MUSCLE FORCE ESTIMATION IN CLINICAL BIOMECHANICS: ANYBODY VS OPENSIM

Ursula Trinler[1], Nathalie Alexander[2], Hermann Schwameder[2], Richard Baker[1]

School of Health Science, University of Salford, Manchester, United Kingdom[1]
Department of Sport Science and Kinesiology, University of Salzburg, Salzburg, Austria[2]

Knowing the force profiles of individual muscles during various functional tasks may help to better identify various neuro-musculoskeletal impairments from functional movement analysis. Different simulation environments exist for this purpose. The aim of this study was to compare gait muscle force estimations (static optimisation) between the simulation environments AnyBody and OpenSim, using two similar musculoskeletal models. Results show mostly similar muscle forces, while some differences exist, resulting out of different anthropometrics and constraints of the generic models. The findings indicate the necessity to carefully analyse results when comparing muscle force estimations from different simulation environments. Future studies will develop a standardised protocol for such analyses.
.

KEY WORDS: musculoskeletal modelling, movement analysis

INTRODUCTION: Knowing the force profiles of individual muscles during various functional tasks may help to better identify various neuro-musculoskeletal impairments from functional movement analysis and give a better understanding about how these affect movement. Patellar femoral pain, for example, is often thought (Herzog, 1998) to be due to an imbalance of force between agonist and antagonist and can lead to excessive loading of knee joint and subsequent risk of developing degenerative joint conditions and injuries (Yavuz, Sendemir-Urkmez, & Turker, 2010). A better understanding of actual muscle forces might help to identify such mechanisms of functional impairments.

More recently computational techniques have made it possible to estimate muscle forces (Lin, Dorn, Schache, & Pandy, 2012). These modelling approaches have already been applied in a variety of research studies related to sport performance or clinical interventions (Anderson & Pandy, 1999; Hatze, 1981). Use of muscle force modelling has, however, not yet become established in a routine movement analysis. A range of musculoskeletal models in different simulation environments are available (Anderson & Pandy, 1999, 2001; Lin et al., 2012). This complicates the comparison of results gained through different simulation environments, as many aspects related to the musculoskeletal model as well as the chosen mathematical approach might differ.

A direct comparison between simulation environments has never been done before in the context of functional movement analysis, although discrepancies in such environments might result in different muscle forces. Therefore, the purpose of this study was to estimate muscle forces of human healthy walking in two different simulation environments (AnyBody, OpenSim) with static optimisation (inverse dynamics approach), including a polynomial muscle recruitment order of 3. Provided musculoskeletal models were used in its standard settings.

METHODS: Ethical approval was granted by the College of Health and Social Care Ethics Panel, University of Salford. A convenience sample of ten healthy adult volunteers with no history of neuro-musculoskeletal impairments was recruited from the university community (28±5 years old, 1.72±0.08m, 69±12kg). Thirty-four markers were placed, which were adapted from the marker set in the OpenSim example data of model *Gait2392*. A ten camera

motion capture system (Nexus 1.8.5, Vicon, T40S cameras, 100Hz) was used to capture walking data at self-selected speed over a walkway equipped with four force plates (Kistler, 2x 9286A, 2x 9253A, 1000Hz). Surface EMG of main muscles of the lower limb were additionally captured (Noraxon, 16 channel, DTS receiver, 1000Hz). Five valid gait cycles for each leg were analysed. Marker trajectories and ground reaction forces were filtered with 5Hz and 12Hz, respectively, and used as input into the estimation process.
Muscle forces were estimated using AnyBody (vers. 6.0, AnyBody Technology, Denmark) and OpenSim (vers. 3.2, OpenSim). Both programmes provide musculoskeletal models for the analysis of walking, the *Twente Lower Extremity Model* included into the *Mocap LowerBody model* (AMMR 1.6.2, AnyBody) and *Gait2392* (OpenSim, Au & Dunne, 2013). Both models were mainly used in its standard settings, similar in number of segments, and set to the same degrees of freedom (DoF) (3DoFs hip, 1DoF knee, 1DoF ankle). Both unscaled generic models had similar properties in the height and weight (both 1.80m, 75.16kg *Gait2392*, 75,46kg *Twente Model*). Virtual markers were placed on the same representative anatomical landmarks.
Standard pipelines were used to estimate joint angles (inverse kinematics), joint moments (inverse dynamics) and muscle forces (static optimisation) in both simulation environments. The LengthMass scaling approach was used with AnyBody to scale the static trial, while a customised similar scaling approach was applied with OpenSim. Static optimisation was solved by minimizing a polynomial muscle recruitment criterion with the exponent of 3. The *Twente Model* (55 muscles divided into 159 muscle-tendon actuators per leg) and *Gait2392* (36 muscles, divided into 46 muscle-tendon actuators per leg) differed in the number of muscles used for the estimation. No force-length-velocity model was applied to estimate muscle forces, to reduce potential influencing factors (Arnold, Hamner, Seth, Millard, & Delp, 2013; Carbone, van der Krogt, Koopman, & Verdonschot, 2016).
Results are compared visually and statistically with the paired t-test after ensuring a normal distribution (SPSS, IBM Corp. 2011).

RESULTS: Similar results were found for all ten participants, thus, results of one participant are presented in Figure 1 and 2 (27 years, ♂, 1.83m, 74.3kg). Group mean sagittal joint angles and moments are similar between simulation environments and lie within each other's standard deviation (SD) band, except ankle dorsi-/plantarflexion and hip flexion/extension (Figure 1). Both angles show an offset throughout the gait cycle between simulation environments, with a significant group mean difference of 12,2±1,5° and 5,5±1,9° for the hip and ankle angle, respectively (p<0,0001).

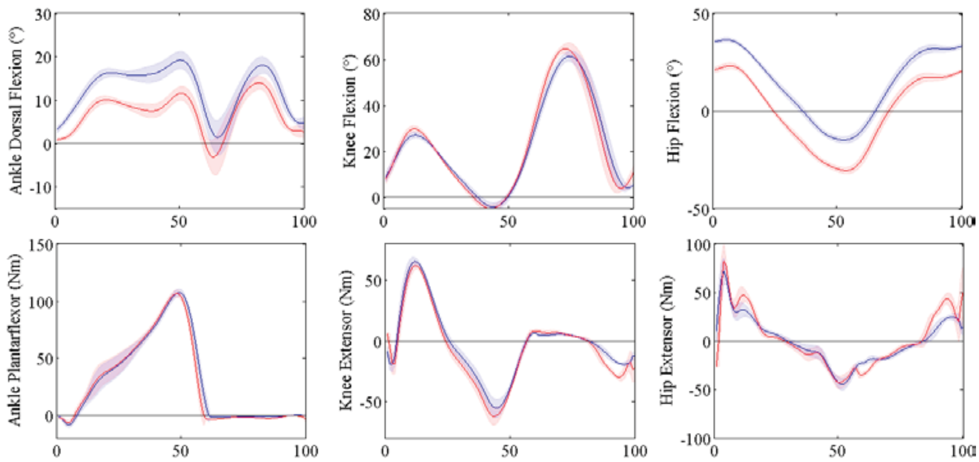

Figure 1. Mean and ± two SD of joint angles and moments of one participant in sagittal plane. Blue and red curves represent results estimated with AnyBody and OpenSim, respectively. X-axes define 100% of a gait cycle with 0% and 100% representing foot contact of the same foot.

On-off muscle pattern are mostly similar between AnyBody and OpenSim except for the Tibialis anterior, Semitendinosus and Biceps Femoris long head in stance and the Gastrocnemius lateralis and Gluteus medius in the end of swing (Figure 2, true for all participants). Estimated muscle forces are higher for AnyBody than OpenSim for the Soleus (group mean peak force difference between AnyBody and OpenSim: 433±218N, *p*<0,0001), Gastrocnemius lateralis, medialis (206±43N and 368±118, both *p*<0,0001), and Semitendinosus (233±53N, *p*<0,0001). Semimembranosus and Biceps Femoris short head develop greater peak forces in stance and swing with OpenSim than AnyBody (62±81N, *p*=0,039 and 218±80N, *p*<0,0001, respectively). Tibialis posterior is visually different for some of the participants, however, without a significant group mean peak (*p*>0,05).

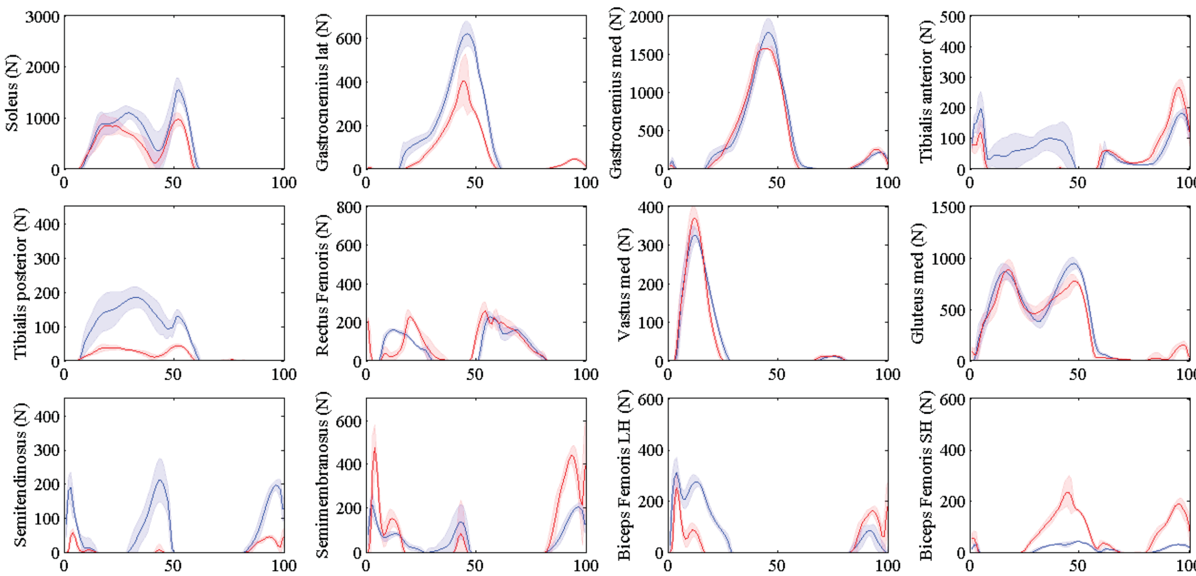

Figure 2. Mean and ± two SD of muscle forces of one participant.

DISCUSSION: Hip kinematics may differ as a consequence of the different definition of the pelvis to the ground (Au & Dunne, 2013), ankle kinematics by using different constraints at the ankle. Sagittal joint moments are similar for all three lower limb joints, indicating similar overall produced muscle forces acting on these joints. The variation in muscle on-off and force production between AnyBody and OpenSim might have multiple reasons, as the number of muscle-tendon actuators, the origin and insertion and the maximum isometric force of each actuator, as well as the mass and inertia of each segments differ between chosen musculoskeletal models (Anderson & Pandy, 1999; Au & Dunne, 2013; Delp et al., 1990; Klein Horsman, Koopman, van der Helm, Prose, & Veeger, 2007).

These model differences might have favoured different muscles of the same agonist/antagonist muscle group (Crowninshield, 1978). Thus, the higher peak muscle force of the gastrocnemii with AnyBody compared to OpenSim in mid stance might be compensated through the higher force of Biceps Femoris short head in OpenSim, which are both knee flexors. Similar pattern is seen in the force distribution between the hamstrings, where Semitendinosus shows higher forces with AnyBody and Semimembranosus with OpenSim. Different constraints at the ankle might induce the continuous activation on the Tibialis anterior in stance with the *Twente model* compared to *Gait2392*.

Estimated muscle forces are mostly similar to parallel captured surface EMG data. However, there is no EMG activation in mid stance of the Semitendinosus and Semimembranosus (Heintz & Gutierrez-Farewik, 2007), whereas muscle forces have here been estimated. This might indicate too weak/too costly intensive muscle-tendon actuators of the triceps surae muscles, which leads to an activation of the agonistic knee flexors the hamstrings. Also, gastrocnemius lateralis and medialis are activated at the end of swing with both AnyBody and OpenSim, which is not true for surface EMG. This small force production could be responsible for an eccentric activation due to the preparation of the ankle before foot contact.

CONCLUSION: Different simulation environments provide their own mathematical and musculoskeletal models which can influence the estimation output. Understanding these differences as well as the limitations of mathematical approaches to estimate muscle forces will help practitioners to operate such simulation environments, to analyse the results and to compare these to other movement laboratories. This study could show, that there is a general agreement between the two simulation environments AnyBody and OpenSim, however, some distinct differences exists in the kinematics and muscle force estimations. One crucial point to consider are the agonist/antagonist interaction when analysing the results, as well as the different anatomical definitions and segmental interactions of the musculoskeletal models. To be able to better distinguish between the performance of AnyBody and OpenSim the same musculoskeletal model will be implemented in both simulation environments in future studies, which might lead to a standardised protocol for such analyses.

REFERENCES:

Anderson, F. C., & Pandy, M. G. (1999). A dynamic optimization solution for vertical jumping in three dimensions. Comput. Meth. Biomech. Biomed. Eng., 2, 201-231.

Anderson, F. C., & Pandy, M. G. (2001). Static and dynamic optimization solutions for gait are practically equivalent. J Biomech, 34(2), 153-161. doi: 10.1016/S0021-9290(00)00155-X

Arnold, E. M., Hamner, S. R., Seth, A., Millard, M., & Delp, S. L. (2013). How muscle fiber lengths and velocities affect muscle force generation as humans walk and run at different speeds. J Exp Biol, 216(Pt 11), 2150-2160. doi: 10.1242/jeb.075697

Au, C., & Dunne, J. (2013). Gait 2352 and 2394 models. Retrieved 18th July, 2016, from http://simtk-confluence.stanford.edu:8080/display/OpenSim/Gait+2392+and+2354+Models

Carbone, V., van der Krogt, M., Koopman, H., & Verdonschot, N. (2016). Sensitivity of subject-specific models to Hill muscle-tendon model parameters in simulations of gait. J Biomech, 49(9), 1953-1960.

Crowninshield, R. (1978). Use of optimization techniques to predict muscle forces J Biomech Engng, 100, 88-92.

Delp, S. L., Loan, P., Hoy, M. G., Zajac, F. E., Topp, E. L., & Rosen, J. M. (1990). An interactive graphics-based model of the lower extremtiy to study orhtopaedics surgical procedures. IEEE transactions on bio-medical engineering, 37, 757-767.

Hatze, H. (1981). A comprehensive model for human motion sumulation and its application to the take-off phase of the long jump. J Biomech, 14, 135-142.

Heintz, S., & Gutierrez-Farewik, E. M. (2007). Static optimization of muscle forces during gait in comparison to EMG-to-force processing approach. Gait and Posture, 26(2), 279-288.

Herzog, W. (1998). Muscle. In B. M. Nigg & W. Herzog (Eds.), Biomechanics of the Musculo-skeletal System (Vol. 2nd, pp. 148-188). West Sussex, UK: John Wiley & Sons Ltd.

Klein Horsman, M. D., Koopman, H. F., van der Helm, F. C., Prose, L. P., & Veeger, H. E. (2007). Morphological muscle and joint parameters for musculoskeletal modelling of the lower extremity. Clin Biomech, 22(2), 239-247. doi: 10.1016/j.clinbiomech.2006.10.003

Lin, Y. C., Dorn, T. W., Schache, A. G., & Pandy, M. G. (2012). Comparison of different methods for estimating muscle forces in human movement. Proceedings of the Institution of Mechanical Engineers. Part H - Journal of Engineering in Medicine, 226(2), 103-112.

Yavuz, S. U., Sendemir-Urkmez, A., & Turker, K. S. (2010). Effect of gender, age, fatigue and contraction level on electromechanical delay. Clin Neurophysiol, 121(10), 1700-1706. doi: 10.1016/j.clinph.2009.10.039