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Impact of scaling errors of the thigh and shank segments on musculoskeletal simulation results

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Highlights

- Scaling errors due to altered thigh length of ±15% were investigated
- Thigh and shank scaling errors influenced simulation results at all joints
- Joint kinematics and kinetics influenced by up to 9.4° and 0.15 Nm/kg respectively
- Maximum muscle and joint contact force errors of 46% and 72% bodyweight

Abstract

Background

Musculoskeletal simulations are widely used in the research community. The locations of surface markers are mostly used to scale a generic model to the participant's anthropometry. Marker-based scaling approaches include errors due to inaccuracies in marker placements.

Research question

How do scaling errors of the thigh and shank segments influence simulation results?

Methods

Motion capture data and magnetic resonance images from a child with cerebral palsy and a typically developing child were used to create a subject-specific reference model for each child. These reference models were modified to mimic scaling errors due to inaccurately placed lateral epicondyle markers, which are frequently used to scale the thigh and shank segments. The thigh length was altered in 1% steps from the original length and the shank length was accordingly adjusted to keep the total leg length constant. Thirty additional models were created, which included models with an altered thigh length of ±15%. Subsequently, musculoskeletal simulations with OpenSim were performed with all models. Joint kinematics, joint kinetics, muscle forces and joint contact forces (JCF) were compared between the reference and altered models.

Results

The investigated scaling error influenced joint kinematics and joint kinetics by up to 9.4° (hip flexion angle) and 0.15 Nm/kg (knee flexion moment), respectively. Maximum muscle and JCF differences of 46% (medial gastrocnemius) and 72% (hip JCF) bodyweight, respectively, were observed between the reference and altered models. Scaling errors mainly changed the magnitude but not the shape of most analyzed parameters. The influence of scaling errors on simulation results were similar in both participants.

Significance

Scaling errors of the thigh segment influence simulation results at all joints due to the global optimization approach used in musculoskeletal simulations. Our findings can be used to estimate potential errors due to marker-based scaling approaches in previous and future studies.

Keywords: musculoskeletal simulation, scaling, cerebral palsy, opensim

Introduction

Musculoskeletal simulations based on commercial (i.e. Anybody, [1]) and open-source software packages (e.g. OpenSim, [2]) are widely used in the research community [3]. Over the last decade, musculoskeletal simulations are increasingly used to answer clinical-relevant questions [4–7]. Furthermore, musculoskeletal simulations in combination with finite-element simulations enable to conduct multi-scale studies to estimate tissue stresses [8] or predict bone growth [9].

The first step in the musculoskeletal simulation workflow is to generate the subject-specific model. The gold standard for personalizing a musculoskeletal model is to create the model based on the segmentation of medical images [10–14]. This, however, requires the collection of medical images (i.e. magnetic resonance images or computer tomography), which is expensive and demanding on the participants, especially in children or pathological populations. Alternatively, biplane x-rays or 3D ultrasound measurements could be used to determine subject-specific anatomical landmarks for personalizing a model [15]. X-rays and computer tomography, however, are invasive and therefore are barely used in children without a clinical reason. Furthermore, creating a model from medical images is very time-consuming and requires a lot of expertise. Hence, this approach is not used in clinical practice and barely used in the research community.

In most cases, a generic musculoskeletal model is scaled to the anthropometry of the participant based on the location of surface markers on the model and the participant. Several marker-based scaling approaches have been proposed to adjust the segment lengths of the model [16,17]. However, all approaches include errors. Scaling is the first step of a musculoskeletal simulation workflow. Hence, scaling errors will impact on all further calculations.

The lateral femoral epicondyle marker is difficult to place and therefore the intra- and inter-examiner precision is only around 10 mm and 19 mm, respectively [18]. This marker, however, is frequently used to scale the thigh and shank segments. Inaccurately placed lateral epicondyle markers can lead to increased femur and decreased tibia length or vice versa. Furthermore, it is very common to scale the thigh segment based on the distance between the lateral femoral epicondyle marker and the anterior superior iliac spine marker. Additionally, to the marker position, this distance, however, depends on the hip flexion angle and therefore can lead to increased errors, especially in people who have difficulties to stand in the anatomical neutral position (e.g. many children with cerebral palsy (CP)). In children, the typically used linear scaling methods lead to errors up to 15% of femur length [16]. The purpose of this study was to quantify the impact of scaling errors of the thigh and shank on the musculoskeletal simulation results in children. A reference model from a typically developing (TD) child and a child with CP was modified to mimic scaling errors of the thigh segment. Joint kinematics, joint kinetics, muscle forces and joint contact forces of the lower limbs during walking were compared between the reference models and models with different amount of scaling errors. It was expected that scaling errors of the thigh and shank segments will influence simulation results at all joints due to the global optimization approach used in musculoskeletal simulations.

Methods

Magnetic resonance imaging data and motion capture data (marker trajectories and ground reaction forces) of a child with CP and one TD child were retrospectively analyzed for this study. The data was captured during a previous study [16] and ethics approval was obtained from the local ethics committee. These two male children were chosen to have the same age (both 8 years), nearly the same body mass (22 kg and 20 kg) and height (125 cm and 124 cm). Detailed information about the participants can be found in supplementary Table S1.

Musculoskeletal models

The torso body and associated muscles were removed from the generic 'gait2392' OpenSim model [2], which included 92 muscles and 23 degrees of freedom. Furthermore, the metatarsophalangeal joints were locked. Subsequently a reference model was created for every participant based on measurements from the MRI images. Pelvis width, height and depth as well as femur and tibia length were calculated from the MRI images as described in detail by Kainz et al. [16]. These 'gold standard' measures were used to scale the musculoskeletal model to create the reference model for each participant. The maximum isometric muscle forces were scaled by Equation 1 [19,20].

$$F_{scaled} = F_{generic} * \left(\frac{m_{scaled}}{m_{generic}}\right)^{\frac{2}{3}} \qquad (1)$$

The reference model was systematically modified to create models, which included different amounts of scaling errors. The thigh length was altered in 1% steps from the original length and the shank length was accordingly adjusted to keep the total leg length constant. Furthermore, the marker positions of the musculoskeletal model were adjusted to match the position of the markers of the 3DGA using a customized MATLAB script. Thirty additional models were created, which included models with an altered thigh length of ±15%. This error range was chosen based on the data of a previous study which evaluated the accuracy of marker-based scaling approaches reporting femur length scaling errors of approximately +/-10% in TD children and up to 12.4% in children with CP [16]. This error span includes errors due to misplaced surface markers and deviation in joint angles from the anatomical position (i.e. hip flexion angle will influence scaling if a pelvis and knee marker is used to scale the thigh segment).

Musculoskeletal simulations

For the musculoskeletal simulations three and four gait trials were used for the TD and the CP child, respectively. The motion capture data was based on the Plug-in-Gait marker set with additional clusters of three markers on each thigh and shank segment and an additional marker at the 5th metatarsal head of each foot. The reference models and the corresponding gait analysis data were used to run inverse kinematics followed by inverse dynamics, static optimization by minimizing the sum of squared muscle activations and joint reaction load analyzes [21] with MATLAB R2020a (Mathworks Inc., Natick, MA, USA) and OpenSim 4.1 [3]. Markers close to joint axes were excluded during inverse kinematics. The remaining markers were weighted equally as detailed in the supplementary Table S2. Afterwards, the same workflow was used to run the simulations for the 30 modified models of each participant.

Data analysis

Joint kinematics, joint kinetics, muscle forces, joint contact forces and joint contact force orientations of the lower limbs during walking were compared between the reference models and models with different amount of scaling errors. First, all results were time normalized to a full gait cycle from initial contact to the following initial contact of the same foot. Subsequently for every model and analyzed parameter, the mean waveform for all gait trials were calculated. Afterwards, the root-mean-square-error (RMSE) between the reference model and every single other model was computed. The orientation of the joint contact forces was defined as the mean value of the contact force during the whole gait cycle.

Results

All plots in figures 1 to 6 show the results for the left leg, results for the right leg can be found in the supplementary material. Detailed numeric results are available as a MATLAB file in the electronic appendix. In general, the RMSE due to scaling error were similar between the TD and CP participant (Figure 1).

Joint kinematics

Errors due to altered femur and tibia length mainly altered joint kinematics in sagittal and frontal plane while effects on transversal plane kinematics were negligible (Figure 2). Maximum absolute differences between the reference and modified models were 9.4° and 7.8° for the hip flexion angle in the TD and the CP child, respectively. All tracking errors of the inverse kinematic simulations were below the OpenSim's best practice recommendations (supplementary Table S3).

Joint kinetics

Scaling errors had a minor impact on joint moments (Figure 3). The largest differences were 0.15 Nm/kg and 0.12 Nm/kg (right leg) for the knee flexion moment of the TD and the CP child, respectively. All other joint moments showed errors lower than 0.08 Nm/kg for both participants.

Muscle forces

The influence of scaling errors on estimated muscle forces varied between muscles (Figure 4 and 5). In the TD child the largest error occurred at the rectus femoris muscle with a value of 34% of the bodyweight (Figure 5). In the child with CP, the biggest influence of the scaling error was observed in the medial gastrocnemius muscle with an error of 46% of the bodyweight.

Joint contact forces (JCF)

A decreased femur length and increased tibia length led to higher hip JCF and lower knee JCF in both participants (Figure 3). The errors in ankle JCF were small compared to the errors in hip and knee JCF.

Maximum differences in JCF of 72% bodyweight and 35% bodyweight were observed for the TD and the CP child, respectively. These values corresponded with an increase of 12% and 14% of the first peak of the hip JCF for the TD child and CP child, respectively.

JCF orientations

Scaling error had an impact on the JCF orientation in transversal plane while the differences in the frontal and sagittal plane were minor (Figure 6). For both participants, the largest errors were observed for ankle JCF orientations in the transverse plane.

Discussion

The purpose of this study was to investigate the impact of scaling errors on simulation results. In general, we found that scaling errors of the thigh segment influence all simulation results, which was in agreement with our assumption. Furthermore, our findings showed that the impact of scaling errors on simulation results is similar for pathological and typical gait.

Simulation results of the reference models were in agreement with previous studies. Joint kinematics and kinetics were comparable to Schwartz et al. [22] and muscle and JCF were in agreement with Modenese et al [13] and Bergmann et al [23]. The simulation results from our participant with CP were similar to previous studies in children with CP, which reported joint kinematics and kinetics [24], muscle forces [20] and JCF [21].

In most analyzed parameters, scaling errors led to a shift of the waveforms and did not change the shape or function of the parameter. This, however, was not the case for the hip adduction angle. Interestingly, increased femur length changed the hip movement strategy from hip adduction to abduction during terminal stance and pre-swing in the participant with CP.

Scaling errors mainly influenced joint moments at the knee joint. Femur and tibia length were altered contrary; therefore, the leg length was constant over all models. This barely affected the ankle and hip joint center locations; hence, the impact was small on these joint moments.

The influence of scaling errors on muscle forces varied between muscles. Due to changed joint kinematics the muscle moment arms were influenced. Hence, this led to altered muscle forces. Furthermore, because of the changed knee joint moment, some muscles had to generate a different force to provide the altered joint moment.

Maximum errors in JCF were observed at the hip joint. The used joint reaction analysis of OpenSim implements an inverse approach [21]. Hence, JCF were calculated from the ankle moving upwards. Therefore, altered ankle JCF influenced the knee JCF which, furthermore, affected the hip JCF. The cumulated error from the ankle to the hip JCF might explain why the maximum error was found at the hip joint.

JCF orientations, especially at the hip and knee joint, barely changed although the magnitude of the JCF were influenced by the scaling errors. Therefore, the magnitude changed in a similar way in all three anatomical directions. This is an important finding for multi-scale studies which use the JCF orientations as an input for further calculations (e.g. bone growth simulations [25]).

In a recent study the influence of a misplaced knee marker on joint kinematics was investigated using the Conventional Gait Model [26]. The Conventional Gait Model, however, uses a direct kinematics instead of an inverse kinematic approach to calculate joint angles and does not allow estimating muscle forces or JCF. Hence, to the best of the authors' knowledge, this is the first study, which investigated the impact of scaling errors on simulation results of musculoskeletal models in a systematic and comprehensive way.

Our error span included worst case scenarios due to misplaced markers and deviation in joint angles from the anatomical position. Scaling errors solely due to misplaced knee markers would likely result in a change in thigh length of less than 2cm [18], which would be approximately 7% of the thigh length of our participants.

Marker-based scaling approaches usually induce errors at every segment, which is scaled based on the location of surface markers. In (lean) children, however, the placement of pelvic, ankle and foot markers are relatively easy compared to the placement of the knee markers. Hence, we did not investigate the concurrent errors due to scaling inaccuracies of different segments.

Several model assumptions and simplifications influence musculoskeletal simulation results, whereas we only investigated the impact of thigh and shank scaling errors on the results. For example, we did not consider the age-specific bony geometry (e.g. femoral neck-shaft angle) in our participants because previous studies already quantified the impact of subject-specific geometry on simulation results in TD and CP children [9,14,27]. Furthermore, we did not consider soft tissue artifacts [28] and different optimization methods to estimate muscle forces [29,30] because we solely wanted to quantify the impact of scaling errors on simulation results independent of any confounding factors.

In summary, this is the first study which quantified the impact of scaling errors due to misplaced lateral epicondyle markers on musculoskeletal simulation results with OpenSim in a systematic and comprehensive way. Scaling errors mainly changed the magnitude but not the shape and function of most analyzed parameters. Our RMSE plots can be used to estimate potential errors due to marker-based scaling approaches in previous and future studies. Furthermore, the gained insights are relevant for multi-scale studies, which use musculoskeletal simulation results as an input for further simulations (e.g. hip JCF magnitude and orientation for femoral growth simulations [25]).

Conflict of interest

The authors declare no conflict of interest.

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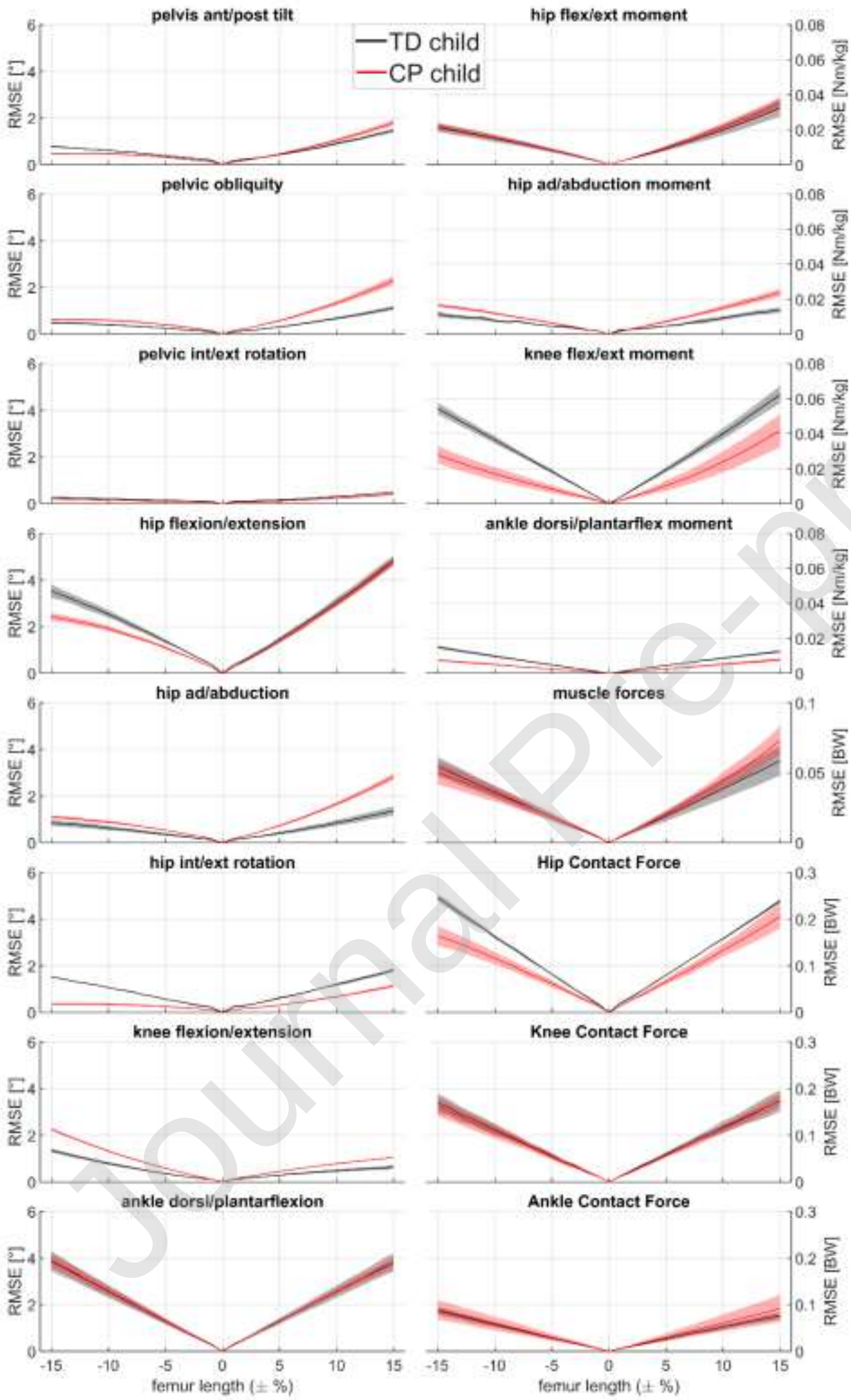

pelvis ant/post tilt
TD child
CP child
hip flex/ext moment
pelvic obliquity
hip ad/abduction moment
pelvic int/ext rotation
knee flex/ext moment
hip flexion/extension
ankle dorsi/plantarflex moment
hip ad/abduction
muscle forces
hip int/ext rotation
Hip Contact Force
knee flexion/extension
Knee Contact Force
ankle dorsi/plantarflexion
Ankle Contact Force
RMSE [°]
RMSE [Nm/kg]
RMSE [BW]
femur length (± %)

Figure 1: Comparison of the root mean square errors (RMSE) between the reference model and the adapted models. The RMSE of the muscle forces were calculated as the mean RMSEs over all observed muscles. The RMSEs for the CP child are shown in red and the RMSEs for the TD child are black.

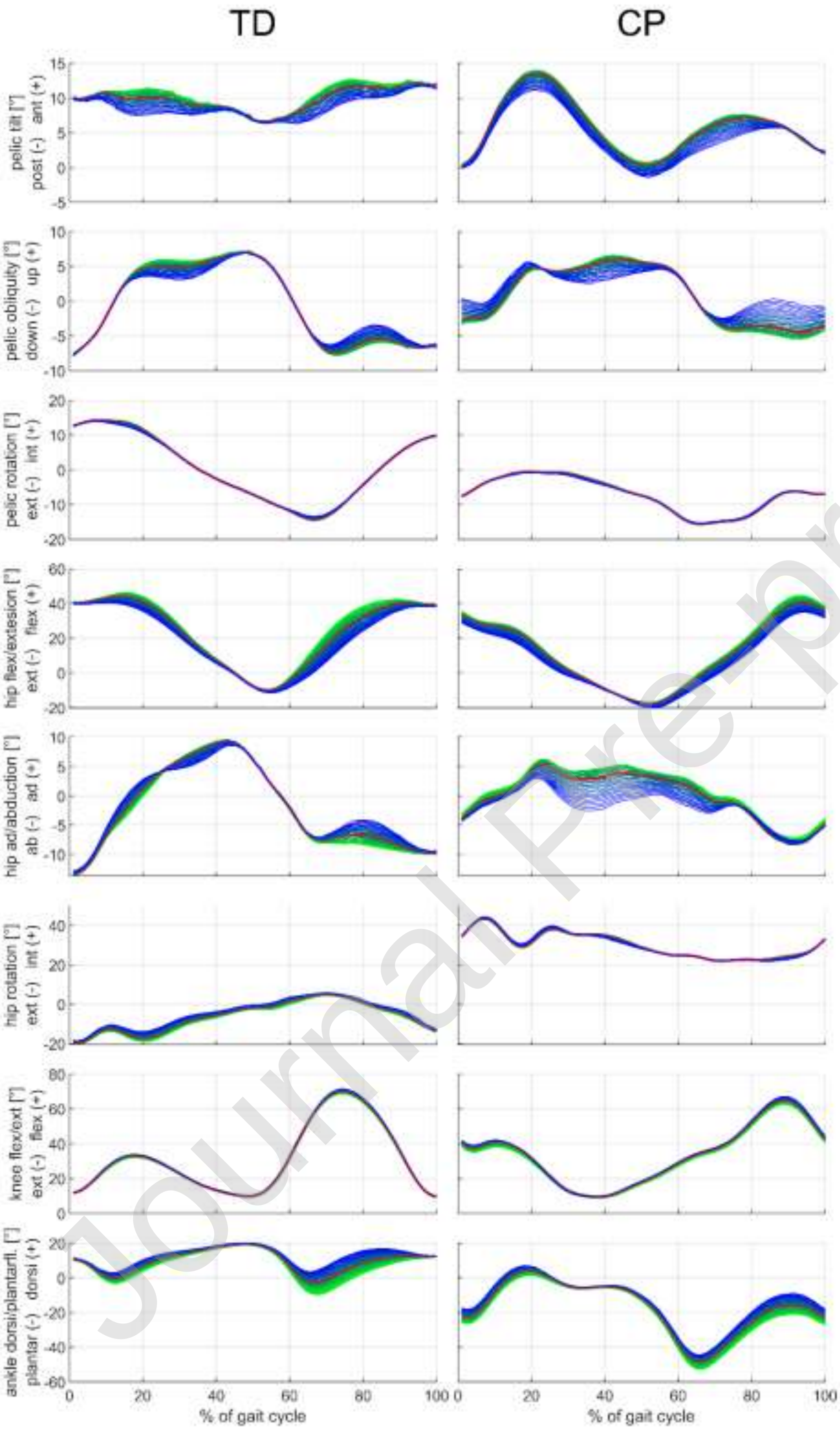

TD
CP
pelic tilt [°]
post (-) ant (+)
pelic obliquity [°]
down (-) up (+)
pelic rotation [°]
ext (-) int (+)
hip flex/extesion [°]
ext (-) flex (+)
hip ad/abduction [°]
ab (-) ad (+)
hip rotation [°]
ext (-) int (+)
knee flex/ext [°]
ext (-) flex (+)
ankle dorsi/plantarfl. [°]
plantar (-) dorsi (+)
% of gait cycle
% of gait cycle

Figure 2: Joint kinematics of the TD (left column) and the CP child (right column) obtained with the reference and modified models. The red line shows the mean kinematic waveforms of the reference model. Green lines represent the models with shorter femur length while blue lines show the models with increased femur length.

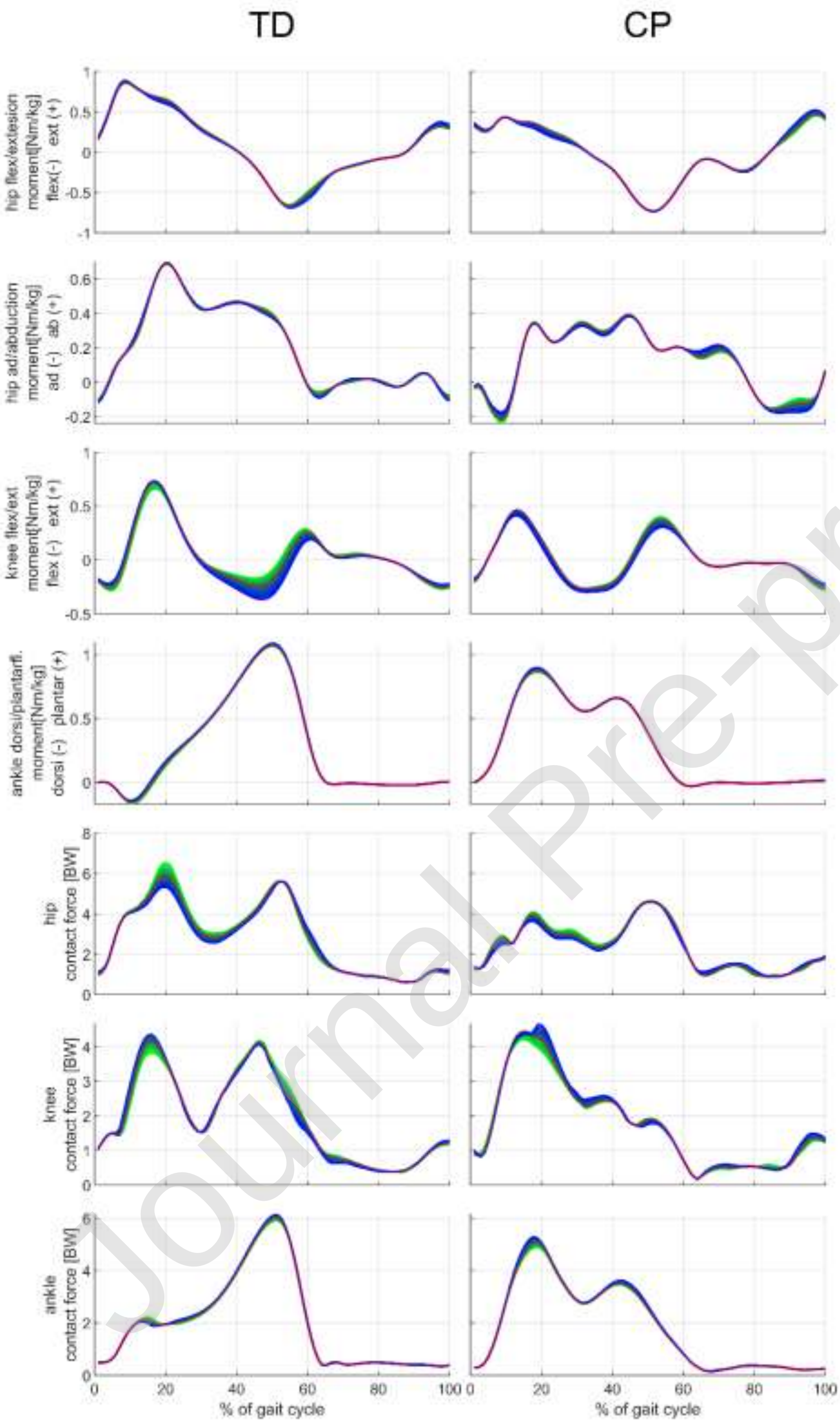

TD
CP
hip flex/extesion moment[Nm/kg] flex(-) ext (+)
hip ad/abduction moment[Nm/kg] ad (-) ab (+)
knee flex/ext moment[Nm/kg] flex (-) ext (+)
ankle dorsi/plantarfl. moment[Nm/kg] dorsi (-) plantar (+)
hip contact force [BW]
knee contact force [BW]
ankle contact force [BW]
% of gait cycle
% of gait cycle

Figure 3: Comparison of joint kinetics and joint contact forces of the TD (left column) and the CP child (right column) including all modified models. The red line shows the mean waveforms of the reference model. Green lines represent the models with shorter femur length while blue lines show the models with increased femur length.

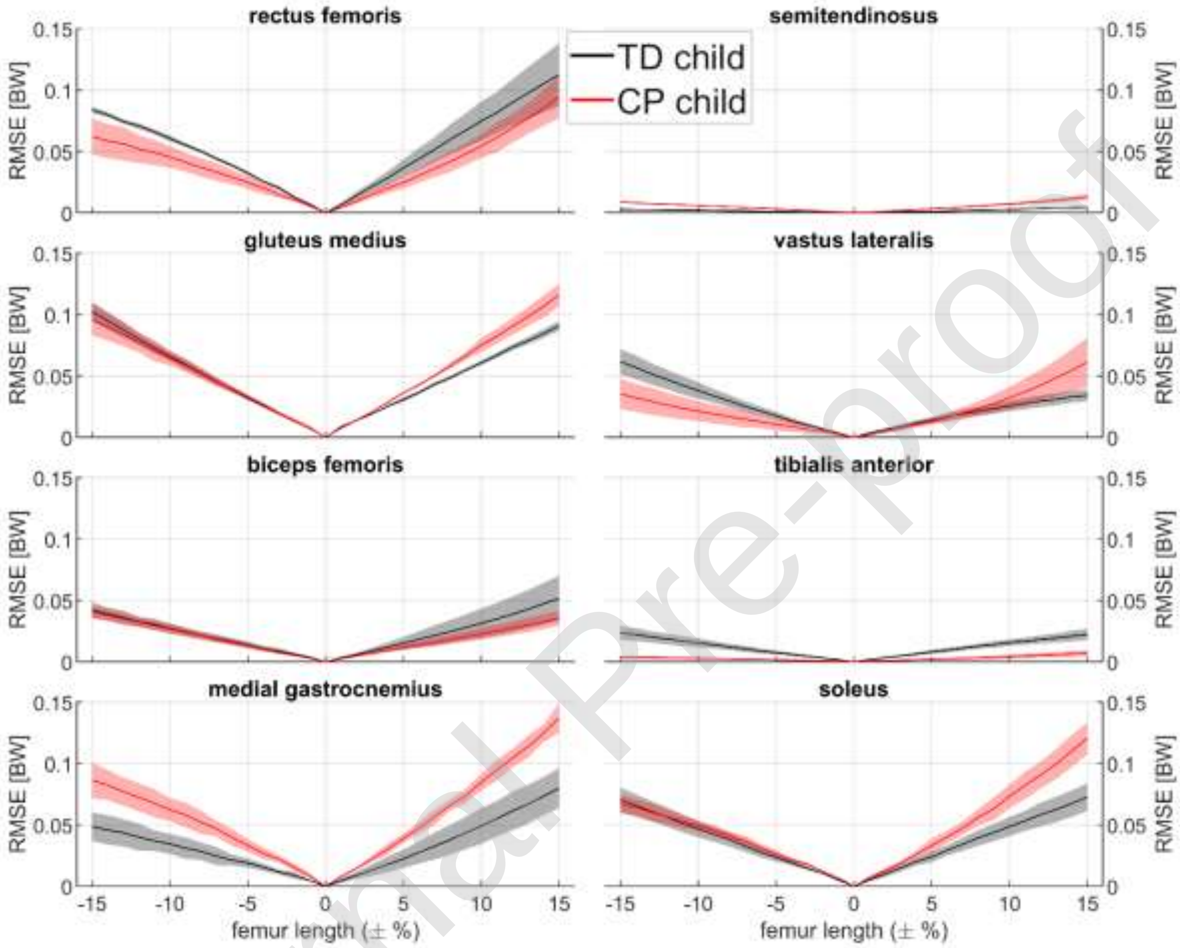


Figure 4: Comparison of the root mean square errors of an example of muscle forces between the reference model and the modified models. The RMSEs for the CP child are shown in red and the RMSEs for the TD child are black.

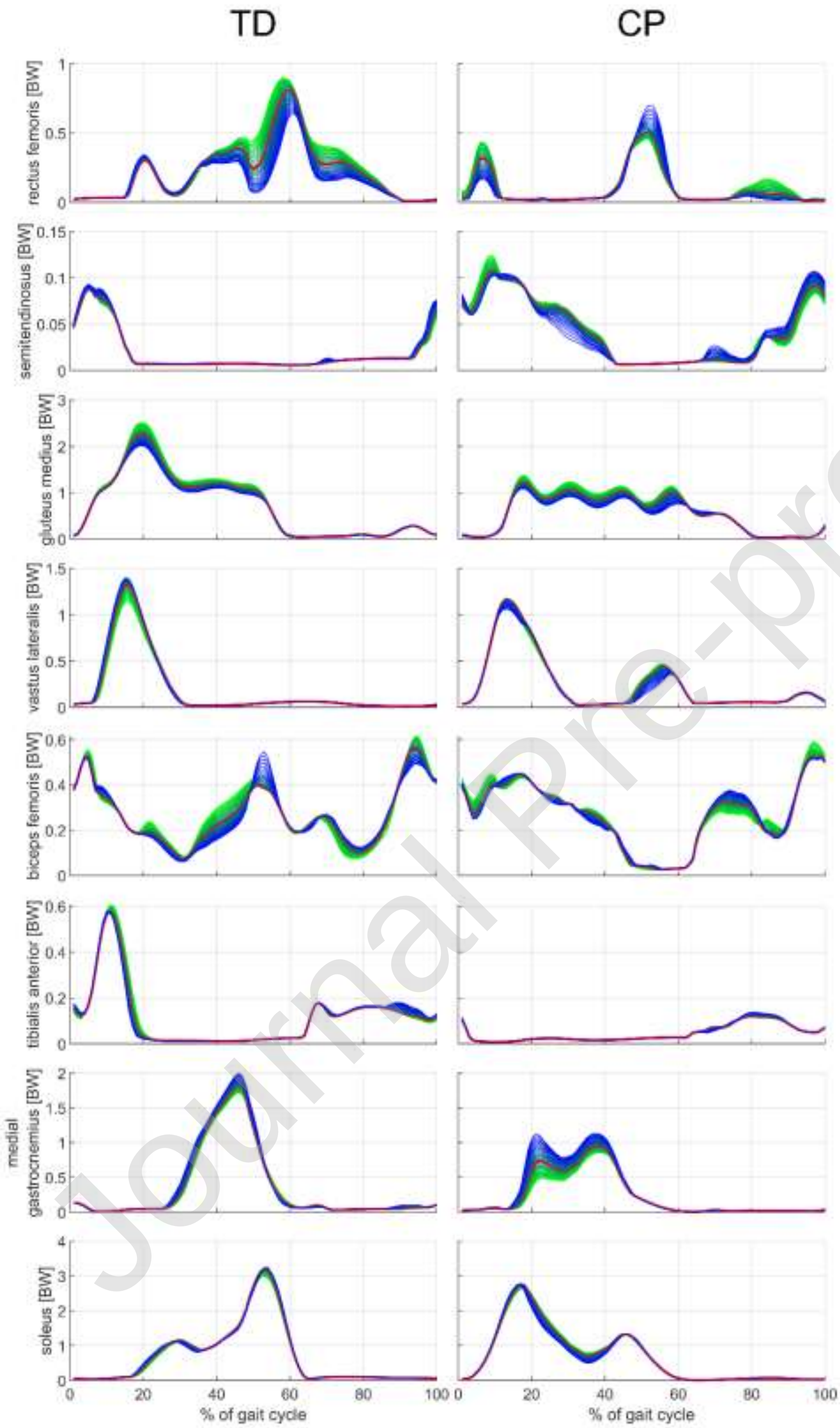
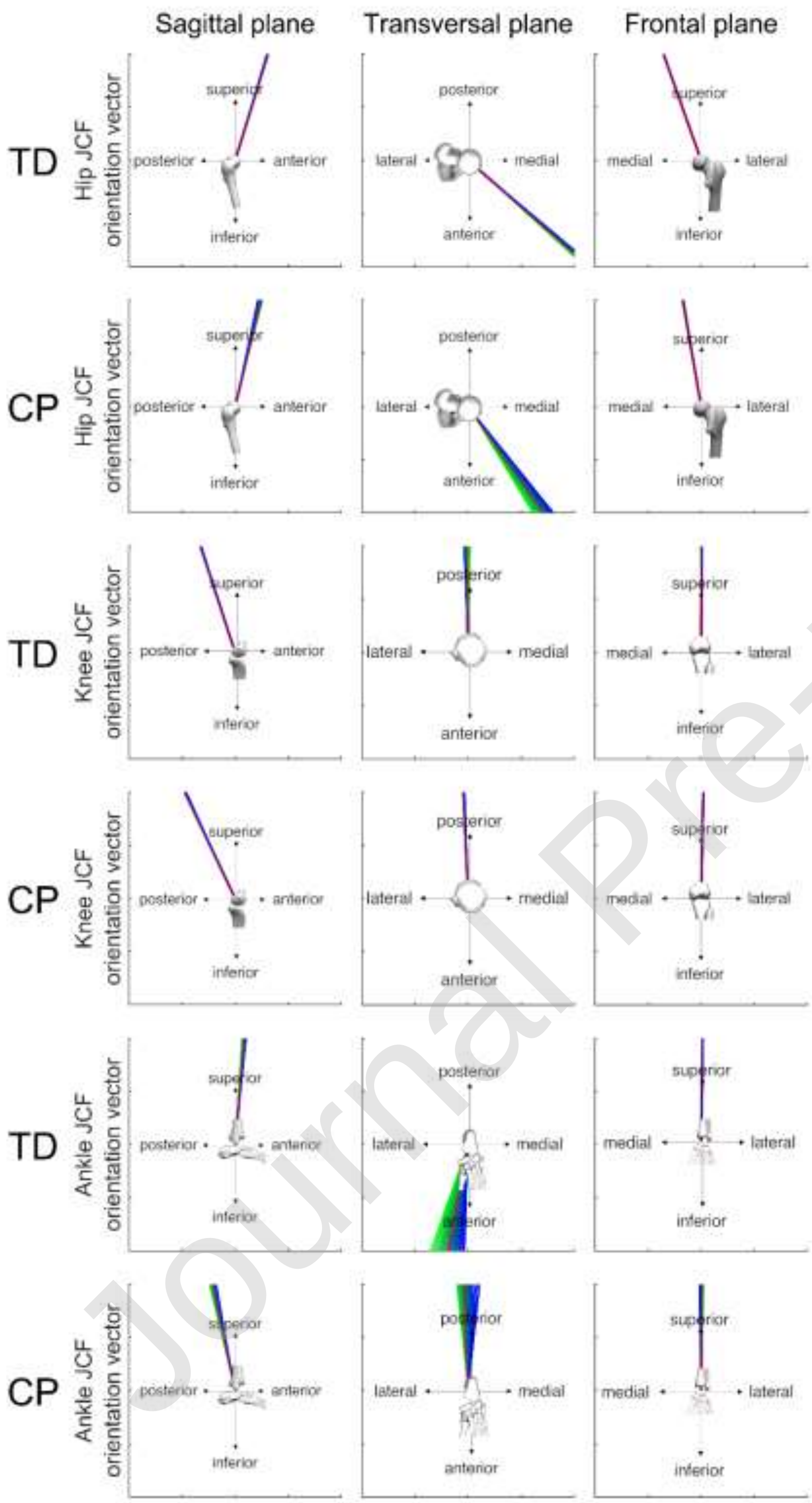

TD
CP
rectus femoris [BW]
semitendinosus [BW]
gluteus medius [BW]
vastus lateralis [BW]
biceps femoris [BW]
tibialis anterior [BW]
medial gastrocnemius [BW]
soleus [BW]
% of gait cycle
% of gait cycle

Figure 5: Comparison of an example of muscle forces obtained with the reference and modified models. The red line shows the mean waveforms of the reference model. Green lines represent the models with shorter femur length while blue lines show the models with increased femur length.



Sagittal plane
Transversal plane
Frontal plane
TD
Hip JCF orientation vector
CP
Hip JCF orientation vector
TD
Knee JCF orientation vector
CP
Knee JCF orientation vector
TD
Ankle JCF orientation vector
CP
Ankle JCF orientation vector
superior
inferior
posterior
anterior
lateral
medial

Figure 6: Joint contact force orientations obtained with the reference and modified models. The red line shows the orientation of the reference model. Green lines represent the models with shorter femur length while blue lines show the models with increased femur length.