Simulated crouch gait by healthy children can help to understand what is behind the crouch gait pattern in CP Children

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Scope Statement

This study aligns with the scope of this special issue in this journal because it leverages motion-tracking–based musculoskeletal simulations to analyze gait deviations in children with cerebral palsy. By comparing real and simulated crouch gait, the work provides biomechanical insights that can directly inform neurorehabilitation strategies and improve clinical decision-making supported by motion-tracking technologies.

Conflict of interest statement

The authors declare a potential conflict of interest and state it below

The author(s) declared that they were an editorial board member of Frontiers, at the time of submission. This had no impact on the peer review process and the final decision

Credit Author Statement

António Veloso: Conceptualization, Formal Analysis, Funding acquisition, Project administration, Resources, Supervision, Validation, Writing – original draft, Writing – review & editing. **Catarina Cardadeiro**: Investigation, Methodology, Software, Writing – original draft, Writing – review & editing. **Filipa Joao**: Conceptualization, Formal Analysis, Investigation, Methodology, Writing – original draft, Writing – review & editing. **Rodrigo Mateus**: Data curation, Formal Analysis, Methodology, Software, Writing – original draft.

Keywords

Cerebral Palsy, crouch gait, inducedacceleration analysis, musculoskeletal modelling, simulation

Abstract

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Neurologic dysfunctions, like cerebral palsy (CP), lead to serious disorders of movement, being walking really affected. Nowadays, the causes associated to crouch gait (CG) are not clearly identified, so being able to differentiate the several gait deviations associated to crouch, may provide guidance for more precise clinical decision-making. Comparing healthy children simulating this pathological gait with CP children with real crouch gait may provide new insight into what is behind the crouch gait pattern. The purpose of this study was to investigate and compare the muscle forces required to walk in simulated crouch and real crouch gait, and to determine how the individual muscle contributions to vertical and fore-aft acceleration of the mass center differ between simulated crouch, real couch, and unimpaired gait, considering just the single support phase of the stance. There were considered three study groups: three children with cerebral palsy walking in severe crouch gait, six typically developing children (TDC) simulating crouch gait, and the same healthy children performing unimpaired gait. The parameters were estimated through musculoskeletal simulations performed in OpenSim software. The results indicate that simulated and real crouch gait show a similar muscle behavior throughout single support in stance phase, relying mostly on the same muscle groups. This suggests that the most significant differences between this pathological gait and normal walking are more likely to be related to the crouch posture adopted than to muscular dysfunctions. The individual muscle contributions to vertical and fore-aft acceleration of the mass center showed that the major contributors to support are the same in all the research groups, being the vasti, soleus and gastrocnemius very important in supporting the crouch posture.

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Ethics statements

Studies involving animal subjects

Generated Statement: No animal studies are presented in this manuscript.

Studies involving human subjects

Generated Statement: The studies involving humans were approved by Ethics Committee for Research of the Faculty of Human Kinetics. The studies were conducted in accordance with the local legislation and institutional requirements. Written informed consent for participation in this study was provided by the participants' legal guardians/next of kin.

Inclusion of identifiable human data

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Generative AI disclosure

No Generative AI was used in the preparation of this manuscript.

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Keywords: Crouch gait, Cerebral Palsy, musculoskeletal modelling, simulation, induced acceleration analysis

Abstract

Neurologic dysfunctions, like cerebral palsy (CP), lead to serious disorders of movement, being walking really affected. Nowadays, the causes associated to crouch gait (CG) are not clearly identified, so being able to differentiate the several gait deviations associated to crouch, may provide guidance for more precise clinical decision-making. Comparing healthy children simulating this pathological gait with CP children with real crouch gait may provide new insight into what is behind the crouch gait pattern. The purpose of this study was to investigate and compare the muscle forces required to walk in simulated crouch and real crouch gait, and to determine how the individual muscle contributions to vertical and fore-aft acceleration of the mass center differ between simulated crouch, real couch, and unimpaired gait, considering just the single support phase of the stance. There were considered three study groups: three children with cerebral palsy walking in severe crouch gait, six typically developing children (TDC) simulating crouch gait, and the same healthy children performing unimpaired gait. The parameters were estimated through musculoskeletal simulations performed in OpenSim software. The results indicate that simulated and real crouch gait show a similar muscle behavior throughout single support in stance phase, relying mostly on the same muscle groups. This suggests that the most significant differences between this pathological gait and normal walking are more likely to be related to the crouch posture adopted than to muscular dysfunctions. The individual muscle contributions to vertical and fore-aft acceleration of the mass center showed that the major contributors to support are the same in all the research groups, being the vasti, soleus and gastrocnemius very important in supporting the crouch posture.

1 Introduction

Cerebral palsy is a permanent neurologic dysfunction caused by serious cerebral damages of the fetal or neonatal brain, primarily leading to disorders of movement and posture. Although the brain damages are not progressive, their expression can change over time [1]. This disorder largely affects the motor control of gait. Consequently, it is a key aspect when it comes to diagnosis. Crouch gait is the most common gait pattern identified in children with this disease, characterized by excessive knee and hip flexion, and increased ankle dorsiflexion. This type of gait overloads the joints, and it requires a much higher energy cost compared to unimpaired gait, so it is extremely inefficient and unsustainable in the long run [2,3]. Furthermore, crouch gait refers to progressive gait deviations that include primary musculoskeletal abnormalities, related directly to neurological disorders, but also secondary deviations that are induced by compensatory effects of the abnormal gait performed. Knowing the primary causes of the gait abnormalities can help clinicians to choose the appropriate corrective treatment and, especially, to define which surgical intervention should be applied.

Motion capture is not enough to study motion with the precision needed in these cases, so musculoskeletal modelling has been widely used as a complementary tool. Previous studies have used this method to investigate muscle activity [4-6] and individual muscle contribution to mass center acceleration [7,8] in crouch gait, by comparing the results obtained with known values for unimpaired gait. Although this method helps in understanding what is behind this pathological gait, comparing it with simulated crouch performed by healthy children may contribute to better distinguish between primary and secondary deviations. Therefore, some studies have aimed to investigate the capacity of neurological intact children to perform crouch gait in a reproducible manner and to characterize the biomechanics of this type of gait, analyzing only the kinetics and kinematics of the motion [9-11]. Inducing physical constraints in healthy subjects to simulate abnormal walking patterns commonly seen in children with neurological disorders as cerebral palsy, has been proven to be useful for a better understanding of the causes behind the pathological gait. This is especially important for progressive gait deviations like crouch gait.

The goals set to this study were to investigate and compare the muscle forces required to walk in simulated crouch and real crouch gait, and to determine how the individual muscle contributions to vertical and fore-aft acceleration of the mass center differ between simulated crouch, real couch, and unimpaired gait, considering just the single support phase of the stance. The analysis was done by using musculoskeletal modeling, performed in the software OpenSim. This work can contribute to improve the diagnosis of crouch gait in children with cerebral palsy and so helping with treatment planning.

2 Materials and Methods

2.1 Participants

The participants were selected from a database of subjects who had previously undergone motion analysis at the University of Lisbon, Faculty of Human Kinetics, as part of an ongoing project. Three children with cerebral palsy were chosen (Table 1) and the selection criteria included: a diagnosis of spastic diplegic CP and classified as presenting a severe crouch pattern. According to Steele's crouch severity classification [6], a crouch pattern is considered severe from a knee flexion angle of 50º. Regarding the typically developing children group, six subjects were chosen as most representative as possible of the age and structure of the CP children selected (Table 1). These subjects performed both simulated crouch gait and their normal walking pattern. They were clinically analyzed, and it was concluded that they did not present any neurological dysfunction. The protocol was approved by and executed in accordance with the Faculty of Human Kinetics Ethics Committee (CEFMH-2/2019). An informed consent was previously signed by the parent or the legal guardian of the participant, and the child assent was also obtained after explaining the entire protocol.

Insert Table 1 here

2.2 Data collection

Firstly, each child was submitted to a clinical exam done by a health professional. A sequence of measures was performed on each subject that aimed to evaluate bone and joint deformities, muscle length, selective motor control, and spasticity. The second part consisted of the motion analysis. The data was collected with Qualisys Track Manager software (Qualisys Inc., Gothenburg, Sweden), version 2.9, operating on an optoelectronic system of 14 Qualisys cameras (Qualisys Oqus 300, Qualisys AB, Gothenburg, Sweden) at a frequency rate of 100 Hz. Ground reaction forces were measured with three Bertec and one Kistler force plates. Each subject had 25 reflective markers and 4 marker clusters placed on specific anatomic places, according to CAST (calibrated anatomical systems technique) protocol and CODA pelvis, used to reconstruct 8 body segments. The gait analysis started with the recording of a static trial barefoot in the standing position. Afterward, the child was instructed to walk along a 10m corridor, at a self-selected speed. The dynamic trials ended when the child successfully achieved a minimum of 10 good kinetic walking cycles for each side, considering the natural variability in kinematic and kinetic gait parameters.

2.3 Data processing

The data processing and inverse kinematics (Supplementary Fig. 1) was performed using Visual3D software. The variables were filtered using a 4th order Butterworth filter at 8Hz. The inverse kinematics problem was solved as a global optimization problem, which means that the pose of the model is computed to best match the data from the motion capture in terms of global criterion. The musculoskeletal modelling was developed using the open-source software OpenSim [12,13], where a musculoskeletal model consists of rigid body segments connected by joints and articulated by actuators, which span these joints and generate forces and motion. It was used a generic model named Gait2392, available in the software. This is a 23 degree of freedom computer model of the human musculoskeletal system in three-dimensions. It features 92 muscle-tendon actuators to represent 76 muscles in the lower extremities and torso. This model represents an average adult subject, which is not ideal in modeling children. However, as there are no generic models for children, this one has been widely used in similar studies [4,7,8,14,15]. The size and inertial properties of all segments were adjusted to represent each subject as well as possible. Logically, all the insertion points of the actuators are also adjusted, as well as joint frame locations. It was done using the scale tool provided by OpenSim. This tool also allows scaling the mass of each segment, which ensures that mass distribution is preserved. The peak isometric force of each muscle was estimated through the Correa and Pandy's scaling approach [16].

Joint moments (Supplementary Fig. 2) were compute using inverse dynamics. The information collected in vivo usually carries dynamic inconsistencies between experimental kinematics and ground reaction forces, normally related to inaccuracies in mass distribution and experimental errors. As the model follows physical laws to simulate the intended movement, it creates non-physical compensatory forces that account for these inconsistencies, called residuals. The residual reduction algorithm (RRA) was used to minimize these effects of modelling and marker data processing errors. It is a form of forward dynamics simulation that uses tracking controllers to follow the model kinematics. The analysis begins by setting the values of the model's generalized coordinates to the values computed by the inverse dynamics tool for the defined initial time. Then, RRA steps forward in time (with each time step of 0.001 s) until the end of the task length. During this process, force values are computed for all the model's actuators at each time step, while the algorithm tries to both reduce the residuals and adjust accelerations according to the original values. The modified musculoskeletal model is used to compute a set of muscle excitations that will drive the dynamic musculoskeletal model to track a set of desired kinematics in the presence of applied external forces, in this case, ground reaction forces. The computed muscle control (CMC) does this by using, not only a static optimization step but also a proportional-derivate control to create a forward dynamic simulation that closely tracks the kinematics from the RRA [12]. The algorithm computes the muscle forces and activations, while accounts for activation and contraction dynamics, which includes the interaction of the force-length-velocity properties of the muscle and the elastic properties of the tendon [17]. Apart from the residuals, reserve actuators are appended to the model to compensate for any possible muscle deficiency during the simulation, for every joint degree of freedom. Finally, the induced acceleration analysis (IAA) was used to compute accelerations induced by individual muscle forces acting on the model. The results represent the contributions of individual muscles for each portion of the movement, especially regarding propulsion and weight-bearing stages. This analysis includes a constraint on both toes that are in contact with the ground, which kinematic behavior is known as pure rolling [18].

2.4 Statistical analysis

Two different statistical tests were applied to test statistically significant differences between the groups. Both are non-parametric tests due to the small number of samples considered and, consequently, to the impossibility of testing the normality of each distribution. Since the typically developing children were performing two different gait patterns, there were different group results for the same subjects, which must be considered as paired samples. When comparing the results of the CP children with any results of the healthy children, they are considered independent samples, so the mean results from each group had to be compared two by two. The Mann-Whitney test was used to compare the group means of children with cerebral palsy performing crouch gait with healthy children, both simulating the pathological gait and performing unimpaired gait. To differentiate the group means obtained from TD children's results, walking in these different gait patterns, it was used the Wilcoxon Signed-Rank Test, as they consist of paired samples. Both statistical tests were performed using the IBM SPSS Statistics software, and the conclusions were taken based on the p-values obtained, considering a 95% confidence interval.

3 Results

The quadriceps and the ankle plantarflexors, in both real and simulated crouch, displayed a sustained force pattern overtime, while in normal gait these muscles presented well-defined peaks of strength related to the stance in which they are expected to be most needed (Fig.1). The crouch gait subjects showed similar muscle contributions, throughout the stance phase, for the gastrocnemius and soleus, with an identified increase during terminal stance and pre-swing.

Insert Figure 1 here

Insert Figure 2 here

Regarding the healthy children simulating crouch gait and performing unimpaired gait, the forces produced by soleus, vasti, rectus femoris, and gluteus maximus were far superior in the simulated crouch (Fig.2). On the other hand, the gastrocnemius, iliopsoas, and ankle dorsiflexors showed higher force values during stance in normal gait. When comparing the average muscle forces results of the simulated crouch with the real crouch gait, only four of the muscle groups reported statistically significant differences. The gluteus maximus and the hip abductors required much more muscle strength during simulated crouch, while iliopsoas and ankle dorsiflexors showed slightly higher demand during real crouch. Finally, by analyzing the normal gait and crouch gait, the results indicate that the only significant differences in the muscle forces between these groups were found in the gastrocnemius, ankle dorsiflexors, and hip abductors. The unimpaired gait required greater muscle forces from the gastrocnemius and hip abductor, but less strength from the ankle dorsiflexors.

The ankle dorsiflexors are the major responsible for the downward acceleration and in slowing the forward progression, in real and simulated crouch (Fig.3). Their contribution is mostly significant during early and mid-stance. In unimpaired gait, this muscle group produced relevant upwards acceleration of the mass center during these gait phases. The hip abductors barely contributed to vertical accelerations in simulated and real crouch gaits during stance, but they generated significant upward acceleration during single support stance in normal gait.

In both unimpaired gait and simulated crouch, the soleus and the gastrocnemius appear to be the muscle groups that contribute the most to the upward acceleration of the mass center (Fig.4). On the other hand, in crouch gait, the major contributors to support are the vasti and the soleus, although the gastrocnemius still have a significant contribution. The upward acceleration produced by soleus was greater during simulated crouch than normal gait and real crouch, while the contribution of the gastrocnemius, was greater during unimpaired gait than simulated and real crouch. The positive contributions of the vasti and rectus femoris to vertical acceleration was greater in simulated crouch than normal gait.

Insert Figure 3 here

Insert Figure 4 here

Regarding fore-aft accelerations, the results were more similar between the research groups. The hamstrings and gastrocnemius produced significant contributions to forward acceleration of the mass center, while the quadriceps contributed to the opposite direction, in all the gait patterns performed. The vasti produced greater backward acceleration during real crouch gait than unimpaired gait. The forward acceleration produced by the soleus was far greater in real crouch, compared to normal gait and simulated crouch. Finally, the ankle dorsiflexors barely contributed to the acceleration of the mass center considering the fore-aft direction, in normal gait and simulated crouch, but in real crouch, they presented a significant contribution to backward acceleration.

Insert Figure 5 here

4 Discussion

The gastrocnemius produced greater muscle forces during unimpaired gait compared with both simulated and real crouch. It was previously suggested that weakness of the gastrocnemius and the hip abductors could contribute to crouch gait [5]. Furthermore, by comparing the TD children simulating crouch with the CP children, the results indicate that the diminished capacity to generate force from the gastrocnemius may be more related to the posture adopted in crouch gait than muscle weakness. In turn, the hip abductors strength required was significantly less during stance in simulated crouch and unimpaired gait than real crouch. So, it is possible that this apparently diminished capacity to produce force is a contributive factor to adopt a crouch posture.

Considering the stance phase in normal walking, the quadriceps, which include the rectus femoris and the vasti muscles, are the major responsible for knee extension and, logically, for the deceleration of knee flexion [19]. The muscle forces from this muscle group are expected mainly during mid-stance [20], which is what is observed in the results from the TD children performing unimpaired gait. On the other hand, in simulated and real crouch gait, the quadriceps have a much more continuous action throughout the stance phase, as was expected based on Steele's work [6]. The results indicate that the forces produced by vasti and rectus femoris were significantly greater in simulated crouch than in unimpaired gait. This was also expected by comparing real crouch gait with normal gait [7,21,22]. Even though the force values were higher in real crouch gait, this difference could not be considered statistically significant due to the high variance of this parameter among the CP children. The higher demand on the quadriceps during simulated crouch and the same expected behavior in real crouch, suggests that the overload on these muscles is necessary to support the crouch posture.

Simulated crouch required a greater demand on the gluteus maximus than unimpaired gait and real crouch gait, which indicates that this muscle may be relevant in the function of counteracting the abnormal posture. The TD children performing this abnormal gait showed a higher capacity in extending the hip compared with the CP children, so it was expected a higher demand on this muscle when comparing these two groups. Muscles are responsible to oppose the effect of gravity in the skeletal, enabling the vertical and forward propulsion of the body, so analysing individual muscle contributions to the mass centre accelerations affords further insight into how support and progression works during gait. Before the foot-flat moment, it is expected that the ankle dorsiflexors are one of the main contributors to support, promoting the upwards mass centre acceleration, in unimpaired gait [20,23,24]. Their function during this stage is to resist the fall of the forefoot because of the weight acceptance. The results of this study are consistent with the assumptions for the normal gait regarding this muscle group, whereas in simulated and real crouch gait this contribution to the upwards mass centre acceleration is not verified, having instead a negative effect in supporting the body. Due to this lack of support during this early stage of the stance, the vasti and soleus appear to be activated earlier to compensate for the downward acceleration generated. This is observed in simulated and real crouch gaits, which suggests that these two muscles are crucial in supporting the body throughout this gait phase.

The quadriceps and the ankle plantarflexors are the major responsible for the upwards acceleration of the mass centre in all the gaits performed. The results suggest that in simulated crouch, the child relies more on the soleus' contribution to upwards acceleration than in real crouch, which indicates that this muscle may be important in supporting the crouch posture. On the other hand, the CP children seemed to rely more on the upwards acceleration produced by the vasti, than the TD children simulating crouch gait. The results are not clear concerning this last assumption because, although the vasti produced greater upwards accelerations during real crouch than simulated crouch, this difference was not proven to be statistically significant due to the high variance of this parameter among the CP children.

The major responsible to modulate the fore-aft accelerations during simulated and real crouch gait are the quadriceps and ankle plantarflexors, accelerating the mass center backward and forward, respectively. This is consistent with that observed in CP children with crouch gait in a previous study [7]. The ankle dorsiflexors produced significantly greater backward acceleration of the mass center in real crouch than simulated crouch and unimpaired gait, which seems to be compensated by the forward acceleration produced by the soleus. These results suggest that weakening or reducing force-generating capacity of the soleus in CP children may disable them to walk or reduce their capacity to progress during gait.

5 Conflict of Interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

6 Author Contributions

C.C., F.J. and RM contributed with the writing of the original draft, data collection and data processing. F.J. and A.V. contributed with the conceptualization and formal analysis of the data. A.V. contributed with supervision of the work, with funding acquisition and project administration.

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8 References

[1] H. K. Graham et al., “Cerebral palsy,” Jan. 07, 2016, Nature Publishing Group. doi: 10.1038/nrdp.2015.82.

[2] P. Rosenbaum, N. Paneth, A. Leviton, M. Goldstein, and M. Bax, “A report: The definition and classification of cerebral palsy April 2006,” 2007, Blackwell Publishing Ltd. doi: 10.1111/j.1469-8749.2007.tb12610.x.

[3] S. Gulati and V. Sondhi, “Cerebral Palsy: An Overview,” Indian J Pediatr, vol. 85, no. 11, pp. 1006–1016, Nov. 2018, doi: 10.1007/s12098-017-2475-1.

[4] R. Dana TUGUI, D. Antonescu, and D. Raluca Tugui, “Cerebral Palsy Gait, Clinical Importance,” Maedica (Bucur), vol. 8, no. 4, p. 388, Sep. 2013, Accessed: Apr. 07, 2025. [Online]. Available: https://pmc.ncbi.nlm.nih.gov/articles/PMC3968479/

[5] M. Sandau, H. Koblauch, T. B. Moeslund, H. Aanæs, T. Alkjær, and E. B. Simonsen, “Markerless motion capture can provide reliable 3D gait kinematics in the sagittal and frontal plane,” Med Eng Phys, vol. 36, no. 9, pp. 1168–1175, Sep. 2014, doi: 10.1016/J.MEDENGPHY.2014.07.007.

[6] T. A. L. Wren, C. A. Tucker, S. A. Rethlefsen, G. E. Gorton, and S. Õunpuu, “Clinical efficacy of instrumented gait analysis: Systematic review 2020 update,” Gait Posture, vol. 80, pp. 274–279, Jul. 2020, doi: 10.1016/j.gaitpost.2020.05.031.

[7] S. Armand, G. Decoulon, and A. Bonnefoy-Mazure, “Gait analysis in children with cerebral palsy,” EFORT Open Rev, vol. 1, no. 12, pp. 448–460, Dec. 2016, doi: 10.1302/2058-5241.1.000052.

[8] S. R. Simon, “Quantification of human motion: Gait analysis - Benefits and limitations to its application to clinical problems,” J Biomech, vol. 37, no. 12, pp. 1869–1880, Dec. 2004, doi: 10.1016/j.jbiomech.2004.02.047.

[9] R. M. Kanko et al., “Assessment of spatiotemporal gait parameters using a deep learning algorithm-based markerless motion capture system,” J Biomech, vol. 122, Jun. 2021, doi: 10.1016/j.jbiomech.2021.110414.

[10] R. Hara, M. Sangeux, R. Baker, and J. McGinley, “Quantification of pelvic soft tissue artifact in multiple static positions,” Gait Posture, vol. 39, no. 2, pp. 712–717, Feb. 2014, doi: 10.1016/J.GAITPOST.2013.10.001.

[11] M. Akbarshahi, A. G. Schache, J. W. Fernandez, R. Baker, S. Banks, and M. G. Pandy, “Non-invasive assessment of soft-tissue artifact and its effect on knee joint kinematics during functional activity,” J Biomech, vol. 43, no. 7, pp. 1292–1301, May 2010, doi: 10.1016/J.JBIOMECH.2010.01.002.

[12] A. Cappozzo, “Gait analysis methodology,” Hum Mov Sci, vol. 3, no. 1–2, pp. 27–50, Mar. 1984, doi: 10.1016/0167-9457(84)90004-6.

[13] A. Peters, B. Galna, M. Sangeux, M. Morris, and R. Baker, “Quantification of soft tissue artifact in lower limb human motion analysis: A systematic review,” Gait Posture, vol. 31, no. 1, pp. 1–8, Jan. 2010, doi: 10.1016/J.GAITPOST.2009.09.004.

[14] D. L. Benoit, D. K. Ramsey, M. Lamontagne, L. Xu, P. Wretenberg, and P. Renström, “Effect of skin movement artifact on knee kinematics during gait and cutting motions measured in vivo,” Gait Posture, vol. 24, no. 2, pp. 152–164, Oct. 2006, doi: 10.1016/J.GAITPOST.2005.04.012.

[15] A. Peters, B. Galna, M. Sangeux, M. Morris, and R. Baker, “Quantification of soft tissue artifact in lower limb human motion analysis: a systematic review,” Gait Posture, vol. 31, no. 1, pp. 1–8, Jan. 2010, doi: 10.1016/J.GAITPOST.2009.09.004.

[16] F. D’Isidoro, C. Brockmann, and S. J. Ferguson, “Effects of the soft tissue artefact on the hip joint kinematics during unrestricted activities of daily living,” J Biomech, vol. 104, May 2020, doi: 10.1016/J.JBIOMECH.2020.109717.

[17] T. Y. Tsai, T. W. Lu, M. Y. Kuo, and C. C. Lin, “Effects of soft tissue artifacts on the calculated kinematics and kinetics of the knee during stair-ascent,” J Biomech, vol. 44, no. 6, pp. 1182–1188, Apr. 2011, doi: 10.1016/J.JBIOMECH.2011.01.009.

[18] A. Leardini, A. Chiari, U. Della Croce, and A. Cappozzo, “Human movement analysis using stereophotogrammetry Part 3. Soft tissue artifact assessment and compensation,” Gait Posture, vol. 21, no. 2, pp. 212–225, 2005, doi: 10.1016/J.GAITPOST.2004.05.002.

[19] M. Moro, G. Marchesi, F. Hesse, F. Odone, and M. Casadio, “Markerless vs. Marker-Based Gait Analysis: A Proof of Concept Study,” Sensors, vol. 22, no. 5, Mar. 2022, doi: 10.3390/s22052011.

[20] Corazza S, Mündermann L, and Andriacchi T, “Markerless Motion Capture Methods for the Estimation of Human Body Kinematics,” Stanford, 2006. Accessed: May 06, 2024. [Online]. Available: https://citeseerx.ist.psu.edu/document?repid=rep1&type=pdf&doi=5e2080eca90de9fdb969088a40b49c4524d5b230

[21] N. Ito et al., “Markerless motion capture: What clinician-scientists need to know right now,” JSAMS Plus, vol. 1, p. 100001, Oct. 2022, doi: 10.1016/J.JSAMPL.2022.100001.

[22] R. M. Kanko, E. K. Laende, E. M. Davis, W. S. Selbie, and K. J. Deluzio, “Concurrent assessment of gait kinematics using marker-based and markerless motion capture,” J Biomech, vol. 127, Oct. 2021, doi: 10.1016/j.jbiomech.2021.110665.

[23] R. M. Kanko, E. Laende, W. S. Selbie, and K. J. Deluzio, “Inter-session repeatability of markerless motion capture gait kinematics,” J Biomech, vol. 121, May 2021, doi: 10.1016/j.jbiomech.2021.110422.

[24] T. A. L. Wren, P. Isakov, and S. A. Rethlefsen, “Comparison of kinematics between Theia markerless and conventional marker-based gait analysis in clinical patients,” Gait Posture, vol. 104, pp. 9–14, Jul. 2023, doi: 10.1016/j.gaitpost.2023.05.029.

[25] A. Cappozzo, F. Catani, U. Della Croce, and A. Leardini, “Position and orientation in space of bones during movement: anatomical frame definition and determination,” Clinical Biomechanics, vol. 10, no. 4, pp. 171–178, 1995, doi: 10.1016/0268-0033(95)91394-T.

[26] R. Baker, “Pelvic angles: a mathematically rigorous definition which is consistent with a conventional clinical understanding of the terms,” Gait Posture, vol. 13, no. 1, pp. 1–6, 2001, doi: 10.1016/S0966-6362(00)00083-7

[27] K. Song, T. J. Hullfish, R. Scattone Silva, K. G. Silbernagel, and J. R. Baxter, “Markerless motion capture estimates of lower extremity kinematics and kinetics are comparable to marker-based across 8 movements,” J Biomech, vol. 157, Aug. 2023, doi: 10.1016/j.jbiomech.2023.111751.

[28] R. Riemer, E. T. Hsiao-Wecksler, and X. Zhang, “Uncertainties in inverse dynamics solutions: A comprehensive analysis and an application to gait,” Gait Posture, vol. 27, no. 4, pp. 578–588, May 2008, doi: 10.1016/J.GAITPOST.2007.07.012.

[29] H. Tang, J. Pan, B. Munkasy, K. Duffy, and L. Li, “Comparison of Lower Extremity Joint Moment and Power Estimated by Markerless and Marker-Based Systems during Treadmill Running,” Bioengineering, vol. 9, no. 10, Oct. 2022, doi: 10.3390/bioengineering9100574.

Tables and Figures

Table 1. Characteristics of the typical developed (TD) and cerebral palsy (CP) children

	N	Age (yrs) Mean ±sd	Height (cm) Mean ±sd	Mass (kg) Mean ±sd
TD children	6	8±1	127±5	25±3
CP children	3	12±3	139±18	35±9

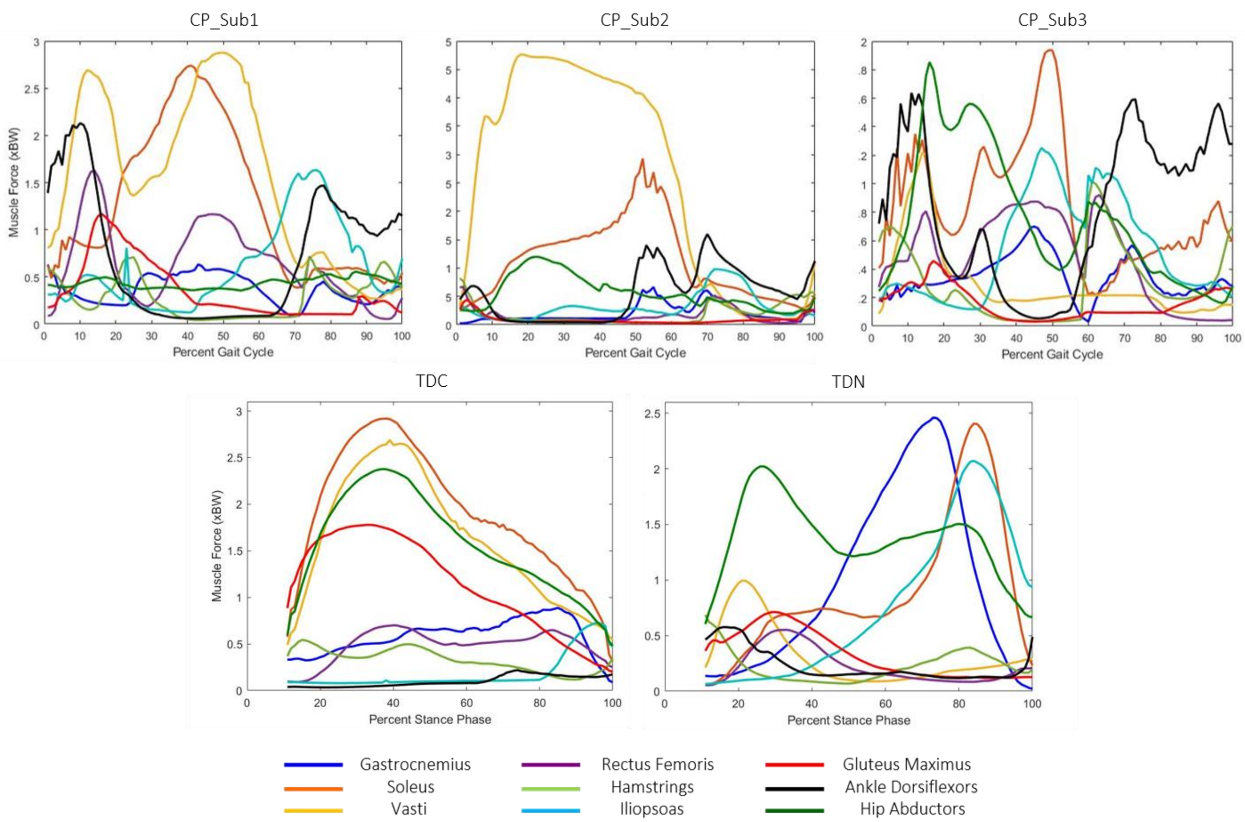

Fig. 1. Average muscle forces normalized by bodyweight (BW) obtained from CMC, during one gait cycle for all the research groups.

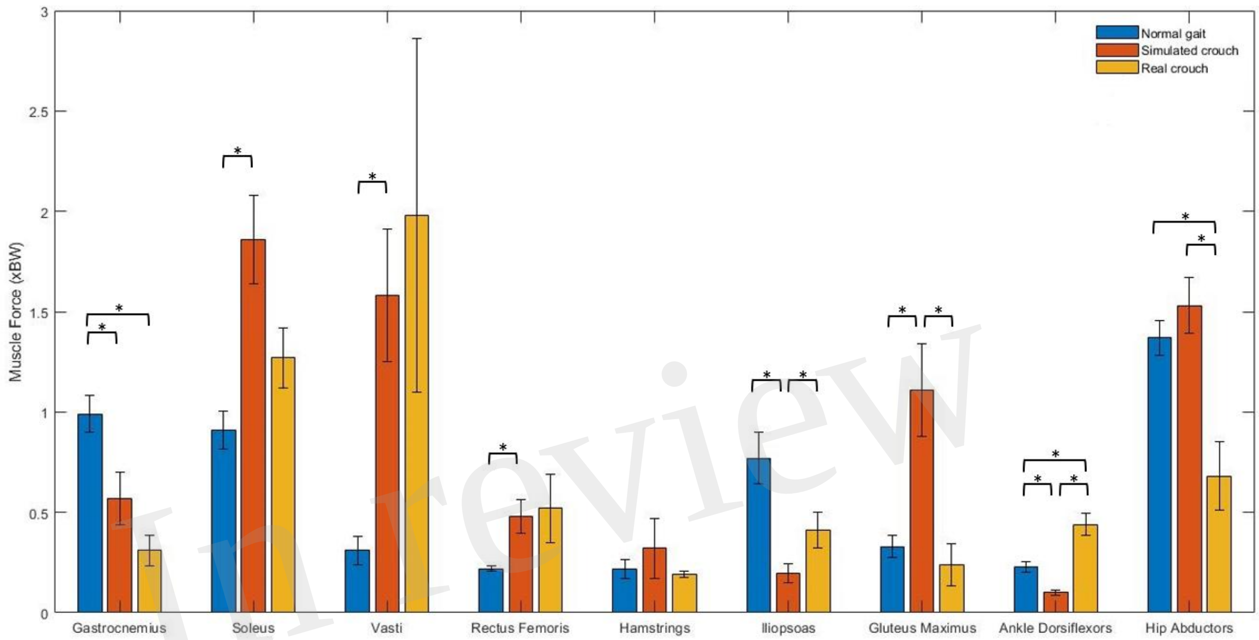

Fig. 2. Average muscle force during single support in stance phase normalized by bodyweight (BW). Error bars are ±1 standard error.

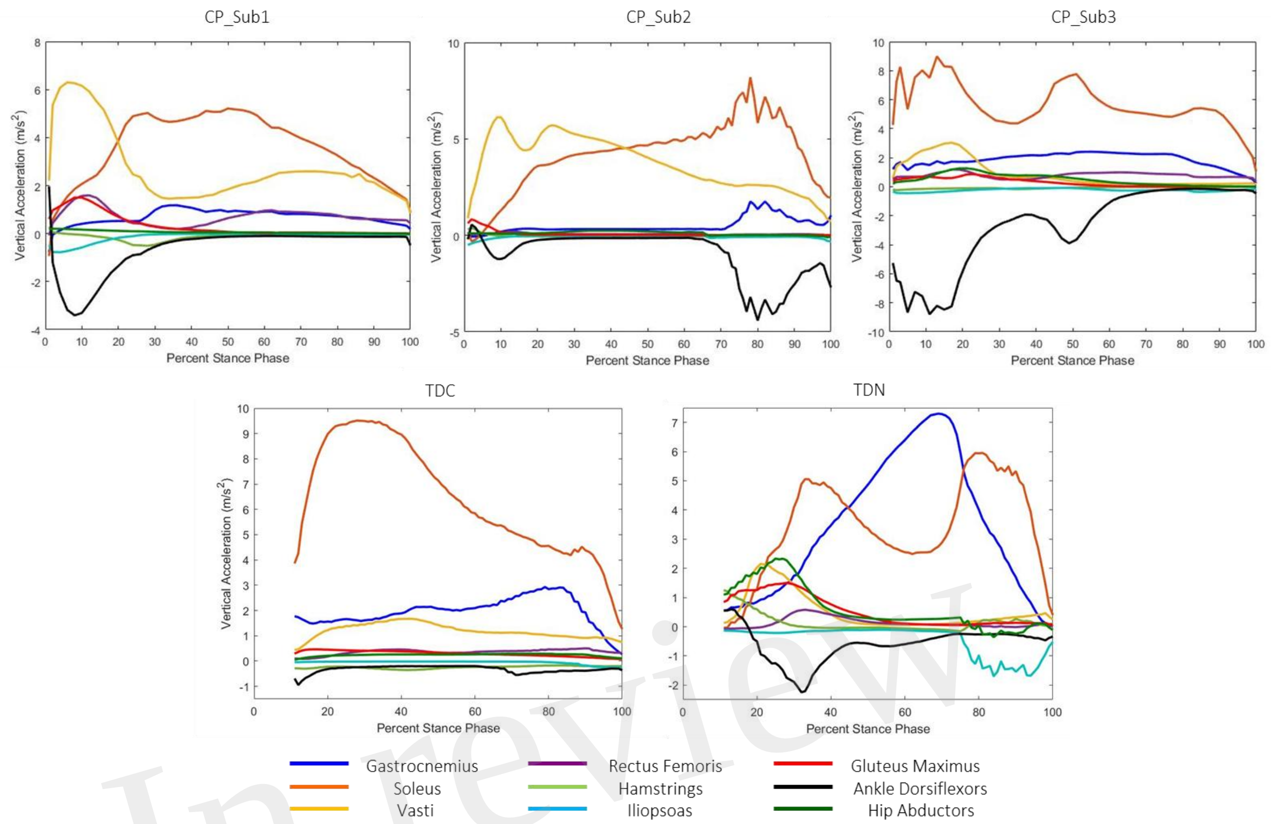

Fig. 3. Contributions of each muscle group to the accelerations of the body's center of mass, along the vertical direction.

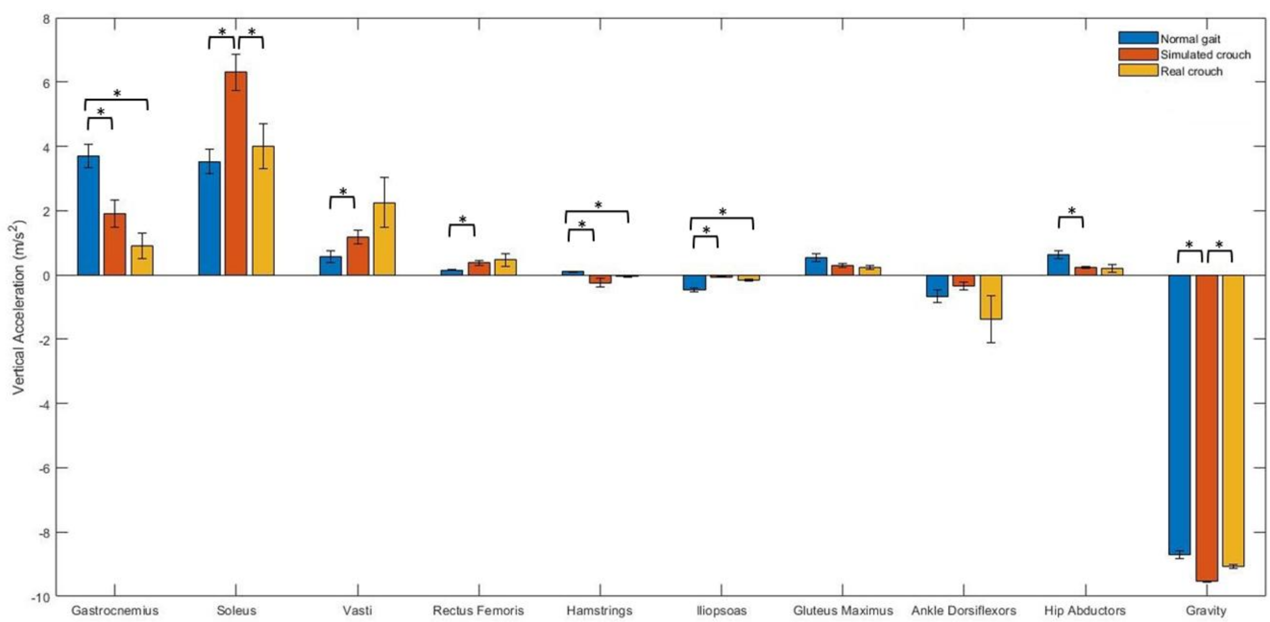

Fig. 4. The average vertical accelerations of the mass center during single support in stance phase produced by each muscle. Gravity indicates the acceleration of the mass center when only gravity is applied. Error bars are ±1 standard error.

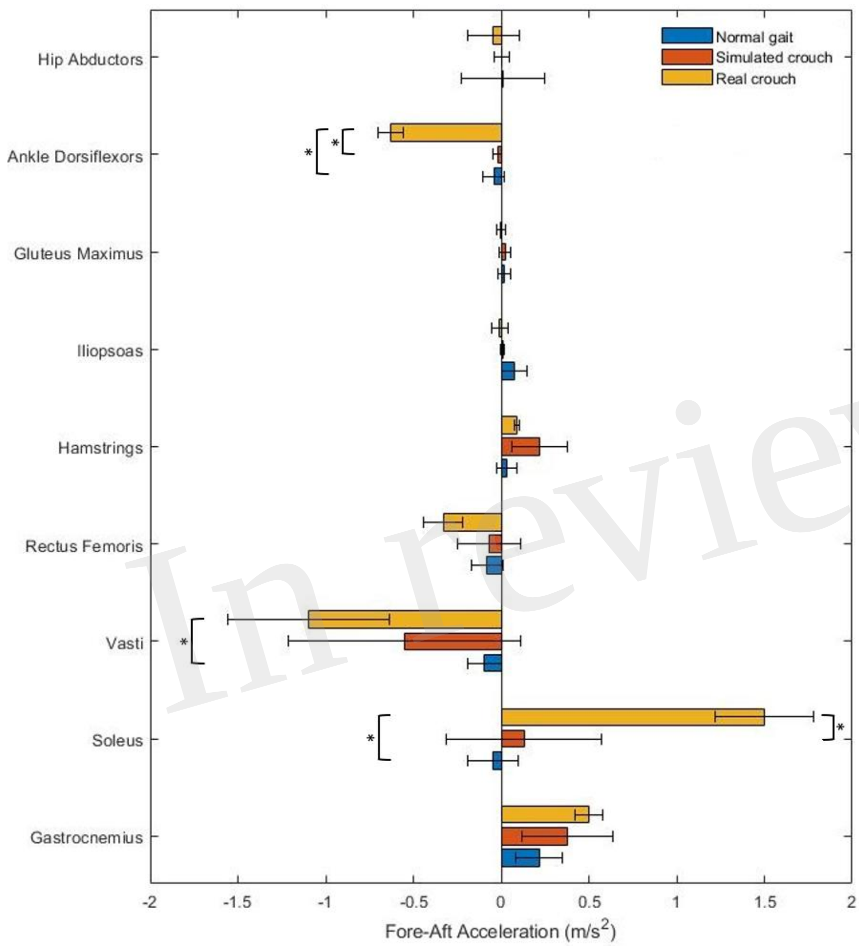

Fig. 5. The average fore-aft accelerations of the mass center during stance produced by each muscle. Error bars are ±1 standard error.

Figure 1.JPEG

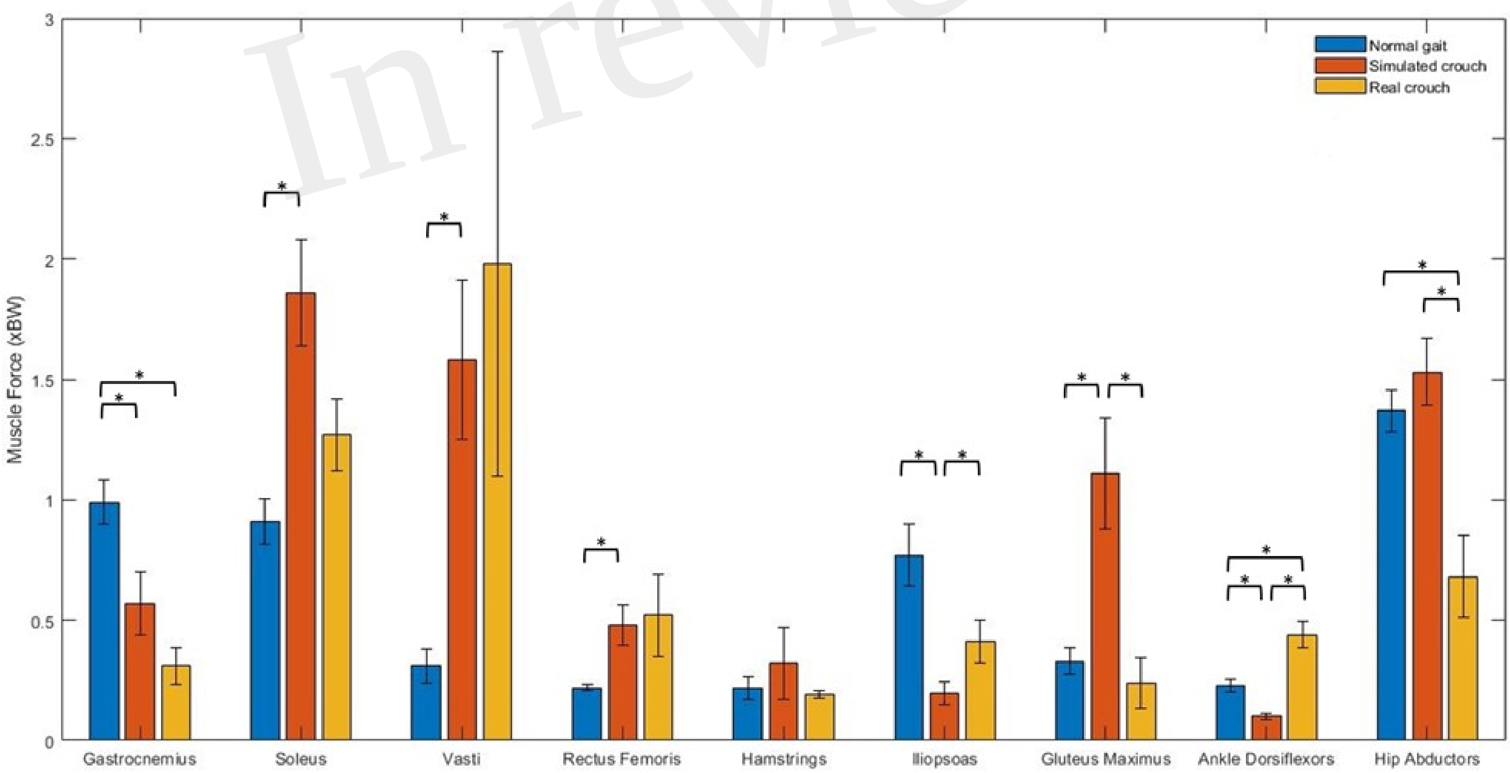

Figure 2.JPEG

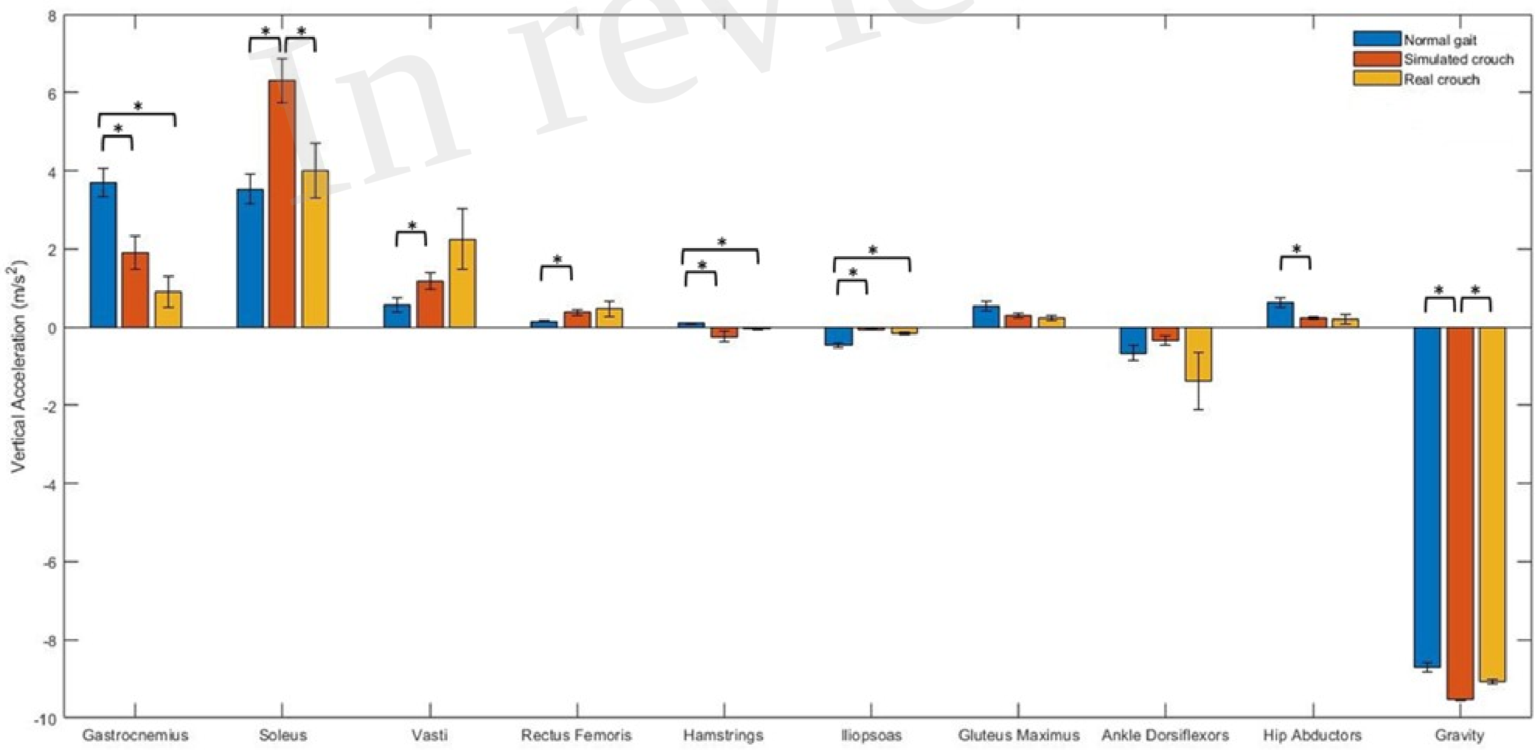

Figure 3.JPEG

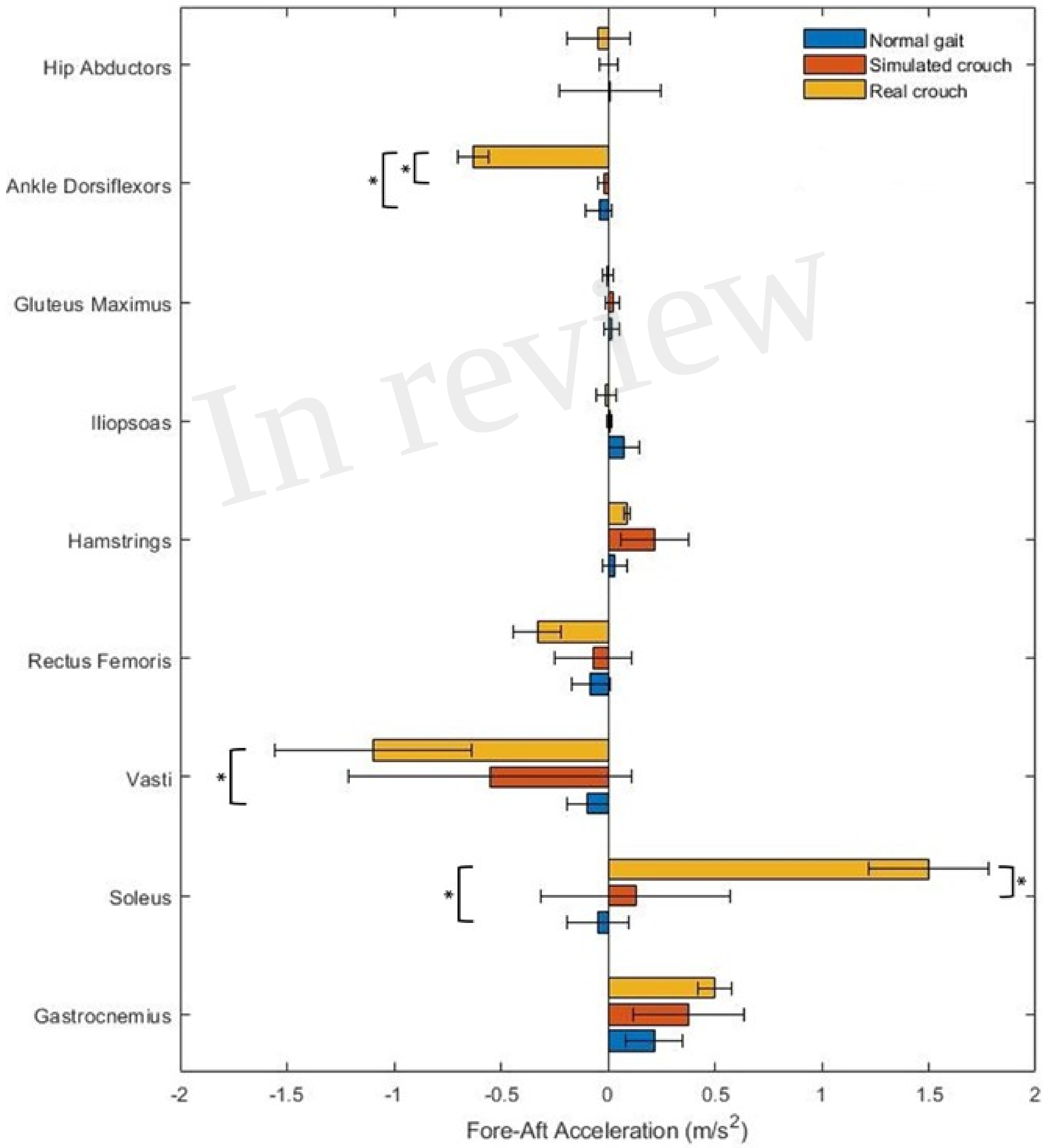

Figure 4.JPEG

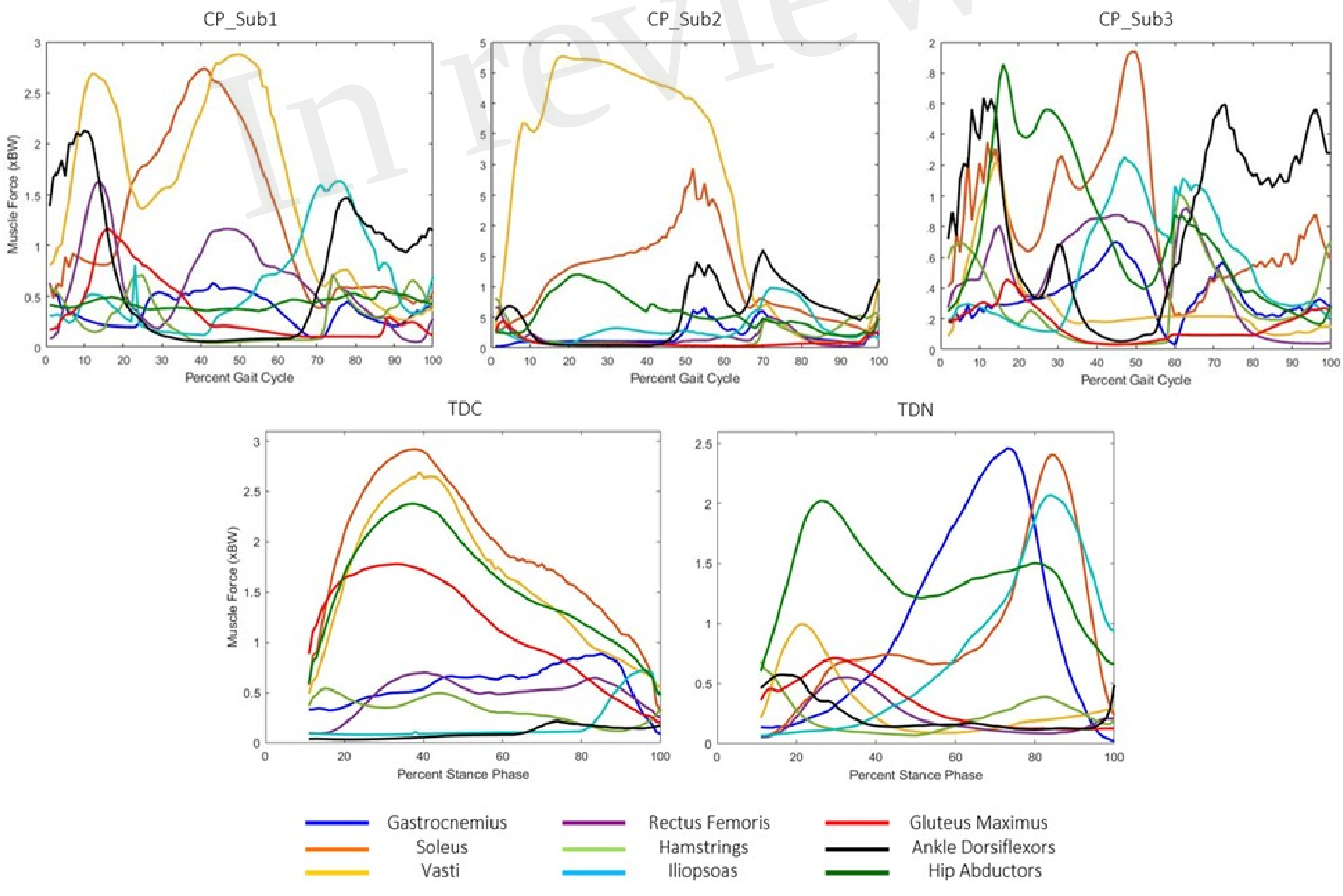

Figure 5.JPEG