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Muscle force estimation in clinical gait analysis using AnyBody and OpenSim

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ABSTRACT

A variety of musculoskeletal models are applied in different modelling environments for estimating muscle forces during gait. Influence of different modelling assumptions and approaches on model outputs are still not fully understood, while direct comparisons of standard approaches have been rarely undertaken. This study seeks to compare joint kinematics, joint kinetics and estimated muscle forces of two standard approaches offered in two different modelling environments (AnyBody, OpenSim). It is hypothesised that distinctive differences exist for individual muscles, while summing up synergists show general agreement. Experimental data of 10 healthy participants (28 ± 5 years, 1.72 ± 0.08 m, 69 ± 12 kg) was used for a standard static optimisation muscle force estimation routine in AnyBody and OpenSim while using two gait-specific musculoskeletal models. Statistical parameter mapping paired *t*-test was used to compare joint angle, moment and muscle force waveforms in Matlab. Results showed differences especially in sagittal ankle and hip angles as well as sagittal knee moments. Differences were also found for some of the muscles, especially of the triceps surae group and the biceps femoris short head, which occur as a result of different anthropometric and anatomical definitions (mass and inertia of segments, muscle properties) and scaling procedures (static vs. dynamic). Understanding these differences and their cause is crucial to operate such modelling environments in a clinical setting. Future research should focus on alternatives to classical generic musculoskeletal models (e.g. implementation of functional calibration tasks), while using experimental data reflecting normal and pathological gait to gain a better understanding of variations and divergent behaviour between approaches.

1. Introduction

Knowing individual muscle force profiles during gait may help to better understand their functional tasks but also neuro-musculoskeletal impairments leading to a better understanding about how these affect movement. Patellar femoral pain, for example, is often thought (Herzog, 1998) to be caused by an imbalance of forces between agonist, synergists and antagonists and can lead to excessive loading of the knee joint and subsequently the risk of developing degenerative joint conditions and injuries (Yavuz et al., 2010). More recently, computational and modelling techniques facilitate to estimate muscle forces (Lin et al., 2012). These techniques have already been applied in a variety of research studies related to sport performance or clinical interventions (Alexander and Schwameder, 2016b,c; Anderson and Pandy, 1999; Hatze, 1981), however, have not yet become established in a routine movement analysis.

One reason is the variety of existing musculoskeletal models applied in different modelling environments, which can lead to substantial differences (Trinler et al., 2018a). The influence of different modelling assumptions and approaches on model outputs are still not fully understood. This complicates the comparison between results gained with different modelling environments, as each approach incorporates different aspects of musculoskeletal morphology, kinematics, kinetics and muscle function using a range of different assumptions (Anderson and Pandy, 1999, 2001; Lin et al., 2012).

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Two of the mainly used simulation tools in biomechanical analyses are *AnyBody* and *OpenSim*. Both modelling environments provide different musculoskeletal models with which estimations including muscle activations and forces can be executed. Muscle activations in gait analysis calculated by AnyBody (Alexander and Schwameder, 2016a) and OpenSim (Liu et al., 2008; van der Krogt et al., 2012; Trinler et al., 2018b) have been previously compared to measured electromyographic muscle activations and have shown overall a good agreement for trials with self-selected walking speed on level ground.

Both simulation tools, however, have only been partly compared with each other to analyse their agreement in muscle force estimation. Sandholm et al. (2011) have used AnyBody and OpenSim to compare two different anatomical knee joints and their influence on kinematics, muscle activations and joint reaction forces during walking. AnyBody has been implemented, however, only to create the geometry-based knee joint for including it further into OpenSim. Wagner et al. (2013) have applied different musculoskeletal models provided by AnyBody, OpenSim and Matlab. Looking at a knee-extension task while only taking the quadriceps muscles into account, they have shown substantial variation among models in the quadriceps moment arms, distributions in quadriceps muscle recruitment and tibio-femoral joint reaction forces.

A direct comparison between estimated muscle forces gained through different modelling environments has to our knowledge not yet been executed in the context of functional movement analysis, although discrepancies in such environments might result in different muscle forces, leading to varying interpretations of the outcome. Therefore, this study seeks to fill this gap in the current literature and robustly compare estimated muscle and joint forces and, additionally, its joint angles and joint moments with two standard approaches offered in two different modelling environments (AnyBody and OpenSim). This is a necessary step to understand differences between approaches and facilitating muscle force estimations in future clinical use. It is hypothesised that joint angles and moments as well as summed up forces of muscle synergists are similar between AnyBody and OpenSim, but that distinctive differences can be detected between individual muscle forces.

2. Methods

2.1. Participants

A sample of ten healthy adults with no history of neuro-musculoskeletal impairments were recruited (5♀/5♂, mean ± SD of age 28 ± 5 years, height 1.72 ± 0.08 m, mass 69 ± 12 kg). All participants provided written informed consent. Ethical approval was granted by the College of Health and Social Care Ethics Panel.

2.2. Experimental setup

A ten-camera motion capture system (Vicon Nexus 1.8.5, T40s cameras, 100 Hz) was used with four force plates embedded into a ten-meter walkway (Kistler, 1000 Hz). Cameras were calibrated and calibration tested before the measurement using the *Caltester* (Holden et al., 2003), while guidelines from the Clinical Movement Analysis Society UK & Ireland (CMAS) were applied (<1° force orientation error, <3 mm CoP displacement error).

Reflective markers were placed on the anatomical landmarks similar to a CAST model (Leardini et al., 2007; Chand et al., 2012) (see Table S1 in supplements). Following a static standing trial, participants walked at a self-selected speed while five valid gait cycles (individual foot fall entirely on a single force plate) for each leg were recorded.

2.3. Data processing

The raw marker trajectories were pre-processed in Vicon Nexus using a customised pipeline for calculating several virtual landmarks and joint centres (Trinler and Baker, 2018). Both kinematics and ground reaction force data were filtered within the AnyBody environment using a 5 Hz and 12 Hz low-pass 2nd order Butterworth filter, respectively. The same filter parameters were used for OpenSim, while ground reaction force data were filtered in MATLAB (2012b) and the kinematics within OpenSim.

2.4. Muscle force estimation

Muscle forces were estimated using AnyBody (vers. 6.0, AnyBody Technology, Denmark) and OpenSim (vers. 3.2, OpenSim) (Fig. 1). Both programmes provide musculoskeletal models to analyse walking, the *Twente Lower Extremity Model* included into the *Mocap LowerBody model* (AMMR 1.6.2) for AnyBody and *Gait2392* for OpenSim (Delp, 1990; Delp et al., 1990). Models were mainly used in its standard settings, similar in number of segments, and set to the same degrees of freedom (DoF) (3 DoFs hip, 1 DoF knee, 2 DoF ankle). Both unscaled generic models had similar properties in the height and mass (both 1.80 m, 75.16 kg *Gait2392*, 75.46 kg *Twente Model*). Virtual markers were placed in both environments on the same representative anatomical landmarks.

Standard pipelines were used to scale the model and to estimate joint angles (inverse kinematics), joint moments (inverse dynamics) and muscle forces (static optimisation) in both modelling environments. To scale individual segments, AnyBody used a dynamic, while OpenSim used a static trial. In OpenSim, segments were scaled separately along the principal axis to fit to joint centres and in the perpendicular planes to fit landmark pairs as specified in Table S2 in the supplements (Trinler and Baker, 2018). In AnyBody, segments are first scaled in the longitudinal dimension and then scaled in width to obtain the specified mass (LengthMass scaling). Muscle tendon unit properties (for example maximum isometric forces) stayed the same across participants and, therefore, have not been scaled to the participant's anthropometrics.

For estimating muscle forces, static optimisation was solved in OpenSim by minimising the sum of muscle activations ($\sum$activactions^3). In AnyBody static optimisation was used to minimise the polynomial muscle recruitment criterion, defined as:

$$G = \sum_{i=1}^{n^{(M)}} \left(\frac{f_i^{(M)}}{N_i} \right)^3 \tag{1}$$

where $n^{(M)}$ is the number of muscles, $f_i^{(M)}$ is the respective muscle force, and N_i is equal to the isometric muscle strength in the simple muscle model. Furthermore, a constraint preventing individual muscle forces from exceeding their physiological maximum (Rasmussen et al., 2001) is included. The number of muscle actuators used differed for the two estimation approaches due to the differences in the musculoskeletal models, Twente Model: 55 muscles divided into 159 muscle-tendon actuators per leg; Gait2392: 36 muscles, divided into 46 muscle-tendon actuators per leg. No force-length-velocity model was applied to estimate muscle forces in order to reduce potentially affecting factors (Arnold et al., 2013; Carbone et al., 2016). Joint moments and joint compression forces were normalised to body mass and body weight, respectively.

The open source one-dimensional statistical parametric mapping (SPM) package (SPM1D, www.spm1d.org) was used to compare the joint angle, moment, muscle and joint force waveforms in Matlab (R2016b, 9.1.0.441655, The Mathworks Inc, Natick, MA). In general, the scalar test statistic, SPM{t} was computed at each point in the time series, forming a SPM (Pataky, 2012) with

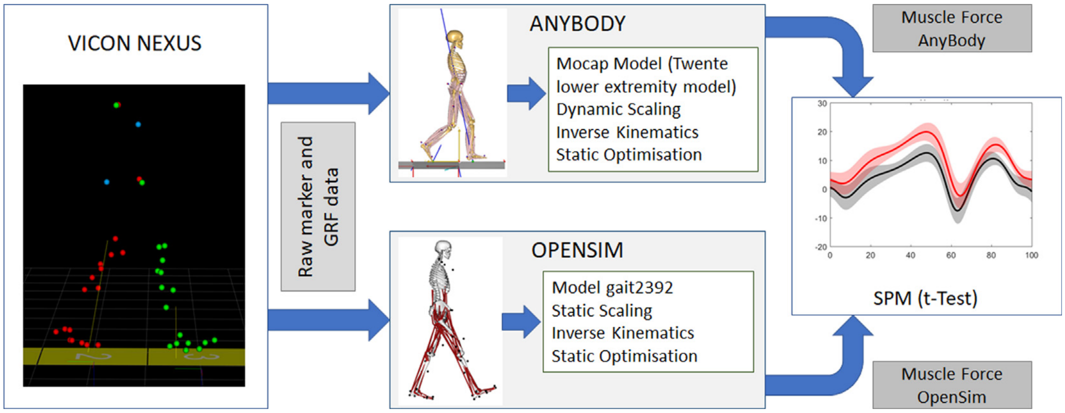

Fig. 1. Flowchart of data processing.

a critical threshold of 5% ($\alpha = 0.05$) (Friston et al., 2007; Adler and Taylor, 2007). The SPM test equivalent to a paired *t*-test was used. Individual muscle forces, however, where only analysed in case at least one of the simulations exceeded 5 N to avoid significant differences due to very small mean and standard deviation bands (i.e. variability). Summated muscles groups' activation threshold was, respectively, set to 5 N times the number of included muscles.

Furthermore, individual muscle actuator moment arms were plotted over a typical range of motion during one gait cycle for each participant. Also, the location of the joint centres in the global reference frame were calculated over the same gait cycle and the mean absolute error (MAE) between AnyBody and OpenSim over 101 data points (n) has been calculated to analyse the mean deviation:

$$\mathrm{MAE} = \left(\frac{1}{n} \sum_{i=1}^{n} |A1_i - A2_i| \right) \tag{2}$$

$A1_i$ represents the joint centre location at the ith time step of the gait cycle of AnyBody, $A2_i$ of OpenSim.

3. Results

Sagittal joint angles and moments are presented in Fig. 2. Shaded areas highlight statistical differences. Sagittal ankle and hip kinematics as well as hip internal rotation and ankle eversion show a steady offset between AnyBody and OpenSim throughout the gait cycle (GC), while sagittal knee flexion is mainly different in swing. Hip adduction differs during stance as OpenSim shows a higher adduction than AnyBody. Displacement of the joint centre locations in the global reference frame are presented in Table 1, while individual traces of the joint centre locations can be found in the supplements (Fig. S1a–i). Especially internal knee flexor and hip abductor moments differ between AnyBody and OpenSim for most of the time of the GC. Ankle and hip moments show small significant differences in stance but are mostly different in swing.

Fig. 3a–c shows muscle actuators' moment arms for the hip, knee and ankle muscles. Except for semitendinosus, semimembranosus (hip, knee) and biceps femoris longhead (hip), all actuators show visible differences out of the one standard deviation band. Gluteus medius during hip flexion has mostly negative values (behind hip joint centre) for OpenSim but positive values for AnyBody.

Muscles of the quadriceps femoris as well as semitendinosus, semimembranosus and tibialis anterior have no or nearly no statistical difference in the stance phase, while soleus, gastrocnemius lateralis, medialis, gluteus medialis and biceps femoris result in significant differences during stance. Significant differences in swing are shown for all muscles except soleus and rectus femoris (Fig. 4). Force summation leads for the hamstrings (semitendinosus, semimembranosus, biceps femoris long head (LH), short head (SH)) to significant differences in terminal stance and swing, while triceps surae summation (gastrocnemius lateralis, medialis and soleus) results in significant differences mostly in stance (Fig. 5).

The greatest differences in joint forces are shown during push off phase at the ankle (∼50% of the GC), while hip compression forces are higher with OpenSim but knee and ankle compression forces are higher with AnyBody (Fig. 6).

4. Discussion

This study aimed to compare lower limb muscle forces which have been estimated in AnyBody and OpenSim using static optimisation for ten healthy participants walking at self-selected speed. Such comparison between modelling environments has been lacking in the literature and is necessary to understand estimation outcomes and facilitate future use of force modelling to augment clinical gait analyses. Our results showed differences between approaches which highlight the importance of careful interpretation and raise the question about the applicability of such models for clinical gait analysis. However, considering the differences of both approaches (e.g. musculoskeletal model, scaling, estimation approach), estimated muscle forces are very similar in their activation pattern.

Compared to the literature (Sutherland, 2002), OpenSim shows on average more hip extension, which might be a consequence of a different definition of the pelvis to the ground (Au and Dunne, 2013). On the contrary, AnyBody shows a slight offset at the sagittal ankle angle compared to the literature (Sutherland, 2002). This is contradictory to the joint centre displacement calculation, which were about 0.7 cm more proximal at the knee and 0.5 cm more distal at the ankle for OpenSim than Anybody (see Table 1, Fig. S1a–i). It could imply greater plantar flexion and greater hip flexion with AnyBody compared to OpenSim. The result might suggest that either the joint centre locations are not exactly at the same anatomical point in the segments or the segment's anatomy (talus) is different, which might have also led to the differences in joint angles. Also, although we took care in placing the virtual model markers on the same anatomical positions, marker placement has been shown to be highly sensitive to the selection of model marker positions (Lund et al., 2015). Therefore, ankle kinematics might be influenced by virtual model marker placement, different constraints or definitions at the ankle (Lund et al.,

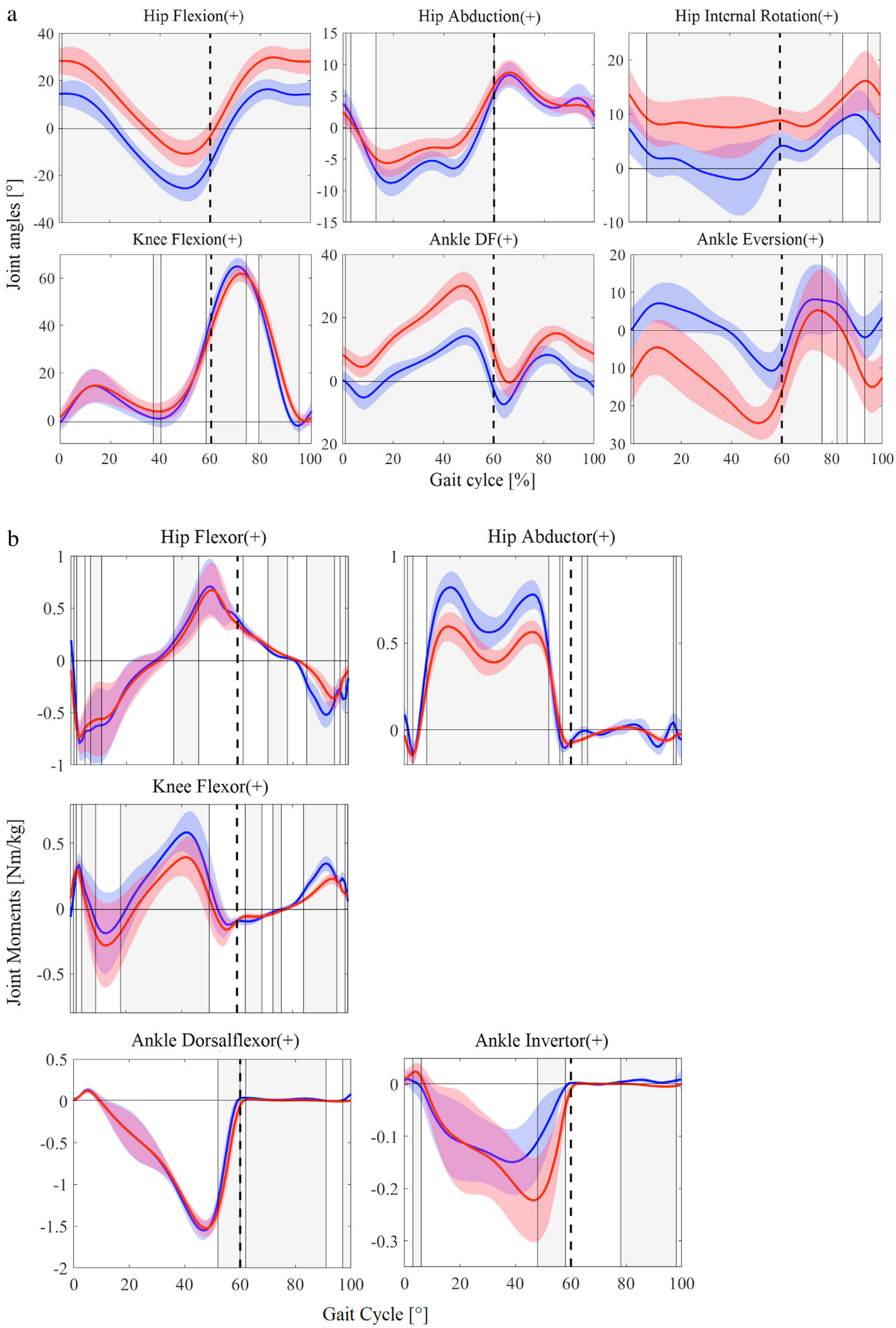

Fig. 2. (a, b). Mean and one standard deviation of sagittal joint angles (a) and internal joint moments (b) of AnyBody (red) and OpenSim (blue) averaged across ten participants. Shaded areas indicate statistical differences. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

Table 1
Mean absolute errors (MAE) between joint centre positions in x,y and z direction of the hip, knee and ankle during one walking trial of each participant. X-direction indicates anterior/posterior, y-direction, vertical, and z-direction medial/lateral displacement.

Participant	Hip			Knee			Ankle		
	x	y	z	x	y	z	x	y	z
P01	2.0 (1.5)	1.5 (0.5)	1.0 (0.5)	2.3 (2.4)	0.6 (0.7)	0.4 (0.4)	2.4 (3.0)	1.7 (1.0)	0.7 (0.3)
P02	2.3 (1.1)	1.4 (0.4)	0.8 (0.4)	1.9 (2.0)	0.7 (0.7)	0.3 (0.3)	1.6 (1.8)	0.8 (0.8)	0.8 (0.2)
P03	2.3 (1.8)	0.5 (0.3)	0.8 (0.2)	2.7 (2.8)	0.6 (0.6)	0.3 (0.3)	2.8 (3.7)	1.0 (1.1)	0.2 (0.2)
P04	1.7 (1.1)	0.2 (0.3)	0.7 (0.3)	1.8 (1.7)	0.5 (0.6)	0.4 (0.3)	1.5 (2.1)	1.4 (0.8)	0.5 (0.2)
P05	1.9 (1.4)	1.2 (0.3)	1.4 (0.4)	2.2 (2.3)	1.0 (0.6)	0.8 (0.5)	1.5 (2.2)	2.0 (0.9)	1.0 (0.3)
P06	2.2 (1.5)	0.6 (0.4)	1.1 (0.4)	2.4 (2.7)	0.7 (0.9)	0.8 (0.4)	2.4 (3.0)	1.2 (1.0)	0.4 (0.2)
P07	2.4 (1.7)	1.0 (0.4)	1.3 (0.2)	2.6 (2.9)	1.0 (0.8)	0.6 (0.5)	2.2 (3.1)	1.8 (1.0)	0.2 (0.2)
P08	2.0 (1.4)	0.3 (0.3)	0.9 (0.3)	3.0 (2.1)	0.7 (0.4)	0.4 (0.2)	2.3 (2.8)	1.3 (0.8)	0.3 (0.3)
P09	1.8 (1.3)	0.3 (0.3)	1.0 (0.2)	2.2 (2.0)	0.6 (0.4)	0.2 (0.1)	1.7 (2.2)	1.5 (0.9)	0.4 (0.1)
P10	1.6 (1.2)	0.2 (0.3)	1.0 (0.2)	1.7 (2.0)	0.8 (0.7)	0.4 (0.2)	1.4 (2.0)	1.1 (0.9)	0.4 (0.1)
Mean	**2.0 (0.3)**	**0.7 (0.5)**	**1.0 (0.2)**	**2.3 (0.4)**	**0.7 (0.2)**	**0.5 (0.2)**	**2.0 (0.5)**	**1.4 (0.4)**	**0.5 (0.3)**

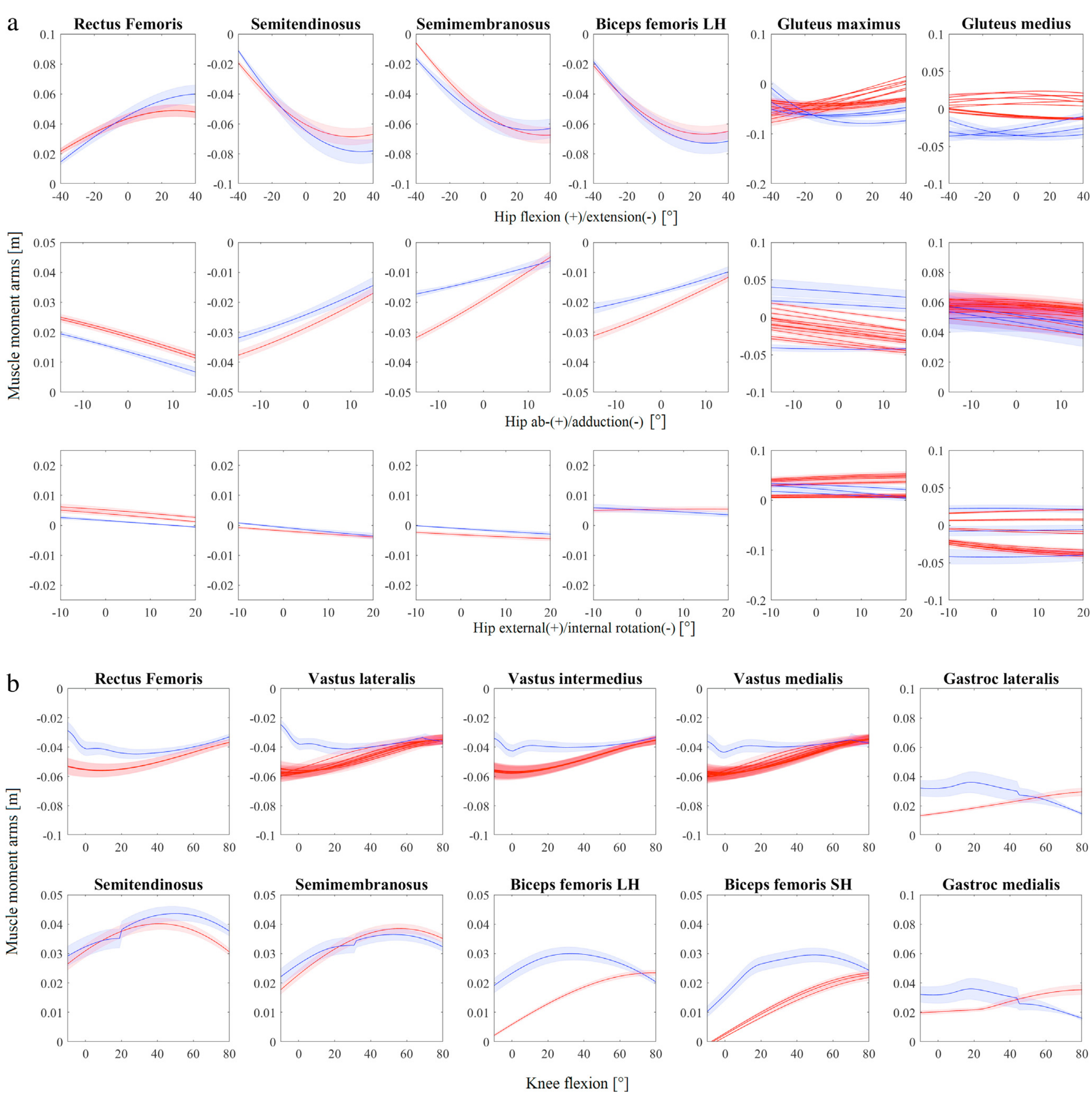

Fig. 3. (a–c). Mean and one standard deviation of moment arms of each muscle actuator of the hip (a), knee (b) and ankle (c) muscles for a typical range of motion during gait (AnyBody red, OpenSim blue). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

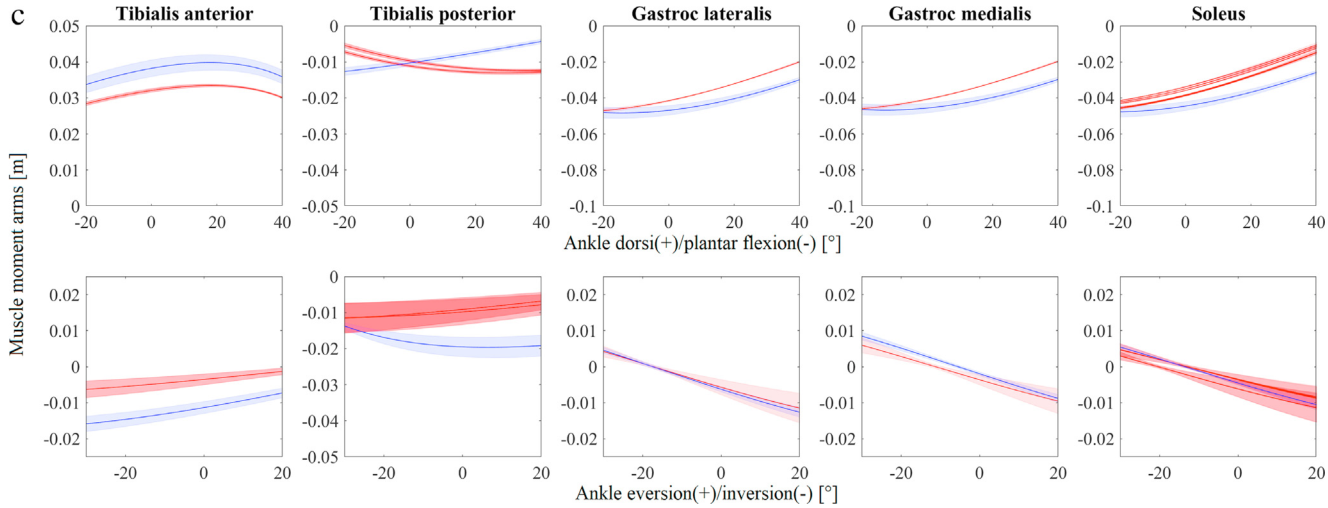

Fig. 3 (*continued*)

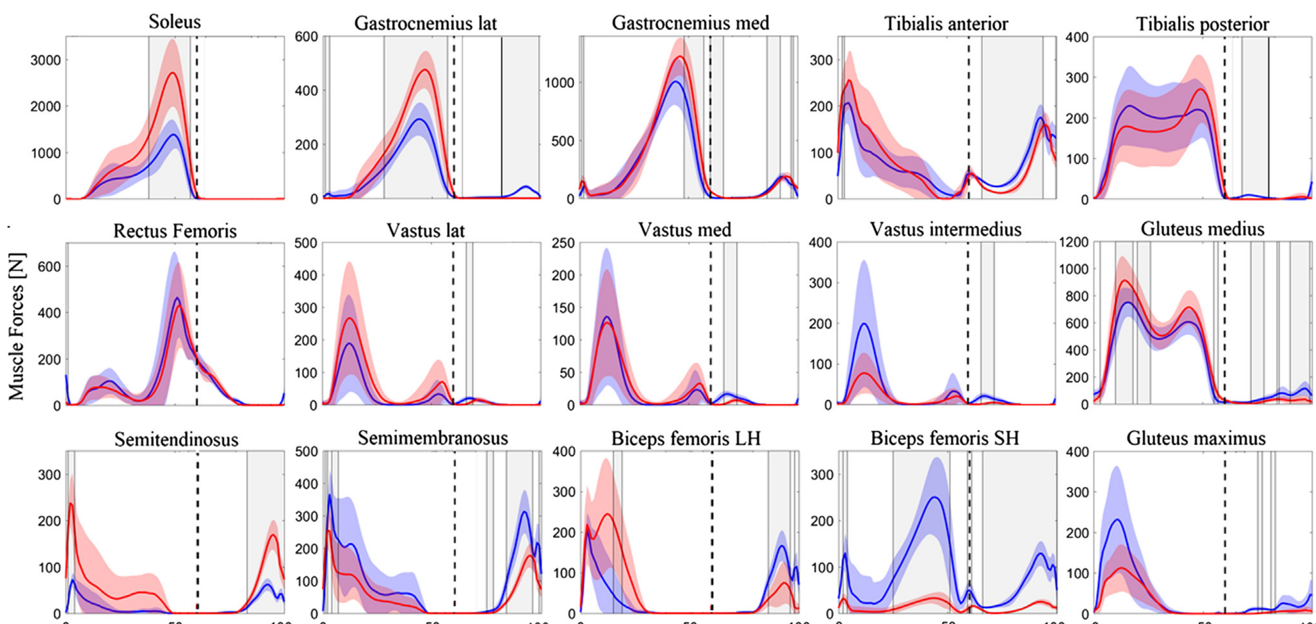

Fig. 4. Mean and one standard deviation of estimated muscle forces of AnyBody (red) and OpenSim (blue) averaged across ten participants. Shaded areas indicate statistical differences, activation threshold was set to 5 N. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

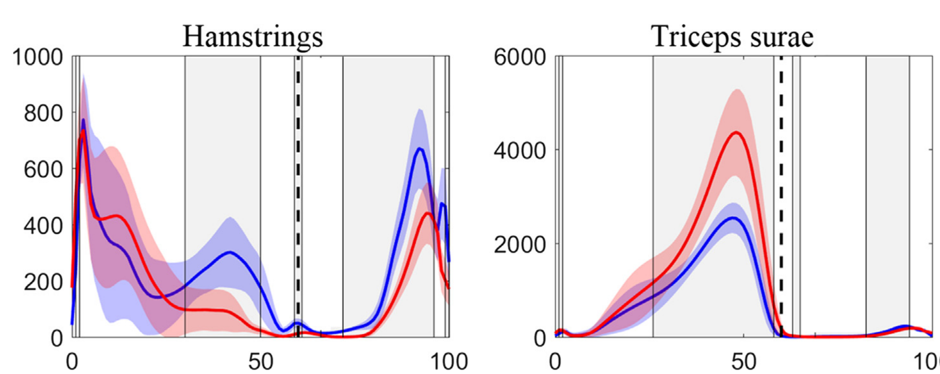

Fig. 5. Mean and one standard deviation of estimated muscle forces of the hamstrings (sum of semitendinosus, semimembranosus, biceps femoris long head and short head) and triceps surae (sum of gastrocnemius laterals, medialis and soleus) averaged across ten participants (AnyBody red, OpenSim blue). Shaded areas indicate statistical differences, activation threshold was set to 20 N and 15 N for the hamstrings and triceps surae, respectively. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

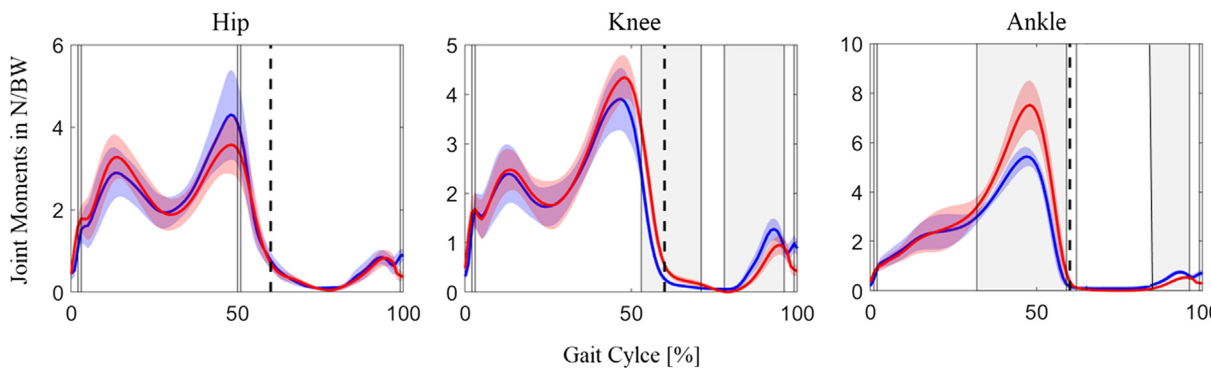

Fig. 6. Mean and one standard deviation of joint compression forces for the hip, knee and ankle (AnyBody red, OpenSim blue). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

2015), e.g. OpenSim uses a static trial for scaling, while a dynamic trial is used in AnyBody.

In comparison to the ankle and the hip, the kinematic differences at the knee are smaller. Nevertheless, AnyBody calculates a more flexed knee angle for a short time period around mid-stance as well as small but statistical significant differences in swing. This might be a result out of either kinematic differences at the ankle and the hip or the anatomical differences at the knee, as a small translation in dependence of the knee flexion angle (tibiofemoral contact point depends on the knee angle) is performed in the OpenSim gait2392 model (Yamaguchi and Zajac, 1989; Delp et al., 1990).

Sagittal joint moments are similar for the ankle and the hip instance, indicating similar overall produced muscle forces for these time frames. The knee moment, however, shows a shift in stance between AnyBody and OpenSim as well as an increased internal knee extensor moment in loading response for AnyBody. Due to kinematic differences, the position of the ground reaction force vector to the knee joint centre might be different and produced, therefore, a higher external flexion moment within the model in AnyBody compared to the model in OpenSim. Joint moment differences between approaches are also shown in swing. Those differences are significant due to the moments' mean being close to zero while moments' standard deviation bands (i.e. variability) are similar between approaches. Knee and hip joint moment differences in terminal swing are most likely caused by different segmental properties (e.g., mass and inertia (Brunner and Rutz, 2013; Krabbe et al., 1997)) of both anatomical models. Since joint moment calculations are not influenced by muscle forces in both modelling environments, differences in swing phase have not been interpreted. Nevertheless, different segment properties, as well as different joint centre definition which possibly cause different joint moment calculations, also have an influence on muscle force calculations.

Variations in kinematics as well as visual differences in moment arms most certainly cause the presented differences for some of the muscle forces. All muscles of the triceps surae show a much higher peak force during stance in AnyBody compared to OpenSim. This might be a result of the more dorsiflexed ankle, as higher force is needed to move the foot in the opposite direction (i.e. plantarflexion). In addition, the knee tends to be more flexed in AnyBody during terminal-stance which might favour the triceps surae over the biceps femoris SH to control the flexion of the knee in preparation for swing. This is in agreement with much higher biceps femoris SH forces using OpenSim compared to AnyBody. The higher biceps femoris SH force in OpenSim could additionally be due to working as an antagonist of the vastii muscles. The vastii muscle forces are similar between AnyBody and OpenSim but not the vastii moment arms which tend to be longer for OpenSim than AnyBody. To account for this imbalance, biceps femoris SH might hold against the vastii group additionally with a longer moment arm compared to the results in AnyBody.

Higher muscle forces observed for the triceps surae muscle group in AnyBody led to higher ankle joint compression forces. Furthermore, visual differences in the ankle-joint-spanning-muscles' moment arms could be observed between the two modelling approaches. Moment arms for muscles of the OpenSim musculoskeletal model were higher than the model in AnyBody which confirms the higher muscles forces of the triceps surae in AnyBody.

The difference found for triceps surae has no influence on tibialis posterior and the antagonist tibialis anterior, where no differences between AnyBody and OpenSim are observed during stance. Tibialis anterior is responsible for a controlled eccentric lowering of the foot to the ground after initial contact (Winter, 1990). Although the plantarflexion/dorsiflexion kinematics show an offset between AnyBody and OpenSim, the foot still needs to be lowered the same way to the ground. Due to the small mass of the foot, changes in kinematics might have only a small influence on the eccentric work of the tibialis anterior during stance.

Except for gluteus maximus (hip ab-/adduction) and medius (flexion/extension, rotation) all muscles show actuator moment arms revealing the same function. Both muscles take over different movements. The ventral part of gluteus medius supports hip flexion while the dorsal parts of gluteus medius support hip extension. In AnyBody the flexion and extension support of gluteus medius are both visible, while in OpenSim only the extension support is presented. Differences in gluteus medius moment arms might be the reason for observed differences of gluteus medius muscle forces in mid stance.

Nearly no differences are observed for the muscles of the quadriceps femoris (rectus femoris, vastus lateralis, vastus medialis, vastus intermedius), as well as for the other muscles of the hamstrings (semitendinosus, semimembranosus, biceps femoris LH). Interestingly, all muscle moment arms of the quadriceps femoris muscles are longer in AnyBody than OpenSim, especially between 0° and 20° of knee flexion. This is the opposite with biceps femoris long and short head. However, altogether, muscle forces resulted in a comparable knee compression force.

Vastii are only spanning the knee joint, whereas all other muscles of the hamstrings (except biceps femoris SH) and rectus femoris additionally cross the hip. Although hip flexion/extension shows an offset between AnyBody and OpenSim, it, has no great effect on these biarticular muscles in stance. Rectus femoris is mostly active at the end of stance to transition of swing for controlling knee flexion (Nene et al., 2004). The hamstrings group (except biceps femoris SH) is active at the beginning and at the end of the GC to serve as hip extensors at initial contact and controllers for knee flexion to slow down leg and foot at the end of swing

(Winter, 1990). Therefore, only the hamstrings at initial contact might be affected through hip flexion, which could explain a small statistical difference directly at heel contact.

It was hypothesised that distinctive differences can be detected between individual muscle forces, but that the summation of synergists would result in similar muscle forces with higher agreement between AnyBody and OpenSim. The hypothesis that muscle force summations result in higher agreement between both modelling environments had to be rejected. The significant difference for the hamstrings in stance between AnyBody and OpenSim occurs due to the high difference shown for the single muscle force of biceps femoris SH.

Altogether, presented results are due to anatomical differences of the two musculoskeletal models as well the scaling approach, which leads to differences in the moment arms and, therefore, muscle forces (Kainz et al., 2017). This shows the high importance of exact individual musculoskeletal information for muscle force estimation, which generic models do not provide (e.g., Wagner et al., 2013). Subject-specific musculoskeletal models take additional individual musculoskeletal anatomy into account, mostly derived through MRI measurements (Valente et al., 2014). However, to-date, MRI- data are expensive and not always available. Another idea is to calculate subject-specific musculoskeletal parameters using a series of experimental functional calibration tasks (e.g., individual isometric and isokinetic contractions) to scale the model (Garner and Pandy, 2003; Lund et al., 2015). Although being time consuming due to multiple tests, the approach has been shown to be a good alternative compared to generic musculoskeletal models (Wu et al., 2016). Nevertheless, accurately applied, the results of this study show that generic models are useful for certain analyses especially for activation patterns and can be compared between approaches.

5. Conclusion

This study shows good agreement between AnyBody and OpenSim for muscle activation patterns, which shows the robustness for such approaches. However, it displays substantial differences in some phases of the GC regarding muscle force estimations between two standard musculoskeletal modelling approaches offered in two different modelling environments, even when summing up synergists of some of the muscle groups (e.g., triceps surae). Understanding these differences and their cause (e.g., sensitivity to model marker placement, different joint definitions, scaling) as well as the limitations of mathematical approaches to estimate muscle forces will help practitioners to operate such modelling environments, to analyse the results and to compare these to other movement laboratories. Crucial points to consider are the synergistic muscle force production when analysing the results, the different anatomical definitions (e.g. pelvis definition to the ground, mass and inertia of the segments), differences in calculated joint centres (e.g., resulted through different scaling approaches, model marker placement) and segmental interactions of the musculoskeletal models. Further analysis should focus on implementing functional calibration tasks while gradually changing experimental data (e.g. different walking speeds to reflect normal and pathological gait), which are used for estimating muscle forces to gain a better understanding of variations and divergent behaviour between approaches.

Conflict of interest statement

All authors had no financial and personal relationships with other people or organisations that could inappropriately influence (bias) this work.

Appendix A. Supplementary material

Supplementary data to this article can be found online at https://doi.org/10.1016/j.jbiomech.2019.01.045.

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