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# EFFECTS OF MODEL PERSONALISATION ON THE STUDY OF CROUCH GAIT BIOMECHANICS

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## 1 INTRODUCTION

Children with cerebral palsy (CP) often exhibit crouch gait (CG), a severe pathological gait pattern [1]. Musculoskeletal (MSK) modelling is used to study CG, typically relying on generic models based on healthy adults, overlooking individual bone deformities caused by CP [2].
Subject-specific MSK models have been developed to address this limitation, with varying levels of detail [3]. OpenSim's torsion tool offers a time-efficient and straightforward method for model personalisation by incorporating solely femoral and tibial torsional angles [2]. This study aims to investigate the impact of this tool on CG biomechanics in children with CP.

## 2 METHODS

This study focuses on a child with spastic diplegic CP, Gross Motor Function Classification System level III, CG, who underwent Single-Event Multilevel Surgery. Clinical examination to measure anthropometric features and ranges of motion, followed by gait analysis were performed one month pre-surgery (age: 13 years; height: 169.7 cm; mass: 60.6 Kg). Clinical gait analysis used marker-based motion capture with 14 Qualysis cameras at 100 Hz and CAST marker set [4]. The participant performed a static trial and walked along a 10 m corridor for dynamic trials. Ground reaction forces were recorded by two force plates at 1000 Hz.
Two models were studied: the generic Gait2392 and a model created with the torsion tool, considering femoral and tibial deformities. Since bone deformities were evaluated using clinical analysis and radiography, both subject to errors, two additional models were developed to incorporate the maximum estimated measurement error. Input values for each model are shown in Table 1. Muscle forces were attained for all models during the single support phase, and muscle strength requirements calculated to compare the percentage of maximum force across models.

Table 1 – Bone deformity angles used as inputs in the torsion tool to generate each model. Personalised values were measured in clinical examination and radiography, with limits of agreement (LoA) lower (- LoA) and upper (+ LoA) bounds sourced from literature. LoA values for femoral anteversion and tibial torsion are between computed tomography (CT) and clinical examination, and between CT and radiography for femoral neck-shaft angle.

| | | Femoral Anteversion Angle (º) | | Femoral Neck-Shaft Angle (º) | | Tibial Torsion Angle (º) | |
|---|---|---|---|---|---|---|---|
| | | Left | Right | Left | Right | Left | Right |
| Model Inputs | Personalised | 44 | 50 | 140 | 140 | 31 | 44 |
| | Personalised - LoA | 18.56 | 24.56 | 129.34 | 129.34 | 21.65 | 34.65 |
| | Personalised + LoA | 73.44 | 79.44 | 142.66 | 142.66 | 39.45 | 52.45 |

## 3 RESULTS

The left side of Figure 1 shows that the generic and personalised models require very similar muscle forces, with slight increases in required strength for the *gluteus maximus*, *gastrocnemius*, and *soleus*, for the personalised model during single support. On the right limb, the torsion model shows noticeable lower demand for the *rectus femoris* and higher for the *gluteus medius*, *vasti*, and *iliopsoas*.

Considering the LoA models, the one with the highest torsional angles overall shows the greatest muscle demand, while the lower bound LoA model closely resembles the generic model.

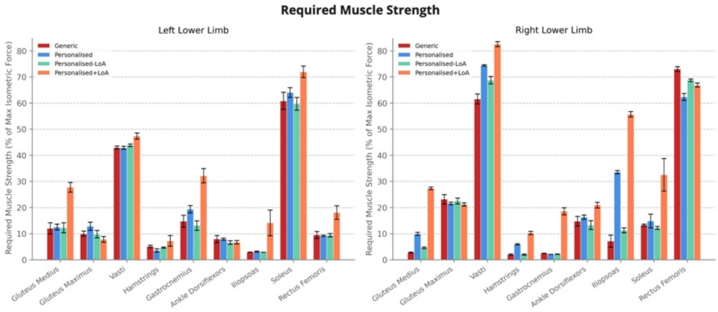


Figure 1 – Pre-surgery required strength for each muscle group in both limbs during the single support period, expressed as a percentage of the maximum isometric force. Error bars represent ±1 standard error.

## 4 DISCUSSION

This study evaluated the impact of torsional variations on muscle force-generating capacity. On the right lower limb, pronounced differences were observed between the generic and personalised models, particularly in the knee extensors and *iliopsoas*, suggesting that anatomical variations influence biomechanical outcomes. In contrast, muscle force predictions for the left limb were consistent across models, except for the one with a greater torsion angle. The generic model's results fall outside the interval defined by the two LoA models, reflecting actual biomechanical differences between the generic and personalised models rather than measurement errors.

Previous research determined that required knee extensor and ankle plantarflexor strength increased with crouch severity, whereas hip abductor strength decreased [1]. When examining the three torsion-based models, our results align with those for the *vasti* and ankle plantarflexors but differ for the *rectus femoris* and *gluteus medius*.

Given the relevance of MSK modelling in clinical applications such as surgical planning and prosthetic design, adopting an approach that improves accuracy could lead to more customized clinical strategies. The observed impact of torsion tool-based models on muscle force estimates highlights their potential and underlines the need for future research involving more participants to compare model accuracy, strengthen reliability, and support broader conclusions.

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