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# OpenSim Versus Human Body Model: A Comparison Study for the Lower Limbs During Gait

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Musculoskeletal modeling and simulations have become popular tools for analyzing human movements. However, end users are often not aware of underlying modeling and computational assumptions. This study investigates how these assumptions affect biomechanical gait analysis outcomes performed with Human Body Model and the OpenSim gait2392 model. The authors compared joint kinematics, kinetics, and muscle forces resulting from processing data from 7 healthy adults with both models. Although outcome variables had similar patterns, there were statistically significant differences in joint kinematics (maximal difference: 9.8° [1.5°] in sagittal plane hip rotation), kinetics (maximal difference: 0.36 [0.10] N·m/kg in sagittal plane hip moment), and muscle forces (maximal difference: 8.51 [1.80] N/kg for psoas). These differences might be explained by differences in hip and knee joint center locations up to 2.4 (0.5) and 1.9 (0.2) cm in the posteroanterior and inferosuperior directions, respectively, and by the offset in pelvic reference frames of about 10° around the mediolateral axis. The choice of model may not influence the conclusions in clinical settings, where the focus is on interpreting deviations from the reference data, but it will affect the conclusions of mechanical analyses in which the goal is to obtain accurate estimates of kinematics and loading.

***Keywords:*** biomechanics, musculoskeletal modeling, simulation, static optimization

Musculoskeletal models for biomechanical simulations have become increasingly popular to analyze human movement. In addition to joint kinematics and kinetics, musculoskeletal models enable researchers and clinicians to assess other biomechanical variables, such as muscle lengths and forces. Different software systems were developed for modeling and analyzing human movement (eg, AnyBody,[1] OpenSim,[2] and Human Body Model[3]), and there is an increasingly large body of literature reporting analyses of motion based on these software systems. OpenSim offers several musculoskeletal models with varying complexity (eg, number of muscles and kinematic degrees of freedom [DOFs]), therefore giving users multiple choices for their study. Roelker et al[4] recently provided valuable information about which OpenSim model to use for studying gait by investigating the effects of using different models on joint kinematics, kinetics, and muscle function. They reported that differences between models were mainly due to different coordinate system definitions and muscle parameters and concluded that the gait2392 model is sufficiently complex to study gait in healthy adults. When interpreting differences in results obtained with different software systems, the added challenge is that discrepancies might result from differences between data processing workflows besides differences between models. To our knowledge, no studies have assessed differences in joint kinematics, kinetics, and muscle forces induced by the use of different models in different software systems. In this study, we compared the clinically-oriented Human Body Model with the research-oriented OpenSim gait2392 model.[5] The goals of this comparison were (1) to evaluate how the model and computational choices influence joint kinematics, kinetics, and muscle forces resulting from processing the same experimental gait data and (2) to relate the outcome differences to the underlying modeling and computational assumptions.

## Methods

Seven healthy adults (3 females and 4 males, age: 30.7 [6.1] y, height: 176.7 [7.1] cm, and weight: 69.4 [6.4] kg) gave informed consent to participate in the study approved by the ethics committee at UZ Leuven (Leuven, Belgium). Each subject was instrumented with 22 retroreflective skin-mounted markers, corresponding to the Human Body Model marker set, excluding arms, head, and torso.[3] Three-dimensional marker coordinates were recorded (100 Hz) using a 10-camera motion capture system (Vicon, Oxford, UK). Ground reaction forces were recorded (1000 Hz) using 2 force plates (AMTI, Watertown, MA). The subjects were instructed to walk at a self-selected speed.

The experimental data were processed with OpenSim 3.3 using the gait2392 model, later referred to as the OpenSim model, and with the Gait Offline Analysis Tool 3.3 (Motekforce Link B.V., Amsterdam, The Netherlands) that integrates Human Body Model. The metatarsophalangeal joints of the OpenSim model were locked so that both models had 21 similar DOFs actuated by 43 muscles per leg. Marker information from a standing calibration trial was used to scale the OpenSim model to the subjects' anthropometry using OpenSim's Scale tool (see Tables S1 and S2 in Supplementary Material [available online] for the marker pairs used to scale the segments' dimensions and for the marker weights used to fit the model's pose to the standing calibration pose, respectively) and to initialize a new model in Human Body Model.[3]

The processing pipeline with both systems consisted of inverse kinematics, kinematic filtering, inverse dynamics, and static optimization. The same weighted least squares problem (see Table S3 in Supplementary Material [available online] for the marker weights) was solved with both systems during inverse kinematics. Details about the different optimization algorithms can be found in Supplementary Material (available online). The resulting root mean square (RMS) and maximum marker errors between modeled and measured marker positions were compared using a Wilcoxon signed-rank statistical analysis. Since OpenSim's effective dual-pass filter cutoff frequency is lower than the user-specified cutoff frequency,[6] a scaling factor was applied to match Human Body Model's effective 6-Hz cutoff frequency when filtering the kinematics and the ground reaction forces (more details in Supplementary Material [available online]). Human Body Model is real time and induces a 37-ms time delay when filtering the kinematics.[3] This delay was corrected when comparing the results.

Different static optimization formulations are available in both systems. Human Body Model enables scaling muscle activity by muscle volume in the objective function (default setting),[3,7] whereas OpenSim enables considering the muscles as ideal force generators or constraining them by their force–length–velocity properties[8,9] (more details in Supplementary Material [available online]). All formulations were tested to investigate their impact on the muscle force estimation. Similar optimization problems are solved in OpenSim and Human Body Model when the muscles are considered as ideal force generators and when muscle activity is not scaled by muscle volume. However, OpenSim enables the use of reserve actuators, whereas Human Body Model does not use upper bounds on muscle activations to guarantee the feasibility of the optimization problem, and both systems use different optimization algorithms (more details in Supplementary Material [available online]). Both models use identical values for maximal isometric muscle forces to relate muscle activations to muscle forces, but there are small differences in moment arms. Human Body Model uses polynomial functions of the joint angles, whereas OpenSim uses muscle-tendon paths (line segments between muscle points defined in segmental reference frames) to compute moment arms.[3] Human Body Model's polynomials are defined such that the moment arms computed based on these polynomials match the OpenSim moment arms within 2 mm for the generic model. Moment arms do not depend on subject size in Human Body Model but are influenced by scaling in OpenSim.

Since the number of gait trials with valid force plate contacts was unevenly divided among subjects, we selected one representative trial for each leg of each subject based on the kinematic errors. We considered each leg apart to increase the size of the data set. Asymmetry between both legs may exist,[10] contributing to the variability in our data. The representative trial was the trial with the RMS inverse kinematic marker error that best matched the error averaged over all trials.[11] This resulted in 14 trials (stride duration: 1.05 [0.06] s) that were used for further analysis. Joint kinematics, kinetics, and muscle forces were time-normalized to the gait cycle duration and averaged over the 14 representative trials. Biomechanical outcomes resulting from the different models and static optimization formulations were analyzed using nonparametric paired $t$ tests with the 1-dimensional statistical parametric mapping package.[12,13] The level of significance was set to $P < .05$.

To evaluate joint center location differences between models, we calculated the transformations between the corresponding segment reference frames that best mapped the OpenSim model markers to the corresponding Human Body Model markers in a least squares sense. We then used these transformations to express the OpenSim model joint centers in the corresponding Human Body Model reference frames and computed the distance between the joint centers of both models. To evaluate pelvic reference frame differences, we similarly calculated the transformation between pelvic reference frames and expressed the difference in orientation in Euler angles (sequence of rotation axes: mediolateral, inferosuperior, and posteroanterior).

## Results

Differences in joint kinematics were found when processing the same experimental gait data with the OpenSim model and with Human Body Model. Joint kinematics showed similar patterns but statistically differed for all DOFs (maximal statistical differences: 9.8° [1.5°], 5.5° [1.0°], 8.5° [3.6°], 5.0° [1.0°], 6.5° [1.5°], and 15.6° [6.2°] for the sagittal hip, frontal hip, transversal hip, sagittal knee, sagittal ankle, and subtalar rotations, respectively) during large intervals ranging from 33% (sagittal ankle rotation) to 100% (sagittal hip rotation) of the gait cycle. An offset in sagittal hip rotation (flexion/extension) was observed (Figure 1). After scaling in OpenSim, the RMS marker error (1.2 [0.1] cm) and maximal marker error (2.2 [0.2] cm) of the markers corresponding to anatomical landmarks were close to OpenSim's recommendations[14] (smaller than 1 and 2 cm, respectively) and had a low sensitivity to user inputs (marker pairs and weights used for scaling; see Table S4 in Supplementary Material [available online]). RMS and maximum marker errors after inverse kinematics were statistically smaller ($P < .001$) with Human Body Model (0.5 [0.1] and 1.1 [0.3] cm, respectively) than with the OpenSim model (0.7 [0.1] and 1.6 [0.4] cm, respectively). Marker errors met OpenSim's best practices[14] (RMS marker error smaller than 2 cm and maximum marker error smaller than 2–4 cm) for both models and had a low sensitivity to user inputs (marker pairs and weights used for scaling and marker weights used for inverse kinematics) in OpenSim (see Table S5 in Supplementary Material [available online]).

Differences in joint kinetics were found between the OpenSim model and Human Body Model. Joint moments showed similar patterns but statistically differed during several intervals of the gait cycle for all DOFs (maximal statistical differences: 0.36 [0.10] N·m/kg, 0.21 [0.03] N·m/kg, 0.09 [0.02] N·m/kg, 0.18 [0.04] N·m/kg, 0.18 [0.03] N·m/kg, and 0.25 [0.11] N·m/kg for the sagittal hip, frontal hip, transversal hip, sagittal knee, sagittal ankle, and subtalar moments, respectively; Figure 2).

Differences in muscle forces were found between the OpenSim model and Human Body Model. Muscle forces computed using similar static optimization formulations showed similar patterns but statistically differed during several intervals of the gait cycle for most muscles (Figure 3 and Figures S1–S4 in Supplementary Material [available online]). The largest statistical differences were observed for the psoas (8.51 [1.80] N/kg), soleus (8.11 [1.30] N/kg), and peroneus longus (6.50 [2.59] N/kg). Maximum absolute reserve actuators were smaller than 4.0e–4 N·m/kg in OpenSim, which met the requirements advocated by Hicks et al.[15] In Human Body Model, muscle activations exceeded 1 (maximum 1.1) for the psoas in 4 out of 14 trials during small intervals of the gait cycle (<5%). Differences in modeling muscle function and performance criteria had an effect on the estimated muscle forces. Constraining the muscles by their force–length–velocity properties in OpenSim induced statistical differences for most muscles (Figures S5–S8 in Supplementary Material [available online]),

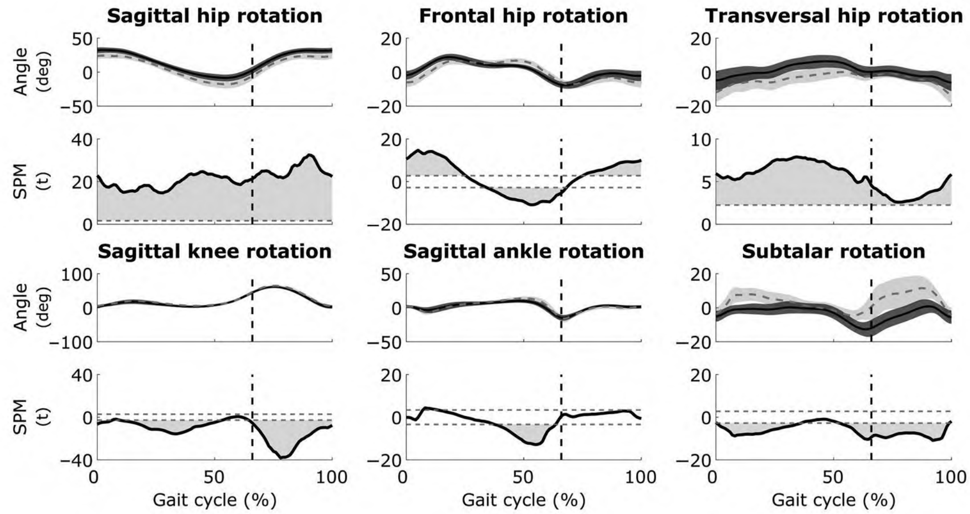


**Figure 1** — (First and third rows) Comparison of joint kinematics calculated with the OpenSim model (dashed gray line) and Human Body Model (black). (Second and fourth rows) Results from the statistical analysis using nonparametric paired *t* tests in SPM1D. Gray-shaded areas above and below the gray dashed lines indicate significant differences. The vertical black dashed line indicates the transition from stance to swing. SPM1D indicates 1-dimensional statistical parametric mapping.

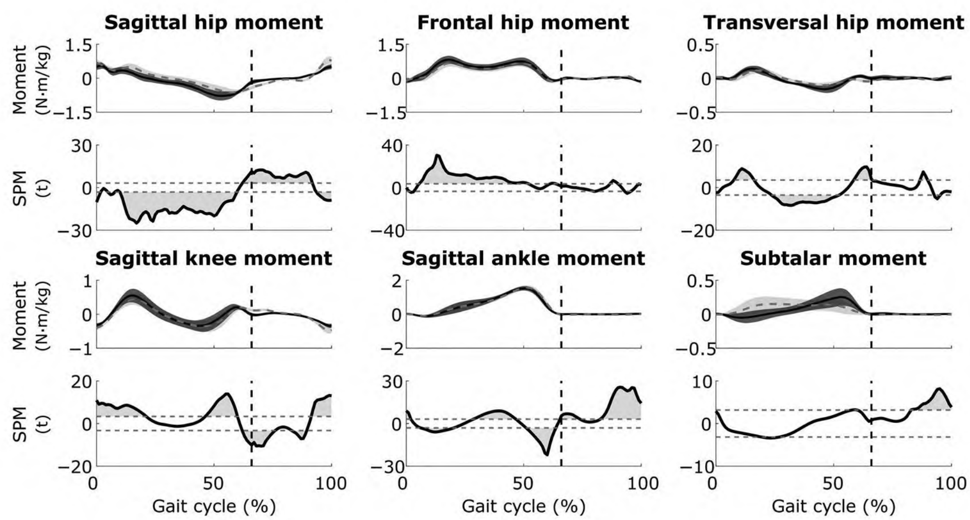


**Figure 2** — (First and third rows) Comparison of joint kinetics calculated with the OpenSim model (dashed gray line) and Human Body Model (black). (Second and fourth rows) Results from the statistical analysis using nonparametric paired *t* tests in SPM1D. Gray-shaded areas above and below the gray dashed lines indicate significant differences. The vertical black dashed line indicates the transition from stance to swing. SPM1D indicates 1-dimensional statistical parametric mapping.

although overall, the impact was relatively limited. The soleus, tibialis posterior, and medial gastrocnemius showed the largest statistical differences (4.27 [1.28] N/kg, 3.84 [2.76] N/kg, and 3.73 [1.21] N/kg, respectively). Scaling muscle activity by muscle volume in the Human Body Model static optimization objective function had a more pronounced influence as the contribution of smaller muscles increased at the expense of the larger muscles (Figures S9–S12 in Supplementary Material [available online]). In particular, we observed a statistical decrease in force for large muscles, including the psoas, gluteus maximus 2 (medial part), and soleus (maximal statistical differences: 6.01 [0.85] N/kg, 3.00 [0.48] N/kg, and 2.40 [0.73] N/kg, respectively) and a statistical increase in force for small muscles, including the piriformis, gluteus minimus 3 (posterior part), and tensor fasciae latae (maximal statistical differences: 3.35 [0.58] N/kg, 2.10 [0.40] N/kg, and 1.48 [0.31] N/kg, respectively).

Definitions of reference frames and joint centers differed between the OpenSim model and Human Body Model. First, the pelvic reference frames had different orientations, as calculated through the Euler angles (Table 1). The largest difference was on average 10.2° about the mediolateral axis. Second, the hip joint centers had different locations (Table 2). In particular, the hip joint center was on average 2.4 cm more anterior in Human Body Model compared with the OpenSim model. Third, the tibia

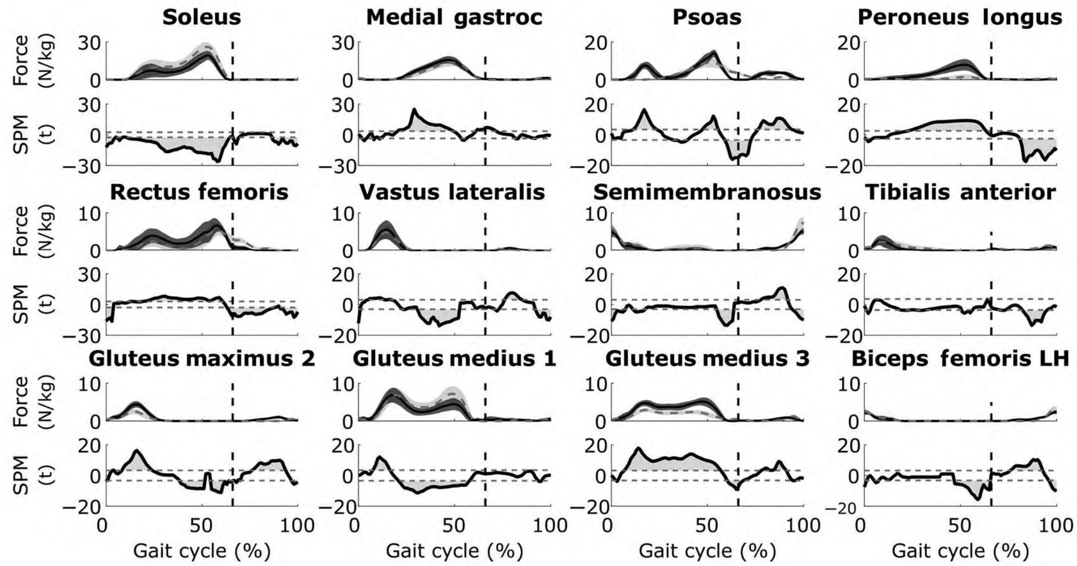


**Figure 3** — (First, third, and fifth rows) Comparison of muscle forces estimated with the OpenSim model (dashed gray line) and Human Body Model (black). Muscle forces estimated without taking force–length–velocity properties into account (OpenSim) and without scaling muscle activity by muscle volume in the static optimization objective function (Human Body Model). (Second, fourth, and sixth rows) Results from the statistical analysis using nonparametric paired *t* tests in SPM1D. Gray-shaded areas above and below the gray dashed lines indicate significant differences. The vertical black dashed line indicates the transition from stance to swing. See Figures S1–S4 in the Supplementary Material (available online) for other muscles. LH indicates Long Head and gastroc states for gastrocnemius; SPM1D, 1-dimensional statistical parametric mapping.

**Table 1 Differences in Pelvic Reference Frame Orientation Between the OpenSim Model and Human Body Model Evaluated Through Euler Angles (in Degrees)**

| | Subjects | | | | | | | |
|---|---|---|---|---|---|---|---|---|
| **Rotation axes** | **1** | **2** | **3** | **4** | **5** | **6** | **7** | **Mean (SD)** |
| Mediolateral | 10.1 | 10.3 | 10.5 | 10.8 | 9.2 | 9.3 | 11.1 | 10.2 (0.7) |
| Inferosuperior | –1.1 | 0.8 | 0.1 | –0.3 | 1.0 | –0.4 | –2.3 | –0.3 (1.1) |
| Posteroanterior | –0.6 | –1.0 | 1.3 | –0.3 | 0.2 | –1.4 | –0.2 | –0.3 (0.9) |

Note: Euler angles in degrees, sequence of rotation axes: mediolateral, inferosuperior, posteroanterior, describing the orientation of the pelvic reference frame of the OpenSim model with respect to the pelvic reference frame of Human Body Model.

**Table 2 Differences in Right Hip Joint Center Location Between the OpenSim Model and Human Body Model (in Centimeters)**

| | Subjects | | | | | | | |
|---|---|---|---|---|---|---|---|---|
| **Axes** | **1** | **2** | **3** | **4** | **5** | **6** | **7** | **Mean (SD)** |
| Posteroanterior | 2.2 | 2.3 | 2.5 | 1.9 | 2.9 | 2.7 | 2.5 | 2.4 (0.3) |
| Inferosuperior | 1.2 | 1.6 | 1.0 | 0.6 | 1.4 | 1.7 | 2.8 | 1.5 (0.7) |
| Mediolateral | 1.0 | 0.7 | 1.3 | 0.9 | 0.8 | 0.8 | 1.5 | 1.0 (0.3) |

Note: Differences in centimeters between the Human Body Model right hip joint center location and the OpenSim model right hip joint center location expressed in the Human Body Model pelvic reference frame. Positive results indicate a more anterior/superior/lateral location in Human Body Model compared with the OpenSim model.

origins, defining the position of the knee joint centers, had different locations in both models (Table 3). In particular, the tibia origin was on average 1.9 cm more superior in the OpenSim model compared with Human Body Model. Finally, the subtalar axis was defined differently in both models. The subtalar axes in Human Body Model and in the OpenSim model are inclined by 42° and 37°, respectively, from the transversal plane and deviate medially by –23° and –9°, respectively, from the sagittal plane.[16]

## Discussion

The primary goals of this study were to compare the OpenSim gait2392 model with Human Body Model based on joint kinematics, kinetics, and muscle forces calculated during gait for healthy adults and to relate the outcome differences to the modeling and computational assumptions. Overall, outcome variables had similar patterns across models, but they statistically differed in large intervals of the gait cycle.

**Table 3 Differences in Right Tibia Coordinate Frame Origin Location Between the OpenSim Model and Human Body Model (in Centimeters)**

| | Subjects | | | | | | | |
|---|---|---|---|---|---|---|---|---|
| **Axes** | **1** | **2** | **3** | **4** | **5** | **6** | **7** | **Mean (SD)** |
| Posteroanterior | 0.1 | 0.3 | 0.4 | 0.2 | 0.3 | 0.2 | 0.5 | 0.3 (0.1) |
| Inferosuperior | −2.3 | −1.8 | −2.2 | −1.7 | −1.9 | −2.0 | −1.8 | −1.9 (0.2) |
| Mediolateral | 1.5 | 0.6 | 0.6 | 1.0 | 0.1 | 1.4 | 0.7 | 0.9 (0.5) |

Note: Differences in centimeters between the Human Body Model right tibia coordinate frame origin and the OpenSim model right tibia coordinate frame origin expressed in Human Body Model femur reference frame. Positive results indicate a more anterior/superior/lateral location in the Human Body Model compared with the OpenSim model.

OpenSim and Human Body Model generate different kinematic models. In particular, we observed large differences in hip and knee joint center locations. Human Body Model estimates the hip joint center locations based on pelvic width and depth using Harrington equations.[17] In OpenSim, the hip joint center locations are scaled with the pelvis. In this study, the hip joint center locations from the generic OpenSim model were scaled in the mediolateral direction with pelvic width and in the inferosuperior and posteroanterior directions with pelvic depth. Kainz et al[18] found that Harrington equations are more accurate than other regression equations, but they suggest the use of functional methods, such as geometric sphere fitting methods,[19,20] in people with sufficient active hip range of motion, such as the subjects in this study. More accurate methods to define subject-specific kinematic models[21] have not been integrated in existing software and are not widely adopted. The OpenSim model and Human Body Model also rely on different joint axis definitions. First, in Human Body Model, the subtalar axis is defined based on the average subtalar joint from Isman and Inman,[22] whereas the OpenSim model subtalar axis is derived from Inman[23] and is in the experimental range of values (20° to 68° and −47° to −4° for the horizontal inclination and the medial deviation, respectively) obtained from cadaver measurements.[22] Second, the OpenSim model uses a moving knee flexion axis[24] to account for the translation of the tibiofemoral joint in the sagittal plane, whereas Human Body Model uses a fixed axis. Finally, there is a large offset between the pelvic reference frames (rotation about the mediolateral axis) in both models. It is worth mentioning that Roelker et al[4] also reported differences in pelvic neutral position definition between different OpenSim models. This suggests, along with the findings of this study, that this modeling feature is highly variable across existing musculoskeletal models. The differences in the pelvic reference frame orientation cause the observed offset in sagittal hip rotation. In combination with the different hip joint center locations, the different pelvic reference frames also explain the different hip rotations in the frontal and transversal planes. The different hip and knee joint center locations and subtalar axis definitions can explain the differences in knee and subtalar rotations. We expect the computational choices (eg, optimization algorithms and stopping criteria) related to the approaches used for solving inverse kinematics in OpenSim and Human Body Model to have contributed to a lesser extent to the differences in kinematic results than the joint definition differences. Given that OpenSim and Human Body Model use the same initial guesses for the optimization algorithms and that Human Body Model allows a relatively long computational time to solve the inverse kinematic optimization problem, we do not think that either converging to different local optima or not achieving convergence contributed to the observed differences in kinematics.

The filter used to process the inverse kinematic results has a different order in OpenSim (third order) and in Human Body Model (second order). The users have no access to this computational feature, nor through the graphical user interfaces of both software systems, nor through to the application programming interface of OpenSim. We therefore choose to present results obtained with the built-in filters since we expect that most users will perform their entire data processing with either OpenSim or Human Body Model. However, we evaluated the impact of using a second-order filter versus a third-order filter by processing the OpenSim inverse kinematic results of 1 trial outside the OpenSim platform before performing inverse dynamics and static optimization. The largest differences in joint moments and muscle forces were 0.06 N·m/kg for the sagittal hip moment and 0.46 N/kg for the rectus femoris, respectively. As a general limitation of this study, due to the limited flexibility of the Human Body Model and OpenSim platforms, we were unable to investigate the influence of each individual modeling and computational choice on the results. As a result, we could only outline important differences in underlying modeling and data processing assumptions without quantifying their relative contributions.

Joint kinematic differences directly affect the joint moments. Other factors, such as different inertial properties[25] and different joint definitions, also play a role. The different joint definitions will result in different locations and orientations of the joint centers and axes in space after inverse kinematics, and hence, the forces and moments applied in the joints to counteract the ground reaction forces and gravity will differ. In particular, we have studied the sensitivity of the joint moments to the knee flexion axis (moving vs fixed) in OpenSim and observed statistical differences (maximal statistical difference: 0.05 [0.01] N·m/kg for the knee; Figure S13 in Supplementary Material [available online]).

Differences in muscle forces result from differences in joint kinematics and kinetics as well as from differences in moment arms. These differences in moment arms are due to the different computation of moment arms in both models, the different joint kinematics that are inputs to this computation, and the influence of the subject size that is taken into account in OpenSim but not in Human Body Model. Differences in joint kinematics between the OpenSim model and Human Body Model induced differences in moment arms up to 1.5 cm (quadratus femoris for hip flexion). Differences in moment arms between the smallest (height: 169 cm) and the tallest (height: 190 cm) subjects were up to 0.9 cm (gluteus maximus 3 [posterior part] for hip flexion) in the anatomical position. In Human Body Model, psoas muscle activations exceeded 1, suggesting an unrealistic muscle force distribution. It was more optimal to activate the psoas above 1 than to increase the contribution of another muscle (eg, rectus femoris). Muscle activations exceeding 1 were dependent on the static optimization objective function. In more detail, piriformis muscle activations exceeded 1 (maximum 1.1) for 2 out of 14 trials during small intervals of the gait cycle (<7%) when scaling muscle activity by muscle volume in the static optimization objective function. This underlines the importance of the criterion used to solve the muscle redundancy problem. However, it is to be mentioned that muscle activations will also depend on the muscle–tendon parameters, which appear in the objective function. We expect that more representative muscle–tendon parameters will result in muscle activations smaller than 1 during gait for both objective functions. Overall, activations larger than 1 are not physiological and should be identified as a limitation of the model. Finally, no experimental muscle activations (electromyography) were available to further validate the static optimization results, which is a limitation of this study.

Modeling assumptions affect the estimation of muscle forces. In particular, the sensitivity of the results to the choice of the objective function was underlined by the differences observed in estimated muscle forces when scaling muscle activity by muscle volume in the static optimization objective function in Human Body Model. Constraining the muscles by their force–length–velocity properties in OpenSim had less influence on the estimated muscle forces. However, this constraint might be more important for faster motions for which muscle properties and dynamics play a more important role.[26,27] Finally, for various reasons, we expect different optimization algorithms and stopping criteria in OpenSim and Human Body Model to have a limited influence on the static optimization results. First, we have studied the sensitivity of the results to the stopping criteria in OpenSim and found that muscle activations differed at most by 1e–4 (biceps femoris short head) when changing the convergence criterion (from 1e–4 to 1e–5) and the maximum number of iterations (from 100 to 10,000). Second, Human Body Model allows a relatively long computational time to solve the static optimization problem, limiting the risks of suboptimal solutions. Third, the static optimization problem is a quadratic programming problem (ie, local optima are global optima), and the initial guesses will therefore not affect the results.

OpenSim and Human Body Model were designed with different applications and target users in mind. Human Body Model is real time and aimed toward clinicians with no particular technical skills. It relies on a predefined muscle model, which may not be suitable when subject specificity is required.[28,29] OpenSim is open source, enables subject-specific modeling, and is aimed more toward researchers with technical backgrounds. Its standard workflow is offline, although an OpenSim-based real-time system was recently developed to compute inverse kinematics and inverse dynamics for lower-limb applications.[30] Finally, Human Body Model does not require user inputs to create a model and is therefore robust against user errors. By contrast, OpenSim provides the users with more flexibility in the scaling and inverse kinematic setups. However, the user choices can have an influence on the results (see Table S5 in Supplementary Material [available online]).

We found differences in joint kinematics, kinetics, and muscle forces resulting from processing the same experimental gait data from healthy adults using the OpenSim model and Human Body Model. Both models are similar in many aspects but differ in the definitions of the kinematic model (joint center and axis definitions), and we expect these differences to be the main causes for the outcome differences. Since different computational choices resulted in different muscle forces, continued efforts for validating models and methods are required.[15,31] Depending on the aim, differences in biomechanical variables between models and software systems may be more or less important. In clinical analyses, focus is on interpreting deviations from reference data. Therefore, processing reference and patient data with the same model and software system is in general sufficient to deal with model and computational uncertainties. We compared SDs of joint kinematics and kinetics between the OpenSim model and Human Body Model as well as which trials deviated more than 1 SD from the mean (see Table S6 in Supplementary Material [available online]). Since we observed similar results, we expect similar interpretations when comparing reference and patient data based on either the OpenSim model or Human Body Model. By contrast, as described by Roelker et al,[4] processing reference and patient data with different models and software systems may result in incorrect interpretations if discrepancies between models and software systems are not taken into account. In mechanical analyses, the goal is to obtain accurate estimates of kinematics and loading, and therefore, discrepancies between models or computational choices may lead to different conclusions. In such cases, musculoskeletal models should be used with care. Similarly, differences in biomechanical variables are important when comparing results from studies in the literature that were obtained with different models and software systems. Differences that are smaller than the differences reported in this study cannot be attributed to differences in the movement execution.

Based on the results of this study, we recommend that researchers aiming to compare their results with results from other simulation studies pay special attention to the definition of the pelvic reference frame, the hip and knee joint centers, and the static optimization cost function. Since it is currently unknown which cost function provides the "best" approximation of the human control strategy, computed muscle activations should be interpreted carefully and, whenever possible, compared with experimentally measured muscle activations. Muscle forces are the main determinants of lower limb contact forces during walking. The large differences in muscle forces might therefore influence the evaluation of joint loading. We previously found differences in knee joint loading of about 8 N/kg between healthy individuals and patients with severe osteoarthritis.[32] Similar differences might be caused by the differences in magnitudes of the muscle forces we report here. However, all muscles spanning a joint determine joint loading, and therefore, additional model comparison is needed to evaluate the effect of the model choice on joint loading. Nevertheless, we advise researchers to be aware of the effect of modeling choices on computed muscle forces when evaluating joint loading. Overall, in model-based biomechanical analyses, users should be conscious of the modeling and computational assumptions and their influence on the biomechanical variables.

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