applied for UT and SA due to violation of sphericity (Mauchly's test  $p < .05$ ); sphericity was assumed for IS ( $p = .149$ ).

UT activation was significantly higher during the Kettlebell Press and Bottoms-Up Press than during swing-based exercises—Kettlebell Swing and Kettlebell Single Arm Swing ( $p < .001$ ). SA activation was greatest during the Kettlebell Swing compared to all other exercises ( $p < .001$ ). IS activation, however, was significantly lower during the Bottoms-Up Press than during the other three exercises ( $p < .001$ ).

## Discussion

Kettlebell exercises generated distinct and clinically relevant muscle activation profiles. The Kettlebell Swing elicited the highest serratus anterior activity, suggesting its potential utility in promoting scapular stability. In contrast, the Bottoms-Up Press produced elevated upper trapezius activation but under-recruited the infraspinatus, indicating a suboptimal recruitment pattern for posterior cuff engagement. These findings underscore the importance of exercise selection and load orientation when targeting specific shoulder muscles and may guide rehabilitation strategies and support further investigation in clinical populations [2].

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## Effects of single- and dual-task training on dual-task gait costs in Parkinson's disease: A randomized controlled pilot trial

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## Introduction

Gait impairments are among the most prevalent symptoms in patients with Parkinson's disease (PD) and tend to worsen as the disease progresses, reducing independence and quality of life [1].

These impairments become more pronounced during dual-task activities, which require simultaneous motor and cognitive performance [2]. Single-task training (STT) and dual-task training (DTT) are commonly used non-pharmacological interventions aimed at improving gait [3]. However, their effects on dual-task walking performance remain unclear.

## Research Question

Do STT and DTT alter the dual-task cost (DTC) of gait parameters in patients with PD?

## Methods

Fifteen patients with PD were randomly assigned to either the STT group (n=7) or the DTT group (n=8). In addition to routine medical care, participants completed 8 weeks of training (2 sessions per week). Gait parameters—including gait speed, cadence, and stride length—were assessed under single-task, cognitive dual-task, and motor dual-task conditions using the BTS G-Walk (BTS Bioengineering S.p.A., Italy) wireless accelerometer system. DTC was calculated as:  $DTC (\%) = [(dual\text{-}task\ gait\ parameter - single\text{-}task\ gait\ parameter) / single\text{-}task\ gait\ parameter] \times 100$ . Within-group changes were analyzed using the Wilcoxon signed-rank test; between-group comparisons were made using the Mann–Whitney U test.

## Results

Baseline demographic and clinical characteristics were comparable between groups ( $p > 0.05$ ). Prior to training, cognitive DTC of cadence was higher in the STT group ( $p = 0.049$ ), while motor DTC of cadence was higher in the DTT group ( $p = 0.021$ ). After training, the STT group showed significant reductions in the cognitive DTC of gait speed ( $p = 0.018$ ), cognitive DTC of cadence ( $p = 0.018$ ), motor DTC of cadence ( $p = 0.018$ ), and motor DTC of stride length ( $p = 0.028$ ). No significant changes were observed in the DTT group ( $p > 0.05$ ). Between-group comparisons revealed greater improvements in cognitive and motor DTC of cadence in the STT group ( $p = 0.049$  and  $p = 0.021$ , respectively).

## Discussion

These findings suggest that STT may enhance gait performance by reducing DTC in patients with PD. The higher initial cognitive demands of DTT may hinder early adaptation due to cognitive-motor interference. In contrast, STT may allow individuals to automatize motor skills more readily, facilitating attention allocation to concurrent tasks and thereby improving dual-task performance.

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## Associations between executive functions and gait speed during single and dual tasks in Parkinson's disease

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## Introduction

Gait disturbances in Parkinson's disease (PD) significantly contribute to fall risk, a major source of morbidity and mortality in this population [1]. Although gait was once considered an automatic motor function, recent evidence suggests that higher-order cognitive processes, including executive functions, play a crucial role in gait control, particularly under dual-task conditions [2]. Understanding this relationship is vital for improving mobility and fall prevention in PD.

## Research Question

Is there a relationship between gait speed and executive functions under different task conditions in patients with PD?

## Methods

Twenty-four patients with PD (9 females, 15 males; median age: 63 years) and no cognitive impairment (Montreal Cognitive Assessment score  $\geq 21$ ) were assessed. Gait speed was measured using the BTS G-Walk wireless accelerometer system under three conditions: single-task (1.06 m/s), cognitive dual-task (0.87 m/s), and motor dual-task (0.96 m/s). Executive functions were evaluated with the Trail Making Test (TMT-A and TMT-B) and the Stroop test (Çapa version). Spearman's correlation analysis revealed significant negative associations between gait speed and several executive function measures.

## Results

Under the single-task condition, gait speed was negatively correlated with TMT-A ( $\rho = -0.661$ ,  $p < 0.001$ ), TMT-B ( $\rho = -0.431$ ,  $p = 0.035$ ), Stroop-A ( $\rho = -0.476$ ,  $p = 0.019$ ), Stroop-B ( $\rho = -0.570$ ,  $p = 0.004$ ), and Stroop-C ( $\rho = -0.616$ ,  $p = 0.001$ ). In the cognitive dual-task condition, gait speed was negatively correlated with TMT-A ( $\rho = -0.435$ ,  $p = 0.034$ ), Stroop-B ( $\rho = -0.504$ ,  $p = 0.012$ ), Stroop-C ( $\rho = -0.502$ ,  $p = 0.012$ ), and Stroop-D ( $\rho = -0.419$ ,  $p = 0.042$ ). Similarly, under the motor dual-task condition, gait speed was negatively associated with TMT-A ( $\rho = -0.534$ ,  $p = 0.007$ ), Stroop-B ( $\rho = -0.443$ ,  $p = 0.030$ ), Stroop-C ( $\rho = -0.559$ ,  $p = 0.012$ ), and Stroop-D ( $\rho = -0.413$ ,  $p = 0.045$ ).

## Discussion

These findings suggest that impaired executive functions are associated with slower gait across different task conditions in PD. Executive dysfunction may compromise key gait-related cognitive processes such as attention shifting, inhibition, and motor planning, potentially increasing fall risk. Interventions targeting executive function could be beneficial in improving gait performance and promoting safety in patients with PD.

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## Investigation of the effect of mental fatigue on hand reaction time in patients with Parkinson's disease: A pilot study

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## Introduction

Fatigue is one of the most common non-motor symptoms of Parkinson's disease (PD) [1]. It is categorized into two groups: peripheral fatigue, which refers to loss of muscle strength caused by repeated muscle contractions, and central fatigue, which

describes the perception of fatigue, weakness, or exhaustion involving physical and/or mental domains [2-4]. In other words, central fatigue can be either mental or physical [5]. Mental fatigue negatively affects productivity and overall cognitive performance by reducing the ability to inhibit responses, process information, and maintain concentration [6]. In patients with Parkinson's disease (PD), mental fatigue is one of the common and debilitating symptoms, occurring as frequently as motor symptoms [7].

## Research Question

Does mental fatigue affect hand reaction time in patients with PD?

## Methods

A total of 25 patients with PD were included in the study (experimental group: 13 patients, control group: 12 patients). In the experimental group, mental fatigue was induced using a modified Stroop task requiring response inhibition and sustained attention for 30 minutes, while the control group watched a documentary for the same duration. Hand reaction time was assessed using the simple and choice hand reaction time tests, and finger tapping test available on the Multi-Operational Apparatus: MOART (Lafayette Instrument Company, Model no. 35600, Lafayette, IN.) reaction time device. Mental fatigue was assessed subjectively using the Visual Analog Scale (VAS). Within-group changes in hand reaction time and mental fatigue were analyzed using the Wilcoxon signed-rank test, between-group comparisons of changes in these variables were analyzed using the Mann-Whitney U test.

## Results

The groups were similar in terms of age, sex, disease duration, and disease stage. In both groups, no statistically significant change was found in hand reaction time tests and mental fatigue when comparing pre-procedure and post-procedure measurements ( $p > 0.05$ ). While no statistically significant difference was observed between the groups in terms of the change in hand reaction time tests before and after the procedure ( $p > 0.05$ ), the change in VAS score was determined to be greater in the experimental group ( $p < 0.001$ ).

## Discussion

Our study is the first research to investigate the effect of mental fatigue on hand reaction time in patients with PD. Although the mental fatigue procedure successfully increased perceived mental fatigue levels of patients, no significant effect was observed on hand reaction time performance. Considering potential learning effects from repeated testing, participants may not have reached the intended objective level of mental fatigue. Future studies with a larger sample size and a different mental fatigue procedure are needed.

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## The influence of childhood sitting habits on gait patterns in individuals with joint hypermobility: A pilot study

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## Introduction

During early childhood, children spend significant time sitting on the floor. These sitting postures shape anatomical structures and neuromuscular adaptations (1). Due to musculoskeletal traits and environmental factors, children adopt specific sitting postures that impose different mechanical loads on the lower extremities, which may affect gait kinematics and locomotor development. Due to a lack of literature, this study aimed to investigate the effects of various sitting postures on lower limb biomechanics and gait kinematics.

## Research Question

Do childhood sitting habits affect gait kinetics in hypermobile children compared to neutral sitters?

## Methods

Generalized joint hypermobility was assessed in all participants using the Beighton score, with a threshold of >5 (2). Femoral anteversion was measured using the Trochanteric Prominence Angle Test (TPAT)  $\leq 30^\circ$  (3). Eleven participants were included in the

study and classified into three groups based on their childhood habitual sitting postures: the W-sitting group (W group,  $n = 4$ ,  $12.5 \pm 3.53$  y.o., Beighton score:  $6.75 \pm 0.95$ , TPAT:  $15.5^\circ \pm 1^\circ$ ), the cross-legged sitting group (C group,  $n = 3$ ,  $11.33 \pm 3.21$  y.o., Beighton score:  $6.67 \pm 1.15$ , TPAT:  $17.83^\circ \pm 4.02^\circ$ ), and the neutral sitting group (N group,  $n = 3$ ,  $11.33 \pm 2.08$  y.o., Beighton score:  $6.33 \pm 1.52$ , TPAT:  $18.5^\circ \pm 4.46^\circ$ ). Kinematic gait parameters were collected using a three-dimensional motion analysis system (VICON, Plug-in Gait Model, UK) and statistically compared using Student's t-tests ( $p < 0.05$ ) (3).

## Results

Gait analysis revealed significant kinematic differences based on childhood sitting postures. Compared to W-sitting and neutral sitting, cross-sitting was associated with decreased anterior pelvic tilt, hip flexion, and ankle dorsiflexion. W-sitting resulted in increased hip adduction and decreased knee flexion, knee extension, and ankle dorsiflexion relative to neutral sitting. Neutral sitters demonstrated the highest values for pelvic tilt, hip flexion, knee flexion/extension, and ankle dorsiflexion. Walking speed did not differ significantly among groups (Table 1).

Table-1: The statistical comparison of walking parameters among groups.

| Kinematic Variables (°)     | W group            | C group           | N group          |
|-----------------------------|--------------------|-------------------|------------------|
|                             | Mean $\pm$ SD      | Mean $\pm$ SD     | Mean $\pm$ SD    |
| Mean Anterior Pelvic Tilt   | 8.27 $\pm$ 4.16    | 4.80 $\pm$ 4.19*  | 10.50 $\pm$ 0.82 |
| Peak Anterior Pelvic Tilt   | 12.95 $\pm$ 0.63   | 7.01 $\pm$ 4.16*  | 10.51 $\pm$ 4.23 |
| Peak Posterior Pelvic Tilt  | 5.44 $\pm$ 4.34    | 1.46 $\pm$ 5.12*  | 7.89 $\pm$ 0.83  |
| Mean Hip Flexion @St        | 5.45 $\pm$ 6.61†   | 2.13 $\pm$ 6.43*  | 11.92 $\pm$ 0.81 |
| Peak Hip Flexion @St        | 32.42 $\pm$ 7.95   | 28.26 $\pm$ 4.19* | 38.87 $\pm$ 2.60 |
| Peak Hip Extension @St      | -14.08 $\pm$ 4.91† | -16.32 $\pm$ 7.92 | -9.11 $\pm$ 3.58 |
| Mean Hip Adduction @St      | 3.87 $\pm$ 1.16†   | 0.21 $\pm$ 3.24 † | 1.58 $\pm$ 2.10  |
| Peak Hip Adduction @St      | 9.92 $\pm$ 1.69†   | 5.90 $\pm$ 3.41 † | 6.67 $\pm$ 2.74  |
| Mean Knee Flexion @St       | 10.34 $\pm$ 4.23†  | 12.78 $\pm$ 4.15* | 16.82 $\pm$ 1.83 |
| Peak Knee Flexion @St       | 0.83 $\pm$ 3.33†   | 1.60 $\pm$ 4.27   | 4.35 $\pm$ 2.23  |
| ROM Knee Rot @St            | 15.83 $\pm$ 2.49†  | 16.21 $\pm$ 4.76* | 21.60 $\pm$ 2.86 |
| Mean Ankle Dorsiflexion @St | 6.06 $\pm$ 2.44†   | 7.45 $\pm$ 3.27   | 10.10 $\pm$ 3.38 |
| Peak Ankle Dorsiflexion @St | 15.35 $\pm$ 3.67†  | 14.45 $\pm$ 3.79* | 19.11 $\pm$ 2.41 |
| Walking Speed (m/s)         | 1.29 $\pm$ 0.09    | 1.46 $\pm$ 0.12   | 1.41 $\pm$ 0.13  |

Significantly different parameters between: \*; N group and C group; †; N group and C group; ‡; W group and C group ( $p < 0.05$ ). Rot: rotation. St.: stance phase

## Discussion

These findings indicate that habitual sitting postures affect the entire lower limb biomechanics during walking in hypermobile individuals. Specifically, W-sitting imposes biomechanical constraints, demonstrated by decreased hip flexion and knee extension, along with increased hip adduction. These alterations suggest potential changes in knee joint kinematics and a heightened risk of musculoskeletal imbalances. Therefore, further research is warranted to fully elucidate the long-term impact of these sitting habits on locomotor efficiency. Previous studies have shown that pelvic orientation plays a crucial role in gait mechanics, influencing lumbar posture and lower limb movement patterns. In contrast, the neutral sitting group exhibited the highest values for peak hip flexion, knee flexion, and ankle dorsiflexion, suggesting that a balanced sitting posture may promote more natural lower extremity kinematics.

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## The effect of asymmetric femoral anteversion on gait kinematics: A pilot study

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## Introduction

Increased femoral anteversion (IFA), the inward rotation of the femoral head relative to the distal condyles, is described as exceeding the typical range (10-20°) of the Trochanteric Prominence Angle Test (TPAT) > 30° (1,2). However, asymmetric femoral anteversion (AFA) is defined as a significant difference between limbs' femoral inward rotation, which may asymmetrically alter gait kinematics. Understanding these alterations could provide valuable insights for developing targeted therapeutic approaches to improve gait mechanics and prevent associated complications.

## Research Question

How does asymmetrically increased femoral anteversion affect gait kinematics?

## Methods

This study included 27 participants; 11 typical developed (N) (11.62 ± 2.50 y.o., TPAT: 18.37° ± 3.29°), 9 with Bilateral IFA (BIFA) (10.33 ± 2.0 y.o., TPAT: 43.54° ± 9.56°) and 7 with AFA (10.06 ± 1.67 y.o.). The AFA group was investigated by the IFA side (UNIFA) (TPAT: 33.62° ± 4.20°), and the typically developed contralateral side (NoIFA) (TPAT: 20.87° ± 3.94°). In AFA, the NoIFA side was defined as TPAT < 30°, and the UNIFA side was TPAT > 30° with a 10° difference between the two sides. Interested kinematic gait parameters were collected by a computerized 3D

motion analysis system (VICON, Plug-in Gait Model, UK)(3). Student's t-tests were used for statistical comparison (p < 0.05).

## Results

Pelvik rotasyon ROM, NoIFA tarafında N grubuna göre daha düşüktü. Tepe pelvik IR ve ROM, UNIFA tarafında BIFA grubuna göre daha düşüktü ve tepe plantar fleksiyon daha yüksekti. Tepe diz varusu, AFA grubunda (UNIFA ve NoIFA) her iki tarafta da sırasıyla N ve BIFA gruplarına göre anlamlı şekilde artmıştı. Tersine, tepe diz valgusu, AFA grubunda BIFA ve kontrollere göre, özellikle NoIFA tarafında azalmıştı (Tablo-1).

Table-1: Comparison of the interested gait parameters

|                            | N Group       | NoIFA Side (AFA Group) | UNIFA Side (AFA Group) | BIFA Group    |
|----------------------------|---------------|------------------------|------------------------|---------------|
|                            | Mean ± sd     | Mean ± sd              | Mean ± sd              | Mean ± sd     |
| Pelvic Rot. ROM            | 18.04 ± 3.91  | 13.28 ± 3.90*          | 13.35 ± 3.36*          | 17.58 ± 3.36  |
| Peak Pelvic IR             | 8.69 ± 2.48   | 7.08 ± 4.46            | 6.27 ± 4.35*           | 10.47 ± 2.89  |
| Mean Hip Flx @St           | 6.92 ± 6.14   | 8.87 ± 3.81            | 10.06 ± 4.71           | 10.41 ± 4.69  |
| Mean Hip Abduction @St     | 3.23 ± 1.34   | 1.98 ± 2.51            | 0.48 ± 3.11            | 2.94 ± 3.35   |
| Mean Hip Rot. @St          | -10.32 ± 8.46 | -4.38 ± 15.02          | -5.89 ± 17.78          | -0.75 ± 17.54 |
| Mean Knee Flx @St          | 12.67 ± 4.02  | 12.02 ± 4.23           | 14.31 ± 4.56           | 14.69 ± 2.89  |
| Peak Knee Varus @St        | 0.98 ± 2.80   | 7.37 ± 6.46*           | 8.06 ± 7.93            | 3.30 ± 8.38   |
| Peak Knee Valgus @St       | -7.44 ± 4.92  | -3.90 ± 5.82           | -4.94 ± 7.16           | -7.05 ± 7.10  |
| Mean Knee Rot. @St         | 4.89 ± 3.34   | 3.69 ± 9.19            | 4.96 ± 12.37           | 4.50 ± 6.45   |
| Peak Knee ER @St           | -2.54 ± 7.98  | -6.03 ± 8.59           | -4.83 ± 10.47          | -6.74 ± 9.72  |
| Peak Ankle Plantar Flx @St | -10.11 ± 5.00 | -7.78 ± 8.72           | -9.53 ± 5.58*          | -3.46 ± 5.23  |
| TPAT                       | 18.37 ± 3.29  | 20.87 ± 3.94           | 33.62 ± 4.20           | 43.54 ± 9.56  |
| Walking Speed (m/s)        | 1.41          | 1.31                   | 1.31                   | 1.26          |

\* Significantly different parameters between the N group and NoIFA side, † UNIFA side and BIFA group (p<0.05). Rot: rotation; ROM: range of motion; IR: internal rotation; ER: external rotation; St: stance phase; Flx: flexion

## Discussion

This study demonstrated that AFA significantly influences transverse plane pelvic and coronal-transverse plane knee kinematics during walking. Reduced pelvic rotation in individuals with asymmetric IFA (the UNIFA and NoIFA sides) may reflect compensatory mechanisms to maintain stability, which is agreed with the literature (4). Increased peak knee varus on both sides of the AFA group may indicate abnormal loading patterns, contributing to joint stress and potential risk of knee injuries. Additionally, reduced peak knee valgus and increased ER on the UNIFA side may show altered knee joint mechanics, which may lead to ligamentous or cartilage problems over time (5). These results underscore the importance of identifying asymmetrically increased femoral anteversion in clinical assessments to tailor appropriate orthopedic interventions and physiotherapy programs, especially for preventive knee injuries. Further longitudinal studies are necessary to investigate the long-term effects of AFA on locomotion and joint health.

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## From healthy to myotonia: tibialis anterior muscle mechanics revealed by shear wave elastography

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### Introduction

Myotonia, characterized by delayed skeletal muscle relaxation following voluntary contraction, is a hallmark of certain hereditary muscle disorders [1]. In severe cases, muscles exhibit weakness and atrophy with reduced tone [2]. There is an increasing need for quantitative, non-invasive methods to evaluate myotonia severity for diagnostic purposes, disease monitoring, and assessing therapeutic interventions. Ultrasound shear wave elastography (SWE) is a promising tool for assessing muscle mechanical properties in both passive and active states [3].

### Research Question

This study employs SWE to assess tibialis anterior (TA) muscle mechanics in myotonia patients, examining responses to passive ankle positioning and active isometric contractions, with comparisons to healthy controls.

### Methods

Fourteen healthy volunteers ( $26.4 \pm 3.7$  years old) and eight patients ( $46.1 \pm 10.3$  years old) participated. Simultaneous electromyography, SWE of the TA, and ankle joint moment measurements were performed at  $-15^\circ$  dorsiflexion,  $0^\circ$  neutral, and  $15^\circ$ ,  $30^\circ$ , and  $45^\circ$  plantar flexion (PF) positions during rest, maximum voluntary contractions (MVC), and submaximal isometric ramp contractions (25%, 50%, and 75% MVC).

### Results

In healthy participants, passive ankle joint moment increased with PF ( $p < 0.001$ ), from  $-9.52 \pm 2.60$  Nm at  $-15^\circ$  to  $1.84 \pm 4.67$  Nm at  $45^\circ$ . Patients consistently showed passive moments in the PF direction across all angles (Figure A). During MVC, healthy participants produced peak moments of  $50.13 \pm 15.54$  Nm at  $15^\circ$ . Among the three representative patients analyzed, only one (Patient 3) exhibited a comparable profile; the others generated substantially lower values.

The TA passive shear modulus in healthy participants increased with plantar flexion ( $p < 0.001$ ), from  $9.67 \pm 5.03$  kPa at  $0^\circ$  to  $33.05 \pm 16.29$  kPa at  $45^\circ$ . This angle-dependent increase was absent in patients (Figure B), with a maximum observed value of 4.41 kPa (Patient 2). Total shear modulus was reduced in patients (Figure C),

averaging 10.32 kPa compared to  $122.96 \pm 9.87$  kPa in the healthy group. No angle effect was observed in the healthy group ( $p = 0.068$ ).

SWE distinguished contraction levels in healthy subjects ( $p < 0.001$ ), with total shear modulus at  $0^\circ$  averaging  $69.86 \pm 23.21$  kPa,  $128.14 \pm 23.53$  kPa, and  $130.26 \pm 25.62$  kPa at 25%, 50%, and 75% MVC, respectively. While similar trends were observed in patients, values remained markedly lower.

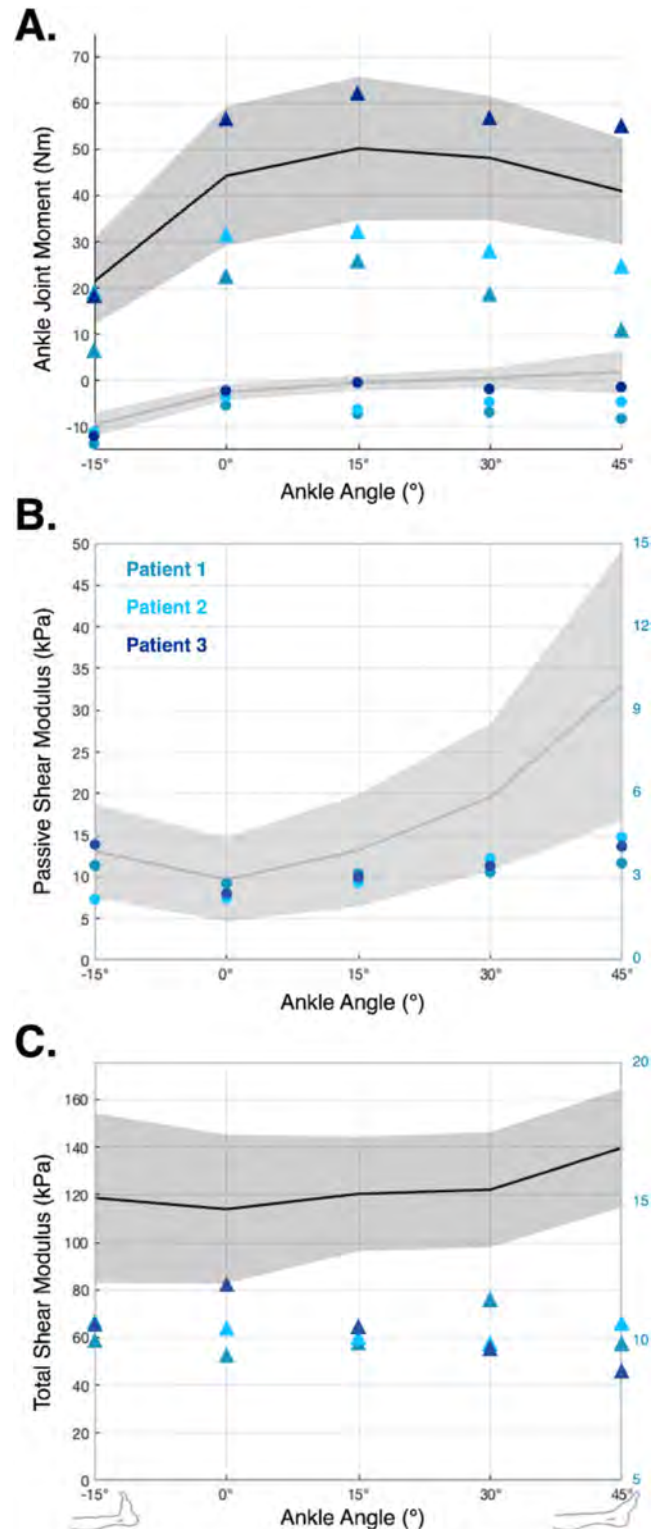


Figure. (A) Ankle moment, (B) passive, and (C) total shear modulus of the TA. Solid lines show healthy group data: black for MVC, grey for rest, and shaded areas indicate standard deviation.

## Discussion

SWE captures passive and activation-dependent mechanical properties of the TA, distinguishing healthy from myotonic muscle. The lower shear modulus in patients reflects muscle weakness and diminished contractility. SWE's sensitivity to contraction intensity supports its potential as a biomarker for functional impairment in myotonia. Although only three patients are presented here, full cohort results, currently underway, will be presented at the conference, offering statistical comparisons and further reinforcing SWE's potential for individualized disease monitoring.

## Acknowledgments

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## Effect of cervicothoracic inhibition test on cervical painful ranges in individuals with mechanical cervical pain

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## Introduction

Mechanical, non-specific neck pain involves heterogeneous anatomical and neurophysiological mechanisms (1). Manual therapists commonly use inhibition tests and cervicothoracic mobilizations to

assess individual responses and influence pain-related mechanisms, though supporting evidence remains limited (2,3,4). This study aims to examine the effect of cervicothoracic posterior to anterior facet joint segmental traction inhibition test (CTPA-FIT) on painful cervical range of motion (ROM) during cardinal and circumduction movements in individuals with mechanical neck pain.

## Research Question

Does the application of CTPA-FIT acutely improve the cervical kinematics performed within painful ranges?

## Methods

Nine individuals (age:  $26.77 \pm 7.85$  y.o.) who have mechanical cervical pain > 3 months and four healthy controls (age:  $24.75 \pm 6.39$  y.o.) were included in the study. Cervical pain was defined by Numerical Pain Rating Scale (NPRS), Body Pain Chart and Neck Disability Index questionnaire (5). Only the individuals with middle and lower cervical pain were included into the study (6). CTPA-FIT was applied to the transverse processes of the T6 to C7 vertebrae using a custom-designed 3D-printed apparatus attached to the tip of an algometer (J-Tech Commander, USA), with an applied pressure of 20 to 40 N for 3 seconds per segment (7). Individuals remained within the ROM of the most painful cervical movement at the beginning of the test at a value of NPRS 2-3, and ROM variations during the test were recorded. Neck movement kinematics including clockwise and counterclockwise circumduction, and primary plane cervical ROMs at sagittal, frontal, and transverse planes were recorded before and after the test by utilizing an instrumented motion capture system (VICON 6xVantage 5, 1xVero MP@420FPS) during sitting position (5,8). Independent samples t-test was calculated for comparison ( $p < 0.05$ ).

## Results

Compared to controls, cervical pain group had significantly lower painful neck movement ranges before the application CTPA-FIT in all three planes and movement types. Although post-CTPA-FIT neck movement ranges were lower compared to the control group, the cervical pain group demonstrated significantly higher post-inhibition values than pre-inhibition, except for lateral flexion and rotation in the cardinal plane movements. Control group had no significant alterations after CTPA-FIT. All interested parameters were compared in Table-1.

Table 1: Statistical comparison of interested parameters.

| Movement Type                   | Neck Movement Range for Groups | Sagittal/Flex-Ext (°) | Frontal/Lat.Flex (°) | Transverse/Rot (°) |
|---------------------------------|--------------------------------|-----------------------|----------------------|--------------------|
| Circumduction Counter-Clockwise | Control Range                  | 141.27 ± 6.29*        | 84.12 ± 8.66*        | 115.49 ± 23.28*    |
|                                 | Painful Range Pre              | 94.86 ± 18.83†        | 69.51 ± 15.44†       | 65.90 ± 24.72†     |
|                                 | Painful Range Post             | 106.19 ± 15.06‡       | 74.72 ± 13.63‡       | 74.37 ± 27.77‡     |
|                                 | Δ Pre and Post                 | 11.32 ± 10.81         | 5.20 ± 8.44          | 8.46 ± 6.88        |
| Circumduction Clockwise         | Control Range                  | 136.14 ± 10.99*       | 82.27 ± 8.13*        | 111.51 ± 13.59*    |
|                                 | Painful Range Pre              | 96.27 ± 19.88†        | 66.39 ± 16.45†       | 65.39 ± 25.44†     |
|                                 | Painful Range Post             | 106.46 ± 12.24‡       | 72.54 ± 15.63‡       | 74.71 ± 31.62‡     |
|                                 | Δ Pre and Post                 | 10.18 ± 12.94         | 6.15 ± 7.60          | 8.58 ± 15.72       |
| Cardinal Planes                 | Control Range                  | 144.22 ± 16.38*       | 100.95 ± 4.78*       | 154.57 ± 4.42*     |
|                                 | Painful Range Pre              | 95.57 ± 15.57†        | 80.23 ± 9.90†        | 125.89 ± 18.11†    |
|                                 | Painful Range Post             | 109.35 ± 12.43‡       | 81.73 ± 12.74        | 128.45 ± 21.84     |
|                                 | Δ Pre and Post                 | 13.78 ± 16.43         | 1.50 ± 6.29          | 3.56 ± 13.02       |

Significant difference between groups illustrated as \* for control and cervical pain pre-inhibition, † for cervical pain pre- and post-inhibition, and ‡ for control and cervical pain post-inhibition

## Discussion

CTPA-FIT was safely used during 3D motion capture, increased the range in all movements and may have therapeutic effects on individuals with cervical pain. Circumduction movements may have more effectively demonstrated the change in the painful range after CTPA-FIT, likely due to the multifunctional kinematics of cervical segments, compared to cardinal movements in the frontal and transverse planes (8,9). This study has shown that the cervical movement kinematics of individuals with mechanical neck pain and their localized need for manual therapy interventions are worth investigating in detail with a large number of participants.

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## In patients with cervical pain, which segments does the inhibition technique applied to increase cervical range?

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## Introduction

Neck pain has multiple underlying mechanisms and is most commonly aggravated by cervical movements. Previous studies have demonstrated the regional interdependence of cervicothoracic segments as a contributing factor to this mechanism (1). Manual therapists often use inhibition tests targeting symptom-related areas

to assess individual responses, and apply facet joint segmental traction mobilizations to modulate pain-related mechanisms during cervical pain; however, supporting evidence remains limited (1,2,3). This study aims to investigate the immediate effect of the cervicothoracic posterior-to-anterior facet joint segmental traction inhibition test (CTPA-FIT) on painful cervical joint range of motion (ROM) in individuals with mechanical neck pain.

## Research Question

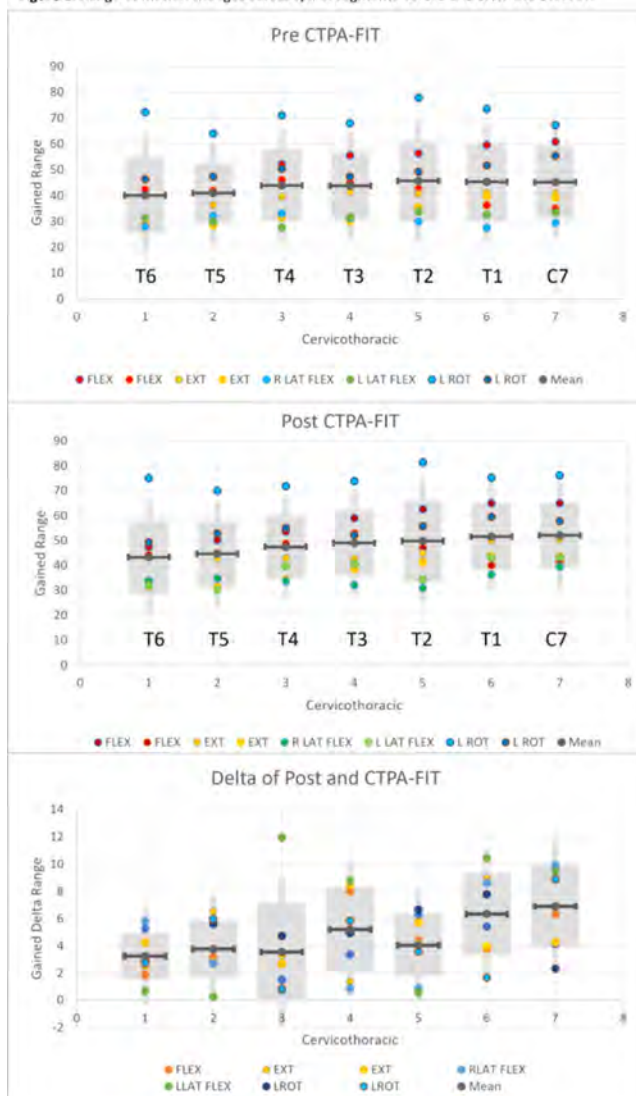
In mechanical neck pain, which cervicothoracic segment, when tested with inhibition techniques, results in greater pain-free cervical joint range ?

## Methods

Eight individuals (age:  $27.5 \pm 8.07$  y.o.) who have mechanical cervical pain > 3 months were included in the study. Cervical pain was defined by Numerical Pain Rating Scale (NPRS), Body Pain Chart and Neck Disability Index questionnaire (5). Only the individuals with middle and lower cervical pain were included into the study (4). CTPA-FIT was applied to the transverse processes of the T6 to C7 vertebrae using a custom-designed 3D-printed apparatus attached to the tip of an algometer (J-Tech Commander, USA), with an applied pressure of 20 to 40 N for 3 seconds per segment (5). Individuals remained within the ROM of the most painful cervical movement at the beginning of the test at a value of NPRS 2-3, and ROM variations during the test were recorded. Neck movement kinematics were recorded during the test by utilizing an instrumented motion capture system (VICON 6xVantage 5, 1xVero MP@420FPS) during sitting position (6). Independent samples t-test was calculated for comparison ( $p < 0.05$ ).

## Results

In eight individuals with neck pain reporting the highest pain levels based on NPRS across three planes, the application of CTPA-FIT resulted in a significant increase in ROM at various segments, specific to each individual. The greatest number of ROM improvements was observed at the C7-T1 segment. However, other segments also demonstrated improvements of 5 degrees or more. Figure 1 illustrates the ROM changes according to spinal segments.

**Figure 1.** Range of motion changes across spinal segments before and after the CTPA-FIT

## Discussion

CTPA-FIT alone was observed to increase cervical ROM in each patient at different segments from T6 to C7 (3), with the greatest ROM improvement most frequently occurring at the C7 and T1 segment. These findings, consistent with the literature, suggest that outcomes may vary across individuals (1). Although the number of participants in this study was limited, the results emphasize the necessity of individualized assessment and treatment planning in manual therapy, even when patients share a common underlying cause such as mechanical neck pain. For future research, we recommend designing studies in which individuals with neck pain are grouped based on differing pathophysiological factors.

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## Effects of using different optimization metrics in selecting the optimal ankle-foot orthosis stiffness in patients with calf muscle weakness

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## Introduction

To maximize the effects of ankle-foot orthoses (AFOs) on walking in individuals with calf muscle weakness due to neuromuscular disorders, their stiffness must be individually optimized<sup>1</sup>. Currently, AFO stiffness optimization is mainly based on minimizing the energy cost (EC) of walking. While humans typically walk to minimize EC, and EC is a clinically valuable indicator of physical effort, patients and clinicians may also aim to improve perceived walking effort, walking speed, or gait kinematics. However, optimizing these metrics influence the optimization outcomes<sup>2</sup>. Understanding how different metrics affect the optimal AFO stiffness may facilitate the provision of optimized AFOs for individual treatment objectives.

## Research Question

What are the effects of using different optimization metrics, including walking EC, perceived walking effort, walking speed, and maximal ankle moment during stance, for selecting the optimal AFO stiffness in calf muscle weakness.

## Methods

Forty-two individuals with non-spastic calf muscle weakness were included (age:  $58.26 \pm 15.02$  years). Individuals received a dorsal leaf AFO (Carbon Ankle7) with manually adjustable stiffness. Participants completed a 3D gait analysis and a 6-minute walk test with breath-by-breath gas analysis at a self-selected speed with five different AFO stiffness (K) settings (range: 2.8-6.6 Nm/degree). From the 6-minute walk tests, for each stiffness walking EC and speed were calculated. Perceived walking effort was measured on a 10-point Borg-scale. From the gait analysis, maximal ankle moment during stance was determined. The optimal stiffness was defined as the lowest EC, highest speed, lowest Borg-score or highest ankle moment. If no minimum occurred, the lowest stiffness was selected.

## Results

Mean optimal stiffness levels were: EC,  $K = 4.3 \pm 1.3$ ; walking speed,  $K = 4.2 \pm 1.2$ ; Borg-score,  $K = 3.6 \pm 1.1$ ; maximal ankle moment  $K = 4.4 \pm 1.3$ . The optimal stiffness using perceived

walking effort as metric was significantly lower than for EC ( $p = 0.009$ ), speed ( $p = 0.008$ ), and maximal ankle moment ( $p < 0.001$ ). Figure 1A illustrates the distribution of stiffness values across metrics. Figure 1B exhibits the percentage of similar optimal AFO stiffness value between metrics.

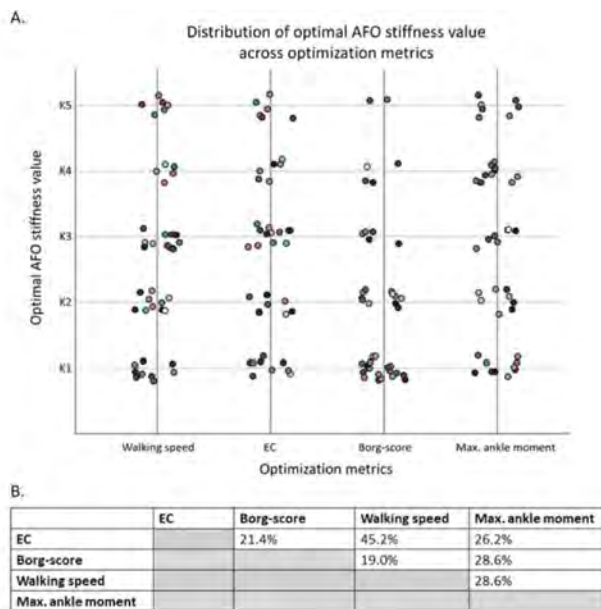


Figure 1: Optimal AFO stiffness values for optimization metrics. Figure A shows the distribution of optimal AFO stiffness values across all metrics. All participants ( $n = 42$ ) show the optimal stiffness for each optimization metric, walking speed, EC, Borg-score, maximal ankle moment. Each dot is one measurement of one patient. Figure B exhibits the percentage of similar AFO stiffness values between the optimization metrics. AFO = ankle foot orthosis EC = energy cost.

## Discussion

Our results show that the optimal AFO stiffness in individuals with calf muscle weakness differs depending on the optimization metrics used, with the optimal stiffness for minimizing perceived walking effort being significantly lower than that of the objective optimization metrics. For many individuals, multiple stiffness values resulted in similar perceived effort scores, leading to selecting the lowest stiffness. It may suggest that this metric lacks sensitivity to determine the optimal AFO stiffness. To maximize the effectiveness of AFOs for calf muscle weakness, it is important to select an optimization metric that aligns with the patient's goal and is sensitive enough to determine individual optimal stiffness.

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## Does the brain utilize the force sensation to determine joint position?

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## Introduction

Proprioception is a significant factor in balance, coordination, joint stability and movement acuity (1). The central nervous system uses signals from sensory receptors located on muscles, tendons, and ligaments to determine the position of joints (2). However, there is no evidence in the literature indicating that the force sensation is used for this purpose.

## Research Question

Are there joints where the brain utilizes the force sensation to determine their position?

## Methods

Thirty-three volunteers (16 men, 17 women, mean age  $21.4 \pm 1.8$  years) participated in the study. Retroreflective markers were placed at the following points: a) the inferior angle of the scapula, b) L5 vertebra, c) ASIS, d) greater trochanter, and e) lateral femoral condyle. Subjects were positioned in: (a) a relaxed upright posture (Pos1), and (b) a plank (Pos2), with the instruction of mimicking the previous upright posture. All participants easily performed planks without being able to observe their body position. Two cameras recorded positions from different heights and distances. Kinovea software was used to calculate the angles from the markers, while XSENS, two force plates and AnyBody (AnyBody Technologies) software were used to create a biomechanical model of the plank position to explain the results.

## Results

All participants incorrectly mimicked the upright plank position with increased lumbar flexion (mean difference  $23.7^\circ$ ,  $p=0.000$ ) by positioning their pelvis higher. The biomechanical model of the plank showed that lumbar spine compression force rises with a higher pelvis position, but posterior-anterior shear force increases with a lower pelvis position, reaching values seven times higher (measured at L4-L5 level) than that of the upright standing position.

## Discussion

Participants likely failed to match the lumbar position between the plank and upright posture due to perceiving "same loading" as "same position". Participants preferred higher pelvis positions during the plank exercise (more flexed lumbar spine compared to the upright posture), where compression force was similar to that of an upright stance, and avoided lower pelvis positions (the lumbar spine is positioned similarly to that of an upright stance), where compression was minimal but posterior-anterior force was extreme. It

appears that the central nervous system prioritizes force sensation over the other proprioceptive signals in the lumbar spine, to avoid positions with unnatural high postero-anterior forces.

Furthermore, the extreme shear forces during the plank position have clinical implications, particularly for spondylolysis and spondylolisthesis, and need further investigation.

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## Effects of myofunctional rehabilitation on pain, function and spinal characteristics in temporomandibular joint dysfunction: preliminary study

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## Introduction

Temporomandibular joint dysfunction (TMD) is a complex musculoskeletal disorder affecting the masticatory system (1-3). It is closely related to spinal pain, especially cervical spine (4-6). Myofunctional therapies, including manual therapy and exercises, are commonly used to relieve pain and improve function in TMD (7). However, the treatment effects on spinal structure and mobility in addition to pain and function are not very well revealed. This study aimed to investigate the effect of an 8-week myofunctional rehabilitation program on pain and the functions of temporomandibular joint and spine with structural and mobility characteristics.

## Research Question

Does an 8-week myofunctional rehabilitation program affect temporomandibular joint pain, function, spinal structure and mobility in patients with TMD?

## Methods

According to the Research Diagnostic Criteria for TMD, 21 participants (15 females, 6 males, age:  $29.24 \pm 4.12$  years, BMI:  $24.12 \pm 3.69$  kg/m<sup>2</sup>) with muscle-joint involvement were included. Mouth opening was assessment with digital caliper, pain levels with Visual Analog Scale, and functional limitation with “Jaw Functional Limitation Scale-20” (8,9). Head position, the angle between the horizontal plane and C7 and the tragus was measured with a goniometer (BASELINE®). Spinal structure and mobility in the

sagittal and frontal planes were evaluated using Valedo®Shape device (Idiag, Switzerland), providing thoracic, lumbar, sacral/hip angles, and trunk inclination (angle between straight line from T1-S1 and vertical line) (10). Thoracic rotation was assessed with the Compass app (iPhone, Apple Inc.) placed at the T1–T2, starting from 0°, and the participant performed active thoracic rotation while maintaining contact with the phone (11). As part of myofunctional rehabilitation, manual therapy methods (including soft tissue and joint techniques) and stretching, strengthening, and stabilization exercises targeting the masticatory and neck muscles were applied twice a week for 8 weeks in 45-minute sessions (12). Pre- and post-treatment assessments were conducted. Normality was tested using Shapiro-Wilk, and paired t-tests were used for comparisons.

## Results

Compared to the pre-treatment, participants showed reduced pain ( $p = 0.003$ ) and functional limitation ( $p = 0.001$ ) and increased maximum mouth opening ( $p < 0.001$ ). Significant differences were observed in the cervical angle ( $p = 0.009$ ), and thoracic rotation right ( $p = 0.045$ ) and left ( $p = 0.023$ ). In the sagittal plane, lumbar and sacral angles ( $p = 0.024$ ;  $p = 0.008$ ), and sacral mobility ( $p = 0.030$ ) also improved. In the frontal plane, a significant difference was noted in the inclination angle ( $p = 0.011$ ). However, regarding angle and mobility, no significant changes were observed in the thoracic ( $p = 0.108$ ;  $p = 0.753$ ), lumbar ( $p = 0.865$ ;  $p = 0.196$ ), and sacral ( $p = 0.766$ ;  $p = 0.713$ ) regions. Details are presented in Table 1.

Table 1. Comparison of Pre- and Post-Intervention Outcomes.

| Parameters                    | Pre-treatment<br>X <sup>±</sup> SD | Post-treatment<br>X <sup>±</sup> SD | 95% CI      | t      | p       |
|-------------------------------|------------------------------------|-------------------------------------|-------------|--------|---------|
| Pain (Rest) mm                | 4.26±1.70                          | 1.58±1.97                           | 1.02-6.50   | 3.414  | 0.003*  |
| Pain (Activity) mm            | 3.89±1.12                          | 2.11±1.29                           | 1.34-5.21   | 3.331  | <0.001* |
| Pain (Night) mm               | 3.21±1.84                          | 2.27±3.07                           | 1.11-4.78   | 3.358  | 0.003*  |
| Functional Limitation         | 48.95±14.25                        | 10.37±22.55                         | 13.71-44.32 | 3.823  | 0.001*  |
| Maximum Mouth Opening (mm)    | 31.85±9.08                         | 38.39±6.83                          | -8.54-4.30  | -4.228 | <0.001* |
| Cervical Angle (°)            | 49.22±8.01                         | 52.87±7.04                          | -9.10-1.01  | -2.331 | 0.009*  |
| Thoracic Rotation (Right) (°) | 51.92±11.81                        | 58.76±15.43                         | -14.28-0.18 | -2.149 | 0.045*  |
| Thoracic Rotation (Left) (°)  | 48.35±13.80                        | 60.23±15.55                         | -22.12-1.76 | -2.453 | 0.023*  |
| Spine Measurements (°)        |                                    |                                     |             |        |         |
| Sagittal Plane (Angle)        |                                    |                                     |             |        |         |
| Thoracic                      | 40.76±10.91                        | 34.42±14.46                         | -1.33-13.82 | 1.859  | 0.111   |
| Lumbar                        | -34.44±11.87                       | -39.86±5.37                         | 0.32-10.81  | 2.439  | 0.024*  |
| Sacral                        | 18.73±9.53                         | 25.10±5.51                          | -10.91-1.83 | -2.940 | 0.008*  |
| Inclination                   | -0.43±4.02                         | -0.85±4.27                          | -1.71-2.86  | 0.483  | 0.634   |
| Sagittal Plane (Mobility)     |                                    |                                     |             |        |         |
| Thoracic                      | 31.47±20.24                        | 38.42±12.78                         | -11.52-1.83 | -1.698 | 0.107   |
| Lumbar                        | 61.32±10.18                        | 64.92±14.27                         | -9.20-3.20  | -1.010 | 0.325   |
| Sacral                        | 91.49±17.26                        | 49.38±18.25                         | 12.72-32.32 | 3.332  | 0.000*  |
| Inclination                   | 127.19±14.40                       | 122.05±13.35                        | -3.02-13.31 | -1.010 | 0.304   |
| Frontal Plane (Angle)         |                                    |                                     |             |        |         |
| Thoracic                      | 8.42±5.08                          | 7.10±4.16                           | -0.80-3.88  | 1.884  | 0.108   |
| Lumbar                        | 5.09±2.99                          | 5.48±11.35                          | -10.24-2.24 | -1.337 | 0.185   |
| Sacral                        | 3.05±1.91                          | 8.31±11.01                          | -5.64-2.21  | -1.308 | 0.758   |
| Inclination                   | 46.85±12.98                        | 52.24±16.48                         | 0.27-1.33   | 2.798  | 0.011*  |
| Frontal Plane (Mobility)      |                                    |                                     |             |        |         |
| Thoracic                      | 58.14±14.72                        | 57.33±17.08                         | -6.97-6.59  | -0.319 | 0.753   |
| Lumbar                        | 32.90±10.41                        | 36.90±12.97                         | -10.24-2.24 | -1.337 | 0.186   |
| Sacral                        | 8.05±7.17                          | 9.05±10.08                          | -5.84-4.88  | -0.374 | 0.713   |
| Inclination                   | 48.85±12.98                        | 52.24±16.48                         | -8.97-1.91  | -1.338 | 0.186   |

Our comparisons were millimeter, X<sup>±</sup> mean, SD: Standard deviation, 95% Confidence Interval, t: Paired sample t-test, p: p-value obtained from statistical comparison between pre- and post-intervention values

## Discussion

Treatment led to reduced pain, functional limitations, and improved mouth opening, along with enhancements in cervical posture, thoracic rotation, and lumbosacral mobility. These findings suggest that targeting postural and mobility issues beyond the jaw may aid symptom relief and functional recovery in TMD.

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### Association of reaction time with trunk muscle strength, endurance and balance in healthy young adults

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### Introduction

Reaction time (RT) is an important sport-related physical fitness parameter, involving the activation of muscle groups after the processing of sensory information by the central nervous system (1,2). In healthy young adults, fast reactions are influenced not only by neural speed but also by good balance and stabilization, which are crucial for controlling distal movements (3,4). Trunk muscle strength and endurance are essential for static and dynamic stability in postural control and movement (5). Investigating the link between trunk muscle function, balance and RT can provide valuable insights into the sensorimotor demands of functional activities (6).

### Research Question

Is there an association between RT, trunk muscle function and balance in healthy young adults?

### Methods

The study included 46 healthy young adults (23 females, 23 males, age:  $21.17 \pm 1.69$ , BMI:  $22.16 \pm 4.15$  kg/m<sup>2</sup>). RT was assessed for dominant, non-dominant side, and bilaterally using Test You Brain Pro system (TestYou, Krakow, Poland) with visual/auditory and perceptual/cognitive tasks (7). Trunk muscle strength was measured with a Takkei back-leg dynamometer (T.K.K. 5102 BACK-D, Japan), and endurance with 'Lateral Bridge' and 'Biering Sorenson' tests (8-10). Static balance was evaluated in a bipedal stance with eyes open using the Prokin system. (Technobody, Prodotti, Italy) assessing average pressure, forward-backward and medial-lateral standard deviations, and average forward-backward speed. Dynamic balance was recorded as stability limits in percentage (11). Data normality was assessed using the Shapiro-Wilk test, and Pearson correlation was performed.

### Results

Trunk muscle strength and bilateral RT were negatively correlated ( $p=0.034$ ,  $r=-0.313$ ). There was no correlation between dominant, non-dominant and bilateral measurements of RT and trunk extensor muscle endurance ( $p=0.362$ ;  $p=0.192$ ;  $p=0.116$ ), while lateral muscle endurance was negatively correlated for right ( $p=0.028$ ,  $r=-0.324$ ;  $p=0.047$ ,  $r=-0.283$ ;  $p=0.021$ ,  $r=-0.340$ ) and left ( $p=0.045$ ,  $r=-0.274$ ;  $p=0.032$ ,  $r=-0.317$ ;  $p=0.005$ ,  $r=-0.410$ ) sides. RT was negatively correlated with left and bilateral average center of pressure in the y-direction ( $r=-0.312$ ,  $p=0.035$ ;  $r=-0.310$ ,  $p=0.036$ ). A significant negative correlation was found between left-sided RT and standard deviation of center of pressure in the frontal plane ( $p=0.049$ ,  $r=0.292$ ). Additionally, the right-sided RT was positively correlated with average mediolateral sway speed ( $p=0.048$ ,  $r=0.293$ ). There was a negative correlation between RT on the right side and the anterior dynamic limits of stability ( $p=0.016$ ,  $r=-0.354$ ). Details are presented in Table 1.

Table 1. Correlation Between Reaction Time and Trunk Muscle Strength, Endurance, and Balance Parameters

| Parameters                                       | Reaction Time (Right) | Reaction Time (Left) | Reaction Time (Bilateral) |
|--------------------------------------------------|-----------------------|----------------------|---------------------------|
| Trunk muscle strength                            | r                     | -0.344               | -0.342                    |
|                                                  | p                     | 0.102                | 0.105                     |
| Lateral Muscle Endurance (Right)                 | r                     | -0.324*              | -0.283                    |
|                                                  | p                     | 0.028*               | 0.047*                    |
| Lateral Muscle Endurance (Left)                  | r                     | -0.274*              | -0.317*                   |
|                                                  | p                     | 0.045*               | 0.032*                    |
| Extensor muscle endurance                        | r                     | -0.138               | -0.190                    |
|                                                  | p                     | 0.362                | 0.192                     |
| Static Balance                                   |                       |                      |                           |
| Average Center of Pressure in X-Direction (COPx) | r                     | 0.215                | 0.033                     |
|                                                  | p                     | 0.133                | 0.834                     |
| Average Center of Pressure in Y-Direction (COPy) | r                     | -0.263               | -0.312*                   |
|                                                  | p                     | 0.077                | 0.035*                    |
| Forward-Backward Standard Deviation              | r                     | 0.154                | 0.292*                    |
|                                                  | p                     | 0.306                | 0.049*                    |
| Medial-Lateral Standard Deviations               | r                     | 0.031                | 0.031                     |
|                                                  | p                     | 0.840                | 0.828                     |
| Average Forward-Backward Speed                   | r                     | 0.101                | 0.140                     |
|                                                  | p                     | 0.505                | 0.354                     |
| Average Medial-Lateral Speed                     | r                     | 0.293*               | 0.030                     |
|                                                  | p                     | 0.048*               | 0.699                     |
| Parameters                                       | r                     | 0.127                | 0.096                     |
|                                                  | p                     | 0.401                | 0.603                     |
| Dynamic Balance                                  |                       |                      |                           |
| Right Side                                       | r                     | 0.080                | -0.017                    |
|                                                  | p                     | 0.597                | 0.917                     |
| Forward-Right Side                               | r                     | -0.124               | -0.105                    |
|                                                  | p                     | 0.412                | 0.272                     |
| Forward                                          | r                     | -0.354*              | -0.239                    |
|                                                  | p                     | 0.016                | 0.109                     |
| Forward-Left                                     | r                     | -0.063               | -0.014                    |
|                                                  | p                     | 0.660                | 0.926                     |
| Left                                             | r                     | -0.011               | 0.109                     |
|                                                  | p                     | 0.944                | 0.485                     |
| Backward-Left                                    | r                     | -0.194               | 0.014                     |
|                                                  | p                     | 0.198                | 0.926                     |
| Backward                                         | r                     | 0.178                | 0.174                     |
|                                                  | p                     | 0.237                | 0.248                     |
| Backward-Right                                   | r                     | 0.009                | -0.053                    |
|                                                  | p                     | 0.960                | 0.727                     |
| Total                                            | r                     | -0.223               | -0.191                    |
|                                                  | p                     | 0.138                | 0.201                     |

r = Pearson correlation coefficient, p = significance level

### Discussion

These findings suggest that better lateral trunk endurance and back strength are linked to faster RT, highlighting the role of trunk

muscle function, particularly lateral control. Additionally, decreased postural stability, indicated by increased anteroposterior sway, was associated with slower RT, while greater mediolateral sway velocity negatively affected RT, emphasizing the importance of precise postural control, especially for lateral tasks. This supports the proximal-distal relationship where trunk stability enhances RT, a distal motor performance.

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## The clinical role of 3d gait analysis on orthotic management for cerebral palsy: a systematic review

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## Introduction

Orthotic management for individuals with cerebral palsy (CP) has traditionally relied on visual gait analysis. The advent of 3D gait analysis (3DGA) offers significant advancements by providing objective and detailed measurable outcomes, which are especially valuable for CP patients whose gait patterns evolve. While experimental studies have explored various orthotic approaches for CP, few systematic reviews have focused explicitly on 3DGA as the primary outcome measure. This study aimed to evaluate the existing literature on the role of 3DGA in orthotic management for CP, offering clinicians evidence-based insights to optimize orthotic recommendations.

## Research Question

What is the primary role of 3DGA in orthotic management for cerebral palsy, and how can clinicians rely on it in 3D gait labs when receiving referrals for the orthotic treatment of CP patients?

## Methods

A systematic review followed PRISMA guidelines, utilizing validated keywords to search PubMed and Web of Science for articles published between 2009 and 2024. Duplicate removal and abstract screening refined the initial pool of 858 articles, resulting in 15 eligible studies for analysis.

Keywords:

Population: "Cerebral Palsy" OR CP

Intervention: orthos\* OR AFO OR orthotic OR brace OR "orthotic management" OR "leaf spring" OR orthoses OR "orthotic devices" OR "floor reaction" OR "ground reaction" OR "solid AFO" OR "hinged AFO" OR orthosis

Comparison: None

Outcome: gait OR "gait disorders" OR "gait analysis" OR "3d gait analysis" OR moment OR power OR "ground reaction force" OR "spatial temporal" OR biomechanic\* OR walking OR kinematic\* OR kinetic\* OR "muscle activity" OR EMG

## Results

The selected studies, involving 567 children with CP, highlighted the primary applications of 3DGA in orthotic management, including evaluating orthotic effectiveness, comparing the biomechanical impacts of various orthoses, and investigating orthotic tuning methods. Notable findings included improved outcomes through

personalized ankle-foot orthosis (AFO) modifications, particularly for individuals with diplegia and hemiplegia. However, unresolved issues such as increased internal foot progression angle with AFO use and the choice between unilateral and bilateral orthoses for hemiplegia were identified. The review emphasized the need for case-specific considerations and revealed gaps in clinical guidelines for interpreting and applying 3DGA findings.

## Discussion

3DGA facilitates fine-tuning orthotic prescriptions for CP, yet clinical guidelines for its integration into orthotic management remain underdeveloped. Future research should prioritize kinetic analyses, electromyography assessments with orthosis, and standardized protocols for orthotic adjustments. Advancements in 3DGA, including AI-driven predictive models and real-time data integration, hold promise for enhancing orthotic design and personalized management strategies.

## Limitations

The inability to conduct a quality assessment or meta-analysis, due to the majority of observational studies that are prone to various forms of bias, with a lack of high-quality randomized controlled trials.

The limited number of studies meeting inclusion criteria, as the focus was specifically on using 3D gait analysis to evaluate biomechanical impacts of orthotic interventions for cerebral palsy.

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## Comparing Theia3D markerless to marker-based lower limb kinetics during walking in typically developing and children with cerebral palsy

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### Introduction

Clinical gait analysis is essential for guiding treatment in children with cerebral palsy (CP), but traditional marker-based (MB) systems require more time, encumbrance, and patient compliance. Theia3D, a video-based markerless system<sup>1</sup>, offers a potential alternative<sup>2-3</sup>, but its accuracy in measuring joint kinetics in children with CP remains understudied.

### Research Question

How does Theia3D compare with MB measures of lower limb joint kinetics during overground walking in typically developing and children with CP?

### Methods

Ten typically developing (TD) children (4M:6F, 13.8 ± 3.7 yrs, 1.6 ± 0.2 m, 49.7 ± 13.8 kg) and 10 children with CP (GMFCS I-IV; 8M:2F, 12.5 ± 3.8 yrs, 1.5 ± 0.2 m, 47.6 ± 21.9 kg) participated. Fifty-four markers were placed on the trunk and lower limbs. Marker and video data were captured at 100 Hz using 12 optoelectronic cameras (Kestrel 4200, Motion Analysis Corp, US) and 10 synchronized video cameras (Miquis Hybrid, Qualisys, SE). Ground reaction forces were recorded at 1,000 Hz using seven embedded force plates (400600HPS, AMTI Inc., US). Participants completed ten walking trials over a 5 m walkway at a self-selected speed. Marker data were labeled and exported to Visual3D (v2024.08.5, HAS-Motion, CA) for model creation and joint kinetics calculation following ISB standards<sup>4</sup>. Video data were exported via the Gait Project Automation Framework (v2023.3, Qualisys, SE) to Theia3D (v2024.1.19, Theia Markerless Inc., CA), where a predefined Visual3D model was used for inverse dynamics. Kinetic waveforms (joint moments and power) were time-normalized to 100% stance phase (e.g., heel strike to toe-off). Eight stance cycles (four per limb) per motion capture system were exported to MATLAB (R2023b, MathWorks, US) for analysis. A multivariate Bayesian model examined group differences in root mean square differences (RMSDs). Waveform and peak value agreement were assessed using one-dimensional Statistical Parametric Mapping (1D-SPM;  $p < 0.05$ )<sup>5</sup> and Bland-Altman analyses, respectively.

### Results

A significant group effect was found for internal ankle plantar-flexion moment, with higher RMSD in children with CP than TD (-0.11 Nm/kg; 95% CI: [-0.20, -0.02]). Waveform analysis showed larger discrepancies in the CP group across the stance phase. In both

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groups, sagittal ankle power had the largest RMSD (TD: 0.33 W/kg; CP: 0.37 W/kg), while hip external rotation moment showed the best agreement (TD: 0.07 Nm/kg; CP: 0.16 Nm/kg). Bland-Altman analysis revealed wider limits of agreement and greater variability in children with CP, particularly for ankle and hip moments (Figure 1).

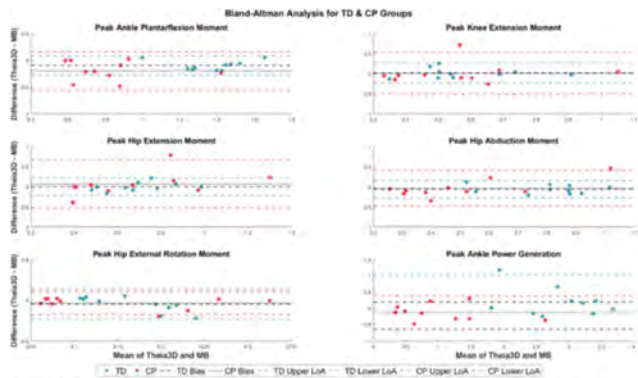


**Figure 1.** Bland-Altman plots comparing Theia3D and MB motion capture for peak joint moments and power in TD children and children with CP. The y-axis represents the difference (Theia3D - MB), while the x-axis shows the mean of Theia3D and MB measurements. Solid lines indicate the bias (mean difference), and dashed lines represent the 95% limits of agreement (LoA). A positive bias indicates that Theia3D overestimates joint moments compared to MB, while a negative bias indicates underestimation. TD data points and LoA are in blue, while CP data points and LoA are in red. The LoA range defines the measurement variability, with narrower LoA indicating stronger agreement and wider LoA suggesting greater variability between the two systems.

## Discussion

Theia3D may be an alternative to MB for measuring lower limb joint kinetics in pediatrics, but caution is needed for atypical gait, especially at the ankle. Despite similar waveform patterns, Theia3D underestimated plantarflexion moment and ankle power during late stance in children with CP. Discrepancies may reflect modeling differences and system-specific limitations.

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## Comparing Clinician-Measured, Marker-Based, and Markerless Leg Length and Margin of Stability Measures in Children With and Without Cerebral Palsy

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## Introduction

Assessment of gait and stability is essential for guiding treatment and evaluating outcomes in individuals with movement disorders. While 3D marker-based (MB) motion capture is a clinical gold standard<sup>1</sup>, it has practical limitations such as the reliance on physical landmarking and specialized laboratory environments. Video-based markerless (ML) motion capture technologies offer a promising alternative by eliminating the need for markers, faster data collection and processing, and can enable assessments in real-world settings<sup>2–6</sup>. The primary aims of this study were to 1) compare clinician-measured (CM) and motion capture estimates of leg length (LL) in children with cerebral palsy (CP) and 2) to quantify differences in LL and mediolateral margin of stability (MoS<sub>ML</sub>) between MB and ML systems in both typically developing (TD) and children with CP.

## Research Question

*Do CM and MB methods yield comparable estimates of LL, and do MB and ML systems provide similar estimates of LL and MoS<sub>ML</sub> in children with and without CP?*

## Methods

Six children with CP (5 males, 1 female; age:  $13.11 \pm 4.15$  years; height:  $1.50 \pm 0.22$  m; weight:  $43.95 \pm 19.62$  kg) and eight TD children (2 males, 6 females; age:  $11.25 \pm 2.80$  years; height:  $1.50 \pm 0.09$  m; weight:  $39.72 \pm 10.85$  kg) performed barefoot 10-meter overground walking trials at their preferred walking speed. Clinician-measured LL was determined by a physical therapist measuring from the anterior superior iliac spine to the medial malleolus bilaterally. Synchronized MB and ML motion capture systems recorded data at 100 Hz using 12 optoelectronic cameras (Kestrel 4200, Motion Analysis Corp., US) and 10 video cameras (Miquis Hybrid, Qualisys, SE). Participants had 54 markers on the lower limbs and trunk. Marker data were processed in Visual3D (v2024.08.5, HAS-Motion, CA); video data were exported via the Gait Project Automation Framework (v2023.3, Qualisys, SE) to Theia3D (v2024.1.24, Theia Markerless Inc., CA), where a predefined Visual3D model was used. Joint centers and segment lengths were exported to MATLAB (R2023a, Natick, MA) to determine LL and calculate MoS<sub>ML</sub> during contralateral toe-off. Agreement between the CM, MB, and ML approaches was evaluated using Bland-Altman plots<sup>7</sup>, with mean bias and 95% limits of agreement ( $\pm 1.96$  SD).

## Results

For both CP and TD groups, most values fell within the 95% LoA, indicating consistency between methods (Figure 1). The MB system underestimated CM LL by 4 cm, and the ML system underestimated MB LL by 4 cm. Measures of MoS<sub>ML</sub> showed a negligible difference (0.01 m) between MB and ML.

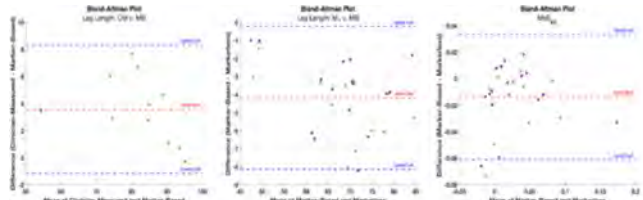


**Figure 1.** Bland-Altman plots comparing CM and MB (left panel) and ML and MB (middle panel) leg length, and medio-lateral margin of stability (MoSML; right panel) for both limbs across each participant. The y-axis represents the difference (CM–MB or ML–MB), while the x-axis shows the mean of the two measurements. The red dotted line indicates the mean bias, and the blue dashed lines indicate the lower and upper limits of agreement (LoA). A positive bias indicates that CM or ML overestimates LL or MoSML, while a negative bias indicates an underestimation. Children with cerebral palsy are indicated with green, while typically developing children are indicated in purple.

## Discussion

Markerless motion capture may be a viable alternative to MB for measuring MoSML. However, caution is warranted when interpreting LL estimates, as MB underestimated CM values, and ML underestimated MB.

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## Association of limits of stability with pain characteristics and functional mobility in adults with chronic neck pain: A case-control study

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## Introduction

Most studies suggest that people with chronic neck pain (CNP) have decreased standing balance. In these studies, static standing balance under different sensory conditions was predominantly assessed [1]. However, research on whether these impairments extend to dynamic balance is limited.

## Research Question

Do limits of stability (LoS) scores and Timed Up and Go test (TUG) times differ between adults aged between 18–65 years with and without CNP? Are LoS scores associated with neck pain characteristics and TUG times?

## Methods

Forty-five adults with CNP (66.7% female, mean age =  $42.5 \pm 10$ , mean BMI =  $27.5 \pm 4.4$  kg/m<sup>2</sup>) and forty-five asymptomatic adults (66.7% female, mean age =  $43.5 \pm 7.9$ , mean BMI =  $26.8 \pm 3.9$  kg/m<sup>2</sup>) were included. LoS scores were obtained for eight directions, with each target positioned at 45° intervals, using a stabilometric platform (Stability-E, TecnoBody, Italy) with a sampling rate of 20 Hz. The time taken to complete the test and the percentage of how much participants reached each of the eight targets were calculated by the device's software. TUG was performed at a self-selected comfortable speed under two conditions: without a concurrent task and while counting backwards by threes from a random number between 100 and 200. Analysis of covariance was performed to compare the groups adjusting for age and BMI. Correlations between LoS scores, TUG times, and pain characteristics were determined. Furthermore, multiple regression analysis was performed to determine whether pain characteristics predict the composite LoS score and TUG time.

## Results

Adults with CNP exhibited lower LoS scores in all directions except forward and backward, longer time taken to complete the LoS test, and longer TUG times under both conditions, with moderate to large effect sizes ( $\eta^2 p =$  between 0.073 and 0.392,  $p < 0.05$ ). In both groups, higher LoS scores were correlated with shorter TUG time while counting backwards ( $r =$  between -0.322 and -0.423,  $p < 0.05$ ), but not with TUG time without a concurrent task ( $p > 0.05$ ). After adjusting for demographics, pain intensity during neck movements and the Neck Disability Index score explained an additional 29.2% of the variance in the composite LoS score and an additional 15.7% of the variance in TUG time while counting backwards ( $p < 0.05$ ). Additionally, neck disability uniquely predicted both the composite LoS score ( $\beta = -0.413$ ,  $p < 0.05$ ) and TUG time while counting backwards ( $\beta = 0.391$ ,  $p < 0.05$ ), after adjusting for demographics and pain intensity.

## Discussion

Similar to previously reported results in older adults with CNP [2,3], the present study further suggested that LoS scores were lower in certain directions, and that TUG times, both with and without the counting backward task, were longer in adults aged 18–65 years,

most of whom had moderate to severe neck pain during neck movements and mild to severe neck disability. LoS can be addressed in the management of CNP, and its association with neck disability should be considered.

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## Postural sway during and within seconds after basic mobility tasks in relation to sway during static standing in young adults

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## Introduction

COP-based sway measures, mostly obtained during static standing, are widely used to assess postural stability [1]. However, their relevance to functional tasks remains unclear.

## Research Question

Does postural sway during and shortly after sit-to-stand, gait termination, and step-down tasks differ between young adults grouped by their static standing sway? Do sway measures within seconds after these tasks differ from those during static standing?

## Methods

Twenty young adults were included. Postural sway data were collected at 40 Hz using a pressure-sensitive mat (HR Mat, Tekscan,

USA) and analyzed using FootMat™ Software for Researchers [2,3]. All measurements were performed barefoot. Area, distance, variability, and anterior-posterior and mediolateral excursions were obtained during static standing, sit-to-stand, step-down, and gait termination. For static standing, participants stood on the mat with feet shoulder-width apart and arms at their sides. For sit-to-stand, they sat on a 40.64 cm-high block and were instructed to stand up within one second and remain stable. For gait termination, participants approached the mat using a two-step method, stepping onto it with the dominant foot first, then the non-dominant, and maintained stable bipedal standing [4]. For step-down, participants stepped down from a 20.32 cm-high wooden step, landing first with the dominant, then the non-dominant foot onto the mat, each step completed within one second, and remained stable. Two analysis periods were defined: an initial 3-second period and a second 3-second period. The initial moment was defined separately for tasks. For sit-to-stand, it was the first frame displaying a force map. For step-down and gait termination, it was the first frame showing force under the non-dominant foot as it was being placed on the mat, following the already positioned dominant foot. Participants were grouped as low- or high-sway based on the median of a composite score derived from the z-scores of area, distance, and variability during static standing. Group differences were analyzed with t-tests, and task differences with repeated measures ANOVA.

## Results

All sway measures during static standing differed between groups ( $p < 0.05$ ). Sway during the first 3 seconds of dynamic tasks did not differ between groups ( $p > 0.05$ ). In the second 3-second period after step-down, the high-sway group showed greater distance and anterior-posterior excursion ( $p < 0.05$ ). When comparing sway obtained in the second 3-second period across tasks, area, anterior-posterior and mediolateral excursions were higher in all mobility tasks than in the static standing task, and distance and variability were higher in the sit-to-stand task than in both the static standing and step-down tasks ( $p < 0.05$ ).

## Discussion

Postural sway measured during the second 3-second period, while participants maintained standing stability shortly after tasks, differed from static standing. Additionally, lower static sway did not consistently indicate lower sway during and within seconds after basic mobility tasks, as between-group differences were found only after step-down.

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## Relationship of regional sagittal and frontal spinal curvatures and mobility with respiratory function and balance in adolescents with idiopathic scoliosis

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## Introduction

Respiratory function and postural balance are impaired in adolescent idiopathic scoliosis (AIS), even in mild cases [1-4]. A skin-surface measurement device, reported to be reliable and valid for assessing spinal curvatures in AIS, offers a practical and radiation-free alternative, though not a replacement for radiography [5]. However, whether spinal measures using a skin-surface device are associated with respiratory function and balance in AIS remains unclear.

## Research Question

Are regional sagittal and frontal spinal alignment and mobility associated with respiratory function and balance in subjects with AIS?

## Methods

Thirty-one participants with AIS (10–18 years, Cobb angles 10°–30°) were included. A skin-surface device (Hocoma, Valedo®Shape, Idig GmbH, Switzerland) was used to assess sagittal and frontal spinal alignment and mobility of the thoracic, lumbar, and sacral regions and trunk inclination. Angles measured in the usual standing posture represented alignment, while flexion-extension and left-right lateral flexion ranges of motion represented sagittal and frontal mobility, respectively. Forced vital capacity (FVC), forced expiratory volume in 1 second (FEV1), FVC/FEV1, peak expiratory flow (PEF), and maximal inspiratory and maximal expiratory pressures (MIP and MEP) were measured using a digital spirometer (Pony FX, COSMED Inc., Italy). Balance was assessed using the Prokin system (ProKin 252, TecnoBody, Italy), including the composite limits of stability (LoS) score and the following sway parameters during quiet standing with eyes open: ellipse area, perimeter, forward-backward and medium-lateral standard deviations, average forward-backward and medium-lateral speeds, average center of pressure displacements on the x- and y-axes (COPX and COPY). Spearman correlation coefficients were calculated to determine the associations of spinal measures with respiratory and balance measures.

## Results

Sacral kyphosis was weakly correlated with FVC% and moderately with MIP% ( $\rho = 0.374$  and  $0.409$ ,  $p < 0.05$ ). Thoracic kyphosis was moderately correlated with FEV1 and weakly with PEF ( $\rho = 0.579$  and  $0.390$ ,  $p < 0.05$ ). Lumbar lordosis showed a moderate correlation with MIP% ( $\rho = 0.413$ ,  $p < 0.05$ ). Sagittal sacral mobility was moderately correlated with FEV1% ( $\rho = 0.428$ ,  $p < 0.05$ ). Sagittal thoracic mobility was moderately and positively correlated with medium-lateral standard deviation and negatively and weakly with the composite LoS score ( $\rho = 0.411$  and  $-0.389$ ,  $p < 0.05$ ). Sagittal inclinational mobility was moderately correlated with FVC%, FEV1%, and the composite LoS score and weakly with PEF% and MEP% ( $\rho = 0.359$ – $0.495$ ,  $p < 0.05$ ). Furthermore, frontal sacral angle was moderately and negatively correlated with PEF, PEF%, MIP, and MIP% ( $\rho = -0.441$ – $-0.506$ ,  $p < 0.05$ ). Frontal thoracic angle was weakly correlated with forward-backward standard deviation and moderately with ellipse area ( $\rho = 0.368$  and  $0.406$ ,  $p < 0.05$ ). Frontal inclination angle and frontal thoracic mobility were weakly correlated with COPX and COPY, respectively ( $\rho = 0.397$  and  $0.375$ ,  $p < 0.05$ ).

## Discussion

In subjects with AIS, increased frontal curvatures were related to poorer respiratory functions and balance; greater sagittal curvatures and mobility to better respiratory function. Increased thoracic mobility was related to poorer balance; increased sagittal trunk inclination mobility to better balance.

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## Relationship between changes in Gait Deviation Index, walking speed and 6-minute-walk-test in adults with cerebral palsy – a 16-year follow-up.

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### Introduction

Previous cross-sectional research shows a low correlation between GDI and 6MWT in children [1] and adults [2] with cerebral palsy (CP). Further, there is a lack of studies investigating changes and associations in walking function variables in adults with CP [3].

### Research Question

What are the associations between changes in gait pattern deviations, changes in self-selected walking speed, and changes in walking capacity in adults with CP over 16 years?

### Methods

Bivariate linear regression analysis of the changes in the Gait Deviation Index (GDI), in distance during the 6-minute walk test (6MWT), and self-selected walking speed during a 10-meter walk test (SSWS) between two time-points 16 years apart.

### Results

29 adults (15 females) with unilateral ( $n = 17$ ) and bilateral ( $n = 12$ ) CP with a mean age of 50 years ( $SD = 10$  years (range 34–65) and mean BMI 25 ( $SD 5.2$ ), gross motor function classification system I ( $n = 11$ ), II ( $n = 15$ ) and III ( $n = 3$ ).

The mean change in GDI was  $-5.6$  ( $SD 8.1$ ),  $-37$  m ( $SD 102$ m) in 6MWT distance and  $-0.08$  m/sec ( $SD 0.18$ m/sec) in SSWS. Bivariate linear regression showed a moderate association between GDI-change and 6MWT-change ( $r^2 = 0.351$ ,  $\beta = 0.593$ ,  $p < 0.001$ ) and between 6MWT-change and SSWS-change ( $r^2 = 0.401$ ,  $\beta = 0.640$ ,  $p < 0.001$ ). A low, non-significant association was found between GDI-change and SSWS-change ( $r^2 = 0.064$ ,  $\beta = 0.253$ ,  $p = 0.194$ ).

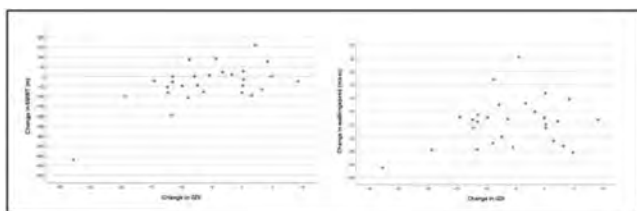


Figure 1. Scatter plot with GDI-change, 6MWT-change (m), and self-selected walking speed-change over 16 years in 29 adults with CP. Abbreviations: 6MWT = 6-minute walk test, GDI = Gait Deviation Index.

### Discussion

The results show that changes in gait pattern deviations are associated with changes in walking capacity (explaining 35% of the variance), but not with changes in self-selected walking speed over a short distance (10 meters). This may indicate that some adults with CP can maintain their self-selected walking speed over short distances despite increasing gait pattern deviations, however, struggle to do so over longer distances or extended periods (such as the 6MWT).

Further, this indicates that the variables investigated are inter-related and that common denominators affect the changes in both gait pattern deviations and walking capacity. Previous research has highlighted that plantar flexor strength [4], and hamstring length [5] influence walking capacity in adults with CP, which are variables also known to affect walking patterns in persons with CP [6].

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### Can Real-Time Feedback Correct Your Exercise Biomechanics? Investigating an AI-Supported Telerehabilitation Platform

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### Introduction

With the advancement of technology and the significant impact of the COVID-19 pandemic, telerehabilitation practices have rapidly evolved [1,2]. In particular, mobile applications and platforms that include only visual or video-based exercise instructions have become widespread to encourage exercise adherence [3]. However, it is well established that supervised exercises are more effective, and exercises performed with incorrect biomechanical patterns may lead to injury [4-6]. In this context, the corrective potential of emerging platforms like *Morla Moves*, which utilize pose estimation algorithms to promote accurate movement patterns during exercise, remains unclear [3].

## Research Question

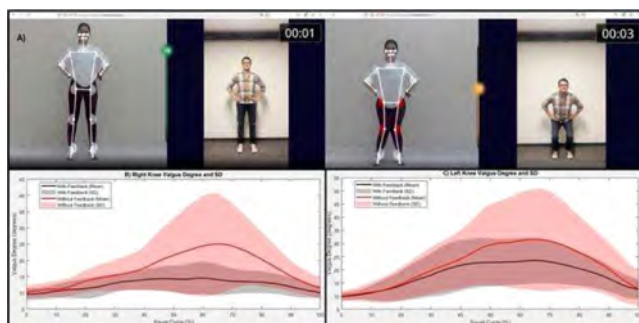
Does performing the squat exercise on an e-supervised feedback platform that analyzes real-time movement using a camera and provides corrective feedback affect the degree of knee valgus during the squat exercise?

## Methods

Thirteen participants (4 female) who met the inclusion criteria (no current injuries or illnesses, ability to perform a squat, and access to a camera-equipped computer) were recruited via social media. Each participant completed the following protocol: Using a beta test link on the *Morla Moves* platform, participants first entered their demographic data and performed a squat analysis in front of their camera. The resulting knee valgus degree (KVD) was recorded as the pre-test value. Participants then completed 10 repetitions of the squat exercise with real-time corrective feedback designed to minimize knee valgus (Figure 1-A). After the exercise session, participants repeated the squat analysis, and the post-test KVD was recorded. The maximum KVD values from the pre- and post-tests were compared using the Wilcoxon Signed Ranks Test. In addition, KVD values across the entire squat cycle were normalized to 101 data points and analyzed using Statistical Parametric Mapping (SPM) via MATLAB (spm1d.org). Statistical significance was set at  $p < .05$ .

## Results

Data from 13 participants (age:  $23.69 \pm 6.56$  years, height:  $1.78 \pm 0.10$  m, body weight:  $77.15 \pm 18.17$  kg) were analyzed. All participants successfully completed the protocol independently on the platform. Post-test maximum KVD values (median; right:  $15.1^\circ$ , left:  $21.1^\circ$ ) were significantly lower compared to pre-test values (median; right:  $26.4^\circ$ , left:  $35.0^\circ$ ) (right  $p = .011$ , left  $p = .013$ ). Although SPM analysis did not reveal statistically significant differences throughout the squat cycle, observable reductions in KVD were present across the movement (Figure 1-B, C).



## Discussion

These findings suggest that online platforms such as *Morla Moves* may help individuals perform exercises with more accurate joint alignment, potentially preventing incorrect movement patterns that could lead to injury. Moreover, such platforms could play an important role in enhancing the effectiveness and accessibility of telerehabilitation services.

Conflict of Interest: UZK, OO, and GB are co-founders of *Morla Moves* and serve as Exercise Domain Experts within the company.

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## Comparison of Running Kinematics Between Recreational Runners With and Without Shin Splints: A Pilot Study

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## Introduction

Shin splints (SS), also known as medial tibial stress syndrome, is a common overuse injury affecting recreational runners (1). Although several studies have been conducted in this area (2-4), there is a need for further research to better understand the biomechanical differences between recreational runners with and without shin splints.

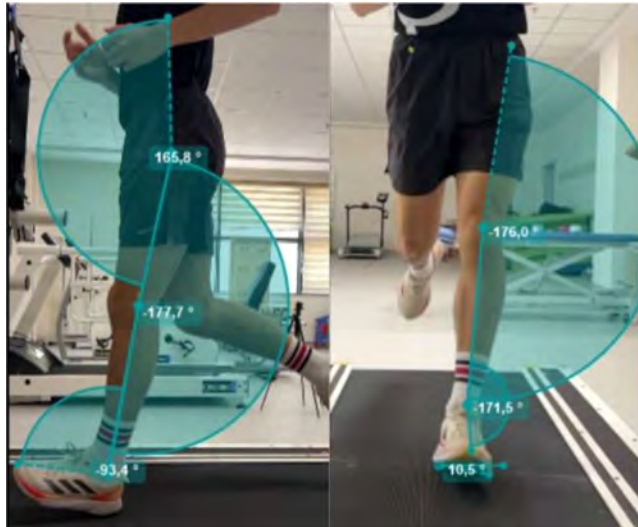
## Research Question

How do the lower limb kinematics differ between recreational runners with and without shin splints while running?

## Methods

A total of 8 recreational runners (4 female, mean age:  $23.7 \pm 3.3$ ; mean body mass index:  $21.9 \pm 1.5$ ; mean weekly running distance:  $8.6 \pm 4.6$  km) were divided into shin splint (SG) and non-shin splint (NSG) groups based on diffuse pain/discomfort on the lower 1/3rd

of the posteromedial border of the tibia induced by running. Sixteen reflective markers were placed on anatomical landmarks of the participants. The running kinematics were analyzed at 10 km/h speed running on treadmill, after a 5-minute warm-up with self-selected pace. Video recordings were captured from the front, rear, and lateral views using three iPhone 14 cameras (Hz, Apple Inc., USA) positioned at a distance of 100 cm from the treadmill during running trials. The recorded videos were analyzed using Kinovea software to measure lower limb kinematics (Figure 1). The Mann-Whitney U test was used to compare running kinematics between groups, with the significance level set at  $p < 0.05$ .



## Results

The median value of foot pronation was  $3.8^\circ$  in the NSG and  $4.5^\circ$  in the SG; foot eversion was  $8.5^\circ$  in the NSG and  $12.6^\circ$  in the SG; dorsiflexion was  $0.7^\circ$  in the NSG and  $1.3^\circ$  in the SG; knee flexion was  $13.9^\circ$  in the NSG and  $11.8^\circ$  in the SG; hip flexion was  $26.6^\circ$  in the NSG and  $26.2^\circ$  in the SG; frontal plane projection angle was  $175^\circ$  in the NSG and  $171.1^\circ$  in the SG. However, no statistically significant differences were observed in any of the kinematic parameters between the NSG and SG groups.

## Discussion

This pilot study did not identify significant differences in lower limb kinematics between recreational runners with and without shin splints while running. However, further research involving larger populations is warranted to better understand kinematic differences between these groups. As a pilot study, the small sample size limits the statistical power and generalizability of the findings, while the use of 2D motion analysis may reduce sensitivity to subtle kinematic differences, and the cross-sectional design with self-reported symptoms may limit diagnostic accuracy and causal interpretations.

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## Sensory Integration is Partially Associated with Gait Speed in Healthy Community-Dwelling Older Adults: A Systematic Review and Meta-analysis

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## Introduction

Integration of sensory inputs from the visual, somatosensory, and vestibular systems within the central nervous system is defined as sensory integration (SI), and it is important for regulating movements such as walking<sup>1,2</sup>. Aging, however, often leads to impairments in these systems, which may contribute to changes in gait among older adults<sup>3,4</sup>. Despite this, the association between gait speed and SI has yet to be thoroughly investigated.

## Research Question

Is there an association between gait speed and SI in healthy community-dwelling older adults?

## Methods

A systematic literature search was conducted in the PEDro, PUBMED, MEDLINE, and Web of Science databases in January 2025. The studies that evaluated SI and gait in healthy community-dwelling older adults were deemed eligible. The quality of the studies was assessed with Quality in Prognosis Studies (QUIPS). Correlation coefficients between gait speed and postural sway during standing with eyes open on a firm surface (EOF), eyes closed on a firm surface (ECF), eyes open on a compliant surface (EOC), and eyes closed on a compliant surface (ECC) were extracted. To examine the association between gait speed and *reliance* on visual and somatosensory inputs, correlation coefficients were calculated between gait speed and the Romberg Quotient ( $RQ = ECF/EOF$ ) and the Proprioception Quotient ( $PQ = EOC/EOF$ ), respectively<sup>5</sup>. The postural sway outcome measurements were grouped as sway area and sway velocity.

## Results

Thirteen studies comprising 719 older adults (mean age 72.5 years, age range 60-102 years, 66.82% female) were included. A high risk of bias was identified across all studies. There were significant associations between gait speed and *sway area* during EOF ( $r: -0.235$ ,  $p < 0.001$ ), ECF ( $r: -0.201$ ,  $p < 0.001$ ), EOC ( $r: -0.198$ ,

$p < 0.001$ ), and ECC ( $r: -0.186$ ,  $p < 0.004$ ). No significant associations were observed between gait speed and sway velocity across any condition, including the Romberg Quotient (RQ) and Proprioception Quotient (PQ) ( $p > 0.486$  for all), except for the in ECS condition, which showed a weak negative correlation ( $r = -0.149$ ,  $p = 0.01$ ).

## Discussion

In conclusion, both greater sway area across all conditions and greater sway velocity in ECS were associated with slower gait speed. On the other hand, gait speed was surprisingly not linked to RQ or PQ. This finding may stem from the fact that SI enables compensation for deficiencies of a sensory system in healthy older adults<sup>6</sup>. It may also be due to the absence of SI measurements in dynamic contexts, as assessing SI in static conditions may not fully reflect the demands of gait. We recommend that future research prioritize the evaluation of SI during dynamic tasks that more accurately reflect the demands of gait.

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## Impact of Femoral Derotation Osteotomies on Gait, Growth Plate Loading, and Femoral Growth Trajectories in Children with Idiopathic Torsional Deformities

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## Introduction

Bone growth and shape is influenced by mechanical loads experienced during daily activities such as walking. Torsional femoral deformities - defined as an increased or decreased anteversion angle (AVA) compared to healthy controls - are common, often causing pain, altered gait, and increase the risk for osteoarthritis [1]. Femoral derotation osteotomies (FDO) are used to correct these deformities, but their impact on bone loading and further growth remains unclear. Importantly, such insight is essential to determine the optimal timing and extent of correction needed for long-term benefit.

## Research Question

How do FDO affect bone loads and future growth trajectories?

## Methods

Three-dimensional gait analysis (3DGA) data, including marker trajectories and ground reaction forces, was recorded before ( $10.2 \pm 8.4$  months) and after ( $18.4 \pm 14.2$  months) an intertrochanteric FDO of five patients with idiopathic torsional deformities - two with increased AVA and three with decreased AVA - aged  $13.4 \pm 1.8$  years at the surgery. Magnetic resonance images (MRIs) were recorded before the surgery. MRIs were used to quantify AVA, neck-shaft-angle, inter-epicondylar distance and tibial torsion to personalize musculoskeletal models. For the post-surgery models, AVA was adjusted based on the surgery report. Subsequently, pre- and post-surgery 3DGA data were used to calculate joint angles, muscle forces and joint contact forces (JCFs) using OpenSim [2,3]. The root-mean-square-difference (RMSD) between patients' and typically developing children' joint angles (from [4]) was quantified.

For each femur, a subject-specific finite element model was created [5]. A second model was created by virtually performing surgery on the segmented femur. Muscle and JCFs from the musculoskeletal simulations (pre- and post-FDO) were applied as nodal forces. Shear and compressive stresses in the growth plate were quantified and used to calculate the osteogenic index (OI) [6]. Subsequently, growth trends, i.e. change in AVA, were predicted based on the OI and our recently calibrated mechanobiological model [4].

## Results

The RMSD of joint kinematics to healthy individuals decreased significantly in all patients from  $12.4 \pm 3.4^\circ$  to  $7.9 \pm 2.5^\circ$ . The hip JCF and its orientation changed in all participants. Interestingly, the OI was very similar before and after surgery. Furthermore, the predicted change of AVA did not significantly change due to the surgery (Figure 1).

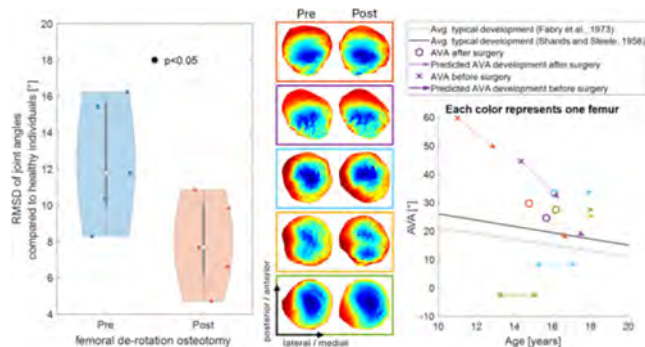


Figure 1: Left: RMSD of joint angles compared to healthy children. Middle: Osteogenic index (red = promoted growth; blue = inhibited growth) within the proximal growth plate before and after surgery. Right: Pre- and post-surgery AVA and predicted growth trajectories within the following two years. Each color represents one femur.

## Discussion

This is the first study which showed how FDOs influence growth plate stresses and femoral growth trends. Our workflow can be used to select the ideal timing for the FDO and identify patients where an over- or under-correction of the AVA is required to achieve the desired long-term outcome.

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## Investigation of the effects of myopia and occlusion disorders on balance

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## Introduction

Balance is the ability to keep the body's center of gravity vertically within the limits of the support surface. (1) The relationship of the visual system, which is one of the important subcomponents in the provision of balance, with occlusion disorders has been investigated in recent years. (2,3) Myopia, which is one of the most common refractive errors among visual system disorders, may lead to poor distance vision and increased balance problems. (4) Although it is not possible to make a clear cause-effect definition in the relationship between visual problems and malocclusions, it has been shown that malocclusions may increase the prevalence of myopia (5). Inappropriately treated myopia and temporomandibular joint occlusion disorders may cause problems in the organization of sensory inputs required for balance.

## Research Question

The aim of this study was to investigate the effects of the relationship between myopia and occlusion disorders on balance.

## Methods

The study enrolled 67 participants. Group1(myopia) n=36, Group2 (control) n=31. Myopia was evaluated using a Nidek autorefractometer device and recorded 30 minutes after cycloplegic drops were instilled into the patients. The occlusion assessment was performed by the dentist according to the Angle Classification in the sagittal plane, using a mirror and probe in the dental unit and under reflector light. Biodex Balance System was used to assess balance, and participants were evaluated with Limits of Stability (LoS) test.

## Results

There was no statistically significant difference between the right (RO) and left occlusion (LO) class distributions of Group 1 and Group 2 subjects (For RO:  $\chi^2(5, N=67)=4.04, p=0.553$ ; for LO:  $\chi^2(5, N=67)=4.88, p=0.458$ ). Statistically significant differences were observed between RO and LO classes for losforward ( $F=33.34, p<0.001$ ), ( $F=14.86, p<0.001$ ) and losbackward ( $F=22.10, p<0.001$ ), ( $F=11.39, p<0.001$ ) parameters. When the relationship between the degree of right refractive error (RRE) in the right eye and left refractive error (LLE) and balance parameters was examined,

it was found that as the RRE and LRE increased, losbackwardright ( $\beta = -4.77$ ,  $p < 0.001$ ,  $R^2 = 0.17$ , SD:  $\pm 12.2$ ), ( $\beta = -3.65$ ,  $p = 0.006$ ,  $R^2 = 0.11$ , SD:  $\pm 13.1$ ) losforwardleft ( $\beta = -4.22$ ,  $p < 0.001$ ,  $R^2 = 0.16$ , SD:  $\pm 11.6$ ), ( $\beta = -3.13$ ,  $p = 0.011$ ,  $R^2 = 0.095$ , SD:  $\pm 12.7$ ) and losright ( $\beta = -3.05$ ,  $p = 0.009$ ,  $R^2 = 0.10$ , SD:  $\pm 11.4$ ), ( $\beta = -2.49$ ,  $p = 0.027$ ,  $R^2 = 0.073$ , SD:  $\pm 11.6$ ) decreased.

## Discussion

According to the results of the study, it was observed that different occlusion classes affected the stability limits especially in the forward and backward directions. In myopic subjects, a significant negative increase was observed in the transverse stability limits as the refractive error increased. In addition to refractive errors, it can be said that occlusion disorders are also effective on balance.

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## Does the anterior tibiofibular gap influence subjective ankle instability?

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## Introduction

Approximately 17.8% of all lateral ankle ligament injuries involve syndesmosis injuries [1]. Most patients with chronic ankle instability (CAI) have subjective ankle instability of the talocrural joint due to damage to the anterior talofibular ligament (ATFL) from repeated ankle sprains. However, the relationship between tibiofibular joint instability and subjective ankle instability remains unknown.

## Research Question

Does the anterior tibiofibular gap (ATFG) differ between individuals with and without CAI, and is it associated with subjective ankle instability?

## Methods

Sixteen healthy young adults were classified into the CAI group (n=8, 16 feet, mean age: 20.9  $\pm$  0.8 years) and the control group (n=8, 16 feet, mean age: 21.9  $\pm$  0.8 years) based on Cumberland Ankle Instability Tool (CAIT) scores. ATFG and the distance between the talus and lateral malleolus were assessed using ultrasonography (US, Konica Minolta Inc., Tokyo, Japan) in sitting and standing positions. The ankle was placed on a tilt table at three different angles: 20° plantar flexion (P20), 0° neutral (N), and 20° dorsiflexion (D20). For ATFG, the US probe was positioned to observe the anterior tibiofibular ligament, defined as the distance between the fibula and tibia on the US image. The distance between the talus and lateral malleolus was assessed by imaging the anterolateral aspect of the lateral neck of the talus and lateral malleolus along the ATFL. Unpaired t-tests compared ATFG and the distance between the talus and lateral malleolus between groups in each position. To confirm the correlations between ATFG and the distance between the talus and lateral malleolus in each position, Pearson's correlation coefficients were applied. Additionally, multiple regression analysis with forced entry was performed to examine the relationship between CAIT scores and both ATFG and the distance between the talus and lateral malleolus.

## Results

The ATFG and the distance between the talus and lateral malleolus were significantly wider in the CAI group than in the control group in both sitting and standing positions at P20 and N (Table1).

|          |     | Anterior tibiofibular gap (mm) |               |         |             | Distance between the talus and lateral malleolus (mm) |                |         |             |
|----------|-----|--------------------------------|---------------|---------|-------------|-------------------------------------------------------|----------------|---------|-------------|
|          |     | CAI group                      | Control group | p value | Effect size | CAI group                                             | Control group  | p value | Effect size |
| Sitting  | P20 | 8.0 $\pm$ 0.3                  | 4.6 $\pm$ 0.2 | <0.001b | 2.955       | 21.2 $\pm$ 4.3                                        | 19.6 $\pm$ 2.0 | 0.015a  | 0.913       |
|          | N   | 5.0 $\pm$ 0.6                  | 5.2 $\pm$ 0.5 | <0.001b | 2.441       | 19.8 $\pm$ 3.3                                        | 17.8 $\pm$ 5.8 | 0.020b  | 0.900       |
|          | D20 | 6.8 $\pm$ 0.2                  | 5.8 $\pm$ 0.2 | <0.001a | 1.996       | 18.6 $\pm$ 2.6                                        | 15.3 $\pm$ 6.1 | 0.139a  | 0.542       |
| Standing | P20 | 8.2 $\pm$ 0.3                  | 5.0 $\pm$ 0.1 | <0.001b | 2.735       | 22.7 $\pm$ 3.7                                        | 20.4 $\pm$ 2.3 | <0.001a | 1.318       |
|          | N   | 6.4 $\pm$ 0.5                  | 5.3 $\pm$ 0.1 | <0.001b | 2.300       | 20.8 $\pm$ 2.8                                        | 18.3 $\pm$ 3.2 | 0.011a  | 0.963       |
|          | D20 | 5.8 $\pm$ 0.2                  | 6.1 $\pm$ 0.2 | <0.001a | 1.355       | 18.5 $\pm$ 6.2                                        | 15.2 $\pm$ 6.2 | 0.190a  | 0.474       |

Values are presented as mean  $\pm$  SD. a shows the results of the unpaired t-test, b the Mann-Whitney U-test.

The CAIT score was lower in the CAI group compared to the control group. Significant positive correlations were found between ATFG and the distance between the talus and lateral malleolus under all conditions. Multiple regression analysis using the forced entry revealed that ATFG in the P20 position was a significant predictor of CAIT scores, whereas the distance between the talus and lateral malleolus in the same position was not. The model explained 64.6% of the variance in CAIT scores.

## Discussion

These findings suggest that US-measured ATFG reflects mechanical ankle instability and may explain subjective instability in individuals with CAI. This approach provides a new perspective on evaluating ankle instability. Importantly, the P20 position, where ankle instability often occurs during daily and sports activities, appears most sensitive for detecting subjective ankle instability.

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## How do different event detection algorithms affect the global scores – gait profile score and gait deviation index?

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## Introduction

Gait event detection is an important step in the evaluation of gait data. In their multicentre study, Visscher et al. [1] showed that event detection differs between laboratories with the same event detection method algorithms for estimating initial contact (IC) and toe off (TO). There are a variety of different event detection methods. Dumphart et al. [2] conducted a comparative analysis of heuristic and machine learning (ML) methods, utilising a common dataset to assess their performance. Furthermore, in a subsequent study, Dumphart et al. [3] demonstrated that even minor errors of  $\pm 2$  frames (13.3 ms @150 Hz) can result in substantial kinematic deviations, which exceeds the minimal detectable change. As demonstrated, different gait event detection methods have a significant influence on the kinematic and kinetic output [4]. In this study, we consider global gait scores, and the effects of event detection methods on GPS and GDS.

## Research Question

To what extent do different event detection methods influence the global GPS and GDI scores?

## Methods

In this retrospective study, 142 patients with various diagnoses (e.g. cerebral palsy, malrotation deformities, club foot, flat foot, X-linked hypophosphatemia, osteogenesis imperfecta) were selected from the existing laboratory database. Four different event detection methods were applied to this dataset: autocorrelation with force plate information (AC), two heuristic approaches (Ghoussayni [5] and Zeni [6]), and an ML approach (IntellEvent [2]). 284 averaged gait cycles, with each session comprising a minimum of 10 trials and

10 force plate hits per side were analysed. Repeated measures ANOVA was used to identify differences in gait profile score (GPS) and gait deviation index (GDI). Values for the minimal clinically important difference were 1.7 [7] for the GPS and 10% for the GDI. [DS1] [DS1]Braucht es hier eine Referenz?

## Results

A significant disparity is evident in GDI between IntelliEvent and AC, in addition to the two heuristic approaches. For GPS, all methods demonstrate highly significant differences from each other. In consideration of the minimal clinically important difference in the GPS, differences above 1.7° were identified between IntelliEvent and Zeni/Ghoussayni/AC in 18/0/0 out of 142 patients. For the GDI, the differences below 10% were present in 33/0/0 patients, and for the GDI-kinetic, in 97/23/1 patients. It is noteworthy that the diagnoses have no recognizable influence.

## Discussion

Our results show an influence of the event detection method on the GPS and GDI scores. Even the global scores exceeded the limits by up to 12% (GPS) and 23% (GDI). This is an important factor to consider for both clinical practice as well as research utilizing multicentre datasets across labs. We, therefore, further recommend to always state in the method section and clinical report which gait detection algorithm was used.

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## Does correcting valgus malalignment always result in a normalised medio-lateral knee joint contact force distribution?

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## Introduction

Genu valgum is among the most common reasons for paediatric orthopaedic consultations. In addition to considerations of aesthetics, the long-term implications of valgus malalignment remain to be fully elucidated. Consequently, the indication for corrective surgery varies significantly among pediatric orthopedic surgeons. Many surgeons tend to advocate surgery based on radiographic features to avoid potential knee osteoarthritis in the long-term. Holder et al. [1] showed that the contact forces normalise post-operatively with a pure frontal malalignment. However, a reduced tibial rotation can have a considerable effect on the frontal knee moment [2,3]. Further, in our clinic we observed an increased internal valgus moment in patients with a reduced external rotation of the tibia. Therefore, this study explores the knee joint contact forces in this population.

## Research Question

Does external tibial torsion following successful correction of a valgus malalignment in adolescents result in an increased medial knee compartment contact force?

## Methods

Patients treated for valgus malalignment with temporary hemiepiphysodesis were selected from our database. The inclusion criteria for patients in the database was as follows: availability of pre- and post-operative 3D gait analysis and a pre-operative mechanical tibiofemoral angle assessed by medical imaging. The results of the 3D gait analysis were processed using an automated OpenSim workflow [4]. It allowed to run musculoskeletal simulations with personalized geometry of the tibial torsion and frontal knee alignment based on the Rajagopal model with the “Lerner knee” to

calculate the medial/lateral knee forces[5]. 20 patients fulfilled the inclusion criteria and were divided into two groups: normal tibial torsion (NTT, n=12 legs) and reduced tibial torsion (RTT, n=28 legs) (lower than 1 SD). Pre and post results were compared by Statistical Parametric Mapping (SPM,[6]). Additionally, the results were compared to a group of typically developing adolescents.

## Results

Before surgery the medial contact force tended to be lower and laterally higher for the NTT. After surgery a significant increased medial force in the RTT was observed compared to the reference and NTT. RTT showed a higher medial and a lower lateral force component in relation to NTT (Fig. 1). All parameters showed a high standard deviation.

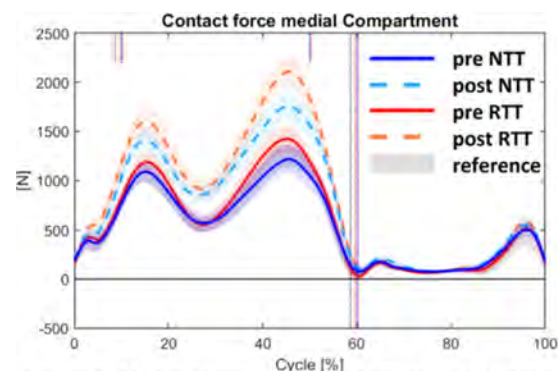


Figure 1: Medial contact force for both groups and the reference

## Discussion

In the case of pure frontal malalignment, successful surgical correction has been shown to result in a normalisation of the contact forces. However, in patients exhibiting reduced external rotation of the tibia, higher medial contact forces are observed following frontal correction of the leg axis. The literature does not currently provide any indication of the level at which an increase in contact force is associated with a higher probability of osteoarthritis. However, we highly recommend considering tibial torsion during frontal leg axis correction to avoid unphysiological knee joint loading postoperatively.

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## Postural stability during specific postural tasks in more and less experienced female figure skaters

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## Introduction

Postural stability is a key aspect of performance in figure skating, particularly during jump landings and the execution of complex coordination elements. The ability to effectively motor and postural control is essential for them not only for optimizing athletic performance but also for injury prevention. However, conventional postural stability tasks often have limitations related to insufficient demands and a lack of specificity for figure skating.

## Research Question

Is there a difference in postural stability during specific postural tasks between more and less experienced female figure skaters?

## Methods

The observed sample consisted of female competitors from figure skating clubs, divided into a more experienced group ( $n = 8$ , age 7 to 10 years) and a less experienced group ( $n = 12$ , age 12 to 15 years). Participants underwent measurements under static conditions through unilateral standing for 30 s (3 attempts per limb and task) in three specific positions: flamingo stance, forward lean and landing position. Postural stability was measured using a force plate (Kistler Instrumente, Winterthur, Switzerland). The output variables derived from centre of pressure (COP) movement include postural sway (standard deviation of COP movement) and COP velocity in the mediolateral (ML) and anteroposterior (AP) directions. Values from all attempts and both limbs were averaged for each subject. Statistical analysis was performed using Statistica (TIBCO software Inc., USA). For assessment of the effect of the task and group, ANOVA for repeated measures was used.

## Results

A significant task effect was observed for the parameters Sway AP ( $p < 0.001$ ) and Velocity AP ( $p < 0.001$ ). In both groups, significantly higher values (indicating poorer stability) for these variables were found during the forward lean task compared to the flamingo stance and the landing position task ( $p < 0.05$ ). For Sway ML, a significantly higher value was found in the more experienced group compared to the less experienced group the forward lean task ( $p = 0.022$ ).

## Discussion

Our results suggest that some forward lean postural task may better distinguish between more and less experienced figure skaters and could be suitable for postural stability testing in this specific group.

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## Predicting joint contact forces using a combination of kinematics, anthropometrics, and demographics with explainable artificial intelligence

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## Introduction

Advanced musculoskeletal (MSK) models, alongside clinical 3D gait analysis (3DGA), provide deeper insights into movement dynamics [1,2]. However, their need for high computational and temporal resources limits clinical utility [3]. Recent studies have shown promising results using machine learning to predict KCFs, but these approaches have primarily relied on kinematics [4–10]. While joint angles are essential, anthropometric and demographic factors, including age, body composition, and tibiofemoral alignment, they also significantly influence knee loading. In this study, we incorporate kinematics, anthropometric, and demographic factors into a machine learning (ML) approach to personalize predictions. We also use an explainability method to identify the factors that contribute most to predictions.

## Research Question

To what extent do anthropometric and demographic data improve prediction performance for medial, lateral, and total knee contact forces, and which features contribute most to the prediction?

## Methods

We used retrospective 3DGA data from 822 patients (male/female: 474/348; age:  $15.8 \pm 10.4$  years) exhibiting various frontal knee malalignments, yielding 12,381 trials. The dataset was randomly split into 70% training, 10% validation, and 20% test sets. We trained Multi-Layer Perceptrons (MLPs) to predict for each trial the medial, lateral, and total KCFs initially derived from musculoskeletal simulations in OpenSim. In a first attempt, an MLP was trained using only 11 kinematic time-series of the pelvis, hip, knee, ankle and subtalar joints as input. Subsequently, an MLP was trained with the kinematic input features and additional anthropometric and demographic features: age, bodyweight, bodyheight, sex, and tibiofemoral angle. As performance metric we utilized the Normalized (by each trial's range) Root Mean Squared Error (NRMSE). The SHapley Additive exPlanations (SHAP) [11] method was applied to the MLPs to evaluate the contribution of input features.

## Results

Including anthropometric and demographic features consistently improved predictions, yielding lower NRMSE across lateral (0.171 vs. 0.198), medial (0.081 vs. 0.098), and total (0.073 vs. 0.080) KCFs. Among kinematics, hip flexion (predominantly early stance) and ankle flexion (predominantly late stance) angles were the most important signals for the prediction (Figure 1). Among kinematics, hip flexion and ankle flexion angles were the most important signals for the prediction (Figure 1).

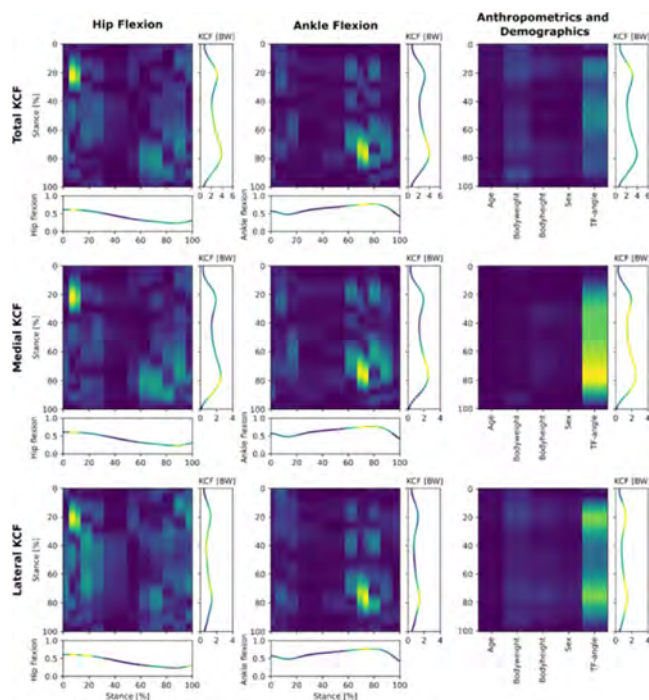


Figure 1: Heatmap of the mean importance for the two kinematic features (mean waveforms) with the highest importance and the anthropometric/demographic features. Warmer colors represent more important features.

## Discussion

Our results demonstrate that including subject-specific anthropometric and demographic data improves prediction performance. According to SHAP explanations, the tibiofemoral angle had the strongest influence on the ML model, followed by bodyweight, suggesting their direct biomechanical relevance. Gender and age play a minor role. We therefore recommend adding subject-specific information about bony geometry whenever possible to ML-based predictions of joint mechanics. Future research may focus on further patient populations, additional personalized data and advanced explainability techniques to validate these initial insights.

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## Isokinetic dynamometers and inverse dynamics provide different moments for the ankle and knee joints

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### Introduction

Accurate measurement of joint moments is of great importance for many applications in sports, rehabilitation, and research. Isokinetic dynamometer (IKD) devices and inverse dynamics approach (ID) are both widely used methods for quantifying joint moments. A vast number of studies using these two methods have provided basis on our understanding of muscle function in various motor tasks among healthy and clinical groups. For decades, IKD and ID are thought to provide equivalent moment values, but this assumption has never been verified.

### Research Question

Whether maximal joint moments of developed by antigravity muscle groups differ when measured using IKD and ID methods.

### Methods

Maximum isometric moments from nine subjects were measured at matched joint angles for ankle extensors (0°), knee extensors (40°), hip extensors (20°) and hip abductors (0°) using a commercially available IKD (Con-Trex) and a force plate integrated apparatus, which enabled ID computation (Fig 1.).

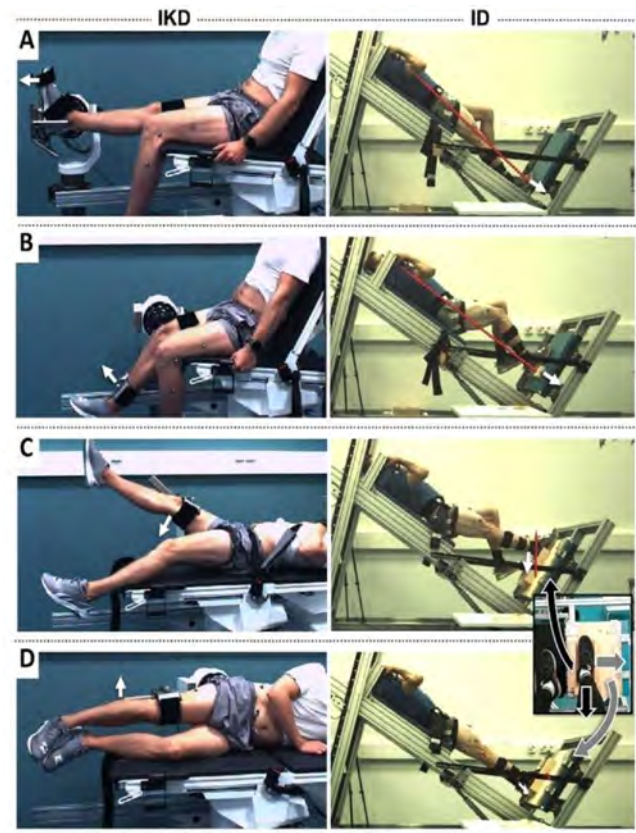


**Fig. 1.** Measurements of maximal moments in IKD and ID apparatus for the ankle extensors (A), knee extensors (B), hip extensors (C), and hip abductors (D).

### Results

A 44% ( $P=0.002$ ) lower ankle extensor moment was obtained in IKD than ID condition ( $1.81 \pm 0.58$  vs  $2.60 \pm 0.42$  Nm/kg), whereas the opposite – a 30% ( $P=0.029$ ) greater extensor moment in IKD – was measured at the knee ( $2.80 \pm 0.78$  vs  $2.16 \pm 0.81$  Nm/kg). No significant differences were found between the two methods for hip abductor and hip extensor moments.

### Discussion

The discrepancy in the ankle extensor moment between IKD and ID may be due to non-optimal force transmission from the pedal adapter to the rotational axis of the dynamometer. However, the explanation for the different knee extensor moments remains unclear, as EMG data did not indicate higher antagonist (hamstrings) muscle co-contractions during the multi-joint leg extension performed in ID.

It is important to consider the differences observed in the ankle and knee extensor moments, when interpreting results derived from IKD and ID measurements. For example, the current results provide a rationale for previous observations of why the ankle plantarflexor moment during walking can often exceed the maximal plantarflexor moment obtained from IKD (Beijersbergen et al., 2013; Brown et al., 2014). Similarly, the present findings can explain why much lower specific tension values (i.e., force per muscle cross-sectional area)

have been previously reported for ankle plantarflexors (11–15 N/cm<sup>2</sup>) (Fukunaga et al., 1996; Maganaris et al., 2001) compared to knee extensors (25–30 N/cm<sup>2</sup>) (Erskine et al., 2010; Reeves et al., 2004).

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## Differences in intersegmental foot and ankle motion between patients with varus and valgus ankle osteoarthritis

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## Introduction

Patients with severe ankle osteoarthritis (OA) not only experience significant pain but also face changes in their range of motion, leading to alterations in their gait. In cases of ankle OA, a significant portion present with varus ankle OA, and valgus ankle OA is also occasionally observed. Both types employ distinct compensatory mechanisms to sustain a plantigrade gait.

## Research Question

This study aimed to investigate how intersegmental foot and ankle motion differs in patients with varus and valgus ankle OA using multi-segment foot gait analysis.

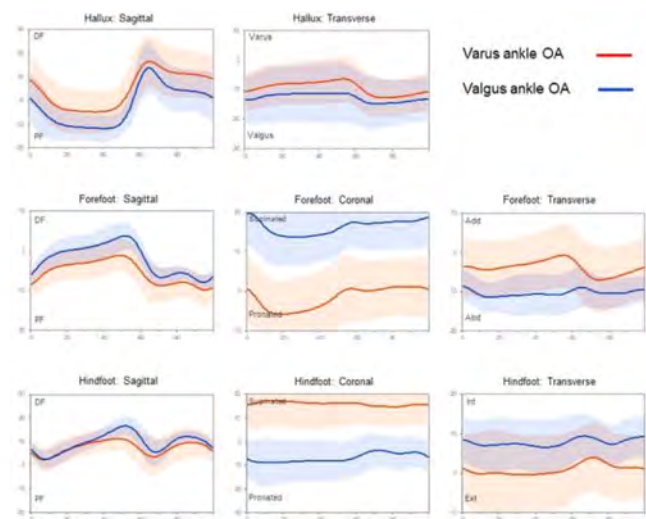
## Methods

The study included 55 patients with varus and 13 with valgus end-stage ankle OA. We classified the varus group as having a talar tilt greater than +4 degrees and the valgus group as having a talar tilt less than -4 degrees. The DuPont foot model was used to analyze intersegmental foot and ankle kinematics during gait.

## Results

Both groups showed displayed comparable temporospatial parameters. In the hallux segment, the varus group showed significant dorsiflexion in the terminal stance phase. In the forefoot segment, the varus group demonstrated significant plantar flexion and adduction during the stance phase and pronation throughout the gait cycle. In the hindfoot segment, the valgus group showed significant pronation throughout the gait cycle and dorsiflexion in the terminal stance phase (Figure 1). Additionally, there was no significant difference in the range of motion of the hindfoot between the varus and valgus groups in the coronal plane.

Figure 1. Comparison of the average kinematics of the distal segment relative to the proximal segment between varus and valgus ankle osteoarthritis groups.



## Discussion

In our study, since the varus and valgus groups were divided based on the talar tilt angle, it is intuitively expected that the position of the hindfoot in the coronal plane would differ significantly, and our study results support this assumption. However, given that the patients were in severe status requiring ankle fusion, despite the tilting of the talus in both groups, there was little difference in the range of motion in the coronal plane. In addition, as we hypothesized, the intersegmental motion of the midfoot (forefoot relative to the hindfoot) during the stance phase showed significant differences between the two groups across all planes. We believe this represents a compensatory movement to maintain a plantigrade in response to each deformity in the hindfoot. An interesting finding in this study was the difference observed between the two groups during the terminal stance phase. Specifically, in the toe-off phase, the varus group exhibited greater dorsiflexion of the hallux, while the valgus group showed more dorsiflexion of the hindfoot. We believe this to be related to the locking mechanism of the transverse tarsal joint when generating propulsion during toe-off.

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## Idiopathic toe walking diagnosed by dynamic foot gait analysis in an adult patient with bilateral heel pain

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## Introduction to clinical case

Idiopathic toe walking (ITW) describes the condition characterized by habitual toe-toe gait without an identifiable cause, usually diagnosed during childhood or adolescence. Patients typically show normal physical examinations except for limitations in passive ankle dorsiflexion, making it challenging to diagnose during static standing evaluation. This clinical case study highlights the importance of dynamic gait analysis in diagnosing ITW in adults. A 21-year-old male presented with bilateral heel pain persisting for several years. Despite more than six months of conservative management including rehabilitation, no significant improvement was observed. Physical examination showed neutral heel positioning in static stance. Ankle dorsiflexion was limited (right/left: knee extension 5°/0°; knee flexion 10°/10°), indicating a positive Silfverskiöld test.

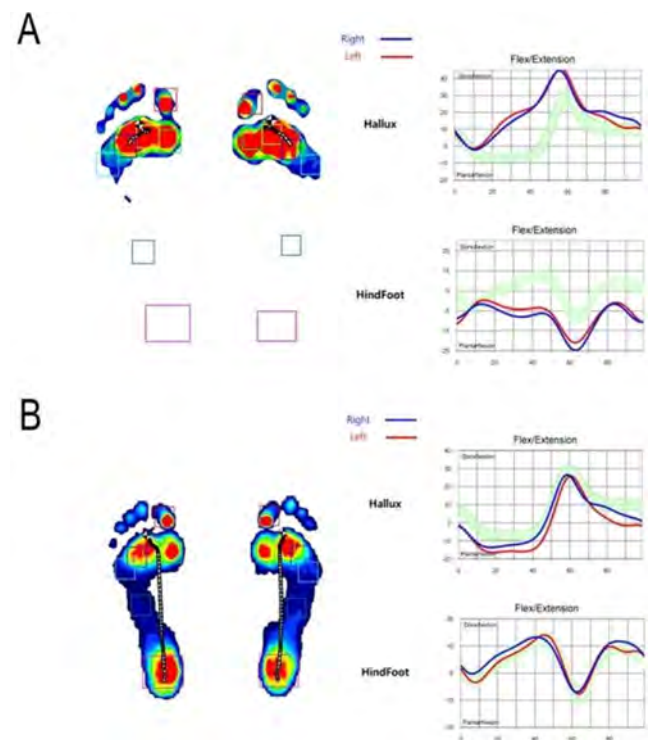
## Methods

Radiographic evaluation revealed no significant foot and ankle pathology. Ultrasonography of the Achilles tendon showed no abnormalities. Static pedobarography using HR Mat (Tekscan Inc., Norwood, MA, USA) demonstrated normal plantar pressure distribution. However, dynamic pedobarography during gait revealed a toe-walking pattern with peak pressures concentrated at the forefoot and contact area decreased at the hindfoot. Dynamic foot gait analysis using the DuPont multisegment foot model with 15 reflective markers (Motion Analysis Co., Rohnert Park, CA, USA) was performed, showing increased dorsiflexion in the hallux segment and decreased dorsiflexion in the hindfoot segment throughout the stance phase (Fig. 1A). Based on these dynamic evaluations, bilateral Achilles tendon lengthening was performed.

## Results

Two years postoperatively, physical examination revealed significantly improved ankle dorsiflexion (right/left: knee extension 10°/10°; knee flexion 15°/15°). Follow-up dynamic pedobarography demonstrated a balanced plantar pressure distribution and contact area similar to normal controls. Gait analysis also revealed normalization of gait patterns comparable to healthy individuals (Fig. 1B).

Fig. 1. Results of dynamic pedobarography and foot gait analysis (A) preoperatively and at (B) 2-year postoperative follow-up.



## Conclusion and clinical relevance

Static evaluations alone failed to identify ITW accurately. Three-dimensional foot gait analysis is an essential tool for diagnosing and managing ITW. This case suggests that gait analysis should be incorporated routinely in clinical assessments, especially for patients with persistent heel pain unresponsive to conservative treatment.

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## Impact of medial longitudinal arch configuration on foot and ankle motion in end-stage varus ankle osteoarthritis

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## Introduction

Varus talar tilt is commonly observed in patients with end-stage ankle osteoarthritis (OA) and has been associated with altered gait compared to those with neutral hindfoot alignment. However, in clinical practice, we have noted considerable variability in the configuration of the medial longitudinal arch (MLA) among such patients. We hypothesized that, despite sharing a similar degree of varus talar tilt, differences in arch morphology may result in distinct intersegmental foot and ankle motion patterns required to achieve plantigrade gait.

## Research Question

Does variation in MLA configuration influence intersegmental foot and ankle motion in patients with end-stage ankle OA presenting with varus talar tilt?

## Methods

This study included 39 consecutive patients with varus ankle OA who underwent tibiotalar arthrodesis. Based on the morphology of the MLA, patients were categorized into a low arch group ( $n=19$ ) and a high arch group ( $n=20$ ), using the lateral talo-first metatarsal angle and calcaneal pitch angle as classification criteria. Intersegmental foot and ankle kinematics during gait were assessed using the DuPont foot model.

## Results

There were no significant differences between the two groups in temporospatial gait parameters. In the hallux segment, intersegmental motion in both the sagittal and transverse planes did not differ significantly between groups. In the forefoot segment, no significant differences were observed in the coronal and transverse planes throughout the gait cycle. However, in the sagittal plane, the low arch group exhibited a significantly more dorsiflexed position from mid-stance to pre-swing phase ( $p=0.003$ ). In the hindfoot segment, no significant differences were found in the sagittal and transverse planes throughout the gait cycle, whereas the high arch group demonstrated a significantly more inverted position in the coronal plane across the entire gait cycle ( $p<0.001$ ) (Figure 1).

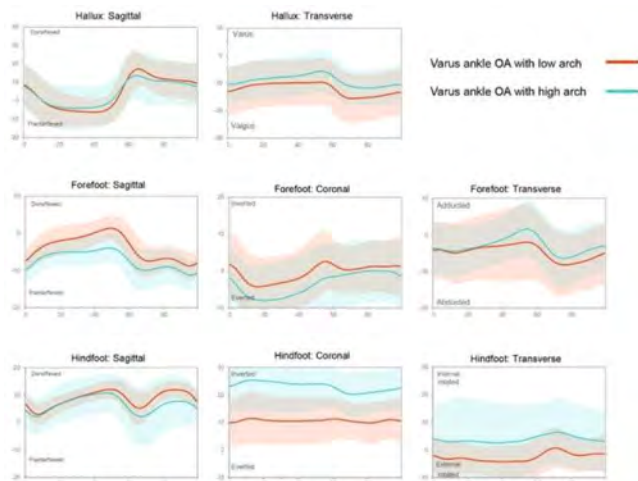
Figure 1. Comparison of the average kinematics of the distal segment relative to the proximal segment between low arch and high arch groups.

## Discussion

Although we initially hypothesized that both groups would exhibit similar kinematic patterns in the hindfoot coronal plane due to the presence of varus talar tilt, the high arch group showed a significantly more inverted position throughout the gait cycle. Given the classification based on MLA morphology, we expected compensatory motion in the coronal plane of the midfoot—particularly in the forefoot relative to hindfoot segment—in the high arch group to maintain plantigrade gait. However, such compensatory motion was minimal, and instead, a more plantarflexed position was observed in the sagittal plane from mid-stance to pre-swing phases. Taken together, the configuration of the MLA, which plays a key role in shock absorption during gait, contributes to variations in kinematic patterns. These results provide foundational evidence for preoperative planning and postoperative alignment strategies in the surgical management of varus ankle OA.

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**Fine-tuning from a clinical gait analysis dataset to predict the effect of botulinium toxin injection during gait**

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**Introduction**

While machine learning models for predicting surgical outcomes in clinical gait analysis are limited [1], only one study forecasts post-botulinum toxin gait trajectories, focusing solely on knee and ankle sagittal kinematics [2]. To date, no study has predicted the full 3D joint kinematics (pelvis, hip, knee, ankle and foot angles) following botulinum toxin injection (BTI). Building deep neural networks from scratch is resource-intensive, particularly when training data is scarce and augmentation is impractical. However, in many real-world applications, transfer learning using pre-trained models from related domains offers a significant advantage. The aim is to pre-train deep learning models from the gait kinematic before BTI and fine-tune them to predict the gait kinematic data after BTI.

**Research Question**

Can a fine-tuned model predict the effect of botulinum toxin injection during gait?

**Methods**

This retrospective study analyzed existing 3D clinical gait analysis (CGA) data acquired at Brest University Hospital. Two distinct datasets were constructed. The first dataset included longitudinal CGA data from 42 children ( $9.97 \pm 4.57$  years) with cerebral palsy (CP), capturing two sessions with a mean inter-session interval of

21.33 months, regardless of any intervening treatments. To ensure consistent analysis, the minimum number of gait cycles available across both sessions for each child was used, resulting in a total of 1 466 gait cycles. The second dataset focused on 17 children ( $9.88 \pm 3.89$  years) with CP who underwent BTI. This dataset included 20 sessions, representing CGA performed within 12 months before and after BTI, resulting in 295 gait cycles. This study developed and validated a prediction tool using two deep learning models: Bi-Long Short-Term Memory (Bi-LSTM) networks and 1D Convolutional Neural Networks (1D CNN). Initially, both models were trained and evaluated on the two distinct datasets using leave-one-patient-out cross validation. Subsequently, the models were fine-tuned by transferring weights learned from the first dataset to the second. The performance of the model was reported as root mean squared errors between the actual and predicted joint angles.

## Results

Applying fine-tuning to both models results in a significant reduction of the average RMSE when predicting the complete 3D joint kinematics following BTI (table 1). While fine-tuning the Bi-LSTM led to RMSE improvements for nearly all patients, it negatively impacted three. However, the 1D CNN achieved RMSE reductions for every patient.

|                     |     | Bi-LSTM      |              | 1D CNN       |              |
|---------------------|-----|--------------|--------------|--------------|--------------|
|                     |     | From scratch | Fine-tuning  | From scratch | Fine-tuning  |
| Average RMSE        |     | 9.50 (0.24)  | 8.81* (0.07) | 11.55 (0.5)  | 8.65* (0.13) |
| RMSE by children ID | 13  | 9.56         | <b>9.18</b>  | 11.85        | <b>8.06</b>  |
|                     | 135 | 7.44         | <b>6.89</b>  | 8.82         | <b>6.44</b>  |
|                     | 167 | 7.90         | <b>7.62</b>  | 10.02        | <b>7.61</b>  |
|                     | 235 | <b>11.72</b> | 12.09        | 15.53        | <b>9.24</b>  |
|                     | 343 | 8.81         | <b>7.59</b>  | 9.00         | <b>8.21</b>  |
|                     | 389 | <b>7.29</b>  | 7.45         | 10.29        | <b>7.20</b>  |
|                     | 428 | 11.75        | <b>8.98</b>  | 11.99        | <b>9.90</b>  |
|                     | 442 | 11.06        | <b>9.39</b>  | 14.20        | <b>9.98</b>  |
|                     | 493 | 8.08         | <b>7.55</b>  | 11.43        | <b>6.92</b>  |
|                     | 506 | 9.29         | <b>8.36</b>  | 10.87        | <b>8.99</b>  |
|                     | 509 | 9.26         | <b>9.08</b>  | 9.78         | <b>9.02</b>  |
|                     | 634 | <b>9.35</b>  | 9.53         | 12.03        | <b>9.07</b>  |
|                     | 661 | 7.66         | <b>7.22</b>  | 9.74         | <b>8.07</b>  |
|                     | 828 | 11.43        | <b>11.16</b> | 13.45        | <b>10.67</b> |
|                     | 836 | 11.33        | <b>10.07</b> | 13.77        | <b>10.97</b> |
|                     | 929 | 12.89        | <b>11.23</b> | 14.51        | <b>10.74</b> |
|                     | 953 | 7.30         | <b>6.44</b>  | 9.01         | <b>5.98</b>  |

RMSE by children for Bi-LSTM et 1D CNN. \* $p < 0.05$ .

## Discussion

This study demonstrated the feasibility of using deep learning models, specifically Bi-LSTM and 1D CNN, to predict the effects of BTI on 3D gait kinematics in children with CP. The results indicate that both fine-tuned models can effectively capture the complex temporal dependencies within gait data, allowing for accurate prediction of post-BTI kinematic changes. Additionally, incorporating other clinical variables, such as muscle strength and spasticity measures, may further enhance prediction accuracy.

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## Dystonia: usefulness of 3D gait analysis and dynamic surface electromyography in discriminating dystonic actions from compensatory mechanism

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## Introduction to clinical case

Dystonia is a movement disorder characterized by involuntary contractions of one or more muscles causing abnormal, often repetitive, movements, postures, or both [1]. In this context, Botulinum Neurotoxin (BoNT) represents an effective treatment option [2], but it is entirely symptomatic and muscle selection for injection is particularly challenging, especially in limb dystonia. In fact, not all hyperactive muscles are necessarily dystonic: some muscles may present hyperactivity as a compensatory mechanism to counteract dystonic movements [3]. Correct Identification of dystonic muscles is therefore critical for successful outcomes in reducing pathological movements or postures without compromising compensatory muscles and their physiological function [3,4].

## Methods

A 35-year-old Caucasian female with a painful dystonic foot posture in dorsiflexion and supination and a marked big toe extension during walking was evaluated by 3D gait analysis (GA) and dynamic surface electromyography (sEMG) of the lower limb [3]. Results confirmed the clinical evidence of Tibialis Anterior (TA) and Extensor Hallucis Longus (EHL) hyperactivation but also showed an abnormal recruitment of Gastrocnemius Medialis (GAM), without plantiflexion, in the Swing-phase (Sw). In the same evaluation session sEMG was registered also during the execution of some segmentary movements including foot eversion (= dorsiflexion + pronation) [5] to investigate muscles behavior reaching the opposite foot posture to the dystonic one.

## Results

Since GAM showed abnormal recruitment both during gait, in Sw, and during eversion at extended knee, with progressive out-of-phase activation and corresponding loss of the eversion and foot sliding in dystonic posture, it was suspected to be the real dystonic muscle while TA and EHL hyperactivation a compensatory

mechanism. By a selective peripheral motor nerve block [6], GAM was temporarily switched off. The patient recovered an almost completely plantigrade support on her right foot during the Stance-phase with disappearance of the big toe extension and, at the request of active eversion, she was able to reach and hold the foot position also at extended knee, without fatigue (Fig. 1).



Fig. 1. Sagittal and frontal view of the right foot posture at the end of MidStance. Comparison between pre(A) and post(B) motor nerve block for GAM branch.

### Conclusion and clinical relevance

Based on the results, GA and dynamic sEMG during personalized motor tasks (as the voluntary attempt to assume the opposite posture to the dystonic one) demonstrated to be helpful in our clinical case: in developing and confirming the suspicion that GAM was the possible dystonic muscle while TA and EHL hyperactivation the compensatory mechanism to counteract GAM dystonic action. Although this is only a single case it could inspire a possible way to proceed in dystonic patients assessment for discriminating hyperactive compensatory muscles, whose contractions are usually clinically observable, being responsible for repetitive movements or abnormal postures, from dystonic muscles, whose actions could be covered and clinically not detectable.

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### Impact of Obesity on Gait Mechanics in Chronic Low Back Pain: Insights from Spatiotemporal Parameters and Vertical Ground Reaction Force

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### Introduction

Obesity and chronic low back pain (CLBP) significantly affect gait mechanics, affecting spatiotemporal parameters and vertical ground reaction force (vGRF). Individuals with obesity and CLBP often exhibit slower walking speed, shorter stride length, and altered cadence, contributing to instability and discomfort. Increased vGRF due to excess body weight and pain-related compensations can lead to abnormal loading patterns, further affecting mobility. Understanding these biomechanical changes is essential for developing targeted rehabilitation strategies to enhance gait efficiency and functional outcomes. This study investigates how obesity influences spatiotemporal gait parameters and vGRF in individuals with CLBP.

### Research Question

How does obesity influence spatiotemporal gait parameters and vertical ground reaction force in individuals with chronic low back pain?

### Methods

Twenty individuals with CLBP were recruited and categorized into two groups based on body mass index (BMI): CLBP with obesity class I ( $n = 10$ ) and CLBP with normal BMI ( $n = 10$ ). Inclusion criteria required participants to have a pain intensity score of less than 3 on the Numeric Pain Rating Scale (NPRS) and a minimal disability level on the Oswestry Disability Index (ODI) to minimize pain as a confounding factor. Each participant completed 10 walking trials at their natural walking pace on a 10-meter walkway. Spatiotemporal gait parameters and vGRF data were collected using a motion capture system (VICON) and force plates (AMTI), with calculations performed using VICON Polygon software. Non-dimensionalization was applied to spatiotemporal parameters to eliminate the effects of leg length and gravity. Normality was assessed using the Shapiro-Wilk test, and group comparisons were conducted using independent t-tests. Statistical Parametric Mapping (SPM) was used to analyze vGRF waveforms across the gait cycle.

### Results

The results showed significant differences in non-dimensional spatiotemporal parameters: Step time was significantly higher in the CLBP obesity group ( $p = 0.012$ ). Cadence was significantly lower in the CLBP obesity group ( $p = 0.010$ ). Walking speed was significantly lower compared to the normal BMI group ( $p = 0.030$ ). Step length showed no significant difference ( $p = 0.323$ ). Kinetic analysis using SPM-based statistical comparisons revealed no significant differences in vGRF throughout the gait cycle.

### Discussion

These findings suggest that obesity in CLBP individuals affects spatiotemporal gait characteristics, with reduced walking speed,

longer step time, and lower cadence. However, the vertical loading forces remain comparable between groups, indicating that the gait adaptations in CLBP individuals with obesity are primarily spatio-temporal kinetic parameters rather than in ground reaction force distribution. This could be attributed to the neuromuscular compensations maintaining vertical loading despite altered gait patterns. Future research should explore muscle activation patterns and joint-level kinetics to further understand the biomechanical adaptations in this population. Clinically, tailored gait interventions should focus on improving walking efficiency rather than reducing ground reaction forces.

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## Effects of multi-targeted electrical stimulation on gait and spasticity in an East Asian individual with multiple sclerosis

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## Introduction

Spasticity is highly prevalent among individuals with multiple sclerosis (MS), and its severity can worsen with progression of disease or during episodic relapses. This results in gait disturbances, limiting independence and decreasing overall quality of life. (Hugos, Cameron 2020) Transcutaneous electrical nerve stimulation (TENS) has been recognized as an effective adjunct modality for reducing spasticity. (Marcolino 2018)

The Exopulse Mollii Suit (EMS) is a non-invasive, multi-channel full-body suit with 58 electrodes for electrical stimulation (ES). It reduces spasticity via the principle of reciprocal inhibition, achieved through low-frequency, low-intensity stimulation of the antagonist muscle.

## Research Question

What are the immediate effects of a single 60-minute session of EMS on gait and spasticity in an East Asian individual with MS?

## Methods

A 41-year old East Asian male with spastic paraparesis and ataxia was diagnosed with primary progressive MS in 2019. His Expanded Disability Status Scale 6.5. He wore the EMS for 60 minutes, with higher frequencies of stimulation parameters applied on muscles with greater spasticity, specifically gluteus medius, gluteus minimus, hamstrings, tibialis anterior and abdominal muscles. The primary outcome measure was observational gait analysis using the Gait Assessment and Intervention Tool (G.A.I.T.). Spatiotemporal parameters were obtained using a markerless, vision-based motion capture system employing the OpenPose BODY25 pose estimation model (Cao Z, 2021). Secondly, the effects on spasticity and ataxia were assessed using the Modified Tardieu Scale (MTS) and the Scale for the Assessment and Rating of Ataxia (SARA), respectively.

## Results

The patient ambulated independently with a walking frame. The main observable gait abnormalities included excessive trunk flexion, posterior pelvic tilt throughout stance and swing phases. During stance, there was bilateral reduced hip extension, knee hyperextension, increased ankle inversion and plantar flexion, with reduced knee flexion and ankle dorsiflexion during swing phase. Post-intervention, the G.A.I.T. score improved from 42 to 31. Observed improvements included reduced trunk flexion, correction of knee hyperextension, improved foot placement during the stance phase, and increased knee flexion during the swing phase. Spatiotemporal parameters improved in double and single support time, step time, and stride length. Walking speed trended upward but did not meet the minimal clinically important difference (Bohannon et al., 2014). Gait variability also improved. There was a trend toward reduced spasticity in bilateral hip adductors, hamstrings, and gastrocnemius muscles, although changes in the MTS score did not exceed the smallest detectable change (Naghdi et al., 2016). Additionally, SARA improved by 1.5 points.

## Discussion

EMS effectively improved gait pattern and reduced spasticity in the hip adductors, hamstrings, and gastrocnemius muscles in an MS individual. However, as the results were based on a single 60-minute session, a dose-response relationship between ES, gait, and spasticity could not be determined. Future studies could examine how varying EMS settings impact gait and spasticity outcomes in MS.

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## Functional motor assessments in children with Autism Spectrum Disorder: A test-retest study of neuromuscular power, strength, and balance.

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## Introduction

Children with Autism Spectrum Disorder (ASD) often exhibit motor skill deficits, such as impaired postural control, poor coordination, and reduced muscle strength. Studies show that 50–88% of children with ASD experience significant motor impairments on standardized assessments, highlighting the need to evaluate motor performance within this population [1,2]. However, administering these motor tests to children with ASD can be challenging due to attention issues, sensory sensitivities, and behavioural variability, which may lead to inconsistent performance [3]. This variability raises concerns about the reliability of repeated assessments in children with ASD. Despite this, few studies have rigorously examined the reproducibility of objective biomechanical measurements in this group.

## Research Question

Are biomechanical variables obtained from assessments of postural control, functional strength and coordination, and neuromuscular power reproducible and reliable in children with ASD?

## Methods

A test-retest design was implemented with 14 children with ASD (age:  $8.57 \pm 2.95$  years, mass:  $37.3 \pm 19.2$  kg). Participants met DSM-5 diagnostic criteria for ASD (support level 1 or 2); those unable to complete testing were excluded. Data were recorded using two synchronized force plates (ForceDecks), capturing vertical ground reaction force and centre of pressure (200 Hz). Data collection utilized the ForceDecks iOS app (V1.8.9) on an iPad (Apple, CA, USA) and exported for analysis. Force plates were zeroed before each test, with the participant's body mass measured in a relaxed standing position. Testing included postural sway (three trials, 30-s, eyes-open/closed), sit-to-stand (STS) and countermovement jump (CMJ) (five trials, ~10 s apart), repeated after a 15-day interval. Intraclass Correlation Coefficients (ICC), Standard Error of Measurement (SEM), and Minimal Detectable Change (MDC) were calculated from all trials, and both sessions were compared to assess reliability.

## Results

Reliability varied across motor tasks and assessment sessions. In the CMJ, concentric mean force showed high reliability during testing (ICC=0.71, %MDC=15.8%) but decreased in the retest (ICC=0.19, %MDC=34.3%). Eccentric mean force followed a similar trend, with high reliability initially (ICC=0.88, %MDC=48.8%) but a notable drop at retest (ICC=0.22, %MDC=137.2%). In STS tasks, reliability ranged from low to moderate (ICC<0.5, %MDC>100%). For postural sway, mean CoP velocity with eyes open had high reliability during testing (ICC~0.78, %MDC~48%) but declined significantly at retest (ICC~0.20, %MDC>200%). Other measures consistently showed lower reliability across sessions.

| FUNCTIONAL TEST |        | VARIABLES                                |  | ICC 2.5 | SEM   | MDC    | Alpha  | OP     | %MDC   | %MDC   |
|-----------------|--------|------------------------------------------|--|---------|-------|--------|--------|--------|--------|--------|
| Sit to Stand    | Test   | Mean sitting force (N/kg)                |  | 0.52    | 1.08  | 2.98   | 7.31   | 1.28   | 15.24  | 41.68  |
|                 |        | Mean standing force (N/kg)               |  | 0.15    | 1.22  | 5.20   | 7.53   | 1.25   | 14.87  | 41.21  |
|                 |        | Peak sitting force (N/kg)                |  | 0.08    | 0.84  | 3.39   | 11.60  | 0.92   | 7.30   | 20.12  |
|                 |        | Peak standing force (N/kg)               |  | 0.14    | 1.05  | 4.74   | 12.35  | 1.43   | 20.81  | 50.11  |
|                 |        | Time to sit (s)                          |  | 0.46    | 0.47  | 2.29   | 1.22   | 0.53   | 39.32  | 105.07 |
|                 |        | Time to stand (s)                        |  | 0.44    | 0.47  | 2.29   | 1.20   | 0.53   | 40.31  | 107.08 |
| Sit to Stand    | Retest | Mean sitting force (N/kg)                |  | 0.20    | 1.01  | 2.83   | 7.39   | 0.99   | 14.36  | 39.88  |
|                 |        | Mean standing force (N/kg)               |  | 0.16    | 1.26  | 5.24   | 7.94   | 1.38   | 15.34  | 44.87  |
|                 |        | Peak sitting force (N/kg)                |  | 0.14    | 0.84  | 3.39   | 11.60  | 1.07   | 8.31   | 22.18  |
|                 |        | Peak standing force (N/kg)               |  | 0.18    | 1.17  | 4.54   | 12.08  | 1.17   | 9.81   | 26.31  |
|                 |        | Time to sit (s)                          |  | 0.43    | 0.47  | 2.28   | 1.27   | 0.59   | 40.07  | 108.76 |
|                 |        | Time to stand (s)                        |  | 0.42    | 0.45  | 2.25   | 1.20   | 0.53   | 39.76  | 104.05 |
| Quiet Stand     | Test   | CoP Range: Anterior-Posterior (mm)       |  | 0.14    | 19.22 | 53.00  | 12.43  | 19.42  | 37.18  | 103.01 |
|                 |        | CoP Range: Medial-Lateral (mm)           |  | 0.27    | 6.92  | 19.58  | 14.88  | 12.12  | 40.30  | 128.92 |
|                 |        | Mean Velocity: Anterior-Posterior (mm/s) |  | 0.80    | 3.70  | 10.25  | 23.83  | 23.73  | 16.91  | 44.98  |
|                 |        | Mean Velocity: Medial-Lateral (mm/s)     |  | 0.77    | 3.38  | 9.82   | 7.81   | 9.44   | 17.64  | 48.90  |
|                 |        | CoP Range: Anterior-Posterior (mm)       |  | 0.12    | 18.22 | 52.91  | 49.34  | 23.83  | 146.58 | 403.38 |
|                 |        | CoP Range: Medial-Lateral (mm)           |  | 0.20    | 28.78 | 82.82  | 15.26  | 19.80  | 189.58 | 542.69 |
| Quiet Stand     | Retest | CoP Range: Anterior-Posterior (mm)       |  | 0.20    | 23.89 | 71.22  | 23.81  | 23.11  | 109.68 | 316.11 |
|                 |        | CoP Range: Medial-Lateral (mm)           |  | 0.21    | 8.37  | 23.73  | 9.21   | 14.22  | 36.97  | 103.98 |
|                 |        | CoP Range: Anterior-Posterior (mm)       |  | 0.40    | 23.08 | 62.74  | 49.14  | 29.74  | 161.58 | 438.27 |
|                 |        | CoP Range: Medial-Lateral (mm)           |  | 0.40    | 7.75  | 21.47  | 14.18  | 7.98   | 34.38  | 111.37 |
|                 |        | Mean Velocity: Anterior-Posterior (mm/s) |  | 0.28    | 19.82 | 55.21  | 23.16  | 7.98   | 144.34 | 260.58 |
|                 |        | Mean Velocity: Medial-Lateral (mm/s)     |  | 0.14    | 3.89  | 10.66  | 6.70   | 2.83   | 147.88 | 368.26 |
| Quiet Stand     | Test   | CoP Range: Anterior-Posterior (mm)       |  | 0.30    | 22.36 | 61.62  | 49.89  | 121.57 | 16.44  | 72.90  |
|                 |        | CoP Range: Medial-Lateral (mm)           |  | 0.32    | 8.20  | 23.97  | 10.14  | 40.38  | 27.22  | 72.90  |
|                 |        | Mean Velocity: Anterior-Posterior (mm/s) |  | 0.32    | 20.41 | 53.88  | 10.21  | 40.32  | 44.81  | 109.21 |
|                 |        | Mean Velocity: Medial-Lateral (mm/s)     |  | 0.21    | 10.44 | 29.00  | 11.25  | 12.92  | 36.34  | 103.01 |
|                 |        | Concentric Mean Force (N/kg)             |  | 0.75    | 1.38  | 3.27   | 10.84  | 2.19   | 1.72   | 13.84  |
|                 |        | Eccentric Mean Force (N/kg)              |  | 0.62    | 2.01  | 5.62   | 13.91  | 2.89   | 14.17  | 40.39  |
| CMJ             | Test   | Concentric Mean Velocity (m/s)           |  | 0.50    | 0.11  | 0.30   | 0.86   | 0.16   | 11.33  | 13.41  |
|                 |        | Eccentric Mean Velocity (m/s)            |  | 0.88    | 0.78  | 2.00   | 3.93   | 1.97   | 17.58  | 48.37  |
|                 |        | Concentric Mean Power (W/kg)             |  | 0.81    | 10.14 | 284.32 | 347.14 | 1.91   | 25.38  | 70.38  |
|                 |        | Eccentric Mean Power (W/kg)              |  | 0.19    | 2.01  | 5.40   | 15.89  | 2.16   | 12.36  | 34.18  |
|                 |        | Concentric Mean Velocity (m/s)           |  | 0.52    | 1.09  | 3.00   | 11.29  | 2.16   | 14.17  | 40.39  |
|                 |        | Eccentric Mean Velocity (m/s)            |  | 0.27    | 0.14  | 0.40   | 0.92   | 0.18   | 11.33  | 13.41  |
| CMJ             | Retest | Concentric Mean Velocity (m/s)           |  | 0.32    | 1.79  | 4.76   | 10.14  | 1.91   | 25.38  | 70.38  |
|                 |        | Eccentric Mean Velocity (m/s)            |  | 0.19    | 0.20  | 0.50   | 1.14   | 0.17   | 12.36  | 34.18  |
|                 |        | Concentric Mean Power (W/kg)             |  | 0.19    | 10.14 | 284.32 | 347.14 | 1.91   | 25.38  | 70.38  |
|                 |        | Eccentric Mean Power (W/kg)              |  | 0.19    | 2.01  | 5.40   | 15.89  | 2.16   | 12.36  | 34.18  |
|                 |        | Concentric Mean Velocity (m/s)           |  | 0.52    | 1.09  | 3.00   | 11.29  | 2.16   | 14.17  | 40.39  |
|                 |        | Eccentric Mean Velocity (m/s)            |  | 0.27    | 0.14  | 0.40   | 0.92   | 0.18   | 11.33  | 13.41  |

Figure 1. Test–retest reliability of biomechanical variables sit-to-stand, quiet-standing, and countermovement jump in children with ASD.

## Discussion

The reliability of biomechanical assessments in children with ASD varied across tasks and sessions. CMJ variables and mean CoP velocity (eyes-open) initially showed high reliability, though it declined at retest and may be influenced by behavioural variability. Sit-to-stand measures showed consistently lower reliability. These tests pose practical challenges for clinicians due to attentional and sensory fluctuations. Adapted protocols, including familiarization and motivational strategies, may help improve consistency and support their clinical applicability in this population.

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## Pain, fear, and movement: Cognitive-behavioural connections in patellofemoral pain

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## Introduction

Pain and movement play essential roles, significantly impacting attitudes and perceptions in an evident and observable way [1]. Cognitive-behavioural factors, such as fear and confidence, may be related to the onset and development of patellofemoral pain (PFP) and are considered important in assessing musculoskeletal complaints. However, given the complexity of PFP, it is crucial to understand whether fear-related behaviour is linked to other factors associated with PFP development.

## Research Question

Is there an association between kinesiophobia and functional domains, movement deviation profile (MDPmean), cognitive-behavioural factors, and pain intensity in patients with PFP?

## Methods

The study included 66 participants with PFP (52 females) age:  $25.56 \pm 5.91$  years; height:  $1.66 \pm 0.08$  m; mass:  $64.39 \pm 11.45$  kg). Scores of Tampa scale kinesiophobia (TAMPA), pain self-efficacy scale (PSEQ-10), chronic pain self-efficacy (CPSE), Catastrophizing Scale, numerical pain rating scale (NPRS) and anterior knee pain score (AKPS) were used to analyze cognitive and function scores. MDPmean was calculated from the trunk, pelvis and lower limb to provide a comprehensive summary of the kinematic data obtained during the lateral step down (SDL) [3]. General linear models were used to analyze the association between the kinesiophobia score and the other factors with a significance level of 0.05.

## Results

The sample characteristics showed that the kinesiophobia score was  $33.5 \pm 7.8$ ; in the cognitive-behavioural factors, the PSEQ-10 was  $35.2 \pm 6.8$ , the CPSE was  $12.5 \pm 3.4$ , and Catastrophizing was  $14.7 \pm 4.2$ , while pain registered an average of  $5.3 \pm 2.0$ , the AKPS was  $82.4 \pm 9.3$  and the MDPmean was  $38.7 \pm 6.5$ . A significant association (Figure 1) was found between the following variables:

kinesiophobia and PSEQ-10 ( $\beta = -0.38$ ,  $p = 0.007$ ), NPRS ( $\beta = 0.89$ ,  $p = 0.039$ ), AKPS ( $\beta = 0.30$ ,  $p = 0.044$ ) and MDPmean ( $\beta = 0.33$ ,  $p = 0.013$ ). In the stepwise regression model, only the PSEQ-10 retained statistical significance ( $\beta = -0.31$ ,  $p = 0.002$ ). Nevertheless, the MDPmean may still be considered relevant, despite not achieving statistical significance ( $\beta = 0.24$ ,  $p = 0.056$ ).

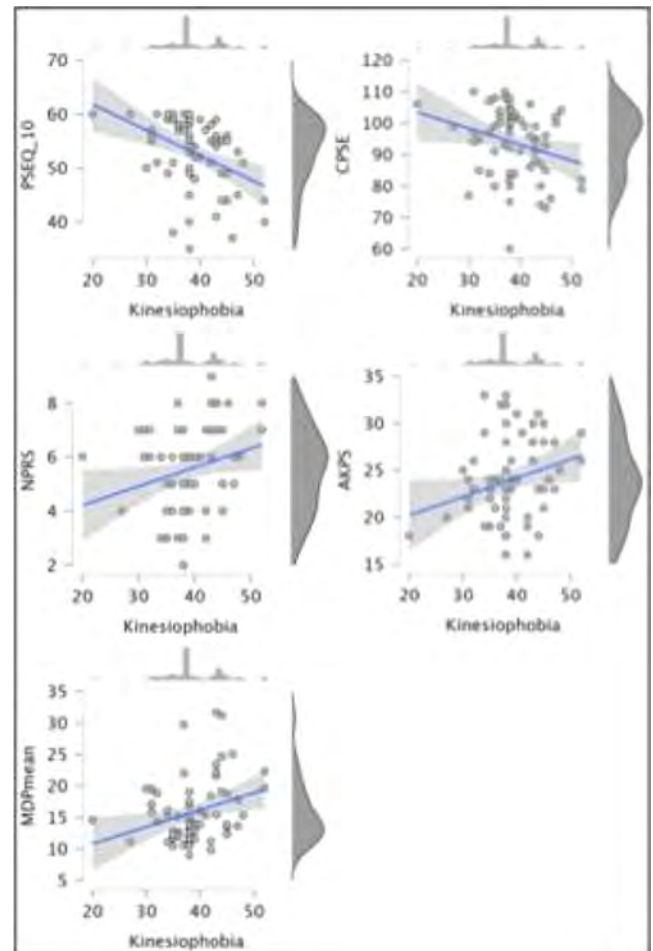


Figure 1: Correlation between kinesiophobia with functional domains, cognitive-behavioural factors, and pain intensity. The regression line (blue) shows a 95% confidence interval for the PSEQ-10, CPSE, NPRS, AKPS and the MDPmean.

## Discussion

Our findings emphasize the multifactorial nature of kinesiophobia in PFP, highlighting the individual associations. A 0.38-point reduction in kinesiophobia leads to a 1-point increase in PSEQ-10; regarding NPRS, a 1-point increase results in a 0.89-point increase in kinesiophobia; in terms of functional capacity, a 1-point increase in AKPS score leads to a 0.30-point increase in kinesiophobia; and for movement deviation (MDPmean), a 1-degree increase leads to a 0.33-point increase. Notably, when all pain-related factors were considered simultaneously, self-efficacy (PSEQ-10) and MDPmean emerged as the predominant factor. This suggests self-efficacy may encompass or be influenced by pain intensity, functional ability, and movement confidence [4], supporting the idea of an interconnected cycle where cognitive beliefs, physical experiences, and movement mutually shape fear-avoidance behaviours in individuals with PFP.

Therapeutic strategies addressing this interrelationship, especially those enhancing self-efficacy related to movement experience, could be crucial for managing PFP effectively.

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<https://doi.org/10.1016/j.gaitpost.2025.07.150>

## Neural network analysis reveals no kinematic effects of knee braces in patellofemoral pain during step-down tasks

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## Introduction

Patellofemoral pain (PFP) is a multifactorial musculoskeletal condition often associated with altered movement during functional tasks, particularly those involving increased joint loading, such as the lateral step-down (LSD) [1]. Among the contributing factors, biomechanical impairments are consistently reported as a key element in the exacerbation of pain symptoms. A commonly observed characteristic in individuals with PFP is patellar malalignment, in which the patella exhibits excessive lateral displacement within the femoral trochlear groove, potentially resulting in nociceptive stimulation [2]. Knee braces have been proposed as a conservative intervention that alleviates kinesiphobia and promotes more optimal patellofemoral joint alignment [3,4]. Despite some evidence supporting its clinical benefits, the extent to which different brace designs influence movement remains unclear.

## Research Question

Does wearing knee braces in patellofemoral pain reduce movement deviation during the LSD?

## Methods

An observational cross-sectional study evaluated 66 individuals with PFP (52 females) age:  $25.56 \pm 5.91$  years; height:  $1.66 \pm 0.08$  m; mass:  $64.39 \pm 11.45$  kg), selected based on a minimum pain score of 3/10 on the Numerical Pain Rating Scale (NPRS), across three separate sessions on the same day: one with no brace and two with different brace designs (closed and open-patella), applied in randomized order. Three-dimensional kinematic data were collected using a motion capture system with the Plug-in Gait model and Nexus 2.x analysis software. The MDPmean [5] was calculated based on angular data from the frontal, sagittal, and transverse plane angles of the trunk, pelvis, and hip, sagittal and frontal plane angles of the knee, ankle dorsiflexion, and foot progression angle. The Friedman test used a significance level set at  $p < 0.05$ .

## Results

No significant differences between conditions  $X^2(=1.121, p=0.571)$  were observed in the MDPmean, as expressed by the median and interquartile range (no brace:  $14.43^\circ$  [ $12.41^\circ$ – $18.63^\circ$ ]; closed-patella:  $14.91^\circ$  [ $12.75^\circ$ – $17.97^\circ$ ] and open-patella:  $14.29^\circ$  [ $12.75^\circ$ – $16.30^\circ$ ]).

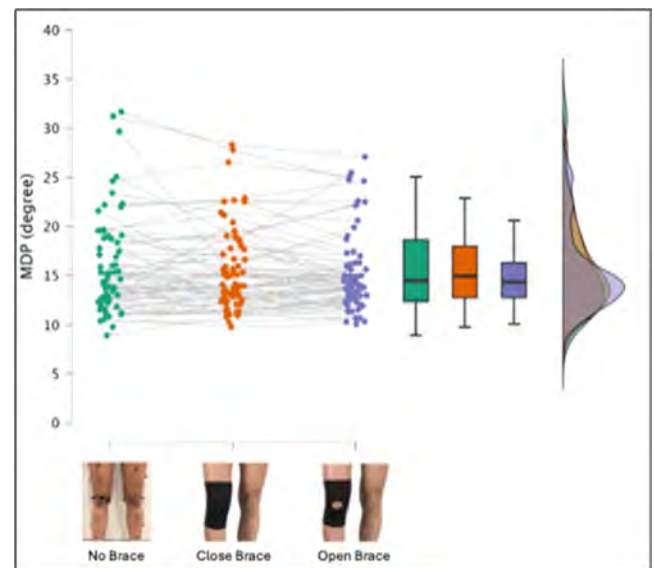


Figure 1. Raincloud plot showing MDPmean (degrees) across three knee brace conditions during the lateral step-down task. Dots represent individual data points connected across conditions. Box plots display median and interquartile range; violin plots illustrate data distribution.

## Discussion

These results suggest that knee braces did not alter movement during the LSD task in individuals with PFP, as indicated by the immediate measures obtained. Despite theoretical expectations that a closed-patella brace might increase patellofemoral contact and that an open-patella brace could improve patellar maltracking, no immediate overall kinematic differences were observed. Although bracing may reduce pain and enhance perceived stability, its effects result from neurosensory mechanisms rather than biomechanical changes [3,4]. However, local biomechanical effects should not be

overlooked, particularly in the coronal and transverse planes during eccentric step descent. [6]. This distinction is clinically relevant, as the immediate reduction of pain may help patients tolerate functional tasks better and engage more actively. Therefore, knee braces may serve as valuable adjuncts in symptom management without necessarily altering overall movement.

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## The influence of spinal posterior instrumentation and fusion on rib cage deformity in adolescent idiopathic scoliosis

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## Introduction

Adolescent idiopathic scoliosis (AIS) is a 3D spinal deformity that affects the chest wall through the costo-vertebral joints and leads to rib cage deformity that significantly impacts self-image in patients with AIS [1,2]. The rib cage deformity can be defined as an increased rib hump (type IIa), an increased spinal penetration into the chest volume (type IIb), or both (type IIc) [3]. In cases of severe scoliosis, spinal fusion is necessary to prevent deformity's progression. The impact of spinal fusion on the rib cage deformity is still unknown.

## Research Question

What is the impact of posterior instrumentation and fusion on rib cage deformity in patients with AIS?

## Methods

35 patients with AIS (right thoracic scoliosis, Cobb =  $51 \pm 12^\circ$ ; 29F) underwent low dose biplanar X-rays before and after spinal fusion, followed by 3D reconstruction of the spine and rib cage. In addition to the classic spinopelvic and scoliosis parameters (Cobb angle and apical vertebral rotation AVR), the following 3D rib cage parameters were calculated: rib cage volumes (total, left and right part of the rib cage from T1 to T10), rib hump at each vertebral level, volumetric spine penetration index (VSPI), and ribs orientation in both sagittal and coronal planes. Pre and postoperative values were compared using the Wilcoxon signed-rank test, and the changes ( $\Delta$  = pre-post) were calculated.

## Results

Before surgery, 71% (25/35) of AIS patients exhibited severe rib cage deformity: 12/35 having rib hump  $> 9^\circ$  (type IIa), 4/35 having VSPI  $> 6.3\%$  (type IIb), 9/35 having both deformities (type IIc). In 51% of patients, T2 was the upper instrumented vertebra and L4 the lowest instrumented vertebra. Postoperatively, Cobb angle decreased by 67% ( $51 \pm 12^\circ$  to  $17 \pm 9^\circ$ ,  $p < 0.001$ ), AVR by 61% ( $18 \pm 7^\circ$  to  $7 \pm 5^\circ$ ,  $p < 0.001$ ), whereas thoracic kyphosis remained unchanged (T1T12:  $26 \pm 10^\circ$  to  $29 \pm 10^\circ$ ,  $p = 0.13$ ). Surgical correction led to 18% reduction in rib hump ( $11 \pm 6$  to  $9 \pm 6^\circ$ ;  $p < 0.001$ ), and 6% increase in rib cage volume ( $5115 \pm 908$  to  $5424 \pm 985 \text{ cm}^3$ ,  $p = 0.001$ ). However, VSPI did not change significantly ( $5 \pm 2\%$  to  $6 \pm 2\%$ ;  $p = 0.25$ ). After surgery, 20/25 patients preserved their severe rib cage deformity (6 type IIa, 8 type IIb, 6 type IIc). The decrease of AVR correlated with the increase of right rib cage volume ( $r = -0.5$ ) and frontal rib orientation ( $r = -0.7$ , both  $p < 0.05$ ; Figure 1).

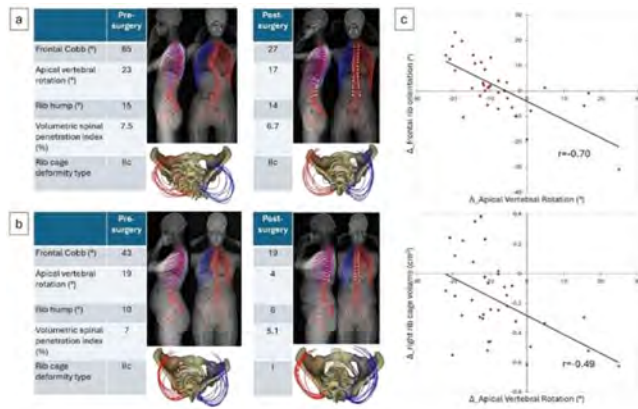
## Discussion

Common corrective maneuvers used in posterior instrumentation and fusion in AIS significantly improve coronal and horizontal spinal deformities but have no substantial impact on thoracic hypokyphosis. The effect on rib cage morphology correction is moderate. Correction of apical vertebral rotation seems to be related to the increase of rib cage volume and correction of frontal rib orientation, potentially reducing rib hump. VSPI does not seem to improve significantly with standard surgical techniques and may require an extensive interlaminar release or the use of sublaminar bands.

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Figure 1: Examples of AIS patients showing no changes in rib cage deformity (a) or reduced rib cage deformity after spinal fusion (b). Correlation between changes in apical vertebral rotation and rib cage parameters (c).



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## Short-Term Treatment Outcomes for Idiopathic Toe Walking

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## Introduction

Several intervention strategies may address idiopathic toe-walking (ITW) including observation, therapy, casting, Botox, and surgery. Literature is less clear in treating ITW exhibiting mild dorsiflexion contracture. Surgical and nonsurgical approaches have been proposed to increase dynamic ankle ROM. A systematic review concluded both serial-casting and surgery improve ankle dorsiflexion range and normalize gait. However, inconsistencies in changes and recurrence rates are reported, and only surgery resulted in sustainable improvement after one year[1]. We are conducting a prospective study to compare the 6 and 12-month efficacy of serial casting versus surgery in children with persistent ITW exhibiting mild contractures. 38/50 (76%) enrolled children currently have 6-month data. A summary of these early results is presented

## Research Question

Does either casting or surgery result in greater changes in gait when treating ITW with mild tendo-Achilles contractures?

## Methods

Children were recruited if they had a diagnosis of persistent ITW, between 6-18 years of age, and maximum ankle dorsiflexion with knee extension between 10° plantarflexion to 5° dorsiflexion. Standard of care treatment was provided based on surgeon-family

consultation. 3-D gait including a multi-segment foot model was performed at baseline, and 6 and 12-months post-treatment[2]. Gait deviation index (GDI), foot deviation index (FDI), and key hindfoot and midfoot ankle kinematics were used to compare pre-intervention to post-intervention gait[3]. Treatment effects were estimated using linear models adjusted for pre-treatment values ( $\alpha=0.05$ ). Data were also compared to age matched typically developing (TD) children.

## Results

Of the 38 children, 26 (51 feet) received casting (C) and 12 (23 feet) surgery (S). Groups were similar at baseline but of note, group S was 20 months older and had 4° more equinus contracture on average. GDI, FDI, and flexion at the hindfoot and midfoot showed 1) large improvements from baseline and 2) no differences between treatment groups. Mean GDI at 6-months were within TD range (C=95, S=92), FDI was just outside (C=89, S=89). Both groups exhibited excess hindfoot dorsiflexion (C=7°, S=9°) compared to TD children ( $p<0.001$ ). In the midfoot, casting improved to 4° more dorsiflexed than TD, surgery remained 4° more plantarflexed ( $p<0.001$ ).

Figure 1

## Discussion

Casting and surgery result in substantial and nearly equivalent improvements in gait in a homogeneous cohort of children with ITW and mild tendo-Achilles contractures. Both groups exhibited some over-correction of hindfoot dorsiflexion as TD values were exceeded. Surgery resulted in 15° mean dorsiflexion change compared to 10° for casting but differences were not significant ( $p=0.07$ ). The study protocol did not randomize treatment but relied on physician/family choice, resulting in more children receiving casting than surgery and some differences between the groups at baseline. Linear models adjusted for baseline differences. Further research will determine if the gains found here at the 6-month interval will last as we continue to follow these children.

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## Development of foot outward rotations in children with clubfoot treated with the Ponseti method and the foot abduction brace.

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## Introduction

The Foot Abduction Brace (FAB) is used to position the corrected clubfoot at 60° outward rotation after Ponseti casting<sup>1</sup>. Initially, it is

worn 23 hours per day for 3 months, then at night until the age of 4–5 years<sup>1</sup>. For unilateral clubfeet, the non-affected foot is positioned at 40° outward rotation<sup>1</sup>.

## Research Question

How do foot outward rotations develop in clubfeet and non-affected feet treated with FAB up to 7 years of age?

## Methods

Outward rotation was measured using the Foot Drawing Method (FDM)2 in 36 children with clubfoot treated with FAB. The children were treated with Ponseti casting followed by FAB, worn 23 hours per day until gait debut, then at night until the age of 4. FDM measures outward rotation in 2 levels -foot outward rotation (FR) and foot-tibia outward rotation (FTR). Measurements were taken at ages 2, 4, 5, and 7 years. Feet were categorized into clubfeet (n=36) and non-affected feet treated with a FAB (n=16). Non-affected feet without brace treatment (n=37) served as controls. The Mann-Whitney U test was used to compare outward rotations between clubfeet and non-affected feet at each age group and between non-affected feet and controls.

## Results

Table 1. Foot outward rotation (FR) and Foot Tibia outward rotation (FTR) measurements of Clubfeet and non-affected feet.

|     | Age (years) | Clubfoot |           | Non affected feet* |           | Control Group |           | P-value 1 | P-value 2 |
|-----|-------------|----------|-----------|--------------------|-----------|---------------|-----------|-----------|-----------|
|     |             | n        | mean (SD) | n                  | mean (SD) | n             | mean (SD) |           |           |
| FR  | 2           | 35       | 25° (9°)  | 14                 | 28° (8°)  | 37            | 24° (5°)  | 0.3       | 0.04      |
|     | 4           | 36       | 19° (7°)  | 15                 | 25° (8°)  | 36            | 22° (5°)  | 0.024     | 0.2       |
|     | 5           | 35       | 16° (6°)  | 15                 | 20° (4°)  | 33            | 20° (6°)  | 0.015     | 0.9       |
|     | 7           | 34       | 15° (6°)  | 14                 | 20° (5°)  | 30            | 18° (5°)  | 0.014     | 0.2       |
| FTR | 2           | 35       | 66° (19°) | 14                 | 73° (17°) | 37            | 56° (7°)  | 0.1       | <0.001    |
|     | 4           | 34       | 55° (19°) | 13                 | 60° (17°) | 36            | 52° (8°)  | 0.3       | 0.08      |
|     | 5           | 35       | 44° (13°) | 15                 | 50° (14°) | 33            | 50° (10°) | 0.2       | 0.3       |
|     | 7           | 34       | 34° (23°) | 14                 | 44° (12°) | 30            | 46° (10°) | 0.02      | 0.3       |

\*non-affected contralateral feet treated with the Foot Abduction Brace. Control group: non-affected contralateral feet not involved in brace treatment. n: number of measurements; SD: standard deviation. P-value 1: difference between clubfoot and non-affected feet. P-value 2: difference between non-affected feet and control group. Mann-Whitney U test was used and the significance level was set to  $p < 0.05$ .

Table 1 presents the FR and FTR measurements. Between the ages of 2 and 7 years of age the mean FR decreased from 25° to 15° and the FTR from 66° to 34°. The mean FR and FTR also decreased for the non-affected feet, however FR decreased at a lower rate compared to clubfeet. At ages 4, 5, and 7, the FR in clubfeet was significantly lower compared to non-affected feet, while the FTR became significantly lower at age 7. There were no significant differences in FR and FR between non-affected feet and controls except at the age of two years. (Table 1).

## Discussion

Clubfeet treated with the Foot Abduction Brace (FAB) maintained outward rotation until age 7, however a noticeable decreasing trend was observed. This suggests that while the FAB preserves correction, some loss of rotation occurs over time. In contrast, the non-affected

feet, which initially showed overcorrection during bracing, returned to normal values after the end of the treatment. These findings indicate that while the FAB is effective, continued follow-up is important to support long-term maintenance of correction in clubfeet

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## Lower leg angular velocity for offline EMG-inertial recordings synchronization and gait segmentation in healthy and cerebral palsy subjects

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## Introduction

The joint use of affordable technologies like surface electromyography (sEMG) and inertial measurement units (IMUs) is increasingly common in clinical and research gait analysis [1]. This multimodal approach is especially useful in cerebral palsy (CP), where muscle activation is atypical, motor control is impaired, and gait patterns are highly variable [2]. Accurate neuromuscular assessment requires precise synchronization of sEMG and kinematic data within the same gait cycle [3]. However, most available tools are costly or proprietary, limiting access in low-resource settings [4]. This highlights the need for open, reliable, and low-cost solutions to align muscle and motion signals in abnormal gait.

## Research Question

Can an offline synchronization method provide an accurate enough alignment of muscular and kinematic signals across populations with both typical and atypical gait patterns?

## Methods

Gait data were collected from 20 participants (ages 4–16), including 10 TD children and 10 with CP. The setup included 4 sEMG sensors (Trigno Discover, Delsys) and 8 IMUs (MVN Awinda, Xsens). Participants walked straight to the end of the room, turned, and returned to the starting point. Synchronization was based on angular velocity from the tibialis anterior (sEMG) and the lateral cranial side of each lower leg (IMUs), assuming anatomical

proximity would yield similar dynamics. A moving variance filter identified the turning point; the closest peaks before and after were used as temporal anchors (Fig. 1A). Gait cycles were segmented by independently detecting heel-strike (HS) and toe-off (TO) events in both modalities (Fig. 1B). Performance was assessed by computing the mean and standard deviation of time differences between corresponding HS and TO events. Additionally, the correlation coefficient between signals was calculated to evaluate synchronization consistency.

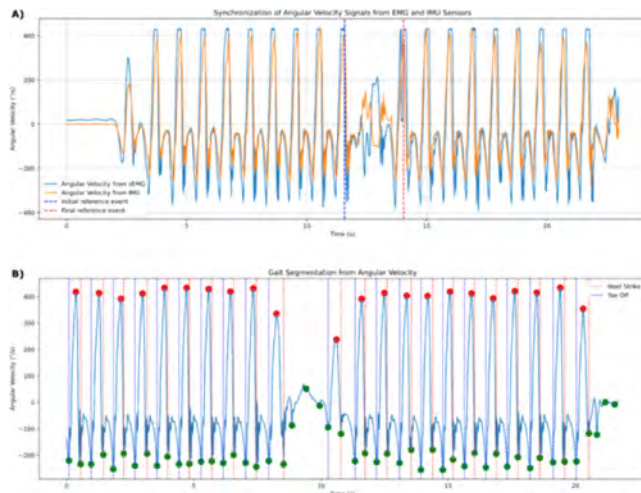


Fig 1. A) Synchronization of angular velocity signals; B) Gait cycle segmentation.

## Results

The method enabled accurate synchronization between EMG and kinematic signals in both groups. In TD children, error values were low (mean error:  $0.043 \pm 0.021$  s for HS,  $0.049 \pm 0.024$  s for TO), with correlation coefficients consistently close to 1.00. In CP group, although slightly higher error values were observed (mean error:  $0.073 \pm 0.044$  s for HS,  $0.080 \pm 0.039$  s for TO), signal agreement remained high correlation. Overall, the data confirm that signal synchronization is accurate and consistent, even in the presence of motor impairments and gait variability.

## Discussion

The proposed method achieved accurate synchronization between EMG and kinematic signals in both groups. Despite slightly higher errors in the CP group, correlation coefficients remained close to 1.00, confirming alignment reliability. Given a sampling rate of 100Hz, mean errors below 80ms represent less than 1% of a typical gait cycle [5], indicating negligible impact on event timing. Minor limitations include some angular velocity signals saturation due to suboptimal sensor configuration. Future work will refine sensor setup and validate the method against gold-standard motion capture systems.

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## Could idiopathic toe walking stem from atypical supraspinal modulation of spinal reflexes?

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## Introduction

Reflexes play an important role in movement control and maintaining posture, as they regulate muscle tone to adapt to internal and environmental changes (1). With the maturation of the nervous system, supraspinal structures play an increasingly significant role in modulating spinal reflexes, enabling smoother and more coordinated movements (2). Deviations in reflex modulation can result in abnormalities in gait and posture (3). Idiopathic toe walking, a pathological gait pattern observed in children, has been associated with functional alterations in spinal motor circuits (4). However, no studies have yet examined whether supraspinal input affects spinal reflex modulation differently in children with idiopathic toe walking.

## Research Question

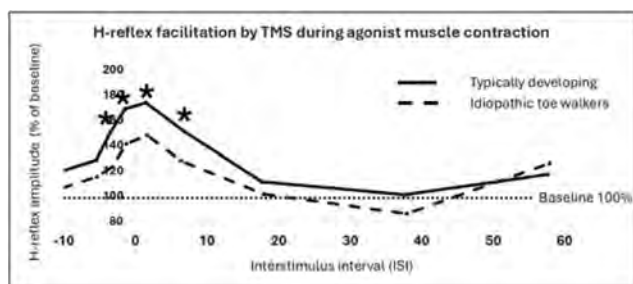
The aim of this study was to investigate whether supraspinal input modulates reflex function during active agonist muscle contraction (ankle plantar flexion) differently in idiopathic toe walkers compared to their typically developing controls.

## Methods

The study involved 11 idiopathic toe walkers aged 8.3–17.6 years and 11 typically developing children aged 7.5–16.2 years. While seated, participants actively maintained plantar flexion of their right ankle at 15% of their maximum voluntary contraction, using a visual force trace as guidance. Transcranial magnetic stimulation (TMS) at 90% of the motor threshold was applied to the left primary motor cortex foot area to examine its impact on spinal excitability, evaluated through the soleus H-reflex, which is an electrical equivalent of the stretch reflex. Eleven interstimulus intervals (ISI) were used: -10, -5, -3, +1, +3, +5, +10, +20, +40 and +60 ms. Negative ISI values represent the time the peripheral stimulus preceded the TMS pulse, while positive values indicate the opposite. Modulation of the H-reflex was captured using electromyography (EMG) from the soleus muscle.

## Results

In the typically developing group, supraspinal TMS during active plantar flexion resulted in a significant increase in spinal excitability within the ISI range of -3 to +10 ms ( $P > 0.05$ ). This was demonstrated by a significant increase of the soleus H-reflex amplitude compared to the baseline condition, in which no TMS was applied. A temporally similar modulation pattern was apparent in the idiopathic toe walkers, but the increase in spinal excitability did not achieve statistical significance.



## Discussion

The weaker increase in spinal excitability to supraspinal efferent input in the idiopathic toe walkers suggests differences in the corticospinal interaction and processing of the peripheral afference in the spinal cord between idiopathic toe walkers and their typically developing peers. This implies that so-called idiopathic toe walking might have underlying problems in the functioning of the central nervous system, particularly within the neural circuits that govern motor control and sensory integration.

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## Three gait parameters are sufficient to evaluate functional impairment in patients with adult spinal deformity

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## Introduction

Patients with Adult Spinal Deformity (ASD) are known to present with postural malalignment, functional limitations and deteriorated quality of life (QOL) [1, 2]. Improving health-related QOL represents a fundamental goal of therapeutic strategies in the management of adult spinal deformity. Remarkably, 3D gait analysis (3DGA) proved to predict QOL scores better than static radiographic evaluation in ASD [3]. However, 3DGA provides multiple kinematic parameters that can be a burden to interpret by spine surgeons. A recent study showed that the most representative gait kinematics of ASD patients were pelvic tilt (k-PT), the forward projection of the head and thorax (k-ODHA: head to hip axis angle with the vertical) and step length (SL) [4].

## Research Question

Can these 3 kinematic parameters alone still reflect deterioration of QOL scores in ASD?

## Methods

197 ASD patients and 143 controls underwent low-dose full-body biplanar Xrays with 3D skeletal reconstruction, full-body 3DGA during walking, and completed QOL questionnaires: SF36 with the physical and mental components (PCS & MCS), Oswestry Disability Index (ODI), Beck's depression inventory (BDI) and the Visual Analog Score (VAS) for pain. 23 ASD patients were followed up postoperatively (6 months to 2 years) and included in the analysis. A random forest machine learning model was used to predict QOL scores in 3 simulations with the following inputs respectively: (1) Xray parameters (spinopelvic and global alignment); (2) k-PT, k-ODHA and SL; (3) Xray parameters and k-PT, k-ODHA and SL (fig.1a). The prediction accuracy was evaluated using 10-fold cross-validation and compared between simulations. The same methodology was applied on the subset of post-operative ASD patients.

## Results

Simulations 1, 2 and 3 had a median accuracy of 81, 85 and 86%, respectively. Simulations 2 and 3 had comparable accuracies of

prediction for all QOL scores and higher predictions compared to Simulation 1 (i.e., accuracy for PCS =  $86 \pm 3$  vs  $90 \pm 2$  and  $91 \pm 3\%$  for simulations 1,2 & 3 resp.,  $p < 0.05$ ; fig.1b). Similar results were obtained for the postoperative ASD patients, with medians of accuracy of prediction of 83, 87 and 87% for simulations 1,2 and 3, respectively.

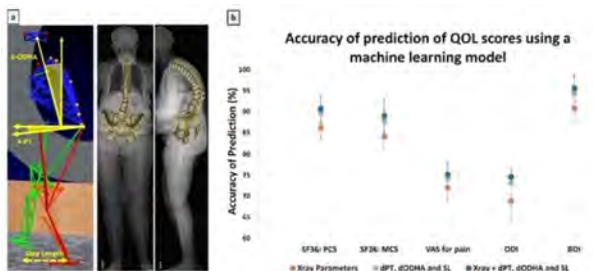
## Discussion

Head and pelvis kinematics and step length are sufficient to predict QOL scores, even postoperatively, with higher accuracies than classic spinopelvic and global alignment parameters. While the latter play an integrating role in the surgical planning of ASD patients, coupling radiographic to functional evaluation would be optimal to provide a complete assessment & postoperative follow-up. Future technologies should focus on capturing these 3 parameters alone to allow surgeons to easily access functional assessment, bypassing the complexity of the complete gait analysis process.

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Fig. 1. (a) k-PT, k-ODHA and Step Length; (b) accuracy of prediction of QOL scores in the 3 simulations.



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## Instrumental gait analysis (IGA) may suggest further neurologic assessments. A case study.

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## Introduction to clinical case

A 40-year-old woman was diagnosed with Parkinson's Disease (PD) in 2010. 123I-ioflupane SPECT imaging confirmed the presence of a reduced striatal presynaptic dopaminergic uptake, while brain

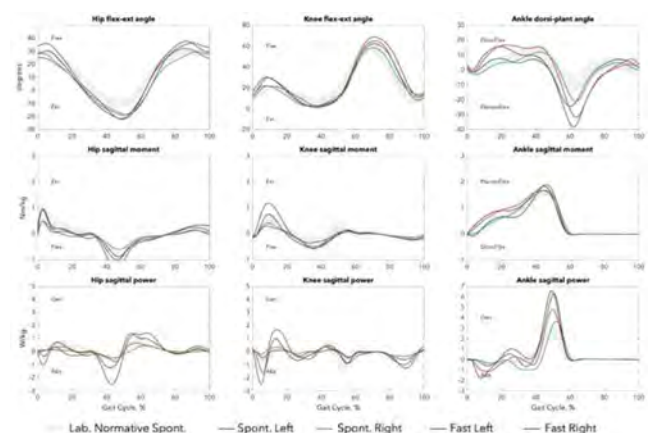
MRI was unremarkable. Treatment with pramipexole at low doses was started with persistent clinical benefit. In 2020, a PM&R clinician referred her to the gait lab because of lower limb pain and a minimally altered walking pattern not consistent with the diagnosis.

## Methods

At the clinical assessment, the patient presented with diffuse leg pain, tremors at rest in both lower limbs, and preserved walking ability (FAC = 5, WHS = 6) with minimal foot supination. Proximally, muscle strength and ROMs were normal, with a mild shortening of hip flexors. Distally, mild shortening was found bilaterally at the soleus and gastrocnemii, with MAS = 1 at the gastrocnemii and MRC = 4/5 for all plantar flexors. IGA with sEMG was carried out at self-selected and fast speed using a ten-camera motion capture system (BTS Bioengineering, Italy). Markers were placed according to the Conventional Gait Model. Electrodes were placed according to [1]. Five trials per limb were acquired.

## Results

During spontaneous walking ( $v = 66 \pm 3$  %height/s), GA showed normal spatiotemporal parameters – unexpected in a PD patient – but altered kinematics and kinetics, mainly at the ankle (Left > Right). Gait was characterized by quick ankle dorsiflexion at the beginning of single support (SS), leading to increased power absorption and subsequent generation. A similar mechanism was present at the end of SS with, again, increased power absorption and subsequent generation, resulting in increased plantarflexion. Dorsiflexion velocity was reduced in swing (see Figure). During fast walking ( $v = 98 \pm 3$  %height/s), this pattern increased bilaterally. Moreover, the hip bent into flexion during loading response, alternating power absorption and generation. We found this pattern – i.e., the replacement of isometric contractions under loading conditions with eccentric and concentric contractions to generate power – in patients with other neurodegenerative diseases aside from PD at a very early stage, and it is reasonably utilized to counteract muscle weakness [2, 3]. Consequently, the patient was referred to the neurologist, who decided to conduct a genetic analysis given the young age at onset. This revealed the presence of a biallelic pathological variant in the PRKN gene, resulting in a diagnosis of autosomal recessive PD, a rare hereditary degenerative condition.



Mean sagittal kinematics and kinetics at spontaneous and fast speeds. Confidence bands have been omitted for better readability.

## Conclusion and clinical relevance

In the presented case, IGA promoted a further neurological assessment that led to the identification of a rare hereditary disease and, consequently, to a more tailored treatment. Furthermore, the family was screened, and two siblings were found to be affected. This contribution from IGA is new to the literature and may open up a new pathway of research for gait laboratories.

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## Exploring lifting coordination in chronic low back pain using modified vector coding: a methodological case-based approach

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## Introduction

Modified vector coding (VC) has been used in biomechanics to analyze movement coordination and variability, including in populations with low back pain (LBP). However, the application of this data analysis approach to lifting task remains limited and its potential to characterize coordination during lifting in LBP remains unclear[1].

## Research Question

This work presents a methodological application of modified VC in lifting, using preliminary data from two individual cases.

## Methods

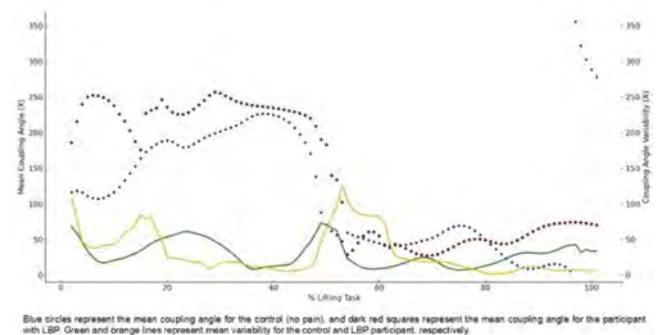
A cross-sectional case-based study was conducted with two female participants: one with LBP (46 years, 1.56m, 70kg, pain intensity 5/10) and a healthy control (28 years, 1.70m, 60kg). Data were collected using a 13-camera system (Oqus 300, Qualisys AB, Sweden) at 200 Hz. Participants lifted a 2.9Kg box at preferred style; six trials were selected. Kinematic data were processed using Visual 3D (Has-Motion, Inc) and filtered with a generalized cross-validatory

spline algorithm. Marker placement followed previously reported protocols[2], and a 9-segment model, with a segmented trunk, was optimized using segment optimization. Coordination was analyzed in the sagittal plane using thorax-lumbar and lumbar-pelvis angles as oscillators. Coupling angles and variability were computed using circular statistics[3]. VC analysis was performed using custom Python scripts (Spyder IDE). Angles were classified into coordination patterns: in-phase proximal dominance, in-phase distal dominance, anti-phase proximal dominance, and anti-phase distal dominance [3]. These describe the relative phase and contribution of proximal (thorax-lumbar) and distal (lumbar-pelvis) joint angles. In-phase patterns denote simultaneous directional change, while anti-phase indicates opposing movement. Dominance reflects the joint contributing most to coordination.

## Results

The participant with pain showed more in-phase coordination (87% vs. 70%), especially during descent (82% vs. 57%). Anti-phase was more common in the control (43% vs. 18% in descent). The LBP participant also showed greater distal dominance (73% vs. 52%), while the control had a more balanced profile. Global variability values were similar between participants with and without pain (30.7° vs. 29.8°, respectively). Coupling angles and variability profiles are illustrated in Picture 1.

Picture 1: Mean Coupling Angle and Variability



## Discussion

Methodological heterogeneity across studies limits comparisons. Ippersiel et al.[1] shares conceptual similarity by using VC, but their inclusion of non-dominant categories and different oscillators (hip-lower lumbar and upper lumbar-lower lumbar) restricts comparison of joint dominance. Thus, only mean coupling angles were comparable, though this overlooks the temporal specificity of coordination analyses. However, frame-by-frame mean coupling angle curves enables qualitative comparisons across studies. Nevertheless, they reported less in-phase coordination and more variable behavior in the LBP group. Frame-by-frame analysis revealed distinct strategies, with both participants showing increased variability during phase transitions; however, the peak was sharper and more abrupt in the participant with pain, suggesting reduced control. This underscores the value of phase-specific analysis, as global measures may miss task-relevant fluctuations.

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## Anticipatory postural adjustments after stroke: characteristics and influencing factors – a systematic review

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## Introduction

Anticipatory postural adjustments (APAs) are crucial for maintaining balance by preparing the body in advance of voluntary movement. These predictive, feedforward mechanisms stabilize posture in anticipation of self-initiated motion. Following a stroke, APAs are often impaired, contributing to postural instability and an increased risk of falls (1). Although APAs are vital for functional recovery, their characteristics across different tasks and the factors influencing their performance in stroke survivors remain insufficiently understood (2).

## Research Question

How are APAs characterized in stroke survivors during various functional tasks, and which key factors influence their generation and execution?

## Methods

A systematic search was conducted in PubMed and Web of Science up to December 2024, following PRISMA guidelines. Studies were included if they involved adults with stroke performing functional motor tasks (e.g., gait initiation, sit-to-stand, reaching)

and reported quantitative APA-related measures (e.g., muscle activation, center of pressure [CoP] displacement). Only clinical trials, cohort, and observational studies in English, Dutch, or French were considered.

## Results

Twenty-nine studies met the inclusion criteria. APAs were assessed using electromyography, force plates, motion capture, and accelerometry during tasks such as reaching, walking, and transitional movements. However, considerable heterogeneity existed in study protocols, including variations in functional tasks, measurement tools, and outcome definitions. Compared to controls, stroke survivors showed delayed muscle activation, reduced APA amplitude, prolonged APA duration, and impaired CoP displacements. Deficits were observed bilaterally, particularly in trunk and lower limb muscles. The presence and quality of APAs were modulated by factors such as motor impairment severity, stroke chronicity, lesion location, and task complexity. APA impairments were more pronounced during complex or dynamic tasks and when using the affected limb. Compensatory strategies involving the non-affected limb and trunk were commonly observed. Interventions such as external attentional focus, neuromuscular training, and functional electrical stimulation improved APA timing and magnitude in some studies.

## Discussion

This review highlights that stroke significantly alters the generation and execution of APAs across various functional tasks, with impairments affecting multiple APA dimensions including timing, amplitude, and coordination. These deficits are influenced by both individual and task-related factors and may evolve over time. The findings underscore the importance of targeted rehabilitation strategies that address APA-specific deficits to enhance motor recovery and functional independence in stroke survivors. Future research should focus on the development of standardized assessment protocols, longitudinal tracking of APA recovery, and the optimization of interventions tailored to individual patient profiles.

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## Establishment of Normative Lower Limb Kinematic and Spatio-Temporal Gait Parameters in Healthy Malaysian Adults Using Three-Dimensional Motion Capture

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### Introduction

Accurate normative gait data are foundational in clinical gait analysis, serving as critical references for diagnosis, assessment, and rehabilitation of individuals with locomotor impairments. Most existing databases are derived from Western and selected East Asian populations; however, anthropometric and biomechanical variations across ethnic and regional groups may render such references less applicable to Malaysians. The absence of localized normative data may limit the accuracy and clinical relevance of gait evaluations in this population. This study aimed to generate normative lower limb kinematic and spatio-temporal gait parameters for healthy Malaysian adults using a three-dimensional motion capture system. The resulting dataset provides essential benchmarks for enhancing individualized gait interpretation in clinical and research settings.

### Research Question

What are the normative values of lower limb joint kinematics and spatio-temporal gait parameters among healthy Malaysian adults, and are there notable gender-specific differences?

### Methods

A cross-sectional observational study was conducted at the PERKESO Tun Razak Rehabilitation Centre in Malacca. Twenty healthy adult participants (10 males, 10 females), aged between 23 and 39 years and with body mass index (BMI) within the normal range (mean BMI: males = 23.8 kg/m<sup>2</sup>; females = 22.0 kg/m<sup>2</sup>), were recruited. Reflective markers (n=16) were placed on anatomical landmarks according to the Plug-in Gait full-body model. Each participant performed multiple trials of level-ground walking at their preferred walking speed (PWS) over a 5-meter walkway. Gait data were captured using an eight-camera Vicon motion capture system (Vicon Motion Systems Ltd., Oxford, UK) and processed using Nexus and Polygon software to extract spatio-temporal and kinematic variables. The analysis focused on 12 key lower limb joint angles and core gait parameters including stride length, cadence, walking speed, and step width. Gender-based comparisons were performed descriptively.

### Results

Gait patterns were symmetrical and consistent across genders, though minor gender-specific differences were observed. Females showed greater angular excursions in hip rotation, knee flexion-extension, and ankle abduction-adduction. Males exhibited longer stride lengths (1.29 ± 0.06 m vs. 1.19 ± 0.07 m) and faster PWS (1.22 ± 0.08 m/s vs. 1.13 ± 0.07 m/s), likely reflecting their greater mean height (1.71 ± 0.05 m vs. 1.55 ± 0.04 m). Cadence was similar (111–114 steps/min). Step width was wider in males (0.20 ± 0.02 m) than females (0.16 ± 0.01 m). Bilateral asymmetries in joint kinematics were minimal.

### Discussion

This study establishes a normative gait dataset specific to healthy Malaysian adults and highlights modest but clinically relevant gender differences. These findings emphasize the need for population-specific references in gait analysis and support the development of sex-specific clinical benchmarks. Broader recruitment across age and BMI ranges is recommended for future studies to improve generalizability and provide deeper insight into gait adaptations over time.

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## Neuromuscular adaptations of quadriceps to cane use during gait in obese women with knee osteoarthritis

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## Introduction

Knee osteoarthritis (KOA) is a prevalent degenerative joint disorder that significantly impairs mobility, particularly in obese older adults. Joint stress plus age-related muscle loss worsens gait pain, stiffness, and limits function [1,2]. In this population, quadriceps muscle dysfunction is a common impairment, contributing to altered movement patterns and reduced joint stability [3]. Assistive devices, such as canes, are frequently recommended to offload the affected limb and improve gait efficiency; however, the precise neuromuscular effects of cane use on quadriceps activation in obese individuals with KOA remain unclear. While previous studies have examined the biomechanical effects of cane-assisted gait in healthy and osteoarthritic populations [4,5], limited research has focused on obese older adults. Furthermore, the influence of cane placement (ipsilateral vs. contralateral) on neuromuscular activation patterns is not well established. This study aimed to investigate the effect of unilateral cane use (contralateral vs. ipsilateral) on quadriceps neuromuscular activation during gait in obese women with KOA.

## Research Question

How does contralateral versus ipsilateral cane use affect quadriceps neuromuscular activation during gait in obese women with knee osteoarthritis?

## Methods

Eight obese women ( $75.5 \pm 8.5$  yrs,  $98.3 \pm 7.5$  kg,  $164 \pm 6.5$  cm) with KOA (Kellgren-Lawrence  $\geq 2$ ) and walking pain walked at preferred speed under three randomized conditions: unassisted, cane on left (contralateral), cane on right (ipsilateral). 12 Vicon motion

captures (100 Hz) tracked markers; Kistler force plates (1000 Hz) measured ground reaction forces. EMG system (Myon, 1000 Hz) recorded right leg quadriceps activity (VL, VM, VI, RF). EMG signals were bandpass-filtered (20–450 Hz), rectified, and smoothed using a 6 Hz low-pass Butterworth filter to create linear envelopes. All data were normalized to percentage cycle time to facilitate the data analysis.

## Results

All four quadriceps muscles (vastus lateralis [VL], vastus medialis [VM], vastus intermedius [VI], and rectus femoris [RF]) exhibited significantly reduced activation ( $p < 0.05$ ) during cane-assisted walking compared to unassisted gait (Figure 1). Contralateral cane use (left side) resulted in the greatest reduction in peak muscle activation across all quadriceps muscles (mean reduction: VL = 12.5%, VM = 7.5%, VI = 21.1%, RF = 20.1%). Ipsilateral cane use (right side) also reduced mean activation for vastus lateralis and rectus femoris.

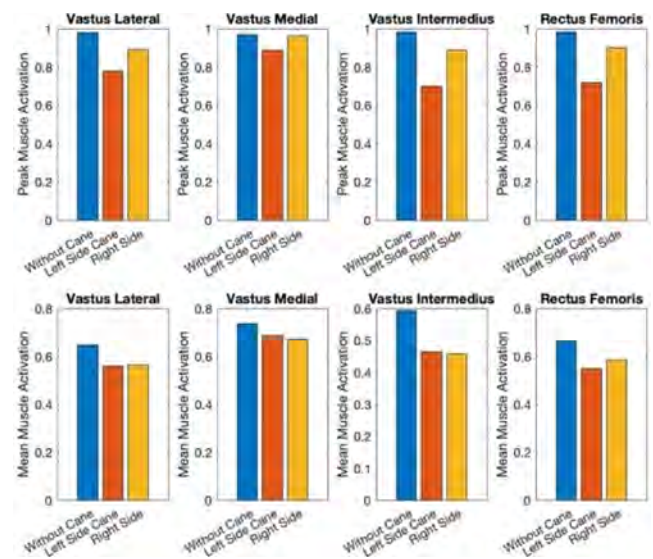


Figure 1: Peak and mean muscle activation of quadriceps muscles during unassisted, cane on left (contralateral), and cane on right (ipsilateral) walking gait.

## Discussion

In obese women with KOA, cane use reduces quadriceps activation during gait, with contralateral placement offering the greatest unloading benefit. These findings provide neuromuscular evidence to guide clinical recommendations, such as contralateral cane use should be prioritized to optimize quadriceps offloading. This study underscores the role of canes as a biomechanical intervention to mitigate quadriceps overuse in KOA. Future research should examine whether these acute neuromuscular changes translate to long-term improvements in pain and function.

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## Computational assessment of knee joint reaction forces in the supporting leg during taekwondo roundhouse kicking

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## Introduction

Taekwondo is a dynamic martial art characterized by high-velocity kicking techniques, with the roundhouse kick being one of the most frequently executed maneuvers [1-3]. While much attention has been given to the biomechanics of the kicking leg, the role of the supporting (stance) leg remains critical yet understudied [4-7]. The supporting leg provides stability, balance, and force transmission during the kick, making it susceptible to significant mechanical loading [8-12]. Excessive or repetitive joint forces in the knee may contribute to acute injuries or long-term degenerative changes, yet quantitative data on knee joint reaction forces (KJRFs) during taekwondo movements remain limited [13-18]. The primary aim of this study was to quantify the knee joint contact forces of the supporting leg during the taekwondo roundhouse kick using musculoskeletal modeling.

## Research Question

How does a taekwondo roundhouse kick affect knee joint reaction forces in the supporting leg across its pre-contact, contact, and post-contact phases?

## Methods

Twenty elite taekwondo athletes (10 male, 10 female; age  $25.5 \pm 4.5$  years, mass  $65.35 \pm 8.5$  kg, height  $174 \pm 4.50$  cm) performed roundhouse kicks in a motion analysis laboratory. Following a static calibration trial, kinematic data were collected at 100 Hz using a 12-camera Vicon MX system with 39 Plug-in-Gait markers, while ground reaction forces were simultaneously recorded at 1000 Hz using two Kistler force plates. The Rajagopal full-body musculoskeletal model was scaled to each participant's anthropometrics. Inverse kinematics and inverse dynamics calculated joint angles and moments, respectively. Static optimization and joint reaction analysis estimated muscle activation and joint reaction load patterns, respectively.

## Results

The analysis showed significant knee joint force variations during the roundhouse kick. Peak medial/lateral forces were highest post-contact ( $-11.86$  BW), while superior/inferior forces also peaked then ( $5.09$  BW). Anterior/posterior forces were stable, peaking pre-contact ( $-0.44$  BW). Mean forces followed similar trends, with

medial/lateral forces highest post-contact ( $-6.03$  BW). Superior/inferior forces increased progressively ( $0.33$  BW pre-contact to  $1.79$  BW post-contact), indicating heavy follow-through loading. The medial/lateral component dominated across all phases, highlighting dynamic multidirectional stress and potential injury implications.

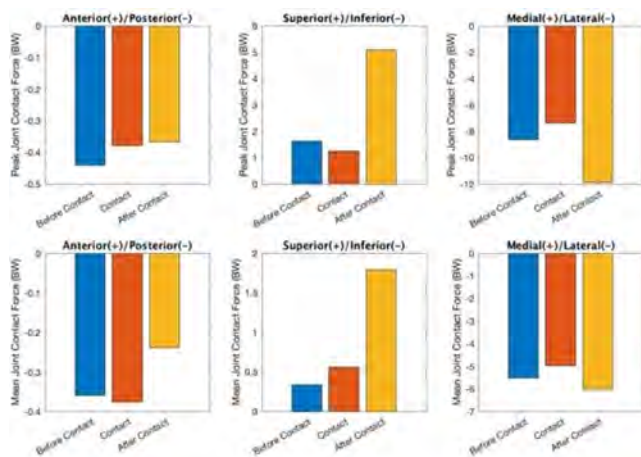


Figure 1: Peak and mean knee joint contact force in Anterior/Posterior, Superior/Inferior, and Medial/Lateral directions during the roundhouse kick of Taekwondo. Blue, red, and orange bar charts show knee joint contact force before, during, and after contact time, respectively.

## Discussion

This study quantified substantial multidirectional knee joint reaction forces in the supporting leg during taekwondo roundhouse kicks, with peak medial-lateral forces exceeding other directions. The post-contact phase generated the highest forces, suggesting greater injury risk during follow-through. These findings align with previous reports of high valgus loading in kicking sports but provide novel taekwondo-specific data. The observed force patterns emphasize the need for targeted strength training of knee stabilizers, particularly during deceleration. Long-term loading adaptations and knee pathologies in taekwondo warrant further research.

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### Three-Dimensional Glenohumeral Joint Reaction Forces During a Middle Direct Punch in Elite Taekwondo Athletes: A Musculoskeletal Modeling Analysis

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## Introduction

Martial arts, like Taekwondo, stress the glenohumeral (GH) joint during high-force movements like punches [1-4]. The middle direct punch (momtong jireugi) generates substantial GH joint reaction forces (GHRF), yet quantitative GHRF analyses, especially in elite athletes, are limited. Musculoskeletal modeling, like OpenSim, can estimate these forces [5-10]. While validated in throwing, its application to martial arts is lacking. Existing punch research neglects internal joint forces across punch phases. This study uses musculoskeletal modeling to quantify three-dimensional GHRF during middle direct punches in elite Taekwondo athletes.

## Research Question

How do glenohumeral joint reaction forces during a middle direct punch vary in various anatomical directions?

## Methods

Twenty elite Taekwondo athletes (10 males, 10 females; age:  $25.5 \pm 4.5$  years; mass:  $65.35 \pm 8.5$  kg; height:  $174 \pm 4.50$  cm; training experience:  $13 \pm 4$  years). 12 Vicon motion captures (100 Hz) tracked 14 markers on bony landmarks during static (anatomical pose) and dynamic Trials (5 maximal punches at sternum height, dominant hand) separated by 5-minute rest. The musculoskeletal model of the shoulder was selected as a base model and scaled by scaling tools of OpenSim software via marker data of static posture

for each subject. Inverse kinematics and Static Optimization were used to calculate the joint kinematics and muscle forces, respectively. Joint Reaction Analysis was used to calculate three-dimensional GHRF via kinematics and muscle forces data.

## Results

Figure 1 shows mean (line) and SD (shaded) glenohumeral joint reaction forces (GHRF, N/BW) during a Taekwondo middle punch in AP, SI, and ML directions. Distinct GHRF patterns were observed. Peak anterior force ( $1.36 \pm 0.63$  N/BW) occurred at  $40 \pm 5\%$  (contact). Superior forces dominated pre- and post-contact; minimum superior force was at contact. Peak medial force ( $1.37 \pm 0.32$  N/BW) occurred late ( $88 \pm 5\%$ ), with lateral forces during retraction. Acceleration (0-40%) showed rapid AP/SI force increases (shoulder flexion). Impact (40-60%) had force peaks (joint compression). Follow-through (60-100%) showed declining forces with posterior deltoid activation.

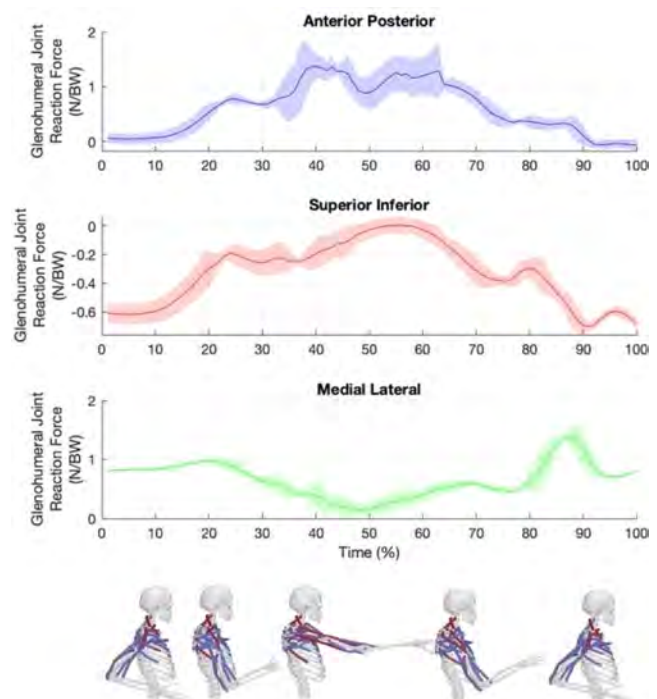


Figure 1: Mean (rigid line) and standard deviation (shaded area) of Glenohumeral joint reaction forces (N/BW) during a Taekwondo middle direct punch in anterior-posterior, superior-inferior, and medial-lateral directions.

## Discussion

This study uses OpenSim to analyze 3D glenohumeral joint reaction forces (GHRF) during the Taekwondo middle punch. Results show anterior force dominates at contact, exceeding superior and medial forces. These forces, though 20-30% lower than boxing, confirm anterior shoulder vulnerability. Superior forces indicate joint compression, potentially causing labral injuries. Strengthening the posterior cuff and emphasizing trunk rotation may reduce injury risk. This quantifies GH joint loads, informing targeted training and technique modifications. Future research should examine different punches and fatigue.

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## Predicting Gait Outcomes in Stroke Rehabilitation: The Contribution of Balance Control

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## Introduction

Chronic stroke survivors often experience gait impairments due to residual motor deficits, affecting their functional mobility and independence. Balance, as measured by the Berg Balance Scale (BBS), plays a critical role in walking ability. Gait dysfunction in stroke survivors is commonly characterized by reduced walking speed, asymmetrical step patterns, and altered temporal parameters, which can hinder community ambulation. Understanding the interplay between balance and gait parameters is essential for optimizing rehabilitation strategies aimed at enhancing stroke mobility. Several studies have explored the relationship between BBS scores and various gait parameters in stroke, consistently showing associations between balance and walking performance. However, despite these associations, there is a lack of research investigating whether BBS can predict specific spatiotemporal gait parameters. This study aims to examine the extend to which balance, as measured by BBS, predict key spatiotemporal gait parameters, including step length, step width, step time, cadence, gait speed, and gait symmetry, in chronic stroke survivors.

## Research Question

How does balance, as measured by the Berg Balance Scale (BBS), predict spatiotemporal gait parameters in chronic stroke survivors?

## Methods

A cross-sectional study was conducted on 85 chronic stroke survivors with ischemic cerebrovascular accident who had completed a structured rehabilitation program. Gait parameters, including step length, cadence, gait speed, step width, symmetry indices, and step time, were measured using a VICON motion capture system. Simple linear regression analyses were performed to assess the predictive value of BBS on step length, step width, step time, cadence, gait speed, and gait symmetry (step length and step time symmetry indices).

## Results

BBS significantly predicted step length ( $\beta = 0.257$ ,  $p = 0.018$ ), step time ( $\beta = -0.486$ ,  $p < 0.001$ ), cadence ( $\beta = 0.218$ ,  $p = 0.045$ ), and gait speed ( $\beta = 0.397$ ,  $p < 0.001$ ), indicating that better balance is associated with longer step length, shorter step time, higher cadence, and faster gait speed. However, BBS did not significantly predict step width ( $p = 0.986$ ), symmetrical step length ( $p = 0.127$ ), or symmetrical step time ( $p = 0.107$ ), suggesting that balance alone may not determine gait symmetry in chronic stroke individuals.

## Discussion

Balance, as measured by BBS, is a key factor influencing step length, step time, cadence, and gait speed in chronic stroke survivors but does not significantly impact gait symmetry. These findings highlight the importance of targeted rehabilitation strategies that address both balance and gait speed to improve overall walking function in stroke survivors. Future research should explore additional factors influencing gait symmetry beyond balance control.

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## Calf muscle morphological adaptations to functional power training in children with cerebral palsy

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## Introduction

Children with spastic cerebral palsy (CP) exhibit neurological impairments resulting in spasticity, paresis, and impaired motor control, which impact their movement patterns and structural muscle development [1]. The gastrocnemius medialis (GM) muscle

plays an essential role in push-off power generation during gait, and shows a reduced muscle volume and muscle length in children with CP [1]. These morphological impairments are presumed to contribute directly to limited walking and running capacity seen in children with CP [2]. Functional power training has been shown to improve walking and running ability [3]. However, it remains unknown how such a training program influences GM morphology. We hypothesized that the program increases GM muscle volume and the physiological cross-sectional area (PCSA), potentially lengthening the muscle belly due to the pennate GM architecture.

## Research Question

What is the effect of functional power training compared to usual care on GM morphology, including muscle belly length, tendon length, fascicle length, PCSA, and muscle volume, in children with CP?

## Methods

Twenty-two children with spastic CP (age:  $8.8 \pm 2$  years, 11 females, 13/9 GMFCS I/II) participated in a twelve-week functional power training program, consisting of three weekly 60-minute training sessions incorporating explosive running and walking exercises. GM morphology was assessed twelve weeks before (T0), just before (T1), and after completion of the training (T2) using 3D ultrasound (Figure 1A, [4]). As secondary measures, isometric GM strength (hand-held dynamometer; 90° hip flexion and extended knee) and running speed (muscle power sprint test) were assessed at T1 and T2. From 3D GM reconstructions, the origin, insertion, and the muscle-tendon junction (MTJ) were identified to calculate absolute muscle belly (origin–MTJ) and tendon lengths (insertion–MTJ); and to define the fascicle plane to measure fascicle length. Muscle volume was manually segmented and used to compute PCSA (volume/fascicle length). A blinded assessor analyzed GM morphology. Time points (T0 vs T1, T1 vs T2) and morphological changes between usual care ( $\Delta T1-T0$ ) and training ( $\Delta T2-T1$ ) were compared with paired t-tests ( $\alpha = 0.05$ ).

## Results

During usual care (T0 vs T1) the muscle belly slightly elongated with 1.3% ( $p = 0.01$ ). During the training period (T1 vs T2) PCSA (12%,  $p = 0.01$ ), muscle volume (10.5%,  $p < 0.001$ ), strength (19%,  $p < 0.001$ ) and running speed (14%,  $p < 0.001$ ) increased. Only muscle volume increased significantly more during the training period compared to usual care ( $p = 0.001$ ).

## Discussion

Functional power training increased GM muscle volume more than usual care but did not alter absolute muscle length, partially confirming our hypothesis. The increased muscle volume may have contributed to the improvements in walking and running ability. In future analyses, we will examine how functional improvements relate to morphology, assess growth effects, and further analyze CP subgroups.

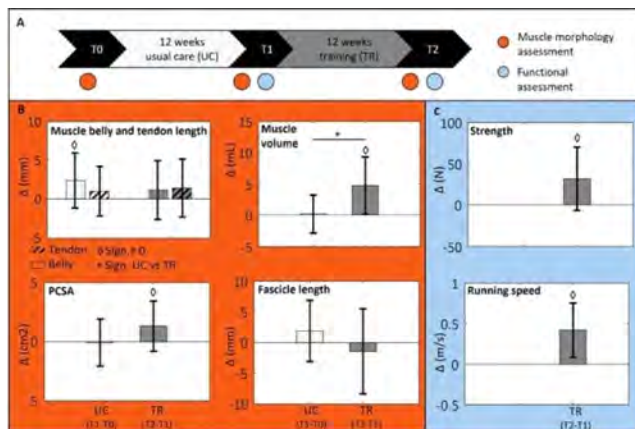


Figure 1. A) Research design. B) Morphological adaptations of the gastrocnemius medialis muscle. C) Functional progress after functional power training.

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## Alterations in gastrocnemius medialis muscle length after lengthening surgery in children with cerebral palsy

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## Introduction

Calf contracture muscles is common in children with spastic cerebral palsy (CP), leading to restricted ankle range of motion (ROM), which contributes to gait abnormalities [1-3]. When conservative interventions such as serial casting, orthotic management and physiotherapy fail to preserve functional mobility, surgical gastrocnemius aponeurotic lengthening may be indicated to improve the ankle ROM and walking ability [4]. Despite the widespread use

of this surgical intervention in CP, its effects on calf muscle morphology, i.e. calf muscle length, remain insufficiently understood. Quantifying muscle morphological adaptations to surgical intervention is important to evaluate treatment efficacy and to inform clinical decision-making.

## Research Question

What is the effect of gastrocnemius aponeurotic lengthening surgery on gastrocnemius medialis (GM) muscle length, i.e. muscle-tendon complex length, muscle belly length, and tendon length, in children with CP?

## Methods

Nine children with spastic CP (age:  $16.2 \pm 3$  years, 3 females, 6/3 GMFCS I/II) who at least underwent gastrocnemius aponeurotic lengthening surgery (Vulpus procedure) on the gastrocnemius medialis (GM) [4] were recruited. Control data from 45 TD children (age:  $11.4 \pm 4$  years, 33 females) were included for comparison [3]. GM lengths were obtained before (T0), and approximately one year after surgery (T1) using 3D ultrasound (Figure 1A, [5]). Foot position was standardized at a net ankle moment of 0Nm, while fixating the talocrural joint [6]. The corresponding footplate angle was assessed. All analyses were performed by the same blinded assessor. GM lengths were normalized to lower leg length. GM length adaptations were compared between T0 and T1 with paired t-tests. GM lengths were compared between TD and CP at T0 and T1 with unpaired t-tests ( $\alpha = 0.05$ ).

## Results

After surgery both the footplate angle ( $16^\circ$ ,  $p < 0.001$ ) and the muscle-tendon complex length (5%,  $p = 0.002$ ) increased, indicating greater dorsiflexion following surgery. This change was primarily driven by a 14% increase in tendon length ( $p < 0.001$ ). At T0, children with CP exhibited a  $23^\circ$  reduced footplate angle ( $p < 0.001$ ) and a 4% ( $p = 0.02$ ) shorter muscle-tendon complex length compared to TD children. However, these differences were no longer significant post-surgery. Muscle belly length was 11% ( $p < 0.001$ ) shorter in CP compared to TD at both T0 and T1. Additionally, tendon length was 23% ( $p < 0.001$ ) longer in CP at T0 compared to TD, and this discrepancy further increased following surgery.

## Discussion

Gastrocnemius aponeurotic lengthening surgery enhances dorsiflexion in CP by increasing tendon length. However, despite improved ankle ROM, the resulting tendon elongation may disrupt optimal muscle-tendon interaction during functional tasks. Future analyses will incorporate morphological parameters—muscle volume, physiological cross-sectional area, and fascicle length—for a more comprehensive understanding, alongside assessments of dynamic muscle-tendon function.

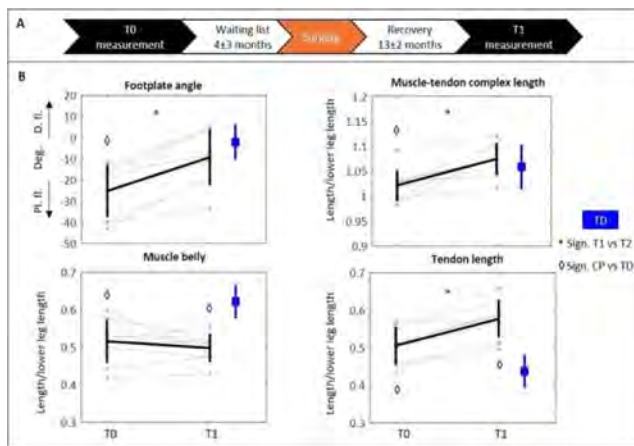


Figure 1. A) Research design. B) Footplate angle and morphological adaptations after surgery. Means  $\pm$  SDs are depicted for T0, T1 and TD data; individual data is grey.

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## Non-ambulatory individuals with severe spastic cerebral palsy show a profound decline in cardiorespiratory fitness and muscle strength

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## Introduction

Individuals with severe spastic cerebral palsy (CP) who depend on assistive devices present low physical activity levels [1], which are assumed to negatively influence their cardiovascular and muscular health [2, 3]. However, information about the extent of their cardiorespiratory fitness and muscle function is still scarce. Since such information is also needed to develop appropriate therapeutic strategies, the aim of this study was to determine the cardiorespiratory fitness and muscle strength of individuals with severe spastic CP in comparison to those of their typically developed (TD) peers.

## Research Question

How do cardiorespiratory fitness and muscle strength differ between individuals with severe spastic CP and TD individuals?

## Methods

Twelve individuals with severe spastic CP (age:  $22.67 \pm 4.9$  years; Gross Motor Function Classification System levels III-IV) and 14 TD individuals (age:  $22.93 \pm 4.71$  years) participated in this ongoing cross-sectional study. Both groups, with all participants naïve to frame running, performed a familiarization session prior to a 6-minute Frame Running test (6-MFRT) on a 400 m running track. The distance covered within six minutes was recorded using a measuring tape, and cardiorespiratory parameters (peak oxygen uptake ( $\text{VO}_{2\text{peak}}$ ), ventilation ( $\text{VE}_{\text{peak}}$ ), breathing frequency ( $\text{BF}_{\text{peak}}$ ), respiratory exchange ratio ( $\text{RER}_{\text{peak}}$ ), heart rate ( $\text{HR}_{\text{peak}}$ ), mean tidal volume ( $\text{VT}_{\text{mean}}$ )) were assessed using spirometry throughout the test. The participants' perceived effort was determined with the Borg Rating of Perceived Exertion (RPE) scale and the isometric muscle strength of the knee extensors and flexors using a strength training machine. Statistical analysis was performed by use of independent t-tests ( $p < 0.05$ ) and effect sizes were calculated using Cohen's d.

## Results

The participants with CP covered a significantly ( $p < 0.001$ ) shorter distance during the 6-MFRT compared to their TD peers (CP group:  $99.97 \pm 94.84$  m, TD group:  $1545.07 \pm 159.64$  m). Despite that, both groups reported similar perceived exertion levels on the Borg scale (CP group:  $15 \pm 3$ , TD group:  $15 \pm 2$ ). Cardiorespiratory data (Table 1) showed significantly lower  $\text{VO}_{2\text{peak}}$  ( $p < 0.01$ ),  $\text{VE}_{\text{peak}}$  ( $p = 0.005$ ) and  $\text{VT}_{\text{mean}}$  ( $p < 0.001$ ) in the CP group. However, both groups exhibited similar  $\text{RER}_{\text{peak}}$ ,  $\text{BF}_{\text{peak}}$  and  $\text{HR}_{\text{peak}}$  (Table 1). In addition, maximal muscle strength was significantly lower ( $p < 0.001$ ) in the CP group for both the knee extensors and flexors.

Table 1. Mean values of cardiorespiratory parameters measured during the 6-minute Frame Running test, and maximal isometric strength values for the knee extensor and knee flexor muscles in individuals with severe cerebral palsy (CP) and their typically developing (TD) peers.

|                                       | CP                       | TD                   | Cohen's d |
|---------------------------------------|--------------------------|----------------------|-----------|
| Cardiorespiratory parameter           | n=9                      | n=14                 |           |
| $\text{VO}_{2\text{peak}}$ [L/min/kg] | $24.11 \pm 6.86^*$       | $44.29 \pm 7.27^*$   | -2.85     |
| $\text{VE}_{\text{peak}}$ [L/min]     | $60.92 \pm 31.04^*$      | $113.31 \pm 33.72^*$ | -1.60     |
| $\text{BF}_{\text{peak}}$ [1/min]     | $56.89 \pm 12.47$        | $54.00 \pm 11.92$    | 0.24      |
| $\text{RER}_{\text{peak}}$            | $1.17 \pm 0.12$          | $1.22 \pm 0.19$      | -0.30     |
| $\text{VT}_{\text{mean}}$ [ml/kg]     | $16.14 \pm 4.11^*$       | $28.62 \pm 4.22^*$   | -2.99     |
| $\text{HR}_{\text{peak}}$ [bpm]       | $159.57 \pm 30.18$ (n=7) | $179.5 \pm 6.85$     | -1.11     |
| Muscle strength                       | n=12                     | n=14                 |           |
| max. knee extensor strength [Nm/kg]   | $1.2 \pm 0.73^*$         | $4.9 \pm 0.93^*$     | -4.38     |
| max. knee flexor strength [Nm/kg]     | $0.41 \pm 0.28^*$        | $2.47 \pm 0.42^*$    | -5.68     |

\* $p < 0.05$

$\text{VO}_{2\text{peak}}$  = peak oxygen uptake;  $\text{VE}_{\text{peak}}$  = peak ventilation;  $\text{BF}_{\text{peak}}$  = peak breathing frequency;  $\text{RER}_{\text{peak}}$  = peak respiratory exchange ratio;  $\text{VT}_{\text{mean}}$  = mean tidal volume;  $\text{HR}_{\text{peak}}$  = peak heart rate; max. = maximum.

## Discussion

The findings indicate severely reduced cardiorespiratory fitness and muscle strength in non-ambulatory individuals with spastic CP. In line with the lower mechanical output, the CP group showed impaired oxygen delivery, reduced oxygen consumption, and decreased muscle strength, suggesting a diminished muscle function, biomechanical alterations, and an impaired respiratory capacity. These findings emphasize the need for interventions targeting muscle function and respiratory capacity to improve overall fitness

and mobility of individuals with severe CP, thereby reducing their risk to develop cardiovascular diseases.

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## Effects of Symmetric and Asymmetric Sitting Postures on Lower Extremity Rotational Walking Kinematics in Healthy Adolescents

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## Introduction

Developing children spend most of their day sitting, with varied sitting behaviors such as cross-legged, W-sitting, long-legged, mermaid sitting, and kneeling [1]. Like standing and walking, active sitting reflects the refined performance of the neuromusculoskeletal system [2]. The loads from sitting behaviors are distributed differently across three planes to the lower extremity joints, mainly influencing the hip and pelvis mechanics. Over time, even minor musculoskeletal changes in the lower extremity can progressively affect gait biomechanics.

## Research Question

How does different sitting postures affect healthy adolescents' rotational lower extremity kinematics during walking?

## Methods

This study included 25 healthy adults (Age:  $21.26 \pm 0.45$ , Height:  $167.85 \pm 11.94$  cm, Weight:  $64.54 \pm 15.27$  kg, BMI:  $23.11 \pm 12.52$  kg/m<sup>2</sup>), determined with a 0.5 margin of error within a 0.95 confidence interval using G\*Power analysis. Participants' families provided photographs of their childhood floor sitting positions (ages 1-5 years). The control group consisted of individuals

sitting neutrally without hip rotation. Based on hip rotation, participants were classified into symmetric (W-sitting, cross-legged) and asymmetric (mermaid sitting) groups. Lower extremity walking kinematics (averaged over three trials) were recorded at similar gait velocities using a 3D motion analysis system (VICON Plug-in Gait Model)[3]. Statistical comparisons were conducted using t-tests ( $p < 0.05$ ).

## Results

The IR sides of the asymmetric (Mermaid sitters) and symmetric groups (W-sitters) exhibited significantly lower hip rotation, pelvic rotation, and foot progression angles compared to the neutral sitting group. The ER sides of the asymmetric (Mermaid sitters) and symmetric groups (cross-legged sitters) showed significantly lower hip rotation and foot progression angles while demonstrating higher pelvic rotation. No significant differences were observed in gait velocity, goniometric measurements, hypermobility, leg length discrepancy, or foot posture. Other relevant parameters are presented in Table 1.

**Table 1:** The statistical comparison of all interested walking parameters

| Kinematic Variables (*)                 | SYMMETRIC IR SITTERS | SYMMETRIC ER SITTERS | ASYMMETRIC IR SITTING SIDE | ASYMMETRIC ER SITTING SIDE | NEUTRAL POSITION SITTERS |
|-----------------------------------------|----------------------|----------------------|----------------------------|----------------------------|--------------------------|
|                                         | Mean $\pm$ SD        | Mean $\pm$ SD        | Mean $\pm$ SD              | Mean $\pm$ SD              | Mean $\pm$ SD            |
| Peak Pelvic: protraction (°)            | 8.14 $\pm$ 3.50      | 5.47 $\pm$ 2.14*     | 6.25 $\pm$ 3.22*           | 6.78 $\pm$ 4.04*           | 9.17 $\pm$ 2.96          |
| Peak Pelvic: Retraction (°)             | 7.29 $\pm$ 3.56      | 4.65 $\pm$ 2.17*     | 5.74 $\pm$ 4.09*           | 5.87 $\pm$ 3.31*           | 8.59 $\pm$ 3.28          |
| Pelvic Rotation ROM (°)                 | 15.61 $\pm$ 5.47*    | 9.99 $\pm$ 2.49*     | 12.34 $\pm$ 3.24*          | 12.19 $\pm$ 2.72*          | 17.77 $\pm$ 3.83         |
| Peak Hip Internal Rotation (°) @Stance  | 3.39 $\pm$ 7.04*     | 11.14 $\pm$ 12.35    | 7.50 $\pm$ 7.42            | 7.01 $\pm$ 5.18            | 11.46 $\pm$ 18.52        |
| Peak Hip External Rotation (°) @Stance  | 10.01 $\pm$ 5.54     | 1.17 $\pm$ 11.23     | 9.08 $\pm$ 7.46            | 5.08 $\pm$ 4.90            | 6.82 $\pm$ 3.56          |
| Hip Rotation ROM (°) @Stance            | 13.41 $\pm$ 4.81*    | 12.32 $\pm$ 4.44*    | 16.58 $\pm$ 1.71           | 17.11 $\pm$ 4.41           | 18.29 $\pm$ 7.24         |
| Mean Foot progression angle (°) @Stance | 6.43 $\pm$ 6.75*     | 10.17 $\pm$ 6.76*    | 1.24 $\pm$ 4.96*           | 0.21 $\pm$ 4.89*           | 0.71 $\pm$ 5.75          |
| Walking Speed (m/s)                     | 1.43 $\pm$ 1.13      | 1.32 $\pm$ 0.12      | 1.31 $\pm$ 0.33            | 1.33 $\pm$ 0.16            | 1.34 $\pm$ 0.15          |

\*: Difference between neutral and IR side in asymmetric sitters ( $p < 0.05$ ), \*: Difference between neutral and ER side asymmetric sitters  $p < 0.05$ , \*: Difference between neutral and IR symmetric sitters ( $p < 0.05$ ), \*: Difference between neutral and ER symmetric sitters ( $p < 0.05$ ), ROM : Range of motion.

## Discussion

Sitting posture inversely affects lower extremity rotation during walking, independent of gait velocity or other factors. The mermaid sitting position may alter lower extremity kinematics, leading to pelvic retraction and hip internal rotation on the externally rotated side, while promoting forward pelvic rotation on the internally rotated side. Individuals with rotational sitting postures (symmetric and asymmetric) exhibit greater external rotation in foot progression angles compared to neutral sitters, indicating significant changes in walking patterns. This increased external rotation is important for understanding lower extremity biomechanics and may impose extra loads on the trunk and joints, contributing to back pain and increased injury risk in high-level athletic activities[4]. Additionally, gait modifications from asymmetric sitting may be influenced by non-musculoskeletal factors like proprioception or environment. Further research on the biomechanical effects of asymmetric sitting

is necessary, in larger samples and for dynamic activities like running and jumping.

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## Effects of Increased Femoral Anteversion with Asymmetric and Neutral Sitters on Pelvis and Hip Rotational Gait Kinematics; A Pilot Study

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## Introduction

Developing children often adopt asymmetric or neutral sitting (NS) postures, such as mermaid position, which can alter hip and pelvic biomechanics and affect movement patterns(1,2). Increased femoral anteversion(IFA) is a lever arm dysfunction which also alters lower extremity biomechanics by increasing internal rotation (IR) during walking. When combined with asymmetric sitting(AS), these changes may disrupt postural control, balance, and gait patterns(3,4). However, the effects of IFA and different sitting postures on lower extremity kinematics remain unexplored.

## Research Question

How does increased femoral anteversion alter pelvis-hip gait rotations on IR and ER sides of mermaid sitters different from neutral sitters ?

## Methods

The study included 10 volunteers (Age:18.7±0.45y.o., Height:164.41±9.94cm, Weight:53.2±15.27kg) who divided into four groups: **IFA-groups**, Asymmetric sitters (**IR and ER sides**)[N:3] and neutral sitters[N:2]; **No-IFA-groups**, Asymmetric sitters(**IR and ER sides**)[N:2] and neutral sitters[N:3]. Parents provided photos of their children (ages 1–5) sitting on the floor. Asymmetric sitting was defined as one hip in IR and the other in ER. IFA was assessed using the TPAT test (IFA: TPAT>30°, No-IFA: TPAT<30°)(5). Each

participant completed three gait trials. Kinematic data were collected using a 3D motion analysis system (VICON Plug-in Gait Model)(6). Data were analyzed using SPSS, and a t-test was performed (p<0.05).

## Results

No significant differences were found between IFA AS and IFA NS in gait velocity, TPAT, internal rotation, tibial torsion, hypermobility, leg length discrepancy, or foot posture. In IFA NS, peak pelvic protraction and ROM remain unaffected, whereas in AS, peak pelvic retraction on the IR side reduces pelvic ROM. In IFA NS, peak hip internal rotation increases during the stance, while in AS, it decreases on the ER side. Peak hip external rotation during the stance is unaffected in IFA NS, whereas in a AS, hip internal rotation increases on the IR side. In IFA AS exhibit increased peak hip ROM during the stance phase, whereas AS show reduced peak hip ROM on the ER side. Other relevant parameters are presented in Table 1.

Table 1 : The statistical comparison of all interested walking parameters

|                                    | IFA                     |                   |                      | No-IFA                  |                   |                      |
|------------------------------------|-------------------------|-------------------|----------------------|-------------------------|-------------------|----------------------|
|                                    | ASYMMETRIC SITTERS (AS) |                   | NEUTRAL SITTERS (NS) | ASYMMETRIC SITTERS (AS) |                   | NEUTRAL SITTERS (NS) |
| Kinematic Variables (°)            | IR SIDE Mean ± SD       | ER SIDE Mean ± SD | Mean ± SD            | IR SIDE Mean ± SD       | ER SIDE Mean ± SD | Mean ± SD            |
| Peak Pelvic protraction (°)        | 6.72 ± 2.23*            | 4.03 ± 3.01       | 9.73 ± 1.96*         | 5.54 ± 4.47             | 10.00 ± 3.26      | 8.70 ± 3.46          |
| Peak Pelvic Retraction (°)         | 3.41 ± 1.62*            | 6.48 ± 2.05*      | 9.48 ± 3.03*         | 9.23 ± 4.28*            | 4.94 ± 4.72       | 7.90 ± 3.39          |
| Pelvic Rotation ROM(°)             | 10.71 ± 3.0*            | 10.53 ± 2.05      | 12.22 ± 2.15*        | 14.77 ± 1.58*           | 14.68 ± 1.30      | 16.61 ± 4.25*        |
| Peak Hip Internal Rotation(°) @St  | 4.66 ± 6.33*            | 9.50 ± 4.82       | 20.18 ± 21.83        | 11.7 ± 7.31             | 15.78 ± 2.83*     | 4.75 ± 13.09*        |
| Peak Hip External Rotation (°) @St | 12.17 ± 6.72            | 5.24 ± 6.32*      | 1.75 ± 15.21         | 4.44 ± 6.32*            | 4.88 ± 1.83       | 10.62 ± 11.68        |
| Hip Rotation ROM(°) @St            | 16.83 ± 1.80            | 14.74 ± 2.44*     | 21.93 ± 8.39*        | 16.21 ± 1.66            | 20.66 ± 4.56*     | 15.38 ± 4.64*        |
| WALKING SPEED(m/s)                 | 1.32 ± 0.34             | 1.34 ± 0.15       | 1.33 ± 0.31          | 1.31 ± 0.33             | 1.33 ± 0.16       | 1.34 ± 0.15          |
| Goniometric Measurement (°)        | IR SIDE Mean ± SD       | ER SIDE Mean ± SD | Mean ± SD            | IR SIDE Mean ± SD       | ER SIDE Mean ± SD | Mean ± SD            |
| TPAT (°)                           | 32.33 ± 2.08            | 34 ± 2.64         | 31 ± 13.57           | 15 ± 2.08*              | 15.5 ± 2.64*      | 20 ± 7.67*           |
| Hip Internal Rotation ROM (°)      | 41.66 ± 16.19           | 45.66 ± 22.27     | 64.66 ± 6.29         | 36 ± 16.19              | 36 ± 22.20        | 60 ± 18.38           |
| Second Toe Test (°)                | 10 ± 3.46               | 9.33 ± 6.35       | 14.33 ± 1.52         | 17 ± 3.46               | 18.5 ± 6.66       | 14.33 ± 1.52         |

\* IR sides differed between the IFA and No-IFA AS groups (p<0.05); & ER sides differed between the IFA and No-IFA NS groups (p<0.05); \* IFA and No-IFA NS differed (p<0.05); \* IR and ER sides of the IFA AS group differed (p<0.05); \*, Difference in IR sides of IFA AS and NS (p<0.05); ^, Difference in ER sides of IFA AS and NS (p<0.05). ROM : Range of motion, SD: Standard deviation, St: Stance

## Discussion

IFA has different effects on the NS and IR side of AS. For Neutral-sitters; although IFA has not significant effect on the pelvis, it increases hip internal rotation and ROM in stance. Contrary, for Asymmetric-sitters, IFA causes stiffness in the pelvis on the IR side and increases hip external rotation during the stance. The mermaid sitting posture, combined with IFA may alter pelvic-hip kinematics and loading behavior which may lead to back pain and higher athletic injury risks(7). Considering sitting posture history for IFA children along with goniometric tests (TPAT, tibial torsion, internal rotation), gait assessment and hypermobility may be beneficial in clinical practice and to prevent injury risk which should be investigated in detail.

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## Impact of reduced hip internal rotation on frontal plane lower extremity biomechanics during landing in volleyball players with pes planus

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## Introduction

Pes planus is a foot deformity characterized by increased pronation due to the collapse of the medial longitudinal arch, often leading to compensatory movement patterns in the lower extremity (1). In volleyball players (VP), these compensations may negatively affect landing performance and biomechanics during jumping. One common compensation observed during jumping in individuals with pes planus is increased hip internal rotation (2). Therefore, this study aimed to investigate the effects of reducing hip internal rotation on landing phase of jumping biomechanics in VP with pes planus.

## Research Question

How does reduced hip internal rotation affect frontal plane lower extremity biomechanics (FPLEB) during the landing phase of jumping in VP with pes planus?

## Methods

Eight VP with flexible pes planus ( $17.85 \pm 1.76$  years old,  $178.37 \pm 7.50$  cm,  $76.69 \pm 9.40$  kg) and 8 healthy VP as Controls ( $18.75 \pm 2.90$  years old.,  $180.68 \pm 8.56$  cm,  $71.87 \pm 9.25$  kg) participated. Athletes with a Foot Posture Index-6 (FPI-6) score greater than +5 were assigned to the pes planus group (PPG) (FPI-6:  $8.31 \pm 1.57$ ), while those with an FPI-6 score between 0 and +5 were classified as controls (FPI-6:  $1.25 \pm 0.96$ ) (3). Functional kinesio taping was applied to the PPG using a modified version of a predefined taping technique from the literature to reduce hip internal rotation (4). The jump task consisted of a countermovement jump (CMJ) with arm free. Jump data collected from PPG before taping were classified as Pre-Kinesio PPG (Pre-PPG), and data collected after taping were classified as Post-Kinesio PPG (Post-PPG). A gold-standard 3D motion capture (Vicon) system was used for assessment of FPLEB during the landing phase (5). Paired samples t-test are used for comparisons ( $p < 0.05$ ).

## Results

After kinesio taping, hip internal rotation, knee valgus, and ankle eversion angles significantly decreased during landing ( $p < 0.05$ ). Knee varus moment increased ( $p = 0.02$ ). No significant changes were found in power absorption. All interested parameters were illustrated in Table-1.

Table-1

| Landing Phase Kinematics             |                             |                              |                              |            |            |
|--------------------------------------|-----------------------------|------------------------------|------------------------------|------------|------------|
|                                      | Pre-PPG<br>(mean $\pm$ std) | Post-PPG<br>(mean $\pm$ std) | Controls<br>(mean $\pm$ std) | p1         | p2         |
| Peak Hip Internal Rotation Angle (°) | $4.04 \pm 12.71$            | $-0.48 \pm 7.91$             | $2.35 \pm 6.47$              | $<0.001^*$ | 0.417      |
| Peak Hip Adduction Angle (°)         | $-4.22 \pm 2.70$            | $-4.46 \pm 3.22$             | $-5.84 \pm 3.93$             | 0.535      | $0.021^*$  |
| Peak Knee Valgus Angle (°)           | $11.63 \pm 4.04$            | $8.18 \pm 4.50$              | $8.42 \pm 4.98$              | $<0.001^*$ | $<0.001^*$ |
| Peak Ankle Eversion Angle (°)        | $-5.72 \pm 4.06$            | $7.00 \pm 3.13$              | $6.74 \pm 1.93$              | $0.015^*$  | $0.048^*$  |
| Landing Phase Kinetics               |                             |                              |                              |            |            |
|                                      | Pre-PPG<br>(mean $\pm$ std) | Post-PPG<br>(mean $\pm$ std) | Controls<br>(mean $\pm$ std) | p1         | p2         |
| Peak Knee Varus Moment (N*m/kg)      | $0.57 \pm 0.36$             | $0.66 \pm 0.29$              | $0.60 \pm 0.33$              | $0.020^*$  | 0.746      |
| Hip Power Abs (W/kg)                 | $-16.75 \pm 11.23$          | $-17.30 \pm 9.72$            | $-20.23 \pm 10.35$           | 0.601      | 0.122      |
| Knee Power Abs (W/kg)                | $-22.03 \pm 11.09$          | $-24.01 \pm 8.01$            | $-25.28 \pm 11.55$           | 0.291      | 0.167      |
| Ankle Power Abs (W/kg)               | $-13.42 \pm 7.40$           | $-15.71 \pm 4.73$            | $-11.87 \pm 5.34$            | 0.075      | 0.244      |

\* $p < 0.05$ , std: standard deviation, abs: absorption, Pre-PPG: Pre-Kinesio Pes Planus Group, Post-PPG: Post-Kinesio Pes Planus Group p1: Pre-PPG vs Post-PPG significance, p2: Pre-PPG vs Control significance, N: Newton, m: meter, kg: kilogram, W: watt

## Discussion

This study shows reducing hip internal rotation through external rotation-directed kinesio taping improves FPLEB during the landing phase in VP with pes planus. Significant reductions in hip internal rotation, knee valgus, and ankle eversion after taping may indicate improved lower extremity alignment and stability (6). Although the changes in power absorption were not significant, the kinematic improvements may help reduce injury risk and enhance landing control by normalizing coronal plane knee kinetics and kinematics (6, 7). These findings highlight the role of proximal joint mechanics in regulating lower limb alignment and support hip external rotation kinesio taping as an acute corrective strategy for athletes with altered foot posture. For players with pes-planus, effects of treatment tools which alter proximal joint kinematics should be investigated in detail with a larger sample size in future studies.

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## Altered Drop Vertical Jump Mechanics in Increased Femoral Anteversion

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## Introduction

Increased femoral anteversion (IFA) is characterized by anterior rotation of the proximal head relative to the transcondylar axis of the femur (1). IFA has been associated with compensatory kinematic patterns such as excessive hip internal rotation, knee valgus, and foot pronation, potentially increasing the risk of sports-related injuries (2,3). The drop vertical jump (DVJ) test is widely used to assess lower extremity function and injury risk, particularly for anterior cruciate ligament (ACL) injuries. However, there is limited literature investigating the biomechanical effects of IFA during landing tasks. This study aims to determine the kinematic and kinetic alterations associated with IFA during DVJ testing.

## Research Question

What is the effect of increased femoral anteversion (IFA) on biomechanics and performance in the drop vertical jump?

## Methods

Five typically developing adolescents ( $13.8 \pm 2.58$  years) and four adolescents with increased femoral anteversion (IFA) ( $13.25 \pm 3.4$  years) participated in the study. Participants with a Craig's test measurement of 30 degrees or greater were classified into the IFA group (4). All participants underwent 3D motion analysis during a drop vertical jump (DVJ) task. The DVJ was performed with free arm movement, starting from a platform positioned 30 cm behind the force plates and elevated 40 cm above ground level. Kinematic and kinetic data were collected and analyzed. Independent t-tests were used for statistical analysis.

## Results

Control and IFA groups were not significantly different in age and BMI. The Craig's test angles were higher in the IFA group ( $37.37 \pm 10.2^\circ$ ) compared to the control group ( $10.8 \pm 6.36^\circ$ ) ( $p = 0.006$ ). The IFA group demonstrated greater peak hip internal rotation during the DVJ (Table 1). Sagittal plane hip and knee joint angles were significantly lower in the IFA group compared to controls. Additionally, hip and knee abductor moments were higher in the IFA group.

Table 1: Lower limb kinematics and kinetics

|                                         | Control            | IFA               | p            |
|-----------------------------------------|--------------------|-------------------|--------------|
| Peak Hip Flexion ( $^\circ$ )           | $86.43 \pm 10.54$  | $67.87 \pm 16.89$ | $<0.05$      |
| Hip Flex-Extension Range ( $^\circ$ )   | $53.99 \pm 8.49$   | $37.56 \pm 14.68$ | $<0.05$      |
| Peak Hip Internal Rotation ( $^\circ$ ) | $9.28 \pm 7.74$    | $20.55 \pm 20.08$ | $<0.05$      |
| Peak Knee Flexion ( $^\circ$ )          | $101.85 \pm 11.75$ | $89.04 \pm 14.92$ | $<0.05$      |
| Knee Flex-Extension Range ( $^\circ$ )  | $106.47 \pm 11.97$ | $63.98 \pm 16.02$ | $<0.05$      |
| Peak Ankle Dorsiflexion ( $^\circ$ )    | $36.69 \pm 4.23$   | $39.31 \pm 6.08$  | 0.08         |
| Peak Hip Abduction Moment (Nm/kg)       | $-0.26 \pm 0.12$   | $-0.46 \pm 0.35$  | $<0.05$      |
| Peak Knee Abduction Moment (Nm/kg)      | $-0.14 \pm 0.08$   | $-0.26 \pm 0.21$  | <b>0.015</b> |

## Discussion

This study demonstrated that adolescents with IFA exhibit distinct biomechanical differences during the drop vertical jump compared to controls. Reduced sagittal plane hip and knee angles in the IFA group may reflect compensatory strategies to stabilize joint motion or minimize discomfort. These results highlight the biomechanical consequences of increased femoral anteversion and suggest that femoral morphology should be considered in injury prevention and rehabilitation programs.

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## Biomechanical Influence of Anthropometry on Stroke Technique and Power Output in Malaysian Elite Rowers: A Motion and Force Analysis Approach

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## Introduction

In rowing, the integration of anthropometric characteristics, stroke technique, and force application plays a critical role in maximizing performance. While biomechanical optimization has been widely studied in Western rowers, limited research has explored the technical and physical demands placed on Asian athletes, particularly those with smaller anthropometric profiles. Malaysian elite rowers, often competing at a physical disadvantage, may adopt unique biomechanical adaptations to maintain performance. Understanding these compensatory mechanisms is essential for developing evidence-based training strategies. This study aimed to examine stroke rate-dependent changes in kinematics, force generation, and power output among elite Malaysian rowers, focusing on identifying upper-limb-driven adaptations that may offset anthropometric limitations.

## Research Question

To what extent do anthropometric characteristics influence stroke kinematics, reaction forces, and power output among Malaysian elite rowers during incremental stroke rate efforts?

## Methods

Ten elite Malaysian male rowers (mean age:  $21.5 \pm 1.6$  years; height:  $1.77 \pm 0.04$  m; body mass:  $77.5 \pm 7.0$  kg) participated in a controlled laboratory analysis. Rowing simulations were performed on a Concept2 Model D ergometer, instrumented with custom-built force sensors on both the handle and foot stretcher. Full-body kinematics were recorded using a Vicon 3D motion capture system (Oxford Metrics, UK) comprising eight Vantage V5 high-speed cameras (5 MP, 100 Hz), one Bonita 720c optical camera (1 MP, 250 fps), 39 retro-reflective markers (14 mm), and Nexus 2.11 software for biomechanical data processing. Each rower completed five trials across stroke rates from 18 to 34 stroke per minute. Key variables included segmental joint angles, angular velocities, reaction forces, and stroke-phase-specific power output. Data were normalized and analysed using repeated-measures ANOVA ( $p < 0.05$ ).

## Results

Increasing stroke rate was associated with significant increases in peak handle power (1309.1 W to 1872.0 W) and foot power (1129.0 W to 1330.8 W) ( $p < 0.05$ ), while peak reaction forces at both handle and foot remained statistically unchanged (handle: 831.2 N to 883.9 N; foot: 996.7 N to 1049.2 N;  $p > 0.05$ ). Handle velocity increased significantly (1.39 m/s to 1.75 m/s;  $p < 0.05$ ), whereas seat velocity showed no significant change (0.55 m/s to 0.69 m/s;  $p > 0.05$ ). Kinematic analysis revealed that shoulder and elbow joint angular velocities were key contributors to increased handle speed and power production, suggesting upper limb dominance in force application.

## Discussion

Despite anthropometric constraints, Malaysian elite rowers demonstrated the capacity to sustain force production and adapt to higher stroke rates through upper limb-driven compensatory mechanisms. These findings underscore the critical role of technique-specific biomechanics in overcoming structural limitations. Coaches and sport scientists should consider individualized kinematic profiling and upper-body power development to enhance rowing efficiency, especially in athletes with shorter statures or lower body mass indices.

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## Wireless Surface EMG Assessment of Trunk Muscle Activity During Lifting With a Passive Back-Support Exoskeleton in Elite Firefighters

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## Introduction

Firefighting is a physically strenuous profession characterized by frequent heavy lifting, awkward postures, and sustained exertion, all of which contribute to an elevated risk of lower back injuries and chronic musculoskeletal disorders (MSDs). Among elite firefighters, particularly those in high-risk sectors such as oil and gas, the cumulative spinal loading from operational tasks can severely impact career longevity and functional performance. Passive lumbar exoskeletons have emerged as a promising ergonomic intervention designed to reduce spinal load and muscular strain during lifting tasks. While prior studies suggest potential benefits, the neuromuscular effects of these devices, particularly in high-performance occupational populations, remain inadequately characterized. Furthermore, inter-individual variability in biomechanical adaptation and device-user interaction may influence their efficacy. Therefore, the objective of this study was to evaluate the effects of a passive back-support exoskeleton on thoracic and lumbar erector spinae muscle activation during standardized lifting tasks in elite firefighters, and to explore individual variability in neuromuscular response across different load conditions.

## Research Question

Does wearing a passive lumbar exoskeleton reduce thoracic and lumbar erector spinae muscle activation during lifting tasks in elite firefighters?

## Methods

Three elite male firefighters specializing in the oil and gas industry (height:  $171.7 \pm 6.1$  cm; body mass:  $67.7 \pm 17.0$  kg; years of service:  $8.7 \pm 8.0$  years) performed standardized lifting tasks under three load conditions (0 kg, 12 kg, 24 kg), both with and without a passive back-support exoskeleton. Surface electromyography (sEMG) sensors (Cometa Picolite) were placed bilaterally over thoracic and lumbar erector spinae muscles. EMG signals were normalized to maximum voluntary contraction (MVC), and root mean square (RMS) and integrated EMG (IEMG) values were calculated. Paired t-tests ( $p < 0.05$ ) were used to assess differences in muscle activation between conditions.

## Results

Significant reductions in lumbar muscle activation were observed in two participants, particularly under 12 kg and 24 kg loads. For example, RMS in the left lumbar erector spinae decreased from 1,735  $\mu$ V to 29  $\mu$ V ( $p < 0.001$ ) in one subject. Another participant showed reductions of 20–40% in IEMG for thoracic and lumbar erector spinae muscles ( $p < 0.05$ ). The third participant exhibited inconsistent results, with some instances of increased muscle activity when wearing the exoskeleton. Effects were minimal during unloaded (0 kg) lifting.

## Discussion

Passive lumbar exoskeletons can effectively reduce erector spinae muscle activation during moderate and heavy lifting tasks, with the most significant unloading observed in the lumbar region. However, the efficacy varies among individuals, likely due to differences in biomechanics, muscle recruitment strategies, and exosuit fit. Personalization and user-specific calibration may be critical to maximize the benefits of such devices.

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## A real-time, joint-specific index to detect compensatory gait in stroke rehabilitation

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## Introduction

Gait impairments are a common consequence of stroke, often reducing independence and quality of life [1]. A key aspect during recovery is compensation: altered movement patterns used to achieve walking function despite motor impairments. While compensation may restore function short-term, it can reinforce abnormal kinematics if not properly managed [2]. A Compensation Index (CI), a low-dimensional measure of compensation, could help quantify the magnitude of compensation patterns and ensure improvements in functional performance don't come at the expense of movement quality. Existing metrics like the Gait Deviation Index (GDI) provide a global gait quality score but are computationally demanding and not suitable for real-time data processing. Also, they mask joint specificity: Compensations in one joint may go undetected if improvements occur elsewhere. This makes it difficult to assess whether improved gait function is driven through compensation or whether gait is improving in a balanced, coordinated manner. To address these limitations, we developed a lightweight, joint-specific CI for real-time use during gait training and assessed its feasibility in a preliminary analysis.

## Research Question

Can a joint-specific, real-time CI effectively detect and monitor compensatory gait strategies during treadmill-based gait training in stroke patients?

## Methods

The CI was tested in eight stroke patients (mean age  $53.4 \pm 10.2$  years) undergoing treadmill-based gait training, split in two phases: baseline and feedback. In the baseline phase, patients walked normally on the treadmill. In the feedback phase, patients received real-time visual biofeedback to improve a patient-specific gait parameter. Movement was continuously assessed using optical motion capture and the Human Body Model 2. In a first calibration

step the CI compares the nine primary lower limb joint angles to normative healthy reference data, calculating an equally weighted deviation score where higher values indicate greater divergence from healthy gait. During the feedback phases, the CI is calculated at each frame for each joint, evaluating if the motion trajectories shifted from baseline towards healthy reference or further away. In this first feasibility evaluation, CI scores for baseline walking were compared to common post-stroke gait impairments, including reduced cadence, step length asymmetry, and shortened stride length.

## Results

Average deviation scores across all joints showed a trend of negative correlation with paretic step length ( $r = -0.61$ ,  $p = 0.11$ ) and cadence ( $r = -0.61$ ,  $p = 0.11$ ). A trend for positive correlation was found in step length symmetry ( $r = 0.41$ ,  $p = 0.32$ ), see Figure 1.

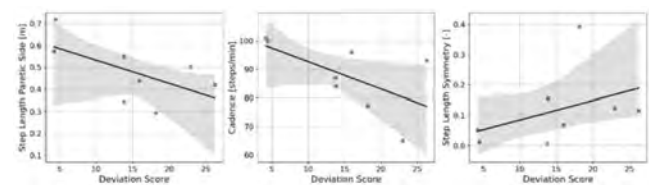


Figure 1: Correlation of baseline deviation score with gait parameters

## Discussion

Preliminary analysis of the CI shows promising trends, aligning with typical post-stroke gait impairments. Higher deviation scores were consistently associated with shorter steps, slower cadence, and higher asymmetry supporting its sensitivity to typical hemiparetic gait characteristics. Future work will compare CI scores to GDI and visual expert ratings to evaluate alignment and added value of real-time, joint-specific compensation detection.

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## Gait compensatory strategies used by a 4-year old cerebral palsy child with genu recurvatum: a biomechanical case study.

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### Introduction to clinical case

Gait impairments caused by neurological conditions disrupt foot clearance mechanisms, which are essential for the swing phase. [1]. To overcome such limitations, certain compensations like hip hiking or hip circumduction, are adopted to facilitate limb progression [2]. These compensations are well reported in adults (e.g. post-stroke hemiparesis) [3], but little is known about its presence in young children. This study presents the case of a young child with hemiparetic cerebral palsy performing a genu recurvatum gait, and describes the compensations used during walking in a biomechanical perspective.

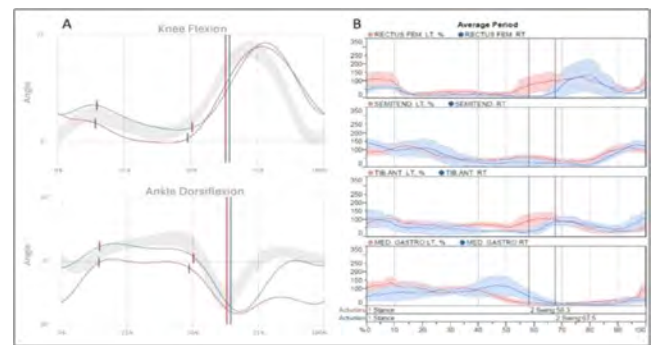
### Methods

Instrumented gait analysis was carried out on a 4-year old child with left hemiparesis due to a pre-natal stroke in the right middle cerebral artery. Gait kinematics were recorded using a 3-D marker-less motion capture system (Miquis, 100 Hz, Qualisys, Sweden), and gait kinetics were measured by two 3-D force plates (1KHz, Arsalis, Belgium) during overground barefoot walking. Muscle activity of the quadriceps, hamstrings, gastrocnemius and tibialis anterior was measured with surface electromyography (Noraxon Ultium, Noraxon, USA).

### Results

Gait spatio-temporal parameters were: speed of 3.3 km/h, cadence of 142 steps/min and longer stance phase on the right (67% vs 58%). Gait kinematics confirmed recurvatum of the left knee (up to 6°) just after loading response and left ankle equinus (average peak dorsiflexion during stance = 1°), which extended throughout the swing phase (average peak dorsiflexion during swing = -17° - Fig.1-A). Kinetics showed an internal flexor moment in the left knee and increased left ankle plantar flexor moment at loading response which was present throughout the remaining stance phase. Early and increased activity of the left triceps surae was observed (Fig.1-B) and was likely the cause for the genu recurvatum and ankle equinus. The compensations used to allow foot clearance were: 1) increased obliquity of the left pelvis, 2) left hip circumduction, evidenced by the higher abduction of the left hip during swing, and 3) contralateral vaulting, confirmed by an early plantarflexion of the right ankle (i.e. starting at 20% of the gait cycle - Fig.1-A), early activation of the right triceps surae and increased internal plantarflexion moment.

Figure 1. A) Knee and ankle kinematics, B) EMG results normalized to the gait cycle. Legend: red = left limb, blue = right limb.



### Conclusion and clinical relevance

The compensations observed in this 4-year old child with genu recurvatum and ankle equinus were: contralateral vaulting, increased pelvic obliquity and hip circumduction. Together, these strategies compensated for genu recurvatum and the lack of dorsiflexion necessary to ensure foot clearance during swing which and represent the most energy-effective compensations[4]. These compensations at a very young age are still not fully understood, therefore future longitudinal studies are important to monitor their development during growth, providing clinicians and families more tools for their decision-making process.

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### Quantifying Gait Smoothness: Are All Metrics Equivalent?

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## Introduction

Gait parameters reflecting movement smoothness, variability or predictability hold important diagnostic and prognostic value. Notably, patients with Parkinson's disease (PD) exhibit significantly less smooth, more variable and less predictable gait compared to age-matched healthy controls [1,2,3]. In PD, gait smoothness warrants particular attention, as its decline has been associated with an increased risk of falls [4] and with the occurrence of freezing of gait episodes [5]. Despite the clinical relevance of gait smoothness, multiple quantifiers have been proposed in the literature [6,7,8], each based on different principles. However, it remains unclear to what extent these quantifiers provide consistent and comparable information when applied to gait in PD.

## Research Question

Are the following movement smoothness quantifiers: log dimensionless jerk, harmonic ratio, and spectral arc length comparable during walking?

## Methods

We enrolled 86 PD patients (55 males, 31 females; mean age  $61.4 \pm 11.8$  years; Movement Disorder Society - Unified Parkinson's Disease Rating Scale, part III:  $25.3 \pm 11.1$ ; disease duration:  $5.1 \pm 2.6$  years) and 65 healthy controls (36 males, 29 females; mean age  $61.0 \pm 11.9$  years).

Six synchronized inertial measurement units (Opal, APDM Inc., USA) were placed on the chest, lower back (lumbar region), wrists, and feet. The sampling frequency was 128 Hz. All participants performed the extended Instrumented Timed Up and Go (iTUG) test. The iTUG test was performed twice, with a one-minute rest interval between trials. The measurement was part of a broader research protocol; however, for the purposes of this study, only data from four sensors—placed on the chest, lumbar region, and both shins—were analyzed.

Three gait parameters related to movement smoothness were computed: the spectral arc length (SPARC) [6], log dimensionless jerk (LJDJ) [7], and harmonic ratio (HR) [8]. Based on gait and turn events automatically detected and reported by the Mobility Lab software (APDM Inc.), only a single segment of the iTUG test was selected for analysis: the gait phase preceding the turn. Metric comparability was assessed using correlation coefficients.

## Results

Correlation analysis revealed multiple significant associations (Figure 1). SPARC and LDLJ showed moderate ( $\rho = 0.40$ – $0.59$ ) to strong ( $\rho = 0.60$ – $0.79$ ) correlations across all sensor locations (feet, lumbar, sternum) in both groups. In contrast, HR demonstrated comparable relationships only at the lumbar and sternum locations.

|         |    |      | Correlation coef. (p-value) |                |                |                |
|---------|----|------|-----------------------------|----------------|----------------|----------------|
|         |    |      | CON                         |                | PD             |                |
|         |    |      | HR                          | SPARC          | HR             | SPARC          |
| Feet    | AP | LDLJ | -0.13 (0.32)                | 0.57 (<0.001)  | -0.21 (0.05)   | 0.24 (0.03)    |
|         |    | HR   |                             | -0.07 (0.60)   |                | -0.10 (0.35)   |
|         | ML | LDLJ | -0.05 (0.69)                | 0.61 (<0.001)  | 0.05 (0.64)    | 0.59 (<0.001)  |
|         |    | HR   |                             | -0.03 (0.82)   |                | 0.01 (0.89)    |
|         | VE | LDLJ | -0.20 (0.10)                | 0.60 (<0.001)  | -0.22 (0.04)   | 0.61 (<0.001)  |
|         |    | HR   |                             | -0.25 (0.05)   |                | -0.14 (0.21)   |
| Lumbar  | AP | LDLJ | 0.18 (0.15)                 | 0.60 (<0.001)  | -0.14 (0.20)   | 0.62 (<0.001)  |
|         |    | HR   |                             | 0.13 (0.32)    |                | -0.42 (<0.001) |
|         | ML | LDLJ | -0.18 (0.15)                | 0.37 (<0.01)   | -0.17 (0.11)   | 0.39 (<0.001)  |
|         |    | HR   |                             | -0.20 (0.11)   |                | -0.12 (0.28)   |
|         | VE | LDLJ | -0.23 (0.07)                | 0.53 (<0.001)  | -0.42 (<0.001) | 0.59 (<0.001)  |
|         |    | HR   |                             | -0.35 (<0.001) |                | -0.56 (<0.001) |
| Sternum | AP | LDLJ | 0.17 (0.17)                 | 0.39 (<0.01)   | -0.14 (0.20)   | 0.54 (<0.001)  |
|         |    | HR   |                             | -0.31 (0.01)   |                | -0.28 (0.01)   |
|         | ML | LDLJ | 0.02 (0.85)                 | 0.52 (<0.001)  | 0.58 (<0.001)  | 0.50 (<0.01)   |
|         |    | HR   |                             | -0.34 (0.01)   |                | -0.34 (<0.001) |
|         | VE | LDLJ | -0.36 (<0.001)              | 0.56 (<0.001)  | -0.49 (<0.001) | 0.59 (<0.001)  |
|         |    | HR   |                             | -0.45 (<0.001) |                | -0.66 (<0.001) |

## Discussion

Given the observed interrelationships, none of the smoothness measures can be considered fully equivalent or interchangeable. Therefore, additional analyses are necessary before drawing direct comparisons between studies employing different metrics. Supported by project National Institute for Neurological Research (Programme EXCELES, ID Project No.LX22NPO5107) - Funded by the European Union – Next Generation EU and by Czech ministry of health (no. NW24-04-00259).

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## A virtual reality balance assessment for central sensory-motor integration

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## Introduction

Sensorimotor integration changes over the lifespan and with disease. Reliable methods to probe the underlying mechanism may be crucial to designing interventions to reduce fall risk, a major health cost factor in aging societies. One approach to identify the feedback dynamics underlying human balance control is to induce sensory conflicts, e.g. by moving the visual scene. The conflict induces sway, where the relation between conflict stimulus and sway response characterizes the system. Model based interpretations of this characterization can then be used to separate sensory, motor, or feedback-dynamics. However, current technologies are prohibitive to measuring these concepts at scale due to their cost and complexity in operation, and current school of thought is based on smaller controlled studies.

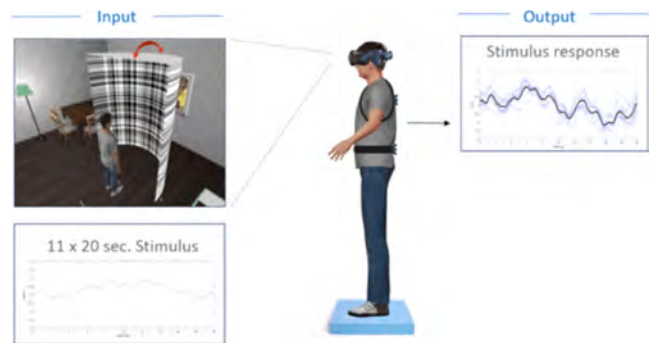
## Research Question

Novel methods development for clinical routine measurement of central sensory motor integration.

## Methods

We introduce a virtual reality (VR) implementation of a Central Sensory-Motor Integration (CSMI) test. Using off-the-shelf virtual reality systems, our freely available, easy to use, yet highly customizable software implementation makes these tests easily accessible for many settings. The in-built motion tracking of VR-systems is sufficiently accurate to measure sway behavior (1). We developed an open-source software solution to implement the experimental setup using a virtual moving screen, data recording, and automated analysis for specific CSMI tests. In this specific analysis, we evaluate test-retest reliability of the virtual CSMI in healthy young (n=20) and elderly (n=20) participants in clinical routine settings.

Figure 1: Visual representation of the vCSMI test



## Results

The virtual CSMI can be easily installed in any room at low cost using off-the-shelf VR systems. Multiple previous studies show good comparability with traditional systems for CSMI assessment using visual perturbations (2). We here additionally report test-retest reliability for recording lengths of below 3min across the lifespan from 20-80 years of age. ICC(3,1) values ranged from 0.89 to 0.96 for most parameters. Only time lag showed medium reproducibility with an ICC of 0.595.

## Discussion

How central sensory motor integration, especially of visual components, changes over the lifespan and with disease is currently an underinformed topic. However, this could be crucial in designing effective and efficient fall-risk reduction interventions. The virtual CSMI is a easy to use, reliable, and free tool to fill the blanks in central sensory motor integration.

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## Immediate and retained effects of single-session multidirectional perturbation-based training on balance recovery in older adults

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## Introduction

Falls are a leading cause of injury and loss of independence in older adults. Typically, falls are caused by an inability to recover balance after unexpected perturbations. Perturbation-based balance training (PBT) has emerged as effective strategy to improve reactive balance responses and reduce fall risk (1). Even a single PBT session demonstrably enhances balance recovery performance (2). However, many parameters regarding optimal dosing, direction of perturbations, and retention of training effects over time remain unspecified (1,3). In particular, the directionality of perturbations is an emerging area of research. While previous studies have demonstrated strong effects on immediate responses, most have relied on a single perturbation direction, which may limit real-life generalizability (1). In this study, we investigate if the strong responses elicited through multidirectional perturbations also confer a prolonged effect on balance recovery performance.

## Research Question

How does a single session of multidirectional PBT influence immediate and post-3month balance recovery responses in older adults?

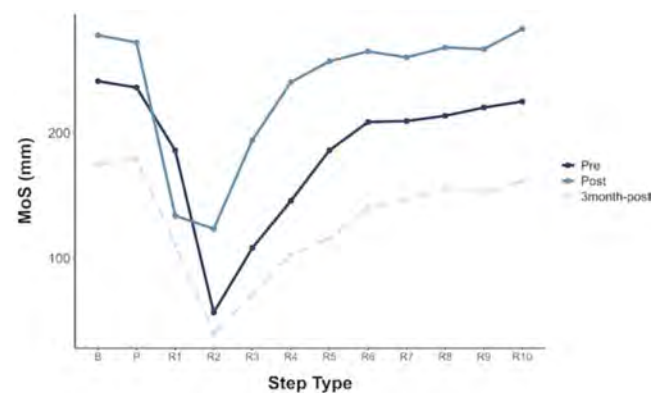
## Methods

Forty community-dwelling older adults ( $\geq 65$  years) without neurological or musculoskeletal impairments were randomly assigned to an intervention ( $n = 20$ ) or control group ( $n = 20$ ). The intervention group completed a single session of multidirectional PBT with three progressively increasing intensity blocks, consisting of 32 perturbations of increasing intensity in both anteroposterior (acceleration/deceleration of treadmill belt: 0.5–2.25 m/s<sup>2</sup>) and mediolateral (platform displacement: 2.35–18 cm) directions. The control group walked without perturbations. Balance recovery was assessed pre-, post-, and at three-month follow-up using number of recovery steps and margins of stability (MoS) in response to a deceleration perturbation (2.25 m/s<sup>2</sup>) during a 4-minute walking block. A one-way repeated-measures ANOVA was used for analysis.

## Results

Baseline characteristics were comparable between the groups, and all participants completed the study without adverse effects. The intervention group required significantly fewer recovery steps immediately after training (mean reduction: 4 steps,  $p = 0.012$ ), but this improvement was not retained at follow-up ( $p = 0.125$ ). The magnitude of destabilization (baseline to minimum MoS), representing the peak instability, remained unchanged across all timepoints ( $p = 0.33$ ). No significant changes were observed in the control group for recovery steps ( $p = 0.73$ ) or destabilization ( $p = 0.15$ ). Figure 1 illustrates MoS trajectories for the intervention group across timepoints.

Figure 1: MoS for the intervention group at all timepoints. B = Baseline, P = Perturbation step, R1–R10 = Recovery steps 1–10.



## Discussion

A single session of multidirectional PBT led to short-term improvements in reactive balance, evidenced by fewer recovery steps immediately post-training. However, these effects weren't maintained at the three-month follow-up, suggesting that continued or repeated exposure may be necessary for long-term retention. Despite faster recovery after training, participants were unable to reduce the initial impact of the perturbation, as indicated by the unchanged magnitude of destabilization. This suggests that the training improved recovery efficiency but not the ability to resist the perturbation itself.

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## Case Study: Treatment of a child with fibular hemimelia and clubfoot

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## Introduction to clinical case

Fibular Hemimelia is a rare congenital limb deficiency characterised by complete or partial absence of the fibula and accompanied with foot and ankle deformity and deficiency. The association of fibular hemimelia and structural clubfoot is not widely reported with the prevalence of clubfoot deformity within the population of fibular hemimelia stated to be 16% while in the reverse situation, ranges

between 1% - 2.5%(1, 2). Treatment of structural clubfoot in children with fibular hemimelia involves Ponseti casting and fitting of the foot abduction brace(1). The aim of the study is to investigate if splinting is effective at treating clubfoot in children with fibular hemimelia.

## Methods

In this case report, a 12-day old boy presents with right type 1A fibular hemimelia, based on the Achterman and Kalamchi classification system, accompanied with absent fourth and fifth ray and ipsilateral structural clubfoot. Child did not have a tarsal coalition on radiograph. The child had received the first Ponseti casting at another hospital on day 2 of birth. On presentation, the Pirani score was 4. Child continued to receive Ponseti treatment however complications arose during the 5th cast, with the problematic foot having slipped. Child was assessed to be ready for surgery with a Pirani score of 2. Post-tenotomy, the ankle achieved plantigrade and achieved a Pirani scoring of 1. The child did not fit the smallest 0000M Markell boots well and required a customised resting splint to hold the foot in position. A resting ankle foot orthosis made of 3 mm thick Orfit and layered with Moleskin, was casted with the foot in dorsiflexion and abduction. The splint was advised to be worn for 23 hours daily. Stretching and weightbearing exercises were taught. Subsequent casts were fabricated to accommodate the growing foot. The fit of the Markell boot remained problematic as the child showed high irritability and could kick it off. Issues arose 3 months into wearing the splint as there were reports from the parents that the foot had been slipping overnight resulting in poor compliance.

## Results

To date, right ankle equinus re-occurred 4 months post-tenotomy, leading to a second tenotomy performed under general anaesthesia, achieving 20 dorsiflexion, 60 abduction and a Pirani score of 0.

## Conclusion and clinical relevance

The gold standard for the management of idiopathic structural clubfeet is the Ponseti method and the application of a foot abduction brace(3-7). There is limited evidence to support the use of unilateral braces in this population(8). In this case report, a customised thermoplastic ankle foot splint was fabricated to fit an undersized foot. The splint demonstrated poor results in maintaining the ankle's range of motion and is dependent on the individual's casting skill. Recasting the child in plaster until the child's foot grows bigger would have been recommended.

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## Observational study of kinematic and ground reaction force distribution strategies in individuals with knee osteoarthritis

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## Introduction

Knee osteoarthritis (KOA) affected around 595 million people in 2020 [1]. It causes chronic pain and limits independence. The sit-to-stand (STS) task, essential for daily activities, is commonly used to assess functional ability in people with KOA. Most research emphasises sagittal plane movement, leaving the coronal and transverse planes underexplored. Research shows individuals with KOA tend to offload the affected leg during STS [2], leading to asymmetry and compensation. A comprehensive biomechanical analysis is needed to understand movement patterns and support personalised and effective rehabilitation strategies.

## Research Question

How does KOA alter lower limb kinematics and ground reaction force (GRF) distribution during STS task, and how does this differ between the affected and contralateral limb?

## Methods

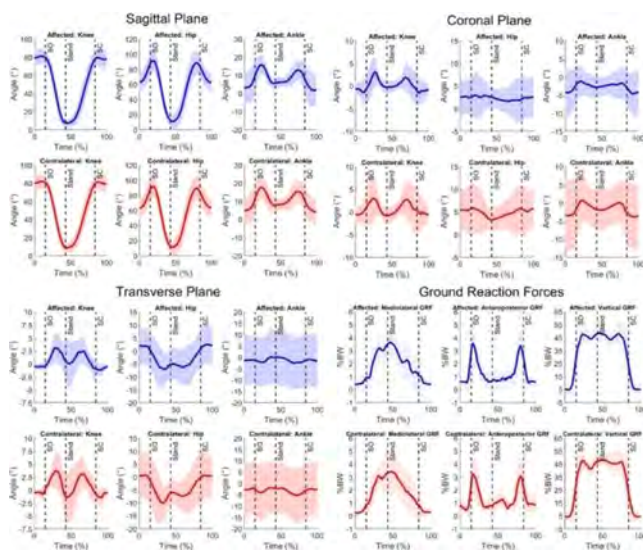
Fifteen individuals with KOA (Age:  $67.3 \pm 6.63$ , Sex: 12F, 3M) performed a 30s STS test at a self-selected pace and strategy. Kinematics was collected using inertial measurement units (Xsens Awinda Mtw, sampling rate: 60Hz), and GRFs were captured using two force plates (BTS Bioengineering P6000, sampling rate: 1000Hz). STS events (sit, sit-off [SO], stand, seat contact [SC]) were identified using custom algorithms (MATLAB version R2024b). Joint angles (hip, knee, and ankle), and GRFs, were statistically compared at each event using paired t-tests using Statistical Package for Social Sciences (SPSS, version 29.0.2.0). Movement waveforms were visually inspected.

## Results

In the transverse plane, the affected knee ( $-1.00 \pm 1.90^\circ$ ) was significantly more externally rotated than contralateral ( $-0.19 \pm 1.43^\circ$ ) at SC ( $t(13) = -2.471, p = 0.028$ ). In the sagittal plane, visual inspection revealed various propulsion and descent strategies. Hip and ankle patterns differed throughout the movement cycle in the coronal and transverse planes, with the knee exhibiting the greatest variability between SO and SC.

The affected leg showed significantly higher vertical GRF than contralateral at sit (affected:  $5.09 \pm 3.12$  %BW; contralateral:  $4.46 \pm 2.61$  %BW;  $t(12) = 2.355, p = 0.036$ ), and higher mediolateral GRF at SO (affected:  $1.22 \pm 1.41$  %BW; contralateral:  $0.91 \pm 1.13$  %BW;  $t(12) = 2.280, p = 0.042$ ) and SC (affected:  $1.28 \pm 1.09$  %BW; contralateral:  $0.85 \pm 0.98$  %BW;  $t(12) = 2.212, p = 0.047$ ). However, high variability can be observed, with some individuals placing more load on the contralateral leg.

**Mean joint angles and GRFs in affected and contralateral legs during STS cycles in individuals with KOA (n = 15)**



## Discussion

Contrary to previous findings, participants exhibited higher loading on the affected limb. However, higher mediolateral loading and greater knee external rotation at SC may indicate instability during descent. This altered strategy could increase shear forces in the knee joint, potentially accelerating KOA progression.

To better understand these compensatory patterns, a larger gender-balanced cohort is needed. Given the variability across STS events, waveform analysis is recommended. This could help identify consistent compensatory strategies used by individuals with KOA to mitigate discomfort, and in turn, inform the design of more targeted interventions.

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<https://doi.org/10.1016/j.gaitpost.2025.07.181>

## Effect of dual task on gait patterns of patients with cervical spinal stenosis

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## Introduction

Degenerative changes of the cervical spine can lead to a narrowing of the spinal canal (degenerative cervical spinal stenosis, DCSS) with subsequent compression of the spinal cord and in some cases a clinically manifested myelopathy (degenerative cervical myelopathy, DCM) [1]. Patients with DCSS or DCM typically have gait and balance impairments [1,2]. Difficulties during dual-task walking have been associated with aging and higher risk of falling [3,4]. Hence, dual-task walking might be more challenging for patients with DCM than patients with DCSS.

## Research Question

Does walking with dual-task change the gait patterns of patients with DCSS? And do these changes differ between patients with DCSS and DCM?

## Methods

Overall, 31 patients with cervical spinal stenosis (14 female, 17 male; age, median [interquartile range] 58 [51, 68.5] years; body mass index, 27.6 [22.5, 29.5] kg/m<sup>2</sup>) performed a gait analysis in a 20m hallway using seven inertial measurement units (RehaGait, Germany) placed on the pelvis, thighs, shanks and feet. In 11 patients DCM (intramedullary hyperintensity in the MRI) was diagnosed. Two gait trials at normal, comfortable walking speed were recorded, followed by two gait trials while naming the months of the year backwards (dual-task). The manufacturer's software calculated spatiotemporal parameters and sagittal ankle, knee and hip kinematics. Range of motion (RoM) of each joint during both gait conditions was calculated for each stride. Means and coefficients of variation (CoV = standard deviation/mean\*100) over all strides were calculated for the discrete gait parameters of each patient and condition. Gait parameters were compared between conditions using Wilcoxon signed rank test ( $P < 0.05$ ). Differences in the effect of dual task on gait between patients with DCSS and DCM were tested using Mann-Whitney U tests (dual task – normal;  $P < 0.05$ ).

## Results

Compared to the normal gait condition, patients walked slower, with a lower cadence and longer stride time in the dual-task condition ( $P < 0.012$ ; Table 1). Additionally, knee RoM in the swing phase was reduced by around  $1^\circ$  and the variability in knee RoM in both stance and swing phase increased compared to normal walking (Table 1). The change from normal walking to dual task walking did not differ between the patient subgroups (patients with CSS vs. patients with DCM,  $P > 0.208$ ).

**Table 1:** Comparison of the gait parameters between the normal gait and dual task gait condition for all patients with cervical spinal stenosis ( $N=31$ )

|                                    | Normal walking<br>Median [interquartile range] | Dual task walking<br>Median [interquartile range] | P value<br>(Mann-Whitney U) |
|------------------------------------|------------------------------------------------|---------------------------------------------------|-----------------------------|
| Walking speed (m/s)                | 1.01 [0.91, 1.19]                              | 0.94 [0.79, 1.09]                                 | <b>0.012</b>                |
| Cadence (steps/min)                | 105.0 [97.8, 111.6]                            | 97.1 [89.4, 105.4]                                | <b>&lt;0.001</b>            |
| Stride time (s)                    | 1.141 [1.077, 1.227]                           | 1.235 [1.138, 1.345]                              | <b>&lt;0.001</b>            |
| CoV stride time (%)                | 3.7 [2.8, 4.3]                                 | 3.7 [2.7, 5.2]                                    | 0.092                       |
| Ankle RoM ( $^\circ$ )             | 29.2 [27.6, 32.7]                              | 28.9 [24.5, 33.2]                                 | 0.256                       |
| CoV ankle RoM                      | 5.9 [4.9, 7.4]                                 | 6.4 [5.1, 8.3]                                    | 0.131                       |
| Knee RoM stance phase ( $^\circ$ ) | 19.7 [16.4, 21.0]                              | 19.0 [15.2, 22.8]                                 | 0.984                       |
| CoV knee RoM stance phase (%)      | 8.4 [7.3, 11.1]                                | 9.7 [7.5, 15.1]                                   | <b>0.034</b>                |
| Knee RoM swing phase ( $^\circ$ )  | 67.7 [61.3, 70.6]                              | 66.5 [58.5, 69.2]                                 | <b>0.013</b>                |
| CoV knee RoM swing phase (%)       | 2.4 [2.1, 3.4]                                 | 3.1 [2.1, 4.2]                                    | <b>0.031</b>                |
| Hip RoM ( $^\circ$ )               | 38.8 [35.1, 42.4]                              | 38.3 [33.9, 42.8]                                 | 0.614                       |
| CoV hip RoM (%)                    | 4.7 [4.0, 5.9]                                 | 4.8 [4.0, 5.9]                                    | 0.583                       |

## Discussion

The lower walking speed and cadence during dual-task compared to normal walking are in line with previous results in healthy older adults [5]. However, the more severely affected patients with DCM did not differ from patients with CSS, indicating that other tasks are needed to investigate the functional differences between these patient groups.

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## Relationship between Foot Mobility Magnitude and navicular bone displacement during the stance phase of gait

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## Introduction

Foot mobility is a critical component of lower limb biomechanics and has been implicated in various musculoskeletal disorders, such as plantar fasciitis, medial tibial stress syndrome, and patellofemoral pain. These conditions are often attributed to repetitive loading during walking and running. Therefore, capturing changes in foot posture during movement is essential for understanding the mechanisms underlying these injuries. However, the relationship between static and dynamic evaluations of foot posture remains inconclusive [1]. Foot mobility magnitude (FMM), regarded as the most reliable current measure of foot mobility [2], reflects the structural response to loading during static assessment and may thus have the potential for predicting dynamic changes in foot posture during movement.

## Research Question

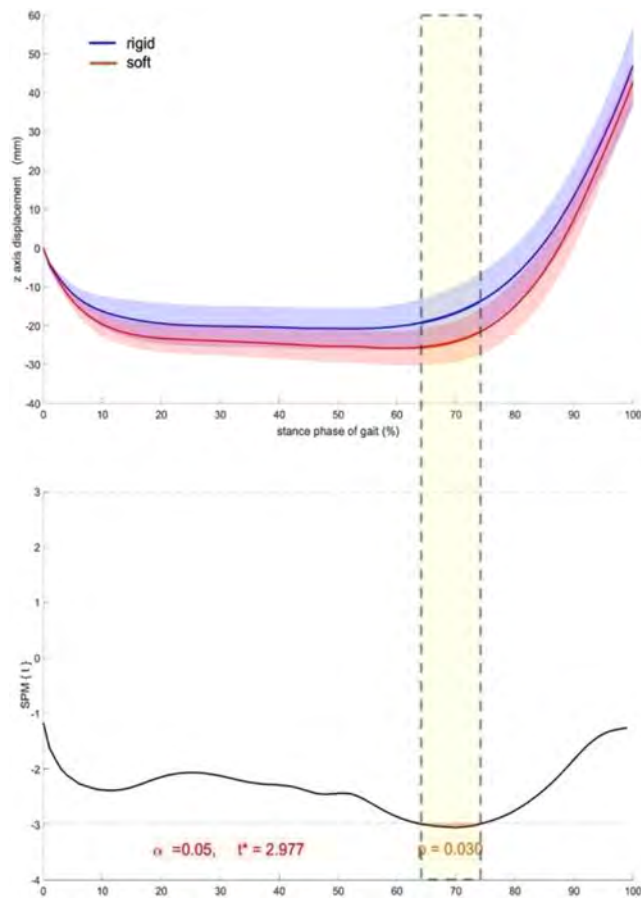
Does midfoot mobility influence the medial longitudinal arch height during gait?

## Methods

Fourteen university students (aged 19–25 years, 28 feet) participated in this study. Sixteen reflective markers were attached to specific anatomical landmarks according to the Plug-in Gait lower body model, with an additional marker placed on the navicular tuberosity. A three-dimensional motion capture system was used to collect data on foot kinematics and navicular bone displacement during the stance phase of gait. To assess midfoot mobility, arch height flexibility (AHF) and FMM were measured. Participants were divided into two groups based on the mean FMM value: a high-FMM (soft) group (14 feet) and a low-FMM (rigid) group (14 feet). Group differences in static foot posture were analyzed. One-dimensional statistical parametric mapping (SPM1d) was employed to evaluate differences in the time-series data of navicular bone displacement and foot kinematics. The level of statistical significance was set at 5%.

## Results

The soft group exhibited significantly greater AHF compared to the rigid group. No significant differences were observed in ankle joint kinematics during the stance phase. However, the soft group showed significantly greater navicular bone displacement along the Z-axis during 64–74% of the stance phase compared to the rigid group ( $p < 0.05$ ).



Navicular Bone Displacement: SPM Analysis of Soft group vs. Rigid group

## Discussion

This is the first study to examine the association between FMM and navicular bone displacement during the stance phase of gait. The 64–74% interval approximately corresponds to the late stance phase, during which extension of the metatarsophalangeal joints increases tension in the plantar fascia [3]. This tension contributes to the elevation of the medial longitudinal arch through the windlass mechanism. A previous study has reported that reduced plantar fascia stiffness impairs the function of this mechanism [4]. Therefore, individuals with greater midfoot mobility may exhibit reduced plantar fascia stiffness, potentially diminishing the effectiveness of the windlass mechanism. To better understand the effect of FMM on the displacement of the medial longitudinal arch during gait, an assessment of the foot muscle and tendon morphology, including the plantar fascia, may be necessary.

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## Injury Mechanisms, Situational Patterns, and Biomechanics of Anterior Cruciate Ligament Injuries in Professional Volleyball Players: A Systematic Video Analysis

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## Introduction

Anterior cruciate ligament (ACL) injuries are prevalent in volleyball, accounting for 26.3%-60.2% of all knee injuries (1,2). Understanding injury mechanisms, specifically focusing biomechanics is essential for developing effective injury prevention programs (3). While systematic video analyses have been conducted to investigate ACL injury mechanisms in various sports (4-6), the specific mechanisms, situational patterns and biomechanics of ACL injuries in women's volleyball remain unclear.

## Research Question

What are the mechanisms, situational patterns, and biomechanics associated with anterior cruciate ligament injuries in professional volleyball players?

## Methods

A total of 10 ACL injuries (8 female, 2 male) were analyzed. A systematic search of online resources was performed to identify ACL injuries occurred during competitions. An injury was included only when we were able to track an official statement from the team stating the nature of the ACL injury sustained by the player. Video footages were obtained from official match recordings and publicly available sources. Two sports physiotherapists have experience with volleyball injuries independently reviewed the videos and completed analysis forms for each injury. The assessment form consisted of variables including player position, court location, injury mechanism, game situation and biomechanical evaluation. Disagreements between reviewers were resolved by a senior reviewer.

## Results

The majority of injured players were outside hitters ( $n=6$ ) and all injuries ( $n=10$ ) were non-contact mechanism injuries. The most common game situations identified for ACL injuries were landing after block ( $n=3$ ) and landing after spike ( $n=3$ ) (Figure-1). 80% of injuries ( $n=8$ ) occurred in front of the net. Knee valgus loading ( $n=10$ ), trunk tilt ( $n=10$ ), hip abduction ( $n=7$ ), and foot eversion ( $n=8$ ) were frequently observed at the injury frame. Most of the analyzed injuries ( $n=8$ ) involved loading of the injured leg.



## Discussion

These findings suggest that non-contact mechanism following landing tasks is the main ACL injury pattern in professional volleyball players. Therefore, improving the ability to maintain frontal and transverse plane control during landing tasks may aid in reducing ACL injury risk in this group. However, kinematic analysis could not be performed in the current study due to the poor quality of the available video recordings, which limited the ability to accurately analyze injury frames. Future research with larger population and kinematic analysis of injuries is warranted to gain deeper insights into ACL injuries in volleyball players.

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## The Effect of Kinesio Taping on Running Kinematics in Recreational Runners With Shin Splint: A Pilot Study

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## Introduction

Shin splint is characterized by pain or discomfort of the lower leg as a result of repetitive stresses and it is a prevalent overuse condition among runners (1, 2). A recent systematic review examining the effectiveness of kinesio taping (KT) on shin splints indicated that the current evidence remains unclear, warranting the need for further high-quality research (3).

## Research Question

What is the immediate effect of KT on the lower limb kinematics of recreational runners with shin splints while running?

## Methods

A total of 4 recreational runners (2 female, mean age:  $25.0 \pm 3.2$ ; mean body mass index:  $22.5 \pm 1.4$ ; mean weekly running distance:  $9.7 \pm 4.8$ ) suffering shin splints were included. The inclusion criteria was presenting pain/discomfort along the lower 1/3rd of the posteromedial border of the tibia, induced by and lasted after running. 16 reflective markers were placed on anatomical landmarks of the participants. Following a warm-up at a self-selected pace for 5 minutes, participants ran at a constant speed of 10 km/h on a treadmill under two conditions: with and without kinesio taping. The KT was applied in a I-shape extending from the medial arch of the foot toward the tibia, by a certified practitioner (Figure 1). Video recordings were captured from the front, rear, and lateral views using three iPhone 14 cameras (Apple Inc., USA) positioned at a

distance of 100 cm from the treadmill during running trials. The recorded videos were analyzed with Kinovea software to measure lower limb kinematics (Figure 1). Wilcoxon signed rank test was performed to compare pre-KT and post-KT running kinematics, with the significance level set at  $p < 0.05$ .



## Results

The median value of foot dorsiflexion was  $11.35^\circ$  in the preKT and  $12.15^\circ$  in the postKT; knee flexion was  $11.75^\circ$  in the preKT and  $11.55^\circ$  in the postKT; hip flexion was  $26.15^\circ$  in the preKT and  $24.8^\circ$  in the postKT; frontal plane foot pronation was  $9^\circ$  in the preKT and  $7^\circ$  in the postKT; knee projection angle was  $171.15^\circ$  in the preKT and  $169.15^\circ$  in the postKT, foot eversion was  $12.6^\circ$  in the preKT and  $10.5^\circ$  in the postKT. However, no statistically significant effect of kinesio taping was observed in any of the lower limb kinematic parameters.

## Discussion

This pilot study did not identify significant effect of kinesio tape application on lower limb kinematics in recreational runners with shin splints while running. However, further research involving larger populations is warranted to better understand kinesio tape applications potential impact on running kinematics in runners suffering shin splint.

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<https://doi.org/10.1016/j.gaitpost.2025.07.185>

## ☆Changes in the distance between the talus and lateral malleolus during the stance phase of gait in chronic ankle instability

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## Introduction

Repeated lateral ankle sprains (LAS) can lead to the development of chronic ankle instability (CAI), which is characterized by mechanical instability, impaired proprioception, and altered joint kinematics. Previous studies have measured the distance between the talus and lateral malleolus using ultrasonography (US), and abnormalities in this distance have been reported in individuals with CAI. However, the dynamic changes in the distance between the talus and lateral malleolus during gait remain unclear.

## Research Question

Does the distance between the talus and lateral malleolus, as well as ankle joint angles, differ during the stance phase of gait in patients with CAI compared with that in healthy individuals ?

## Methods

This cross-sectional study included 12 individuals with the CAI group (16 feet) and 10 control group (20 feet). CAI participants met criteria including a history of repeated LAS and a CAIT score  $\leq 24$ . Controls had no history of ankle sprain and a CAIT score  $\geq 25$ . The distance between the talus and lateral malleolus was evaluated using B-mode US. The US probe was carefully positioned along the course of the anterior talofibular ligament (ATFL). Ankle joint angles, including dorsiflexion/plantarflexion and eversion/inversion, were measured during gait using a three-dimensional motion analysis (MA) system (Vicon Motion Systems Ltd., Oxford, UK). These measurements were performed simultaneously using a synchronized system that combined US and MA system. The participants walked 6 m over eight force plates (OR-6, 1,000 Hz; AMTI, Watertown, MA, USA), and the average of three trials was used for analysis. A two-way split-plot ANOVA was conducted to assess differences between groups and across stance phases. The significance level was set at 5%.

## Results

The CAI group demonstrated a significantly greater distance between the talus and lateral malleolus throughout the stance phase compared to the control group: early (18.96 mm vs. 16.06 mm,  $p < 0.001$ ), middle (17.45 mm vs. 14.45 mm,  $p < 0.001$ ), and terminal stance (20.01 mm vs. 15.47 mm,  $p < 0.001$ ). Additionally, the CAI group showed significantly smaller ankle dorsiflexion angles in the middle and terminal phases ( $p < 0.001$ ), and significantly greater inversion angles during stance phases ( $p < 0.001$ ).

## Discussion

The increased distance between the talus and lateral malleolus in the CAI group likely reflects mechanical instability due to laxity or elongation of the ATFL. Limited dorsiflexion and increased inversion

angles may contribute to a heightened risk of recurrent LAS. This study is the first to dynamically assess ankle joint behavior during gait using synchronized US and MA system, offering clinically relevant insights that cannot be captured through static evaluations. These findings may be crucial in developing targeted treatment and rehabilitation strategies for individuals with CAI.

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## Identifying the type and number of sensors required to capture behavioral complexity in daily activities in chronic stroke patients

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## Introduction

Tracking behavior in patients with stroke (or other motor impairments) outside the clinic or laboratory is becoming more feasible given the development of new sensor technologies. Data captured at home might have a higher ecologic validity, support interventions and increase the understanding of rehabilitation processes. There is, nonetheless, an issue in how to capture the richness of behavior and the changes over time that patients may display at home, while maintaining the acceptability of using sensors. Information-theoretic measures can be used to establish how much information (entropy) there is in a given behavior (e.g., how many bits of information a given movement demonstrates) and derive the minimum number of sensors and type of tracking devices necessary to describe such behavior.

## Research Question

What is the type and the required number of sensors to capture movement behavior of stroke patients during upper-limb activities?

## Methods

We reanalyzed a data set (1) from 19 patients with stroke performing three trials of 30 upper-limb tasks of varied difficulties (from a “thumbs-down” gesture to reaching and manipulating

objects). Ten Inertial Measurement Units (IMUs) that provide acceleration and orientation data were placed at the pelvis (P), trunk (Th8), both shoulders (SH), upper arms (UA), forearms (FA), and hands (HA). All tasks were considered as one, concatenating all trials and tasks time series to obtain a task-free measure of movement entropy. We aimed to identify the type (accelerometry or IMUs) and number of sensors required to express similar information (movement's Shannon's entropy). The comparisons from 10 sensors was made against the movement entropy from 6 (P, Th8, SH (active), UA, FA, HA), 4 (Th8, UA (active), FA, HA) or (SH (active), UA, FA, HA), 3 (Th8, UA (active), FA) or (UA (active), FA, HA), and 1 sensor (HA (active)). The acceleration or both acceleration and orientation of the sensors were considered. Cronbach's  $\alpha$  was used in the analysis with cutoff of  $\alpha = 0.8$  to consider the outcome to be sufficiently reliable (2).

## Results

The 4 IMUs placed at Th8, UA (active side), FA and HA was the best compromise to capture the behavioral complexity both for the paretic and non-paretic limb (Fig 1).

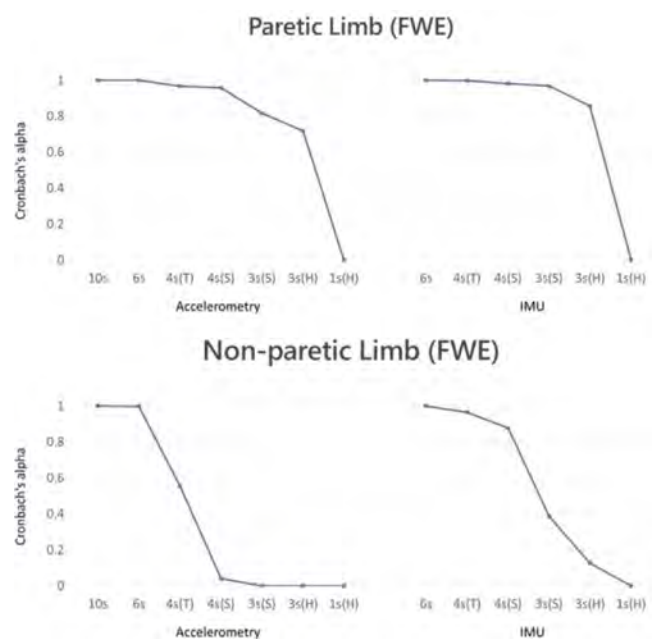


Fig. 1. Comparisons of movement entropy during upper limb activities measured with 10 sensors and 6s, 4s(Trunk), 4s(Shoulder), 3s(Shoulder), 3s(Hand) and 1s(Hand), respectively. Cronbach's alpha was used.

## Discussion

The use of 4 IMUs on the upper extremity was sufficient to describe the amount of movement entropy in the data set. This information can be used to increase patients' adherence in wearing sensors, as 4 sensors are likely to be more tolerable than a full set of 10 sensors while maintaining data quality. This study represents an initial step towards identification and quantification of upper limb movements in people after stroke.

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## Paediatric rehabilitation: Integrated instrumental evaluation of the upper limb in children with neuromotor disabilities.

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## Introduction

Autonomy and social integration are fundamental goals for improving the quality of life of patients affected by neuro-motor disabilities, in this scenario the Upper Limb (UL) rehabilitation is an important tool to achieve them. Several studies [1][2][3] demonstrate the efficacy of the robotic rehabilitation treatment for UL. The outcome measures generally include the output from the integrated sensors of the robotic device and clinical scales specific for UL. It appears not enough data are available of the integration of the kinematic evaluation with the robotic treatment and the clinical scales measures.

## Research Question

The aim of this study is to verify the correlation between the kinematic index by optoelectronic system (Vicon), the Armeo Spring Pediatric measures, and the clinical scale scores.

## Methods

15 hemiparetic children (10 cerebral palsy, 5 miscellaneous; age from 8 to 16 ys) were involved, evaluated before (T0) and after the treatment period (T1) which lasted 4 weeks and included a daily session of robotic treatment, a physiotherapy and occupational therapy. The outcome measures includes the Arm profile Score (APS) from the kinematic, Armeo specific measures and clinical score from Quest, Cheq and the Modified Melbourne scales. The study provides a 6 months (T2) and 1 year (T3) follow up.

## Results

A statistical analysis was conducted on 15 patients with Spearman's Rho at T0, T1 and T2. At T0 it is observed a significant correlation between APS Score of the hemiplegic hand and clinical scales (ADL tasks vs QUEST:  $p < 0.013$ ; ADL task vs CHEQ "two hands":  $p < 0.045$ ). At T1 a significant correlation between APS Score and Armeo measures (reaching task vs deviation:  $p < 0.001$ ; bimanual task vs deviation:  $p < 0.003$ ). A significant correlation is found between APS Score and clinical scales (ADL task vs QUEST:

$p < 0.006$ ; ADL task vs CHEQ "two hands":  $p < 0.024$ ; bimanual task vs QUEST:  $p < 0.024$ ). It is particularly interesting the significant correlation between the global APS and the QUEST Scale ( $p < 0.001$ ). At T2 a significant correlation between APSs and Armeo (reaching tasks vs instability:  $p < 0.005$ ; bimanual task vs instability:  $p < 0.019$ ). The APS correlates well with the QUEST Scale, in particular with the two bimanual tasks (task 1:  $p < 0.001$ ; task2:  $p < 0.019$ ) and the global index ( $p < 0.002$ ). 9 patients completed the T3, not enough for a statistical analysis.

## Discussion

Even though the data presented are preliminary, they illustrate a significant correlation between the clinical evaluations, the robotic measures, and the kinematic index. That said, kinematic analysis enables to quantify ULs movements in more ecological way compared the Armeo's or clinical scales tasks separately, providing a solid support for its clinical application.

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## Plantar Pressure Reconfiguration via Custom CAD/CAM Insoles in Pes Planus: A Pedobarographic and PCA-Based Study

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## Introduction

Pes planus (flatfoot) is a prevalent foot deformity characterized by a reduced medial longitudinal arch, often resulting in altered plantar pressure distribution, impaired postural control, and compensatory gait adaptations. Custom-made orthotic insoles manufactured via CAD/CAM technology offer structural support tailored to individual foot morphology. However, the extent of their biomechanical impact remains underexplored.

## Research Question

Can custom CAD/CAM foot orthoses effectively redistribute plantar pressure in individuals diagnosed with flexible pes planus?

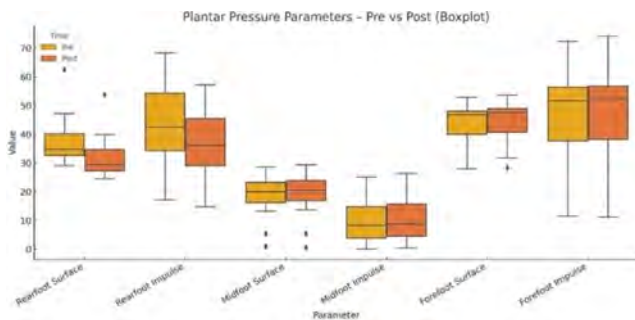
## Methods

Fifteen participants (mean age:  $29.6 \pm 9.3$  years; 11 females, 4 males) with clinically confirmed flexible pes planus were recruited

from Kale Prosthetics and Orthotics Center, Istanbul, Türkiye. Each participant received custom-made CAD/CAM insoles fabricated using 3D foot scanning and CNC milling. The intervention lasted 8 weeks, during which participants consistently used the insoles in daily life. Pedobarographic data—surface area and impulse across rearfoot, midfoot, and forefoot—were collected before and after the intervention. Paired t-tests and Cohen's d effect sizes were calculated. Principal Component Analysis (PCA) was used to evaluate overall shifts in pressure distribution profiles.

## Results

Post-intervention, rearfoot pressure metrics significantly decreased (surface area:  $p < 0.01$ ,  $d > 0.8$ ), while midfoot impulse showed a moderate increase ( $p < 0.05$ ,  $d \approx 0.5$ ), indicating enhanced medial arch engagement. Forefoot parameters showed minimal change. PCA revealed a distinct shift in plantar pressure profiles post-intervention, with a clear separation from baseline. The first two components explained 57.2% of the total variance (PC1: 37.1%, PC2: 20.1%). Variables with the highest loading on PC1 included rearfoot impulse and midfoot surface area.



## Discussion

Custom CAD/CAM orthoses produced meaningful redistribution of plantar pressure in individuals with pes planus. The primary biomechanical adaptation—reduction in rearfoot overload and increased midfoot activation—suggests improved medial arch function and enhanced postural stability. PCA results reinforced these findings by identifying a quantifiable pattern shift in pressure distribution. The strong contributions of rearfoot impulse and midfoot contact area to PC1 underscore the role of the orthosis in altering load-bearing dynamics. These findings support the use of personalized, CAD/CAM-manufactured orthoses as an effective intervention for biomechanical realignment in pes planus, and highlight the value of multivariate techniques like PCA in clinical gait analysis.

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## Evolving Trends in Motion Analysis: The Last 5 Years of the ESMAC Congress

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## Introduction

Although numerous motion analysis studies are regularly published in academic journals (1), conferences remain a key platform for highlighting the latest priorities in motion analysis labs (2). The research papers and abstracts presented at the conference serve as an essential resource for scientists, offering insights into emerging trends and the future of the field (3). This study aims to identify and analyze evolving trends in motion analysis research by applying text mining techniques to abstracts presented at ESMAC congresses from 2020 to 2024.

## Research Question

What are the dominant research themes in motion analysis, and how have they evolved at ESMAC congresses from 2020 to 2024?

## Methods

We analyzed the titles of abstracts presented at ESMAC congresses between 2020 and 2024. Using Python, we applied text mining techniques—including TF-IDF vectorization and k-means clustering (with  $K=10$ )—to identify research themes. Trends were analyzed by tracking cluster prevalence over the five-year period.

## Results

A total of 1160 ESMAC abstracts spanning 2020 to 2024 were analyzed, revealing ten dominant research clusters (Fig. 1) that capture the breadth of motion analysis topics presented at the congresses. The word cloud (Fig. 1) highlights the most frequently occurring terms, with “gait,” “child,” “cerebral,” “palsy,” “analysis,” “muscle,” and “kinematic” featuring prominently. Together, these findings underscore a sustained interest in pediatric populations, neurological conditions, and methodological advancements in motion analysis.

Fig.1. Clusters and Word Cloud of ESMAC Titles (2020–2024)



## Discussion

This study provides a comprehensive overview of motion analysis research trends at ESMAC congresses from 2020 to 2024. The analysis revealed a strong and consistent focus on gait and kinematic studies in children with cerebral palsy, alongside a growing interest in advanced technologies such as machine learning and markerless motion capture. While pediatric research, particularly on cerebral palsy and orthotic interventions, remains a cornerstone of the field, the increasing prevalence of studies on adult populations (e.g., osteoarthritis, spinal deformities) and neurological conditions (e.g., Parkinson's disease, multiple sclerosis) highlights the broadening scope of motion analysis. The integration of machine learning and 3D technologies suggests a shift toward more automated and precise methods for clinical assessment, which could enhance the diagnosis and treatment of movement disorders. However, the field continues to prioritize traditional areas like joint kinematics and muscle function, indicating a balance between innovation and foundational research. These insights can guide researchers in identifying emerging priorities and fostering interdisciplinary collaborations in motion analysis.

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## Effects of ankle exoskeleton on biomechanics and oxygen consumption of walking in typically developed children and children with cerebral palsy

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## Introduction

Exoskeleton devices have potential to improve the economy of walking in children with cerebral palsy (CP) by applying assistive torques to the lower limb joints [1,2]. However, previous studies in CP show mixed results regarding their effects on walking, and little is known about whether children with CP adapt differently to exoskeletons compared to typically developed children [3,4]. This study investigates the acute effects of ankle exoskeleton assistance and resistance on the biomechanics and oxygen consumption of walking in typically developed children and children with CP.

## Research Question

Whether changes in biomechanics and oxygen consumption of walking differ when using an exoskeleton to assist or resist ankle movement in typically developed children and children with cerebral palsy?

## Methods

Eleven children with CP (11–18 years old; 9 unilateral CP) and twelve typically developed (TD) children (8–18 years old) participated in this study. Each subject performed three treadmill walking trials: normal walking, exoskeleton-assisted walking, and exoskeleton-resisted walking. The bilateral ankle exoskeleton (Biomotum SPARK) assisted plantarflexion and dorsiflexion, and in resistance mode resisted plantarflexion. Subjects walked at 90% of their comfortable overground walking speed. Oxygen consumption was measured using a wearable metabolic system, and 3D marker-based motion analysis captured lower body gait biomechanics on an instrumented treadmill.

## Results

Exoskeleton assistance increased oxygen consumption (VO<sub>2</sub>) by 7.9% ( $p=0.005$ ) in TD children and decreased it by 2.3% ( $p=0.471$ ) in children with CP. Resistance mode increased VO<sub>2</sub> by 17.2% ( $p=0.003$ ) in TD children and by 2.3% ( $p=0.512$ ) in children with CP (Figure 1.). Ankle dorsiflexion during initial contact tended to increase in both assistance and resistance modes, but the change was significant only in resistance mode on the affected side in children with CP ( $2.71^\circ$ ,  $p=0.034$ ), in assistance mode on the less affected side in children with CP ( $2.16^\circ$ ,  $p=0.023$ ), and in TD children ( $1.77^\circ$ ,  $p=0.013$ ). Peak ankle plantarflexion moment did not change significantly in either assistance or resistance mode. Peak knee extension moment increased significantly on the less affected side in children with CP in both assistance (+0.11 Nm/kg,  $p=0.006$ ) and resistance mode (+0.17 Nm/kg,  $p=0.001$ ).

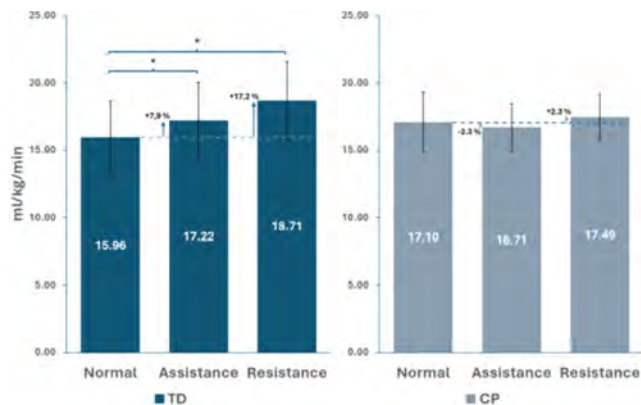


Figure 1. Changes in oxygen consumption during walking using the ankle exoskeleton. Average oxygen consumption of walking (ml/kg/min) is shown for normal, assistance, and resistance conditions. Error bars represent one standard deviation.

## Discussion

The results indicate that TD children and children with CP respond differently to ankle exoskeleton assistance and resistance. In TD children, the exoskeleton increased oxygen consumption, suggesting initial disturbance to their natural gait. Conversely, children with CP showed a decrease in oxygen consumption with assistance, which may imply potential benefits. The increased peak knee extension moments on the less affected side in CP suggest a compensation strategy from ankle to proximal muscle groups. These findings highlight the need for further research to optimize exoskeleton design and training protocols. Additional analysis of joint kinetics and muscle activity will provide further insights into possible distal-to-proximal redistribution of joint mechanics and compensation strategies in different conditions.

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## Kinematic signatures of knee joint reaction load patterns during self-paced level walking in healthy adults

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## Introduction

Understanding knee joint reaction load (JRL) patterns during walking is crucial for developing targeted prevention, rehabilitation, and training strategies. However, the association between visually observable lower limb kinematics and internal joint reaction loads remains unclear.

## Research Question

Do knee JRL patterns have distinct lower limb kinematic signatures during self-paced level walking?

## Methods

We analyzed self-paced level walking of 39 female/32 male healthy participants. 3D ground reaction forces (GRF, 3 AMTI force plates) and kinematic data were recorded using an adapted Cleveland Clinic marker set with a 3D motion analysis system (Vicon Nexus). In OpenSim 4.5, the Lerner/Rajagopal lower body model [1] was scaled [2] and inverse kinematics and static optimization were performed. Medial and lateral tibiofemoral and patellofemoral loads were estimated, normalized to bodyweight (BW), and clustered using soft dynamic time warping time-series k-means ( $k=2$ ) [3]. Bayesian optimization maximized the clustering silhouette score, and statistical difference was assessed with Bonferroni corrected 1D statistical parametric mapping (spm1d) two-sample t-tests alongside Kruskal-Wallis tests for point estimates ( $p_{crit}=0.0028$ ).

## Results

The average scaling RMSE was 0.4 cm ( $SD<0.01$ ), the inverse kinematics RMSE was 0.82 cm (0.16) with a maximum error (ME) of 2.0 cm (0.53), and the static optimization vertical GRF residual was 5.2% BW (1.1) with an ME of 16.4% BW (4.8). The silhouette score was 0.27 ( $\gamma_{opt}=0.26$ ). Cluster A participants had a height of 1.74 m (0.09), an BMI of 23.6 kg/m<sup>2</sup> (2.5), and an age of 34.0 years (13.0) (555 strides of 32 females & 30 males; 50% exercise-experienced). Cluster B participants had a height of 1.73 m (0.09), an BMI of 23.7 kg/m<sup>2</sup> (2.0), and an age of 35.8 years (13.7) (95 strides of 7 females & 2 males; 67% exercise-experienced). Cluster B had 32.0% BW (2.0) lower patellofemoral loads, 40.5% BW (4.8) lower total tibiofemoral loads with a 5.5% (0.7) lateral shift compared to cluster A during loading response (Figure 1). Direct kinematic show that cluster B had 5.7° (0.5) less knee and 3.3° (0.6) less hip flexion and 1.8° (0.4) more ankle plantar flexion.

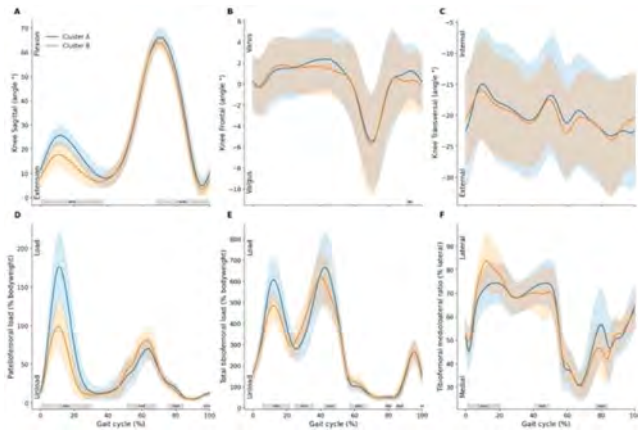


Figure 1: Strides clustered by JRL time series patterns. Top: Direct kinematic knee sagittal (A), frontal (B), and transversal (C) joint angles. Bottom: Patellofemoral (D) and total tibiofemoral (E) resultant JRLs and the mediolateral ratio (F). spm1d differences between clusters are highlighted with gray bars.

## Discussion

Although cluster separation was weak, unsupervised clustering identified knee JRL patterns with sagittal plane direct kinematic angle differences associated with patellofemoral and total tibiofemoral JRLs. Without seeing distinct frontal plane direct kinematic knee angle patterns, we observed JRL differences in the tibiofemoral mediolateral JRL ratio. While the differences in stance phase kinematics were on average  $<6^\circ$ , the interpretation of subtle but potentially clinically relevant knee JRL variations from kinematics seems challenging.

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## Applicability of a 3D hand model to investigate the spastic hand

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## Introduction

Limited functionality of the wrist and fingers affects the execution of activities of daily living (ADLs) in patients with spasticity. 3D motion analysis as an objective approach is rarely used to observe functionality in this population. For clinical applications, biomechanical hand models using one marker per segment are preferred due to reduced patient discomfort, faster marker application and more consistent placement [1]. However, contractures and hyperextension complicate marker application, and accuracy is limited due to skin artifacts. Nevertheless, a hand model can be beneficial for some clinical questions regarding surgical planning, rehabilitation, and further understanding of hand and finger movement patterns.

## Research Question

Can we apply 3D hand models to assess surgical outcomes in spastic hands?

## Methods

The overall study includes 20 healthy adults and 8 patients with upper limb spasticity. Participants completed a test battery of ADLs focusing on hand function. 38 markers were placed on the affected limb of the patient, following the models of Metcalf et al. [1], Carpinella et al. [2], and Jaspers et al. [3], to capture the entire upper limb. Surface electromyography (sEMG) was recorded at the forearm and upper arm. Measurements in the laboratory were conducted preoperatively (Pre) as well as 6 (6post) and 12 months (12post) postoperatively. For this analysis, preliminary kinematic data from the metacarpophalangeal (MCP) joints were examined, as their functionality is essential for fist closure. The 'grasp a baton' task was analyzed in one single patient (28 years old) with traumatic brain injury at three measurement points. Kinematic Data from 10 healthy adults performing the same task and the movement duration of movement repetitions were compared between the patient and healthy controls.

## Results

The movement duration (in seconds) of the ADL grasp baton Pre (13.5) is higher compared to 6post (8.5) and 12post (8.9). At all three measurements, movement duration remained elevated compared to that of healthy adults (mean:  $2.39 \pm 0.36$  s). The range of motion of MCP increased from Pre  $38.7 \pm 0.4^\circ$  to  $46.9^\circ \pm 0.29$  12post (Fig.1). ROM of all MCP joints gradually approached normative data ( $42^\circ \pm 10.7$ ;  $50^\circ \pm 12.5$ ).

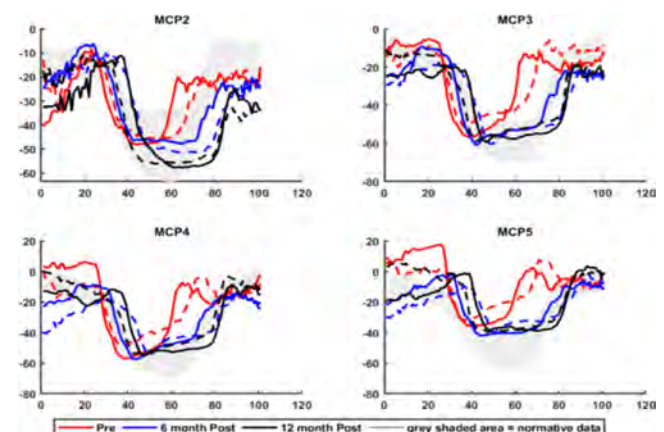


Figure 1: overview of all MCP joint angles

## Discussion

The preliminary results for MCP 2-5 demonstrate that a 3D hand model in patients with spasticity enhances our understanding of functional improvements over time and provides meaningful kinematic data for the joints distal to the wrist. Analyzed joint angles and movement durations show slight improvements 6 and 12 months post-surgery, pointing to functional improvements in the MCP. However, the current model has not yet been tested for reliability. Further research, including additional information such as sEMG data, is essential to develop a more comprehensive understanding of the functional capacity. These objective data are crucial for assessing postoperative outcomes following tendon or nerve transfers in the spastic hand.

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## sEMG to investigate anticipatory, predictive, and reactive strategies of healthy subjects walking on a yielding platform with different a-priori information

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## Introduction

Anticipatory, predictive, and reactive strategies (AS, PS, RS) enable individuals to fine-tune motor responses to perturbations, integrating previous experiences with current tasks and unexpected environmental demands [1]. According to Feldman's theory [2], the modulation of muscle stretch reflex thresholds is a key neurophysiological mechanism underlying these strategies and should result in different muscle activation timing and amplitude among AS, PS, and RS.

## Research Question

Is it possible to differentiate between AS, PS, and RS in the sEMG activity of the leg muscles while walking on a movable platform, both with and without knowledge of future platform behavior?

## Methods

Healthy participants repeatedly walked at a self-selected pace along a linear pathway (10m), provided with an embedded yielding platform that lowers proportionally to the subject leg length and weight [3] (Figure 1a). Trials were randomly sequenced to induce both expected and unexpected platform displacements: baseline walking (B), expected-displacing (ED), expected-nondisplacing (END), unexpected-displacing (UD), unexpected-nondisplacing (UND), confounding-displacing (CD), confounding-nondisplacing (CND).

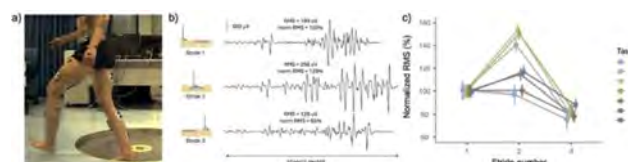
Full body kinematics and sEMG data (SENIAM protocol) from TA, SOL, and GM were collected. For each trial, three gait strides were analyzed: (#1) before the platform, (#2) on the platform, and (#3) after the platform. RMS amplitude was computed for stance and swing phases and normalized to the RMS value of step #1 (Figure 1b) for all trials to allow for between-subject comparisons [4].

Mixed Model Analysis was carried out on RMS amplitudes, using tasks and strides as fixed factors and subjects as random intercepts. Results from the stance phase are presented below.

## Results

Ten subjects (3/7 M/F), mean age 31 (7), were included. SOL normalized RMS amplitude significantly increased by about 50% in all trials involving platform movement, compared to B ( $p < 0.001$ ), increased by about 20% in the UND and CND tasks ( $p < 0.05$ ), while no difference was found in the END condition, where participants were informed that the platform would remain stable ( $p = 0.70$ ) (Figure 1c). Amplitude reduction in stride #3 is due to the subjects' deceleration at the end of the walkway. At GM, amplitude increased by about 30% ( $p < 0.01$ ) only in the trials involving platform movement. TA activity increased by about 50% only in the UD condition ( $p < 0.001$ ) and by about 30% ( $p < 0.01$ ) in the trials when the platform was moved as expected and also in the one when the platform did not move, despite expected.

**Figure 1:** a) Yielding platform; b) Soleus activity during stance (UND task); c) Plot of the mixed-model analysis of soleus RMS in stance (estimated marginal means and standard errors are shown)



## Discussion

Mechanical perturbations and a priori information both affected, at a different level, the activity of the three calf muscles presented in this study. sEMG could be a valuable tool for investigating AS, PS, and RS in research protocols.

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## Implementing Motor Control Theory in Robotic Orthosis Design for Children's Gait Recovery

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## Introduction

Introduction Gait is a dynamic function influenced by factors such as objectives, terrain characteristics, and walking speed. Robotic solutions addressing gait dynamics in children have been proposed through both end-effector devices and wearable exoskeletons, although their effectiveness remains uncertain [1]. To enhance understanding of sensorimotor integration during locomotion and to improve gait rehabilitation, we developed a robotic knee orthosis [2].

## Research Question

Our objective was to test a control hypothesis concerning locomotor function and to clarify the relationship between force interactions and the information they convey. Additionally, the developed controller aimed to target critical temporal windows identified in biological gait development across the lifespan.

## Methods

The orthosis controller was designed based on Feldman's hypothesis about the reference frame of body segments [2]. Two inertial sensors measured leg and thigh posture, an encoder recorded rotation angles, and actuator currents determined and measured exerted torque. Ten young adults walked on a treadmill at a comfortable speed under three conditions: natural gait without interaction (transparent mode), induced increases or decreases of 20% in maximum target flexion angles, and interactions at low and high torques (2 and 4 Nm). Torque and posture control prevented motor activation until the knee transitioned from initial to target posture, reflecting known muscle activation and CNS roles during gait control [2]. We analysed 20 gait cycles per condition.

## Results

Results We observed a significant increase in knee flexion angle compared to free walking, rising from  $53^\circ \pm 9^\circ$  to  $60^\circ \pm 11^\circ$  and  $62^\circ \pm 13^\circ$  under lower and higher torque conditions, respectively. However, the differences between the two torque levels were not statistically significant. Conversely, applying torques aimed at

reducing knee flexion consistently yielded significant reductions to  $43^\circ \pm 8^\circ$ . Increased knee flexion was associated with higher rotational speeds and increased actuator current absorption (pushing) during the initial flexion phase. In contrast, reduced knee flexion involved actuator current generation (braking) throughout the flexion phase, accompanied by decreased rotational speeds (Figure 1a, 1b). These findings led to the development of a new wearable exoskeleton prototype, suitable for use starting from two years of age. This device will be evaluated for improving gait rehabilitation by initially adopting a transparent mode—following the pathological gait pattern—to facilitate gait reorganization (Figure 1c).

## Discussion

Applying different torque magnitudes yielded similar results, suggesting that gait modification depends on the information embedded within force interactions, not merely force magnitude. This indicates that the CNS automatically adapts to context modifications guided by implicit information. These findings have implications for developing rehabilitation devices and physiotherapy practices during life span, emphasizing functional training specificity and the dynamics inherent in designed interventions.

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## Free-living daily physical activity levels in children with cerebral palsy: an EMG and accelerometry study

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## Introduction

Cerebral palsy (CP) is the most common childhood disability in developed nations (1), leading to reduced physical activity (PA) and increased health risks (2). To complement traditional PA assessment methods (e.g. accelerometers and self-reports), electromyography (EMG) provides physiologically relevant information on muscle activity during free-living (3). This study used shorts with embedded EMG electrodes for assessing daily muscle inactivity and activity in individuals with CP and their typically developing (TD) peers.

## Research Question

How do daily EMG activity levels differ between individuals with CP and their TD peers? Do daily activity levels assessed by EMG an

accelerometry differ from each other? Are there differences in EMG activity between thigh muscles (hamstrings vs. quadriceps femoris) and between the most affected and least affected legs? What is the influence of Gross Motor Function Classification System (GMFCS) levels on muscle activity patterns?

## Methods

Shorts with embedded EMG electrodes and waist-worn tri-axial accelerometer recorded daily PA in 8 children with spastic CP (mean age 14y 7mo, Gross Motor Function Classification System (GMFCS) I (n=5), III (n=3)) and 6 TD children (mean age 15y 4mo) during free-living. Sedentary time and light, moderate and vigorous PA were reported relative to recording time, using established cut-off values for accelerometry, and EMG amplitude categorized based on a two-minute average during the 6-minute walk test (6MWT). Mann-Whitney U-tests were used to compare group differences (CP vs. TD, GMFCS I vs. III, the most affected vs. the least affected leg in CP, quadriceps femoris vs. hamstrings). Significance level was set at  $P < 0.05$ .

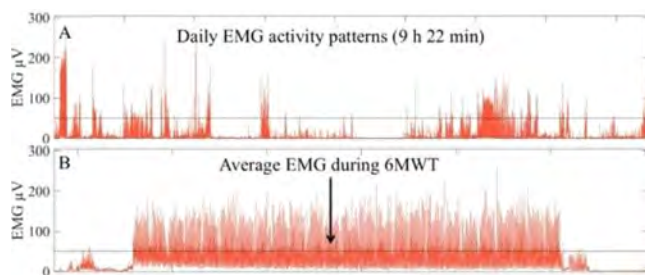


Figure 1: Example of baseline corrected, average daily EMG for right leg from one participant with CP (A) and during 6MWT (B). The average EMG during 6MWT (50.5  $\mu$ V, horizontal lines) was used to normalize the daily EMG.

## Results

Daily EMG inactivity (CP: 58.4%, TD: 50.4%) and activity levels did not differ between CP and TD groups. Accelerometry provided greater sedentary time than EMG in both CP ( $p=0.021$ ) and TD ( $p=0.010$ ). Light activity in the hamstrings was greater in the less affected leg compared to the most affected leg ( $p=0.046$ ) and there was more moderate activity in the quadriceps femoris in GMFCS level I compared to level III ( $p=0.025$ ).

## Discussion

Individuals with CP are not significantly more inactive compared to their TD peers. EMG detects light PA that accelerometry may underestimate, offering a more detailed view of daily muscle use, which can inform interventions to improve PA levels in CP. In CP, only incidental findings between legs, muscles, and GMFCS levels were observed. The detailed insights on muscle activity can inform the development of targeted interventions to improve PA levels, function and overall health outcomes in individuals with CP.

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## 4D External Body Scanning: a new tool for functional analysis of scoliosis.

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## Introduction

Scoliosis is one the most common spinal disorders and can lead to abnormal functionality, which affect gait and other activities. Besides, literature has been demonstrating that patients with scoliosis can experience complex biomechanical changes.

Traditionally, radiology has been the gold standard for spine assessment, but it can be harmful due its radiation. Furthermore, the static assessment of spine deformities is sometimes not enough to predict functional performance, so a combination of static and dynamic assessments should be considered for a full assessment.

Current methods for static and dynamic assessment of spine kinematics, based on motion capture and 3D body surface analysis, struggle with inherent constraints regarding the manual detection of landmarks or limitations on motion plane direction, which can affect their reliability, validity and usability. 4D scanning technology may face the challenge of achieving accurate information to overcome those limitations.

## Research Question

Can 4D external body scanning provide useful metrics to improve the accuracy of scoliosis functional interpretation?

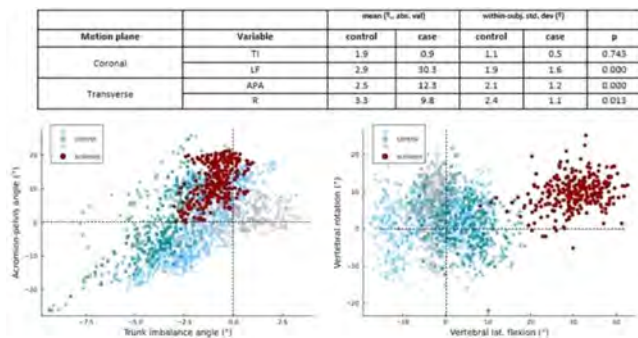
## Methods

3 healthy participants and 1 patient with scoliosis were scanned with Move4D scanning system during level gait [1]. Midline of spine and positions and orientations of the vertebrae from C7 to L5 were estimated at each frame through the topological analysis of the back surface [2-5]. Trunk imbalance (TI) and Acromion-pelvis angles (APA) were calculated for frontal and transverse plane analysis respectively, using homologous anatomical points of the processed mesh. Additionally, Vertebral lateral flexion (LF) and rotation (R) were also analyzed [6] tracking anatomical landmarks on the pelvis, neck and shoulders that were obtained in the process of spine identification [2,3]. The mean value and within-subjects variability of each variable were estimated using a mixed linear model. The values obtained for the case with scoliosis were compared to those of the control group.

## Results

There were asymmetries and differences in the ranges of motion between the case with scoliosis and controls. The variables based on gait parameters (TI, APA) were correlated to each other (Picture 1), and APA was asymmetrically distributed in the case with scoliosis. The underlying vertebral rotations (LF, R) were uncorrelated, and differentiated the case from the controls in both planes. Within-subjects variability was smaller for the case than for the controls in all variables.

Picture 1. Distribution of variables measured in the study



## Discussion

Alterations observed in gait parameters are the result of accumulated abnormal rotations of the spine, which when analysed in detail can differentiate better a case with scoliosis from healthy people, and explain the mechanical cause of those alterations. 4D scanning may allow to carry out static and dynamic studies in order to evaluate adaptations and compensations in patients with scoliosis. 4D scanning brings accuracy detecting functional differences, providing automated processing techniques and analyzing free planes motion facing off current constraints.

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## From metrics to meaning: Exploring clinical perspectives on upper limb kinematics in neurological rehabilitation

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## Introduction

The integration of movement measurement technologies into clinical practice has the potential to transform healthcare delivery by enabling objective assessment of motor function (1). Electronic reports derived from kinematic analyses and real-life behaviour monitoring offer valuable biomarkers for tracking disease progression and evaluating treatment response.(2) While these technologies are well-established in research settings, the perspectives of clinical professionals on the interpretability and clinical utility of kinematic parameters remain underexplored.(3) As part of the broader IMPACT study on digital health integration, this abstract presents preliminary findings focused specifically on upper limb kinematic parameters.

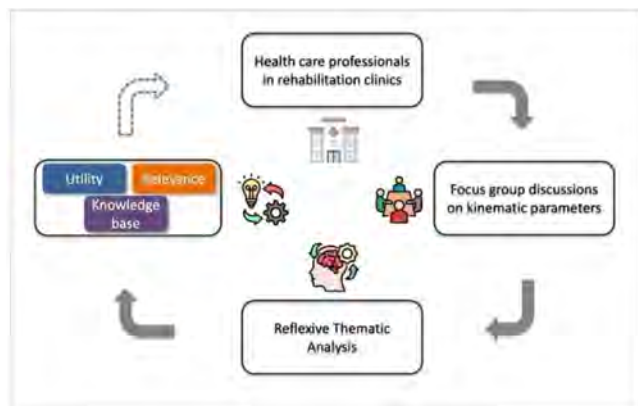
## Research Question

This sub-study investigates clinicians' perspectives on the meaning, relevance, and interpretability of kinematic parameters related to the upper limbs' movement quality and activity performance. The aim is to support knowledge translation strategies that facilitate the clinical integration of movement analysis tools in neurological rehabilitation(4).

## Methods

Healthcare professionals from three leading Swiss rehabilitation centres participated in expert-led focus group discussions to share their views on upper limb kinematic parameters. During semi-structured interviews, participants were asked to identify and justify movement constructs they considered relevant for neurological rehabilitation. Established upper limb kinematic parameters—quantifying movement quality(5) and real-life performance(6)—were introduced, discussed, and prioritised. Audio recordings were transcribed, coded, and analysed using reflexive thematic analysis(7). A mixed-methods design allowed for comparison between qualitative themes and prioritisation data.

Figure 1. Workflow of user-centered integration of kinematic parameters.



## Results

Three focus group discussions were conducted, involving ten healthcare professionals: physiotherapists (n=4), occupational therapists (n=3), clinical movement scientists (n=2), and one medical doctor. Only one group had routine experience with upper limb kinematic assessments or parameter interpretation. Across all discussions, participants expressed limited knowledge and confidence in interpreting upper limb kinematic data. Nonetheless, these parameters were perceived as a valuable complement to standard clinical assessments. Movement quality metrics were considered promising indicators of real-world motor behaviour, including inactivity and compensatory strategies. Most participants expressed a generally positive attitude toward incorporating upper limb movement analysis into rehabilitation practice.

## Discussion

These preliminary findings offer early insights into healthcare professionals' views on the clinical utility of upper limb kinematic parameters. The results highlight a substantial knowledge gap in interpreting established movement biomarkers, underscoring the need for targeted educational initiatives (Figure 1). Positive attitudes among clinicians toward movement analysis may support user-centred implementation strategies for integrating upper limb kinematic assessments into routine rehabilitation practice.(4,8)

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## Exploring morphological variations of the subtalar joint in children with cerebral palsy – a statistical approach

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## Introduction

The development of foot deformities in children with cerebral palsy (CP) remains poorly understood. Although often associated with abnormal joint orientation, soft tissue pathology, and muscle imbalances [1, 2], the contribution of bone morphology is largely unexplored. Identifying subtle variations could help improve our understanding of the foot's functional anatomy, relating key discriminating features to altered loading patterns and joint instabilities [3]. Traditionally two-dimensional radiographs have been the primary tool to assess bone morphology and identify bony landmarks. For bones in the hindfoot which lack a distinguishable long axis however, planar views fail to represent anatomical complexity of the region. This makes it difficult for clinicians to understand and accurately assess secondary foot deformities.

## Research Question

Do neuromuscular abnormalities in CP affect the morphology of developing subtalar joint bones?

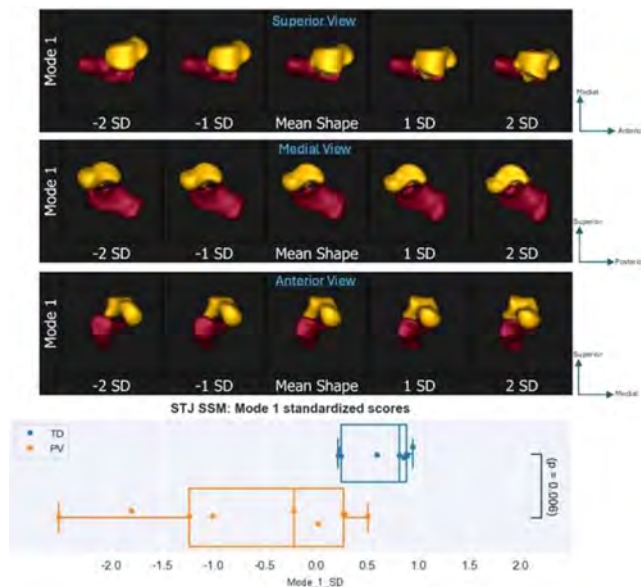
## Methods

This study applied statistical shape modelling (SSM) to explore morphological variation in the talus and calcaneus bones of the subtalar joint (STJ). SSM is an emerging method which can characterize and quantify bone morphology, with the benefit of encompassing the full 3D structure. Weightbearing-bearing CT scans were used to build SSMs from a population of 9 typically developing (TD) controls (age: 13.1 ± 2.3 years) and 9 CP subjects (N=9, age: 10.8 ± 2.9), all with planovalgus foot deformity. Further analysis

was then performed to calculate specific joint measurements including joint coverage, space, and facet maturity.

## Results

For each model built the dominant mode of variation (mode 1) statistically discriminated between groups ( $p=0.017$  for the talus,  $p=0.008$  for the calcaneus, and  $p=0.006$  for the combined STJ model). Noteworthy morphological differences in CP include flatter articulating surfaces in both bones, poorly defined facet boundaries, lateral growth of the talar neck, and a thinner less pronounced sustentaculum tali (medial shelf of calcaneus). In the combined talo-calcaneal model there were also alignment variations, with the talus rotating internally relative to the calcaneus. This causes the inferior surface of the talus to slide beyond supporting regions and allows joint space to open up. Inferior talus coverage was quantified as 53.89% and 45.09% for the typically developing and CP cohorts respectively. A consistent trend across all data in this study was a tendency for more severe planovalgus foot deformities to have scores which deviated further from the control population.



## Discussion

These findings enhance our understanding of foot deformity development in CP. The less complex morphological structures observed in CP may diminish the mechanical constraints required to support end range of motion, thus contributing to STJ instability. This may help explain inconsistent treatment outcomes reported in CP and warrants further discussion [4]. The described SSM approach also shows promise as a clinical evaluation tool, with shape metrics offering a measure of severity. This could ultimately inform better treatment decisions and aid surgical planning.

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## Determining the minimal level of detectable change (MLDC) in normalised muscle tendon lengths (nMTL) and velocities (nMTV) from reliability studies

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## Introduction

Musculoskeletal (MSK) modelling outputs, nMTL and nMTV, have potential clinical use in treatment planning and evaluation. Although input kinematic data reliability is documented (Kainz et al., 2017), assessments of nMTL and nMTV appear unpublished, likely due to dependency on selected MSK models and processing parameters. Reliability data for nMTL and nMTV (derived using specific MSK models and approaches) would provide a baseline for evaluating current outputs and alternative modelling strategies.

## Research Question

What is the minimal level of detectable change (MLDC) in nMTL and nMTV outputs across the gait cycle, as determined by reliability studies?

## Methods

Within-session standard deviation (WSsd) and Standard Error of Measurement (SEM) were used to evaluate reliability, representing inherent subject variability and procedural error, respectively. These metrics, and the subsequent derivation of 95% confidence limits as an indicator of MLDC, were computed using Baker's (2016) methods.

Retrospective 3D gait analysis data was used from three typical adults, measured by three assessors, across two sessions, five trials per session. Marker trajectory data from Vicon Nexus was converted, prior to processing within OpenSim v4.4. Outputs were generated using the generic RajagopalLaiUhlrich2023.osim MSK model, modified based on the approach of Kainz et al. (2017) to align kinematic outputs (polarity, order, and offset of pelvic rotations; 3 degrees of freedom at the knee). Pelvic width, height and joint centres were utilised for model scaling. MTL and MTV outputs were normalised to anatomical neutral position and peak velocity from age-matched typical cohort, respectively.

Unlike marker placement error, inherent subject variability cannot be assumed comparable across typical and pathological cohorts. As such WSsdCYP was also derived from 3 intra-session trials of 20 children and young people (CYP) both typically developing (TD  $n=10$ ) and with diagnoses of Cerebral Palsy (CP  $n=10$ ). A conservative estimation of MLDC was determined from the square

root of the sum of squares of the WSsdCYP and SEM, utilising a study design-based multiplying factor of 1.3.

## Results

Table 1. SEM and WSsd from a reliability study, WSsd from a mixed cohort of CYP, calculated MLDC for nMTL and nMTV

|                          | nMTL  |                     |                     |       | nMTV  |                     |                     |       |
|--------------------------|-------|---------------------|---------------------|-------|-------|---------------------|---------------------|-------|
|                          | SEM   | WSsd <sub>rel</sub> | WSsd <sub>cyp</sub> | MLDC  | SEM   | WSsd <sub>rel</sub> | WSsd <sub>cyp</sub> | MLDC  |
| Adductor Brevis          | 0.019 | 0.012               | 0.020               | 0.036 | 0.077 | 0.111               | 0.184               | 0.260 |
| Adductor Magnus (distal) | 0.008 | 0.004               | 0.009               | 0.016 | 0.199 | 0.182               | 0.327               | 0.498 |
| Gracilis                 | 0.007 | 0.005               | 0.008               | 0.014 | 0.097 | 0.133               | 0.172               | 0.257 |
| Psoas                    | 0.006 | 0.007               | 0.006               | 0.011 | 0.068 | 0.085               | 0.151               | 0.215 |
| Rectus Femoris           | 0.004 | 0.004               | 0.005               | 0.008 | 0.055 | 0.085               | 0.130               | 0.183 |
| Semi-membranosus         | 0.005 | 0.004               | 0.006               | 0.010 | 0.045 | 0.051               | 0.081               | 0.120 |
| Medial Gastrocnemius     | 0.006 | 0.004               | 0.005               | 0.011 | 0.098 | 0.140               | 0.212               | 0.304 |
| Soleus                   | 0.006 | 0.005               | 0.008               | 0.013 | 0.062 | 0.101               | 0.159               | 0.221 |
| Tibialis Anterior        | 0.006 | 0.005               | 0.007               | 0.012 | 0.057 | 0.089               | 0.136               | 0.192 |
| Tibialis Posterior       | 0.003 | 0.003               | 0.003               | 0.006 | 0.088 | 0.150               | 0.198               | 0.282 |

## Discussion

For all nMTL outputs, WSsd and SEM were similar, indicating comparable impacts of subject variability and procedural error. For nMTV outputs, WSsd exceeded SEM, suggesting subject variability affects nMTV more than procedural error. The derived MLDC offers a threshold for assessing meaningful changes in nMTL and nMTV during the gait cycle, based on the described models and methods.

Clinical MSK modelling is evolving rapidly, with advancements in generic models, personalisation tools, and methods to minimise inverse kinematic errors prior to generating outputs. While the models and processes used in this study could be further optimised, the presented MLDC provide pragmatic thresholds for evaluating current outputs and a baseline for assessing future models and methods.

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### Relationship between pelvic tilt and 3D acetabular orientation in patients with adult spinal deformity: the role of pelvic morphology

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## Introduction

Patients with Adult Spinal Deformity (ASD) present with postural malalignment mainly compensated through pelvic retroversion, radiographically measured by pelvic tilt (PT) [1,2]. Some patients have concurrent degenerative changes in their hips described as spine-hip syndrome. ASD undergoing total hip replacement (THR) have a higher risk of THR dislocation than the general population, due to their spino-pelvic malalignment and changes that occur after spinal realignment surgery [3]. Previous studies suggested that acetabular anteversion increases linearly with PT: 1° increase in PT leading to 0.7 to 0.9° increase in acetabular anteversion [4]. However, those studies assumed linearity between these positional parameters without accounting for patient-specific pelvic morphology.

## Research Question

What are the 3D geometric relationships between positional acetabular and morphologic pelvic parameters?

## Methods

A mathematical model was first established as follows. Parameters of interest (Figure 1a) included the acetabular morphological angle (AMA), measured between the acetabular axis and the hip axis, and the sagittal acetabular inclination (SAI), measured between the acetabular axis projected on the sagittal plane and the horizontal axis. While AMA is a morphological patient-specific angle, SAI increases incrementally with PT (1° increase in SAI corresponds to 1° increase in PT). The 3D geometrical relationships between AMA, SAI, and acetabular parameters (abduction and anteversion) were first established and then graphically represented (Figure 1b). The previous relationships were validated in 97 ASD patients who underwent spino-pelvic realignment surgery. Pelvic and acetabular parameters were calculated on 3D reconstructions obtained from biplanar X-rays. Examples were presented in Figure 1c.

## Results

As illustrated in Figure 1b, acetabular abduction and anteversion can be directly calculated from the AMA and SAI angles. The associated graphical representation demonstrates that acetabular parameters exhibit a non-linear relationship with SAI and is impacted by the acetabular morphology (AMA). Specifically, for a given decrease in pelvic retroversion (SAI), changes in abduction and anteversion vary significantly depending on pelvic morphology (AMA). Case examples (Figure 1c) highlight this non-linearity: two ASD patients with similar SAI variations (i.e. similar decrease in pelvic retroversion of 16°) exhibited markedly different changes in acetabular abduction (-14° for case-1 vs. -3° for case-2) and anteversion (-10° vs. -2°). Errors between theoretical and radiological acetabular abduction and anteversion did not exceed 5°.

## Discussion

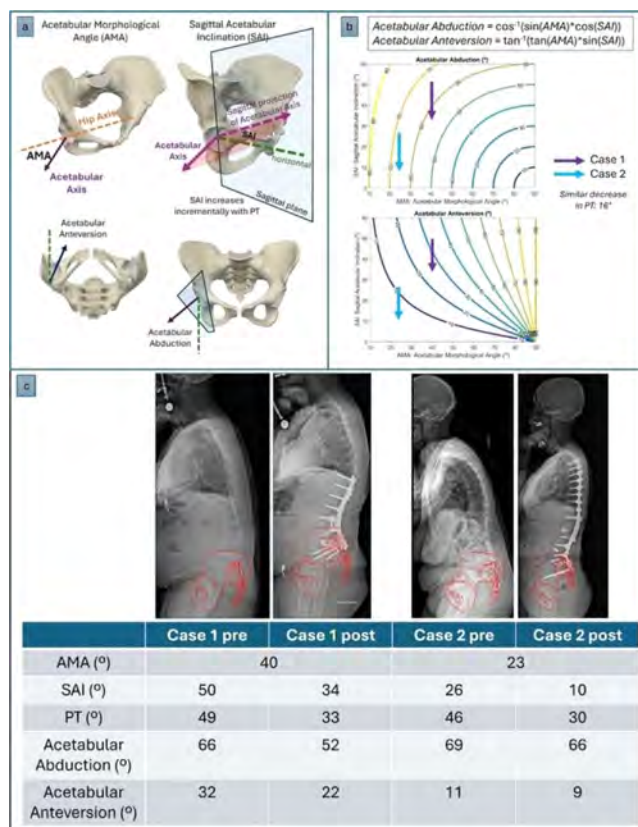
This study establishes that 3D pelvic morphology influences changes in acetabular abduction and anteversion following spinal realignment, resulting in a non-linear relationship between PT and

acetabular orientation. Recognizing this interplay is essential for spine and hip surgeons who can use this knowledge to anticipate postoperative shifts in cup orientation and establish patient-specific hip "safe zones" to minimize THR dislocation risk in ASD patients.

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Figure 1: 1a) Definition of acetabular and pelvic angles;1b) Non-linear transformation of acetabular abduction and anteversion with SAI (PT);1c) examples of post-operative changes.



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## Prediction of crutch-assisted gait using data from instrumentalized crutches: towards the generation of synchronization signals for personalized exoskeleton control

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## Introduction

Wearable lower-limb exoskeletons are used for gait training in subjects with neuromuscular impairments. However, most cannot provide enough stability for patients to walk, therefore, assistive devices such as crutches, walkers and canes are normally used for support [1]. This offers an opportunity for adaptive control using real-time measurements from instrumentalized crutches, which could be used to generate synchronization signals [2,3,4]. To generate them, knowing the point of the cycle and the pattern performed is required, which is done off-line in this proof-of-concept.

## Research Question

Is it feasible to reconstruct the crutch-assisted movement using a musculoskeletal model, based solely on crutch-ground contact forces measurements?

## Methods

Data of healthy subjects performing four different types of crutch-assisted gait patterns were collected prior to this work [5]. Crutch-ground reaction forces (CRFs) and orientations of both crutches were collected thrice for each pattern. Two subjects were selected to perform predictive simulations of crutch-assisted gait by optimising performance (e.g., minimizing metabolic cost) using a three-dimensional musculoskeletal model [6]. Simulations did not rely on movement data, except for CRFs, which were tracked by minimising the difference between measured and simulated magnitudes. To evaluate prediction accuracy, two metrics were used: a qualitative comparison between predicted and reference gait patterns, and the cross-correlation coefficient between experimental and predicted crutch sagittal angles.

## Results

Realistic simulations of assisted gait were achieved using the model with crutches. When adding the tracking, simulated CRFs accurately matched the references. However, the predicted gait patterns did not seem to reliably reenact the reference patterns, although they majoritarily displayed similar behaviors: the averaged cross-correlation coefficient for the crutches' sagittal orientation for all the trials was of 0.684 (Fig 1.a). Half of the predicted gaits were considered non-physiological, however, qualitative analysis showed that in 9 out of 12 trials, the simulated gait patterns would be classified as the same pattern as the reference (Fig 1.b).

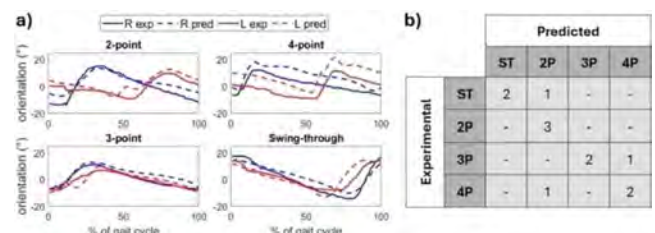


Figure 1. a) Experimental (exp) and predicted (pred) right (R) and left (L) crutch sagittal angles, for four different trials of different patterns. b) Number of trials that converge to each pattern according to the reference pattern.

## Discussion

Although half of the simulated gaits were considered non-physiological, the framework seems capable of identifying the reference gait pattern. To improve results, a subject-specific model should be used instead of a generic one. More predictions with different reference CRFs for all four patterns should be done, to assess repeatability. Besides, collecting additional experimental data, using a setup that captures the exact movement of the subjects, and comparing with the simulation results, would allow better validation. These results may allow the development of algorithms for generating high-quality synchronization signals and, in the future, real-time reconstruction, both resulting in an improvement in personalized exoskeleton control.

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<https://doi.org/10.1016/j.gaitpost.2025.07.202>

## Do patients with midfoot Charcot deformity have better walking efficiency than patients with Ankle/Hindfoot Charcot deformity?

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## Introduction to clinical case

Plantar Pressures and Gait characterisation in patients suffering with progressive deformity secondary to Charcot Arthropathy is an area less explored in literature. The aim of this study is study the plantar pressure and gait characterisation in patients with progressive Charcot deformity of midfoot vs. hind foot/ ankle.

## Methods

This was a single centre case series. Patient who suffered with progressive Charcot deformity and ankle / midfoot collapse underwent a structured biomechanical assessment. Plantar pressure analysis were carried out in standing and walking using Matscan (TekscanTM Inc., USA) walkway system. The standing assessment consisted of 30 seconds, walking recording included 2 x 3 meters walking trial at patient's self-selected speed.

## Results

6 patients who presented with midfoot collapse (Group A), 3 patients with ankle/hindfoot collapse (Group B) were included. There was reduced loading of the affected foot (44.8% vs. 54.5%) when compared to opposite foot (55.1% vs. 74%) for Group A and B respectively (p=0.011 and p=0.016). Although overall loading of the affected foot was reduced on the affected side for both groups, Group A showed similar peak pressure recordings (361 KPa vs. 361 KPa) at region of deformity when compared to Group B (184 KPa affected foot vs. 219KPa non-affected side). Group A showed increased average cadence (66.5 steps/mt vs. 49.4 steps/mt) and walking velocity (0.1 m/s vs. 0.4 m/s) when compared to Group B.

## Conclusion and clinical relevance

Plantar pressure measurements along with spatial temporal parameters of gait has highlighted the trend that although patients with charcot midfoot collapse demonstrate peak plantar pressures at the midfoot area, they demonstrate better walking efficiency in terms of cadence and walking velocity when compared to patients with charcot hindfoot / ankle collapse. Further studies including 3D kinematics and centre of mass measurements is required to explain the difference in pathomechanics between the groups.

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## Radiographic assessment of shoulder imbalance and its correlation with gait patterns

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### Introduction

Adolescent idiopathic scoliosis (AIS) the most common three-dimensional spinal deformity in adolescents AIS patients often present with shoulder imbalance, a common cause for consultation [1]. More recently, it was shown that AIS patients present kinematic impairment during gait at both the shoulders' level and the scoliotic segments that are compensated in the adjacent spinal segments, the pelvis or the lower limbs [2]. The aim was to evaluate radiographic shoulder imbalance in AIS and its relationship to radiographic parameters, QOL scores and gait kinematics.

### Research Question

Is radiographic shoulder imbalance associated with functional gait impairment in AIS?

### Methods

64 AIS (N=42 Lenke-1 with a right convexity of major thoracic scoliosis, N=22 Lenke-5 with left convexity of major thoracolumbar/lumbar scoliosis, average Cobb:36° ± 23° resp.) and 23 healthy subjects (with similar age & height) filled the SRS22r questionnaire. 3D gait analysis was performed to calculate full-body joint and spine segment kinematics during walking [3,4]. All participants underwent low-dose biplanar X-rays from which the classic 3D spinopelvic and scoliosis parameters were calculated. The acromial angle was measured on the frontal Xray between the bi-acromial line and the horizontal. Nine operators measured the acromial angle twice on a subset of 20 AIS patients in order to evaluate its reliability by calculating the intraclass correlation coefficient (ICC) and the confidence interval at 95% (CI95%). The acromial angle was compared between groups and its correlation with the scoliotic parameters, 3D gait kinematics and QOL scores were evaluated.

### Results

The acromial angle showed to be a highly reliable parameter (ICC=0.99 & CI95%=0.6°). Lenke-1 patients had elevated right shoulders, while Lenke-5 patients had elevated left shoulders (acromial angle:  $-1 \pm 2^\circ$  vs  $1 \pm 2^\circ$  resp.,  $p=0.02$ ). AIS patients had several gait impairments among which, an increased shoulder lateral bending (Lenke-1:  $-2^\circ$ , Lenke-5:  $3^\circ$ , controls:  $0^\circ$ ) and axial rotation (Lenke-1:  $6^\circ$ , Lenke-5:  $2^\circ$ , controls:  $0^\circ$ ) as well as a thoracic hypokyphosis during walking (C7T9-T9T12 Lenke-1:  $20^\circ$ , Lenke-5:  $22^\circ$ , controls:  $27^\circ$ ; all  $p<0.05$ ). The radiographic acromial angle was correlated to the Cobb angle ( $r=-0.30$ ), SRS22r pain score ( $r=-0.25$ ), the shoulder lateral bending ( $r=0.48$ , Fig.1) and axial rotation during gait ( $r=0.38$ , all  $p<0.05$ ).

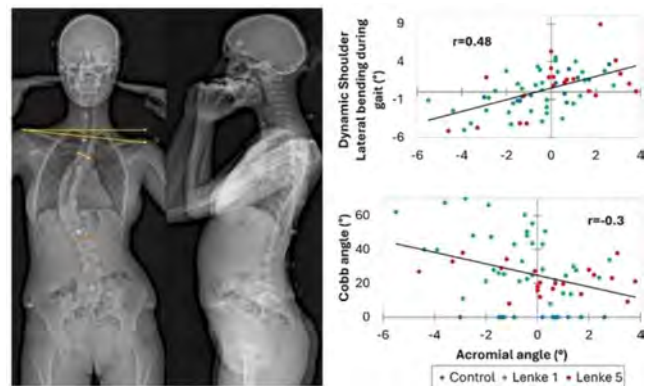
### Discussion

Shoulder imbalance in patients with AIS is critical for preoperative assessment and surgical planning, as it may contribute to suboptimal outcomes after spinal fusion. Beyond its radiological and clinical significance, this study demonstrates that shoulder imbalance is associated with gait functional impairment. Therefore, it is important to consider these associations in the overall evaluation of AIS patients.

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Fig.1- Shoulder imbalance in AIS and its relationship with the Cobb angle and the dynamic shoulder bending during gait.



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### A comparison of different methods of centre of mass extraction across various walking tasks

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### Introduction

Reduced order models provide a simplified but mechanistically insightful approach to understanding how human movement is maintained across various behaviours. In its simplest form, the

centre of mass (CoM) is represented by a point mass attached to two mass-less legs, virtual legs. Movement stability is shown to be enhanced by simple adjustments among virtual leg parameters (e. g. CoM position, leg angle, and leg length) (1,2). Hence, accurately estimating CoM position is vital in order to apply reduced order models on experimental data.

## Research Question

Despite the importance of CoM in human movement analysis, it is not clear how accurate simplified marker sets are for extracting virtual leg parameters across various walking tasks. Here, we aimed to compare various methods of CoM extraction, during different walking tasks, to evaluate their effect on reduced order model parameters.

## Methods

Healthy adults ( $n = 26$ ; age  $27.38 \pm 5.1$ ) performed walking at self-selected speed, fast walking speed, ascending/descending stairs, and stepping on-and-over a step. CoM was estimated using the marker sets, Visual3D (control, 42 markers; 8 segments), the DeLeva 13-segmental model (18 markers; 13 segments) (3), a sacral marker (4), a mid-PSIS marker (5), and PSISil (average of four pelvic markers; 1 segment) (6). Repeated measure ANOVA was used to evaluate the difference of CoM position estimated for each marker set compared to Visual3D, in CoM and virtual leg parameters.

## Results

CoM position at initial contact (IC) and foot-off (FO) in medio-lateral (ML) direction estimated by PSISil marker set did not significantly differ from Visual3D for all the tasks ( $p > 0.05$ ). In the anterior-posterior (AP) direction at FO the difference between those marker sets was not significant ( $p > 0.05$ ) but was at IC ( $p < 0.05$ ). In the vertical direction Visual3D marker set differed significantly from the other 4 marker sets ( $p < 0.05$ ). Angle between CoM velocity vector and virtual leg () was not significantly different between PSISil and Visual3D at IC and FO ( $p = 0.49$ ) for walking and fast walking, but variability increased by task complexity (Coefficient of Variation increased from 0.13 (walking) to 0.24 (ascending stair)).

## Discussion

CoM positions varied significantly dependent on marker set, with PSISil demonstrating the smallest difference to Visual3D across all tasks. DeLeva has known issues with joint centre estimations, segment length and mass distribution as they are based on cadaver anthropometric data (3). Sacral and midPSIS methods neglects CoM movement relative to the pelvis and pelvic tilt (7). In conclusion, as PSISil minimises occlusion issues and appears to approximate the body's CoM, so it could be used as a simplified representation (6), reducing the burden of some marker sets.

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## Pain-driven gait alterations: the role of the movement deviation profile in evaluating joint kinematics and pain intensity in patellofemoral pain

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## Introduction

Individuals with patellofemoral pain (PFP) exhibit alterations in their biomechanical gait. However, a recent systematic review indicates that the evidence supporting these findings is poor to moderate quality [1]. A previous study has established a strong relationship between pain intensity and deviations from typical gait patterns, as measured by the Movement Deviation Profile (MDP) [2]. Specifically, higher pain levels are associated with gait deviations, which may arise from compensatory or primary movement alterations. Nonetheless, it remains unclear whether gait deviations vary based on the severity of pain or whether different movement patterns are associated with milder versus more severe pain.

## Research Question

Does the Movement Deviation Profile (MDP) identify and is associated with gait kinematic differences among women with mild, moderate, and severe patellofemoral pain? Which movements significantly contribute to the identified deviations?

## Methods

A retrospective chart review was conducted to evaluate 237 women with PFP (age:  $25.84 \pm 5.3$  years, height:  $1.67 \pm 0.6$  m, mass:  $62.1 \pm 3.4$  kg) who underwent a three-dimensional gait analysis between 2012 and 2024. Participants were divided into three groups ( $n = 79$ ) according to reported pain intensity in the Visual Analogue Scale (VAS 0-10): mild(3-4), moderate(5-6), and severe(7-8). The same biomechanical model (Plug-in Gait) and analysis software

(Nexus 2.x) were used. The MDPmean [4] was calculated from the frontal, sagittal, and transverse plane angles of the trunk, pelvis, and hip, sagittal and frontal plane angles of the knee, ankle dorsiflexion, and foot progression angle. Group comparisons were conducted through one-way ANOVA and Bonferroni post hoc tests for mean and statistical parametric mapping (SPM) to identify angular kinematics differences. Pearson's correlation coefficient was used to investigate associations between pain intensity (VAS) and MDPmean. All statistical analyses adopted a significance level of  $p < 0.05$ .

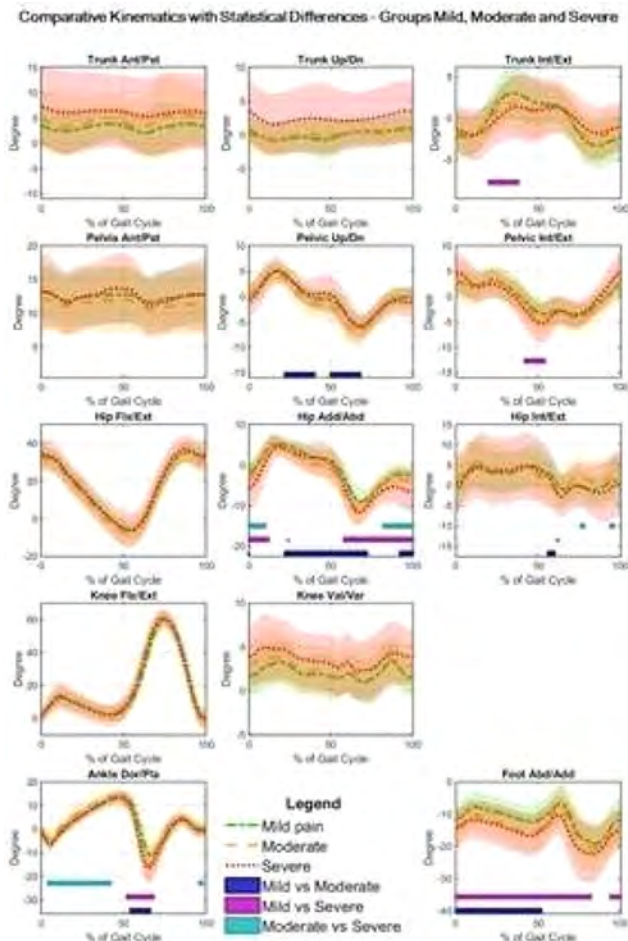


Figure 1. A comparison of mean  $\pm$ SD gait kinematics between groups. Statistically significant results are indicated by horizontal bars.

## Results

Significant differences in MDPmean were observed, with mild-moderate pain showing a deviation of  $3.56^\circ$  (95%CI:3.12-4.01), moderate-severe pain at  $6.48^\circ$  (95%CI:6.04-6.93), and mild-severe pain at  $2.92^\circ$  (95%CI:2.47-3.36). Moreover, MDPmean was strongly correlated with pain intensity (VAS,  $r=0.82$ ), and SPM analysis revealed phase-specific joint kinematics differences (Figure 1). Compared to the mild group, the moderate and severe groups showed greater pelvic obliquity, hip adduction, ankle plantarflexion, and foot progression angle, respectively. Between the moderate and severe groups, the severe group showed greater pelvic obliquity, hip adduction, less ankle dorsiflexion, and greater foot progression angle.

## Discussion

This study reveals that the MDP effectively differentiates gait kinematic changes in women with PFP, with more pronounced deviations correlating with higher pain levels. The strong relationship between MDPmean and pain intensity highlights its utility as an objective clinical tool. Furthermore, SPM analysis identified specific kinematic changes associated with varying pain levels, which may support future interpretation and highlight the potential for personalized rehabilitation strategies in managing PFP.

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## From real-world mobility to predictive simulation: fall risk assessment

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## Introduction

Falls pose a significant health risk in older adults, especially those with knee osteoarthritis (OA), which has been associated with increased fall incidence due to joint instability and muscle weakness. Although several biomechanical factors—such as muscle strength, power, motor control, and mobility—are known to influence fall risk, the most reliable predictor remains unclear.

## Research Question

The main objectives of the Power-Aging study are (1) to investigate the correlation between (the decline in) muscle force, muscle power and mobility performance over time in an elderly population with knee OA, and (2) to identify the most informative parameter (or combination of parameters) for fall risk prediction.

## Methods

A longitudinal clinical study has been designed to address the research question. The experimental protocol includes repeated assessments over time (4 in total, 6 months apart) on 50 older adults with knee OA, using gold standard measures and novel technologies

including a gait assessment, dynamometry, mobility monitoring through wearable sensors. After each visit, upon data processing, a linear mixed-effects model (LMM) will be developed, to estimate inter-subject variability and subject-specific trajectories over time. In addition, at baseline, a musculoskeletal model will be developed for each participant (by scaling a generic template model) and used to estimate biomechanical quantities—such as joint kinematics, kinetics, muscle forces, and joint contact forces—via an inverse approach. The same MSK model will later be employed to perform predictive simulations of gait in OpenSim Moco [1], informed by DMOs extracted from the inertial sensor data [2]. Prior to their use in predictive models, a statistical analysis will be conducted to perform dimensionality reduction and variable selection, thus capturing the most relevant aspects of real-life mobility. This will allow us to explore how changes in muscular capacity, joint loading, or motor control strategies (to simulate the positive effect of rehabilitation programmes) may influence future mobility performance and stability, providing a tool for testing hypothetical interventions or disease progression scenarios *in silico*.

## Results

Data collection is currently ongoing, and to date, approximately 20% of the baseline target has been completed (number of gait sequences longer than 10 seconds [%]:  $64.72 \pm 5.31$  [53.88, 68.86]; Cadence mean [step/min]:  $80.71 \pm 1.80$  [78.21, 82.52]; Steps [#]:  $35.31 \pm 4.65$  [28.11, 39.93]; Stride length mean [m]:  $0.95 \pm 0.08$  [0.85, 1.10]; Distance mean [m]:  $23.79 \pm 3.64$  [19.13, 28.14]; Speed mean [m/s]:  $0.64 \pm 0.06$  [0.56, 0.74]). A preliminary analysis of the baseline dataset will be presented at the conference.

## Discussion

This study aims to advance our understanding of the mechanisms behind fall risk and to support the development of predictive tools for monitoring motor control and informing preventive strategies.

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### ☆Accurate steady-state VO<sub>2</sub> estimation with less metabolic data using calibrated patient-specific Bayesian regression models

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## Introduction

Measuring steady-state oxygen consumption ( $\dot{V}O_2$ ) during walking is time-consuming and challenging for individuals with gait impairments. Typical protocols require six minutes of constant-speed walking to allow  $\dot{V}O_2$  to stabilize, limiting the feasibility of repeated testing. Previous work demonstrated that a dual-exponential Bayesian regression model can accurately estimate steady-state  $\dot{V}O_2$  with only three minutes of data by modeling the early metabolic rise of the  $\dot{V}O_2$  response [1]. However, this approach relies on a general prior for the exponential time constant ( $\tau$ ), the key parameter that characterizes metabolic dynamics. This study investigates whether patient-specific priors for  $\tau$ , derived from a short calibration trial, can further reduce the time required for accurate steady-state  $\dot{V}O_2$  estimation.

## Research Question

Can patient-specific priors on  $\tau$  reduce the time required for accurate estimates of steady-state  $\dot{V}O_2$ ?

## Methods

We simulated breath-by-breath  $\dot{V}O_2$  data for 50 patients transitioning from steady-state resting ( $SS_r$ ) to steady-state walking ( $SS_w$ ). Two simulated trials were generated for each patient—a calibration trial and a test trial—with different  $SS_r$  and  $SS_w$  values but the same  $\tau$ , mimicking real-world repeated tests under different conditions.

To assess the impact of patient-specific  $\tau$  priors on  $SS_w$  estimation, we:

1. Used six durations of calibration trial data (1–6 min) with a generic Bayesian model to generate six posterior  $\tau$  distributions for each patient.
2. Used these  $\tau$  distributions to build six ‘calibrated’ patient-specific Bayesian models.
3. Estimated  $SS_w$  using six durations of test trial data (0.5 min to 3.0 min) with each calibrated model as well as by an uncalibrated generic model.
4. Assessed model accuracy as the percentage of  $SS_w$  estimates that were within 10% of the true  $SS_w$  value.

## Results

Longer calibration trials yielded more accurate and precise  $\tau$  estimates (mean error reduced from 8.1 to 1.6 sec; SD reduced from 25.0 to 4.2 sec). Incorporating these refined  $\tau$  priors into patient-specific models enabled accurate  $SS_w$  estimates with less metabolic data. For example, with a six-minute calibration trial, using only 1.5 and 2.0 minutes of test data resulted in 92% and 96% of  $SS_w$  estimates to be within 10% of the true value respectively—comparable to 96% and 100% achieved using 2.5 and 3.0 minutes of trial data with the uncalibrated model.

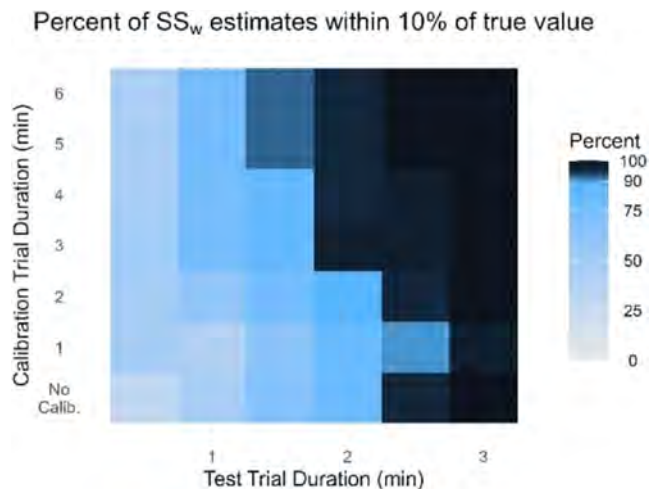


Figure 1:  $SS_w$  prediction performance using varying amounts of calibration and test data

## Discussion

Patient-specific  $\tau$  priors significantly enhanced the predictive performance of the Bayesian regression model, enabling accurate steady-state  $\dot{V}O_2$  estimates with as little as 1.5 minutes of data. While this method may not benefit protocols requiring only a single steady-state estimate, it holds substantial promise for reducing test duration in studies involving repeated measurements.

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## Non-invasive gait analysis approaches for Parkinson's disease: assessing the reliability of various markerless methods

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## Introduction

Parkinson's Disease (PD) is the second most common neurodegenerative disorder, with cases expected to double in 15 years [1]. Gait analysis has successfully demonstrated its ability to provide key indicators of PD progression [2]. Current data collection relies mainly on lab-based setups, limiting its accessibility. Advances in AI-driven markerless motion capture, such as MediaPipe [3] and AlphaPose [4], offer an alternative for real-time, home-based assessments. While stereophotogrammetry remains the gold standard, the reliability of AI-driven markerless solutions must be evaluated, especially in the case of pathological subjects, whose movements differ from the healthy subjects included in the available training datasets.

## Research Question

This study aims to assess the accuracy of different markerless motion capture approaches—single vs. multi-camera setups—on PD individuals. This is obtained by comparing joint kinematics and spatiotemporal parameters across different approaches.

## Methods

Ten PD individuals (Age:  $73.50 \pm 6.64$  years, BMI:  $25.39 \pm 1.73$  kg/m<sup>2</sup>) were recruited from Fresco Parkinson Foundation Center, Villa Margherita upon signing informed consent (ClinicalTrials.gov NCT04778852).

Data collection involved simultaneous recordings of gait trials through an optoelectronic system (Vicon, 120 Hz) and two GoProHero® v7 cameras (60 Hz, 1080p) positioned sagittally and frontally. The IORGait protocol [5] served as the gold standard for the comparison.

Different joint kinematics approaches were applied based on the available kinematic model from MediaPipe and AlphaPose:

- 2D body pose estimation (BPE) obtained from a single-camera view (solution 2D-BPE);
- 3D BPE obtained through triangulation of the different views (3D-BPE);
- 3D BPE achieved through AI-based depth estimation available from MediaPipe obtained from the 2D-BPE (depth2D-BPE).

A simplified kinematic model (i.e. based on angle between vectors defined between consecutive joints) was adopted for Marker-based joint kinematics estimation in order to match the markerless one. For statistical analysis RMS, CMC [6] and statistical SPM [7] t-test were computed.

## Results

Figure 1 shows the results for the BPE obtained through multiple views' triangulation applied to joint trajectories obtained for both ML technologies (solution 3D-BPE). While both solutions report large errors in the estimate of the sagittal joint kinematics, AlphaPose reports joint angle profiles closer to the gold standard.

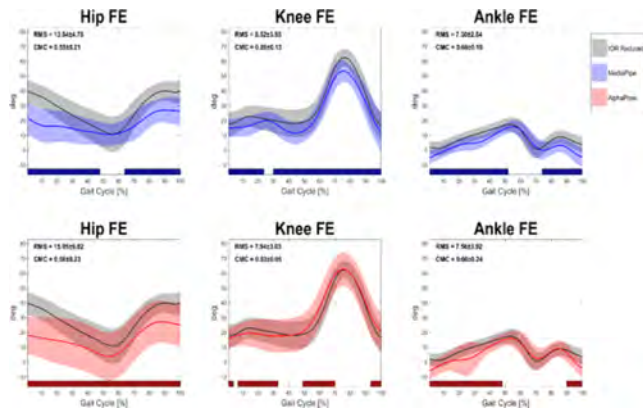


Figure 1: Lower limbs kinematics from multiple view triangulation, MediaPipe (blue) and AlphaPose (red): mean  $\pm$  st. dev. \*statistical significance ( $p < 0.05$ ). Stereophotogrammetry simplified model's kinematics in grey as reference.

## Discussion

While most studies evaluated these tools on healthy individuals, research on their performance in pathological conditions remains limited. In this work we assessed the performance of different AI driven markerless solutions in reconstructing lower limb joint kinematics in a PD population. As shown in Figure 1, both MediaPipe and AlphaPose reported large RMSE when compared with the stereophotogrammetric based kinematics. Future development might include developing adequate AI driven markerless solutions trained on pathological gait data.

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<https://doi.org/10.1016/j.gaitpost.2025.07.209>

## Reliability assessment of markerless-driven inverse kinematics and inverse dynamics in Parkinson's disease

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## Introduction

Parkinson's Disease (PD) is the second most common neurodegenerative disorder, with cases projected to double within the next 15 years [1]. Gait analysis has the potential to serve as a crucial tool for monitoring PD progression by providing key biomechanical indicators (i.e., joint kinematics, spatiotemporal parameters, ground reaction forces and joint torques)[2]. Current gait analysis methods rely on laboratory-based measures, which comes with some limitations. Recently significant efforts have been directed toward developing markerless (ML) motion capture technologies [3] that enable motion tracking without requiring physical markers, allowing for a more natural and non-intrusive assessment. At the same time musculoskeletal modeling (MSM) has proven to provide valuable information for tracking disease progression and assessing treatments' efficacy [7].

In this scenario, new ML approaches have been developed that can be applied to inform MSM simulations.

## Research Question

This study aims to evaluate the sensitivity of MSM simulations to the introduction of ML based measures. In particular the impact of ML on Inverse Kinematics (IK) and Inverse Dynamics (ID) results is assessed with respect to state of the art (i.e., stereophotogrammetry and force plate based) ones.

## Methods

Ten individuals with Parkinson's Disease (PD) (Age: 73.50  $\pm$  6.64 years, BMI: 25.39  $\pm$  1.73 kg/m<sup>2</sup>) were recruited from the Fresco Parkinson Foundation Center, Villa Margherita, as part of an approved clinical trial (ClinicalTrials.gov NCT04778852).

Data collection involved simultaneous recordings using an optoelectronic system (Vicon, 120 Hz), two GoPro Hero<sup>®</sup> v7 cameras (60 Hz, 1080p) positioned sagittally and frontally and two force plates (960 Hz, Bertec). The IORGait protocol [4] was employed.

Pose2Sim [3] markerless was applied on AlphaPose [3] estimates to reconstruct the 3D body pose and perform the IK in OpenSim. The GoPro cameras were extrinsically calibrated to ensure alignment with the global reference frame.

Simultaneously, MB data were exported to OpenSim for comparison. Both MB- and ML-driven kinematics were integrated with ground reaction forces and joint torques estimated through the ID.

Results were compared between methods through the Statistical Parametric Mapping (SPM) [6]. Furthermore, Root Mean Square Error (RMSE) and Coefficient of Multiple Correlation (CMC) [5] were computed.

## Results

The estimated ID results are presented in Figure 1. A good agreement could be observed for the majority of lower limb joints' time-series, with a CMC higher than 0.70. However, important differences in magnitude were detected, as indicated by the RMS values.

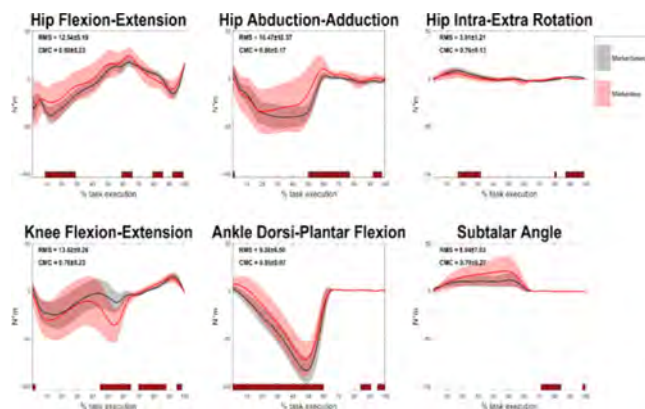


Figure 1: Lower limbs joint torques computed through ID, Markerbased (grey) and Markerless (red): mean  $\pm$  st. dev. \*statistical significance ( $p < 0.05$ ).

## Discussion

From the statistical analysis, the ML-driven ID results were closer to the MB counterparts compared to the IK ones.

In this regard, the Residual Reduction Algorithm could be used to leverage GRF data for refining ML-based joint kinematics, potentially enhancing overall accuracy and biomechanical consistency.

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<https://doi.org/10.1016/j.gaitpost.2025.07.210>

## Thorax and hip kinematics can differentiate limping severity in patients undergoing total hip arthroplasty

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## Introduction

Limping is a significant gait impairment in patients undergoing total hip arthroplasty (THA), that often persists after surgery, and affects postoperative patient satisfaction and quality of life[1]. Therefore, a better understanding of this phenomenon is crucial for patient care. Currently, limping and its severity are mostly assessed with the Harris Hip limping subscore, a clinician-reported outcome measure consisting of a four-point Likert scale (ranging from none-0 to severe-3)[2]. However, the intra-rater reliability of this subjective measure was recently found moderate to strong and weak to moderate in the case of inter-rater reliability[2]. This highlights the need for an objective measure of the limping severity for future research and clinical applications.

## Research Question

The present study aims to identify kinematic features of gait that describe the limping severity in THA patient

## Methods

Twenty-four patients undergoing total hip arthroplasty underwent gait analysis at three-time points: before surgery, three months postoperatively, and one year postoperatively. Additionally, 33 asymptomatic participants completed a single gait analysis, resulting in a total of 97 visits. During each assessment, full-body kinematics with an optoelectronic system and 2D video recordings were collected. The limping severity was assessed by an experienced

surgeon using the Harris Hip limping subscore on 2D video recordings (front and back views) of a 10-meter walk at a self-selected speed. Spatiotemporal parameters and the kinematics of the trunk, spine, pelvis, hip, and knee were extracted from the marker-based motion capture. For each joint and segment, joint angles, angular velocities (AV), and angular accelerations (AA) were computed in the frontal plane. Data reduction was performed in two steps exclusively on the THA patient's data. First, each kinematic time series was discretized in four simple features: amplitude, minimum, maximum, and mean values [3]. Then, a Principal Component Analysis was performed, selecting components with elbow methods and features with a threshold of 0.7 on cos2. These selected variables were used to compare limping severity groups within the THA patients and control group, using Kruskal-Wallis ANOVA with Holm post-hoc adjustment.

## Results

One THA patient scored 3 in a single session and was reassigned to the limping score 2 group for consistency. Over the 4 principal components, AV and AA amplitude of the thorax, spine, pelvis, and hip were selected. The AV amplitude of the hip and thorax and the AA amplitude of the thorax were significantly different between groups (Figure 1).

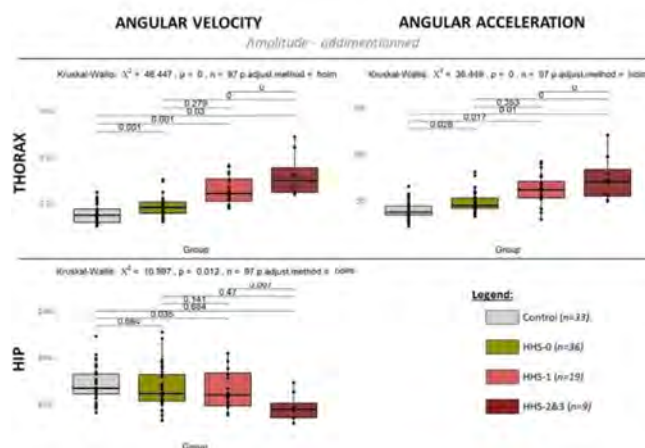


Figure 1: Group comparisons of the selected features

## Discussion

Amplitude of the hip and thorax AV, and the thorax AA appear to be promising features to differentiate levels of limping severity in THA patients. These findings highlight the potential of using objective biomechanical outcomes to objectively characterize limping severity in this population. Further validation is needed to define the full psychometric properties of these outcomes.

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## Can game-based metrics serve as relevant measures to evaluate upper limb impairments in children with neuromotor disorders? A systematic review

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## Introduction

Children with neuromotor disorders often experience upper limb impairments, affecting their daily activities and participation [1]. Videogames, including virtual reality and computer-based games, have emerged as engaging tools for upper limb rehabilitation, showing potential benefits for movement improvement and motivation [2]. However, the capacity of using video games to assess limb movements through in-game performance metrics has not yet been explored, and their validation remains unclear.

## Research Question

can game-based metrics serve as relevant measures to evaluate upper limb impairments in children with neuromotor disorders?

## Methods

A systematic review (PROSPERO: CRD42024550469) was conducted across PUBMED, MEDLINE, Web of Science, and Cochrane database according to PRISMA guidelines [3]. Articles were screened according to inclusion/exclusion criteria and data were extracted focusing on game characteristics, outcomes measured, and their metrological properties.

## Results

Out of 1033 articles screened, 23 studies met the inclusion criteria and were retained for analysis. A summary of the extracted data is presented in Figure 1. Altogether, 447 children—mainly with cerebral palsy (n=355), with a mean age of  $9.9 \pm 2.4$  years—participated in video game-based tasks involving quantitative assessments. Across these studies, 83 distinct upper limb performance metrics were identified, measured either directly from the game itself or from external measurement tools. These outcomes encompassed kinematic measures (58%), game scores (24%), actimetry

(11%), electromyography (6%), and kinetic measures (1%). Only two studies examined reliability, reporting test-retest values across 8 metrics, with results ranging from moderate to excellent. For discriminant validity, 8 studies included control groups, evaluating 20 outcomes, of which 13 successfully differentiated between populations. Responsiveness was addressed in 10 studies using pre-/post-intervention designs; among 45 assessed outcomes, 23 showed significant improvements following therapy. Convergent validity was explored in five studies, revealing moderate to strong correlations with standard clinical assessments.

Figure 1: Description of the extracted outcomes.



SP: spatiotemporal; EMG: electromyography; IMU: inertial measurement unit; ROM: range of motion; For measurement properties section: colored: property investigated; letter: which property; number: number of involved outcome; R: reliability; Rp: responsiveness; D: discriminant validity; C: criterion validity; +: positive result reported; -: negative result reported

## Discussion

The present systematic review identified a wide range of game categories and upper limb game performance outcomes, capturing a broad spectrum of motor tasks. Despite their growing use, measurement properties (reliability, validity, and responsiveness) of these game-based measures have been insufficiently addressed, limiting the interpretability and clinical relevance of the reported outcomes. To be used as research outcomes and to guide therapies in clinical practice, important work on the measurement properties has to be done.

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## Alterations of the flexion relaxation phenomenon in asymptomatic participants induced by hamstring mobility restrictions.

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## Introduction

The flexion-relaxation phenomenon (FRP) – defined by the reduction of paraspinal muscle activity at maximum trunk flexion – is frequently altered in individuals with non-specific chronic low back pain (NSCLBP) [1]. This phenomenon is believed to result from reflex inhibition of the erector spinae muscles, triggered by stretch receptor activation in posterior spinal tissues [2]. Hamstring flexibility influences pelvic tilt and trunk motion during flexion, and reduced extensibility may limit anterior pelvic rotation, leading to compensatory spinal movements [3,4]. This altered lumbopelvic rhythm could potentially affect neuromuscular control and disrupt normal FRP patterns. However, the specific role of hamstring tightness in modulating FRP remains unclear.

## Research Question

Can isolated hamstring mobility restriction, emulated with a passive exoskeleton alter the FRP in asymptomatic participants?

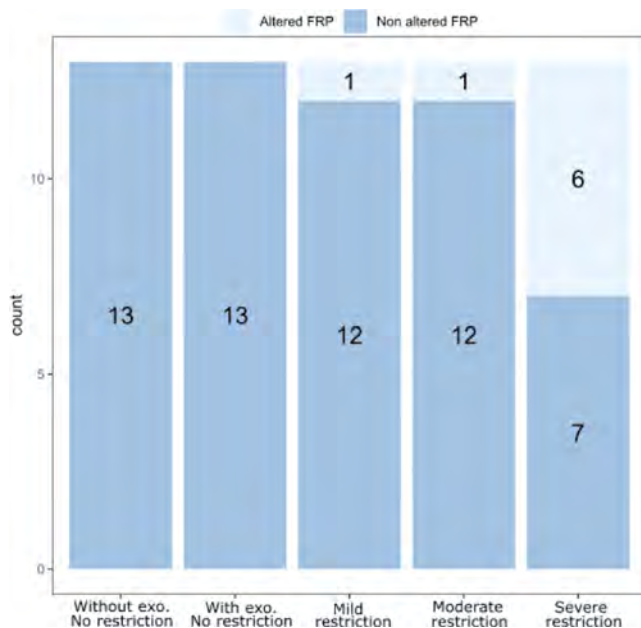
## Methods

Asymptomatic participants were prospectively enrolled and evaluated in a human movement laboratory. The inclusion criteria were 1) no history of back pain in the last six months, 2) age between 18 and 60 years old, 3) body mass index inferior to 30 kg/m<sup>2</sup>, 4) no musculoskeletal injury in any other body parts, 5) no prior back surgery. Three levels of hamstring mobility restriction were simulated using a passive exoskeleton, corresponding to hip flexion

limitations of mild (30°), moderate (15°), and severe (0°). Two reference conditions—without the exoskeleton, and with the exoskeleton but no restriction—were also tested. For each condition, the participants performed three consecutive maximal trunk flexions with their legs straight. Surface electromyographic (sEMG) was used to record bilaterally erector spinae longissimus activity. FRP presence or alteration was visual assessed from raw sEMG signals. The number of participants exhibiting altered FRP was reported for each condition.

## Results

Thirteen participants (mean age  $28.3 \pm 7.8$  years; 62% female) were recruited. In both reference conditions, none of the participants (0/13, 0%) showed altered FRP. Under mild and moderate simulated restrictions, 1/13 participants (8%) exhibited FRP alteration. Under the severe restriction condition, 6/13 participants (46%) demonstrated altered FRP (Figure 1).



## Discussion

Severe emulated restriction on hamstring mobility led to FRP alteration in nearly half of the asymptomatic participants who otherwise presented normal FRP patterns. These results suggested that spino-pelvic imbalance, induced by hamstring tightness, may play an important role in FRP alterations. Further analyses are required to evaluate the impact of the simulated hamstring lack of mobility on kinematic adaptations.

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## Machine learning-driven preoperative evaluation of lower limbs in spinal sagittal imbalance using IMU sensors

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## Introduction

The spinal sagittal imbalance (SSI) is a multifaceted condition resulting in a disruption of the spine's natural curvatures, leading to back pain associated with nerve impingement and weakness, as well as a debilitating effect, resulting in diminished quality of life [1-2]. The initial cause of SSI is misalignment of the spine in the sagittal plane, leading to the potential, as in degenerative conditions, for nerve impingement and muscular strain. Imaging studies remain a key part of investigating SSI, but the method has inherent limitations. Imaging studies do not provide information about functional impairments [3-4].

## Research Question

Can preoperative lower limb kinematic patterns, captured via IMU sensors, predict using machine learning?

## Methods

This study investigated SSI by comparing 50 female patients ( $58 \pm 8.94$  years old) and 50 control ( $55 \pm 3.5$  years old) diagnosed with SSI (SVA > 5 cm). The study followed strict ethical protocols, including informed consent and adherence to the Declaration of Helsinki. In this study, gait measurements were collected using a 9-axis IMU system (MyoMotion, Noraxon USA). Each IMU captured linear acceleration (LA) and angular velocity (AV) along three axes. Seven IMU sensors were placed on the pelvis, both thighs, shanks, and feet to comprehensively capture lower limb motion. To ensure the signals are collected accurately throughout the gait cycle, the

signals received from the IMU sensors underwent noise removal and were then divided into units corresponding to each gait cycle [5].

## Results

In this study, machine learning models (SVM) were trained on sensor data to classify SSI patients versus controls [6-7]. Each sensor was analyzed separately, and model performance was evaluated based on training and test results shown in Table 1.

Table 1: SVM Classification Accuracy

| Body Segment | Train  | Test   |
|--------------|--------|--------|
| Pelvic       | 99.07% | 98.75% |
| Left Thigh   | 99.12% | 97.34% |
| Right Thigh  | 99.00% | 97.80% |
| Left Shank   | 99.20% | 98.09% |
| Right Shank  | 98.40% | 96.70% |
| Left Foot    | 99.01% | 97.80% |
| Right Foot   | 98.07% | 97.09% |

The high classification accuracy observed across all body regions suggest that ML model are a valuable method for diagnostic assessments and monitoring patient conditions in clinical conditions.

## Discussion

In this study, we investigated the kinematics of the lower limbs of SSI patients using IMU sensors. The SVM model showed the highest classification accuracy and achieved a highly accurate classification of both the control and the patient group. The primary findings were:

**Pelvic:** We observed alterations in motion that suggested a loss of postural control and a decrease in pelvic control while in motion.

**Thighs:** Primary impairment on both sides was observed with rotational and vertical control impairing forward propulsion and stability.

**Shanks:** Impairment of lateral and forward/backward stability was observed, leading to an unsteady gait pattern and increased sway.

**Feet:** Patients exhibited difficulty controlling foot motion for walking and instability, as has previously been discussed involving heel strike and push-off.

From the results, it was concluded that there was a broad range of deficits in motion control in Prader-Willi syndrome patients affecting balance, coordination, and efficiency of gait.

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## Surgical impact on walking patterns in patients with sagittal spinal misalignment

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## Introduction

Spinal issues can be treated in many ways, including surgery, physiotherapy, medication, and lifestyle change. All treatment of the condition must start by accurately diagnosing the issue or ailment through proper clinical evaluation and testing [1-2]. The integrity of soft tissues may have an effect on the intervertebral discs and contribute to postural deformities. Gait parameters of step time, step length, cadence, and step velocity all provide a quantitative measurement of the effect of spinal disease on functional ambulatory data and balance data. Spinal disease changes gait characteristics, resulting from the pain producing muscle weakness and structural deficits, increasing fall risk and limiting the ability to complete daily activities. Existing assessment techniques, such as inertial measurement unit (IMU) sensors, are a valid, efficient, and cost-effective assessment alternative to current motion assessment systems [3-4].

## Research Question

How does gait variability change in patients with spinal sagittal imbalance following surgical intervention?

## Methods

The primary aim of this study was to examine gait variability in 50 female patients ( $58 \pm 8.94$  years old) and 50 controls ( $55 \pm 3.5$  years old). Patients had an SVA of  $> 5$  cm with no history of spinal or gait-related disorders in controls. All participants were screened for conditions that could affect cardiovascular, neurological, and motor function. All data was collected using IMU sensors (accelerometers and gyroscopes) that were placed on the feet, which provided spatiotemporal gait parameters. All raw gait data was pre-processed; specific components included formatting the data, segmenting the data into gait cycles, removing outliers, and applying a filter to eliminate noise using a 4th-order Butterworth low-pass filter [5].

## Results

Statistical analysis of the pre-processed data consisted of using a two-way repeated-measures ANOVA to assess the gait parameters between control, pre-surgery, and post-surgery groups. Time (pre vs. post-surgery) served as the within-subjects factor and group (control vs. patients) as the between-subjects factor. Bonferroni-corrected post hoc tests [6] were provided, and alpha was set at  $p < 0.05$  in Table 1.

Table 1: Spatiotemporal gait parameters in SSI patients and control group (right and left foot).

| Parameter           | Control<br>(Mean $\pm$ SD) | Pre-surg.<br>(Mean $\pm$ SD) | Post-surg.<br>(Mean $\pm$ SD) | p-value<br>(C. vs. Pre / C. vs. Post) |
|---------------------|----------------------------|------------------------------|-------------------------------|---------------------------------------|
| <b>Right Leg</b>    |                            |                              |                               |                                       |
| Stride Time (s)     | 1.12 $\pm$ 0.08            | 1.45 $\pm$ 0.12              | 1.35 $\pm$ 0.10               | .001 / .002                           |
| Velocity (m/s)      | 1.42 $\pm$ 0.10            | 0.98 $\pm$ 0.14              | 1.30 $\pm$ 0.12               | .001 / .003                           |
| Cadence (steps/min) | 112.5 $\pm$ 5.2            | 89.3 $\pm$ 7.1               | 105.6 $\pm$ 6.5               | .001 / .001                           |
| <b>Left Leg</b>     |                            |                              |                               |                                       |
| Stride Time (s)     | 1.12 $\pm$ 0.07            | 1.46 $\pm$ 0.13              | 1.34 $\pm$ 0.11               | .001 / .003                           |
| Velocity (m/s)      | 1.41 $\pm$ 0.09            | 0.97 $\pm$ 0.13              | 1.28 $\pm$ 0.12               | .001 / .002                           |
| Cadence (steps/min) | 113.0 $\pm$ 5.1            | 88.9 $\pm$ 6.8               | 104.9 $\pm$ 6.3               | .001 / .002                           |

## Discussion

Postoperative evaluation revealed mild improvement in spatiotemporal gait parameters in SSI patients. Progress in gait velocity, cadence, and stride length, with reduced double support time and more normalized stance/swing phases, is an indicator of improved balance and walking efficiency after surgery. These parameters did not fully normalize, which indicates residual motor deficits in the lower limbs. These findings emphasize the value of surgery for lessening biomechanical constraints but also its partial success in normalizing gait. The study underlines the need for targeted post-surgical rehabilitation with a focus on strength, balance, and neuromuscular retraining to optimize mobility and functional results. The persistent gait pathologies emphasize the need for long-term measures of rehabilitation [7].

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## Machine learning for surgical outcome evaluation: gait analysis in spinal sagittal imbalance patients

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## Introduction

Sagittal spinal imbalance (SSI) shifts the spine from its normal arc and can change the biomechanics of walking, including speed and stride length [1-2]. These gait adaptations may lead to progressive joint degeneration and increased fall risk. Treatment for SSI includes surgical, therapeutic, pharmacologic, and lifestyle modifications. Care and treatment will be optimized based on thorough clinical assessments, including medical imaging and other diagnostics, such as motion capture and analysis [3-4].

## Research Question

How effective is ML-based gait analysis in detecting and quantifying changes in gait patterns in patients with SSI before and after surgical intervention?

## Methods

The aim of this study was to analyze gait variability in 50 female patients ( $58 \pm 8.94$  years old) and control groups ( $55 \pm 3.5$  years old) and compare it with the results of controls. Patients were included with a sagittal vertical axis exceeding 5 cm. Controls did not have a history of spinal surgery and had no gait-affecting disorders. Every participant was screened for cardiovascular, psychological, and psychoactive diseases that could affect motor function. These signals were then low-pass filtered to keep only the gait-relevant information while filtering out high-frequency noise [5].

## Results

Postoperative outcome for SSI patients was classified by the SVM model [6] using pre-surgical frame and control group data according to the classification of control and pre-surgery group data that is shown in Table 1. This was represented by the largest identity with control data in the right shank sensor output, meaning that it looked most like a control person following surgery. The right thigh sensor showed the lowest alignment with healthy adults, indicating that this site may need additional attention in future clinical studies. Pelvic sensor data analysis demonstrated that 65% of patients had unaltered pathological gait features after surgery, reflecting no significant benefits in dynamic over static outcomes. Data recorded for the left thigh were 80% similar to preoperative gait patterns, while data collected from the right thigh demonstrated an even better outcome with an indistinguishably high value of similarity (70%) not significantly less than behavior before surgery. 78% of left shank data from post-surgical patients and 68% from the right resembled preoperative gait patterns when it came to measurements extracted using lower leg sensors. Foot sensor analysis showed that after surgery, 75% had left foot and 80% had right foot data, no different from before the operation.

Table 1. Performance metrics for SVM method across different datasets

| Dataset     | Accuracy | Precision | Recall | F1-score |
|-------------|----------|-----------|--------|----------|
| Pelvic      | 99%      | 97.7%     | 98%    | 98.4%    |
| Left Thigh  | 99.2%    | 98.1%     | 98.00% | 99.2%    |
| Right Thigh | 99.5%    | 99%       | 99.4%  | 99.20%   |
| Left Shank  | 99%      | 99.60%    | 99.2%  | 99.8%    |
| Right Shank | 98.20%   | 99.02%    | 99.2%  | 99.20%   |
| Left Foot   | 99%      | 99.5%     | 99%    | 99%      |
| Right Foot  | 98.5%    | 99%       | 95%    | 98%      |

## Discussion

The SVM model demonstrated strong performance in distinguishing between pre-surgery and control group gait patterns, achieving classification accuracies exceeding 97%. The high accuracy of the ML models indicates that the extracted gait features effectively differentiate between the gait profiles of SSI patients and controls. Among the various models, the SVM model showed optimal performance.

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## Biomechanical analysis of lower limb muscle forces during lifting of incremental loads using OpenSim

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## Introduction

Lifting activities form an important part of daily life and work activities. Proper and safe lifting of objects requires the coordinated forces of muscles and ligaments to maintain posture while also limiting the risk of musculoskeletal injury [1-2]. Excessive strain or injury can occur in lifting movements if movements are performed poorly or if intensities exceed normal tolerances; this is particularly the case with the lower limb [3-4]. Due to the importance of understanding this as well as minimizing the risk of these types of awkward postures/injuries [5], this investigation investigates the biomechanics of lifting based on an analysis of muscle forces by using OpenSim [6].

## Research Question

How do lower limb muscle forces change during the lifting of incremental loads, as analyzed through biomechanical simulation using OpenSim?"

## Methods

The study involved two healthy, young adult volunteers (18-25 age cohort) who performed controlled lifting maneuvers, lifting 3 kg, 6 kg, and 9 kg weights. A motion analysis imaging system was used to collect kinematic data of their lower limbs while lifting each of the

weight conditions. The motion data were imported into OpenSim v4.4, correlation of musculoskeletal dynamics occurred, and muscle forces for 93 lower limb muscles were calculated. The raw muscle force data were analyzed statistically with SPSS software, using a one-way ANOVA.

## Results

The muscle force became increasingly significantly greater (from 3 kg to 9 kg) for the gastrocnemius ( $p = 0.004$ ), rectus femoris ( $p = 0.001$ ), and biceps femoris ( $p = 0.003$ ), which all exhibited progressive increases in force at greater weight in average amount in both legs. Furthermore, in the right leg, the vastus medialis ( $p = 0.007$ ), vastus intermedius ( $p = 0.009$ ), and vastus lateralis ( $p = 0.01$ ) muscles, and the left soleus ( $p = 0.015$ ) muscle also showed significant increases in muscle force across all weight changes. The other muscles in the vastus of the left leg (the vastus medialis ( $p = 0.04$ ), vastus intermedius ( $p = 0.03$ ), and vastus lateralis ( $p = 0.04$ )) and the semitendinosus muscle of the right leg all showed significant change only at 6 kg to 9 kg and no significant difference from 3 kg to 6 kg. The right soleus ( $p = 0.03$ ), left semitendinosus ( $p = 0.04$ ), and semimembranosus muscles in both legs ( $p = 0.04$ ) all showed significant force change only from 3 kg to 6 kg and no further significant change in force from 6 kg to 9 kg.

## Discussion

This study emphasizes the different responses of lower limb muscles to increases in lifting loads. Some muscles exhibited linear increases in force as weight increased, while others only responded at specific thresholds. Analyzing these responses adds to safety in the design of lifting protocols and rehabilitation strategies.

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## Investigation of lumbar paraspinal muscle loading in relation to arm swing variability during human gait

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## Introduction

Arm swings during walking affect different parts of the body, leading to increased stride length, walking speed, and step width [1–3]. This improves the stability of the lumbar spine across different age groups. Restricting arm swings can change load transfer and increase trunk flexion [4–7]. Arm swings also dampen trunk and head rotation while aiding in diagnosing diseases [8]. This study explores the role of arm swing in human walking and its effect on the lumbar muscles. Previous research has not examined how different arm swing patterns directly affect the force exerted on the lumbar muscles. To address this gap, the study investigates the impact of three arm swing conditions (large, normal, and small) on the muscular forces in the lumbar spine.

## Research Question

Do different states of arm swing affect the muscle forces of the lumbar spine?

## Methods

This study involved young, healthy men who met specific inclusion criteria, such as no previous surgery or injuries in the spine and lower limbs and not using psychotropic medications. Five male students between 18 and 30 years old were chosen. Data was collected using the Vicon motion analysis device with 6 infrared cameras. model was created in OpenSim and scaled for each person. The results obtained from inverse dynamics were analyzed using SPSS and the ONE-WAY ANOVA test.

## Results

The study analyzed 11 lumbar muscles, including 8 surface muscles (for quadratus lumborum and longissimus) and 4 deep muscles (for multifidus spinous). Similarity was observed in 5 participants (Table 1). This study found a significant difference in muscle force between "large arm swing" and "normal arm swing" in surface muscles. However, no significant difference was found between "normal arm swing with small arm swing" and "small arm swing with large arm swing." In deep muscles, two significant differences were observed. There was a significant difference in muscle force between "normal arm swing and large arm swing" with "small arm swing," but not between "normal arm swing and large arm swing" with each other.

Table 1 - Results from SPSS Software for 11 Selected Muscles

|                 | The name of the muscle | LA/SA    | LA/NA     | NA/SA     |
|-----------------|------------------------|----------|-----------|-----------|
| Surface muscles | QL_post_1_1-L3         | 1.82120  | 2.37359*  | -0.55239  |
|                 | QL_post_1_3-L1         | 1.08225  | 1.69720*  | -0.61494  |
|                 | QL_post_1_3-L3         | 2.06525  | 2.97997*  | -0.91427  |
|                 | QL_post_1_2-L3         | 0.83835  | 1.10978*  | -0.12758  |
|                 | LTrpL_L1               | 1.30950  | 6.99011*  | -5.68062  |
|                 | LTrpL_L2               | 3.24110  | 9.91768*  | -6.67657  |
| deep muscles    | LTrpL_L3               | 7.93036  | 19.62527* | -11.69491 |
|                 | MF_m1_1                | 3.53430* | 0.51421   | 3.02009*  |
|                 | MF_m1_2                | 2.88683* | 0.66205   | 2.22478*  |
|                 | MF_m1_laminar          | 1.19823* | 0.02982   | 1.16841*  |
|                 | MF_m2_laminar          | 1.42228* | 0.01105   | 1.41123*  |

The numbers with an asterisk (\*) indicate a significant difference.

## Discussion

This study aimed to analyze the impact of different arm swing positions on lumbar spinal muscle forces. Past studies have not examined how arm swings directly affect muscle forces in the lumbar region, which is important due to the proximity of these muscles to the body's center of mass. Findings reveal that the large arm swing state results in greater force in all muscles studied (both surface and deep) in the lumbar region. This is because the large arm swing causes movement of the body's center of mass, requiring the lumbar muscles to exert more effort for balance during gait.

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## Classification of daily human activities based on IMU data and machine learning models

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## Introduction

The classification of human daily activities has become a significant focus within the fields of data science and artificial intelligence due to its wide range of applications [1-2], particularly in healthcare [3-4], sports, and wearable technology [5]. By leveraging motion data captured from sensors, it becomes possible to automatically recognize physical activities such as walking, running, sitting, and stair navigation. Among the most commonly used sensors for this purpose are inertial measurement units (IMUs), which combine accelerometers and gyroscopes to detect and record body movements in three-dimensional space [6].

## Research Question

How can daily human activities be accurately classified using IMU sensor data in combination with optimized machine learning models?

## Methods

In the current study, the participants were given a total of six different activities to complete as part of data collection. These activities were running a distance of 15 meters, walking for 7 meters, sitting, standing, climbing stairs, and descending stairs. The participants completed the activities according to standardized conditions so that the dataset was consistent and of high quality. A single IMU sensor was placed at the body's center of mass, which was in the sacrum region of the body, as a partial stationary device to explore linear acceleration signals. There were a number of ML algorithms applied in this study to find the best classifier for the six target activities, including a hybrid model of Random Forest and k-Nearest Neighbors (k-NN), a self-contained k-NN classifier and, for comparison purposes, a Naive Bayes classifier [7-9].

## Results

The study concluded that Random Forest coupled with k-NN offered the highest classification accuracy of 96%. This demonstrates that the hybrid approach can likely be effective for capturing more complex relationships in the feature space. The k-NN also performed well alone, achieving 93.7% accuracy; this suggests that distance-based methods are effective for classifying human movement patterns. However, the Naive Bayes classifier had the lowest accuracy of 72%. There is a simple explanation for this likely result: the Naive Bayes classifier specifically assumes the features are independent of one another when making a classification. Accuracy results of all models are shown in Table 1. This assumption does not hold well for the dependent features derived from IMU data.

Table 1. Accuracy results of all models.

|                     | SVM   | k-NN  | RF    | LDA   | NAIVE | HYBRID |
|---------------------|-------|-------|-------|-------|-------|--------|
| ACCURACY            | 87.55 | 93.7  | 92.53 | 83.4  | 72.61 | 96.2   |
| CROSS<br>VALIDATION | 83.76 | 90.07 | 90.86 | 76.51 | 68.24 | 92.7   |

## Discussion

This study demonstrates the wide applicability of machine learning techniques, in particular ensemble and/or non-linear models, to effectively classify daily human activity signals collected from a single source of wearable sensor data. The proximal placement of the IMU sensor at the sacrum allowed it to capture whole-body motion effectively while allowing for a practical method of deployment. This work emphasizes the promising, novel role of sensor-based activity recognition in the area of health-based monitoring, personal fitness tracking, and intelligent technologies.

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## ☆Functional relevance of the spine-hip radiographic classification in patients with adult spinal deformity

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## Introduction

With ageing, the musculoskeletal system is affected by tissues degeneration, especially in the spine. Adult spinal deformity (ASD) is known to affect the lumbo-pelvic complex and therefore the relationship between the spine and the hip [1]. Several authors have attempted to describe this relationship to better understand both spine and hip mobility and evaluate risks of dislocation after total hip replacement. The Bordeaux team suggested a classification by including values of pelvic incidence (PI) and pelvic tilt (PT) changes between sitting and standing radiographs [2]. Patients are classified as spine-users if they exhibit harmonious movement between their spine, pelvis and hip, or as hip-users if they mostly recruit their hips.

## Research Question

Does the Bordeaux classification apply on pelvic and hip kinematics during the sit-to-stand movement?

## Methods

189 ASD and 52 controls underwent biplanar X-rays in standing and sitting positions, with the calculation of 3D spinopelvic and global alignment parameters and their adaptation between positions ( $\Delta$  = sitting-standing). ASD were radiographically classified according to the Bordeaux classification as spine-users if both their  $PI \geq 40^\circ$  and  $\Delta PT > 10^\circ$ , otherwise as hip-users. 3D movement analysis of the sit-to-stand movement was performed to calculate pelvic and hip kinematics [3, 4] along with their corridor of normality. ASD patients were classified as having a retroverted pelvis and extended hips attitude (Retro-Ext) compared to controls, an anteverted pelvis and flexed hips attitude (Ante-Flex), or normal pelvis-hip kinematics. Distribution of the Bordeaux classification among pelvis and hip kinematics was evaluated (Fisher's test).

## Results

ASD patients were distributed as 70 hip-users and 119 spine-users. When analyzing the pelvis and hip kinematics, 71 patients presented an Ante-Flex pattern (26 hip-users, 45 spine-users), 77 patients presented a Retro-Ext pattern (30 hip-users; 47 spine-users) and 41 patients presented with a normal pattern (14 hip-users; 27 spine-users). Spine and hip users were equally distributed among the kinematic patterns (Fisher's test:  $p=0.8$ ). 55% of ASD with a retroverted pelvis-hip extended kinematic during the sit-to-stand movement had a sagittal malalignment (Fisher's test:  $p=0.04$ ); an increased SVA was correlated to more extended hips during movement ( $r=0.2$ ,  $p=0.02$ ).

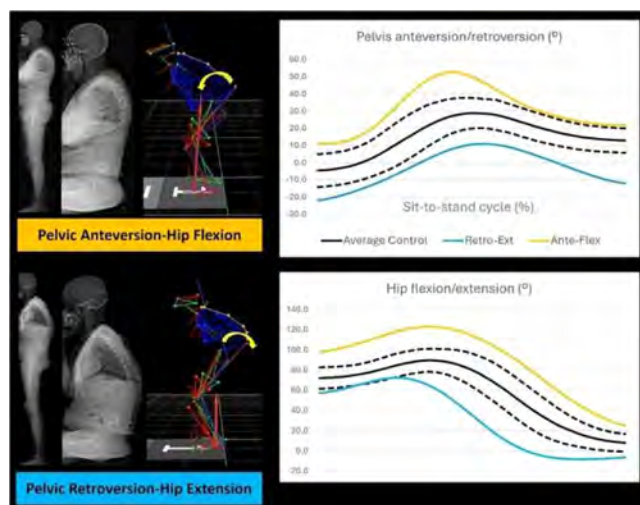
## Discussion

This study showed that spine users and hip users can have the same kinematic pattern during the sit-to-stand movement. The existing spine-hip classifications are all based on standing and sitting radiographs to classify patients according to their spine and hip mobility. As demonstrated in this study, static radiographs show only the start and ending position of the patient and hide the exact movement of the spine, pelvis and hip. Thus, a new classification should incorporate 3D kinematics of the pelvis and hip in addition to the static radiographs in standing and sitting.

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Figure 1: Kinematic patterns of the sit-to-stand movement in patients with adult spinal deformity.



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## Spine-hip adaptation during the sit-to-stand transitioning in patients with Adult Spinal Deformity

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## Introduction

Patients with Adult Spinal Deformity (ASD) are known to present with concomitant degenerative problems in the hip, described as the spine-hip syndrome [1]. In severe cases, total hip replacement (THR) is performed. However, the risk of cup dislocation is known to be

elevated in the presence of spinopelvic malalignment, mainly due to inadequate cup anteversion [2]. This conflicting spine-hip interaction happens during daily life activities such as the transition between standing and sitting [3]. Thus, it is essential to understand the spine-hip adaptation during the sit-to-stand (STS) transition.

## Research Question

What are the determinants of abnormal spine-hip adaptation during the STS transition in ASD patients?

## Methods

204 ASD patients and 53 controls underwent full body bi-planar X-ray in the standing and sitting positions, with the calculation of 3D spinopelvic and hip parameters (pelvic tilt PT, sagittal vertical axis SVA, pelvic incidence - L1S1 lumbar lordosis mismatch PI-LL, C7 coronal plumbline to the sacrum C7-CSL, knee flexion, acetabular anteversion). They all filled the Oswestry Disability Index (ODI). 3D movement analysis was performed during the STS movement and kinematics of the head, thorax, spinal segments, pelvis and lower limbs were calculated [4]. The difference ( $\Delta$ ) between sitting and standing L1S1 lumbar lordosis (LL) and acetabular anteversion (AcetAntev) were calculated on radiographs. Patients having increased  $\Delta$ LL and/or decreased  $\Delta$ AcetAntev (outside [10th-90th] percentile in controls) were classified in the abnormal spine-hip adaptation (Abnormal-SHA) group, otherwise as Normal-SHA.

## Results

131 ASD were classified as Abnormal-SHA and 73 as Normal-SHA (figure 1a). On standing radiographs, patients in the Abnormal-SHA group had an increased pelvic retroversion (PT:24.4 vs 14.1°), PI-LL mismatch (8.9 vs -9.4°), SVA (52 vs 3mm), C7-CSL (15 vs 8mm), knee flexion (8 vs 2°), acetabular anteversion (24 vs 18°) and a decreased LL (44 vs 61°; all  $p < 0.05$ ) compared to ASD in the Normal-SHA group. On sitting radiographs, ASD in the Abnormal-SHA had a reduced PT (34 vs 40°) and increased LL (36 vs 26°; both  $< 0.05$ ) with a similar acetabular anteversion (28°) compared to the Normal-SHA group (figure 1b). During the STS movement, ASD with Abnormal-SHA had an increased thorax flexion (25 vs 18°). Patients in the Abnormal-SHA had an increased ODI score (37 vs 30).

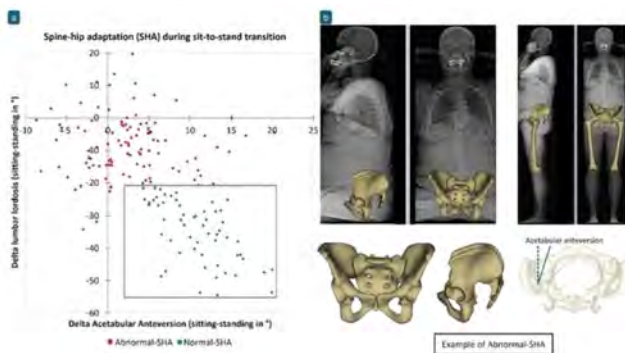
## Discussion

This study showed that severe sagittal and coronal spinal malalignment in ASD is related to abnormal spine-hip adaptation during the STS transition. This is mainly due to the lack of spinal flexibility and pelvic retroversion, limiting the increase of acetabular anteversion, when transitioning from standing to sitting. These findings can help in establishing specific recommendations for cup positioning that take into account spine-hip adaptation in patients with spinopelvic malalignment.

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Figure 1: a) Spine-hip adaptation in the sit-to-stand transition:  $\Delta$ LL relatively to  $\Delta$ acetabular anteversion in abnormal and normal SHA ASD groups; b) Example of Abnormal-SHA patient in standing and sitting positions.



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## Can 3D kinematic parameters be predicted by radiographic and quality of life scores in patients with adolescent idiopathic scoliosis?

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## Introduction

Adolescent idiopathic scoliosis (AIS) is a 3D spinal deformity that impacts one or more segments of the spine [1]. Physicians tend to base their therapeutic plan on frontal and sagittal X-rays and patient's quality of life (QOL). However, patients with AIS were found to have different kinematic strategies during gait depending on their type of scoliotic deformity [2]. Knowing that spinal movement is reduced after spinal fusion, 3D gait analysis (3DGA) could be important to evaluate treatment impact on patient's functionality [3]. While radiographic evaluation is easily accessible, 3DGA remains limited in clinical settings and difficult to access.

## Research Question

Can radiographic parameters and QOL scores predict kinematic parameters in AIS?

## Methods

72 patients with AIS underwent low-dose biplanar X-rays with the calculation of the following radiographic parameters: Cobb angle, apical vertebral rotation (AVR), frontal odontoid-hip axis angle (ODHA), T1T12 thoracic kyphosis (TK), and T10L2 thoracolumbar junction (TLJ). In addition, 3D gait analysis was performed with the calculation of the head, thorax, spinal segments, pelvis, and lower limb kinematics [4,5]. All subjects filled QOL questionnaires: Visual Analog Scale (VAS) for pain, Clinical Outcomes in Routine Evaluation- Outcome Measures (CORE-OM), Hospital Anxiety & Depression Scale (HADS), and Scoliosis Research Society-22 revised (SRS-22r) questionnaire. A machine learning random forest regression model was applied to predict kinematic parameters of interest in AIS [6] based on radiographic parameters and QOL scores, with calculation of prediction accuracy and percentage of contribution to the prediction. A 10-fold cross-validation method was used to validate the results of the model and reduce its bias.

## Results

The model predicted the kinematic frontal ODHA angle during gait with a median accuracy of 82%, walking speed and step length at 81%, pelvic obliquity at 79% and the shoulder-pelvis axial rotation at 78%. It also predicted the kinematics of the spinal segments in the frontal plane T12L3-L3L5, T6T9-T9T12, L3L5-Pelvis, C7T12-T12L5, and T12L5-Pelvis with a median accuracy ranging between 72 and 83%. In the sagittal plane, it predicted C7T12-T12L5 and T3T6-T6T9 with respective median accuracies of 81% and 78%. Cobb angle and thoracic kyphosis were the most predictive radiographic parameters of the previous kinematics (up to 43%).

## Discussion

The median accuracy of the model in the prediction of the kinematic parameters varied between 70 to 80%. However, this accuracy remains insufficient for clinical guidance to evaluate impact of surgical procedures on patient's functionality and cannot substitute the data provided by 3D movement analysis. Thus, kinematic parameters obtained from 3DGA remain a reliable standard for the functional evaluation of patients with AIS. Future technologies should focus on making 3DGA more accessible to both patients and spine surgeons.

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## Turning Task in Unilateral Cerebral Palsy: Does Direction Matter?

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### Introduction

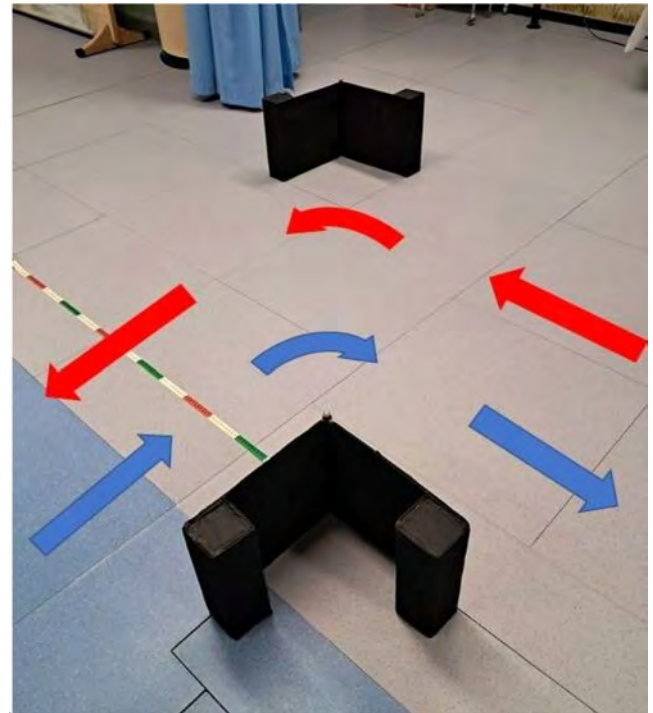
Individuals with cerebral palsy (CP) experience gait impairments that limit their mobility and negatively impact their overall quality of life. Walking is a complex task composed of several components, and among them, turning plays a critical role, especially in indoor environments where frequent directional changes are necessary. Turning task imposes unique biomechanical demands and may amplify gait asymmetries in CP patients. Falls during turning are especially dangerous, as approximately 70% result in injury (2), further emphasizing the need for targeted assessment and intervention strategies.

### Research Question

Do individuals with unilateral spastic CP exhibit specific kinematic and temporal asymmetries during 90° turns?

### Methods

Seven individuals (Median age = 25, IQR = 18–41 years) diagnosed with unilateral spastic CP, classified as GMFCS levels I–II, with sufficient cognitive ability to follow instructions and walk barefoot without assistive devices, participated in the study. Each subject performed at least five trials of preplanned left and right 90° turning tasks on a 7-meter walkway. Turns were initiated at a marked point located at least 4 meters from the start and end of the walkway to minimize acceleration/deceleration effects. Participants were allowed to choose their preferred walking speed, initial step, and turning radius. A 12-camera Vicon Nexus 3D motion capture system (Vicon, Oxford Metrics, UK) using the Plug-in Gait marker model was used to collect spatiotemporal and kinematic data. Figure 1: Location of the pylons in the turning procedure.



### Results

There was no statistically significant difference in maximum pelvic rotation angles during turning strides between the hemiparetic side (Median = 44.12°, IQR = 37.14–45.54) and the non-hemiparetic side (Median = 40.80°, IQR = 39.04–57.60). During left turns, no significant differences were found between the hemiparetic and non-hemiparetic sides in cadence, velocity, double support time, stride length, step length, or step width. Additionally, stance phase ( $p = 0.018$ ) and single support time ( $p = 0.018$ ) were shorter on the hemiparetic side. During right turns, no significant differences were found in cadence, velocity, double support time, stride length, step length, or step width. Stance phase ( $p = 0.028$ ) and single support time ( $p = 0.028$ ) were shorter on the hemiparetic side. The median number of patient-reported falls per year was 3.5 (IQR = 1–12).

### Discussion

The findings suggest that although pelvic rotational symmetry is maintained during turning in individuals with unilateral CP, distinct spatiotemporal asymmetries persist, particularly on the hemiparetic side. These asymmetries may reflect underlying neuromuscular deficits or limitations in dynamic balance control. Clinical gait evaluations incorporating turning assessments could inform targeted rehabilitation to improve directional stability and reduce fall risk.

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## Effect of selecting different biomechanical gait model on outcomes of the AI model

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## Introduction

Artificial intelligence (AI) models have proven useful across various scientific fields, including motion analysis [1]. However, the models and their outcomes can be affected by several factors [2]. Identifying and assessing these factors may help achieve more accurate outcomes.

## Research Question

Do different biomechanical gait models influence the outcomes of AI models in typically developing individuals (TD) and patients with cerebral palsy (CP) during walking?

## Methods

Retrospective gait data from 50 patients with bilateral CP (100 legs; age:  $15.5 \pm 3.7$  years) and 46 TD individuals (92 legs; age:  $26.2 \pm 2.5$  years) were collected for this study. All participants were able to walk barefoot without any assistive devices. Data were captured using a 12-camera Vicon Nexus system and three Kistler force plates. Kinematic and kinetic data, including hip, knee, and ankle joint angles (degree) and moments (Nm/kg)—were obtained using the Plug-in-Gait (PiG) model [3] and the OpenSim Hamner full-body gait model [4]. A one-dimensional convolutional neural network with five convolutional layers was trained using a 10-fold cross-validation on the kinematic data to predict joint moments for both the PiG and OpenSim models [5], for each joint separately to optimize performance. The normalized root-mean-square error (nRMSE) between measured and predicted joint moments was calculated and the AI outputs were classified into three labels—Green (G, acceptable), Yellow (Y, acceptable with caution), and Red (R, unacceptable)—using lower and upper kinetic limits (LKL and UKL) [6]. All data were normalized to a gait cycle and the AI model was trained using sagittal plane data during the stance phase.

## Results

The PiG model shows larger populations in the G and Y across all joints compared to the OpenSim as shown in Table. 1. The G+Y population for OpenSim is lower for every joint. Its difference to PiG ranges from 12% (TD-Ankle) to 39% (CP-Knee). The best prediction outcomes for hip, knee, and ankle joints were for TD using the PiG model (G; 100%, 95%, and 92%, respectively), while the worst were found in CP using the OpenSim model (R; 25%, 49%, and 28% respectively).

|            | Hip (CP)   |            | Knee (CP)  |            | Ankle (CP) |            | Hip (TD)   |            | Knee (TD)  |            | Ankle (TD) |            |
|------------|------------|------------|------------|------------|------------|------------|------------|------------|------------|------------|------------|------------|
| Population | PiG        | OpenSim    | PiG        | OpenSim    | PiG        | OpenSim    | PiG        | OpenSim    | PiG        | OpenSim    | PiG        | OpenSim    |
| G (%)      | 68         | 82         | 64         | 7          | 38         | 20         | 100        | 86         | 95         | 49         | 92         | 78         |
| Y (%)      | 29         | 42         | 38         | 44         | 46         | 41         | 0          | 22         | 5          | 29         | 8          | 15         |
| R (%)      | 3          | 31         | 16         | 49         | 23         | 39         | 0          | 17         | 0          | 22         | 0          | 15         |
| n (%)      | 38         | 75         | 34         | 31         | 72         | 82         | 100        | 85         | 95         | 78         | 100        | 48         |
| nRMSE      |            |            |            |            |            |            |            |            |            |            |            |            |
| G          | 0.245/0.04 | 0.310/0.03 | 0.185/0.04 | 0.310/0.02 | 0.195/0.03 | 0.310/0.02 | 0.085/0.04 | 0.095/0.04 | 0.185/0.04 | 0.310/0.04 | 0.075/0.04 | 0.085/0.03 |
| Y          | 0.345/0.08 | 0.245/0.05 | 0.345/0.05 | 0.345/0.05 | 0.245/0.05 | 0.245/0.04 | 0.245/0.05 | 0.245/0.05 | 0.245/0.05 | 0.245/0.05 | 0.245/0.05 | 0.245/0.05 |
| R          | 0.485/0.07 | 0.305/0.04 | 0.485/0.02 | 0.485/0.02 | 0.485/0.02 | 0.485/0.02 | 0.485/0.02 | 0.485/0.02 | 0.485/0.02 | 0.485/0.02 | 0.485/0.02 | 0.485/0.02 |

Table 1. Population of predicted moments (acceptable (G), acceptable with caution (Y), and unacceptable (R)) of the hip, knee and ankle joint in the sagittal plane for OpenSim vs PiG. The results are for both groups of TD and CP. The moments categorized according to LKL and UKL, criteria.

$$LKL = 10\% \cdot MaxMom$$

$$UKL = 20\% \cdot MaxMom$$

$$MaxMom = \frac{(MaxMom + MinMom)}{2} \quad [N]$$

## Discussion

The results indicate that the AI model achieved near-perfect performance for TD using the PiG model, while it was moderate for CP. In contrast, the AI model showed poor performance for CP and moderate results for TD when using the OpenSim model. This discrepancy may be attributed to several factors, such as differences in joint definitions. OpenSim models the knee and ankle as hinge joints, whereas PiG treats them as ball joints. Additionally, the use of optimization algorithms to determine kinematics and kinetics, along with the inclusion of virtual markers and different mass distribution in OpenSim, may have contributed to its lower accuracy. However, further research is required to identify the primary causes.

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## Dynamic ankle joint stiffness assessment in typical and pathological gait

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## Introduction

Lower extremity kinematics (joint angles) and kinetics (joint moments) are key components in clinical gait analysis. However, unlike kinematics, interpreting and understanding joint moments often remains challenging for clinicians [1]. To this aim we propose

joint stiffness, reflecting how much the joint resists changes in angle when subjected to external loads during walking as a more suitable parameter to rate and interpret impairment in clinical decision making [2,3].

### Research Question

How does dynamic ankle dorsi/plantar joint stiffness during walking help for the interpretation of joint kinetics?

### Methods

Retrospective gait data of 26 patients with pes planovalgus (PV, age  $20.8 \pm 10.7$  years), 31 with pes equinovarus (EV, age:  $17.3 \pm 10.8$  years), and 29 typically developed (TD,  $22.5 \pm 11.1$ ) subjects were collected for this study. All participants were able to walk barefoot without any assistive device. Joint moment (Nm/kg) and angle (deg) for the ankle joint were determined using the Plug-in-Gait model. All data were normalized to a gait cycle. To evaluate the control or resistance to angular motion, the dynamic ankle dorsi/plantar joint stiffness (Nm/deg/kg) during the stance phase was calculated as follow. A 5th-degree polynomial was fitted on data of ankle dorsi/plantar angle (horizontal-axis) vs. moment (vertical-axis). Ankle stiffness was then estimated by analytically deriving the polynomial allowing for a smooth while minimized numerical instability result. Additionally, correlations between ankle stiffness and both the ankle angle and moment were calculated to assess their interdependence.

### Results

Fig1. shows the averaged dorsi/plantar ankle angle, moment, and stiffness during stance phase for TD (black), PV (blue) and EV (red). The averaged and standard deviation stiffness value (Nm/de/kg) across each group during the stance were  $0.08 \pm 0.02$ ,  $0.05 \pm 0.02$ ,  $0.09 \pm 0.05$  for PV, EV and TD respectively. Correlation coefficients between the ankle angle-stiffness were 87%, 65% and 95%, and the ankle moment-stiffness 79%, 73% and 91%, for PV, EV and TD respectively.

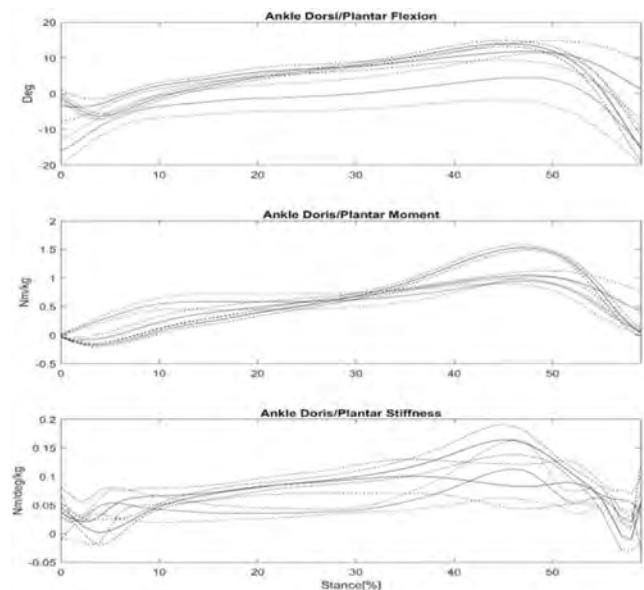


Fig1. The ankle dorsi/plantar flexion angle, moment and stiffness during stance for TD (black), PV (blue) and EV (red). The solid lines are mean and dash line are standard deviation.

### Discussion

Fig 1 illustrates that ankle joint stiffness varies dynamically throughout the stance phase. The observed stiffness values show that increases in joint moment are accompanied by increases in joint angle—and vice versa—indicating a tightly coupled force-displacement relationship. In TD, the average ankle stiffness aligns well with values reported in previous studies [4]. For TD, stiffness is strongly correlated with both joint moment and angle, suggesting a smooth transition of force from maximum plantarflexion to dorsiflexion. In contrast, this correlation is reduced by approximately 20–30% in individuals with PV and EV. Notably, in EV, stiffness decreases during midstance, which may indicate to increased energy dissipation and a tendency toward gait instability. This may be attributed to lack of proper heel contact, and calf muscle toughness.

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## Clustering using fuzzy C-means: Challenges in identifying clinically meaningful clusters in the gait of stroke patients

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### Introduction

Soft clustering techniques, such as fuzzy c-means (FCM), are particularly well-suited for handling the complex and partially overlapping characteristics typical of GA data and for identifying clusters of patients with similar characteristics. In cases where the number of clusters is unknown, it is determined using validity functions (VFs) as the first stage of the study. Briefly, VFs identify the optimal number of clusters by evaluating metrics that measure the compactness and separation of the clusters. Despite being crucial for the result, none of the published papers on the clustering of GA data discussed the reason for selecting one VF among others [1].

### Research Question

How do various fuzzy validity functions affect the identification of clinically meaningful patterns in stroke patients using FCM?

### Methods

In a preliminary analysis, the seven fuzzy VFs available in the literature [2] were tested alongside FCM on well-known datasets from the UCI Repository [3], and the two best ones were selected (named Vw and VHY). The selected VFs then were tested on a set of spatiotemporal parameters (see Table 1) typically altered in stroke patients, from three known groups: household walking strokes (HWsp: vel  $\leq 0.4$  m/s, N=19, 12 F, 51(15) y, median FAC=4.5, WHS=5, RMI=12), community walking strokes (CWsp: vel  $> 0.4$  m/s, N=27, 14 F, age 50(10) y, median FAC=5, WHS=6, RMI 13.5) and healthy subjects (HS, N=7). To assess VFs robustness across different dataset compositions, we used Generative Adversarial Networks to augment the number of healthy subjects and create three datasets with varying stroke-to-healthy ratios. To address the effect of cluster size on the cluster numbers, a new VF (Vw\*) was defined by combining the strength points of Vw and VHY. Vw\* is based on an Euclidean distance metric, with a term to enhance the impact of cluster numbers on variance and incorporates an additional term to capture intra-cluster variation (as in VHY). We first verified Vw\* on the UCI dataset and then applied it to cluster the healthy-stroke dataset.

### Results

Vw\*, Vw, and VHY outperformed the other VFs on the UCI datasets, each successfully identifying the correct number of clusters in all datasets. However, Vw and VHY failed to separate HWsp from CWsp when the number of HS increased (See Table 1), merging them into a unique cluster. The proposed VF outperformed other functions, providing a more stable and accurate cluster determination across different dataset compositions.

Table 1. Clusters Characteristics.

| Features                                                                          | Datasets | Determined No. of Clusters for Validity Functions |    |     | Clustering Performance for three clusters: Predicted Clusters (rows) vs. Actual Clusters (columns) |                 |      |      |
|-----------------------------------------------------------------------------------|----------|---------------------------------------------------|----|-----|----------------------------------------------------------------------------------------------------|-----------------|------|------|
|                                                                                   |          | Vhy                                               | Vw | Vw* | Predicted Clusters                                                                                 | Actual Clusters |      |      |
|                                                                                   |          |                                                   |    |     |                                                                                                    | HS              | CWsp | HWsp |
| Affected side Stance Duration, Cadence,                                           | HWsp: 19 | 3                                                 | 3  | 3   | HS                                                                                                 | 16              | 0    | 0    |
|                                                                                   | CWsp: 27 |                                                   |    |     | CWsp                                                                                               | 1               | 23   | 3    |
|                                                                                   | HS: 16   |                                                   |    |     | HWsp                                                                                               | 0               | 3    | 16   |
| Normalized Stride Length, Normalized Velocity, Contralateral side Stance Duration | HWsp: 19 | 2                                                 | 3  | 3   | HS                                                                                                 | 24              | 0    | 0    |
|                                                                                   | CWsp: 27 |                                                   |    |     | CWsp                                                                                               | 1               | 24   | 2    |
|                                                                                   | HS: 24   |                                                   |    |     | HWsp                                                                                               | 0               | 5    | 14   |
|                                                                                   | HWsp: 19 | 2                                                 | 2  | 3   | HS                                                                                                 | 32              | 0    | 0    |
|                                                                                   | CWsp: 27 |                                                   |    |     | CWsp                                                                                               | 1               | 24   | 2    |
|                                                                                   | HS: 32   |                                                   |    |     | HWsp                                                                                               | 0               | 5    | 14   |

### Discussion

This study demonstrates the effectiveness of FCM combined with the proposed VF in identifying clinically meaningful clusters in GA data based on spatiotemporal parameters of stroke patients. The proposed VF improved robustness in determining the number of clusters across datasets with varying stroke-to-healthy ratios.

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### Machine learning-based classification of FAIS patients from healthy subjects and feature analysis using explainable AI

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### Introduction

Femoroacetabular impingement syndrome (FAIS) is characterized by abnormal contact between the femoral head and the acetabulum, leading to pain and reduced hip function. Discriminating FAIS patients from healthy subjects based on biomechanical features is essential for a better understanding of the biomechanical alterations in this disorder and improving clinical decision-making

[1]. Given the complexity of FAIS biomechanics, accurately distinguishing patients remains challenging [2]. Machine learning (ML) methods in gait analysis can offer great potential for classifying FAIS patients and identifying key biomechanical features through Explainable AI (XAI) [3].

## Research Question

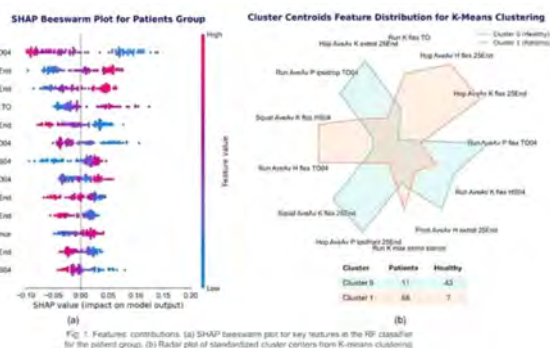
Can ML and XAI methods classify FAIS patients from healthy subjects and highlight key clinical features?

## Methods

The dataset comprises pain data scaled from 0 to 10 (Visual Analogue Scale) and 3D kinematics from stereophotogrammetry (Vicon, Oxford Metrics Ltd, Oxford, UK) from surgically treated (N=46, 19 F, 27.7(6.7) y, 173.7(9.8) cm, 70.9(12.3) kg) and conservatively treated (N=23, 14 F, 31.8(6.9) y, 175(7.9) cm, 74.1(15.2) kg) FAIS patients, as well as healthy controls (N=50, 20 F, 27.3(4.6) y, 175(10.4) cm, 71.6(12.6) kg) performing nine dynamic tasks (walk, run, stand-up, sit-down, step-up, drop jump, pivot, squat, hop). Four tasks (run, pivot, squat, hop) were selected based on the frequency and variability of pain during execution. Joint angles and angular velocities were calculated as features. Significant features for each task were identified using the Mann-Whitney U test and correlation analysis. A 10-fold cross-validation Random Forest (RF) classifier and SHAP analysis were applied to each task, selecting the top features contributing to 80% of cumulative SHAP importance. The combined feature set from the tasks was then used to classify patients and controls with the same classification method, followed by SHAP analysis to interpret the importance of features. The top contributing features from tasks were then used for subject clustering via K-means to validate the method's robustness.

## Results

The proposed method achieved classification performance in distinguishing FAIS patients from healthy subjects across 10-fold cross-validation, with an average recall of 0.86 and a precision of 0.89. K-means clustering also demonstrated discrimination between patients and controls, confirming the selected features' relevance. Angular velocities in phase-transition points contributed more significantly to classification than joint angles (Figure 1.a). The alignment between SHAP-based feature contributions and K-means cluster centroids highlights the robustness of the selected features and the resulting classifications.



## Discussion

This study demonstrated the effectiveness of ML and XAI methods in classifying FAIS patients from healthy controls, offering insights into key features to better understand movement alterations in this disorder. Angular velocities were strong contributors and, therefore, important to include in the analysis of movement patterns in FAIS patients. Future work should involve using a broader dataset to improve the reliability and generalizability of the model.

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## Stand up and walk! Are we missing information about trunk movement?

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## Introduction

The trunk represents almost 50% of the total mass of a person [1] and, because it comprises multiple segments, has a large range of motion [2]. Trunk posture and movement are important in the execution of activities of daily living (ADL), especially those related with arm function [3]. In movement analysis, the trunk is usually defined as a single rigid, segment between the shoulders and pelvis. This representation ignores the trunk's large movement potential [2], preventing a complete evaluation of its movement; however, it is not clear how much information is missed when assessing performance of ADL.

## Research Question

Are there differences in the movements quantified by a single- and multi-segment trunk model?

## Methods

The University Ethics Committee (ref:47565) approved the project. Eleven people (7 male; (mean  $\pm$  SD) age: 27.82  $\pm$  3.18years,

height:  $1.74 \pm 0.11\text{m}$ ; weight:  $75.0 \pm 12.7\text{kg}$ ) participated after signing the consent form. An upper-body marker-set was used: left/right acromion, iliac-crest, ASIS; manubrium, S1; five inverted "L" clusters of 3 markers: two 2.5cm lateral either side of C7, T3, T7, T11 and L3, with the third marker on the long end of the "L" with the length adjusted based on the participant's size. These defined a single-segment-trunk (acromia to iliac-crests), and upper-, mid- and lower-thoracic, and upper- and lower-lumbar segments (multi-segment-trunk). Participants were asked to stand from a bench set so the thigh was parallel to the floor and walk towards the left. Motion capture data were recorded (100Hz), and analysed to provide segmental angles (in relation to the absolute coordinate system) in each trial. Change in each segmental angle [flexion-extension (X), inclination (Y), rotation (Z)] was calculated between the start of the trial and the maximum angle (part A), and between the maximum angle and the end of the trial (part B). A repeated measures ANOVA was done between the single-segment-trunk angles and each of the other segments.

## Results

The statistical analysis showed no overall difference between the single-trunk-segment and each of the segments of the multi-segment-trunk for either part A or B. Looking at the independent components the analysis (figure 1) showed that for part A there was a mean difference between the single-trunk-segment and the upper-thoracic segment for component X of  $26.13^\circ$  (SE  $3.4^\circ$ ); and for part B a mean difference with the upper-lumbar segment component X of  $-8.06^\circ$  (SE  $2.1^\circ$ ), and the lower-thoracic component Z of  $-29.42^\circ$  (SE  $17.8^\circ$ ), which is double the difference with the other segments.

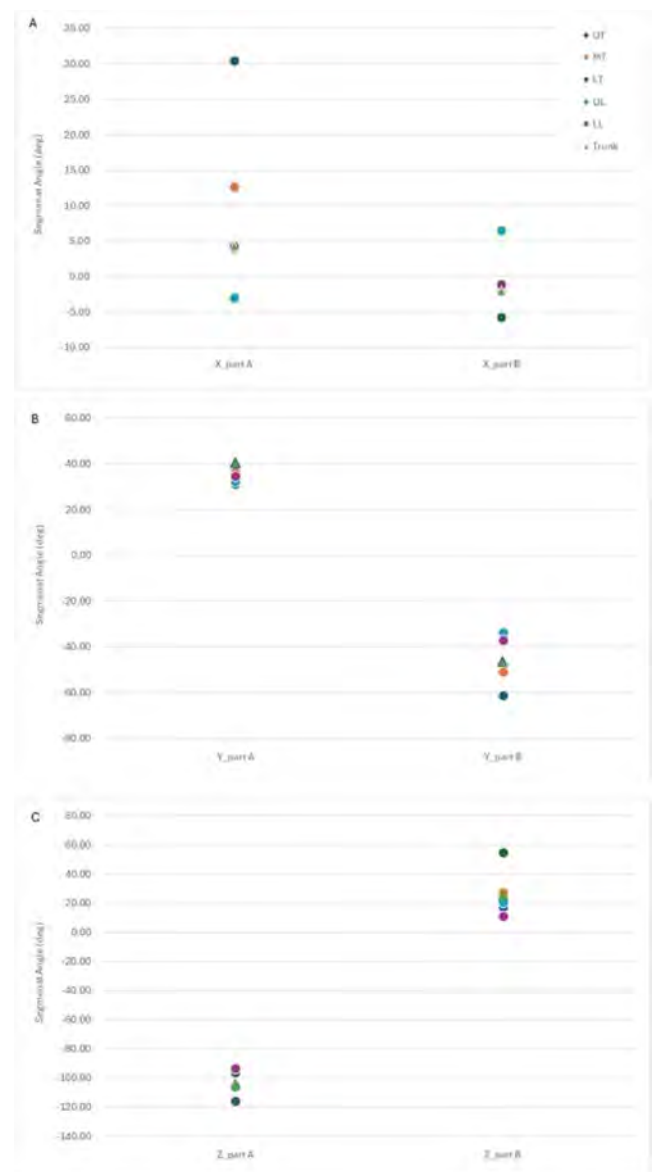


Figure 1 Segmental angle change

## Discussion

The single trunk model does quantify the general behaviour of the trunk during the tasks assessed. There is variation in the amount of movement occurring across the different segments, but how useful this additional information is, or what it may contribute to, is yet to be determined.

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## Longitudinal Evaluation of Functional Mobility in Different Knee Prostheses Using an Inertial Sensor

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## Introduction

Knee osteoarthritis is a progressive, multi-etiological degenerative musculoskeletal condition that primarily affects older adults, significantly limiting mobility and quality of life (1,2). Total knee arthroplasty (TKA) is performed to relieve pain and restore functional mobility (3,4). Various prosthetic designs, such as fixed-bearing (FB) and mobile-bearing (MB) prostheses, have been developed to improve joint mobility (5). However, aging and the progression of knee osteoarthritis may impact spatiotemporal gait parameters and functional mobility (6). Nonetheless, the longitudinal impact of these prosthetic designs on functional mobility remains unclear.

## Research Question

Do different types of knee prostheses (FB and MB) affect functional mobility during longitudinal assessment?

## Methods

Thirty-six patients who underwent unilateral TKA for advanced knee osteoarthritis were evaluated at 6 months and 1 year

postoperatively. Spatiotemporal variables (antero-posterior, latero-lateral, and vertical axes) were collected with an inertial sensor (G-sensor, BTS Bioengineering) positioned at L2 level during the Timed Up and Go (TUG) test assessment. We analyzed and compared the total time (in seconds) and sub-phases (time and velocity) of the TUG between the two types of prosthesis.

## Results

A two-way ANOVA indicated that both groups improved their TUG times after 1 year compared to 6 months; however, no significant differences were observed between groups ( $p > 0.05$ ). Although patients with MB prosthesis tended to perform turning tasks more quickly than those with the FB prosthesis after 1 year, this trend was not statistically significant (Fig. 1). Despite the lack of significant differences between groups, our findings could suggest potential advantages for MB prostheses due to their design, which allows for greater freedom of movement.

## Discussion

The results indicate that prosthetic design may influence specific aspects of functional mobility, particularly in tasks that require turning. Despite the absence of statistically significant differences, the consistent trend toward faster rotational performance in the MB prosthesis group suggests a possible biomechanical advantage associated with its design which might highlight their potential benefits in dynamic movement. Importantly, the use of wearable inertial sensors enabled detailed spatiotemporal analysis of functional tasks, providing precise, real-world insights into post-TKA mobility. This technology-driven approach offers a valuable scalable tool for long-term monitoring, making it an important addition to rehabilitation and prosthesis evaluation protocols. Future studies should investigate how these differences impact daily activities and quality of life.

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## Cerebellar transcranial direct current stimulation and gait training in children with Down Syndrome: preliminary results

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## Introduction

Neurological alterations associated with Down syndrome (DS) include diffusedysfunctions of brain activity and reduced cerebellar volume, resulting in compensatory motorpatterns and strategies that affect gait. 1 Transcranial direct current stimulation (tDCS) applied tothe cerebellum has shown promise as a therapeutic tool to optimize the results of gait training inchildren. 2,3 However, the scientific literature on its application in children with DS remainsscarce.

## Research Question

This study aims to analyze the effects of ten sessions of cerebellar anodaltDCS during treadmill gait training on functional mobility and gait in children with DS.

## Methods

Participants in this pilot study were assessed one week before, one week after, and one month after completing ten sessions of cerebellar anodal tDCS (1 mA) administered during 20 minutes of treadmill gait training. Data were collected using the Timed Up and Go (TUG) test and the 10-Meter Walk Test (10MWT), with the aid of the G-Sensor. The intervention was conducted at a frequency of five sessions per week over two consecutive weeks.

## Results

Four children (age:  $3.3 \pm 0.4$  years) participated in the study. The analysis revealed a reduction in the time taken to perform the TUG (before:  $28.7 \pm 5.3$  s, after:  $20.1 \pm 5.0$  s, and follow-up:  $13.4$

$\pm 2.5$  s,  $p < 0.001$ ), as well as increases in speed (before:  $0.82 \pm 0.32$  m/s, after:  $0.81 \pm 0.32$  m/s, and follow-up:  $0.73 \pm 0.07$  m/s,  $p = 0.003$ ) and cadence (before:  $108.6 \pm 33.2$  steps/min, after:  $143.2 \pm 24.8$  steps/min, follow-up:  $145.3 \pm 8.9$  steps/min,  $p = 0.01$ ). These results demonstrated satisfactory effects immediately after the intervention, which were maintained at the follow-up evaluation.

## Discussion

Changes in the gait of children with DS present a significant challenge for rehabilitation professionals. Interventions such as treadmill gait training have been proven effective in improving gait, although they often require a considerable number of sessions to yield significant improvements in speed and cadence.4 To our knowledge, these are the first results reported on the potential of cerebellar anodal tDCS to enhance the effects of gait training. The data suggest that the effects on the spatiotemporal gait parameters were sustained one month after the intervention, indicating that tDCS may be a promising approach to enhance rehabilitation outcomes in children with DS. However, further studies are needed to deepen the understanding of the mechanisms involved and to optimize this intervention.

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## Transcranial Direct Current Stimulation and Treadmill Training in Children with Spastic Cerebral Palsy: Predictors of Positive Effects

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## Introduction

Anodal transcranial direct current stimulation (atDCS) has gained prominence in the scientific literature due to its potential to enhance the effects of motor training. 1 When applied during treadmill gait training, atDCS has proven to be a safe and promising approach for gait rehabilitation in children with spastic cerebral palsy. 2,3 However, the variability observed in the results of previous studies underscores need to identify factors that can predict which children will benefit from this therapeutic modality.

## Research Question

This study aimed to identify the clinical and neurofunctional factors that can influence gait responses in children with spastic cerebral palsy who underwent atDCS during treadmill training.

## Methods

A retrospective study was conducted involving 148 children with spastic cerebral palsy, aged between 3 and 7 years, whose gross motor functions were classified at levels I to III of the Gross Motor Function Classification System (GMFCS). For inclusion in the analysis, children were required to have completed ten sessions of atDCS (1 mA) applied over the primary motor cortex or cerebellum during 20-minute treadmill training sessions, conducted at a frequency of five sessions per week. Assessments were performed before and after the intervention, using the Timed Up and Go test (functional mobility), the 10-meter walk test (gait velocity), and motor evoked potentials. Logistic regression analyses were employed to identify clinical, neurophysiological, and neuroanatomical predictors associated with treatment responsiveness.

## Results

The presence of motor evoked potentials during the initial assessment and subcortical lesions were associated with positive effects on gait speed and functional mobility, specifically in the intervention that utilized atDCS on the primary motor cortex. When treadmill gait training was combined with cerebellar atDCS, only subcortical injury emerged as a predictor of the effects on gait. Furthermore, GMFCS level III was identified as a significant predictor of functional mobility response, yielding superior results compared to children in levels I and II, especially during the application of cerebellar atDCS.

## Discussion

This preliminary study analyzed a substantial sample of children with spastic cerebral palsy who received transcranial stimulation in conjunction with treadmill gait training. Important predictors of

favorable responses to atDCS were identified. Consistent with prior literature, the presence of motor evoked potentials, indicating corticospinal tract integrity, was a critical determinant of positive responses to atDCS applied to the primary motor cortex. Conversely, children with moderate impairment of gross motor function, who exhibit postural instability, may benefit more from cerebellar atDCS. A better understanding of predictors of positive tDCS responses specificity and quality of clinical treatment recommendations.

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## RehaBot: Enhancing Cerebral Palsy Rehabilitation with a Chatbot and Assessment of Video-Based Activity Recognition

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## Introduction

Cerebral Palsy (CP) is a major cause of physical disability in childhood, with physical rehabilitation crucial for improving function [1]. Telerehabilitation is increasingly important, driven by the growth of ICT and the COVID-19 pandemic [2]. Chatbots offer a promising way to support home-based therapies and improve patient adherence [3-5]. RehaBot, a chatbot-based system, was developed to provide this support. Future versions aim to integrate automated activity recognition to enhance the platform.

## Research Question

Can a chatbot-based telerehabilitation system (RehaBot) effectively support home-based rehabilitation for patients with CP, and can activity recognition using video data be a feasible addition to this system?

## Methods

RehaBot's usability was assessed with 10 therapists and 10 patients using the System Usability Scale (SUS) and a supplementary questionnaire designed to evaluate RehaBot's specific features. RehaBot consists of a chatbot for patients (Fig. 1, left) and a web/mobile interface for therapists (Fig. 1, right), designed to facilitate home-based rehabilitation by providing exercise guidance and therapist monitoring. The chatbot delivers therapist-designed sessions with video demonstrations and collects patient feedback. The web interface allows therapists to create sessions, manage exercises, and track patient progress. To explore activity recognition, recordings from 16 healthy subjects performing 8 upper-limb activities from the VIDIMU dataset [6] were used. A fine-tuned Resnet3D model (R3D\_18) was evaluated.

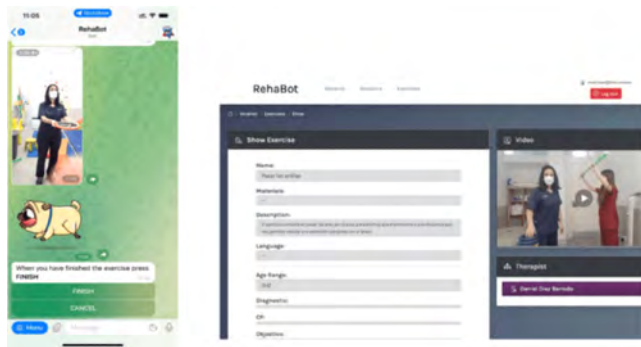


Fig 1. RehaBot Interfaces. Left: chatbot for patients. Right: web/mobile for therapists.

## Results

RehaBot demonstrated positive usability, with therapists and patients scoring 68.41 and 70.25 respectively on the SUS, despite having no prior training. Patients found the chatbot intuitive, particularly valuing the exercise videos and feedback features. Therapists appreciated the session creation and patient monitoring capabilities of the web interface. The fine-tuned R3D\_18 model demonstrated that raw video can be effectively used for activity recognition, achieving 96.30% accuracy and a 0.96 F1-score.

## Discussion

RehaBot shows potential as a tool for home-based CP rehabilitation, facilitating patient engagement and therapist collaboration. Integrating activity recognition could further enhance the system by enabling automated exercise assessment, personalized adjustments, and progress tracking. The performance metrics obtained with the R3D\_18 model suggests that video-based activity recognition is a feasible addition to the platform[7], offering a non-intrusive and simple way to measure the time spent by the patient performing the prescribed exercises[MM1]. Future work will focus on integrating and evaluating activity recognition within the RehaBot platform.

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## Comparative evaluation of monocular deep learning pose estimation and IMU-based systems for remote kinematic assessment

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## Introduction

Remote assessment of human motion is increasingly pivotal in clinical [1], sports [2], and rehabilitation contexts, particularly given the rise of telemedicine. While traditional motion capture systems deliver high-precision data, their dependence on expensive equipment and controlled laboratory conditions limits their broader application [3]. Advances in computer vision have enabled the development of monocular video-based 3D human pose estimation methods, which leverage ubiquitous camera technologies to offer cost-effective and accessible kinematic analysis [4]. This study systematically benchmarks joint angles derived from both video-based models and IMUs, addressing the gap in comparative evaluations under realistic, out-of-the-lab conditions.

## Research Question

Do the kinematics extracted from video-based pose estimation models match those of inertial sensors in capturing clinically relevant joint angles during everyday activities?

## Methods

Video recordings and IMU data collected from 16 subjects performing 5 daily-life activities from VIDIMU dataset [5] was used to evaluate representative movements. Joint angles were computed from the video-derived 3D keypoints. Then, the pipeline converted model outputs into a unified format, applied interpolation and smoothing filters to mitigate noise, and synchronized the time-series data with the reference IMU joint signals extracted from OpenSim [6]. To quantify the accuracy of the video-based estimations, two metrics were computed: RMSE to assess the magnitude of the deviations between both modalities, and Pearson Correlation Coefficient to evaluate the linear relationship between them. These metrics were calculated separately for each of the five selected activities, thereby providing a comparison of performance across diverse tasks.

## Results

In the analysis of five upper- and lower-limb selected tasks, performance varied across models. For *walk\_forward*, MMPose [7] achieved the lowest RMSE of  $6.54^\circ$  and highest correlation coefficient of 0.88. In *walk\_along* and *move\_right\_arm*, MotionAGFormer [8] excelled, recording error values of  $5.98^\circ$  and  $6.14^\circ$  with correlations of 0.90 and 0.92, respectively. In the *drink\_left\_arm* task, although BodyTrack [9] had the best correlation of 0.97, MotionBERT [10] delivered the smallest RMSE of  $8.39^\circ$ . For *tear\_both\_arms*, MotionAGFormer showed the lowest error  $4.99^\circ$  and highest correlation 0.98.

Table 1. Per-activity RMSE and PCC across all models

| Root Mean Square Error (RMSE) |                       |                 |                  |                  |                  |
|-------------------------------|-----------------------|-----------------|------------------|------------------|------------------|
| ID                            | Legend                | MotionAGFormer  | MotionBERT       | MMPose           | BodyTrack        |
| A01                           | <i>walk_forward</i>   | $8.12 \pm 3.42$ | $7.40 \pm 2.56$  | $6.54 \pm 2.64$  | $7.07 \pm 2.73$  |
| A03                           | <i>walk_along</i>     | $5.66 \pm 2.72$ | $7.09 \pm 2.54$  | $6.25 \pm 2.61$  | $6.02 \pm 2.26$  |
| A05                           | <i>move_right_arm</i> | $6.14 \pm 2.64$ | $13.47 \pm 3.90$ | $10.19 \pm 2.70$ | $13.02 \pm 3.02$ |
| A08                           | <i>drink_left_arm</i> | $8.94 \pm 3.21$ | $8.39 \pm 3.14$  | $8.49 \pm 2.62$  | $8.40 \pm 2.97$  |
| A13                           | <i>tear_both_arms</i> | $4.99 \pm 2.48$ | $16.79 \pm 6.42$ | $17.57 \pm 3.53$ | $7.61 \pm 2.55$  |

| Pearson Correlation Coefficient (PCC) |                       |                 |                 |                 |                 |
|---------------------------------------|-----------------------|-----------------|-----------------|-----------------|-----------------|
| ID                                    | Legend                | MotionAGFormer  | MotionBERT      | MMPose          | BodyTrack       |
| A01                                   | <i>walk_forward</i>   | $0.83 \pm 0.12$ | $0.87 \pm 0.10$ | $0.88 \pm 0.11$ | $0.87 \pm 0.12$ |
| A03                                   | <i>walk_along</i>     | $0.92 \pm 0.06$ | $0.89 \pm 0.07$ | $0.90 \pm 0.06$ | $0.91 \pm 0.05$ |
| A05                                   | <i>move_right_arm</i> | $0.92 \pm 0.06$ | $0.79 \pm 0.11$ | $0.87 \pm 0.06$ | $0.64 \pm 0.19$ |
| A08                                   | <i>drink_left_arm</i> | $0.91 \pm 0.13$ | $0.94 \pm 0.06$ | $0.93 \pm 0.05$ | $0.97 \pm 0.02$ |
| A13                                   | <i>tear_both_arms</i> | $0.98 \pm 0.02$ | $0.88 \pm 0.10$ | $0.93 \pm 0.05$ | $0.94 \pm 0.06$ |

## Discussion

Our findings highlight that model performance is intrinsically linked to task dynamics and architectural design. MotionAGFormer's hybrid design, merging transformer-based global context with GCN-driven local refinement, excels in capturing complex movements, while sequential lifting methods like MMPose perform better in steadier, predictable tasks. While MotionBERT is trained to recover 3D motion representations from noisy 2D inputs, it may inadequately generalize to fine-grained joint kinematics in dynamic tasks. These trade-offs in accessibility, inference speed, and performance emphasize the need to select models aligned with specific clinical applications. Additionally, our work affirms the viability of video-based approaches as accessible and cost-effective alternatives to IMUs for kinematic assessment in real-world telerehabilitation. Future work could integrate a comparison against gold-standard measurements.

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## Relationship between lower limb muscle size and gait pathology in children with cerebral palsy

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## Introduction

Gait impairments are common in children with cerebral palsy (CP), primarily due to neuromuscular problems such as spasticity, weakness, impaired control and secondary muscular deformities like shorter muscle lengths (ML) and reduced muscle volume (MV) compared to typically developing (TD) peers [1]. While studies have associated distal muscle morphology, such as the medial

gastrocnemius (MG) and tibialis anterior (TA), with gait impairments, there is limited research on proximal muscles like the rectus femoris (RF) and semitendinosus (ST) [2-6]. Understanding the relationship between muscle morphology across lower limb muscles and gait performance is essential to guide clinical decision-making and support targeted interventions aimed at improving gait outcomes.

## Research Question

Is gait pathology associated with the muscle size of four lower limb muscles in children with CP?

## Methods

Thirty-one children with CP (Gross Motor Function Classification System levels I-III), aged 4-15 years, were recruited. Overground walking at a self-selected speed was recorded using three-dimensional gait analysis. Gait pathology was quantified using the gait profile score (GPS) and gait variable score for the ankle (GVSa) and knee (GVSk), based on an age-matched TD reference group ( $n = 51$ ). ML and MV of MG, TA, RF and ST were assessed via 3D-freehand-ultrasound. MV and ML were converted to z-scores using anthropometric-related percentile curves of TD children [7], representing standardized muscle size deficits for each muscle. Lower limb muscle size deficits were defined as the sum of the 4 muscle-specific z-scores. Pearson or Spearman correlation coefficients, depending on data distribution, were calculated between muscle size deficits and gait scores using SPSS. Statistical non-parametric mapping identified regions in the gait cycle where MV or ML deficits correlated with knee or ankle kinematics.

## Results

A moderate negative correlation was observed between MG ML z-scores and GVSa, as well as between GVSk and ST MV z-scores (Figure 1a). In terminal stance, reduced MG ML and MV were associated with increased knee flexion. Additionally, lower MG ML associated with increased knee flexion at initial contact and during terminal swing (Figure 1b). No correlations were found between muscle size deficits and ankle kinematics.

Figure 1: Correlations between gait pathology and muscle size deficits.

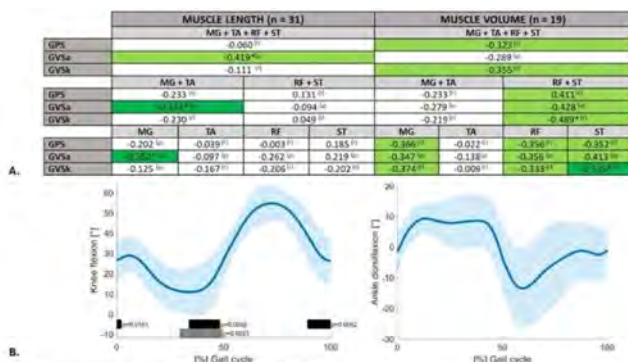


Figure 1: Correlations between gait pathology and muscle size deficits.

(A) Correlations of overall muscle size deficits (sum of z-scores of 4 lower limb muscles) and muscle-specific deficits with gait indices (GPS, GVSa and GVSk).

(B) Averaged knee and ankle kinematics with indications of significant regions in the gait cycle where there is a significant correlation between the muscle size deficits and sagittal kinematics (highlighted by horizontal bars: BLACK bars refer to associations with MG ML deficits and GREY bars refer to associations with MG MV deficits).

Abbreviations: ML = Muscle length; MV = Muscle volume; GPS = Gait Profile Score; GVSa = Gait variable score ankle; GVSk = Gait variable score knee; MG = Medial gastrocnemius; TA = Tibialis anterior; RF = Rectus femoris; ST = Semitendinosus; \* = Correlation is significant at 0.05 level; <sup>ns</sup> = Not significant; <sup>\*\*</sup> = Moderate correlation; <sup>\*\*\*</sup> = Strong correlation; p = Spearman correlation; r = Pearson correlation.

## Discussion

Shorter MG was associated with increased pathological knee and ankle kinematics, underscoring the importance of preserving MG length through interventions (e.g. stretching). The link between reduced ST MV and altered knee kinematics highlights the value of combining assessments of muscle morphology and gait. These findings emphasize the need for joint-specific evaluation rather than relying solely on global gait scores. As reduced MG ML and MV were linked to increased knee flexion during terminal stance, and muscle demands vary across the gait cycle, phase-specific interventions are essential. Early identification of muscle-specific deficits may support more effective gait rehabilitation and functional development in children with CP.

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## Temporal sequence of muscle activation in the glute bridge exercise

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## Introduction

The glute bridge (GB) is a foundational exercise strengthening lower posterior muscles and valuable for fitness, strength training, rehabilitation, and physical therapy. While the contributions of the muscles involved in GB have been investigated (Gasibat et al., 2023; Lehecka et al., 2017; Youdas et al., 2017), the temporal sequence of their activation remains unexplored.

## Research Question

To address this gap, we conducted a statistical analysis of 1,272 muscle onsets from surface electromyography (EMG) data. The central research question is: What insights can be gained from the statistical analysis of muscle onset timings in lower posterior muscles during glute bridges?

## Methods

We recruited five healthy participants (age 29–38, three male, two female). EMG data were collected using a FREEEMG system (BTS Bioengineering) at 1,000 Hz. Electrodes were attached to left and right rectus femoris (RF), gluteus maximus (GM), erector spinae (ES), and biceps femoris (BF), following the SENIAM guidelines (<http://seniam.org/>). Participants completed 1–4 cycles of 20 GBs, contributing a total of 159 GBs. The study was approved by the Ethics Committee at Setanta College Ltd. We processed EMG data and determined muscle onsets following Schuermans et al. (2017). Noise levels for each test cycle were estimated as the standard deviation of a three-second resting period with all eight muscles at rest. Muscle onsets were defined as the first 0.1s with signal consistently above  $3 \times$  noise level. The resulting 1,272 muscle onsets, shown in Fig. 1, were analysed with a linear mixed effects model (Bates et al., 2015) of the form  $\text{Onset} \sim \text{Muscle} * \text{Side} + (1 | \text{Participant/Rep})$ .

## Results

The fixed effects from Table 1 with confidence intervals (CIs) that do not span zero show that the typical GB is initiated by the left BF and both ESs and GMs, all of which activate in the first 0.1s. The right BF activates 0.1s later, followed within 0.25s by the left and finally,  $<0.1$  s later, the right RF. Random effects (Table 2) revealed that 13.4% of variance in onset timing was due to participants, 22.1% to repetitions, and 64.5% remained as residual variance (SD = 0.127s).

## Table/Figure

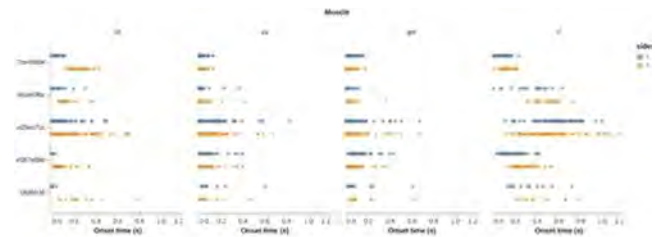


Figure 1: Muscle onset time (s) by participant, muscle, and side

Table 1: Fixed effect estimates, standard errors (SE), *t* statistic, and 95% confidence intervals (CI), rounded to three significant digits

| Effect    | Estimate | SE     | <i>t</i> | 95% CI Lower | 95% CI Upper |
|-----------|----------|--------|----------|--------------|--------------|
| Intercept | 0.0734   | 0.0288 | 2.55     | 0.0169       | 0.13         |
| ES        | 0.00285  | 0.0143 | 0.199    | -0.0251      | 0.0308       |
| BF        | 0.101    | 0.0143 | 7.08     | 0.0731       | 0.129        |
| RF        | 0.296    | 0.0143 | 20.7     | 0.268        | 0.324        |
| Left      | 0.0131   | 0.0143 | 0.917    | -0.0149      | 0.0411       |
| ES:Left   | 0.00235  | 0.0202 | 0.116    | -0.0372      | 0.0419       |
| BF:Left   | -0.146   | 0.0202 | -7.24    | -0.186       | -0.107       |
| RF:Left   | -0.0478  | 0.0202 | -2.37    | -0.0874      | -0.00824     |

Table 2: Random effects variance and standard deviation (SD), rounded to three significant digits

| Effect          | Variance | SD     |
|-----------------|----------|--------|
| Participant:Rep | 0.00556  | 0.0746 |
| Participant     | 0.00338  | 0.0582 |
| Residual        | 0.0162   | 0.127  |

## Discussion

Findings indicate that GB muscle activation sequence is remarkably consistent both within and across individuals. This contrasts with other posterior muscle exercises, such as prone hip extension (Bonacci et al., 2009; Carlos et al., 2024; Kiseljak & Medved, 2022; Lirio-Romero et al., 2021; Suehiro et al., 2014) where onset timing varies and has been controversially linked to low back pain or hamstring injury risk (Emami et al., 2014; Ryan & Bruno, 2024; Schuermans et al., 2017). The consistency in GB activation timing suggests it may provide a more stable basis for injury risk modelling and clinical assessment compared to other posterior chain exercises.

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## The Effect of Selective Dorsal Rhizotomy on the Incidence of Orthopedic Surgery: A Matched Cohort Study

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### Introduction

The effect of a selective dorsal rhizotomy (SDR) on the incidence of subsequent orthopedic surgery is unknown. Proponents of SDR often assert that the procedure lowers the need for orthopedic surgery, including surgery to address muscle contractures. There is no strong evidence to support or refute this claim. A prior multicenter study evaluating long-term outcomes with- and without SDR was inconclusive [1].

### Research Question

Does an SDR change the incidence of subsequent orthopedic surgery among children with cerebral palsy?

### Methods

This was a matched cohort study. Using historical data from a single center we identified two groups of individuals based on their history of SDR (Yes-SDR and No-SDR), matched at baseline on a large number of key clinical variables. We defined 10 orthopedic surgery categories that accounted for >95% of surgery recorded in our gait center database. We then fit Kaplan-Meier estimates of cumulative incidence for each surgery, stratified by SDR status. We chose to consider inter-group differences larger than 15% to be "clinically meaningful", though this is an arbitrary threshold.

### Results

We obtained excellent comprehensive baseline matching, including the overall probability of undergoing an SDR, which was identical at 90% (70% - 90%) in both the Yes-SDR and No-SDR groups.

Foot and ankle bone surgery and femoral derotation osteotomy exhibited 15% - 20% higher cumulative incidence after SDR, while rectus femoris transfer exhibited ~30% lower cumulative incidence [Fig. 1]. The remaining surgeries — including all surgeries for treating muscle contractures — were largely unaffected by SDR, exhibiting only small (< 15%) or insignificant ( $p > .05$ ) differences in cumulative incidence.

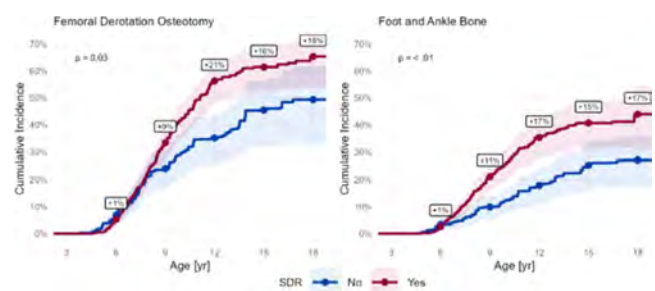


Figure 1. Kaplan-Meier curves for Femoral Derotation Osteotomy and Bony Foot/Ankle Surgery.

## Discussion

An SDR does not consistently lower the cumulative incidence of orthopedic surgery, and does not lower the rate of any surgery for muscle contracture. The lower incidence of rectus femoris transfers following SDR is consistent with the generally accepted mechanism for stiff-knee gait. Further investigation into the mechanisms underlying higher rates of bony foot surgery and femoral derotations is needed.

This study relied on observational data. Thus, the study is limited by the usual confounders in observational data including sample bias and follow-up bias. We have taken steps to adjust for the most obvious source of confounding through comprehensive baseline matching on a large number of indications for an SDR.

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## Surgical decision-making in children with walking impairments: comparing a surgeon's algorithm and two machine learning models

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## Introduction

Clinical gait assessments play a key role in guiding surgical decisions for children with walking impairments. These assessments typically involve reviewing videos, tables, and graphs of gait data, followed by discussions among clinicians about the most appropriate treatment. This process leads to significant variation in treatment decisions between surgeons. To reduce this variability, there have been efforts to establish consensus-based clinical guidelines [1-2]. Despite these efforts, what is stated in consensus papers may not align with actual clinical practices. Explicit treatment algorithms could help address this issue and provide several benefits such as (1) benchmarking various treatment strategies, (2) improving the training of new surgeons, (3) creating a permanent record of decision logic in the patient medical record, and (4) allowing better communication of treatment plans to patients and families. While machine learning (ML) algorithms have been successfully developed to predict treatment decisions based on historical data, their performance compared to clinician-derived algorithms remains unclear [3].

## Research Question

How do clinician-derived algorithms compare to automated ML algorithms in predicting surgical treatment decisions for children with walking impairments?

## Methods

A single surgeon explicitly described his clinical decision algorithm (Clinical) for determining two common surgeries: femoral-derotation-osteotomies and calf-muscle-lengthening. We then fit two ML models – classification and regression trees (CART) and Bayesian additive regression trees (BART) – to predict the treatment decisions for both surgeries based on over 7000 retrospective cases from our clinical gait center. We compared the sensitivity and specificity of the three models (Clinical, CART, and BART) to assess their utility for predicting the surgical treatments.

## Results

The BART models outperformed CART for both surgeries. The Clinical model performed reasonably well for calf-muscle-lengthening but exhibited significant imbalances in sensitivity and specificity for femoral-derotation-osteotomy.

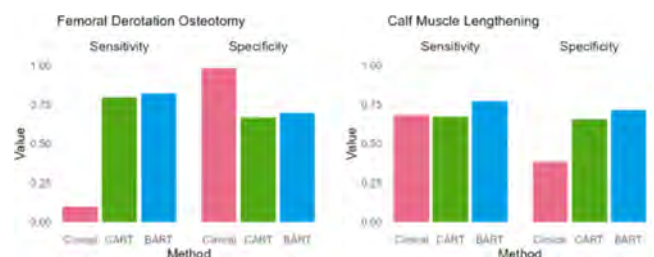


Figure 1. Performance of clinical and ML algorithms.

## Discussion

Our results suggest that clinician-derived algorithms may not align well with historical treatment decisions, which could be due to several factors. First, clinical decision-making varies widely among surgeons, so a single surgeon's algorithm may not reflect treatment decisions made by other clinicians. Additionally, while clinical algorithms can be represented as decision trees, this may not be reflective of how many surgeons – including the one in this study – make decisions. On the other hand, ML models, especially BART, performed well in predicting treatment decisions with sensitivity and specificity comparable to those seen in established medical diagnostic tests. However, the superior performance of BART comes at the cost of added complexity, which should be considered when selecting models for clinical use. These findings highlight the variability in clinicians' surgical philosophies and the potential for ML models to improve the consistency of treatment decision-making. The practical challenges of model complexity and implementation remain important considerations.

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## Disparities in gait between adolescent idiopathic scoliosis patients and healthy matched controls

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## Introduction

Adolescent idiopathic scoliosis (AIS), a three-dimensional deformity of the spine<sup>1</sup>, influences body posture and gait, leading to reduced health-related quality of life<sup>2-4</sup>. Inconsistent results have been reported regarding the gait pattern of patients with AIS compared to matched controls<sup>5</sup>. Moreover, knowledge on gait variability of AIS patients is scarce, while this provides relevant information on gait control and stability which may be useful for clinical decision making<sup>6</sup>.

## Research Question

Are there differences in gait in terms of joint angles, spatiotemporal parameters and variability between patients with AIS and healthy matched controls?

## Methods

Gait of 19 AIS patients indicated for brace treatment and 19 age and gender matched healthy controls was assessed at the Computer Assisted Rehabilitation ENvironment (CAREN; Motekforce Link, Amsterdam, the Netherlands). Measurements were performed after brace fitting but before brace treatment commenced and consisted of walking at comfortable speed without wearing a brace. After a six-minute familiarization period<sup>7</sup>, 250 steps were recorded. Spatiotemporal and variability parameters, and 3D pelvis and sagittal hip, knee and ankle angles were determined using custom-made algorithms programmed in Matlab (Mathworks V2016)<sup>8</sup>. Joint angles were normalized for time (gait cycle (GC) 0-100%). Variability was measured as coefficient of variation (CoV, SD/mean\*100). Spatiotemporal parameters, CoV and range of motion (ROM) in joint angles of the two groups, were compared using independent t-test or Mann Whitney U test. Statistical parametric mapping (SPM) was used to compare kinematic waveforms for both groups using independent t-tests.

## Results

AIS patients and controls were predominantly female (74%), having an average [SD] age of respectively 15 [1.5] and 15 [7] years and a similar average [SD] height of 1.7m [0.1]. Compared to controls, AIS patients walked significantly slower, with reduced cadence, shorter stride length and increased stride time. Additionally, AIS patients showed significantly increased variability in stride time and stride length. Furthermore, AIS patients showed significantly decreased ROM in pelvic obliquity (GC: 9-23%) and reduced internal pelvic rotation (GC: 0-5% and GC: 84-100%) (Figure 1). Patients walked with significantly reduced hip flexion (GC: 0-18% and 80-100%) resulting in limited sagittal hip angle ROM. Additionally, AIS patients had a decreased stance phase knee flexion (GC: 3-24%).

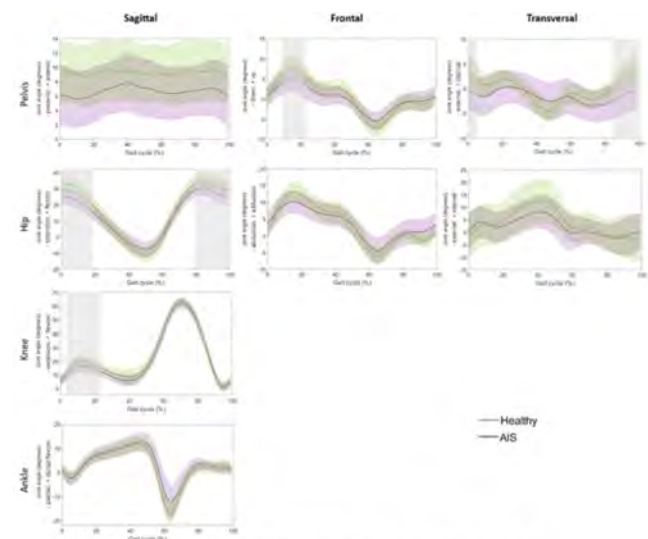


Figure 1: 3D joint angles (%) of the pelvis, hip, knee and ankle during the complete gait cycle (%) Presented as mean with standard deviations during complete gait cycle (0-100%). Green: Healthy controls, Purple: AIS patients, Grey shaded area indicates the part of the gait cycle (%) where the kinematic waveforms significantly differ between groups tested with Statistical parametric mapping (SPM; independent t-test). AIS: adolescent idiopathic scoliosis

## Discussion

Compared to controls, patients with AIS walk significantly different in terms of joint kinematics, spatiotemporal parameters and gait variability. The significantly reduced joint movements in patients with AIS suggests a stiffer gait, which may be a compensatory mechanism for maintaining balance or minimizing energy expenditure<sup>9, 10</sup>. With insight into the dynamic functioning, more tailored interventions can be provided aimed at improving mobility and thereby health-related quality of life<sup>11</sup>. Additionally, assessing gait before and after treatment interventions such as bracing may improve the understanding of the treatment effectiveness in correcting posture and gait.

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## Human movements during unsuccessful threat encounters

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## Introduction

Understanding movement behaviour under threat is vital for emergency response and managing vulnerability. Virtual Reality (VR) offers an immersive, ethical way to investigate these behaviours [1, 2]. Building on previous work using unsupervised machine learning for movement motif identification in VR [3, 4], here we investigate how environmental factors and threat characteristics, influence movement during unsuccessful threat encounters in VR.

## Research Question

How do environmental factors (e.g., scenario, visibility, safe-house presence), threat characteristics (e.g., type, behaviour), and personal factors, influence human movement patterns during unsuccessful threat encounters?

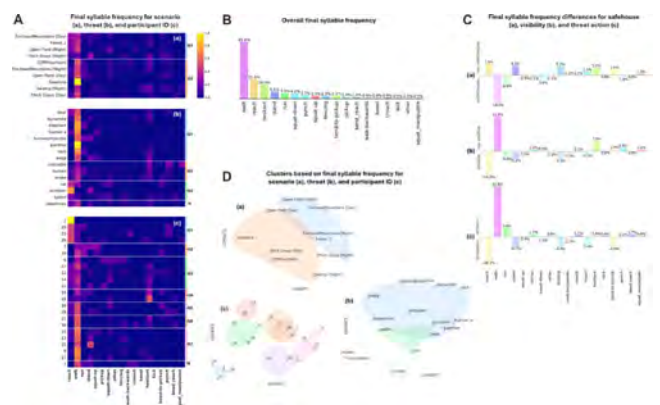
## Methods

Twenty-three participants performed a fruit collection task in a 5 x 10 m VR environment while avoiding threats [setup based on 2, 4]. A Qualisys motion capture (MoCap) system (200 Hz) recorded full-body movements. VR epochs were streamed wirelessly to an HTC Vive Pro headset. From the MoCap data, 311 spatio-temporal kinematic features were extracted. Unsupervised clustering (UMAP [5], HDBSCAN [6]) identified movement patterns within features [following 4]. A researcher assigned 22 "movement syllable" labels (e.g., walk, reach) to representative pattern segments. Subsequent frequency analysis and statistical testing focused specifically on the final movement syllable executed immediately prior to each unsuccessful threat encounter. Chi-squared tests identified experimental factors significantly influencing syllable distribution. Unsupervised clustering was applied to syllable frequency as a data-driven way to group external factors that similarly influence behaviour, as well as participants with similar behaviour.

## Results

We identified 221 unsuccessful threat encounter epochs. The top 5 syllables immediately preceding contact with the threat were: 'walk' (~43%), 'reach' (~15%), 'hesitant' (~11%), 'stand' (~5%), 'run' (~4%) (Fig.1B). Chi-squared tests ( $p < 0.05$ ) showed threat type, action, scenario, visibility, and safehouse presence significantly influenced final syllable frequency. There was also significant variability between participants. Notably, 'walk' (>18%)/'reach' (>7%) frequencies differed by threat action, presence of a safe house, and visibility (Fig.1C). Clustering syllable frequencies yielded distinct groups: 3 for threat type (differing by use of 'walk', 'hesitant', 'reach', 'punch'), and 2 for scenario. Similarly, 7 participant groups emerged, with patterns from high 'reach'/'walk' dominance to mixed 'hesitant'/'stand'/'run' use (Figs.1A, 1D). Findings highlight varied strategies influenced by conditions and individual tendencies.

Figure 1. (A) Final syllable frequency a. Scenario, b. Threat, c. Participant-ID, (B) Overall frequency, (C) Frequency differences a. Safehouse, b. Visibility, c. Threat behaviour, (D) Groups of external factors that similarly influence behaviour a. Scenario, b. Threat, c. Participant-ID.



## Discussion

Environmental context (e.g., scene type, visibility, refuge availability) and threat characteristics (type, actions) significantly shape movement strategies in virtual threat encounters. These factors influence final behaviours like walking, reaching, and hesitation. Infrequent 'punch'/'kick' syllables suggest some attempted active

defence before failed encounters. Clustering revealed distinct behavioural phenotypes linked to conditions and individuals, showing diverse coping strategies. This VR/MoCap/ML approach offers objective insights into behaviour in threatening situations, potentially enhancing safety training and crowd models.

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## Evaluating Phase Sensitivity and Cost Function Criteria in Predictive Simulation of Stumble Recovery

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## Introduction

Understanding stumble recovery mechanisms is crucial for preventing falls, improving assistive device technologies, and enhancing rehabilitation strategies. However, collecting empirical data on stumble recovery is both labor-intensive and ethically challenging. Predictive simulations provide a viable alternative in such cases. Previous studies indicate that stumble recovery strategies vary depending on the swing phase of the gait at which perturbation occurs [1, 2].

## Research Question

This study aims to address two core questions: (1) How does the timing of anterior perturbations within the swing phase affect the ability to recover from a stumble? (2) How do different physiologically based cost function criteria influence the predicted recovery strategy and timing?

## Methods

The "gait2392" musculoskeletal model was adapted by constraining motion to the sagittal plane and grouping muscles into 18 functional units. A reflex-based gait controller was employed for neural control. Stumble perturbations were introduced as a constant anterior force applied to the distal right toe at nine discrete swing phase intervals (10% to 90%, in 10% increments), generating nine distinct perturbation scenarios. Each was optimized using four physiologically motivated objective functions: minimization of head acceleration, muscle activation, cost of transport, and ground reaction force (GRF) jerk, resulting in a total of 36 predictive simulations. The simulations were executed in the SCONE platform [3] and evaluated based on: (1) recovery success, (2) classification of the recovery strategy employed (elevating, lowering, or delayed lowering), and (3) the time required to re-establish a stable gait cycle.

## Results

Simulation results revealed that perturbation timing significantly influenced both recovery success and strategy selection. Early swing phase perturbations elicited elevating responses, while late swing favored lowering strategies. Mid-swing perturbations, particularly at 50%, were the most destabilizing as none of the simulations achieved successful recovery at this phase. Among the cost functions, muscle activation was least effective, failing to produce viable solutions between 30–60% of swing. GRF jerk optimization yielded the fastest recovery times, whereas cost of transport resulted in the slowest stabilization.

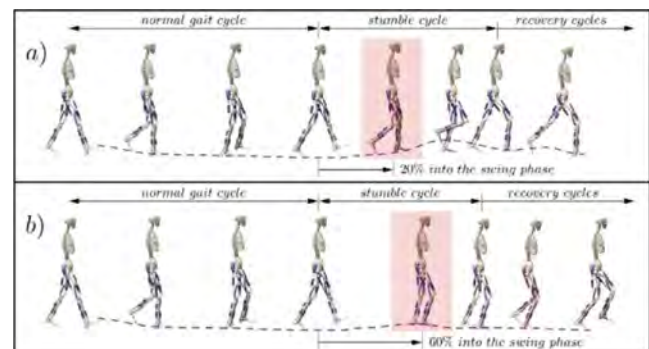


Figure 1: Two distinct stumble recovery strategies: a) elevating and b) lowering (red frames: perturbation onset)

## Discussion

This study enhances the understanding of stumble recovery through a predictive neuromechanical simulation framework. The recovery strategies were in line with the experimental work [1, 2]. The findings indicate that the model was unable to recover stumbles at mid-swing (at 50%) perturbations, deserving particular consideration in fall prevention training. Moreover, muscle fatigue has minimal influence on stumble response, indicating its limited impact on the nervous system's recovery prioritization. Future research will aim to enhance cost function formulations and validate the simulations through the similarity of the recovered gait cycle to normative gait data. The insights gained from this study will inform rehabilitation and fall prevention programs.

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## One sequence to show them all? Choosing the appropriate Euler sequence for shoulder kinematics in the Modified Mallet Scale

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## Introduction

The Modified Mallet Scale (MMS), which includes six specific movements, is a widely used tool for assessing shoulder movement in children with brachial plexus birth injury (BPBI) due to its several advantages [1, 2]. Although MMS provides clinically actionable scores, its observational nature introduces subjectivity and may fail to capture complex three-dimensional motions [3, 4]. As motion capture technologies gain prominence, a critical need emerges for consistent and clinically interpretable methods to report shoulder kinematics during MMS movements. However, the choice of rotation sequence can substantially influence the reported angles, limiting comparability and interpretability across studies [5].

## Research Question

This study aims to determine the optimal rotation sequence for accurate and consistent reporting of shoulder kinematics during the six MMS movements. We hypothesize that the most appropriate reporting method may vary by movement due to the diverse biomechanical demands of each task.

## Methods

Upper extremity kinematics from the unaffected limbs of ten children with BPBI (Narakas Type 2a) were captured using inertial measurement units (IMUs) and analyzed using OpenSim's IMU inverse kinematics tool. Three Euler rotation sequences (ZXY, XZY, YXY) were evaluated. Each method was assessed using two criteria: (1) occurrence of discontinuities, including gimbal lock and phase angle discontinuities (PADs), and (2) amplitude coherence, comparing reported angles with clinically expected values.

## Results

The ZXY sequence showed the highest rate of discontinuities, including gimbal lock during overhead movements (M1–M3) and PADs in M1. XZY and YXY showed no gimbal lock, though YXY exhibited minor PADs. In terms of amplitude coherence, XZY and YXY produced angles most consistent with clinical expectations for M1 and M2, while YXY was most suitable for M3 and M5. For M4, XZY demonstrated optimal coherence. ZXY was the only method to report clinically coherent values for M6. After analysing the performance of all considered rotation sequences in terms of discontinuity occurrence and amplitude coherence, the final recommendations for each MMS movement are presented in Table 1.

Table 1. Recommended Reporting Methods for each MMS movements. M1: global abduction, M2: global external rotation, M3: hand-to-neck, M4: hand-to-back, M5: hand-to-mouth, M6: hand-to-belly.

|                               | M1       | M2       | M3  | M4  | M5  | M6  |
|-------------------------------|----------|----------|-----|-----|-----|-----|
| Recommended Reporting Methods | XZY, YXY | XZY, YXY | YXY | XZY | YXY | ZXY |

## Discussion

These findings emphasize the critical impact of reporting method selection on the accuracy and clinical relevance of shoulder kinematic data. Notably, no single method proved universally superior; instead, movement-specific recommendations offer the best alignment with clinical needs. The adoption of such tailored reporting strategies enhances interpretability and facilitates standardized data sharing in both clinical and research settings. This work establishes a foundation for harmonizing shoulder kinematics reporting, ultimately supporting better assessment, treatment planning, and monitoring in BPBI and other upper extremity disorders.

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## Kinematic analysis of gait and levodopa response in young-onset and middle-aged Parkinson's disease

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### Introduction

Parkinson's disease (PD) is more commonly diagnosed in middle-aged individuals (MPD) between 50 and 69 years of age [1]. However, approximately 10% of cases occur before the age of 50, classified as young-onset Parkinson's disease (YPD). Although motor symptoms are shared across age groups, YPD patients may exhibit distinct gait patterns compared to MPD. Previous studies [2] have reported gait differences between these subgroups using portable systems without strict medication control. This study explores those differences using gold-standard 3D motion capture under both ON and OFF levodopa conditions. Characterizing gait kinematics with this level of precision may support the development of targeted clinical interventions for YPD.

### Research Question

Do patients with rare young-onset Parkinson's disease present gait kinematic and levodopa response differences when compared to typical middle-aged cases?

### Methods

Sagittal gait kinematics were analyzed in 20 PD patients: 10 with YPD ( $41.8 \pm 4.3$  years; disease duration:  $5.9 \pm 2.7$  years; UPDRS:  $27.5 \pm 11.3$ ) and 10 with MPD ( $57.7 \pm 3.8$  years; disease duration:  $7.8 \pm 4.3$  years; UPDRS:  $28.3 \pm 12.7$ ). Gait analysis was conducted using a motion capture system with a 7-camera Vicon system, 28 reflective markers and processed in Nexus using the Conventional Gait Model 2.3. Participants walked along a 10-meter track at a self-selected speed in ON and OFF conditions. Spatiotemporal variables were compared using a factorial two-way ANOVA in R (R Core Team). Sagittal kinematic waveforms were analyzed using Statistical Parametric Mapping (SPM) with a factorial two-way ANOVA in Matlab (The MathWorks, USA). Alpha was set to 0.05.

### Results

YPD exhibited a significantly higher percentage of single stance phase compared to MPD ( $p = 0.001$ ), indicating increased gait stability. MPD patients walked with increased hip and knee flexion and ankle dorsiflexion throughout most of the gait cycle duration (Figure 1). Levodopa significantly increased gait speed ( $p = 0.029$ ), enhanced hip extension and ankle plantarflexion during propulsion, reduced anterior pelvic tilt, and increased knee flexion during weight acceptance (Figure 1). The only interaction effect (group  $\times$  medication) occurred in the ankle dorsiflexion angle during the mid-balance phase, where increased dorsiflexion was observed only in the MPD group. Thus, medication primarily affected pelvic motion, whereas Parkinson's onset age influenced the distal joints (hip, knee, and ankle).

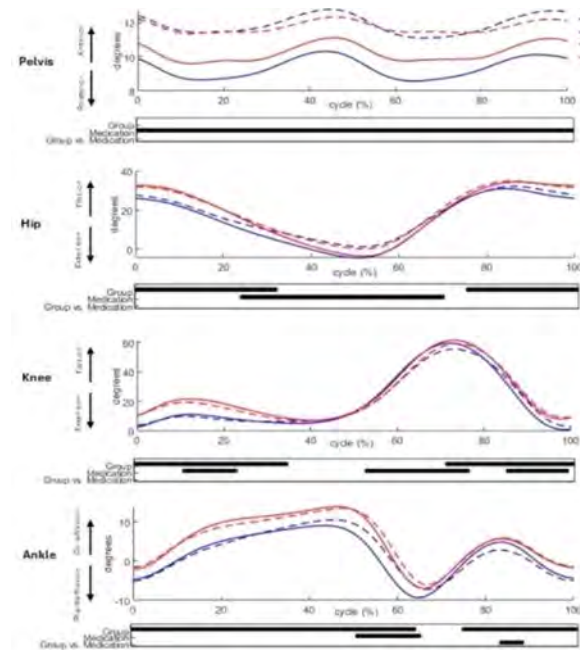


Figure 1: Sagittal averaged joint angles. The horizontal bar highlights regions where SPM {t} exceeds the critical threshold

### Discussion

Our findings highlight positive levodopa's effect on gait. Patients with YPD showed a larger single stance relative time, and thus a more stable gait, while MPD patients adopt a crouched gait. Levodopa primarily influenced pelvic and proximal joint kinematics in both groups, while distal joint behavior varied with age of onset. These results underscore the need for distinct rehabilitation strategies for individuals with YPD and MPD.

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### Clinical recommendations for orthotic provision and follow-up; a Norwegian evidence-based guideline

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## Introduction

In Norway, many people with cerebral palsy (CP) use orthoses. However, there is a lack of comprehensive evidence-based clinical guidelines focusing on orthoses (1), and variations have been found in provision practice between clinicians and healthcare regions (2).

An evidence-based guideline for diagnosing and follow-up of people with CP was developed by the Norwegian Quality and Surveillance Registry for Cerebral Palsy (NorCP) in collaboration with the Regional Competence Service for Habilitation and the Norwegian CP Association. The guideline encompasses 12 main themes, including recommendations to improve quality of orthotic prescription and follow-up. The target group is health professionals in municipal and specialist health services.

## Research Question

What is recommended practice for provision, follow-up, and use of orthoses among people with cerebral palsy?

## Methods

A working group of three certified prosthetist orthotists from major clinics formulated relevant questions with PICO format: (P) population CP; (I) intervention orthoses for ankle-foot, arm, trunk, hip, standing, dynamic textile, and functional electrical stimulation; (C) comparators; and (O) outcomes according to body functions and structure, activity, and participation. University librarians performed the literature search in June 2022. Inclusion criteria: clinical guidelines, systematic reviews, and articles with at least CP study population of 50%, published in English language, peer-reviewed journals between 2012-2022.

Two group members independently screened titles and abstracts, followed by full text reading and inclusion of studies, consulting a third member in case of disagreement. Included articles were assessed for methodological quality and risk of bias using AGREE II, AMSTAR2 and CASP. Evidence synthesis and development of clinical recommendations concerning each orthosis type involved structured panel discussions with health professionals and user representatives according to the Evidence to Decision framework. Grading was based on the strength and magnitude of the scientific evidence along with views and values of the panel members.

## Results

Based on 35 included studies, main and extended recommendations were developed for 5 of 7 defined intervention areas, and published in Metodebok.no

Methodological quality of the scientific evidence was overall high for the international clinical guidelines, and overall low for the systematic reviews, ranging from moderate to critically low. Main recommendations for ankle-foot orthoses were strong, whereas the scientific evidence-base was limited for the effect of arm and hand, trunk, dynamic textile, and functional electrical stimulation orthoses.

## Discussion

Whereas recommendations for ankle-foot orthoses were supported by sound scientific evidence together with professional and user experience, knowledge gaps were identified for the other areas. To develop clinically meaningful recommendations for practice based on research evidence it is essential to maintain good dialogue

with both professionals who apply the recommendations and users who have personal experience with assessments and interventions — within their unique contexts.

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## Kinematics and kinetics during sit-to-stand and stair-walking in healthy individuals using robot-assisted body weight unloading

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## Introduction

Body weight unloading (BWU) is the application of a vertical upwards force acting on the body centre of mass thereby reducing kinetic requirements during walking (8). Clinical neurorehabilitation often features elements of both gait training and tasks such as sit-to-stand and stair-walking (4). However, only a few studies have investigated the biomechanics of robot-assisted BWU during stair-walking and sit-to-stand (1). Such assessments may provide important clinical information regarding the task-specificity and relative loading-intensity during neurorehabilitation.

## Research Question

The aim of the present study was to investigate changes in movement patterns evaluated as kinematic and kinetic trajectories with increasing dynamic robot-assisted BWU in young healthy adults during stair-walking and sit-to-stand movements.

## Methods

Twenty participants performed standardised sit-to-stand movements and stair-walking barefoot at a self-selected speed in a 3-D motion-capture lab (VICON Inc., Oxford, UK, 8\*T40 and 8\*T20 cameras, 200 Hz sample rate) at 0, 10, 20, 30, 40, and 50% BWU. Ground reaction forces (GRFs) (AMTI, Watertown, MA, USA, 1000 Hz sample-rate), sagittal plane joint angles (ankle, knee, hip), body centre of mass (BCM) displacement and BCM velocity were obtained from the stance phase of stair-walking (ascending and descending)

and during sit-to-stand (eccentric and concentric phase). Kinematic and kinetic trajectories were time-normalised using linear interpolation and analysed using Statistical Parametric Mapping (5) to compare different BWU conditions through One-Way Repeated-Measures ANOVA univariate analyses.

## Results

Main effects of different BWU intensities were observed for all variables ( $p < 0.05$ ) both in sit-to-stand and stair-walking. Post-hoc analyses revealed progressive reductions in GRFs (Fig. 1) with increasing BWU ( $p < 0.01$ ), whereas changes in kinematic trajectories mainly occurred at 30–50% BWU ( $p < 0.01$ ). Both amplitude and temporal effects were observed for kinematic and kinetic signal trajectories, when nonlinear registration was performed (6).

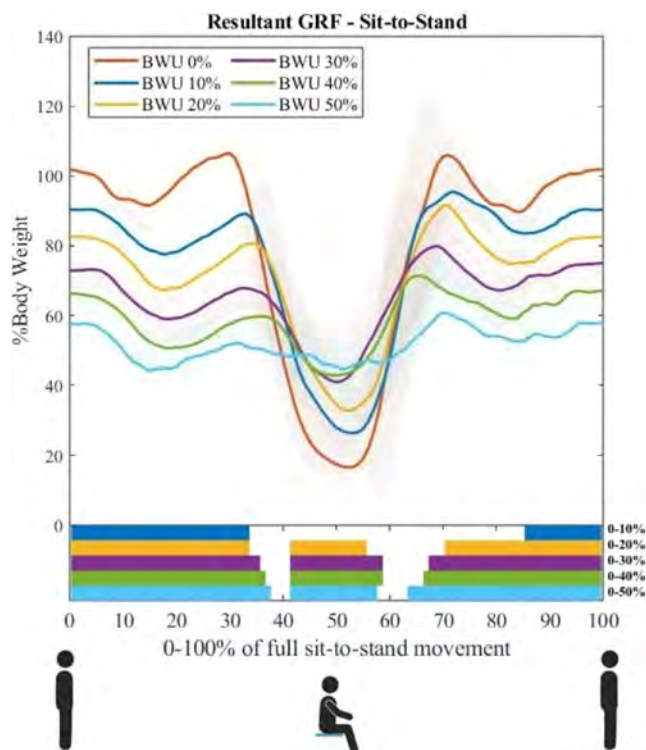


**Fig. 1** Resultant GRFs obtained during sit-to-stand at 0, 10, 20, 30, 40, and 50% BWU (group mean). Bottom panel: Bars represent phases of statistically significant signal-differences relative to 0% BWU.

## Discussion

As hypothesized, dynamic robot-assisted BWU progressively reduced kinetic requirements during stair-walking and sit-to-stand in young healthy adults. However, especially at higher BWU intensities, movement trajectory shapes were significantly affected, raising concerns about maintaining habitual movement patterns. These findings contrast with our previous research, which demonstrated preserved movement trajectories across various BWU intensities during overground walking in both healthy (8) and neurologically impaired individuals (7,8). This suggests that robotic unloading dynamics should be tailored for sit-to-stand and stair-walking tasks and incorporate modulated support throughout different movement phases (e.g., eccentric versus concentric). However,

while the observed temporal shifts were statistically significant, their clinical relevance remains uncertain. Nonetheless, reducing the kinetic demands through robot-assisted dynamic BWU during functional tasks may enable a greater training volume (9), which is important for facilitating experience-dependent plasticity in neurorehabilitation (2,3).

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## AI applications and data annotation practices in clinical gait analysis: Initial insights from a survey of ESMAC and GAMMA members

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## Introduction

Artificial intelligence (AI) holds great potential to support clinical gait analysis (CGA). To develop meaningful and transparent AI systems, it's essential to involve domain experts early in the design process to identify relevant applications and mark-up annotations (e.g., waveform features) that could serve as a basis for explaining these complex systems. A survey was conducted within the "TrustAI" [1] project at St.Pölten University of Applied Sciences, which develops explainable and interactive AI for CGA.

## Research Question

What are the most relevant AI applications in CGA, and is Baker's [2] annotation framework suitable to support transparent AI systems?

## Methods

Following an initial expert workshop, an online survey with 19 questions was collaboratively developed by experts in CGA, AI, and human-machine interaction, and distributed to members of the ESMAC and GAMMA societies. The survey included: 1) demographic questions, 2) rating of seven suggested AI applications in CGA, 3) assessment of Baker's annotation framework, and 4) open-ended questions on additional AI applications and current annotation practices.

## Results

The survey was completed by 30 individuals (only four responding anonymously). The respondents reported an average of 12.23 years of experience (range: 2–30 years; SD: 7.95) and working in both clinical settings and research ( $n = 26$  each), with fewer involved in sports movement analysis ( $n = 5$ ) or other areas ( $n = 1$ ). Participants regarded all seven proposed AI applications as very important, with "predicting outcomes" and "identifying waveform-specific gait patterns" rated highest (Figure 1A). Participants suggested additional AI applications, including data exploration (e.g., clustering), personalized treatment planning, error detection, event detection, and markerless motion capture. 18 participants were aware of Baker's annotations, and 9 had already used them. Most participants rated the annotations as very suitable for annotating data (Figure 1B). Users familiar with the annotations were more positive, while those without experience were more skeptical. Two participants who rated them unsuitable utilized textual descriptions. Most use textual descriptions ( $n = 11$ ) or custom annotations ( $n = 9$ , e.g., [3,4]), while only three currently use Baker's annotations. While many participants found all annotations useful, some questioned "within normal limits", "abnormal slope", and "other feature", or suggested adding a "symmetry" annotation. Two participants suggested alternative annotation frameworks, proposed by Chia et al. [3] and Sangeux et al. [5]. Participants also raised concerns about AI bias and data privacy in clinical applications.

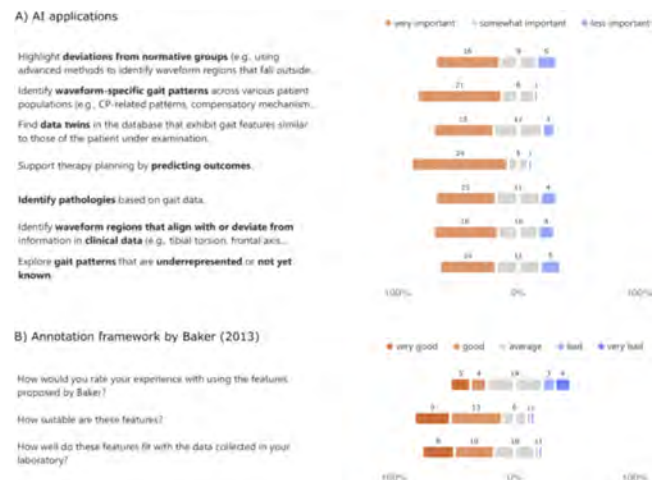


Figure 1. Aggregated survey results.

## Discussion

The survey confirms the high relevance of AI in CGA and indicates that it is largely accepted in the clinical community, with an awareness of its potential benefits. We conclude that Baker's framework offers a useful representation for annotating gait signals and could serve as a solid basis for future transparent AI systems. However, integrating this framework into an AI system remains an open question, which we will explore in future research.

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## Walking conditions change biomechanical gait signatures

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## Introduction

Gait signatures are unique movement patterns that reflect individual differences in walking and can be identified *within* various walking conditions (e.g., different speeds) [1]. At self-selected speeds, walking patterns allow differentiating individuals with >99% accuracy in both healthy (671 individuals) [2] and clinical (2,092 individuals) [3] cohorts. This highlights their potential for individualised gait assessments and interventions. However, it remains unclear whether gait signatures generalise *across* different conditions; specifically, if a signature obtained at preferred walking speed (PS) and cadence (PC) remains similar in other walking conditions.

## Research Question

To what extent are gait signatures invariant to changes in walking speed and cadence?

## Methods

We analysed bilateral 3D ground reaction force (GRF) signals from treadmill walking in 18 neurotypical adults [4], recorded at PS and three standardized speeds (0.4, 0.8, 1.1 m/s), each combined with five cadences (−10% to +20% of PC). Linear support vector machines (SVMs;  $C=1$ ) were used for person classification. For each condition, 40 gait trials per individual were randomly selected from a 5-min walk for training and 10 for testing. As a baseline, separate SVMs were trained and tested within each condition. Subsequently, the SVM trained on PS&PC data was tested across all walking conditions. GRF signals were time-normalised to 101 points and max-normalised per trial. For visualisation, t-distributed stochastic neighbor embedding (t-SNE) was applied to the mean test data per individual and condition.

## Results

The classification accuracy was near perfect (98.3–100.0%) when the SVMs were trained and tested within the same walking condition (dotted bars, Figure 1B). Accuracy decreased when SVMs trained on PS&PC were tested across other conditions (solid bars, Figure 1B). The smallest decrease occurred at 1.1 m/s, close to individuals' mean preferred speed ( $1.09 \pm 0.12$  m/s [4]). Accuracy declined further at slower speeds: moderately at 0.8 m/s and substantially at 0.4 m/s. The t-SNE plot (Figure 1A) illustrates why the generalisation performance drops. The preferred gait used for training the SVMs (PS&PC, red) is predominantly in the vicinity of 1.1 m/s gait (blue), less so of 0.8 m/s gait (green), and very distant from 0.4 m/s gait (purple). In most cases, individual gait across cadences is grouped together (different hues of a symbol).

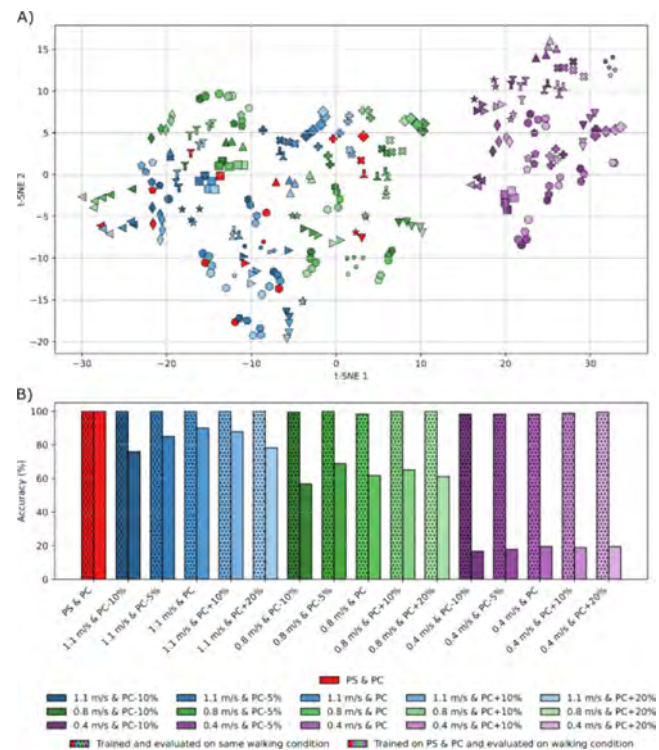


Figure 1: A) t-SNE plot showing individual gait signatures across walking conditions. Each symbol type represents an individual, each colour a different speed, and each hue a different cadence. B) Performance evaluation across conditions.

## Discussion

Our results show that gait signatures are highly individual within walking conditions (consistent with [1]) but change across conditions. This indicates that gait signatures are not invariant to changes in speed and cadence, being primarily dependent on speed. Reliable applications in clinical settings may require condition-specific modelling or normalisation strategies to account for variability in walking conditions.

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## Acceptability of wearing IMU sensors in a home environment in stroke patients

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## Introduction

Integrating rehabilitation into daily life and directly in the patient's home is key to promoting upper limb use and preventing decline in function. Movement sensors can facilitate this process by enabling remote monitoring and feedback. Their success depends not only on data quality, but also on patients' acceptability to wear the sensors. Previous studies in hospital settings reported high usability (i.e., easy to attach, easy to use), but wearability (i.e., comfort), which is critical for using sensors in the home, is unclear.

## Research Question

What is the acceptability (usability, wearability and willingness) of using movement sensors in a home-like rehabilitation setting for people with stroke?

## Methods

Eleven participants in the chronic phase after stroke were recruited from a rehabilitation hospital and outpatient clinic in Oslo, Norway. They were observed in their home or in a home-like setting in the hospital while attaching/detaching sensors themselves and performing Activities of Daily Living. The participants received help if they could not fit all the sensors independently. Nine IMU sensors (Xsens Technology B.V., Enschede, The Netherlands) were placed on their trunk (Th8 vertebra), shoulders (scapular spine), upper arms (lateral mid-humerus), forearms (mid-ulnar), and hands of both paretic and non-paretic limbs. Acceptability was assessed with a custom-made questionnaire that captures usability, wearability and willingness to use sensors. This was derived from a combination of the System Usability Scale (SUS) (Brooke, 1996) and the Auepanwiriyaikul et al. (2020) questionnaire. The questionnaire had 14 items scored on a 1-to-5 Likert scale.

## Results

Two participants managed to attach the full set of sensors themselves, while 9 participants were able to attach and detach the sensors only on their paretic limb. Participants reported high scores for wearability (comfort [mean = 4.18, SD = 1.25] and

unrestricted movement [mean = 4.46, SD = 1.21]) and for willingness to wear the sensors (continuously, when deemed beneficial [mean = 3.82, SD = 1.33] and when recommended [mean = 3.82, SD = 1.25]). They reported lower scores for usability (ease of attaching/detaching [mean = 1.91, SD = 1.22], independent use [mean = 2.36, SD = 1.63] and ease of daily integration [mean = 2.82, SD = 1.40]).

## Discussion

The results of this study show usability issues with sensors, particularly with attaching and detaching the sensors themselves. Although usability concerns, participants found the sensors comfortable to wear and they did not limit their movements. The participants further expressed willingness to wear them continuously if deemed beneficial for rehabilitation. This suggests that the ease of use of the sensors should be improved to enhance patients' acceptability. Practitioners should think carefully about the necessary amount and usability of future sensors. These findings highlight the importance of tailoring rehabilitation tools to patients' preferences. Personalizing sensors according to the patients' needs could increase long-term adherence and improve rehabilitation outcomes.

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## IMU-based balance assessment: alternative or complementary to gold standard GRF-based posturography?

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## Introduction

Inertial measurement units (IMUs) have been proposed as a portable alternative to force plates for posturography, being the most common sensor location at L5 to approximate centre of mass (CoM) acceleration [1,2]. Based on the inverted pendulum model of quiet standing [3], trunk acceleration (ACC) in the horizontal plane is proportional to the difference between CoP and CoM position. Therefore, ground reaction force (GRF-) and IMU-based metrics may have a similar mathematical formulation but quantify different aspects of posture, hindering their direct comparison. Before attempting to provide a physiological interpretation of IMU-based metrics, comparative sensitivity of GRF- and IMU-based metrics to testing conditions must be assessed, and IMU acquisition and processing protocols require standardisation. This work compares IMU- and GRF-based metrics of the postural response to varying visual and surface conditions in healthy individuals also analysing sensitivity of results to different filtering of IMU signals.

## Research Question

Is IMU-based posturography an alternative to GRF-based posturography?

## Methods

Twenty-one healthy participants ( $24 \pm 3$  years) were asked to maintain static posture under 8 different randomized conditions (4 support surfaces: without-foam, soft-foam ( $E = 33\text{ kPa}$ ), medium-foam ( $E = 52\text{ kPa}$ ), rigid-foam ( $E = 139\text{ kPa}$ ); 2 visual conditions: eyes-open, EO, eyes-closed, EC). Posture data were collected using a force platform (Bertec, USA, 800Hz) and an inertial sensor at L5 level (Cometa, Italy, 200Hz) [1,2]. COP signals were lowpass-filtered at 10Hz. Based on literature review [2], ACC were lowpass-filtered at: 0.5Hz, 3.5Hz, the maximum frequency containing 95% of the signal power ( $f_{95\text{max}}$ ),  $2 \cdot f_{95\text{max}}$ , 50Hz. Time- and frequency-domain postural parameters [1] were extracted from both COP and ACC with different filtering. As distribution was not normal, Scheirer–Ray–Hare test (significance level, 0.05) was applied to test the effect of surface and visual condition on COP and ACC.

## Results

$f_{95\text{max}}$  for IMU-signals resulted 5Hz. Correlations between GRF- and IMU-based metrics resulted weak to moderate ( $0 < |\rho| < 0.7$ ): 3.5Hz- and 5Hz- low pass filtered ACC signals exhibited the highest number of moderate positive correlations ( $\rho > 0.40$ ). A 5Hz cut-off frequency resulted the best compromise to allow noise reduction maintaining 95% of the signal power. Both GRF- and IMU-based metrics showed increased postural oscillations on foam surfaces, but opposite behaviours in frequency, with no significant difference among foams. GRF-based metrics highlighted higher postural oscillations under eyes-closed conditions, especially on foam, whereas IMU-based metrics showed no significant change except range and root mean square in the medio-lateral direction that decreased with eyes closed.

## Discussion

Results highlighted differences in GRF- and IMU-based posturography supporting the hypothesis that GRF- and IMU-based metrics target different specific manifestations of postural control. According to the inverse pendulum approximation [4] i) GRF-based metrics target postural adjustments (i.e., the trajectory of CoP); ii) IMU-based ones postural control performance (i.e., the CoM-CoP distance, proportional to CoM-ACC [4]).

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## A digital twin approach for monitoring neuro-motor developmental trajectory in preterm infants

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## Introduction

Preterm births lead to an increased risk of neuro developmental disorders (NDDs) [1]. Early longitudinal monitoring of motor development, strongly associated with neurodevelopmental outcome [2], is crucial for a timely identification of deviations from the reference path and the prediction of possible NDDs. However, the current understanding of the causal pathways through which motor difficulties emerge and evolve is limited by the lack of longitudinal, quantitative, and interpretative measures for infant motor development, and a complex multidisciplinary examination of medical history is needed. To overcome these limitations, we propose an approach based on Digital Twins (DTs) [3,4] and quantitative interpretative metrics for motor assessment [5,6] to support holistic longitudinal evaluations of infant development.

## Research Question

Can digital health technologies support NDDs prediction in preterm infants?

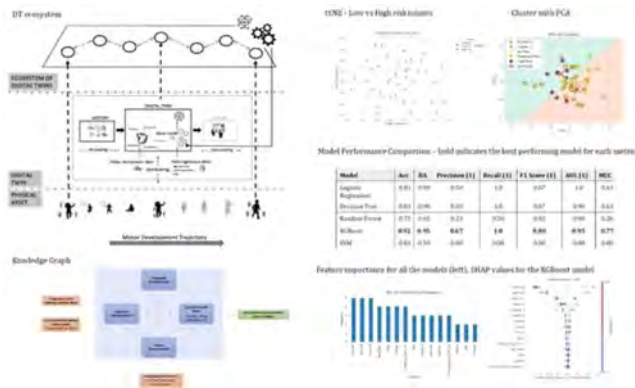
## Methods

The designed DT framework integrates i) heterogeneous data collection, ii) data processing for quantitative motor assessment [5,6] and iii) advanced AI methods for an efficient holistic data analysis (Fig.1). Each DT manages available data and is dynamically updated, building a complete history of the infant's state. Longitudinal clinical data (i.e., pregnancy-, neuroimaging-, growth-, nutritional-data, developmental status) are collected at term equivalent age, T0, and at 3-, 6-, 12-, 18- and 24-months corrected age (CA), matching the timing of clinical follow-ups. Longitudinal data measuring motor functions are acquired on specific milestones: i) video-based recordings of spontaneous movements [6] (T0 and 3months), ii) sensor-based [5] assessment of motor performance (sitting posture, independent posture, independent gait) from 6 to 24months CA, when present. Data are stored according to Fast-Healthcare-Interoperability-Resources [7]. A domain-specific knowledge graph was defined to map interactions among pieces of information, allowing higher-level analyses. A first case study application was conducted exploiting data presented in [5] (longitudinal clinical data and sensor-based gait data at 24months,  $n = 46$ ) to demonstrate the range of possible analyses. Data were normalized (Min-Max Scaler) and automatically analysed using: i) t-distributed stochastic neighbour embedding (t-SNE); ii) clustering (k-means,  $k = 2$ ; Kernel Density Estimation); iii) classification (list in Figure 1, train/test: 75/25%) and SHAP analysis.

## Results

Figure 1 shows i) on the left, the DT ecosystem and the Knowledge Graph, ii) on the right, case study results. Decreased early motor complexity resulted among the most relevant biomarkers of motor delay in preterm toddlers.

Figure 1



## Discussion

Case study results confirmed the feasibility and the potential of the approach. Complexity results, in agreement with literature on gait [5] and other tasks [8], reinforced the vision of a longitudinal perspective that overcomes task specificity and considers age-appropriate tasks within a unified motor trajectory. When fed with longitudinal data, the DT will enable the identification of interpretative quantitative biomarkers of NDDs, advancing our understanding of the factors that contribute to long-term consequences in preterms, enabling effective, targeted interventions.

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## An IMU-based ambulatory assessment for spinal decompressive surgery

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## Introduction

Lumbar spinal stenosis (LSS) is a common cause for chronic pain and disability. Treatment begins with conservative options, but often requires surgery. Globally, outcomes seem better for surgical neural decompression than for non-operative treatment. Recent randomized controlled trials have produced perceived conflicting results [1]. Given the significant clinical and economic implications of LSS treatment [2], tools to evaluate the effectiveness of treatment options for this population are a matter of priority [1].

In addition to radiological and clinical assessment, methods for the quantification of functional impairment are necessary. IMU-based methods allow to quantify posture and motor performance for routine clinical application, due to their portability and ease of use.

The aim of the present work is to present an IMU-based protocol for in ambulatory functional assessment and characterize the pre- and post-intervention functional alteration of LSS patients undergoing instrumented (Mid-F) and non-instrumented (Mid) laminectomy.

## Research Question

Can functional IMU-based ambulatory assessment characterize pre- and post surgical motor function in LSS patients?

## Methods

Eighty patients with clinical and radiological diagnosis of symptomatic degenerative LSS, for whom surgical decompression was prescribed (Mid or Mid-F based on standard clinical practice), were recruited. On hospital admission (T0) and 90 days post-op (T90), patients were assessed by physical therapist using 3 triaxial IMUs (MetamotionR, mBientLab, USA) for: 1) Posture (Eyes Open and Closed, 60 s each); 2) Normal Walking (15m 3 times); 3) Tandem Walking (15m). EO and EC postural parameters [3], NW and TW temporal parameters and variability, and recurrence and complexity indexes [4-5] were calculated and statistically compared (Mann-Whitney, 0,01 significance) pre-op for LSS patients vs healthy adults at T0, and for Mid-F vs Mid patients at T90.

## Results

At T0, in PSS vs healthy: postural COM-CoP oscillation increased for EO in ML, for EC in AP direction too; NW resulted slower, with increased DS, Stance, and temporal variability (Figure a), and increased recurrence (Figure b); TW resulted slower, with increased temporal variability and recurrence, and decreased entropy in all time scales. At T90, postural parameters improved in Mid, but not in Mid-F for EO, but not for EC; in NW, a general normalization of Mid-F (blue) and Mid (red) parameters was observed (Figure c and d); in TW, Mid-F temporal parameters resulted normal, while Mid still showed some increased variability, and both Mid-F and Mid recurrence and entropy improved, but still differed from normal.

Figure 1 consists of four panels, (a) through (d), each showing a schematic representation of an experimental paradigm. Each panel features a central target (red) and a surrounding area (green) containing distractors. The panels are labeled as follows: (a) Visual search, (b) Visual search with a distractor, (c) Visual search with a distractor, and (d) Visual search with a distractor. The panels are arranged in a 2x2 grid. The top row (a and b) shows a central target (red) and a surrounding area (green) containing distractors. The bottom row (c and d) shows a central target (red) and a surrounding area (green) containing distractors. The panels are labeled as follows: (a) Visual search, (b) Visual search with a distractor, (c) Visual search with a distractor, and (d) Visual search with a distractor.

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Restricting the forefoot did not significantly impact step width compared to NOR walking on either ground type, but significantly decreased step length on level ground (Fig. 1). The restriction also significantly changed the range of ankle flexion in the sagittal plane, knee and hip abduction, and knee rotation.

There was only a significant interaction effect between ground and restriction type on step period.

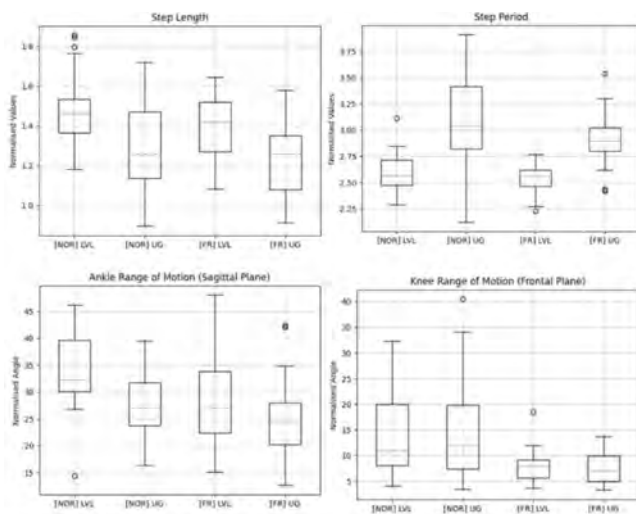


Figure 1: Box plots showing the variables with the most prevalent changes between conditions. Key: NOR = no restriction, FR = full restriction, LVL = level ground, UG = uneven ground

## Discussion

Walking on uneven ground and walking with forefoot motion restricted both cause changes in a person's gait compared to walking with an unrestricted foot on level ground. The implication for prosthetic design is that the inclusion of a more flexible forefoot could be beneficial in reducing superfluous movements below the hip and improving walking efficiency.

Further investigation into how these variations in gait characteristics affect walking efficiency may provide more detailed insight into the power and stability functions of the forefoot.

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## Recovery of Muscle Fatigability After a Progressive Exercise Task in Children in Children with Cerebral Palsy

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## Introduction

The experience of fatigue limits activity and participation in youth with cerebral palsy (CP).<sup>(1,2)</sup> Its improvement is often a high-priority goal.<sup>(3)</sup> Muscle fatigability in youth with CP has been documented during walking tasks and intense exercise,<sup>(4-6)</sup> Recovery has not been well documented. Knowledge of recovery is important in the design and success of rehabilitation protocols.

## Research Question

Is recovery of muscle fatigability in youth with CP after a progressive exercise task similar to the recovery in typically developing (TD) peers?

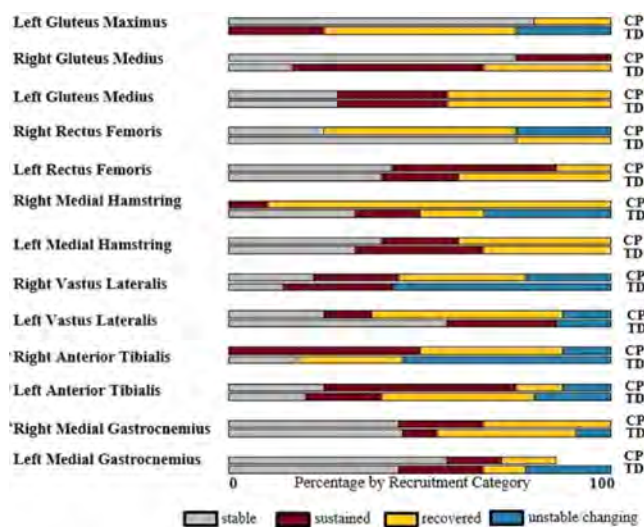
## Methods

Youth with CP and age- and sex-matched TD peers completed a four-phase testing protocol on an instrumented split-belt treadmill with shoes, without orthotics: 1) 8-minute level walk at 85% of self-selected over-ground speed (pre-exercise); 2) 30-minute rest; 3) progressive maximal exercise test using a modified-Bruce protocol; 4) repeat of phase 1 (post-exercise). Surface EMG were collected from 7 lower extremity muscles bilaterally throughout all phases. Median frequency (MF) of the power spectrum was computed by stride for each muscle. Two 60-second epochs of post-exercise (30-90 and 420-480 sec) were evaluated for evidence of fatigability (drop in MF) and recovery (change from drop in MF) based on: a) percent change and b) the percentage of strides that fell below the pre-exercise 95% confidence interval (CI). Four categories of muscle recruitment were used to characterize fatigability based on MF: stable (<5% strides below CI both epochs); sustained ( $\geq 5\%$  strides below CI both epochs), recovered ( $\geq 5\%$  strides below CI 1st epoch only), and unstable/changing (variable). Paired T-tests assuming unequal variances were used to test differences in percentages between groups by category.

## Results

Thirteen youth with CP (7 GMFCS I; 6 GMFCS II; mean age (14.7 years)) and thirteen age- and sex-matched TD peers participated. No differences between groups were present in age, height, weight, or BMI. 62% of youth with CP and 44% TD peers exhibited signs of fatigability in the early post-exercise epoch. Both groups exhibited signs of recovery between early and late post-exercise. The CP group demonstrated less percent change in MF than TD group (5.3% CP vs 10.9% TD;  $p < .01$ , effect size.851). A higher proportion of children with CP had sustained fatigability compared to TD peers (.328 CP vs.188 TD;  $p < .02$ ). Distal muscles demonstrated a greater percentage of sustained fatigability in the CP group. Figure 1.

Figure 1: Post-Exercise Recruitment Category Percentages by Individual Muscles and Group



## Discussion

Both groups exhibited signs of fatigability as measured by MF in individual muscles immediately after an intense progressive exercise task. Sustained fatigability was more prevalent in youth with CP. The magnitude of fatigue of individual muscles in the initial post-exercise epoch appears to influence the time needed for recovery. Distal muscles tended to exhibit less recovery. Further study is needed to understand the impact on rehabilitation protocols.

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## Combining immobilization and activity to influence calf muscle morphology in children with cerebral palsy: Randomized controlled trial using 3D ultrasound

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## Introduction

Children with cerebral palsy (CP) frequently develop muscle contractures due to neuromuscular impairments and altered musculoskeletal morphology [1,2]. Orthotic treatment using ankle-foot orthoses (AFOs) worn for extended periods is commonly used to apply sustained stretch to the spastic calf muscles [3]. While this strategy may promote muscle lengthening, prolonged immobilization can also contribute to muscle atrophy and reduced fascicle length due to disuse [4].

## Research Question

Does orthotic treatment regime combining immobilization using AFO with daily phases of permitted activity lead to a superior effect on spastic calf muscle morphology and function compared to continuous AFO immobilization?

## Methods

Fourteen ambulant children with CP and equinus deformity (8 unilateral, 6 bilateral; mean age  $9.93 \pm 3.0$  years; GMFCS level I = 10, level II = 4) participated in this randomized controlled trial. Participants were assigned to either a control group treated with AFOs for 23 h/day or an intervention group using the same AFOs at night and for 6 hours during the day, combined with a foot stabilizing orthosis 10 h/day to allow ankle mobility. The 8 weeks control phase, 8 weeks intervention with 4 weeks of follow-up were assessed at four time points (familiarization (week 0)/pre-intervention (week 8)/post-intervention (week 16)/follow-up (week 20)) using a 3D ultrasound imaging technique and dynamometry [5]. Ankle joint function, isometric muscle strength, and muscle-tendon properties of the gastrocnemius medialis (GM) muscle-tendon unit (MTU) (e.g., muscle belly and tendon lengths, fascicle lengths, muscle thickness and volume, tissue length changes) were assessed at each time point. We used repeated measures ANOVA for statistical analysis.

## Results

Within the intervention group, significant increases were observed in normalized MTU length at 0 Nm ( $p = 0.032$ ,  $\eta^2 = 0.446$ ) and 4 Nm ( $p < 0.001$ ,  $\eta^2 = 0.611$ ), as well as in tendon length at both 0 Nm ( $p = 0.041$ ,  $\eta^2 = 0.319$ ) and 4 Nm ( $p = 0.014$ ,  $\eta^2 = 0.389$ ). However, no significant group  $\times$  time interaction effects were found for any parameter.

## Discussion

Although time-dependent improvements in selected morphological parameters (e.g. MTU length) were observed in both groups, there was no statistically significant interaction effect indicating that the addition of daytime activity would result in superior outcomes compared to continuous immobilization only. On the other hand, these findings suggest that prolonged immobilization may not be essential to achieve structural adaptations in spastic muscle. Allowing periods of mobility and calf muscle strengthening could reduce disuse-related effects and support more individualized and potentially more tolerable orthotic treatment approaches [6]. Further research with larger sample sizes is warranted to evaluate long-term functional implications and to optimize orthotic strategies in children with cerebral palsy.

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### The effects of a prosthetic walking practice using virtual reality on gait kinematics: A pilot study

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### Introduction

Prosthetic limbs play a crucial role in preserving the mobility and quality of life among individuals with lower limb amputations [1]. Recent advancements suggest that virtual reality (VR)-based training using simulated transfemoral prostheses may outperform conventional rehabilitation in promoting motor learning and symmetry [2]. However, the biomechanical effects of VR-assisted prosthetic gait training remain unexplored. This pilot study employed three-dimensional (3D) motion analysis to examine the impact of VR-based simulated prosthetic walking on gait kinematics and kinetics.

### Research Question

How do VR-assisted simulated prosthetic walking influence joint angles and ground reaction forces during gait?

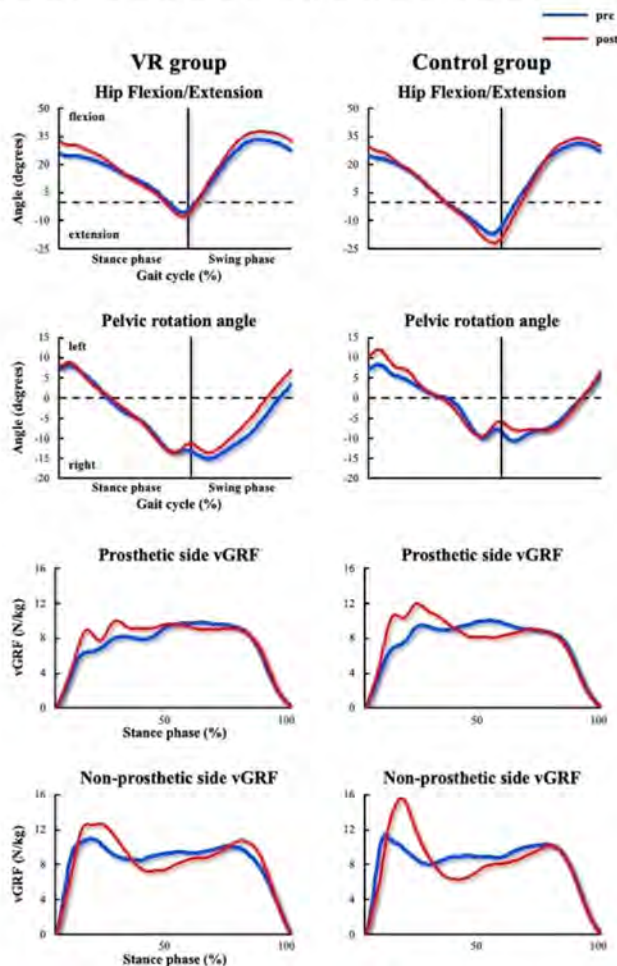
### Methods

Six healthy adults (four men and two women) with no experience of using simulated prostheses were recruited. A transfemoral prosthesis featuring a 3R80 knee joint (Ottobock, Japan) was used on the right leg to simulate amputation. Participants were randomly assigned to either the VR group (n = 3) or the control group (n = 3), each comprising two men and one woman. Both groups performed five minutes of in-place stepping training twice with parallel bars. The VR group viewed a first-person walking video using a head-mounted display, whereas the control group practiced the video without any visual input. Gait performance was evaluated using the 10-meter walk test, administered immediately before and after the training. Motion data were captured using a 3D motion analysis system (Vicon, UK) equipped with 10 infrared cameras. One gait cycle per participant was analyzed to calculate the stride length (normalized to the lower limb length), prosthetic-side hip joint angle, pelvic rotation, and vertical ground reaction force (vGRF) on both sides. Due to the small sample size, the outcomes were descriptively compared using graphical analysis.

### Results

After training, both groups showed an increased walking speed. The VR group demonstrated a more pronounced improvement in stride length. Additionally, only the VR group exhibited increased prosthetic-side hip flexion from late swing to early stance, along with greater leftward pelvic rotation during the swing phase. Early stance vGRF increased bilaterally in both groups, whereas only the VR group showed a further increase in late stance vGRF on the non-prosthetic side (Results).

|         | Walking speed (km/h) |      | Stride length |      |
|---------|----------------------|------|---------------|------|
|         | pre                  | post | pre           | post |
| VR      | 3.28                 | 4.32 | 1.34          | 1.63 |
| Control | 3.64                 | 4.86 | 1.61          | 1.82 |



## Discussion

The VR group exhibited a higher peak vGRF on the non-prosthetic limb during late stance compared to the control group, indicating enhanced push-off. This improved forward and upward propulsion may have facilitated increased hip flexion from the late swing to the early stance, as well as greater leftward pelvic rotation during the prosthetic-side swing. These kinematic and kinetic alterations can be attributed to the vection effect induced by VR visual stimuli, consistent with findings in previous research [2]. These findings suggest that VR-assisted prosthetic walking may enhance push-off on the non-prosthetic side, contributing to a more effective swing phase on the prosthetic side.

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## Gait Rehabilitation Using Intensive Robotic Exoskeleton Training in Individuals with Incomplete Spinal Cord Injury (ASIA D): A Pilot Study

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## Introduction

Spinal cord injury (SCI) significantly impacts mobility, leading to gait impairments, muscle weakness, and reduced functional independence. Individuals with incomplete SCI classified as ASIA D retain some motor function below the level of injury, making them suitable candidates for rehabilitation aimed at restoring walking ability. Traditional gait rehabilitation methods often involve manual-assisted therapy, which can be labor-intensive and limited in terms of precision and intensity.

Robotic exoskeletons have emerged as a promising intervention in neurorehabilitation, offering precise, repetitive, and intensive gait training. These devices provide controlled movement assistance, promoting neural plasticity and facilitating motor recovery. Cyberdyne's Hybrid Assistive Limb (HAL) is one such robotic exoskeleton that utilizes bioelectrical signals to enhance voluntary movement. However, there is limited evidence regarding the efficacy of HAL in improving gait parameters specifically in individuals with ASIA D SCI. This study aims to evaluate the effectiveness of bilateral robotic exoskeleton training in improving spatiotemporal gait parameters in this population.

## Research Question

Does gait training with Hybrid Assistive Limb (HAL) robotic exoskeleton provide improved gait parameters in spinal cord injury population?

## Methods

A total of 10 individuals diagnosed with ASIA D SCI, who had the ability to walk independently with assistive devices and were willing to participate, took part in this study. The intervention consisted of 16 sessions of bilateral robotic exoskeleton training using the Cyberdyne HAL lower limb system. Each session was conducted three times per week, with a duration of one hour per session.

Gait assessment was performed pre- and post-intervention using the Vicon motion analysis system, equipped with 8 infrared cameras and 2 real-life cameras. Expert therapists conducted the assessments, focusing on spatiotemporal gait parameters, including stance percentage, swing percentage, walking speed, step width, step length,

stride length, cadence, step time, and stride time. Data were analyzed using the Wilcoxon Signed-Rank Test in SPSS to determine statistical significance.

## Results

Significant improvements were observed in stance percentage ( $p = 0.022$ ), swing percentage ( $p = 0.022$ ), walking speed ( $p = 0.028$ ), and stride time ( $p = 0.005$ ). Other parameters showed trends of improvement but did not reach statistical significance. These findings suggest that bilateral robotic exoskeleton training has a positive impact on selected gait parameters in individuals with ASIA D SCI.

## Discussion

Bilateral robotic exoskeleton training using the Cyberdyne HAL system demonstrated effectiveness in enhancing selected gait parameters in individuals with ASIA D SCI. However, further research with a control group is required for comparison to determine the true efficacy of this intervention. Additionally, more comprehensive analyses, including kinetic and electromyographic (EMG) data, are necessary to better understand the underlying mechanisms of robotic-assisted gait training and its long-term rehabilitation benefits.

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## Sex and Side Differences in Muscle Stiffness of Lower Limb Muscles in Healthy Adults

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## Introduction

Increased stiffness and decreased elasticity in skeletal muscles in the lower limbs are commonly observed following sports-related injuries and may contribute to a heightened risk of re-injury [1]. Therefore, accurate assessment of muscle stiffness in healthy individuals is critical for developing rehabilitation strategies. However,

several factors can influence muscle stiffness, including sex and limb laterality, which must be considered in clinical assessment. Despite this, healthy individuals' reference data on muscle stiffness remains limited [2].

## Research Question

- 1) Do muscle stiffness parameters differ between men and women in the lower limb muscles of healthy adults?
- 2) Are there left-right differences in muscle stiffness in the lower limb muscles of healthy adults?

## Methods

Seventeen healthy adults (12 men, 24 limbs; mean age:  $23.5 \pm 2.1$  years; 5 women, 10 limbs; mean age:  $24.4 \pm 1.4$  years) were recruited. Muscle stiffness parameters of four lower limb muscles—rectus femoris (RF), biceps femoris (BF), medial gastrocnemius (MG), and tibialis anterior (TA)—were quantitatively assessed using the MyotonPro device (Myoton AS, Tallinn, Estonia). The device delivers a brief (15 ms), low-force (0.4 N) mechanical impulse to the skin surface, producing damped oscillations in the underlying muscle. A built-in accelerometer captures these oscillations to calculate five parameters: tone (Hz), stiffness (N/m), elasticity (arb), relaxation time (ms), and creep (arb). Sex-related differences were analyzed using unpaired t-tests or Mann-Whitney U tests. Left-right differences were examined using paired t-tests or Wilcoxon signed-rank tests.

## Results

Men showed significantly higher tone and stiffness and significantly lower relaxation and creep in the RF, BF, and MG compared to women. However, no significant sex-related differences were observed in the parameters of the TA. In contrast, most of the muscles had no significant differences in muscle stiffness parameters between the left and right sides.

## Discussion

The findings indicate that men generally show greater muscle stiffness than women, consistent with previous studies. However, this pattern was not observed across all muscles, as the TA did not show significant sex differences. Furthermore, left-right asymmetries were minimal, suggesting bilateral symmetry in muscle mechanical properties among healthy adults. These findings emphasize the importance of considering sex-related differences in clinical assessments of muscle stiffness using MyotonPro. Contributing factors to higher muscle stiffness in men may include greater muscle strength and mass, lower fat mass, differences in muscle fiber composition, and larger muscle cross-sectional area [3]. Future research is needed to better understand muscle stiffness while combining it with ultrasound assessment to inform rehabilitation strategies after sports injuries.

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## Clustering gait and functional measures in adults

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## Introduction

As we age, we walk slower, and older adults often employ a more cautious gait (1). It remains unclear whether this phenomenon is consistent at an individual, rather than population, level. Studies tend to investigate biomechanical changes with age (1,2) and compare young to older adults, yet walking speed declines with age (3) and within each decade (4). Therefore, having one group of older adults as a comparator may not be appropriate and could hide decade-specific related changes in function.

## Research Question

To use a clustering approach on spatiotemporal and clinical function data to determine whether patterns of function (strength, movement, quality of life) can be determined based on individuals' gait data.

## Methods

698 participants (18-92 years) were included from 1000 Norms Project (5); a cross-sectional, observational study, collected gait, clinical function, and self-reported health data. Where appropriate all data were normalised to body dimensions (height and mass). Velocity was normalised to Froude number. All statistical analysis was carried out using JASP (v 0.17.3). An unsupervised K-means nearest neighbour clustering analysis optimised to BIC was performed resulting in 17 clusters. This was not a practical nor feasibly relevant. Therefore, a principal component analysis (PCA) approach was used, on spatiotemporal and functional data, to determine the number of components to use to inform the number of clusters (i.e. a supervised method).

## Results

The PCA yielded 3 oblique components (77% of variance; eigen values >1) and these informed the number of clusters for the K-means clustering. Three participants were missing a functional measure and were excluded from further analysis. The 695

participants were partitioned into three-clusters of n=277 for cluster 1, n=208 for cluster 2, and n=210 for cluster 3. The AIC and BIC were 3853.9 and 3962.9 respectively with an R<sup>2</sup> of 0.314. There was no significant difference between clusters for Froude walking speed (p=.406), yet there were significant differences (p<.001) between clusters for all functional/strength measures.

## Discussion

Our clusters defined 3 groups, "higher functioning", "age average" and "cautious gait", whose spatiotemporal, clinical function, strength and quality of life measures vastly differed, independent of walking speed and including a wide range of ages. Our analysis suggests that age should not be used to separate individuals into groups, but rather greater focus should be in function of movement. Future work should determine whether individuals remain within a group throughout the life span or whether interventions can modify functional groups.

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## Machine Learning Model for Discriminating Hallux Valgus based on Plantar Pressure Measurements

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## Introduction

Hallux Valgus (HV) is a foot deformity where the big toe bends outward, causing pain and functional disability in severe cases, impacting quality-of-life. Early diagnosis is essential to prevent complications [1]. Traditional diagnostic methods (visual inspection, radiography, physical examination) are often insufficient for detailed evaluations [2] or costly. This underscores the need for objective reliable diagnostic tools. Deep learning (DL) and machine learning (ML) offer promising alternatives by supporting decision-making.

## Research Question

ML/DL application to HV detection remains limited, primarily focusing on radiographs and intrinsic factors (e.g., age, arc height) [3-5]. This study compares ML/DL models to distinguish between individuals with and without HV, using plantar pressure measurements as a non-invasive alternative to radiographs.

## Methods

Plantar pressure data from 50 HV pre-surgery patients and 55 healthy individuals were collected using the RSscan Footscan System at 200 Hz [6]. This dataset (CADWALK) contains dynamic measurements from both feet, totalling 4,106 foot images. Statistical features (mean, standard deviation, 25-50-75th percentiles, maximum values) were extracted at specific points (10-30-50-70-90%) across a gait stride. Key gait parameters (walking speed, stride length, time) were also included. Several ML models were tested: Support Vector Machine (SVM), Random Forest (RF), Logistic Regression (LR), k-Nearest Neighbors (kNN), and Gradient Boosting (GB). A Convolutional Neural Network (CNN) – Table 1 – was applied to raw images from the 50% percentile slice. Data from same subject were assigned into training, validation, or test sets without overlap. Features were standardized, and images scaled, padded, and augmented (vertical flipping). Models were fine-tuned using grid search and GroupKFold cross-validation. Evaluation metrics included accuracy, precision, recall, and F1-score. ML models were trained using data from individual trials; trial-level predictions were aggregated via majority voting to form an overall prediction at subject-level.

## Results

GB outperformed other ML models (F1-score: 0.7881), with GB and RF demonstrating the highest accuracy (87.66% and 87.43%, respectively) - Table 2. Subject-level analysis produced similar results (accuracy: 86.67% - Table 3). CNN achieved 89.41% accuracy on single-image predictions (confusion matrix in Figure).

Picture/Tables 4

| Layer (type)                   | Output Shape        | Num of Params |
|--------------------------------|---------------------|---------------|
| sequential_2 (Sequential)      | (None, 47, 32, 1)   | 0             |
| conv2d_3 (Conv2D)              | (None, 45, 30, 64)  | 640           |
| max_pooling2d_3 (MaxPooling2D) | (None, 22, 15, 64)  | 0             |
| conv2d_4 (Conv2D)              | (None, 20, 13, 128) | 73,696        |
| max_pooling2d_4 (MaxPooling2D) | (None, 10, 6, 128)  | 0             |
| conv2d_5 (Conv2D)              | (None, 8, 4, 256)   | 295,168       |
| max_pooling2d_5 (MaxPooling2D) | (None, 4, 2, 256)   | 0             |
| flatten_1 (Flatten)            | (None, 2688)        | 0             |
| dense_2 (Dense)                | (None, 128)         | 262,272       |
| d Dropout                      | (None, 128)         | 0             |
| dense_3 (Dense)                | (None, 1)           | 128           |

Table 1: Summary of the CNN model architecture (single image).

| Model | Accuracy | Precision | Recall | F1 Score |
|-------|----------|-----------|--------|----------|
| SVM   | 0.8381   | 0.7838    | 0.7632 | 0.7743   |
| RF    | 0.8476   | 0.8919    | 0.7333 | 0.8049   |
| LR    | 0.8090   | 0.6488    | 0.7500 | 0.6957   |
| kNN   | 0.8090   | 0.6257    | 0.7353 | 0.7042   |
| GB    | 0.8667   | 0.9459    | 0.7447 | 0.8333   |
| NN    | 0.8429   | 0.8243    | 0.7531 | 0.7871   |

Table 2: Performance metrics for different models (subject level).

| Model | Accuracy | Precision | Recall | F1 Score | Accuracy Variance |
|-------|----------|-----------|--------|----------|-------------------|
| SVM   | 0.8624   | 0.7532    | 0.7323 | 0.7426   | 0.0026            |
| RF    | 0.8743   | 0.8484    | 0.7228 | 0.7886   | 0.0024            |
| LR    | 0.8463   | 0.6488    | 0.7366 | 0.6889   | 0.0022            |
| kNN   | 0.8217   | 0.6100    | 0.7104 | 0.6564   | 0.0014            |
| GB    | 0.8706   | 0.8715    | 0.7103 | 0.7801   | 0.0014            |
| NN    | 0.8639   | 0.7763    | 0.7343 | 0.7547   | 0.0023            |

Table 2: Performance metrics for different models (trial level).



Figure 1: Confusion matrix across all trials for CNN (single slice).

## Discussion

Combining plantar pressure images and gait data proved effective in predicting HV. CNNs performed best, demonstrating the potential of using raw image data for accurate HV identification. This supports the feasibility of using AI for non-invasive HV diagnosis. Though previous studies [7-8] analyzed this dataset, no ML/DL models were used for HV detection. In [9], the same dataset achieved a classification rate >90%. However, we achieved our results by using data from just five images instead of using all the pressure maps from the

full gait stride. More impressively, we reached 89.41% accuracy using just a single image, which is a significant advantage over analyzing the entire pressure map of a full stride as done in [9].

## Funding

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## Estimation of Ground Reaction Forces in Running via Deep Learning Models: A Comparative Analysis

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## Introduction

Ground Reaction Forces (GRFs) are fundamental to the analysis of running biomechanics and play a critical role in performance assessment, injury prevention, and rehabilitation [1]. While GRFs

are traditionally measured using expensive and non-portable systems such as instrumented treadmills, there is growing interest in more accessible, scalable alternatives [2]. Inertial Measurement Units (IMUs), which record acceleration, angular velocity, and orientation, have demonstrated potential in this context [3-5]. Recent research has applied deep learning techniques to estimate GRFs from IMU-derived kinematic data [4-6]; however, a standardized benchmark for comparing different models using a publicly available dataset remains absent, limiting reproducibility and cross-study evaluation.

## Research Question

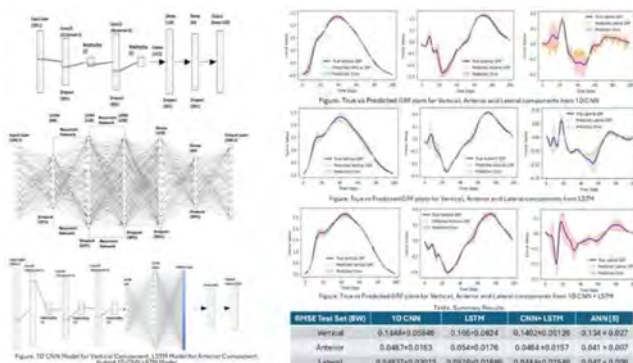
This study aimed to develop and compare multiple deep learning models for predicting vertical, anterior, and lateral GRFs using a publicly available dataset. The goal was to provide a reproducible benchmark for GRF estimation from kinematic data, enabling fair comparison across modeling approaches.

## Methods

The analysis utilized a publicly available dataset by Fukuchi et al. [7], comprising kinematic and kinetic running data from 28 recreational runners. Each participant completed treadmill running trials at three fixed speeds (2.5 m/s, 3.5 m/s, 4.5 m/s) for 30 seconds per speed. GRFs were recorded via an instrumented treadmill, and kinematic data were captured using a 12-camera motion capture system. Acceleration profiles were computed from the trajectories of reflective markers as performed in [8-9]. Data were partitioned into training (60%), validation (20%), and test (20%) sets by subject. Three model architectures were implemented: a one-dimensional convolutional neural network (1D-CNN), a Long Short-Term Memory network (LSTM), and a hybrid 1D-CNN + LSTM model. Each model was separately optimized for each GRF component, tuning hyperparameters such as number of filters, kernel size, number of layers, number of neurons in the fully-connected layers, presence of batch normalization and/or L2-regularization, batch size, epochs, etc. Leave-One-Subject-Out Cross-Validation (LOSO-CV) was employed for evaluation, with Root Mean Square Error (RMSE) used as the loss function. Some examples of the developed architectures are shown in Figure.

## Results

Prediction results for each GRF component were evaluated visually (shown in Figure) and are summarised in Table and compared to the results obtained from analysis on the same dataset [8].



## Discussion

All three models demonstrated performance comparable to previous work using simpler artificial neural networks applied to the same dataset. The findings indicate that complex deep learning models, when properly trained and optimized, can perform competitively for GRF prediction using relatively small biomechanical datasets. However, the study also underscores the limitations imposed by data scarcity, a common challenge in biomechanics. In this context, model simplicity and careful architectural design may yield more practical and generalizable results than overly complex solutions. These results reinforce the importance of balancing model complexity with dataset size in biomechanical machine learning applications.

## Funding

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## Limb lengthening surgery solves anatomical and functional leg length discrepancy during gait

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## Introduction

Anatomical leg length discrepancy (aLLD) is common in paediatric orthopaedics. Deviations from the typical gait pattern often occur in response to aLLD. These may include increased pelvic obliquity, anatomical short side down, with repercussions in the lower spine [1]. The functional leg length discrepancy (fLLD) during gait may be reduced compared to the aLLD by various deviations at the hip, knee, or ankle [2,3]. Apart from conservative treatment with a shoe lift and/or insole, leg lengthening surgery may be an option to remedy aLLD.

## Research Question

Does leg lengthening surgery in adolescents with aLLD improve gait function?

## Methods

15 Adolescents (excluding neuromuscular syndromes or significant deformities) with at least 2 cm aLLD scheduled for an intramedullary lower-limb lengthening with the Fitbone™ nail (femur: 11, tibia: 4) were included prospectively. They received a pre- and post-surgical 3D gait analysis. We defined fLLD as the average difference in height between the left and right anterior superior iliac spine markers over the gait cycle. In addition to fLLD, primary outcome parameters included aLLD and the gait profile score (GPS). Control data were from 15 age-matched peers. Statistical comparisons were conducted using the two-samples (LLD vs controls) or paired (pre vs post) Wilcoxon test.

## Results

Mean patient's age pre-surgery was  $15.2 \pm 1.9$  years and the mean follow-up between gait analyses was  $2.2 \pm 0.7$  years. Preoperatively, individuals had a mean aLLD of 32.3 mm and postoperatively this was 4.7 mm ( $p < 0.001$ ) (Figure 1A). Mean fLLD over the gait cycle

decreased from 19.4 mm preoperatively (vs controls:  $p < 0.001$ ) to 7.6 mm postoperatively ( $p < 0.001$ ) (vs controls:  $p = 0.015$ ) (Figure 1B). The GPS decreased postoperatively ( $6.2^\circ$ ) compared to preoperatively ( $7.2^\circ$ ) but this difference was not significant ( $p = 0.110$ ). However, GPS was different from controls preoperatively ( $p = 0.021$ ) but not anymore postoperatively ( $p = 0.358$ ).

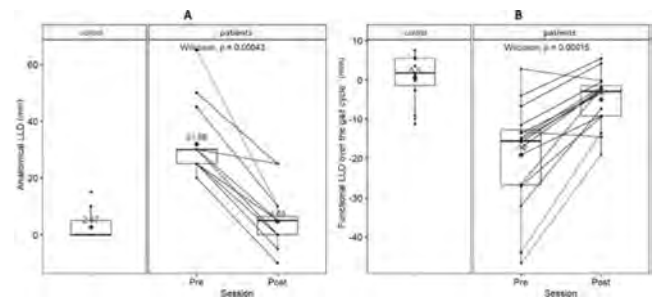


Figure 1. Anatomical (A) and functional (B) leg length discrepancy (short leg) for the controls and patients pre- and postoperatively.

## Discussion

Larger leg length discrepancies can have a significant impact on gait function and increase the risk of related health conditions. The aim of aLLD treatment, either conservative or surgical, is towards functional leg length equalisation. fLLD was more than 1cm smaller than aLLD preoperatively, highlighting that gait compensations tend to minimise the impact of aLLD during gait. Postoperatively, aLLD and fLLD were similar (less than 3mm difference) and both were significantly improved by the leg lengthening surgery. Our results also show that the GPS postoperatively decreased by  $1^\circ$  and approximated the GPS of a normal control group implying a positive gait change towards the norm achieved by surgical leg lengthening. Pelvic obliquity is one of the nine parameters included in the GPS, suggesting that reducing not only aLLD but also fLLD plays a key role in bringing GPS values closer to normal in patients with LLD.

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## Video-based screening for clubfoot recurrence – A first step towards remote assessment

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## Introduction

Children with idiopathic clubfoot often experience high recurrence rates and frequent clinical visits [1]. The Pirani Böhm Sinclair (PBS)-score is one of the 20 outcome measurements included in the published core outcome set for clubfoot research [2], and is proposed as a tool for early detection of recurrence [3]. It evaluates seven dynamic and static items while the child is walking, standing, and sitting. Using video to assess the PBS-score remotely may reduce the need for in-person visits.

## Research Question

Can the PBS-score, when assessed through video recordings, provide reliable inter-rater agreement for both individual items and the total score? It was hypothesized that dynamic items would benefit from reduced playback speed and repeated viewings that video-recordings offer, and further, that items requiring manual input would render lower agreement.

## Methods

Children aged 4–14 years with idiopathic clubfoot were consecutively included from two pediatric orthopedic centers (one in Sweden and one in Pakistan) from January to December 2023. Each child was filmed following a standardized protocol to complete all seven items of the PBS-score, which includes five dichotomous items and two four-point scale items. The total score ranges from 7 (no foot deformity) to 18 (fixed foot deformity). Four orthopedic surgeons from different countries reviewed the videos and scored the feet to assess inter-rater reliability. Agreement, consistency, and reliability were evaluated using intraclass correlation coefficients (ICC) and Kappa statistics.

## Results

55 children with a total of 85 clubfeet were included (64% bilateral). ICC agreement for total PBS-score was 0.82 (95% CI 0.72–0.89) and consistency was 0.86 (95% CI 0.81–0.90). Fleiss' unweighted Kappa for dichotomous items ranged from fair to substantial (Table 1). Cohen's unweighted Kappa ranged from slight

to substantial (Table 1). Cohen's weighted (quadratic) Kappa for items with ordinal data ranged from substantial to almost perfect agreement 0.74–0.82 ('passive dorsiflexion'), and moderate to almost perfect agreement 0.45–0.85 ('subtalar abduction').

Table 1

| Items of the PBS-score     | Fleiss' unweighted Kappa (95% CI) | Unweighted Cohen's Kappa (95% CI) |                      |                      |                      |                      |                      |
|----------------------------|-----------------------------------|-----------------------------------|----------------------|----------------------|----------------------|----------------------|----------------------|
|                            |                                   | Rater 1 vs. 2                     | Rater 1 vs. 3        | Rater 1 vs. 4        | Rater 2 vs. 3        | Rater 2 vs. 4        | Rater 3 vs. 4        |
| Heel View                  | 0.60<br>(0.51, 0.69)              | 0.59<br>(0.43, 0.75)              | 0.77<br>(0.61, 0.93) | 0.85<br>(0.47, 0.93) | 0.54<br>(0.38, 0.70) | 0.54<br>(0.38, 0.70) | 0.58<br>(0.38, 0.78) |
| Standing Supination        | 0.48<br>(0.39, 0.57)              | 0.59<br>(0.47, 0.70)              | 0.52<br>(0.32, 0.72) | 0.47<br>(0.29, 0.68) | 0.30<br>(0.13, 0.47) | 0.35<br>(0.18, 0.52) | 0.80<br>(0.63, 0.99) |
| Walking Supination         | 0.38<br>(0.29, 0.46)              | 0.60<br>(0.43, 0.78)              | 0.28<br>(0.13, 0.41) | 0.34<br>(0.18, 0.50) | 0.32<br>(0.17, 0.47) | 0.30<br>(0.18, 0.43) | 0.97<br>(0.86, 0.97) |
| Heel Rise                  | 0.69<br>(0.60, 0.78)              | 0.77<br>(0.63, 0.91)              | 0.67<br>(0.51, 0.83) | 0.82<br>(0.46, 0.79) | 0.59<br>(0.34, 0.83) | 0.85<br>(0.55, 0.83) | 0.72<br>(0.57, 0.88) |
| Active Ankle Dorsiflexion  | 0.52<br>(0.43, 0.61)              | 0.62<br>(0.40, 0.78)              | 0.68<br>(0.46, 0.91) | 0.86<br>(0.22, 0.98) | 0.61<br>(0.44, 0.78) | 0.61<br>(0.45, 0.77) | 0.41<br>(0.39, 0.43) |
| Passive Ankle Dorsiflexion | 0.45<br>(0.36, 0.50)              | 0.54<br>(0.41, 0.67)              | 0.45<br>(0.21, 0.60) | 0.42<br>(0.28, 0.56) | 0.52<br>(0.30, 0.80) | 0.36<br>(0.23, 0.49) | 0.38<br>(0.24, 0.53) |
| Subtalar Abduction         | 0.41<br>(0.35, 0.47)              | 0.18<br>(0.08, 0.30)              | 0.23<br>(0.05, 0.40) | 0.35<br>(0.17, 0.58) | 0.51<br>(0.47, 0.79) | 0.41<br>(0.27, 0.56) | 0.58<br>(0.40, 0.71) |

## Discussion

This study found near-perfect ICC for the total PBS-score, supporting its reliability. However, the fair inter-rater agreement for the clinically important item 'swing phase supination' suggests variability in interpretation, highlighting the need for clearer scoring guidelines – especially given its role in decisions about tendon transfer surgery. The high weighted Kappa values for the four-point scale items indicate only minor disagreements, suggesting the scale steps for 'passive dorsiflexion' and 'subtalar abduction' may need revision for video-based assessments. In conclusion, video-based assessments of the PBS-score could, following further development of the scoring instructions, reduce the need for in-person consultations and facilitate assessments of patients in remote locations or countries with limited access to pediatric orthopedic surgeons.

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## Evaluation of dataset robustness for video-based automated General Movement Assessment: A protocol for video inclusion via visual clustering

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## Introduction

General Movement Assessment (GMA) is a non-invasive method for early detection of neurodevelopmental disorders, based on visual evaluation of infants' spontaneous motions [1]. GMA widespread adoption is limited by the need for licensed evaluators, subjectivity, and time constraints [2]. Several methods have been proposed for automated, video-based GMA, exploiting anatomical landmark tracking and different evaluation metrics [3].

Anatomical landmark tracking with convolutional neural networks (OpenPose, AlphaPose, DeepLabCut) requires a well-structured training-set to ensure accurate trajectory extraction, both on labeled and unseen videos. To achieve reliable performances, the dataset must be large and diverse, accounting for variability in infant poses, clothing, and lighting.

This work proposes a protocol for assessing, without network processing, whether a newly acquired video requires manual annotation or if the existing dataset is sufficient for accurate trajectory extraction. By avoiding full trajectory reconstruction, this method aims to optimize the dataset update process and labeling efforts.

## Research Question

How can we assess the robustness of a training-set for accurate kinematic reconstruction?

## Methods

Frames extracted from GM videos (GoPro Hero 9, 240fps, 1920x1080p) of 116 very preterm infants at 40 weeks and 3 months of corrected age were manually labelled and used to train a DeepLabCut network (14-point model, 15-50 training frames per video, k-means).

To characterize the training-set, extracted frames were pre-processed with white balance correction, blue channel removal, and color normalization. Infant segmentation was performed using DeepLabV3, pre-trained on COCO dataset, and refined using GrabCut and morphological operations to eliminate residual background artifacts. Histogram of Oriented Gradient (HOG) features were extracted from the masks and, to maintain consistency with DeepLabCut pipeline, clustered into 32 groups using k-means.

The same procedure was applied to 31 new videos. HOG frame features were compared to the existing clusters using a distance-based threshold: if over 70% of a video's frames fell within cluster boundaries, the video was considered included.

To validate this protocol, each new video was also processed through the trained DeepLabCut network. Videos with more than 30% of frames with reference points below 95% confidence level were considered to require manual annotation. Agreement between the two methods was evaluated.

## Results

Trajectories extracted from training-set videos had a mean of 29.55% missing frames (median = 27.92%, IQR = 18.95%). Of 31 new videos, the clustering protocol identified 13 requiring manual annotation, showing moderate agreement (accuracy = 38.7%) with DeepLabCut analysis, where 17 videos needed manual labels. The protocol showed a precision of 38.1% and recall of 57.1% in detecting videos needing inclusion.

## Discussion

Moderate agreement with DeepLabCut analysis highlights that pose-based clustering alone is insufficient, as lighting and color variations significantly affect performance. It is necessary to integrate additional visual features and define objective thresholds to enable more reliable decisions during dataset expansion and reduce manual annotation efforts.

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## Detection of antalgic gait simulations using an AI model in forensic medicine: A lie detector for walking

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## Introduction

After traffic accidents with injuries, the compensation to be paid after the incident is brought to the legal platform is regulated according to the disability rates determined by the Forensic Medicine Department. One of the parameters taken into consideration when determining this disability rate is the presence of "antalgic gait". In addition to increasing the disability rate, antalgic gait also increases the amount of compensation to be received. Unfortunately, some people pretend to simulate antalgic gait during the disability assessment in order to abuse this situation. Although the Forensic Medicine Department (FMD) has mechanisms for determining disability rates, there is a need for systems supporting these mechanisms for individuals simulating pathological gait to obtain higher rates.

## Research Question

This study aimed to develop mechanisms for detecting simulated gaits using motion analysis and artificial intelligence (AI).

## Methods

This study was carried out under the TUBITAK 1001 project (grant no. 221S678). The study was conducted with three groups: forensic medicine patients with unilateral lower extremity fractures (F), orthopedic patients with unilateral lower extremity fractures (O), and healthy individuals (H) with similar demographic characteristics. Demographic information of all participants, as well as fracture-related data for the F and O groups, were recorded. Pain and fatigue levels were assessed before, during, and after walking. During ten walking trials, spatiotemporal gait parameters were recorded using a GAITRite electronic walkway, while whole-body angular kinematics were simultaneously captured using a three-dimensional motion analysis system with eight cameras. The collected data were processed to create comparable datasets. Based on the gait-related data, a decision support system using artificial intelligence was developed through deep learning models. The data of O and H groups were used to train the artificial intelligence model. Then, the decisions of the developed artificial intelligence model were compared with FMD and PTR opinions, and the consistency rates were determined.

## Results

Demographic characteristics of the groups were similar. For the cases in group A, the decisions made by FMD and PTR were found to be 75% compatible. It was seen that the decisions of the developed artificial intelligence model were 60% compatible with FMD and 75% compatible with PTR. For five patients in which FMD and PTR expressed different opinions, the decisions of the artificial intelligence model were in agreement with PTR in four patients and with FMD in one patient.

## Discussion

While the results derived from the model demonstrate a higher level of agreement with the PTR findings, it is believed that further refinement is necessary. Nevertheless, in its present form, it is considered feasible to utilize the model as a supplementary evaluator, to be consulted in cases where discrepancies arise between the opinions of the FMD and PTR.

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## Analyzing Gait Stability in Parkinson's Disease: Event-Specific Differences in Margin of Stability

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## Introduction

Gait instability is a key motor symptom of Parkinson's disease (PD) and one of the main contributors to falls, injury, and loss of independence [1], [2]. Therefore, early identification of dynamic instability is critical for fall prevention and targeted intervention. The margin of stability (MoS), calculated as the distance between the extrapolated center of mass (xCOM) and the base of support (BoS), reflects the capacity to maintain balance during walking. While typically assessed at heel strike (HS), MoS may vary across the gait cycle, and its discriminatory power at different time points remains unclear.

## Research Question

Does anteroposterior MoS (MoSAP) differ across key gait events (HS, toe-off (TO), and mid-swing (MS)), and can it detect differences in dynamic stability between people with PD and healthy controls (HC)?

## Methods

Sixty-nine participants: 35 HC (16M/19F, 67.1 ± 7.0 yrs), and 34 patients with PD (21M/13F, 67.7 ± 8.7 yrs) walked four times along a 4-meter path, turned, and returned. All provided informed consent (ethics approval no. ENM095/25260220251782025). Full-body kinematics were recorded using 17 inertial measurement units. MoSAP was computed at HS, TO, and MS following Hof's approach (MoSAP = BoSAP - xCOMAP) [3]. Repeated measures ANOVA assessed within-group differences across events and linear mixed models were used to compare groups at each event.

## Results

MoSAP differed significantly across the three gait events in both groups (HC:  $p < 0.0001$ , PD:  $p < 0.0001$ ). Post-hoc tests revealed that HS had the highest MoSAP, followed by TO and MS. Significant differences between PD and HC groups were observed at HS ( $p = 0.0271$ ) and MS ( $p = 0.0277$ ), but not at TO ( $p = 0.2003$ ). Specifically, the PD group showed less negative MoSAP values, suggesting that their xCOM was positioned closer to the BoS boundary than controls. These results are reflected in Figure 1.

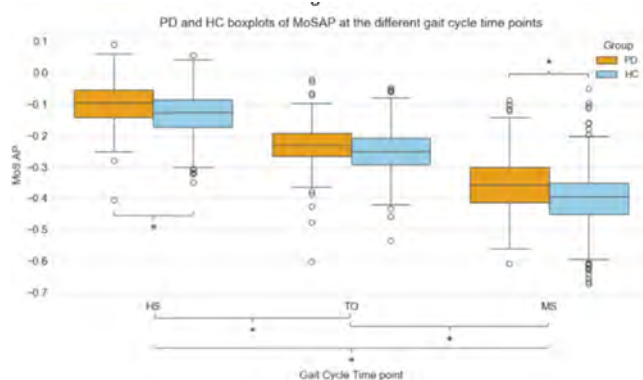


Figure 1. PD and HC boxplots of MoSAP at the different gait cycle time points. Significant differences are highlighted with asterisks.

## Discussion

MoSAP in people with PD was less negative than in healthy controls, indicating their xCOM is positioned closer to the edge of the BoS and reflecting reduced dynamic stability and a narrower margin for error during gait. This may be explained by typical Parkinsonian gait features [4],[5]: slower walking speed reduces xCOM velocity, shorter step length limits forward progression, and axial rigidity restricts trunk movement. These factors likely contribute to a more cautious but less stable gait. Differences were most evident at heel strike and mid-swing, while toe-off showed no group differences. This emphasizes the need for event-specific MoSAP analysis to better detect subtle gait impairments in PD.

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## Test-Retest Reliability of Inertial Measurement Unit-Derived Trunk and Arm Swing Metrics Across Different Walking Conditions

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## Introduction

Wearable inertial measurement units (IMUs) have transformed gait analysis by enabling continuous, objective, and non-invasive monitoring of human movement. The APDM Mobility Lab system, with its validated multi-sensor configuration, facilitates simultaneous assessment of limb, trunk, and gait dynamics. While IMU-derived metrics are widely adopted in research and clinical settings, the reliability of trunk and arm swing measurements under varying walking conditions remains insufficiently studied.

## Research Question

This study aimed to assess the test-retest reliability of trunk kinematics, arm swing velocity, arm swing asymmetry, arm range of motion (ROM), and spatiotemporal gait parameters across four walking tasks in healthy adults, using a standardized repeated-measures protocol.

## Methods

Twenty-four healthy adults (18–65 years) performed walking trials under four conditions: (1) comfortable speed, (2) fastest speed, (3) walking with constrained arm swing, and (4) dual-task walking. Six IMUs (Opal, APDM Mobility Lab, USA) were positioned on the wrists, feet, lumbar spine, and trunk. Parameters included arm swing velocity (°/s), arm ROM (°), trunk ROM in three planes (°), and spatiotemporal measures such as cadence, stride length, speed (m/s), and stride length asymmetry. Each walking condition was repeated after seven days. Test-retest reliability was evaluated using Intraclass Correlation Coefficients (ICC) and Bland-Altman plots.

## Results

Reliability varied by parameter and walking condition. Right arm ROM and velocity demonstrated excellent reliability during comfortable and fast walking (ICC > 0.75, 95%CI). Transverse trunk ROM showed strong reproducibility in all tasks, particularly during high-

speed and dual-task walking (ICC up to 0.89, 95%CI). Gait speed, cadence, and stride length exhibited good to excellent ICC values (e. g., cadence ICC = 0.91, 95%CI; speed ICC = 0.87, 95%CI), with the highest reliability during dual-task trials. Bland-Altman analysis confirmed consistency across sessions without systematic bias. In contrast, parameters reflecting asymmetry—arm swing asymmetry and stride length asymmetry—showed poor reliability (ICC < 0.40, 95%CI).

## Discussion

IMU-based assessment using wearable sensors provides reliable data on gait, trunk, and arm movements under varied walking tasks. Metrics such as gait speed, cadence, transverse trunk ROM, and right arm swing appear particularly stable and suitable for longitudinal monitoring. However, asymmetry measures demonstrated low reproducibility and should be interpreted with caution. These findings support the application of IMUs in clinical gait evaluation and research on motor function.

**Table 14 : Reliability of Gait, Trunk and Arm Swing Parameters for Different Walking Conditions According to ICC Values**

|                            |                               | Comfortable Speed Walking | High Speed Walking | Walking with Bound Arm | Walking with Dual Duty |
|----------------------------|-------------------------------|---------------------------|--------------------|------------------------|------------------------|
| Arm Oscillation Parameters | Left Arm Swing Speed (°/s)    | Good                      | Weak               | Good                   | Good                   |
|                            | Right Arm Swing Speed (°/s)   | Good                      | Good               | Good                   | Good                   |
|                            | Arm Swing Speed Asymmetry     | Weak                      | Good               | Good                   | Weak                   |
|                            | Left Arm Movement Deficit     | Weak                      | Good               | Good                   | Good                   |
|                            | Right Arm Range of Motion (°) | Good                      | Good               | Good                   | Good                   |
|                            | Arm Movement Aperture         | Weak                      | Good               | Good                   | Weak                   |
|                            | Asymmetry                     | Good                      | Good               | Good                   | Good                   |
|                            | Cadence                       | Good                      | Good               | Good                   | Good                   |
|                            | Speed(m/s)                    | Good                      | Good               | Good                   | Good                   |
|                            | Step Length(°)                | Good                      | Good               | Good                   | Good                   |
| Walking Parameters         | Step Length                   | Weak                      | Weak               | Weak                   | Weak                   |
|                            | Asymmetry                     | Good                      | Good               | Good                   | Good                   |
|                            | Coronal (°)                   | Weak                      | Good               | Good                   | Good                   |
| Body Parameters            | Sagittal (°)                  | Good                      | Weak               | Weak                   | Weak                   |
|                            | Transvers (°)                 | Good                      | Good               | Good                   | Good                   |

Perfect Good Weak

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## The Contribution of the Dominant Arm to Gait Dynamics: A Reliability Study of Arm Swing Using Wearable Sensors

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## Introduction

Arm swing plays a crucial role in enhancing gait stability and energy efficiency. However, the specific contribution of the dominant arm during walking has received limited attention. It is known that the dominant arm typically exhibits greater motor control and a wider range of motion, which may influence gait asymmetry and dynamic balance, especially under demanding conditions. This study aimed to examine the influence of dominant arm swing on gait performance and to evaluate the reliability of swing-related parameters using wearable sensors across different walking conditions.

## Research Question

To investigate the effect of dominant arm swing on gait parameters in healthy adults and assess the test-retest reliability of these measurements.

## Methods

Twenty-four healthy adults participated in this observational reliability study. Each participant completed walking trials under four conditions: (1) comfortable walking speed, (2) maximum safe speed, (3) dual-task walking (with a cognitive subtraction task), and (4) walking with restricted arm motion. The dominant arm was identified based on participant self-report. Arm swing velocity, range of motion, swing asymmetry, cadence, and gait speed were measured using APDM Opal wearable motion sensors. All tests were repeated after 5–7 days to assess test-retest reliability.

## Results

In all walking conditions, the dominant arm demonstrated significantly higher swing velocity and greater range of motion compared to the non-dominant arm ( $p < 0.01$ ). These differences were most pronounced during high-speed and dual-task walking, where dominant arm performance remained more consistent. The asymmetry index increased notably under dual-task conditions, indicating a greater compensatory role of the dominant arm in maintaining dynamic balance. Intraclass correlation coefficients (ICC) for dominant arm swing parameters exceeded 0.90 across conditions, confirming excellent test-retest reliability.

**Table 14 : Reliability of Gait, Trunk and Arm Swing Parameters for Different Walking Conditions According to ICC Values**

|                            |                                 | Comfortable Speed Walking | High Speed Walking | Walking with Bound Arm | Walking with Dual Duty |
|----------------------------|---------------------------------|---------------------------|--------------------|------------------------|------------------------|
| Arm Oscillation Parameters | Left Arm Swing Speed (°/s)      | Good                      | Weak               | /                      | Good                   |
|                            | Right Arm Swing Speed (°/s)     | Good                      | Good               | /                      | Good                   |
|                            | Arm Swing Speed Asymmetry       | Weak                      | Good               | /                      | Weak                   |
|                            | Left Arm Movement Deficit       | Weak                      | Good               | /                      | Good                   |
|                            | Right Arm Range of Motion (°)   | Good                      | Good               | /                      | Good                   |
|                            | Arm Movement Aperture Asymmetry | Weak                      | Good               | /                      | Weak                   |
|                            | Cadence                         | Good                      | Good               | Good                   | Good                   |
|                            | Speed(m/s)                      | Good                      | Good               | Weak                   | Good                   |
|                            | Step Length(°)                  | Good                      | Good               | Good                   | Good                   |
|                            | Step Length Asymmetry           | Weak                      | Weak               | Weak                   | Weak                   |
| Walking Parameters         | Coronal (°)                     | Weak                      | Good               | Good                   | Good                   |
|                            | Sagittal (°)                    | Good                      | Weak               | Weak                   | Weak                   |
|                            | Transvers (°)                   | Good                      | Good               | Good                   | Good                   |
| Body Parameters            |                                 |                           |                    |                        |                        |
|                            |                                 |                           |                    |                        |                        |
|                            |                                 |                           |                    |                        |                        |

■ Perfect    ■ Good    ■ Weak

## Discussion

The dominant arm plays a key role in preserving gait efficiency and balance, particularly in cognitively or mechanically demanding walking conditions. Its enhanced motor performance supports compensatory mechanisms under challenging scenarios, such as dual-tasking or rapid gait. These findings emphasize the importance of including dominant-side assessments in gait analysis and suggest that early asymmetry detection via wearable sensors may offer

clinical insights into emerging gait disturbances. Monitoring dominant arm contributions could be particularly valuable in populations at risk for balance impairments or neurological conditions, helping clinicians tailor interventions more precisely.

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### Modeling tibialis anterior muscle activity of children with cerebral palsy during passive stretches using plantar flexor sensory feedback

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## Introduction

In children with cerebral palsy (CP), muscle responses to stretch are often exaggerated, typically attributed to hyperexcitability of the stretch reflex of the agonistic (lengthened) muscle [1]. However, increased activity of the antagonistic (shortened) muscle has also been observed, potentially due to reduced reciprocal inhibition [2]. In this study, we used a musculoskeletal model to investigate whether antagonistic muscle activity can be explained by sensory information from the agonistic stretched muscle.

## Research Question

Can tibialis anterior activity in response to stretch of the plantarflexors be described based on sensory information from plantarflexors?

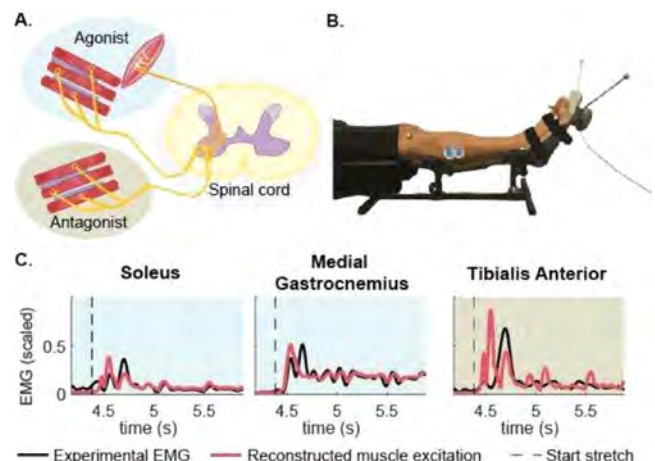
## Methods

Kinematics and muscle electromyography (EMG) of the medial gastrocnemius, soleus and tibialis anterior were collected while manually dorsiflexing the ankle joint of a relaxed child with CP as fast as possible (1B; [3]). We extended a previously proposed model of the muscle response to stretch [4] to capture the response of the antagonistic tibialis anterior muscle. We scaled a musculoskeletal model to the dimensions of the child [5], used the measured joint kinematics to compute muscle-tendon length, and simulated the muscle's force and yank based on a Hill-type muscle model with the measured EMG and muscle-tendon length as inputs. Sensory feedback gains were estimated by minimizing the difference between modeled and measured muscle activation, using force and yank as inputs. Tibialis anterior gains were estimated from soleus sensory information, while plantarflexors gains used their corresponding sensory input. Computed reflex activity and constant background activity, recorded 0.5 s before stretch onset, were summed to reconstruct total muscle activity.

Here, we present preliminary results based on data of one child (Boy, 11 years, diplegic CP, GMFCS 2) as a case study, but the analysis will be extended to 85 children based on existing data.

## Results

The model captured the response of both plantarflexors and tibialis anterior to the stretch of the plantarflexors (Fig. 1C).



## Discussion

Our preliminary results suggest that the activity of the antagonistic tibialis anterior are driven by sensory information from the stretched agonistic soleus. However, the model underestimates the magnitude and timing of the initial stretch response. This may be due to modeling errors. For example, the muscle model does not yet account for short-range stiffness (SRS), i.e., a movement-history-dependent steep increase in muscle force [7]. SRS is especially pronounced when muscles are held isometric, as in our experiment, and when background activity is high, as is often the case in CP. Future directions, therefore, include updating the muscle model to account for SRS. In addition, we will use our model to evaluate whether reciprocal activity during isolated joint motion relates to that observed during walking.

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## Feasibility of foot posture assessment within ankle-foot orthoses using weight-bearing CT scans and plantar pressure insoles

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## Introduction

Foot deformities commonly develop during growth in individuals with cerebral palsy (CP). Early and adequate correction of foot posture may prevent deterioration of the deformity [1], and improve foot function during walking by optimizing its role as a lever arm [2]. Many children with CP wear ankle-foot orthoses (AFOs) to improve gait and correct deformities of the foot. However, because AFOs enclose the foot, foot posture assessment is challenging, especially when walking. Weight-bearing cone-beam Computed Tomography (WBCT) [3] and plantar pressure insoles [4] have the potential to measure static and dynamic foot posture while wearing AFOs. However, metal components of the AFO may distort the CT image, while fitting of insoles in AFOs may be challenging.

## Research Question

Is it feasible to assess foot posture within AFOs using WBCT and plantar pressure insoles?

## Methods

Two cone-beam CT scans (Planmed Verity, Planmed Oy, Helsinki, Finland) were made of an adult cadaveric lower limb, one with a dorsal AFO, and the second with a ventral hinged AFO. Although the weight-bearing CT scanner was used, the scans were made in a non-weight-bearing position. Bones were semi-automatically segmented using Disior Bonelogic (Helsinki, Finland) and visually assessed for potential artefacts. Additionally, a gait analysis with plantar pressure insoles (Pedar-X, Novel GmbH, Munich, Germany) was performed in one adult participant (EU shoe size 40) without foot pathology. The participant walked with a custom-made dorsal AFO on the right foot, and confection shoes. A medial wedge was added to the AFO. The pressure insole was placed inside the AFO. The participant also walked at a similar walking speed with confection shoes only.

## Results

The bone segmentations from the CT scans were not affected by artefacts (Figure 1a). Fitting the pressure insoles into the AFOs proved feasible, and we were able to detect changes in the plantar pressure related to the AFO (Figure 1b-c). Higher pressures were measured under the medial wedge of the AFO, which likely caused unloading of the medial hindfoot. Additionally, under the head of the first and second metatarsal, and under the lateral forefoot and hindfoot differences were shown.

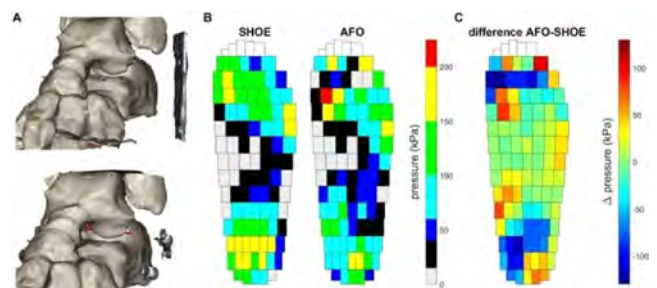


Figure 1: (a) The bone segmentations with metal components; (b) the maximum plantar pressure per sensor during the gait cycle, averaged over 12 steps (most distal sensors removed due to technical errors); (c) difference in plantar pressure per sensor.

## Discussion

The results of this feasibility study show that assessing foot posture within AFOs using WBCT scans and plantar pressure insoles is feasible, which opens up new possibilities to assess foot posture within AFOs. Future studies will apply these techniques in children with CP, with the aim of evaluating the precision and effectiveness of the AFO to align the foot and eventually of improving orthotic treatment for foot deformities in this population.

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### Favourable effect of selective dorsal rhizotomy on energy cost of walking in children with spastic cerebral palsy

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### Introduction

Children with spastic Cerebral Palsy (CP) often experience walking problems, including an increased energy cost of walking (ECw) compared to typically developing peers (1). An increased ECw negatively and persistently impacts daily physical activity (2). Therefore, interventions are aimed at improving walking efficiency. Selective dorsal rhizotomy (SDR), a neurosurgical treatment to reduce spasticity, appears to improve gait patterns (3), which may reduce the ECw. However, its effect on ECw is unclear due to limited studies and varying outcome measures (4, 5).

### Research Question

What is the effect of SDR on gross ECw in children with CP?

### Methods

Thirteen children with spastic CP (mean age:  $8.9 \pm 2.5$  y; Gross Motor Function Classification System levels I (n=1), II (n=11), III (n=1)), who underwent SDR to improve walking function, were included. Gross ECw (total energy used per unit of distance covered) was measured pre-SDR and 2-years post-SDR during six-minutes of walking at a self-selected walking speed (SWS), with simultaneous gas-analysis. Children walked with Ankle-Foot-Orthoses (AFOs) or walking aids if needed. Differences in gross ECw and SWS between pre-SDR and 2-years post-SDR were analysed using a Wilcoxon Signed Rank Test. Changes in gross ECw were compared against the expected natural developmental decline per year in children with CP ( $-0.2$  J/kg/m (6)).

### Results

Gross ECw decreased by 22.8%, from  $10.1 \pm 4.4$  J/kg/m pre-SDR to  $7.8 \pm 2.9$  J/kg/m at 2-years post-SDR ( $p=0.002$ ) (Figure 1). The mean reduction in gross ECw of  $2.3$  J/kg/m exceeded the expected natural developmental decline of  $0.4$  J/kg/m by 475%. SWS did not increase significantly due to large variation ( $p=0.075$ ; Figure 1). At the pre-SDR assessment, five children wore AFOs, one used a walking aid, and one used both. At the two-year post-SDR assessment, five children wore AFOs, and two used walking aids and AFOs consistent with the pre-SDR assessment.

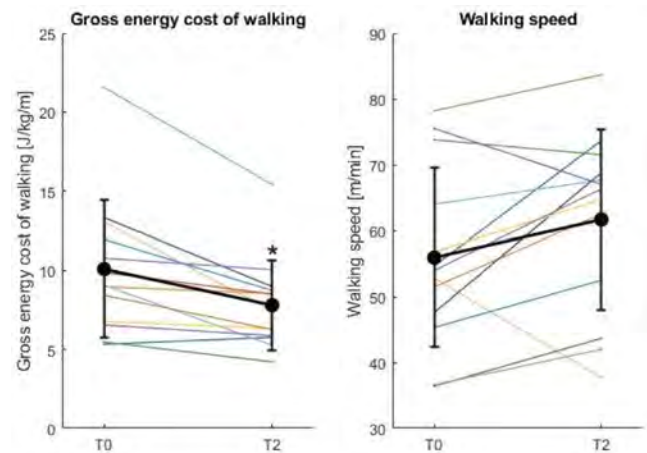


Figure 1. Gross energy cost of walking (left) and walking speed (right) before and 2-years post-SDR (n=13). Coloured lines represent individual values; black line represents the mean with standard deviation. \* significant difference.

### Discussion

Two years after SDR, children with CP showed a pronounced reduction in gross ECw, suggesting that this reflects a beneficial effect of SDR, not solely maturation. This aligns with previous findings, which reported a decrease in the energy expenditure index (an indirect measure of ECw) in favour of the SDR group (5). Our results contrast previous findings, which reported a modest reduction in energy consumption (-12%) without a change in walking speed and no significant difference from a matched control group (4). Possible explanations for these favourable results are the use of different SDR selection criteria and the intensive postoperative rehabilitation program (4). In conclusion, in our cohort of children with CP, SDR seems to have a more pronounced effect on the energy cost of walking compared to natural development.

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## Biomechanical gait changes in relation to change in walking energy cost following orthosis provision in usual and specialized orthotic care

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## Introduction

Lower limb orthoses are frequently prescribed to improve walking in people with neuromuscular disorders (NMD). Guideline-based orthotic provision in expert centers (i.e. specialized orthotic care) has been shown to improve functioning more than usual orthotic care in this population. It also resulted in an approximate 10% reduction in walking energy cost, compared with no change after usual orthotic care [1]. However, group mean changes in biomechanical gait parameters 6 months after orthosis provision did not differ between the specialized and usual care groups, leaving the underlying mechanisms of improvement in walking energy cost for specialized orthotic care unclear.

## Research Question

The aims of this study were to (1) assess changes in biomechanical gait parameters following specialized and usual orthotic care separately for treatment groups and (2) to explore the relationship between changes in biomechanical gait parameters and change in walking energy cost.

## Methods

This post hoc analysis included 50 participants (mean(SD) age: 62(12), 15 women) with complete data from a randomized controlled trial evaluating cost-effectiveness of specialized orthotic care (intervention) versus usual orthotic care (control) in adults with NMD [2]. Biomechanical gait parameters and walking energy cost (in J/kg/m) at comfortable speed were assessed at baseline (with shoes or previous orthosis) and six months after orthotic treatment in both groups. For each group separately, sagittal ankle and knee biomechanics were analyzed using Statistical Parametric Mapping; changes in walking speed and ankle power timing were evaluated with paired t-tests. Correlations between sagittal ankle and knee

biomechanical changes at specific points during the stance phase and walking speed with change in walking energy cost were analyzed using Spearman's correlations.

## Results

Following specialized orthotic care (n=25), significant differences were observed between baseline and follow-up in ankle angle, ankle moment, knee angle, knee moment (p<0.039), walking speed (p<0.001) and ankle power timing (p<0.001) but not in ankle power. Usual orthotic care (n=25) only resulted in a significant change in knee moment (p=0.042), walking speed (p=0.039) and ankle power timing (p<0.001). Change in walking energy cost was significantly correlated with change in walking speed (r=-0.618, p<0.001) and change in timing of peak ankle power (r=0.416, p=0.003). No other biomechanical parameters showed significant correlations with walking energy cost (Figure 1).

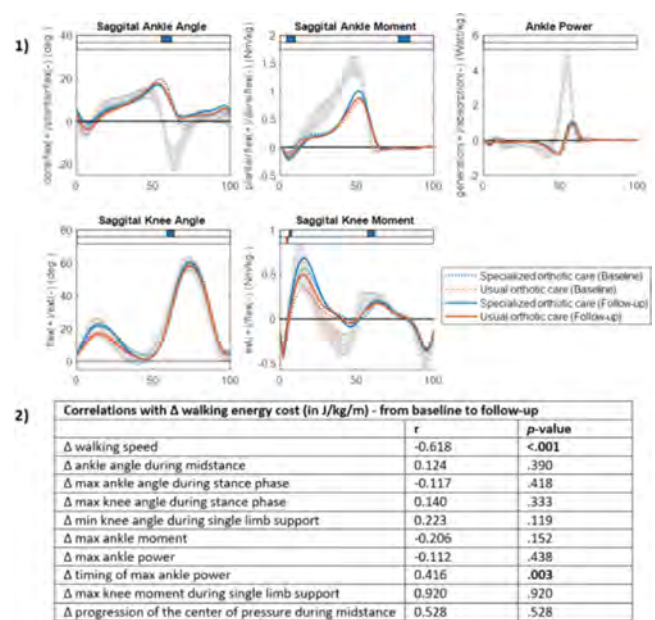


**Figure 1:** (1) Gait biomechanics before and after specialized orthotic care (blue) and usual orthotic care (orange) at baseline (dotted line) and follow-up (solid line), SPM outputs between baseline and follow-up are displayed in the bars above each trajectory (blue = significant following specialized orthotic care, orange = significant following usual orthotic care). (2) Correlations between changes in biomechanical gait parameters and changes in walking energy cost.

## Discussion

Specialized orthotic care led to more biomechanical gait improvements than usual care, although changes were small in both groups. Nonetheless, increased walking speed and earlier timing of peak ankle power correlated with reduced energy cost of walking. As increased speed is beneficial for gait efficiency, and an earlier timing of push-off is also associated with gait efficiency by enhancing propulsion, both factors could underlie the observed reductions in energy cost. Future research should focus on individual biomechanical gait responses to orthotic interventions to better understand mechanisms of effectiveness.

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## Optimizing gait with bidirectional tuning of ankle-foot orthosis stiffness in people with neuromuscular disorders: preliminary results

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## Introduction

Ankle-foot orthoses (AFOs) are commonly prescribed to improve standing and walking in people with neuromuscular disorders (NMD) affecting dorsiflexor and plantarflexor strength [1]. Their effectiveness depends largely on spring type and bending stiffness [2]. Most prefabricated AFOs provide equal bending stiffness in both dorsiflexion and plantarflexion direction, while spring-hinged AFOs allow independent tuning of stiffness in both directions. This may enhance walking outcomes, particularly in loading response during the first ankle rocker. Additionally, precompression of spring units provides basic resistance that could improve standing balance by providing immediate support.

## Research Question

Do custom made spring-hinged AFOs with independently optimized dorsiflexion and plantarflexion stiffness improve walking and standing outcomes in people with NMD compared to prefabricated spring-like AFOs with similar stiffness in both directions?

## Methods

An ongoing pre-post pilot study is being conducted in 10 adults with NMD with at least plantarflexor weakness and an AFO indication. At baseline, 3D gait analyses and 6-minute walk tests were performed using three prefabricated spring-like AFOs (Matrix®, Matrix Max2®, WalkOn Reaction®) of different stiffness levels (0.58, 1.35, 2.88 Nm/degree). Participants were then fitted with a custom-made spring-hinged AFO (NEURO SWING®), with stiffness individually optimized for dorsiflexion and plantarflexion direction separately (possible stiffness levels ranging from 0.7–8.4 Nm/degree) based on a predefined selection algorithm. After 6 weeks of use, assessments were repeated with the optimized spring-hinged AFO. Outcomes included velocity of ankle plantarflexion movement during the first ankle rocker (primary outcome) and ankle range of motion during the first ankle rocker (ROM), maximal knee flexion during loading response, walking speed (in m/s), energy cost of walking (in J/kg/m), and standing balance (root mean square amplitude of CoP displacements (aCOP) in anteroposterior and mediolateral direction).

## Results

Preliminary data from 6 participants (mean age:  $73 \pm 7$  years, 3 women) showed no significant difference between conditions in the primary outcome, ankle plantarflexion velocity ( $p = 0.420$ ). For secondary outcomes, no condition effect was found on ankle range of motion ( $p = 0.830$ ). Maximal knee flexion differed significantly between conditions ( $p = 0.010$ ), though not significantly lower for the spring-hinged AFO compared to the spring-like AFOs (all three post-hoc tests:  $p > 0.05$ ). Walking energy cost and walking speed did not differ significantly between conditions ( $p = 0.131$  and  $p = 0.803$ ). Regarding standing balance, aCOP in anteroposterior direction differed significantly between conditions ( $p < 0.001$ ), with lower amplitudes for the spring-hinged AFO compared to each spring-like AFO (post-hoc tests  $p \leq 0.016$ ), while mediolateral aCOP did not differ between conditions ( $p = 0.084$ ).

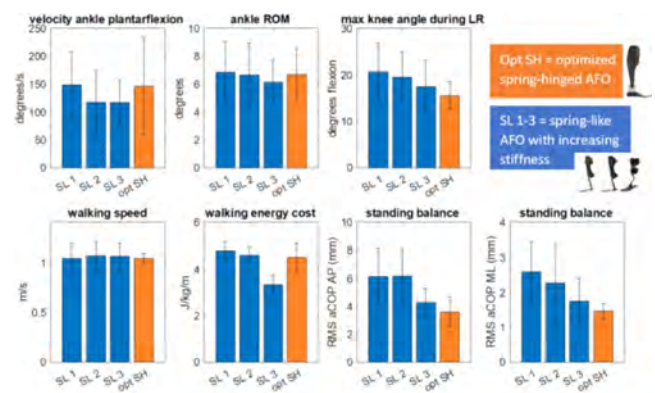


Figure 1: Above: comparison of gait biomechanics in loading response (velocity of ankle plantarflexion movement and range of motion during first ankle rocker and knee flexion) for the optimized spring-hinged AFO (orange) versus the three spring-like AFOs with increasing stiffness (blue); below: comparison on walking speed, walking energy cost and standing balance. Abbreviations: SL = spring-like AFO; opt SH = optimal spring-hinged AFO; ROM = range of motion; LR = loading response; RMS = root mean square; aCOP = amplitude of CoP displacements; AP = anteroposterior; ML = mediolateral.

## Discussion

Preliminary findings from this pilot study indicate that spring-hinged AFO with bidirectional stiffness tuning may potentially reduce knee flexion during loading response compared to prefabricated spring-like AFOs with similar stiffness in both directions in people with NMD. Moreover, reduced anteroposterior COP movement amplitudes while standing with the spring-hinged AFO suggest improved standing balance.

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## Validation of automated fascicle tracking algorithms for functional gait tasks in children

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## Introduction

Sonomyography facilitates dynamic assessment of muscle function by quantifying the length changes and interactions of muscle fascicles, tendons, and the muscle belly during movement. The gastrocnemius medialis (GM), a key contributor to push-off power during gait, is frequently examined using this technique [1, 2]. Manual fascicle tracking may be considered the gold standard for assessing fascicle behavior; however, it is labor-intensive and susceptible to human error [3]. The semi-automated Kanade-Lucas-Tomasi (KLT) and fully automated UltraTimTrack (UTT) algorithms have been proposed as viable alternatives [4, 5, 6]. However, their accuracy and computational efficiency require further evaluation for both regular walking and more challenging gait tasks.

## Research Question

What is the concurrent validity and computational efficiency of the KLT and UTT algorithms compared with manual tracking for GM fascicle tracking across various functional gait tasks in children?

## Methods

For three Typically Developing (TD) children (age:  $11.00 \pm 1.00$ ), dynamic ultrasound data were collected of treadmill walking and running, walking and running while pulling weights attached to a pulley system, and upwards incline walking. Manual tracking was performed by a single experienced assessor, and the KLT and UTT algorithms were applied. Reliability of manual tracking as a gold standard was assessed across five trials with the criteria of  $RMSE \leq 10\%$  and  $r \geq 0.68$ . Computational efficiency was evaluated based on the absolute time required for manual tracking and the two algorithms. Statistical Parametric Mapping (SPM) two-way repeated measures ANOVA with post-hoc comparisons was conducted to analyze fascicle length differences across manual tracking, KLT, UTT and across functional tasks.

## Results

Manual tracking showed good reproducibility ( $RMSE = 4.63$  mm and  $r = 0.78$ ), confirming its reliability as the gold standard. Compared to manual tracking ( $60 \pm 15$  min) and KLT ( $60 \pm 10$  min), UTT ( $35 \pm 5$  s) exhibited superior computational efficiency. Compared to manual tracking, KLT data revealed significant differences for all functional tasks ( $p < 0.05$ ). UTT data revealed significant differences

for walking tasks ( $p < 0.05$ ) and no significant differences for running tasks compared to manual tracking. UTT showed better concurrent validity for pulley-walking and -running and slope-walking tasks than for regular walking (Fig.1).

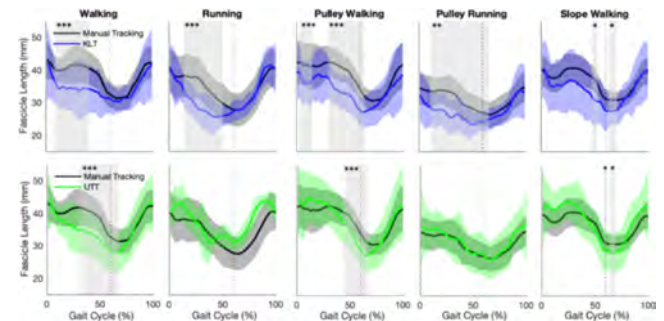


Fig. 1. Average fascicle length across the gait cycle for three children, with standard deviation calculated across five steps per child. Light grey areas indicate significantly different sections of the curves. (\* $p \leq 0.05$ , \*\* $p \leq 0.01$ , \*\*\* $p \leq 0.001$ ).

## Discussion

UTT was the most efficient and had the highest concurrent validity compared to manual tracking, particularly for running tasks. Both algorithms underestimated fascicle lengthening during stance phase of regular walking. Future studies should refine the UTT algorithm to account for aponeurosis curvature, which is especially relevant during low activation (walking tasks) or complex movements [1, 5]. Furthermore, a higher acquisition frequency would help reduce noise in running tasks, likely caused by faster fascicle contractions and greater fascicle length changes than during walking tasks [7].

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## Usability and feasibility of the web-based GAIT.SCRIPT interpretation tool for systematic gait analysis in children with cerebral palsy

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## Introduction

Gait analysis is frequently performed in children with cerebral palsy (CP) to diagnose gait impairments and assist in their treatment. Interpretation of clinical gait data is time-consuming, requires extensive expertise and often lacks transparency. We recently developed the GAIT.SCRIPT method<sup>1</sup>, based on impairment focused interpretation<sup>2</sup>, which automatically links gait features to potential underlying impairments and combines this with supporting clinical data. The improved web-based version of GAIT.SCRIPT is designed to improve standardization and transparency of gait data interpretation and to guide professionals in the interpretation process.

## Research Question

This study aimed to explore the usability and feasibility of the web-based GAIT.SCRIPT interpretation tool.

## Methods

Professionals in pediatric gait analysis, with various work backgrounds from different Dutch rehabilitation clinics, were asked to use the web-based GAIT.SCRIPT tool to interpret clinical gait data from children with CP (intervention). Their expectations, needs, preferences and long-term outlook regarding the tool were assessed using a mixed-method approach with pre- and post-intervention questionnaires, including the System Usability Scale (SUS, score range 0-100), and semi-structured interviews. Two authors (SD and AW) independently analyzed the interview data applying a thematic analysis approach and using MAXQDA software.

## Results

Nineteen participants were included (84% female; 9 pediatric rehabilitation physicians, 5 pediatric physiotherapists, 4 gait lab technicians and 1 resident in rehabilitation medicine), with an average experience of eleven years (SD 6.9, range 1-24 years) in clinical gait analysis. All participants completed the questionnaires, and eleven semi-structured interviews were performed until data saturation was reached. Participants indicated a great need for a structured and uniform method for gait analysis interpretation. The GAIT.SCRIPT tool met the participants' high expectations and mean scores in 15 out of 17 items on usability exceeded the scale's midpoint, suggesting a generally positive evaluation of the tool. The mean overall SUS score was 62.4, reflecting a usability that hovers between acceptable and good. Many participants expressed challenges with adopting a new interpretation method. They emphasized that the tool could offer significant value in educating novices on gait data interpretation. To improve usability and potentially reduce the processing time required for gait interpretation, participants suggested expanding the automation of GAIT.SCRIPT annotations, providing additional user guidance, and enabling personalization of the interface layout.

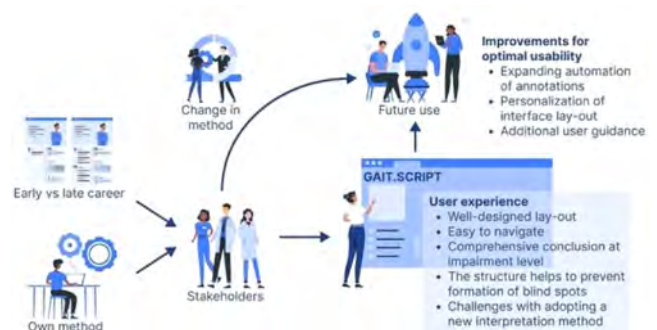


Figure 1: Semi-structured interview data implied that user experiences with GAIT.SCRIPT were influenced by stakeholders' work experiences and the method that they currently used. Stakeholders expressed generally good usability of the tool, but anticipated challenges with adopting a new method. Several improvements of the GAIT.SCRIPT tool were suggested for optimal usability.

## Discussion

The complex and time-consuming process of interpreting clinical gait data of children with CP demands an automated, structured and uniform method.<sup>3</sup> The web-based version of GAIT.SCRIPT is a user-friendly tool that meets professionals' expectations and has the potential to save time, educate novices and improve transparency of clinical gait analysis. Further implementation of the tool should acknowledge the challenge of engaging professionals in a new work method. Additional changes to the tool's automation, user guidance and personalization may optimize GAIT.SCRIPT's usability.

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### Clinical and instrumented assessments of trunk movement and posture in spastic and dyskinetic cerebral palsy – a scoping review

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### Introduction

Trunk control is fundamental for enabling functional movement and supporting normal musculoskeletal development. In individuals with cerebral palsy (CP), particularly those classified at GMFCS levels IV-V, trunk control is often severely impaired, significantly impacting participation and quality of life. Despite its clinical importance, the assessment of trunk control remains underexplored, hindering accurate identification and targeted management. A wide range of tools exist, yet psychometric evidence is limited and consensus on appropriate tool selection is lacking. This study presents a scoping review on clinical and instrumented tools used for assessing trunk control in individuals with spastic and dyskinetic CP, critically evaluating existing tools and highlighting gaps in the field.

### Research Question

(I) Which clinical and instrumented assessment tools and features are used to assess trunk control in individuals with spastic and dyskinetic CP? What are the key characteristics of studies evaluating these tools, including their measurement approach, test protocol, and study sample?

(II) Which psychometric characteristics of these tools and features have been assessed?

### Methods

A systematic search was conducted across seven databases in accordance with the PRISMA-ScR guidelines and using the combined concepts 'Cerebral Palsy' and 'Trunk Control'. Studies were screened by two independent reviewers at title/abstract and full-text level. Inclusion criteria required studies to evaluate reliability, validity, or responsiveness of tools designed to assess trunk movement and posture in individuals with spastic or dyskinetic CP. Risk of bias was independently assessed using the AXIS tool.

### Results

The search yielded 27.332 publications, of which 110 met the inclusion criteria (see Figure). Twenty-two distinct assessment tools were identified, including 12 clinical and ten instrumented tools. Instrumented assessment tools were more frequently evaluated (65%) than clinical tools, with 3D motion analysis being the most common. Validity was the most frequently reported psychometric property (62%), followed by reliability (30%), whereas responsiveness was not reported. Most studies focused on the spastic CP population (85%), with only four addressing dyskinetic CP. Only 28% of studies included individuals classified at GMFCS levels IV-V, reflecting a significant research gap potentially hindering clinical implementation.

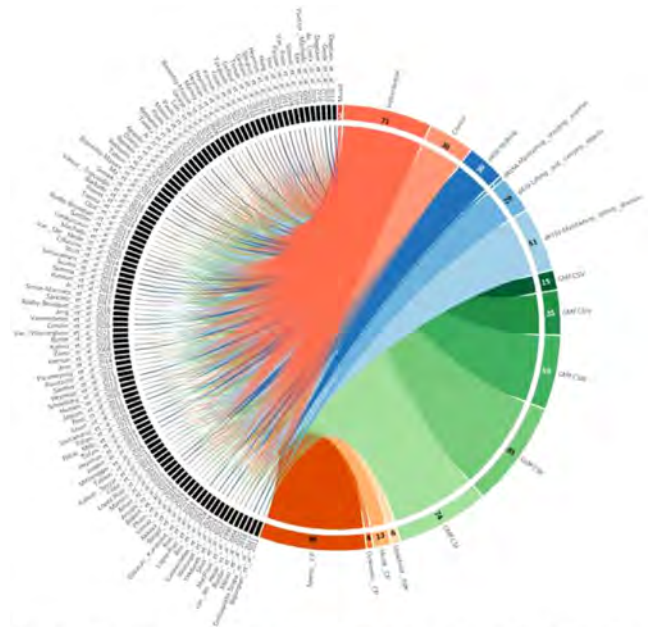


**Figure:** Circos plot showing links between included studies (black), assessment approach (orange), tasks (blue), GMFCS levels (green), and CP diagnosis (red).

### Discussion

This review highlights critical research gaps on trunk control assessment in CP, particularly for individuals with severe motor impairment. The limited psychometric evidence, underrepresentation of high-need populations, diverse and non-inclusive test protocols, and limited coverage of trunk control facets hinder both scientific progress and clinical implementation. By mapping existing tools and critically evaluating strengths and limitations, this review offers first practical guidance for tool selection and identifies priorities for tool development. Future research should address feasibility, reliability, validity and responsiveness of assessment tools across diverse presentations and applications, towards targeted management strategies of trunk control.

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## Assessing Trunk Control in Dyskinetic Cerebral Palsy Using Pressure Mapping: A Pilot Study

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## Introduction

Trunk control, the ability to stabilize and orient trunk movement and posture, is essential for daily life activities and musculoskeletal development. However, in individuals with dyskinetic cerebral palsy (DCP), presenting with involuntary movements and abnormal postures, trunk control is often significantly impaired. While the Segmental Assessment of Trunk Control (SATCo) is commonly used to clinically evaluate trunk control in DCP, it lacks objectivity and may not detect subtle postural differences. Validating objective and clinically feasible assessment tools such as pressure mapping is crucial to improve understanding and tailor management of trunk control in DCP.

## Research Question

What is the known-group validity of pressure mapping parameters – specifically peak pressure, center-of-pressure (CoP) distance, anterior-posterior (AP) sway and left-right (LR) sway – for differentiating facets of trunk control in individuals with DCP versus typically developing (TD) peers?

## Methods

This cross-sectional study included ten adolescents with DCP (ages 15–25 years, GMFCS levels II–V) and ten age- and gender-matched TD controls. Participants completed 20 seated trunk tasks designed to assess static, active and reactive trunk control under seven decreasing levels of external support, following the SATCo as reference clinical protocol. A pressure-sensitive mattress (XSensor) was used to objectively quantify trunk alignment and stability during these tasks, via pressure mapping parameters. Known-group validity was examined by comparing DCP and TD participants using Mann-Whitney U tests.

## Results

Significant differences were found in most tasks between the DCP group (N = 10; mean age 19.9 years, range 17.1–25.5 years; GMFCS levels II–V) and the TD group (N = 10; mean age 21.2 years, range

15.1–24.11 years). Specifically, LR sway differed in 19 out of 20 tasks ( $U = 0-22$ ,  $p < .001-.035$ ), CoP distance in 18 tasks ( $U = 0-22$ ,  $p < .001-.035$ ), AP sway in 14 tasks ( $U = 0-23.5$ ,  $p < .001-.043$ ), and peak pressure in six tasks ( $U = 13-23$ ,  $p = .004-.043$ ). Data exploration (see Figure) revealed substantial variability in the individual presentation of impaired trunk control across individuals with DCP, ranging from near-typical to excessive sway and/or pressure responses within the same task.

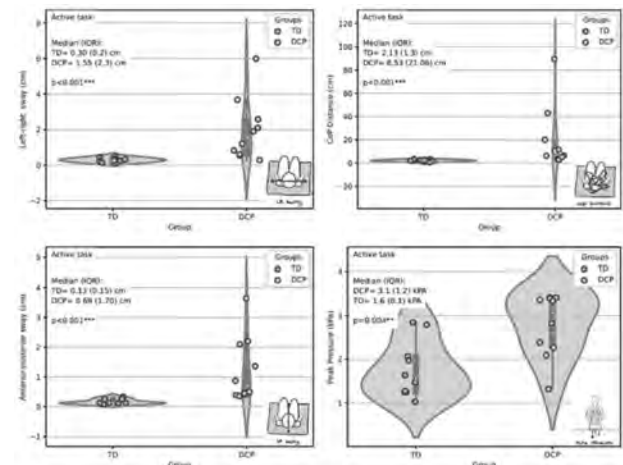


Figure 1 Differences between individuals with dyskinetic cerebral palsy (DCP) and typically developing peers (TD) in an active trunk task for left-right (LR) sway, center-of-pressure (CoP) distance, anterior-posterior (AP) sway, and peak pressure. Individual values (green/purple spots) as well as medians and inter-quartile ranges are presented. Within the violin plot the median is plotted as white spot, the inter-quartile range as grey bar and the upper and lower adjacent value as thin grey line.

## Discussion

This pilot study provides preliminary yet compelling evidence supporting the use of pressure mapping as an objective and sensitive tool to detect impaired trunk control in individuals with DCP. Across a structured set of trunk tasks, individuals with DCP consistently showed significantly greater LR sway, CoP distance, AP sway, and peak pressure compared to their TD peers. Findings suggest that pressure mapping offers a valuable augmentation to traditional clinical tools as SATCo, advancing precision and objectivity. Moreover, the methodology enables more nuanced insights into the complex multifaceted nature of trunk control. The substantial heterogeneity observed within DCP further underscores the clinical relevance of personalized trunk management. To enhance robustness of findings, identify most informative tasks, and validate additional parameters, research with larger and GMFCS-stratified samples is needed.

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## Longitudinal relation between muscle impairments and gait pathology in growing boys with Duchenne muscular dystrophy

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## Introduction

Children with Duchenne muscular dystrophy (DMD) develop progressive gait pathology due to progressive muscle weakness and contractures. Only one cross-sectional study (n = 15) has investigated the relationship between muscle weakness and gait, finding no meaningful associations [1]. Given the progressive and heterogeneous nature of DMD, larger and longitudinal studies are needed. Understanding how the progression rate of muscle impairments relates to the progression rate of gait deviations, could clarify the potential role of muscle impairments in gait deterioration in DMD.

## Research Question

How do specific muscle impairments relate to specific gait deviations, both at starting levels and in progression rates?

## Methods

Thirty-one boys with DMD (aged 4.6-16.4 years), were repeatedly assessed (1-10 times) over 0.5-6.6 years, with intervals of 0.4-3.5 years, yielding 152 total observations. Fixed dynamometry, goniometry, and 3D gait analysis were used to assess hip extension (HE), hip abduction (HA), knee extension (KE) and dorsiflexion (DF) strength, passive DF range of motion (ROM), and gait. Strength and ROM outcomes were converted into unit-less z-scores using anthropometric-related strength percentile curves [2,3] and age-related ROM reference values [3,4] of typically developing children, reflecting strength and ROM deficits. Spatiotemporal parameters were non-dimensional. Predefined gait features (e.g. maxima, minima) were calculated from the mean gait waveforms. Joint random-effects models between gait and muscle outcomes were fitted in SAS®(v9.4). The correlation between the random intercepts ( $r_a$ ) and random slopes ( $r_b$ ) indicated the degree to which the starting levels and progression rates over time of two outcomes were related, respectively.

## Results

Correlations between gait features and muscle impairments are shown in Fig.1(A) with two model visualizations in Fig.1(B-C). At starting levels, reduced HE strength was moderately associated with increased anterior pelvic tilt (Fig.1(B)) and reduced HE moment. Reduced HA strength was moderately associated with increased step width. Reduced KE strength was strongly associated with decreased KE moment and knee power absorption. Reduced DF ROM was strongly associated with decreased DF at initial contact (Fig.1(C)) and moderately with decreased DF in swing. Regarding progression rates, the rate of decrease in HE strength was moderately associated with the rate of increase in posterior trunk lean and strongly with the rate of increase in anterior pelvic tilt (Fig.1(B)). The rate of decrease in HA strength was moderately associated with the rate of increase in internal foot progression. The rates of decrease in KE strength and DF ROM were moderately associated with the rate of decrease in DF at initial contact (Fig.1(C)).

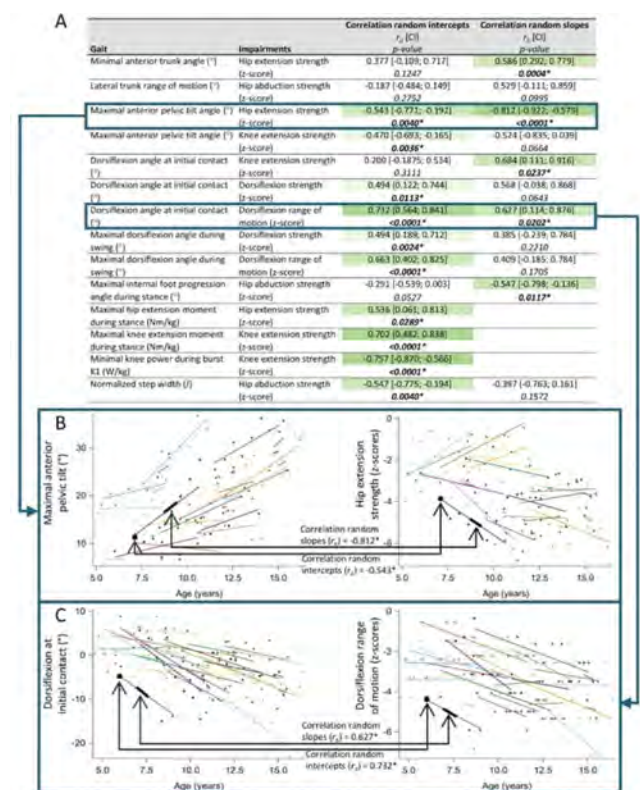


**Fig. 1.** (A) Correlation coefficients of random intercepts and slopes of joint random-effect models between gait features and muscle impairments. By linking the random effects of two outcomes in a joint model, Pearson correlations between random intercepts ( $r_a$ ) and random slopes ( $r_b$ ) can be calculated.  $r_a$  and  $r_b$  indicate the degree to which the starting levels and progression rates over time of two outcomes are related, respectively.  $p$ -values in bold indicate significance at  $p < 0.05$ . The colors indicate the strength of the relationship: dark green, high; middle green, moderate; light green, low. (B-C) Visualizations of joint random-effect models between (B) maximal anterior pelvic tilt angle and hip extension strength, and between (C) dorsiflexion angle at initial contact and dorsiflexion range of motion. The individual predicted profiles (colored lines) with starting level (l) and progression rate (r), and the actual observed values (colored dots) of the children with DMD are displayed with each color representing a different subject. CI, 95% confidence interval; DMD, Duchenne muscular dystrophy; kg, kilogram; Nm, Newton meter; W, Watt;  $r_a$ , Pearson correlation coefficient of random intercepts;  $r_b$ , Pearson correlation coefficient of random slopes.

## Discussion

The findings provide valuable insights into the underlying mechanisms of pathological gait in DMD. The observed associations highlight the relevance of targeted clinical assessments. These findings offer a foundation for optimizing rehabilitation strategies, orthotic management, and orthopedic interventions, ultimately improving clinical decision making and enhancing mobility outcomes in DMD.

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<https://doi.org/10.1016/j.gaitpost.2025.07.277>

## Medial gastrocnemius muscle growth in children with cerebral palsy compared to typically developing children: Deficit trajectories and impact of severity

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## Introduction

Muscle growth is hampered in children with cerebral palsy (CP). We recently modelled medial gastrocnemius (MG) muscle volume (MV) trajectories [1,2]. However, the absence of longitudinal typically developing (TD) data, covering a large age range, has hindered direct comparisons of MG-MV growth rates between CP and TD. Moreover, Gross Motor Function Classification System (GMFCS) levels were pooled. Hence, it remains unclear how MG-MV deficits relative to TD evolve with age and whether these deficit trajectories differ by GMFCS-level.

## Research Question

- (1) How does MG-MV growth differ between GMFCS-I, -II, -III, and TD?
- (2) How do MG-MV deficits evolve with age and do they differ by GMFCS-level?

## Methods

The longitudinal CP database included 383 measurements (GMFCS-I/-II/-III=203/113/67) of 116 children with CP (GMFCS-I/-II/-III=60/35/21) and 236 measurements of 62 TD children. MG-MV was measured by 3D freehand ultrasound, and converted into unit-less z-scores using anthropometric-related TD percentile curves to express MV deficits [3]. (Non-)linear mixed-effect models were fitted in SAS®(v9.4) to investigate the change in absolute MV (mL) and MV deficit (z-score) with age per group. The significance of the regression coefficients, both within and between groups, was assessed using F-tests ( $p < 0.05$ ).

## Results

At 12 months, MG-MV was smaller in CP versus TD ( $p = 0.0083$ ) (Fig.1(A)). Before age 2-3, MG-MV grew rapidly and at similar rates in TD, GMFCS-I, and GMFCS-II, while MG growth in GMFCS-III was already lower than in TD, GMFCS-I, and -II ( $p < 0.0001$ ). After age 2-3, MG growth in TD, GMFCS-I, and -II decreased compared to earlier ( $p = 0.0008$ ). From this point, MG growth in GMFCS-I and GMFCS-II diverged from TD with lower growth rates ( $p_{\text{GMFCS-I}} = 0.0004$ ;  $p_{\text{GMFCS-II}} = 0.0098$ ), but was comparable to GMFCS-III ( $p_{\text{GMFCS-I}} = 0.1155$ ;  $p_{\text{GMFCS-II}} = 0.2372$ ). At age 10, TD showed a marked increase in MG growth compared to earlier ages, unlike the CP groups ( $p < 0.0001$ ). At 12 months, the MG-MV deficit was -0.43, -0.84, and -1.47 in GMFCS-I, -II, and -III, respectively, with a significant difference between GMFCS-I and GMFCS-III ( $p = 0.0098$ ) (Fig.1(B)). Deficits increased with age in all GMFCS-levels ( $p < 0.0001$ - $p = 0.0034$ ). GMFCS-III trajectory showed larger and a more rapid increase in deficits compared to GMFCS-I ( $p < 0.0001$ ) and GMFCS-II ( $p = 0.0085$ ).

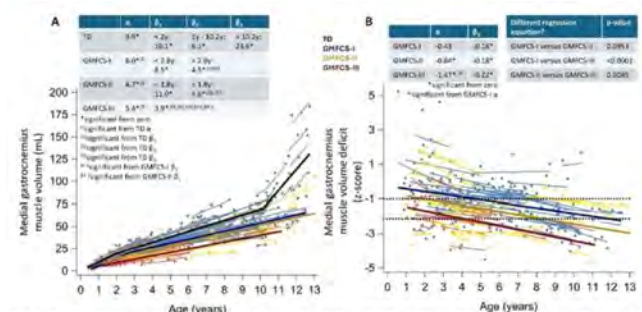


Fig. 1. (A) Medial gastrocnemius muscle volume (mL) trajectories with age for TD, GMFCS-I, GMFCS-II, and GMFCS-III. (B) Medial gastrocnemius muscle volume deficit (z-score) trajectories with age for GMFCS-I, GMFCS-II, and GMFCS-III. The z-scores represent muscle volume deficits in reference to anthropometric-related TD percentile curves. A z-score of 0 means that the absolute muscle volume equals the median absolute muscle volume of TD peers. Z-scores of -1 and -2 (black dashed horizontal lines) indicate that the absolute muscle volume deviates by one and two standard deviations below the median of TD children, respectively. The solid colored lines are the average trajectories for each group. The thinner colored lines are the individual predicted profiles, and the dots are the actual observed values. Significance level was set at  $p < 0.05$ . a, intercept at 12 months; b, regression slope; GMFCS, gross motor function classification scale; mL, milliliter; TD, typically developing; y, years.

## Discussion

This study highlights the impact of GMFCS-levels on muscle growth and is the first to describe MG-MV deficit trajectories in CP. GMFCS-III exhibited the largest initial MG-MV deficit and the lowest initial growth, potentially related to delayed ambulation, emphasizing the need for early stimulation of muscle development in this group. In contrast, GMFCS-I and -II demonstrated smaller initial

deficits but growth rates similar to TD during the first 2-3 years. However, their growth rates decreased thereafter, highlighting a critical time window where stimulating muscle growth becomes particularly crucial. Ongoing analyses of additional muscle size parameters will provide a more comprehensive understanding of muscle development in CP.

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## From motion to muscle: exploring the influence of movement mechanics on muscle activation patterns in common exercises

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## Introduction

Bodyweight training is a widely adopted strength training modality in rehabilitation for children with cerebral palsy (CP). Bodyweight exercises engage multiple joints and coordinate activation of various muscle groups [1], and are thus considered effective for strengthening multiple muscle groups simultaneously [2-4]. However, inadequate exercise performance can affect muscle activation and reduce the effectiveness of strength training. Altering joint positions can shift muscle recruitment patterns, potentially decreasing the activation of the target muscle or engaging alternative muscles [5-8]. A review of current literature identified sit-to-stand, squats, step-ups, bridging, and heel raises as commonly prescribed exercises for children with CP. Despite their frequent use, biomechanical evaluations of bodyweight exercises in this population are lacking. Hence, the impact of exercise performance characteristics on muscle activation patterns and intensity remains unclear in CP.

## Research Question

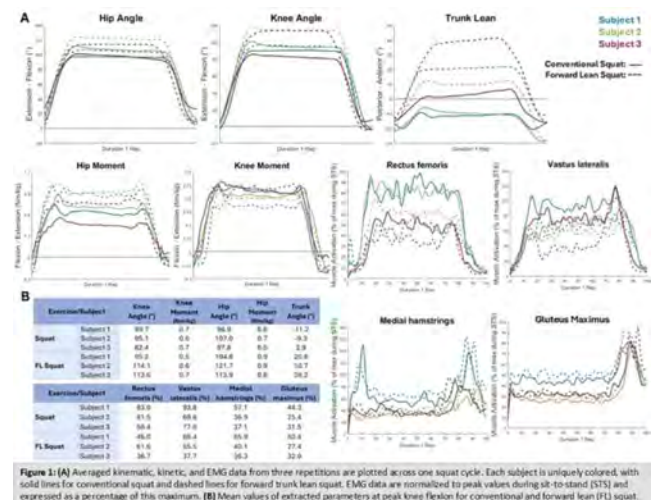
What is the effect of variations in movement mechanics on muscle activation patterns during common bodyweight exercises?

## Methods

The current pilot study was conducted on healthy adults to assess the validity of an assessment protocol involving two variations of a squat exercise. Three healthy females (mean age:  $25.6 \pm 0.5$  years) performed three repetitions of a conventional and a forward trunk lean squat. Kinematic and kinetic data were collected using 3D motion analysis (VICON®, Plug-in-Gait) and two force platforms (AMTI®). Surface electromyography (Cometa®) was recorded from 11 lower limb muscles (incl. rectus femoris, vastus lateralis, medial hamstrings, and gluteus maximus), and normalized to peak activation during a sit-to-stand task. Joint angles and internal moments (normalized to bodyweight) at the trunk, hip and knee, as well as muscle activation at peak knee flexion, were averaged across repetitions and compared between squat variations and participants.

## Results

Preliminary findings confirmed increased anterior trunk displacement relative to the pelvis during the forward trunk lean squat. Moreover, kinematic data showed increased hip and knee flexion compared to the conventional squat. Kinetic data suggest higher hip extension moments during the forward lean squat, while knee extension moments were reduced. EMG data indicated reduced muscle activity in the rectus femoris and vastus lateralis and a slight increase in gluteus maximus activation during the forward lean squat (Figure 1).



## Discussion

Results from this pilot study demonstrate consistent alterations in muscle activation patterns by adjusting joint positions in healthy adults. Combined kinematic, kinetic, and EMG data indicated that moving the trunk forward during a squat shifts muscle activation from the knee extensors to the hip extensors. Understanding muscle recruitment patterns associated with exercise performance has potential implications for exercise selection and the effectiveness of training programs. We will assess squats and additional exercises, such as step-ups, bridging, and heel raises in typically developing

children and children with CP with the ultimate goal to define a biomechanically justified lower limb exercise program for CP.

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## Biomechanical evaluations of bodyweight exercises: a scoping review of current practices

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## Introduction

Bodyweight training is an accessible strength training modality, with demonstrated applicability and effectiveness across both healthy and clinical populations [1–4]. However, inadequate exercise performance can lead to injuries, and may reduce the effectiveness of the strength training [5,6]. Biomechanical evaluations of bodyweight exercises are therefore an important focus in research, providing a foundation for optimal exercise selection. Nevertheless,

current literature lacks biomechanical frameworks to define standardized performance criteria and optimize exercise selection across clinical populations.

## Research Question

Which biomechanical methods are commonly reported in the literature to assess the impact of exercise performance on the effectiveness of bodyweight exercises?

## Methods

A search strategy was developed around three key concepts: exercise, biomechanical analysis, and lower limbs. Searches were conducted across four databases: Pubmed (including MEDLINE), Web of Science, Scopus and Embase. Studies were screened by three independent researchers based on predefined eligibility criteria. Exclusion criteria encompassed investigations of intervention outcomes, sport-specific movements (e.g., karate kicks, pole vaulting), use of external resistance (e.g. barbells or machines), and studies employing musculoskeletal modeling. Inclusion was limited to studies published within the last decade and quality appraisal was performed using the modified Downs and Black checklist [7]. Relevant data were extracted on study objectives, exercise performance, target muscle groups, data collection methods and analysis techniques.

## Results

The initial search yielded 3909 results, of which 53 studies met the predefined eligibility criteria. Across studies, a total of 1121 participants were evaluated, with 94% of studies including healthy individuals. Patient populations were limited to one publication including children with cerebral palsy, one study on post-stroke adults, and one study evaluating older adults with osteoarthritis. Study aims varied from comparing exercise modifications to identifying optimal exercises for specific target muscles. Most studies (93%) thoroughly described experimental methods, containing detailed outlines of exercise protocols and performance criteria. Surface electromyography was the most frequently used evaluation method (85%), followed by three-dimensional motion capture (32%), kinematic (30%), and kinetic data collection (23%). Seven studies combined movement mechanics and muscle activation assessments. Normalization of EMG data was most often performed using maximal voluntary contractions (89%). Notably, the study on stroke patients used gait trials, while the study involving children with cerebral palsy employed functional exercises for normalization purposes.

## Discussion

The biomechanical evaluation of bodyweight exercises is a well-established research area. Despite detailed and replicable protocols, key limitations in literature persist. Integrated analyses that capture both movement mechanics and muscle activation are uncommon. Furthermore, pediatric and patient populations are significantly underrepresented. In patient populations, the current literature lacks standardized normalization procedures using functional tasks rather than maximal contractions.

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## Video observations are reliable for gait pattern assessment in Children with cerebral Palsy

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## Introduction

Gait alterations are common in children with cerebral palsy (CP). Three-dimensional gait analysis (3DGA) is the gold standard for assessing gait and evaluating interventions aiming to enhance and maintain walking ability. Its accessibility is however limited in many clinical settings. Video observations are widely used as alternative, yet no reliable classification system exists for this method.

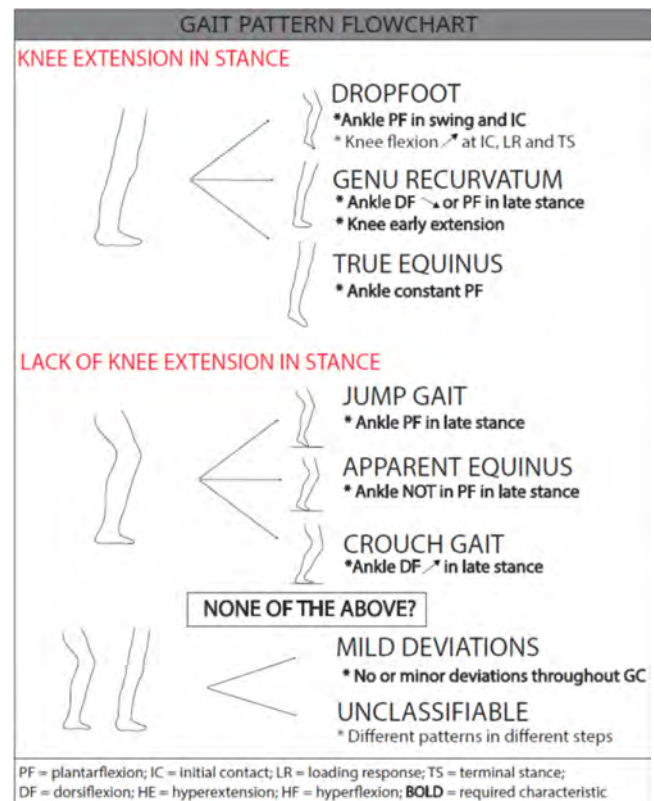
## Research Question

Can gait patterns be reliably recognized from sagittal plane gait videos?

## Methods

A convenience sample of children with spastic CP age 4–18 was obtained from retrospective clinical gait analyses at Karolinska Hospital between 2008 and 2025. Sagittal plane gait pattern was scored on a test set of 30 videos (15 left legs; 15 right legs) individually by 5 raters: kinesiologist, trainee medical doctor, orthotist, pediatric orthopaedic surgeon and physiotherapist (3 experienced in 3DGA, 2 unexperienced). Six gait pattern definitions

(drop foot, genu recurvatum, true equinus, jump gait, apparent equinus, crouch) described by Papageorgiou et al. [1], and two additional scoring options (mild, unclassifiable) were used. A scoring flowchart (Fig.1) was created based on knee and ankle segment orientation defining the key characteristics differentiating between gait patterns.



The original definitions by Rodda et al. [2] were used for clarifications. The leg closest to the camera was scored by looking at the child walking at natural speed, followed by observing gait in slow motion. Prior to scoring, training sessions were conducted including group evaluation of 10 videos and two test rounds of 10 individually scored videos each, leading to flowchart adjustments. All videos were quality-checked and sorted for right -and left leg. Primary study outcomes were inter-rater agreement among five raters scoring eight gait patterns from sagittal plane videos, using Fleiss' Kappa (k) in SPSS.

## Results

The dataset consisted of 25 individuals with spastic CP, aged 4–16 years: mean age 9.9 + /- 3y; (unilateral 7; bilateral 18), GMFCS I–III (I = 12; II = 10; III = 3). Three individuals were scored bilaterally, two on different occasions. Preliminary results indicated good between-rater agreement;  $\kappa = 0.65$  (95% CI [0.60–0.70];  $p < 0.001$ ) with k values for individual category ranging from 0.56–0.84 except for 'unclassifiable' (0.18). Prevalence of the gait patterns (for which 3/5 raters agreed; n = 28) was: genu recurvatum/mild deviations n = 6; apparent equinus n = 5; crouch gait n = 4; jump gait/dropfoot/true equinus n = 2; unclassifiable n = 1.

## Discussion

This study contributes to the development of a cost-effective, practical and accessible system for describing gait patterns in children with CP using standardized video recordings. Our results demonstrating good inter-rater agreement, support the reliability of the video-based approach. Further analysis on a larger sample and on the validity of the descriptions in relation to kinematic data are required. A system to describe gait patterns in a reliable way could enhance clinical decision-making, improve intervention monitoring and ultimately contribute to improved functional outcomes for children with CP.

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## The complex interplay between knee kinematics, motor functioning and passive knee range of motion in children with spastic cerebral palsy

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## Introduction

Three-dimensional gait analysis (3DGA) data enables the objective evaluation of gait and the effect intervention strategies to maintain and improve walking ability in children with cerebral palsy (CP). The complex interplay between motor functioning, walking and passive range of motion is relatively unexplored in this population, despite its importance for intervention strategies and clinical decision making.

## Research Question

What are the associations between motor functioning, gait pattern and available passive knee extension in children with CP?

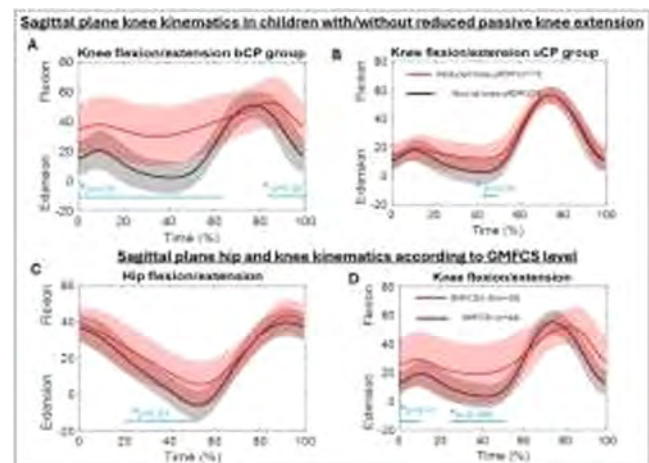
## Methods

A convenience sample of participants with spastic CP, Gross Motor Functioning Classification System (GMFCS) level I-III; age 4-18 was recruited. Retrospectively collected 3D-gait data between 2008-2025 at the Karolinska University Hospital, Sweden was included after consent. Primary outcomes were sagittal plane kinematics and maximum passive knee extension. Kinematics were obtained from infra-red markers placed on the skin, processed with a plug-in-gait lower limb model. Angles were time-normalized and averaged over 3 trials. Canonical correlation with statistical parametric mapping was used to associate passive maximum knee extension as measured with a goniometer with sagittal pelvis, hip,

knee and ankle angles. T-tests assessed the kinematic differences between GMFCS levels I and II-III.

## Results

3D gait data of 72 children (age 4-16 years; mean age 10.4  $\pm$  2.9), 40 bilateral (bCP, GMFCS I=16; II=18; III=6) and 32 unilateral CP (uCP, GMFCS I=28; II=4) was included. Passive knee extension was below neutral in 13/40 and 11/32 children in the bCP/uCP group respectively, with a median knee extension deficit of -10° (n=24). Pelvis angles and passive knee extension correlated for the whole gait cycle in the bCP group [0-100%;  $p < 0.001$ ], but not in the uCP group. For the hip, the uCP and bCP group showed similar associations with passive knee extension [30-65%;  $p = 0.009$  & 29-65%;  $p = 0.02$ ]. For the knee, there was a significant association between knee angles and passive knee extension for most of the gait cycle for the bCP group [0-65%;  $p < 0.01$  & 84-99%;  $p = 0.02$ ] (Fig 1A), but only in 3% for the uCP group [41-44%;  $p = 0.04$ ] (Fig 1B). Hip and knee flexion were significantly different between GMFCS groups during most of stance (hip: 28-61%;  $p < 0.01$  & knee: 0-7%;  $p < 0.001/24-55%$ ;  $p = 0.008$ ) with reduced hip and knee extension for higher GMFCS levels (Fig. 1 C/D).



## Discussion

The associations between structural properties of the knee and knee position during gait in the bCP group imply that restricted range of motion may be driven by functional constraints. Especially, since the available knee range of motion exceeded the knee position during walking. These findings are clinically important since knee flexion during walking is commonly addressed for orthopedic management. Additionally, our results highlight the importance of differentiating between GMFCS levels when tailoring interventions to improve walking ability in children with spastic CP.

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No references

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## Exploring gait adaptations during dual-tasking in typically developing children: A systematic review and meta-analysis

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### Introduction

Gait analysis is typically performed in clinical settings, focusing solely on walking performance. However, in daily life, we walk while performing additional tasks simultaneously (dual-tasking). This could come with a cost, particularly in children who are still refining their motor skills [1] and even more for children with motor automatization impairments like Developmental Coordination Disorder (DCD). In typically developing (TD) children distinct effects based on task type have been reported [2], but the effect of age and a comparison between gait parameters has yet to be studied. This is important to understand developmental differences between TD children and children with DCD.

### Research Question

What is the effect of dual-tasking on gait performance in TD children and is there a difference based on age, type of additional task, and between different gait parameters?

### Methods

Four databases were searched and studies were independently screened by two authors against the eligibility criteria: TD children (<18y), walking only condition (single-task condition), walking combined with an additional task (dual-task condition), and gait parameters. Relevant population characteristics, information on the additional task, and means and standard deviations on gait parameters during single- and dual-task conditions were extracted to perform a meta-analysis. Only gait parameters reported at least five times in at least three studies were analyzed. Effect sizes were calculated as the Standardized Mean Differences (SMD) and data were transformed so that positive values represented a dual-task cost. The effect of age and task-type were assessed through subgroup analyses ( $p < 0.05$ ).

### Results

A total of 62 articles were included, yielding sufficient data on 256 parameters including spatiotemporal ( $n=190$ ) and variability ( $n=66$ ), measured in 1,109 children (age range = [4.8y,15.1y]). Random effects models showed significant dual-task effects on both spatiotemporal (SMD=1.28,  $p < 0.0001$ ) and variability (SMD=1.48,  $p < 0.0001$ ) parameters (Figure 1). SMD values were higher for temporal (1.36) and temporal-variability parameters (1.71) than spatial (1.04) and spatial-variability parameters (1.11). There were significant effects of task-type on both spatiotemporal

and variability parameters ( $p < 0.01$ ), with cognitive tasks causing greater dual-task interference (1.52 95%CI = [1.18,1.85]; 1.77 [1.04,2.49], respectively) than motor tasks (0.90[0.67,1.12]; 0.36 [-0.13,0.85]). There was no effect of age ( $p = 0.18$ ;  $p = 0.29$ ).

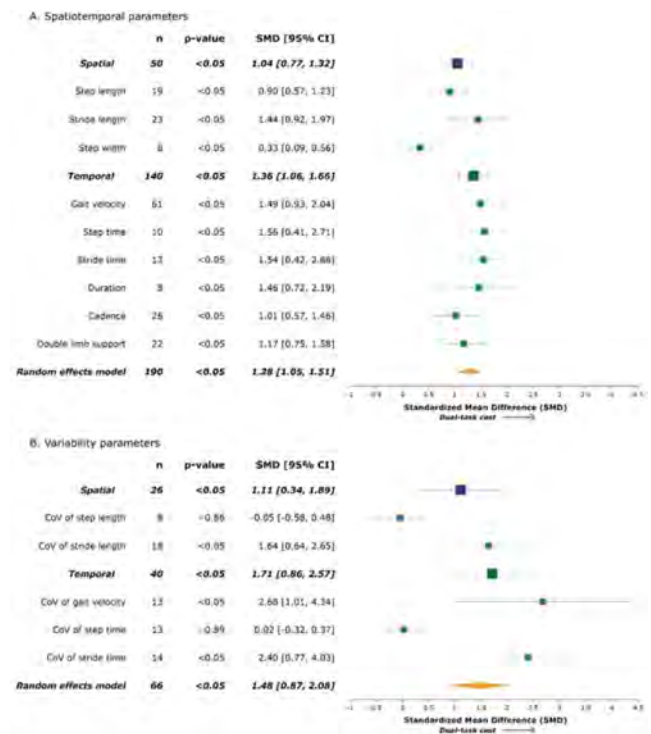


Figure 1. Forest plots visualizing the Standardized Mean Differences (SMD) for the spatiotemporal (A) and variability (B) parameters.

### Discussion

When dual-tasking, TD children adopt a distinct spatiotemporal gait pattern with increased variability to meet the task demands. Our results confirm previous findings that cognitive tasks have a greater effect on gait than additional motor tasks [2]. Moreover, children tend to prioritize temporal over spatial adjustments. This distinctive gait profile highlights the importance of studying gait in functional, dual-task contexts. Incorporating analyses of kinetic, kinematic, symmetry, and balance parameters remains essential. An encompassing understanding of gait in these functional contexts is also crucial for the assessment and treatment of pathologies like DCD.

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## Lower limb muscle fatigue after uphill walking in children with unilateral spastic cerebral palsy

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### Introduction

Fatigue during walking is a common complaint in cerebral palsy (CP).

### Research Question

The primary purpose of this study is to investigate muscle fatigue from surface electromyography (sEMG) measurements after a treadmill-based fatigue protocol with increasing incline and speed in children with CP with drop foot. The secondary purpose is to investigate whether changes in sagittal kinematics of hip, knee and ankle occur after fatigue.

### Methods

Eighteen subjects with unilateral spastic CP performed the protocol while wearing their ankle-foot orthosis and scored their fatigue on the OMNI scale of perceived exertion. The median frequency (MF) and root mean square (RMS) were used as sEMG measures for fatigue and linear mixed effects model were applied.

### Results

The MF was significantly decreased in fatigued condition, especially in the affected leg and in the tibialis anterior and peroneus longus muscle. The RMS did not change significantly in fatigued condition, while the OMNI fatigue score indicated patients felt really fatigued. No changes in sagittal kinematics of hip, knee and ankle were found using statistical non-parametric mapping.

### Discussion

In conclusion, the current fatigue protocol seems promising in inducing fatigue in a population with CP with drop foot and it could be used to expand knowledge on muscle fatigue during walking in CP.

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## Gait Phase Importance in Affected Side Prediction for Cerebral Palsy via Gradient-Based Analysis

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### Introduction

Gait impairments in cerebral palsy (CP) are often asymmetric, and accurately identifying the affected side is crucial for treatment planning and intervention [1]. Clinical gait analysis may not always provide the granularity needed to detect subtle differences in gait patterns [2,3]. Machine learning offers a promising solution, but many models are considered "black boxes" [4,5], making it difficult for clinicians to interpret the results. This study aims to bridge this gap by applying gradient-based sensitivity analysis to gait patterns, offering interpretable insights into the specific phases of the gait cycle that most influence the classification of the affected side.

### Research Question

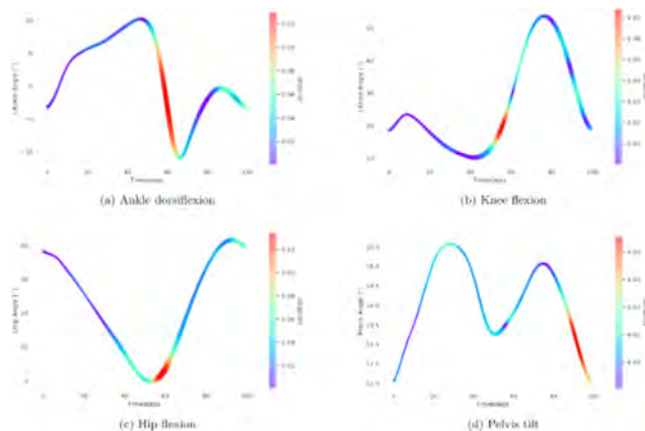
Can a gated recurrent unit (GRU) [6] accurately classify affected side in patients with CP using lower-body gait angles, and which temporal regions of the gait cycle contribute most to the model's decisions?

### Methods

We analyze gait kinematics from patients with CP: hemiplegia (n=76, mean age=12.39 years) and diplegia (n=73, mean age=12.58 years). The kinematics data was split by side, with each side labeled as either affected or non-affected. Our prediction task leverages these side-specific kinematics comprising ankle, knee, and hip flexion/extension as well as pelvis tilt angles. These angles were temporally normalized to 100% of the gait cycle duration. A GRU was trained to classify sides based on these curves. To interpret model behavior, we computed the absolute gradient of the loss with respect to each input feature, identifying which gait phases most influenced predictions. Feature importance was averaged across the training set to reveal consistent patterns.

### Results

On an unseen test set, the model achieved 91% accuracy in classifying the affected side. Sensitivity analysis revealed that the most informative segments of the gait cycle occur around 60% and 90%—corresponding to the toe-off and terminal swing phases, respectively. These regions exhibited the highest gradient magnitudes, indicating their strong influence on model predictions.



## Discussion

Our findings highlight biomechanically critical gait phases that are most discriminative in classifying affected sides in CP. The use of gradient-based sensitivity not only enhances interpretability but also aligns well with clinical understanding of gait impairments in CP. This method offers a promising path toward transparent, data-driven tools for supporting clinical gait assessments.

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## Individuals with hip Osteoarthritis walk with less symmetry and stability during natural walking measured by a single trunk IMU.

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## Introduction

Hip osteoarthritis (OA) affects about one in four individuals during their lifetime [1] and is characterized by joint cartilage degeneration, leading to pain, stiffness and altered gait dynamics [2,3]. Traditional gait analysis in labs may not reflect natural walking behaviors and is impractical for regular clinical use. Assessing gait through a single trunk IMU in clinical settings offers a more practical alternative to complex laboratory settings, allowing for capturing natural gait movements with valuable biomechanical insights in a clinical hospital setting.

## Research Question

Do individuals with hip OA exhibit altered gait characteristics compared to healthy controls during natural walking, as measured by a single trunk IMU?

## Methods

Fifty-five individuals with end-stage hip OA ( $63 \pm 11.2$  years) and forty-four healthy controls ( $66 \pm 8.14$  years) underwent gait assessment while walking freely in clinical settings and wearing a single trunk IMU (Byteflies, Antwerp, Belgium). This device recorded 3D acceleration and angular velocity at 200 Hz and was attached to the lower back. Four key gait parameters including step symmetry, stability, smoothness, and regularity were calculated using specific algorithms including autocorrelation, Lyapunov exponents, SPARC and sample entropy.

Gait quality between persons with hip OA and healthy controls was compared using independent sample t-tests or the Wilcoxon rank sum test. Statistical analyses were conducted in RStudio (version 2023.6.0.421, RStudio, Inc.)

## Results

Persons with hip OA showed distinct gait characteristics during natural walking compared to healthy controls (Figure 1). They showed reduced step symmetry and stability in all three directions. Moreover, persons with hip OA walked more smoothly in the VT and ML directions. In terms of gait regularity, persons with hip OA walked more regularly in the ML direction and less so in the AP direction.

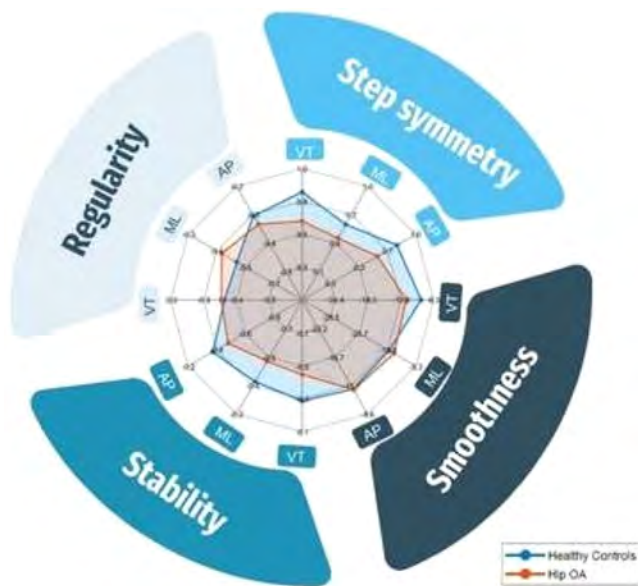


Figure 1. Spider plot to illustrate group differences in gait parameters, including step symmetry, stability, smoothness and regularity across the three directions—anteroposterior (AP), vertical (V), and mediolateral (ML). Persons with hip osteoarthritis (OA) are shown in red and healthy controls in blue.

## Discussion

All 55 participants had unilateral hip OA, which typically results in degenerative changes that unevenly affect both sides of the hip joints, leading to asymmetries [4]. The decrease in stability aligns with existing literature that associates severe hip OA with deteriorated gait stability and an increased risk of falls [2,5].

The higher regularity and smoothness in the ML direction, indicates a lower variability that might suggest an impaired ability to adjust their gait when needed in ML direction [6]. However, greater variability in the AP direction might suggest a loss of control in the AP plane, potentially reflecting impairments caused by hip OA progression [7].

The findings highlight that using an IMU in clinical settings effectively assesses altered gait quality in individuals with hip OA.

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## Lower Limb Joint Moments in Children with Haemophilia

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## Introduction

Haemophilia is a hereditary bleeding disorder characterised by deficiencies in clotting factors VIII (Haemophilia A) or IX (Haemophilia B) [1]. The most common clinical manifestation of haemophilia is recurrent joint bleeds (haemarthrosis), which leads to joint damage and, over time, the development of haemophilic arthropathy [2]. Joint damage in haemophilia is progressive and often occurs through asymptomatic bleeds, which can go undetected until the disease reaches advanced stages [3]. Early identification of biomechanical alterations could enable timely intervention, potentially improving long-term outcomes for affected individuals. Studies using three-dimensional gait analysis (3DGA) to investigate joint kinetics in haemophilia are insufficiently explored [4]. This study aims to investigate whether knee and ankle moments differ between children with haemophilia (CWH) and age-matched typically developing controls.

## Research Question

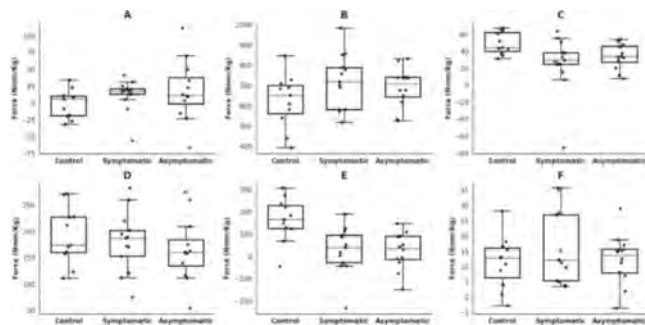
How do knee and ankle joint moments differ between children with haemophilia and age-matched, typically developing individuals, and to what extent are these differences influenced by a history of symptomatic joint bleeds?

## Methods

A prospective cross-sectional study was conducted at the One Small Step Gait Laboratory at Guy's Hospital, involving 25 male CWH (aged 7–11 years). Participants were divided into two groups: those with at least one clinically observed joint bleed and those without clinically recorded joint bleeds. An age-matched, typically developing control group was used for comparison. Motion capture data were collected using a Vicon system employing the conventional gait model (version 1), and joint moments were calculated. A Shapiro-Wilk test assessed data normality. An unpaired t-test was conducted to compare knee and ankle joint moments between haemophilia and control groups, while a one-way ANOVA with Tukey's HSD post-hoc test examined differences between haemophilia subgroups and controls.

## Results

No statistically significant differences were found in coronal plane knee and ankle moments between CWH and controls ( $p = 0.314$ ,  $0.767$ ), nor between haemophilia subgroups ( $p = 0.3356$ ,  $0.8795$ ). However, CWH exhibited significantly lower sagittal plane knee extensor moments compared to controls ( $p = 0.020$ ), with reductions observed in both symptomatic and asymptomatic subgroups ( $p = 0.0039$ ,  $0.0047$ ). Additionally, ankle rotation moments were significantly different between symptomatic haemophilia participants and controls ( $p < 0.0448$ ).



## Discussion

While coronal plane moments did not significantly differ between groups, the findings suggest that early biomechanical adaptations in haemophilia may be more prominent in the sagittal and transverse planes. The reduction in sagittal plane knee moments across both haemophilia subgroups suggests that altered gait mechanics may precede clinically observable joint damage, reinforcing the potential for 3DGA as an early screening tool. Future longitudinal research should investigate whether these alterations in joint kinetics progress over time and whether 3DGA can be used for the screening for and monitoring of haemophilic arthropathy.

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## Learning to walk with passive ankle foot orthoses – Does it happen spontaneously?

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## Introduction

Ankle foot orthoses (AFOs) are provided to people with lower leg impairments and can improve their walking ability. AFOs limit the ankle range of motion, creating a new walking context. People are expected to adapt their gait to this new context, in order to minimize walking energy cost. This requires changes in gait and likely variability, which may not occur spontaneously in some individuals, for example due to fear of falling. Insights into the adaptation process with AFOs is crucial to identify the necessity and targets of AFO-related gait training. To minimize the effects of motor control impairments on adaptation, we first explored gait adaptations in healthy individuals.

## Research Question

1. Does walking energy cost, gait and gait variability change spontaneously after provision of an AFO?
2. What is the required time to minimize walking energy cost after AFO provision?

## Methods

We included 14 healthy participants (age  $24 \pm 2$ ) without motor impairment and previous experience with AFOs. To acclimatize to treadmill walking (GRAIL, Motek), participants first walked 20-minutes with normal shoes. Thereafter, participants were fitted bilaterally with a dorsal leaf AFO (4.5 Nm/degree). With AFOs, participants walked for another 20-minutes at 5 km/h. Simultaneously VO<sub>2</sub> and VCO<sub>2</sub> were measured (Cosmed K5) to determine walking energy cost. Step length, step width, ankle angle at midstance, and peak ankle power, and their respective variability, were determined for the third and last minute.

People that demonstrated a >5% reduction in walking energy cost between the third (highest point) and last minute were classified as spontaneous adapters. Time to adaptation was determined for the adapters by calculating the zero point of the derivative of a second order fit of walking energy cost, which can be longer than 20 minutes. A  $p < 0.10$  was considered significant due to the exploratory nature of this analysis.

## Results

Between the third and last minute, walking energy cost, step width and variability of the maximal ankle angle significantly reduced while step length step length and variability of step width increased (see table). The average estimated time to adaptation was  $24.2 \pm 9.3$  minutes. Adapters showed larger reductions in the variability of ankle angle at midstance and peak ankle power variability (see table).

|                                | Over time (whole group (n=14)) |                         | Adapters vs non-adapters |                    | p-value      |
|--------------------------------|--------------------------------|-------------------------|--------------------------|--------------------|--------------|
|                                | 1 <sup>st</sup> minute         | 20 <sup>th</sup> minute | Adapters (n=7)           | Non-adapters (n=7) |              |
| Walking energy cost (J/kg/m)   | 4.41±0.31                      | 4.22±0.33               | <b>0.020</b>             | 0.02±0.17          | <b>0.001</b> |
| Step length (cm)               | 74.1±3.3                       | 75.1±2.6                | <b>0.100</b>             | 1.5±2.5            | 0.445        |
| Step Width (cm)                | 13.4±2.5                       | 12.5±2.4                | <b>0.021</b>             | -1.1±1.2           | 0.794        |
| Ankle angle midstance (deg)    | 9.7±3.2                        | 10.2±3.2                | <b>0.008</b>             | 0.2±1.3            | 0.349        |
| Maximal ankle power (W)        | 1.55±0.34                      | 1.33±0.35               | 0.533                    | -0.04±0.16         | 0.727        |
| <b>Variability measures</b>    |                                |                         |                          |                    |              |
| Step length SD (cm)            | 1.5±0.3                        | 1.3±0.3                 | 0.511                    | 0.5±0.3            | 0.762        |
| Step Width SD (cm)             | 2.2±0.5                        | 2.3±0.5                 | <b>0.026</b>             | 0.4±0.5            | 0.475        |
| Ankle angle midstance SD (deg) | 0.6±0.2                        | 0.5±0.1                 | <b>0.026</b>             | -0.2±0.2           | <b>0.008</b> |
| Maximal ankle power SD (W)     | 0.12±0.03                      | 0.11±0.05               | 0.727                    | -0.02±0.01         | <b>0.004</b> |

## Discussion

On group level, walking energy cost reduced and gait biomechanics changed considerably during the first 20-minutes of AFO use. However, only half of the participants adapted spontaneously towards minimizing walking energy cost. Spontaneous adapters reduced ankle angle and peak ankle power variability more suggesting they learned a new gait pattern. Gait training focussing on promoting exploration of new gait patterns, such as gait variability training, might therefore be indicated for AFO users.

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## Do footwear properties matter when walking with AFOs? A predictive simulation study.

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## Introduction

Ankle-foot orthoses (AFOs) are often provided to improve gait and reduce walking energy cost in people with plantarflexor weakness. Usually, AFOs are used within footwear (AFO-Footwear-

Combination; AFO-FC). Footwear properties, such as rocker radius, apex position and heel-toe differential can affect ankle kinetics and kinematics<sup>1</sup>, possibly dependent on AFO properties.

## Research Question

What is the effect of rocker radius, apex position and heel-toe differential, in interaction with AFO stiffness, on walking energy cost, kinematics and kinetics in people with plantar flexor weakness?

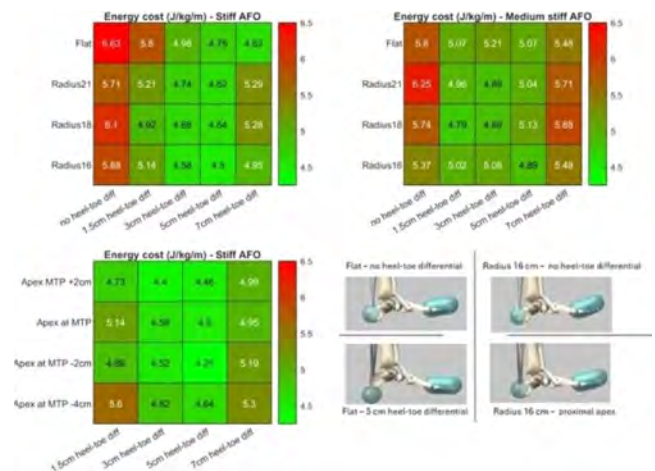
## Methods

We employed a planar musculoskeletal-model with nine degrees of freedom and nine muscles per leg to perform forward-dynamic predictive walking simulations using a reflex-based muscle controller. Plantar flexor weakness was modelled by reducing maximal strength of the gastrocnemius and soleus by 80%, while an AFO with neutral alignment was modelled as massless torsional spring around the ankle<sup>2</sup>. As default, footwear (24.5 cm long) was modelled without heel-toe differential, apex position at the MTP joint and no radius (e.g. flat). For a medium-stiff (150 Nm/deg) and a stiff AFO (350 Nm/deg), the rocker radius (flat, 21, 18 and 16 centimeter), the heel-toe differential (adding 1.5, 3, 5 and 7 centimeters), and the apex position (at MTP joint, 2 and 4 cm proximal, 2 cm distal of MTP joint) were varied by relocating the contact spheres. Effects on walking energy cost and lower limb gait kinematics and kinetics were analyzed.

## Results

For the medium-stiff AFO, a radius of 18 cm with a heel-toe differential of 1.5 cm maximally reduced walking energy cost. For the stiff AFO, the effects of shoe properties on walking energy cost were more pronounced, with the lowest energy cost found at a 16 cm radius, 5 cm heel-toe-differential, and an apex position 2 cm proximal of the MTP joint (see figure).

For both AFO stiffness levels, increasing rocker radius led to a small reduction in maximal ankle dorsiflexion angle and moment during stance. Increasing the heel-toe differential substantially reduced ankle dorsiflexion, ankle moments and increased internal knee extension moments in the loading response. Moving the apex position proximally reduced maximal ankle dorsiflexion, maximal ankle moment, and knee extension and internal knee flexion moment during terminal stance.



## Discussion

Footwear properties can have a large effect on gait outcomes when walking with AFOs. Based on the simulations, having a forefoot rocker and some heel-toe differential is most beneficial, although the best combination depends on AFO stiffness. For clinical practice this means that both footwear and AFO properties should be optimized in conjunction to maximize results, which warrants future experimental studies.

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## Clusters of frontal knee angle range-of-motion in healthy people during level walking

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## Introduction

Level walking (LW) is among the most investigated human motor tasks. Various papers on normative gait data of healthy participants [1,2] and characteristic deviations of subgroups [3] or in comparison to pathological conditions [4,5] have been published. However, control of frontal leg alignment, its kinematic and kinetic indicators, and their cut-offs concerning sustainable joint health are still not fully understood [6]. Therefore, we started establishing a database on LW for in depth investigation of normal gait in healthy adults.

## Research Question

Is frontal knee range of motion (fKROM) a suitable parameter for classifying subgroups of dynamic functional leg alignment qualities during LW in healthy adults?

## Methods

Seventy-five (40 female / 35 male) healthy participants aged 19 to 65 (mean  $34 \pm 13.6$ ) years with BMI 18.5 to 30.39 (mean  $23.6 \pm 2.5$ ) kg/m<sup>2</sup> performed LW. Data were captured by a 3D marker-based system (Vicon Nexus) using the modified Cleveland Clinical markerset and three force plates (AMTI). Processed gait cycle trials were time-normalized. Based on dominant leg fKROM a k-means cluster analysis ( $k=3$ ) identified subgroups of participants. Discrete

gait parameters were analyzed for differences between the three clusters using one-way ANOVA with Bonferroni posthoc testing. Further, lower limb kinematics over the entire gait cycle were assessed using one-dimensional statistical parametric mapping (1D-SPM) with one-way ANOVA ( $p < 0.05$ ) to detect differences between clusters.

## Results

fKROM ranged from  $0.8^\circ$  to  $4.4^\circ$  (mean  $2.3^\circ \pm 1.0^\circ$ ) with cluster centers of  $1.4^\circ$  ( $n = 31$ ),  $2.3^\circ$  ( $n = 26$ ) and  $3.7^\circ$  ( $n = 18$ ). Neither, spatio-temporal parameters, nor other relevant frontal and transversal knee, hip and pelvis angles, and external frontal moments differed significantly between the three clusters. Concurrent 1D-SPM analysis revealed no significant differences across clusters for any of the tested kinematic parameters. (Figure 1)

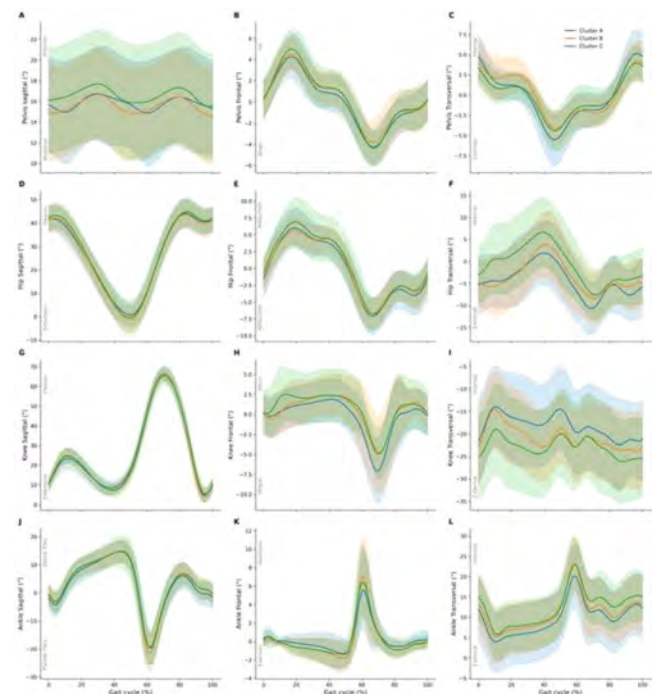


Figure 1: 1D-SPM plots of kinematic parameters by clusters of fKROM

## Discussion

Findings from this preliminary analysis suggest that individuals do not differ in relevant loading-response phase gait parameter point estimates nor across the entire gait cycle, despite the identification of distinct fKROM clusters. A combined parameter other than fKROM during loading phase, which is not yet established, may be more sensitive to characterize dynamic functional leg alignment. Furthermore, kinematic patterns indicative of unfavorable dynamic alignment should be investigated in relation to kinetic overloading to enhance clinical relevance.

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## Perturbation training improves muscle coordination during reactive standing balance in trained and untrained conditions in children with cerebral palsy

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## Introduction

Balance impairments are common in cerebral palsy (CP). When standing balance is perturbed using support-surface displacements, children with CP step at lower perturbation amplitudes and have higher levels of muscle co-activation than typically developing (TD) children (1–4). Perturbation training is a promising tool to improve reactive balance (5). However, it is unclear whether perturbation training normalizes muscle coordination in children with CP.

## Research Question

What is the effect of perturbation training on reactive balance performance and reactive muscle activity during trained and untrained tasks in children with CP and TD children?

## Methods

Children with CP (GMFCS: I-II; age 9-16) and TD children (age 8-16) were assessed before and after a training intervention consisting of nine sessions over three weeks. Participants were asked to maintain standing balance in response to toe-up rotations (96 per session; fig1). Perturbation magnitude was adapted to the

participant's performance level. We assessed balance performance and muscle coordination in trained toe-up rotational and untrained backward translational perturbations of standing balance. Balance performance was assessed as the number of perturbations completed without stepping. The magnitude of muscle activity was computed as the average filtered and scaled EMG (lateral gastrocnemius (LG), medial gastrocnemius (MG), soleus (SOL), and tibialis anterior (TA)) from 200 to 450ms after perturbation onset, the time bin in which we previously observed the largest differences in reactive muscle activity between children with CP and TD children (4). Muscle co-activation was assessed using the co-contraction index (CCI), i.e. the overlap between TA and respectively LG, MG, and SOL filtered and scaled EMG 0 to 1.5s after perturbation onset.

## Results

Preliminary results are based on data from 6 (out of 11) children with CP and 4 (out of 11) TD children. Perturbation training improved the ability of children with CP to withstand perturbations without stepping. Reactive muscle activity in plantarflexors and dorsiflexors decreased after training during rotational (mean pre-post LG activity of 0.02 vs. 0.01 in TD, 0.51 vs. 0.07 in CP) and translational (mean pre-post LG activity of 0.37 vs. 0.22 in TD, 1.45 vs. 0.76 in CP) perturbations (fig1). Muscle co-activation also decreased during rotational (CCI pre-post of 0.17 vs. 0.10 in TD, 0.68 vs. 0.38 in CP) and translational perturbations (CCI pre-post of 0.19 vs. 0.12 in TD, 0.46 vs. 0.32 in CP) (fig1).

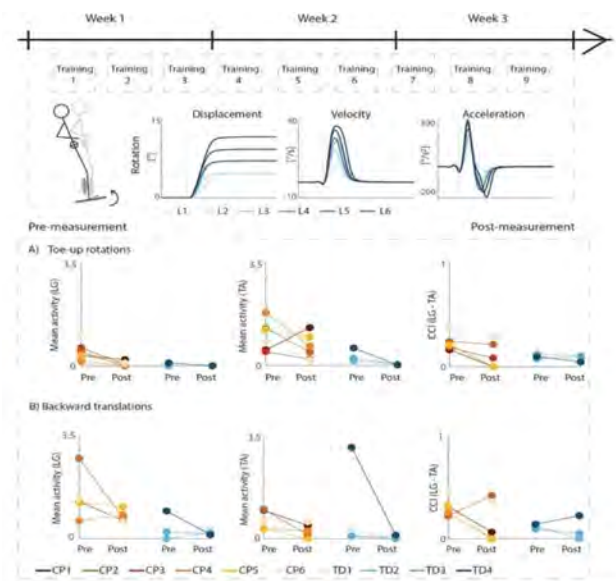


Figure 1: Experimental set-up (top panel). Platform kinematics for the different perturbation levels (L1-L6) for toe-up rotations used during training. Average normalized EMG for the lateral gastrocnemius (LG) and tibialis anterior (TA) and co-contraction index (CCI) for level 2. (A) toe-up rotational perturbations and (B) backward translational perturbations. Children with cerebral palsy (CP) in red-yellow and typically developing (TD) children in blue.

## Discussion

Our preliminary results suggest that reactive balance control can be trained in children with CP. After training, the muscle response to perturbations was smaller in both children with CP and TD children and co-activation was reduced in children with high levels of co-activation pre-training. Overall, the response to perturbations of children with CP was more similar to the response of TD children after training.

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## Evaluating gait patterns from a predictive simulation framework against gait variations observed in knee osteoarthritis patients

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## Introduction

Knee osteoarthritis (KOA) is a disease characterized by progressive cartilage degeneration(1), exacerbated by excessive knee joint loading. Gait retraining has emerged as a potential intervention to modify gait patterns and reduce joint loading(2). However, current gait retraining interventions are usually generic and do not account for patient-specific anatomical malalignment, which is known to be critical when estimating medial compartment loading(3). Alternatively, predictive computational workflows (4) may facilitate identifying gait retraining patterns that minimize joint loading while accounting for patient specific joint alignments.

## Research Question

Are gait retraining strategies identified by predictive simulations more effective in reducing knee joint contact forces than compensatory strategies observed in KOA patients?

## Methods

Motion capture data of overground walking from an early KOA cohort (n = 7) was collected. Models for each patient were scaled and modified to include patient-specific alignment, derived from medical imaging. These models along with measured patient-specific kinematics were used as input to a predictive simulation framework.

Standard settings of the predictive simulation formulation were modified to include minimization of medial compartment knee contact forces while loosely tracking patient-specific input kinematics. Medial compartment contact forces of the patient-specific gait pattern, the optimized gait pattern as well as the population-based gait patterns(6) were calculated using the COMAK simulation framework(5) to verify the extent to which the predictive simulation derived gait retraining strategies further reduced the medial compartment contact forces compared to population-based strategies.

## Results

Compared to the patient-specific gait pattern, predictive simulation derived gait retraining strategies successfully reduced medial knee contact force mainly by modifying hip abduction and rotation, trunk forward & lateral bending. These changes in the kinematic pattern are consistent with the characteristics of the population-based synthetic gait patterns which presented with the largest reduction in medial compartment loading in these patients. Predictive simulation derived gait kinematics resulted in an average decrease of 13.7% ( $\pm$  22.0%) at first peak loading and 42.6% ( $\pm$  14.1%) at second peak loading. Kinematic changes were highly heterogeneous between patients. For example, subject 1 (Figure 1), would benefit from walking with more internal hip rotation while subjects 2,3 and 4 would profit from an outward rotation in the hip.

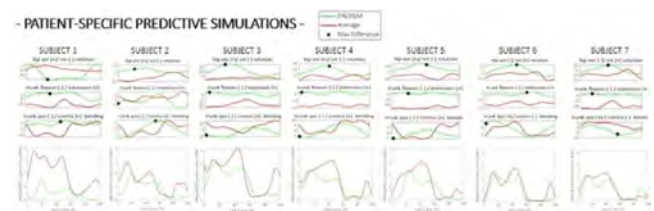


Figure 1. Patient-specific predictive simulation output (green) and average experimental kinematics used for tracking (red) are presented in 3 degrees-of-freedom (y-axis: joint angle (%), x-axis: gait cycle (%)) along with corresponding medial compartment contact forces (y-axis: bodyweights (BW), x-axis: gait cycle(%)).

## Discussion

This analysis demonstrates the ability of the predictive simulation framework to generate gait patterns that reduce medial compartment knee contact force with changes to the patients' preferred gait pattern. The amount of kinematic change is highly heterogeneous between patients, highlighting the important interaction between joint alignment and gait patterns. Future work will assess the ability of individuals to learn and retain these gait patterns over time. The integration of predictive simulations into gait retraining protocols has the potential to enhance patient outcomes by providing targeted, data-driven strategies to mitigate knee joint loading in KOA patients.

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## Is Gait Symmetrical in Hereditary Spastic Paraplegia?

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### Introduction

Hereditary spastic paraplegia (HSP) is a genetic neurological disorder characterized by spasticity and weakness due to corticospinal tract degeneration causing gait impairments. [1] Anecdotally, asymmetry has been noted when comparing the right and left sides of the body.

Asymmetry is the amount of divergence between the left and the right side of the body, often associated with pathology, but also present because of limb dominance [2,3].

3-dimensional gait analysis (3DGA) provides kinematic and kinetic gait data. The Gait Deviation Index (GDI) and the Gait Profile Score (GPS) allow gait deviations to be quantified with respect to variability from typical; however, one assumes the right and left are similar to allow the mean to be used as a representative value. [4].

### Research Question

Are the gait deviations in hereditary spastic paraplegic symmetrical when evaluated with objective 3-dimensional gait analysis?

### Methods

A retrospective review of prospectively collected gait data was performed on genetically confirmed patients with HSP. Gait data were processed and calculation of their GDI, GPS and associated Movement Analysis Profile (MAP), as well as cadence parameters were completed for each extremity. The Symmetry Index (SI) [5] was used to compare left/right GDI, GPS, MAP scores, and left/right cadence parameters and gait symmetry is acceptable when the SI <10%. [5,6]

### Results

40 HSP patients, 18 female and 22 male, mean age  $10.37 \pm 5.66$  years were analyzed. HSP genotypes included 14 SPG4, 9 SPG3a, and 17 miscellaneous types. Acceptable symmetry values and very strong correlations ( $>0.85$ ) for the GDI, Step Time, Step Length, and Double Support Time were found. Symmetry for the GPS is considered good with a strong correlation. Individual MAP scores show a greater level of asymmetry. Mann-Whitney comparisons did not find significant differences between genders, age groups, or different genotypes.

| Gait Variable:         | Value: Left | Value: Right | SI %  | rho (p-value)  |
|------------------------|-------------|--------------|-------|----------------|
| GDI                    | 67.10       | 69.93        | 9.21  | 0.88 (<0.0001) |
| GPS                    | 11.18       | 10.54        | 15.35 | 0.89 (<0.0001) |
| Cadence Parameters:    |             |              |       |                |
| Step Time              | 115.68      | 117.03       | 5.85  | 0.85 (<0.0001) |
| Step Length            | 84.20       | 83.93        | 6.46  | 0.93 (<0.0001) |
| Single Support %       | 94.73       | 94.35        | 5.17  | 0.62 (<0.0001) |
| Double Support %       | 124.10      | 125.85       | 7.96  | 0.92 (<0.0001) |
| GPS MAP Variables:     |             |              |       |                |
| Pelvis-Sagittal        | 6.83        | 6.93         | 15.73 | 0.95 (<0.0001) |
| Pelvis-Coronal         | 3.28        | 3.15         | 23.46 | 0.86 (<0.0001) |
| Pelvis-Transverse      | 6.24        | 6.38         | 25.30 | 0.84 (<0.0001) |
| Hip-Sagittal           | 9.16        | 9.17         | 26.70 | 0.79 (<0.0001) |
| Hip-Coronal            | 5.02        | 4.55         | 37.56 | 0.53 (<0.0001) |
| Hip-Transverse         | 11.23       | 10.80        | 36.76 | 0.56 (<0.0001) |
| Knee-Sagittal          | 13.55       | 13.59        | 21.96 | 0.92 (<0.0001) |
| Ankle-Sagittal         | 13.60       | 12.97        | 22.78 | 0.82 (<0.0001) |
| Foot Progression Angle | 15.20       | 13.64        | 40.78 | 0.67 (<0.0001) |

Table 1: Comparison between gait variables including the Gait Deviation Index (GDI), the Gait Profile Score (GPS), the GPS Movement Analysis Profile (MAP) scores and cadence parameters on the left and right sides of patients with Hereditary Spastic Paraplegia (HSP). Comparisons were completed using the Symmetry Index (SI) for all variables, as well as Spearman's Rank Correlations (rho)  $p < 0.05$  for the GDI, GPS, and cadence parameters.

### Discussion

Despite heterogeneity within the HSP population, individual gait deviations appear symmetrical, suggesting corticospinal tract injury is also symmetrical. Good to acceptable SIs and moderate to strong correlations were found for GDI and GPS scores as well as cadence parameters. The decreased symmetry found across the various MAP scores shows increased variability across joints and/or planes of motion and overall gait scores created through a consolidation of kinematic data may hide the smaller asymmetries present throughout a heterogeneous group such as with HSP.

Gait deviations in patients with HSP appear symmetrical when comparing GDI, GPS, and temporal parameters. Individual joints and planes may show differences when comparing sides, as shown by increased variability in the MAP scores.

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## Selective Motor Control Correlates with Gait Function in Patients with Hereditary Spastic Paraplegia

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### Introduction

Selective motor control (SMC) is the ability to isolate the activation of muscles in a specific pattern in response to demands of a voluntary movement. Hereditary spastic paraplegia (HSP) is a heterogeneous group of neurologic disorders that present with bilateral lower extremity spasticity and weakness due to CST degeneration. [1]. SMC is related to the functioning of the corticospinal tracts (CST) and degeneration alters control of force, speed, timing, and patterns of voluntary movements [2].

SMC can be reliably assessed using the Selective Control Assessment of the Lower Extremity (SCALE) and has been shown to correlate to Gross Motor Function Classification System (GMFCS) levels as well as Gait Profile Scores (GPS) in patients with cerebral palsy [3,4].

### Research Question

Do alterations in selective motor control predict gait deviations in patients with hereditary spastic paraplegia?

### Methods

A retrospective review of prospectively collected data from genetically confirmed HSP patients with comprehensive 3D gait Analysis (3DGA) and SCALE results within 3 months of 3DGA was reviewed. Cadence parameters and GDI was calculated for each extremity. Spearman's Rank Correlations were calculated looking for associations between GDI and SCALE scores with the entire cohort, between young and older ages, and within SPG4 and SPG3a genotypes.

### Results

23 patients, mean age  $12.87 \pm 6.61$  years eight females and 15 males met inclusion criteria. HSP genotypes included 10 SPG4, 7 SPG3a, and 6 miscellaneous types. Spearman's Rank Correlations showed a strong correlation ( $>0.70$ ) between the SCALE scores and the GDI scores for the entire HSP cohort. (Table 1) Within SPG4 and SPG3a patients a very strong correlation was found between average GDI and total SCALE.

| Total Cohort (n=23)             |               | Spearman Correlation | p-value |
|---------------------------------|---------------|----------------------|---------|
| Left SCALE Score                | Left GDI      | 0.729                | 0.0001  |
| Right SCALE Score               | Right GDI     | 0.714                | 0.0001  |
| Total SCALE Score               | Avg GDI       | 0.719                | 0.0001  |
| Males (n=15)                    |               |                      |         |
| Left SCALE Score                | Left GDI      | 0.650                | 0.0087  |
| Right SCALE Score               | Right GDI     | 0.679                | 0.0053  |
| Total SCALE Score               | Avg GDI       | 0.669                | 0.0064  |
| Females (n=8)                   |               |                      |         |
| Left SCALE Score                | Left GDI      | 0.383                | 0.3490  |
| Right SCALE Score               | Right GDI     | 0.685                | 0.0609  |
| Total SCALE Score               | Avg GDI       | 0.482                | 0.2265  |
| Pediatric Age <10 (n=10)        |               |                      |         |
| Left SCALE Score                | Left GDI      | 0.883                | 0.0294  |
| Right SCALE Score               | Right GDI     | 0.767                | 0.0096  |
| Total SCALE Score               | Avg GDI       | 0.744                | 0.0136  |
| Total SCALE Score               | Walking Speed | 0.895                | 0.0256  |
| Adolescent/Adult Age ≥10 (n=13) |               |                      |         |
| Left SCALE Score                | Left GDI      | 0.669                | 0.0125  |
| Right SCALE Score               | Right GDI     | 0.462                | 0.1123  |
| Total SCALE Score               | Avg GDI       | 0.606                | 0.0283  |
| SPG4 Genotype (n=10)            |               |                      |         |
| Left SCALE Score                | Left GDI      | 0.707                | 0.0221  |
| Right SCALE Score               | Right GDI     | 0.704                | 0.0231  |
| Total SCALE Score               | Avg GDI       | 0.669                | 0.0345  |
| Total SCALE Score               | Walking Speed | 0.869                | 0.001   |
| SPG3a Genotype (n=7)            |               |                      |         |
| Left SCALE Score                | Left GDI      | 0.764                | 0.0457  |
| Right SCALE Score               | Right GDI     | 0.946                | 0.0013  |
| Total SCALE Score               | Avg GDI       | 0.852                | 0.0148  |

Table 1. Comparison between various Selective Control Assessment of the Lower Extremity (SCALE) scores and various indicators of gait function, including the Gait Deviation Index (GDI) for patients with Hereditary Spastic Paraplegia (HSP). \*p<0.05

### Discussion

Impairments in gait in HSP are due to spasticity, weakness, possible joint contracture, and poor selective motor control. As an assessment of selective control, SCALE can be used as a predictive factor of functional mobility for patients with HSP, especially SPG4 and SPG3a patients. HSP patients with better selective motor control are likely to have fewer gait deviations.

Assessing preoperative selective motor control may be a consideration in treatment, as patients with better control may have better outcomes with both surgical and nonsurgical interventions.

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## Gait Deviations in Hemiplegic Cerebral Palsy: Comparing Outcomes of Achilles Lengthening with and without Medial Hamstring Lengthening

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### Introduction

The most common deformity in hemiplegic cerebral palsy patients is an equinus contracture [1] and is compensated for by a flexed knee during stance, leading to a shortened step length as full knee extension is not achieved either in stance or in terminal swing. Equinus gait pattern also results in instability, increasing the risk of tripping and falling due to insufficient foot clearance in swing phase [2]. Medial hamstring lengthening (MHL) is performed for early knee contractures and may increase terminal knee extension in swing. The purpose of this study is review objective changes gait in patients with hemiplegic cerebral palsy who underwent a tendoachilles lengthening (TAL) or gastrocnemius recession (GSR) for equinus contracture correction, with or without the addition of an MHL.

### Research Question

Will the addition of medial hamstring lengthening improve objective measures of gait including step length and temporal parameters in hemiplegic cerebral palsy patients surgically treated for an equinus contracture?

### Methods

Retrospective review of an IRB approved study was conducted and data for 24 hemiplegic patients with both a preoperative and 1-year postoperative gait analysis visit were analyzed. Of these patients, 16 underwent a TAL or GSR procedure alone, while 8 had a simultaneous TAL/GSR and MHL. The addition of MHL was considered in patients with early knee contracture, usually measuring less than 10 degrees. Kinematic data were collected using a Vicon motion capture system (Vicon, CO) and processed in Nexus with the plug-in-gait model. Temporal parameters were calculated and normalized to age-matched, typically developing individuals. Preoperative and postoperative gait data, including step length, cadence, and ankle and knee range of motion, were compared. Statistical analysis was performed using the Mann-Whitney U Test. Alpha was set to 0.05.

### Results

The addition of MHL to surgery for an equinus contracture led to a statistically increased walking speed that was not seen in the equinus procedure alone cohort. Increased step length also approached statistical significance for this cohort. Both cohorts experienced statistically increased ankle range of dorsiflexion motion, improved knee extension, and ankle position at initial contact. A statistically significant improved ankle maximum dorsiflexion during single leg stance was also seen in both groups. Without MHL, patients showed a statistically significant worsening popliteal angle over time (Table1). As expected with equinus

surgery, patients demonstrated improved ankle motion and kinematics. The addition of a MHL, when indicated, improved walking speed, had some improvement on step length, and was protective against worsening popliteal angles.

| Temporal Parameters                  | TAL/GSR        |               |         | TAL/GSR MHL    |               |         |
|--------------------------------------|----------------|---------------|---------|----------------|---------------|---------|
|                                      | PRE            | POST          | P VALUE | PRE            | POST          | P VALUE |
| Cadence (% age matched normal)       | 94.56 ± 10.21  | 96.44 ± 8.17  | 0.622   | 89.33 ± 10.51  | 95.63 ± 11.35 | 0.141   |
| Walking speed (% age matched normal) | 81.38 ± 11.45  | 85.38 ± 12.78 | 0.407   | 72.13 ± 16.76  | 83.75 ± 14.93 | 0.011   |
| Step Length (% age matched normal)   | 88.19 ± 10.67  | 85.56 ± 11.97 | 0.532   | 76.00 ± 14.22  | 87.88 ± 10.48 | 0.052   |
| Range of Motion and Kinematics       |                |               |         |                |               |         |
| Knee Extension (°)                   | 1.56 ± 3.01    | 0.00 ± 1.83   | 0.06    | -1.13 ± 5.30   | -1.23 ± 4.43  | 0.983   |
| Popliteal (°)                        | 48.44 ± 12.74  | 54.69 ± 10.72 | 0.013   | 50.63 ± 6.23   | 43.75 ± 6.94  | 0.027   |
| Dorsiflexion w/ knee flexed (°)      | -16.25 ± 10.88 | 4.69 ± 7.85   | 0.001   | -11.25 ± 15.98 | 14.38 ± 5.63  | 0.012   |
| Dorsiflexion w/ knee extended (°)    | -20.31 ± 15.78 | 1.25 ± 7.19   | 0.002   | -15.63 ± 16.35 | 8.13 ± 5.30   | 0.012   |
| Maximum Knee Flexion (°)             | 55.91 ± 5.55   | 58.87 ± 4.18  | 0.159   | 52.88 ± 7.23   | 48.47 ± 9.57  | 0.547   |
| Ankle Position IC (°)                | -35.67 ± 14.56 | -15.54 ± 6.67 | 0.001   | -33.00 ± 14.98 | -8.59 ± 5.86  | 0.008   |
| Maximum Dorsiflexion SLS (°)         | -21.23 ± 17.63 | 7.22 ± 5.23   | 0.001   | -21.83 ± 25.41 | 11.94 ± 5.26  | 0.008   |

Table 1. Gait results following surgery in patients who underwent TAL or Gastroc lengthening with or without a medial hamstring lengthening. (IC – initial contact, SLS – single leg stance)

### Discussion

Surgical considerations for isolated TAL/GSR vs. adding MHL should focus on which procedure better improves gait efficiency. Further recruitment of additional patients will improve the power of the study and allow more definitive recommendations for the inclusion of MHL in hemiplegic patients with equinus contracture.

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### Inter-rater agreement of systematic gait data interpretation in children with cerebral palsy using the GAIT.SCRIPT tool

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### Introduction

The interpretation of clinical gait analysis (CGA) data demands a substantial amount of time and expertise, and methods are often not standardized and highly expert dependent. The development of standards would support evidence-based clinical decision-making, ensuring safe and effective healthcare. ESMAC emphasizes this importance with their standard initiative, demonstrating there is

already some agreement on methods for obtaining CGA data across Europe[1]. However, methods for data interpretation have not yet been addressed. To better standardize this interpretation process, we have developed the GAIT.SCRIPT interpretation tool[2,3], based on the impairment focused interpretation method[4]. The tool aims to structure the interpretation of CGA data by automatically linking identified gait features with potential underlying impairments. To bring the tool to the clinic, it is essential to know its reliability.

## Research Question

What is the inter-rater agreement of healthcare professionals and novices using GAIT.SCRIPT for systematic interpretation of CGA data in children with cerebral palsy (CP) to identify 1) deviating gait features; and 2) potential underlying impairments?

## Methods

Nineteen professionals in pediatric CGA and five novices (students with biomechanical knowledge) interpreted two CP cases with the web-based GAIT.SCRIPT tool. The assessors were divided into three groups: novices (no experience,  $n=5$ ), semi-advanced ( $n=8$ ) with  $\leq 10$  years of experience (average 4.5(3.3) years) and experts ( $n=8$ ) with  $> 10$  years of experience (average 15.8(4.5) years). Inter-rater agreement was assessed for identified gait features and included gait impairments within the three groups with Krippendorff's Alpha, ratio observed matches[5], and percentage agreement. Ratios and percentage agreement scores  $\geq 0.8/80\%$  were considered substantial, 0.67/67% to 0.8/80% acceptable and below 0.67/67% low[6].

## Results

Agreement scores among healthcare professionals were considered low for identifying deviating gait features, but acceptable to substantial for the included gait impairments (Table 1). Agreement among assessors was higher in healthcare professionals than in novices. Differences in agreement between experts and semi-advanced groups were less clear, semi-advanced assessors tended to score slightly higher on gait feature selection, where impairment inclusion was almost equal.

|                              | Gait Features     |                      |                  | Potential Impairments |                      |                  |
|------------------------------|-------------------|----------------------|------------------|-----------------------|----------------------|------------------|
|                              | Case              | Krippendorff's Alpha | Observed matches | % Agreement           | Krippendorff's Alpha | Observed matches |
|                              |                   | Alpha [95% CI]       | Ratio            | %                     | Alpha [95% CI]       | Ratio            |
| All healthcare professionals | Case 01<br>$N=16$ | 0.41 [0.39, 0.43]    | 0.62             | 74.2%                 | 0.44 [0.41, 0.47]    | 0.77             |
|                              | Case 02<br>$N=16$ | 0.32 [0.30, 0.34]    | 0.55             | 68.9%                 | 0.41 [0.36, 0.44]    | 0.73             |
| Expert group                 | Case 01<br>$N=8$  | 0.36 [0.32, 0.41]    | 0.59             | 73.3%                 | 0.39 [0.32, 0.46]    | 0.78             |
|                              | Case 02<br>$N=8$  | 0.30 [0.26, 0.34]    | 0.54             | 69.8%                 | 0.50 [0.45, 0.56]    | 0.78             |
| Semi-advanced group          | Case 01<br>$N=8$  | 0.47 [0.43, 0.51]    | 0.65             | 76.5%                 | 0.50 [0.44, 0.55]    | 0.78             |
|                              | Case 02<br>$N=8$  | 0.33 [0.29, 0.37]    | 0.56             | 70.9%                 | 0.33 [0.27, 0.40]    | 0.69             |
| Novice group                 | Case 01<br>$N=5$  | 0.22 [0.14, 0.29]    | 0.52             | 69.8%                 | 0.02 [-0.09, 0.13]   | 0.55             |
|                              | Case 02<br>$N=5$  | 0.19 [0.11, 0.26]    | 0.48             | 67.4%                 | 0.11 [-0.01, 0.22]   | 0.59             |

## Discussion

The Krippendorff's Alpha scores being low, is likely the consequence of skewed data and scores frequently falling into one category. The observed matches and percentage agreement showed consistently higher scores, and better represent the actual agreement, which can be considered acceptable to substantial. Thorough instructions and training sessions for the end-users are likely to increase inter-rater agreement. The lower values in novices suggest

that some expertise in clinical gait analysis remains required for interpretation. Identifying gait features had a systematically lower agreement than included impairment, which implies that the tool guides the assessors towards a more uniform conclusion on underlying impairments. This agreement about included gait impairments by healthcare professionals shows promising results for future implementation of the GAIT.SCRIPT tool into clinical care, providing a structured method for CGA interpretation in children with CP.

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## Influence of walking conditions on spatiotemporal and kinematic gait symmetry following unilateral transtibial amputation

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## Introduction

The primary objective of post-amputation rehabilitation in individuals with lower limb loss is to facilitate the restoration of a biomechanically efficient and symmetric gait pattern. Gait symmetry is a critical determinant of energy expenditure, musculoskeletal health, and functional mobility in daily life [1,2]. It has been observed that amputees tend to exhibit an asymmetrical gait pattern, with less load on the prosthetic limb and shorter stance phase time on this limb when walking, and a longer stance time and more loading on the intact side [3-7]. Furthermore, an asymmetrical gait pattern in amputees has the potential to impact energy consumption, general health and quality of life. Many studies have highlighted the potential benefits of achieving symmetrical gait patterns in prevent-

ing secondary health problems and improving rehabilitation outcomes [6-10]. This study aimed to investigate the effects of various walking conditions on gait symmetry in individuals with unilateral transtibial amputation.

## Research Question

Among gait velocity, slope inclination, and surface characteristics, which parameters significantly influence gait symmetry in individuals with unilateral transtibial amputation?

## Methods

Ten individuals with unilateral transtibial amputation who ambulate independently with prosthetic limbs in daily life were included. Participants performed walking trials under varying conditions of speed, surface texture, and slope. Spatiotemporal gait parameters and lower extremity joint kinematics were recorded using an inertial measurement unit (IMU)-based motion capture system. Symmetry indices were calculated for relevant gait variables to quantify interlimb asymmetries [11].

## Results

Knee joint range of motion in the frontal plane demonstrated greater asymmetry on uneven surfaces compared to level ground. Step length asymmetry was more pronounced during slow-speed walking relative to fast-speed walking. Furthermore, uphill walking was associated with increased asymmetry in swing phase duration and sagittal plane ankle joint kinematics.

## Discussion

The findings suggest that environmental and task-specific modifications, such as surface irregularity, slope gradient, and walking speed, exert differential impacts on gait symmetry in individuals with unilateral transtibial amputation. These results underscore the necessity of incorporating variable walking conditions into clinical gait assessment and prosthetic rehabilitation protocols.

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## Barriers and facilitators of the sedentary parents' exercise behaviours

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## Introduction

The increase in sedentary behavior, which has become a public health problem in our country as well as in the world, has reached even more dangerous dimensions with the Covid-19 Pandemic. As a result, while the severity of chronic health problems increased, some systemic problems began to be encountered at an early age. While the interruption of almost all support systems and services under pandemic conditions deeply affects families with children with special needs (for example, autism spectrum disorder, down syndrome, etc.), it has been observed that physical activity behaviors have begun to create changes in both parents and children. In this context, physical activity behavior modification programs, which have theoretical foundations, are thought to be important for both individual and public health. It is expected that potential changes in parents' physical activity behaviors may also have an impact on their children.

## Research Question

The aim of this study is to create an exercise behavior modification program based on the Web 2.0, transtheoretical model (TTM) in order to increase the physical activity levels of parents with and without

special needs children and to analyze the effectiveness of the program. The first aim of the study was to determine the barriers and facilitators perceived by sedentary parents with and without children with special needs regarding exercise behaviours in order to create a physical activity behaviour change programme based on the transtheoretical model.

## Methods

The facilitators and barriers perceived by the parents were examined using focus group interviews from qualitative research methods (Parker & Tritter, 2006). The data obtained from 18 parents (three focus groups of six parents each) with children with Autism Spectrum Disorder, Down Syndrome and typical development were analysed by content analysis method (Graneheim & Lundman, 2004).

## Results

The findings were visualised with descriptive tables and Venn diagrams to identify the points of agreement and disagreement between each participant group. The most intensely perceived barriers belong to the contemplation stage (Table 1). Here, it was observed that parents of children with special needs perceived the lack of support for childcare and the intensity of housework as barriers. On the other hand, it is seen that the facilitators perceived by the parents of children with special needs are doing exercise with their children, doing physical activity in the home environment and receiving professional support.

**Table 1**  
Sedentary Parents' Perceived Barriers and Facilitators in Contemplation Stage

| Stage | Down Syndrome                                             |                                                    | Typical Development                    |                                           | Autism Spectrum Disorder                           |                                                    |
|-------|-----------------------------------------------------------|----------------------------------------------------|----------------------------------------|-------------------------------------------|----------------------------------------------------|----------------------------------------------------|
|       | Barrier                                                   | Facilitator                                        | Barrier                                | Facilitator                               | Barrier                                            | Facilitator                                        |
| C*    | Unstable environment (n = 3)                              | Encouragement of family members (n = 3)            | Not being used to doing (n = 3)        | Support for housework (n = 4)             | Intensity of housework (n = 4)                     | Coping with previous behaviours (n = 4)            |
|       | Being able to do FA with the child (n = 4)                | Being able to do FA with the child (n = 4)         | Distance of the FA environment (n = 3) | Having an FA partner (n = 4)              | Being able to do it with the child (n = 4)         | Being able to do it with the child (n = 4)         |
|       | Intensity of housework (n = 4)                            | Being able to do FA with the child (n = 4)         | Distance of the FA environment (n = 3) | Subsidization of the FA programme (n = 3) | Distance of the FA environment (n = 3)             | Improvement of physical appearance (n = 4)         |
|       | Anxiety of not being able to continue the routine (n = 3) | Receiving professional support (n = 4)             | Financial difficulties (n = 3)         | Doing FA in the (n = 4)                   | Receiving professional support (n = 4)             | Receiving professional support (n = 4)             |
|       | Continuation of health (n = 4)                            | Continuation of health (n = 4)                     |                                        | Continuation of health (n = 4)            | Lack of support for children (n = 4)               | Supporting motivation (n = 4)                      |
|       | Doing able to continue children in old age (n = 4)        | Doing able to continue children in old age (n = 4) |                                        |                                           | Doing able to continue children in old age (n = 4) | Doing able to continue children in old age (n = 4) |

C\*, Contemplation

## Discussion

When the findings were analysed in general, it was the common conclusion of the barriers and facilitators that parents need a solution where they can do exercise with their children's care and daily responsibilities. When preparing a programme based on the transtheoretical model (Prochaska & Velicer, 1997), the programme should be customised taking into account the barriers and facilitators that the participants have.

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## The effects of frequency of Subthalamic Nucleus Deep Brain Stimulation on postural control in Parkinson's Disease

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## Introduction

Subthalamic nucleus deep brain stimulation (STN-DBS) is a standard treatment for motor complications in Parkinson's disease (PD). Its impact on axial symptoms is still not fully understood. It is suggested that different frequencies of STN-DBS should attenuate postural control differently in this manner, with a lower frequency likely providing more effective residuals to stabilize posture. Therefore, this study hypothesizes that different frequencies of STN-DBS will have various effects on postural control in individuals with PD.

## Research Question

This study aimed to quantitatively evaluate the effect of frequency changes within the therapeutic window on postural control performances of individuals with PD who underwent bilateral STN-DBS.

## Methods

Postural control was assessed using Computerized Dynamic Posturography with randomized DBS frequency parameters, low (60 Hz), high (130 Hz), and very high (180 Hz), across six sensory organization test (SOT) conditions (NeuroCom International, Inc., Clackamas, OR, USA). In the SOT test, participants were asked to stand still on a platform on both feet and maintain their posture for 20 seconds and they were evaluated on the force platform under six conditions (Figure 1). Twenty PD participants with a mean age of  $61.2 \pm 10.1$  years were included.

## Results

There were no differences in equilibrium scores of SOT conditions between 60, 130, and 180 Hz frequencies ( $p > 0.05$ ), except the SOT6 score ( $p = 0.003$ ), where 60 Hz showed better equilibrium performance in SOT6, indicating an advantage in postural control when visual cues are disturbed (Figure 2).

## Discussion

Results from the current study indicated that though low-frequency stimulation (60 Hz) produced similar results compared to high- (130 Hz) and very-high-frequency (180 Hz) stimulation for postural control, the low frequency may be helpful for individuals with PD who are over-reliant on visual cues and ineffective use of somatosensory and vestibular cues. In addition to the sensory organization test, various motor control tests should be applied to verify low-frequency stimulation's short- and long-term effects on the target group. Importantly, this finding underscores the possibility that low-frequency stimulation may have some clinical utility in patients who have become visually dependent yet show a diminished ability to flexibly use somatosensory and vestibular inputs for balance control and modulation. This reliance on somatosensory function may compromise their overall posture control and may render these patients suitable targets for low-frequency stimulation as an intervention to improve balance. A tailored approach to the DBS frequency setting could optimize postural stability and reduce fall risk in these patients. Future research is needed to explore these mechanisms to enhance therapeutic strategies.

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## Effectiveness of Model-Based Digital Intervention for Physical Activity Behavior of Parents with and without Children with Special Needs

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## Introduction

Physical activity behavior change program based on the trans-theoretical model (TTM) to reduce sedentariness have gained more importance especially after COVID-19.

## Research Question

The aim of this study was to examine the effect of a 12-week TTM-based program on physical activity variables to reduce sedentary behaviors of parents with and without children with special needs.

## Methods

A total of 69 parents, 23 parents each group (experimental, placebo and control) participated in this study. Physical activity behaviors were measured by accelerometer (ActiGraph GTX9, USA). The experimental group received the TTM-based intervention program as digital content (video and jpeg) via WhatsApp, the placebo group received a digital physical activity guide via WhatsApp, and the control group did not receive any intervention. The intervention programs were maintained for 12 weeks. A 3×3 mixed-design MANOVA was conducted to test the effects of the intervention.

## Results

Although the time × group interaction was not statistically significant, it approached significance, indicating potential variability in how participants responded to the two different types of programs. Univariate analyses showed significant decreases over time in step count, MET value, light and moderate physical activity, and increases in sedentary time and bout duration across the entire sample. However, when pre-test and post-test mean values of each group were examined descriptively, a more favorable pattern was observed in the experimental group (Table 1). For example, the decline in step count was less pronounced in the experimental group (from 89,931 to 84,922 steps/day) compared to the placebo group (from 91,372 to 77,946 steps/day). Similarly, the experimental group showed a smaller reduction in MET values (from 1.579 to 1.551) than the placebo group (from 1.626 to 1.530). Light physical activity decreased from 2238.0 to 2146.9 minutes/day in the experimental group, whereas it dropped more sharply from 2101.8 to 1890.4 in the placebo group. The same pattern was observed in moderate physical activity, with the experimental group showing a reduction from 1516.1 to 1422.4 minutes/day, compared to a reduction from 1482.0 to 1320.5 minutes/day in the placebo group. In terms of sedentary behavior, although all groups experienced an increase, the experimental group had a more modest rise. Total sedentary time increased from 5200.2 to 5527.1 minutes/day in the experimental group, while it rose from 5212.2 to 6012.1 minutes/day in the placebo group.

## Discussion

These findings suggest that while seasonal or contextual factors may have contributed to an overall decline in physical activity and an increase in sedentary behavior, the TTM based intervention may have helped to buffer against these effects. Participants in the experimental group exhibited a more stable trajectory across most key variables, indicating that theory-driven, tailored interventions may offer protective benefits in populations facing elevated caregiving demands.

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## Active lidar-assisted spatial sensing: performance analysis of a bio-inspired system for navigation and obstacle detection

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## Introduction

Echolocation enables individuals with impaired or no vision to perceive spatial features by analyzing sound reflections [1]. However, mastering this skill requires extensive training, and its effectiveness is influenced by environmental factors. Additionally, users must continuously generate sounds while interpreting auditory feedback, which can be cognitively demanding [2][3]. Previously, we introduced a proof-of-concept light detection and ranging (LIDAR) assisted spatial sensing (LASS) system that converts LIDAR-detected spatial data into stereo sound with varying pitches. The relative pitch conveys angular orientation and horizontal distance, enhancing spatial perception for visually impaired users [4]. However, LASS operates passively, restricting user control over scanned regions. To overcome this limitation, we present the bio-inspired active LIDAR assisted spatial sensing (ALASS) system, which enables users to dynamically select specific regions for LIDAR scanning. This active control reduces scanning delays and allows

users to adopt sensing strategies similar to biological echolocation, particularly in bat navigation [5][6]. By allowing users to prioritize spatial sensing, ALASS enhances spatial awareness, responsiveness, and overall navigation efficiency.

## Research Question

This study addresses questions regarding both system efficacy and human-system interaction:

The Effectiveness of ALASS in assisting navigation.

How do users develop spatial cognition strategies through active scanning control, particularly in complex navigation tasks?

What learning curve patterns emerge when transitioning from passive to active sensing modalities?

## Methods

Three experiments assessed ALASS's capacity to aid blindfolded users in developing spatial awareness, navigating structures, and identifying/approaching objects.

1. Hallway Navigation: Blindfolded participants traversed a hallway path with turns using ALASS auditory cues. They completed ten trials from randomized starting points to a fixed finish line. Completion time was recorded to assess navigation efficiency.

2. Wheelchair Navigation: Participants operated an electric wheelchair (1 m/s) blindfolded, controlling ALASS with one hand, eliminating tactile feedback from the feet. Distance and time were recorded over ten trials.

3. Object Approach: In open spaces, participants located and walked toward targets (variable distances/angles) using ALASS stereo pitch cues until physical contact. Ten trials recorded approach time.

## Results

ALASS improved spatial perception efficiency by enabling active region selection for LIDAR scanning. Participants successfully navigated complex environments and located objects using controlled spatial scanning. Performance was evaluated through three tasks: foot-based hallway navigation, wheelchair navigation, and open-space object approach. Participants completed all tasks collision-free with minimal training. Adaptive strategies emerged, such as preferential use of specific regions for maintaining spatial awareness and navigating turns, highlighting effective utilization of active control features.

## Discussion

Future research could integrate Brain-Computer Interface (BCI) technology with ALASS to enable mental selection of scanning regions without physical controllers.

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## Which joint drives ball velocity during throw-in in football? A kinematic analysis.

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## Introduction

The techniques used to score in football are crucial. Like shooting and corner kicks, throw-ins are fundamental. The farther a player throws the ball, the larger the area for teammates to receive it, increasing scoring opportunities (1). To perform a long throw-in, a player must release the ball at high velocity and an optimal angle (2,3). Prior research examined the biomechanics of football throw-ins; however, the link between joint biomechanics and ball release velocity remains unexplored (4,5). Thus, our study aims to investigate the relationship between upper and lower extremity variables and release velocity.

## Research Question

During a standing throw-in, which joint's kinematic data shows the most significant correlation with the ball's release velocity in football?

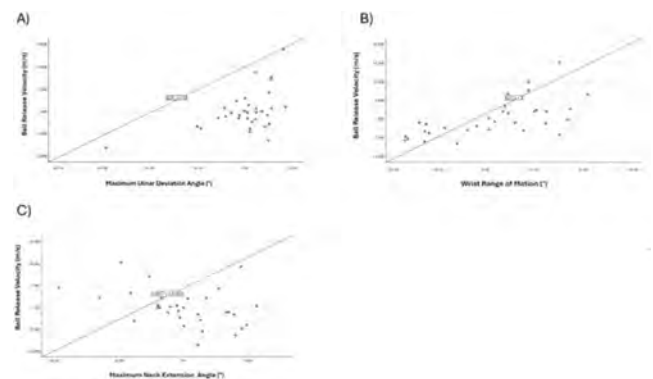
## Methods

Thirteen adolescent male football players (mean  $\pm$  SD: age  $15 \pm 0.5$  y.o., height  $174.76 \pm 5.91$  cm, weight  $62.23 \pm 4.7$  kg) participated in this pilot study. Participants performed a standing throw-in with a step onto a force platform, following official football rules. Three high-performance trials, compliant with game rules, were recorded under laboratory conditions. Full-body kinematics were captured using a 3D motion analysis system (VICON Plug-in Gait Model) (6). Each throw was also recorded by a camera (iPhone XR) positioned to the participant's right. Spearman's coefficients were used for correlation ( $p < 0.01$ ). The Spearman correlation coefficient ( $\rho$ ) is considered moderate between 0.4–0.6, strong between 0.6–0.8, and very strong between 0.8–1 (7).

## Results

This study analyzed the kinematic data of upper and lower extremities during a standing throw-in. Among 13 adolescent male soccer players, the mean ball velocity during the throw-in was  $10.83 \pm 6.02$  m/s. A strong positive correlation was found between ball release speed and both maximum ulnar deviation angle ( $\rho = 0.784$ ,  $p < 0.01$ ) and wrist range of motion ( $\rho = 0.784$ ,  $p < 0.01$ ) (Fig 1; A, B). Conversely, a strong negative correlation was observed between ball speed and maximum neck extension angle ( $\rho = -0.780$ ,  $p < 0.01$ ) (Fig 1; C).

Fig. 1: Partial Regression Plot, Dependent Variable: Ball Release Velocity



## Discussion

This study demonstrated that the most correlated joint kinematics with ball-release velocity were coronal plane wrist and sagittal plane neck. Maximum ulnar deviation and ROM and peak neck extension had strongly correlated with ball velocity during throw-in. These findings indicated that an increase in wrist ulnar deviation and decreased peak neck extension angles during throw-in enhances ball release speed may provide a better chance to create a score in football which should be included in training. On the other hand, increasing velocity might create compensations on different joints which should also be investigated by higher number of volunteers and by more sophisticated ball-release velocity measurement tools not only for definition purposes but also for preventing the injury risks on football players.

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## Foot posture and gait parameters in individuals with bruxism: A preliminary study

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## Introduction

Considering the fascial chain, it is well-established that muscle groups move in relation to one another. A change in one muscle group can, therefore, primarily shorten or lengthen the muscles in close proximity, and subsequently lead to alterations in other muscle chains located in distant regions that share the same fascial connections, potentially resulting in various pathologies (1). One of the most significant risk factors is bruxism, commonly observed in TMD patients, in which the anterior tilt of the head frequently occurs, accompanied by hyperactivity in the head and neck muscles. This exemplifies the close-range relationship (2). Furthermore, studies showing that individuals with a concave plantar arch experience a secondary reduction in masseter and temporalis muscle activity provide an example of the distant-region relationship (3). Although there are studies in the literature addressing the relationship between TMD and distant site, there is no study focusing directly on the bruxism population.

## Research Question

What are the differences in foot posture and gait parameters in individuals with and without bruxism?

## Methods

Ten participants, including 6 people with bruxism (Group 1) and 4 people without bruxism (Group 2), were included in this study.

The "Bruxism Questionnaire (4)" was applied to the participants and individuals who responded positively to at least two questions from the questionnaire were included in Group 1, and the others were included in Group 2. The "Foot Posture Index (5)" was used to evaluate the foot posture of the individuals distributed to the groups once and the "Kineovea software program" was used to evaluate the spatiotemporal parameters (right step length, left step length and speed) of gait (6).

## Results

There was no significant statistically difference between the height ( $p=0.914$ ), weight ( $p=0.610$ ), BMI ( $p=0.610$ ), gender ( $p=0.667$ ) and educational status ( $p=1.000$ ) of the participants in Group 1 and Group 2. Similarly, there is no statistically significant difference between foot posture and gait parameters in Group 1 and Group 2 ( $p>0.05$ ) (Table 1).

**Table 1. Comparison of foot posture and gait parameters of Group 1 and Group 2**

|                         | Group 1 (n=6) | Group 2 (n=4) | p     |
|-------------------------|---------------|---------------|-------|
|                         | Mean ± SD     | Mean ± SD     |       |
| Foot Posture Index      | 5.17±3.12     | 2.25±1.18     | 0.352 |
| Right step length (cm.) | 63.72±7.72    | 69.85±6.76    | 0.352 |
| Left step length (cm.)  | 60.17±8.19    | 66.96±1.72    | 0.114 |
| Speed (m/sec.)          | 1.01±0.25     | 0.77±0.05     | 0.067 |

cm., centimeter; m/sec., meter/second; SD, standard deviation

## Discussion

Although the results showed that individuals with bruxism had greater pronation, shorter right and left step lengths, and lower walking speeds than those without bruxism, these results were not statistically significant. This is a preliminary study, we believe that the difference will become statistically significant once the sample size is completed.

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## Improving Pelvic Orientation Estimation for Real-Time Frontal-View Motion Capture systems

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### Introduction

Gold-standard Motion Capture (MoCap) systems rely on multi-camera setups to track infrared-reflective markers with high accuracy, but are costly and impractical for routine clinical use. Recent advances in time-of-flight (ToF) depth cameras and computer vision have enabled low-cost, portable alternatives that track reflective markers with a minimal setup [1-5]. In this context, we developed HoloMoCap [1], a standalone marker-based MoCap application for HoloLens 2 that captures patients' motion while the headset is worn by the clinician.

Frontal-view systems like HoloMoCap cannot capture posterior landmarks, making pelvic orientation estimation challenging. A common approach, adopted by markerless systems, is the SternumAsis method, which estimates pelvic tilt using ASIS and sternum landmarks. However, this approach leads to large errors during torso-tilting movements. In this study, we evaluate an alternative method—IlioAsis—that uses ASIS and iliac crest markers.

### Research Question

- 1) How stable and accurate is the SternumAsis method in estimating pelvic orientation?
- 2) Does IlioAsis offer better stability and accuracy under dynamic conditions?

### Methods

Motion data from 15 healthy adults (8 male, 7 female; weight =  $67 \pm 9$  Kg, height =  $173 \pm 9$  cm), were collected using a Vicon system. Participants performed six exercises: treadmill walking, Sit-to-Stand (STS), 30 Second Sit to Stand (STS30s), and dynamic right and left hip abductions.

Pelvic orientation was estimated using both methods and compared against the Plug-in Gait (PiG) reference, with accuracy quantified by the angular deviation ( $\alpha$ ) between each method's vertical axis and that of PiG (Figure 1). The IlioAsis configuration defines a pelvic plane that is tilted relative to PiG. However, this tilt was consistent across participants and tasks, allowing a fixed offset correction—computed from the average  $\alpha$ —to be applied for alignment.

Intra-exercise standard deviation of  $\alpha$  ( $SD\alpha$ ) was used to evaluate configuration stability. To determine which method yields lower alpha, either paired t-tests or Wilcoxon signed-rank tests were applied, depending on Shapiro-Wilk normality results.

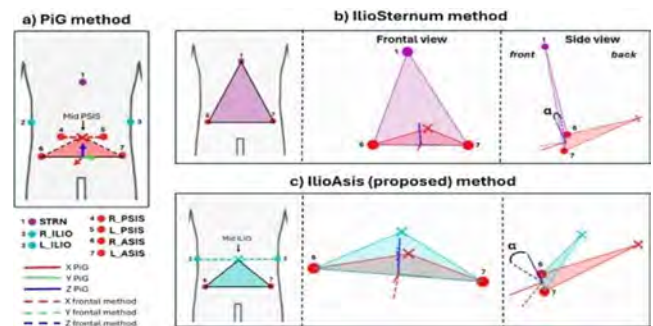


Figure 1: Comparison of pelvic orientation methods, showing marker configurations, coordinate systems, and angular deviation  $\alpha$  between methods.

### Results

The SternumAsis method showed limited stability and accuracy, particularly during torso-tilting exercises.  $SD\alpha$  was  $<3^\circ$  for treadmill walking and hip abductions, but exceeded  $5^\circ$  during STS and peaked at  $10^\circ$  in STS30s. Average  $\alpha$  across exercises was  $10^\circ$ , with the largest deviation observed in male STS30s trials ( $20^\circ$ ).

In contrast, IlioAsis showed higher stability, with  $SD\alpha < 3^\circ$  across all exercises. After applying a fixed correction, IlioAsis achieved significantly lower orientation errors than SternumAsis in all exercises ( $p < 0.05$ ), with average  $\alpha$  values  $<5^\circ$  during walking and hip abductions, and  $<6.5^\circ$  during STS and STS30s.

### Discussion

This study presents a simple yet effective approach for estimating pelvic orientation using only frontal markers. The proposed IlioAsis approach outperformed the conventional SternumAsis approach.

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## The impact of amputation level on gait patterns, mobility, and daily life: a comparison between transtibial and transfemoral amputees

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### Introduction

In Germany, approximately 60,000 lower limb amputations are performed annually. To assess the rehabilitation process, prosthesis acceptance, health-related quality of life and functional deficits, the German Exo-Prosthesis Registry (AMP Register) was established in 2013. This registry plays a key role in evaluating care quality, identifying structural deficiencies and enabling evidence-based treatment adaptations. Data are collected via standardized questionnaires covering clinical parameters such as amputation level and cause, mobility, pain, stump condition, prosthesis type, and overall quality of life. Additionally, functional tests and 3D gait analyses provide objective mobility assessments.

### Research Question

This study investigates gait kinematics in prosthetic users, correlating them with factors such as quality of life and stump condition. The main goal is to identify typical gait patterns associated with different amputation levels, focusing on transtibial and transfemoral amputees. These findings aim to support improved therapy planning, prosthesis fitting, and rehabilitation.

### Methods

Retrospectively, patients with above- and below-knee amputation were included having received a 3D-gait analysis. A total number of 258 amputees was analyzed, consisting of 127 transtibial amputees (109 men, 18 women, median age 48 years) and 131 transfemoral amputees (109 men, 22 women, median age 47 years). The 3D gait analysis was conducted using Vicon Nexus, capturing kinematic, kinetic, and spatiotemporal parameters. Patient data were extracted from the registry and analyzed with SPSS. The data were tested for normality, and due to some non-normal distributions, the Mann-Whitney U test was applied to assess statistical significance for all variables.

### Results

Both transtibial (G1) and transfemoral (G2) amputees face challenges due to the loss of the ankle joint, affecting roll-off during stance phase. Transtibial amputees, however, demonstrate a larger range of motion in dorsiflexion/plantarflexion (19.9 °; 17.5 °). Additionally, transfemoral amputees exhibit reduced knee flexion during loading and mid-stance (9.7 °; 0.9 °), as well as increased pelvic tilt (3.3 °; 7.1 °). They also show wider step widths (10.2; 10.7 [cm]), slower walking speeds (1.3; 1.2 [m/s]), and lower cadence (105.1; 103.1 [steps/min]).

### Discussion

The preserved lower leg musculature in transtibial amputees enables greater dorsiflexion/plantarflexion, facilitating better compensation with a well-fitted prosthesis. Moreover, many artificial knee joints, except for microprocessor-controlled ones, limit flexion to maintain stability, restricting knee movement in transfemoral amputees. The increased pelvic tilt compensates for the loss of ankle and knee function, aiding leg shortening and improving balance. The broader step width, slower gait, and reduced cadence observed in transfemoral amputees suggest decreased gait stability. [1] In conclusion, this study highlights the significant impact of amputation level on gait patterns, compensatory movement strategies, and mobility. The identified biomechanical differences have direct implications for daily function and rehabilitation, emphasizing the need for targeted therapy and optimized rehabilitation planning.

Table 1: Spatio-temporal parameters for transtibial (G1) and transfemoral (G2) amputees

|                                                             | G1     |              |              | G2     |              |              | p-value |
|-------------------------------------------------------------|--------|--------------|--------------|--------|--------------|--------------|---------|
|                                                             | Median | 25% Quartile | 75% Quartile | Median | 25% Quartile | 75% Quartile |         |
| Cadence [steps/min]                                         | 105.1  | 98.2         | 111.7        | 103.1  | 96.6         | 108.0        | 0.021   |
| DorsiPlanFlex (ROM)                                         | 19.9   | 15.76        | 26.9         | 17.5   | 13.9         | 21.6         | 0.001   |
| KneeFlexEx (Maximum during Loading Response and Mid Stance) | 9.7    | 4.0          | 15.1         | 0.93   | -4.2         | 4.7          | 0.001   |
| PelvicTilt (ROM)                                            | 3.4    | 2.7          | 4.2          | 7.1    | 5.4          | 9.7          | 0.001   |
| Speed [m/s]                                                 | 1.25   | 1.15         | 1.36         | 1.21   | 1.04         | 1.31         | 0.003   |
| StepWidth [cm]                                              | 10.2   | 8.7          | 11.9         | 13.5   | 10.7         | 16.45        | 0.001   |

ROM = Range of motion

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### Postural balance while standing at virtual heights in children with and without cerebral palsy

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### Introduction

Postural balance is commonly measured by recording the center of pressure (CoP) trajectories via force plates. Fear of falling can impact postural balance, particularly in populations at risk of falling such as children with cerebral palsy (CP) [1]. Thus, virtual height exposure may be used to assess threat-related changes in postural balance.

## Research Question

Does standing in virtual reality (VR) at a certain height affect postural balance in children with CP compared to those with typical development (TD)?

## Methods

Included were children with spastic CP aged 7-18 years, classified GMFCS I/II, without orthopedic surgeries (<1year) and botulinum toxin A treatment (<6months), and age-matched children with TD. Participants were immersed in VR using a head-mounted display and instructed to stand still for 40s on a virtual plank, both at ground level (VR0) and at 10m above ground (VR10). Conditions were applied in a randomized order. Recordings displaying movements that deviated from maintaining a quiet standing posture, evaluated by two reviewers, were excluded from the analysis. As a result, 11 children with CP (out of 22) and 20 children with TD (out of 24) were analyzed. Postural balance was quantified by considering the middle 20s of each trial, using (1) the 95% CoP prediction ellipse area, (2) the average CoP velocity, and (3) the CoP amplitude in anteroposterior (CoP\_AP) and mediolateral (CoP\_ML) directions. The Kruskal-Wallis test was applied to compare groups and conditions.

## Results

No statistically significant differences were found between groups and conditions. The results showed that children with CP tended to increase ellipse area, velocity and CoP\_AP from VR0 to VR10, whereas children with TD exhibited no changes (Figure 1). At VR10, children with CP tended to increase ellipse area, velocity and CoP\_AP compared to those with TD, while the parameters for VR0 were similar across both groups.

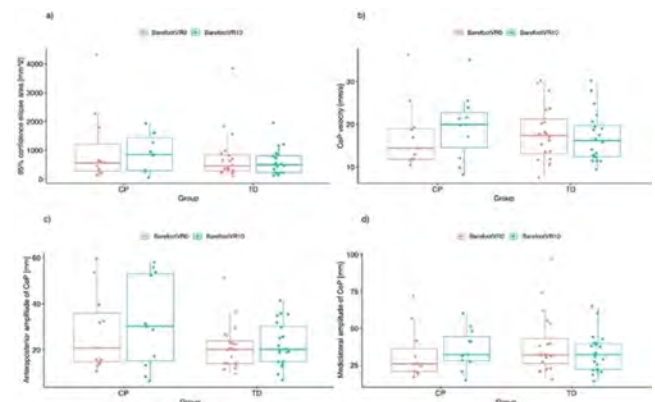


Figure 1: Comparisons for center of pressure (CoP) 95% ellipse area (a), CoP velocity (b), and CoP amplitude in anteroposterior (c) and mediolateral (d) directions. CP: cerebral palsy, TD: typical development, VR0: standing on a virtual plank at ground level, VR10: standing on a virtual plank at 10m above ground.

## Discussion

Our findings indicate that children with CP demonstrated postural adaptations similar to those of children with TD, suggesting they were able to maintain balance even in more challenging conditions. It also appears that VR10 may not have imposed as much of a challenge as anticipated. However, the small sample size may have limited statistical power. This study highlights the challenges of recording standing postural sway data in VR with young children, particularly due to fidgeting and the excitement of being immersed in VR. Our findings offer valuable insights that can guide future research aimed at addressing this issue.

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