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Consistency Among Musculoskeletal Models: Caveat Utilitor

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Abstract—Musculoskeletal simulation software and model repositories have broadened the user base able to perform musculoskeletal analysis and have facilitated in the sharing of models. As the recognition of musculoskeletal modeling continues to grow as an engineering discipline, the consistency in results derived from different models and software is becoming more critical. The purpose of this study was to compare eight models from three software packages and evaluate differences in quadriceps moment arms, predicted muscle forces, and predicted tibiofemoral contact forces for an idealized knee-extension task spanning −125 to +10° of knee extension. Substantial variation among models was observed for the majority of aspects evaluated. Differences among models were influenced by knee angle, with better agreement of moment arms and tibiofemoral joint contact force occurring at low to moderate knee flexion angles. The results suggest a lack of consistency among models and that output differences are not simply an artifact of naturally occurring inter-individual differences. Although generic musculoskeletal models can easily be scaled to consistent limb lengths and use the same muscle recruitment algorithm, the results suggest those are not sufficient conditions to produce consistent muscle or joint contact forces, even for simplified models with no potential of co-contraction.

Keywords—Musculoskeletal models, Muscle moment arm, Joint contact force, Muscle recruitment, Musculoskeletal simulation, Knee flexion.

INTRODUCTION

Software packages specifically designed to facilitate the development and analysis of musculoskeletal models (e.g., AnyBody,[13] BoB,[57] LifeModeler (http://www.lifemodeler.com), Opensim,[15] SIMM[16]) have led to the expansion of musculoskeletal simulations. Additionally, model repositories (e.g., AnyBody Repository (http://forge.anyscript.org/gf/), PhysiomeSpace (www.physiomespace.com/), Simtk.org) have made possible the sharing and distribution of musculoskeletal models, which have allowed different researchers and users to more easily expand or incorporate previous work not developed locally. One early example of such a repository that contained model parameters of the lower limb (http://isbweb.org/data/delp/index.html) demonstrates the potential and impact that musculoskeletal data, made available to the research community, can have with the primary manuscript associated with the data-set[17] currently having 533 citations (Scopus, accessed 5/9/2013). The widespread use of this data set over the past two decades can in part be explained by the considerable time and effort required to develop mathematical representations of anatomical structures.

Musculoskeletal models have been used to investigate a wide range of research topics including physiological loading,[33,45,57,58,64] wheelchair propulsion,[20] reaching,[60] ergonomic evaluation,[1,49,63] and design optimization.[31,50] Musculoskeletal simulation software, which can be used to estimate quantities difficult to measure non-invasively (e.g., muscle force, joint contact force), has not only been developed to quantify absolute internal body forces, but also with the intent of examining the effect of an environmental or postural change on model performance (e.g., stability, muscle function).[13,55] Analysis of such cause–effect relationships has great potential for incorporating internal body measures into device and component design.[30,32] The same relationships have also been

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proposed as a method for validating certain components of musculoskeletal simulations.[42] Generic human figure models widely used in the related field of ergonomics can be scaled to population-based anthropometric measurements to evaluate accommodation and other engineering-based design goals.[22] In a similar capacity, the use of scaled generic musculoskeletal models has the potential to be used as an engineering tool in which individualized patient assessment is not required. Additionally, compared to image-based models defined using individual-specific scan data, analyses with generic models are not burdened by expensive scan costs and lengthy image processing times.[9,62]

Verification and validation of newly developed and currently available musculoskeletal models are non-trivial tasks and remain topics of ongoing research.[14,27,42] Recent studies have investigated the comparative accuracy of scaled generic musculoskeletal models to that of subject-specific geometry, and the effect of those differences on computed muscle moment arm,[6,52–54] functional roles of muscles during gait,[12] and joint contact force.[27,45] Validation among models is also necessary, with the expectation of users that the same analyses performed with different models or software will produce consistent results.[61] It is not known whether this expectation is currently being met and/or to what capacity users of different models must scale or adapt those models to yield consistent results.

Mathematical models of the knee joint and its surrounding muscles have been used to better understand a wide array of topics including cruciate ligament function,[2] the interaction of muscle activation and knee injury during frontal car crashes,[11] and knee joint reaction loading during walking.[27] One application relevant to our laboratory is the use of generic musculoskeletal models for evaluating exercise therapies and interventions for individuals with spinal cord injury (SCI). Joint reaction force at the knee has previously been used to compare different exercises and quantify internal loading during exercise participation, including those with a functional electrical stimulation component.[5,21,28,35,44] In the context of skeletal health, an issue particularly relevant to individuals with SCI,[36] both trend and absolute estimates of knee force can aid in the design or adaptation of an exercise. To our knowledge, there exist no directly measured data (e.g., instrumented endoprostheses) that can be used to compare to the knee joint reaction force output of musculoskeletal models simulating exercise therapies or interventions for individuals with SCI. Therefore, indirect validation of the overall musculoskeletal model appears to remain the optimal method for gaining confidence in the simulation results. The model may in fact provide the best available estimate to the internal loading within the actual system. However, in the context of this application, it remains unclear if the selection of the generic model substantially influences the accuracy and/or interpretation of the results.

The overall goal of this study was to compare the results of several commonly available generic musculoskeletal models, scaled to consistent anthropometry, in determining moment arms, muscle force contributions, and predicted knee joint contact force during an idealized knee-extension task for postures spanning an extended and substantially flexed knee. To simplify the comparisons, simplified musculoskeletal models that only included the quadriceps muscles were used. Our first study aim was to quantify the differences in the lengths of the quadriceps moment arms between models, particularly at postures of high knee flexion. Our second study aim was to explore absolute and trend differences in simulated muscle recruitment and joint contact force between models. Our final aim was to identify future research questions and topics that will aid in the consistency of results produced by different musculoskeletal modeling models and software packages.

MATERIALS AND METHODS

Quadriceps muscle moment arms and tibiofemoral joint contact for a simulated knee extension task were computed for several musculoskeletal models spanning three unique musculoskeletal simulation software environments. Models were anthropometrically scaled to have consistent limb length dimensions. Muscle moment arms were computed for eight unique musculoskeletal models. Tibiofemoral joint contact loads were computed for a subset of five models that had the capability for computing tibiofemoral joint loading during a simplified isotonic knee extension task. Results are presented over knee angles ranging from $-125°$ to $+10°$ knee extension. Knee angles of $-20°$, corresponding to peak knee flexion during mid-stance of normal gait,[46] and $-100°$, corresponding to peak or sub-peak knee flexion during activities that include stair ascent, stair descent, cycling, leg press, sit to stand, power lifting, squatting and FES rowing,[23–26,34,43,67] are also used to compare intra and inter-model differences for minimal and deep knee flexion postures.

Musculoskeletal Models

Eight musculoskeletal models that included lower extremity musculature (Table 1) were evaluated (see Appendix—Table 7 for model accessibility). The selected models were implemented in the AnyBody (http://

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www.anybodytech.com), OpenSim (http://opensim.stanford.edu/), or Biomechanics of Bodies (BoB) modeling software packages. Prior to testing, each model was scaled to the joint-to-joint dimensions listed in Table 2. Off-axis bone dimensions were scaled isometrically. Each model was simplified to only include representations for four quadriceps muscle groups (vastus lateralis, vastus intermedius, vastus medialis, and rectus femoris). All muscles were modeled using a Hill-type representation.[70] The model-defined values of maximum muscle strength at optimal fiber length (Table 3) were not changed. Additional differences between muscle model representations and parameters (e.g., optimal fiber length, pennation angle, *etc.*) are not presented.

Muscle path representation, a component that contributes to the effective muscle moment arm, varied among models. The AnyBody and Biomechanics of Bodies musculoskeletal models represented muscle paths as line segments defined by insertion, origin, and intermediate *via* points. *Via* points are frictionless constraints at one or more locations along the path of the muscle. The Delp 1990, Gait 2392, and Steele 2012 models used *via* points that depended on posture. The London Lower Limb and Lower Limb 2010 models defined the path of each quadriceps muscle based on insertion and origin points and idealized surface geometry used to represent underlying physiological structures around which a muscle wraps.[6] The AnyBody—LegTD and London Lower Limb models, based on the same cadaver dataset,[37] represent each quadriceps muscle using multiple muscle fascicles while the remaining models represent each quadriceps muscle with a single muscle unit. For example, in both models with multiple muscle fascicles, the vastus intermedius is represented as 6 separate fascicles attached at two insertion points on the proximal aspect of the patella, and multiple muscle origins along the femur. The reported muscle strengths are the sum of all the muscle fascicles representing that single muscle (Table 3).

The kinematic knee joint definition, another component that contributes to the effective muscle moment arm, was not consistent among all models. The AnyBody—Leg, AnyBody—LegTD, and London Lower Limb models define the tibiofemoral joint kinematics as an idealized hinge (revolute) joint. The Delp 1990, Gait 2392, and Steele 2012 models define the tibiofemoral kinematics as a single coordinate with coupled rotation and translation.[69] The Lower Limb 2010 model defines the tibiofemoral kinematics based on experimental data presented in Walker *et al.*[65] The BoB model defines the tibiofemoral kinematics as two rolling cylinders with radii approximated from Leszko *et al.*[41] The AnyBody—LegTD and London Lower Limb models define the patellar kinematics as a circular path defined in the local femur reference frame and is prescribed by the tibiofemoral knee angle. For those models, the patellar position maintains a constant patellar tendon length throughout the knee range of motion. The AnyBody—Leg model does not have a patellar body but includes a quadriceps muscle *via* point in the approximate location of the patella with the quadriceps muscles attached to the proximal tibia. The Gait 2392 model does not include a patella. The Delp 1990 model includes a patella body with its position defined by 4 coordinates (3 translational, 1 rotation), each functionally prescribed by the tibiofemoral knee angle, with respect to the local tibial

TABLE 2. Lower extremity scaled model dimensions.

Scaled dimension[a]	Value	Definition
Pelvis width (m)	0.166	Left to right hip joint center
Thigh length (m)	0.434	Hip to knee joint center
Shank length (m)	0.428	Knee to ankle joint center
Body mass (kg)	74	Whole body mass
Body height (m)	1.75	Not used in scaling, for reference only

[a]Dimensions based on scaled 'AnyBody—Leg' model to 50th percentile male by stature.

TABLE 1. Musculoskeletal models used to compute model-predicted moment arms.

Model name	Software package	References
AnyBody—Leg	AnyBody (v 4.1.0)	Damsgaard *et al.*[13]
AnyBody—LegTD	AnyBody (v 4.1.0)	Andersen *et al.*[4]
Biomechanics of Bodies (BoB v3.0)	Matlab (v 7.12)	Shippen and May[57]
Delp 1990[a]	Opensim (v 2.4.0)	Delp *et al.*[17]
Steele 2012	Opensim (v 2.4.0)	Steele *et al.*[58]
Gait 2392	Opensim (v 2.4.0)	http://simtk-confluence.stanford.edu:8080/x/54Mz
London Lower Limb	Opensim (v 2.4.0)	Modenese *et al.*[45]
Lower Limb 2010	Opensim (v 2.4.0)	Arnold *et al.*[7]

[a]As implemented in the Opensim model 'BothLegs.osim'.

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TABLE 3. Maximum quadriceps muscle strengths for the different models.

	Maximum isometric strength at optimal fiber length (N)			
Model name	Vastus lateralis	Vastus intermedius	Vastus medialis	Rectus femoris
AnyBody—Leg[a]	1852	1224	1283	773
AnyBody—LegTD[b]	1882	1029	1617	780
Biomechanics of Bodies	1870	1235	1295	780
Delp 1990	1871	1365	1294	779
Steele 2012	1871	1365	1294	1169
Gait 2392	1871	1365	1294	1169
London Lower Limb[b]	2579	1410	2216	1069
Lower Limb 2010	2255	1024	1444	849

[a]Muscle strengths were scaled based on thigh mass using standard software pipeline.
[b]Muscle strengths are the sum of individual fascicles used to represent each muscle.

reference frame. The Steele 2012 and Lower Limb 2010 models include a patella body with its kinematics defined by 3 coordinates (2 translational, 1 rotational), each functionally prescribed by the tibiofemoral knee angle, with respect to the local femur reference frame. The BoB model includes a patella with its kinematics defined from Azmy *et al.*[8] with the patella translations and rotations defined as a function of knee flexion angle encoded using a cubic interpolating look-up table.

Muscle Moment Arms—Quadriceps

Model-predicted moment arm data were obtained using the same method for all models using a direct load measurement method, previously summarized by An *et al.*[3] Sub-models of each musculoskeletal model were constructed with only the single muscle (or group of muscle fascicles representing a single muscle) to be evaluated. An external unit torque was applied about the rotational axis of the knee. Knee flexion was varied between −125° and +10° (knee extension) over a time of 1000 s to approximate a quasi-static analysis at each analyzed posture. Hip flexion, abduction, and internal rotation were defined to be 90, 0, and 0°, respectively. Muscle and tendon force for each model was computed using a static optimization procedure incorporated into each software package that the models were constructed in. Although an optimization procedure was used for the moment-arm analysis, the results are deterministic since only one muscle was included in each model and the muscle and connected skeletal linkage was modeled as a deterministic system (as opposed to a stochastic representation). The muscle moment arm at each knee angle was computed as the applied torque divided by the computed tendon force. The force of gravity was reduced to zero for each model. The computed moment-arms for the models implemented in OpenSim were essentially equivalent to the moment-arms given by the software's muscle moment arm calculation function.[56]

Tibiofemoral Joint Contact Force

Model-predicted tibiofemoral joint contact forces were obtained for a simulated task of knee extension. The method of load application and evaluated knee postures was similar to the muscle moment arm derivation previously described. A constant external knee flexion torque of 90 N-m was used in each simulation. Each musculoskeletal model included representations of all four components of the quadriceps. Individual muscle strengths, paths, and muscle model parameters were not changed from their default values following anthropometric scaling. Muscle forces were computed using a static optimization procedure that minimized the sum of squared muscle activations. Tibiofemoral joint contact forces were computed within each musculoskeletal software program and reported in the local tibial reference frame defined by each model. The overall magnitude of the joint contact force is reported here to facilitate comparisons between models.

RESULTS

Moment Arms

The difference between the moment arms for the individual quadriceps muscles within a single model was relatively small. The maximum intra-model moment arm difference was 1.33 cm and occurred for the BoB model with a knee extension angle of 10°. At each knee angle, the quadriceps moment arms were equal for the AnyBody-Leg model with the exception of the rectus femoris, which was not able to produce a knee extension torque between −22 and +10° of knee extension. The mean intra-model quadriceps muscle moment arm difference across models (excluding the AnyBody-Leg model) over the range of motion tested (−125 to +10 knee extension) was 0.44 cm. For knee flexion angles greater than 20°, the maximum moment arm difference for all models was 0.68 cm, which

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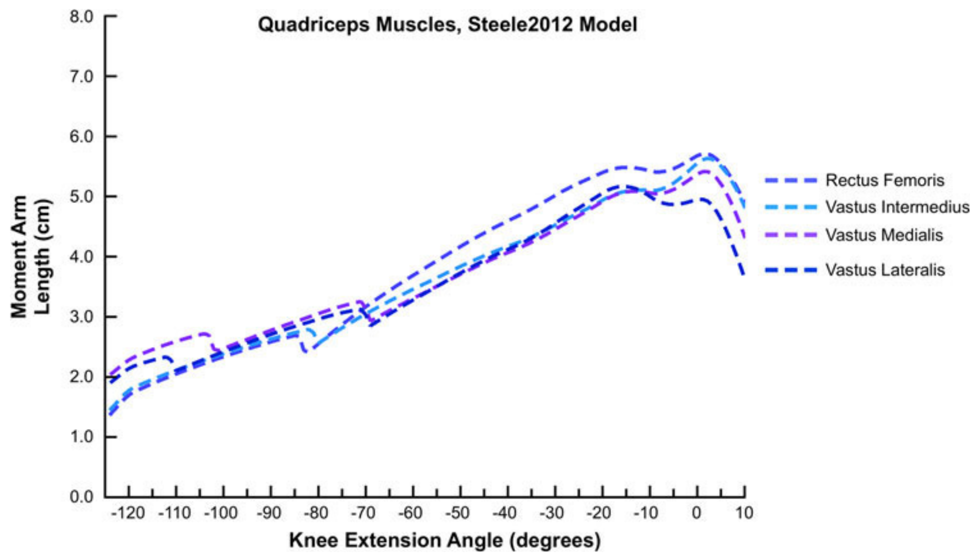

FIGURE 1. Quadriceps muscle moment arms for the Steele 2012 model.

TABLE 4. Maximum difference in quadriceps muscle moment arm for each musculoskeletal model at 20° and 100° of knee flexion.

Model	Moment arm difference at 20° (cm)	Moment arm difference at 100° (cm)
AnyBody—Leg	0.0[a]	0.0
AnyBody—LegTD	0.15	0.35
BoB	0.55	0.30
Delp 1990	0.46	0.19
Steele 2012	0.48	0.14
Gait 2392	0.57	0.08
London Lower Limb	0.65	0.12
Lower Limb 2010	0.40	0.27

[a]Excluding rectus femoris, which could not produce a knee extension moment in this posture.

TABLE 5. Maximum muscle moment arm change observed in the quadriceps muscle group for knee extension angles spanning -125 to +10° for each model.

Model	Muscle(s)	Minimum (cm)	Maximum (cm)	Range (cm)
AnyBody—Leg	All vasti	1.50	5.03	3.53
AnyBody—LegTD	Vastus medialis	1.65	6.17	4.53
BoB	Vastus medialis	3.07	3.84	0.78
Delp 1990	Rectus femoris	2.02	5.11	3.09
Steele 2012	Rectus femoris	1.27	5.70	4.43
Gait 2392	Rectus femoris	4.73	7.53	2.80
London Lower Limb	Vastus medialis	2.67	6.87	4.20
Lower Limb 2010	Vastus lateralis	1.33	4.93	3.60

occurred in the Steele 2012 model at maximum knee flexion ($-125°$) between the vastus medialis and rectus femoris (Fig. 1). Table 4 summarizes the intra-model moment arm differences at 20 and 100° knee flexion.

In general, the quadriceps moment arms decreased as the knee extended beyond $-20°$. The exception to this trend occurred in the BoB model, which exhibited consistent moment arms throughout the evaluated range of motion. Within each model, the maximum length change of a single quadriceps muscle moment arm over the evaluated knee range of motion (Table 5) spanned from 0.78 cm (BoB) to 4.53 cm (AnyBody—LegTD). Table 6 summarizes the computed quadriceps moment arms for each model at 20 and 100° knee flexion.

The eight scaled musculoskeletal models have both different absolute lengths of the quadriceps moment arms and different trends over the evaluated knee range of motion. The moment arms of the vastus lateralis for the different models are presented in Fig. 2. Similar results were observed for the vastus medialis, vastus intermedius, and the rectus femoris (not shown). No single model resulted in either the highest or lowest moment arm limits over the range of knee angles evaluated.

The greatest inter-model agreement, identified by the coefficient of variation (COV), was observed between knee flexion angles of -10 and $-60°$, angles nearly spanning those observed in normal gait[46] (Fig. 3). For knee flexion angles approaching either end of the range of motion limits, the coefficient of variation exceeded 2.5 times the minimum value observed at 23° knee flexion. Excluding the BoB and Gait 2392 models, which have different qualitative trends for the moment arm versus knee extension angle as the other models and previously reported data,[10] the minimum coefficient of variation value decreases from 0.16 to 0.11, the maximum coefficient of variation for deep knee flexion

TABLE 6. Quadriceps muscle moment arms for each musculoskeletal model at 20° and 100° of knee flexion.

	Moment arms at 20° knee flexion (cm)				Moment arms at 100° knee flexion (cm)			
Model	VL[a]	VM	VI	RF	VL	VM	VI	RF
AnyBody—Leg	4.72	4.72	4.72	0.0	3.28	3.28	3.28	3.28
AnyBody—LegTD	5.66	5.61	5.72	5.77	2.50	2.35	2.64	2.70
BoB	3.65	3.68	3.91	4.20	3.62	3.81	3.91	3.92
Delp 1990	4.77	4.64	4.64	5.10	2.77	2.87	2.77	2.69
Steele 2012	5.05	4.91	4.93	5.39	2.41	2.46	2.38	2.33
Gait 2392	6.04	5.66	5.63	6.20	4.85	4.93	4.93	4.85
London Lower Limb	6.22	6.87	6.57	6.55	2.80	2.68	2.80	2.75
Lower Limb 2010	4.93	4.80	4.82	5.20	1.58	1.57	1.76	1.84

[a]VL—Vastus Lateralis, VM—Vastus Medialis, VI—Vastus Intermedius, RF—Rectus Femoris.

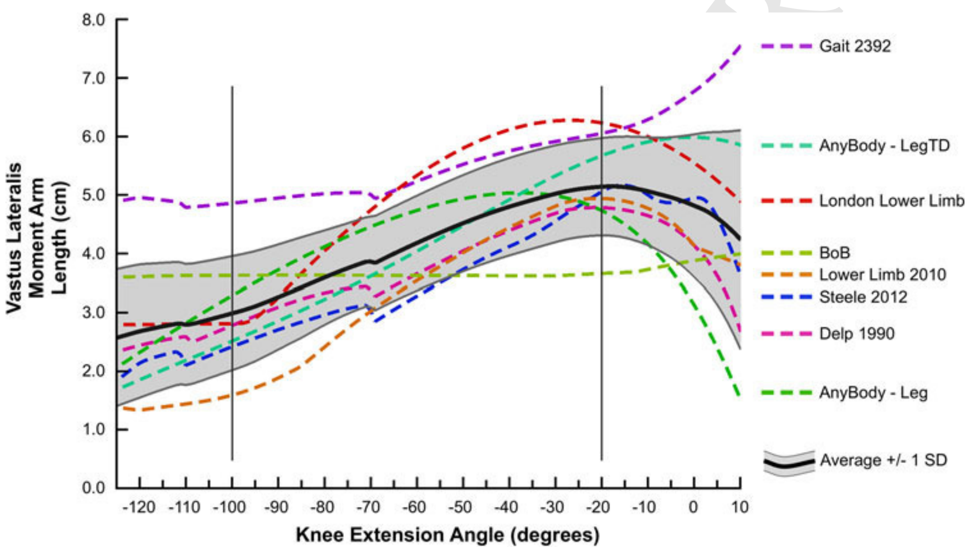

FIGURE 2. Muscle moment arms for the vastus lateralis.

decreases from 0.46 to 0.25, and the inter-model agreement remains relatively unchanged for straight and hyper-extended knee postures. The variation among moment arms between similar anthropometrically scaled (isometrically) musculoskeletal models is comparable to the variation previously reported between subjects from previous studies (Fig. 3).

Muscle Force

Quadriceps muscle recruitment was compared for seven of eight musculoskeletal models. Muscle forces for the vastus lateralis, vastus medialis, vastus intermedius, and rectus femoris were computed for each model during the same simulated knee-extension task. For each knee angle, the distribution of quadriceps muscle forces to produce a 90 N-m knee extension torque was computed. Results are presented for all the models for knee flexion angles of 20 and 100° (Fig. 4).

All models had an increase in the combined quadriceps muscle force between 20 and 100° knee flexion, with an average increase of 1351 N. At 100° knee flexion, the contribution of the vastus lateralis to the combined quadriceps force was reasonably consistent between 30 and 49 percent. In contrast, at 20° knee flexion, the contribution of the vastus lateralis ranged from 14 to 82 percent. Of the combined 14 evaluated models and postures, the vastus lateralis contributed the largest percentage of all muscles in 11 of the analyses. Within each model, the contribution of force associated with the vastus medialis and vastus intermedius was fairly consistent. The difference in force contribution between those muscles within each model was always less than 6%, with the exception of the AnyBody—Leg (20° knee flexion) and the AnyBody—LegTD models, which had differences between the force contribution from the vastus intermedius and vastus medialis upwards of 20%.

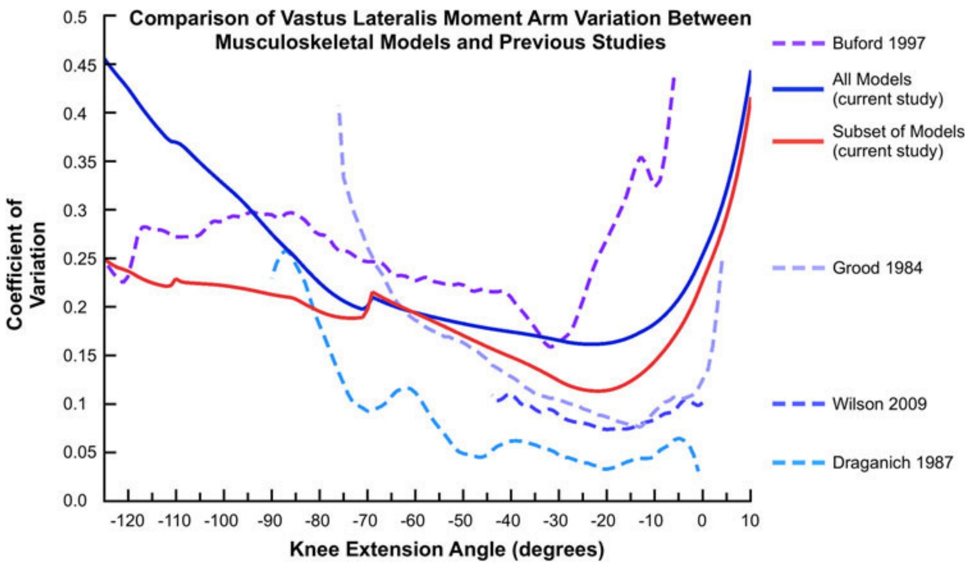

FIGURE 3. Coefficient of variation for the vastus lateralis moment arm for the musculoskeletal models and previous studies. The 'subset of models' group excludes the Gait 2392 and BoB musculoskeletal models.

Tibiofemoral Joint Contact Force

The magnitude of the resultant force vector of the tibiofemoral joint contact force was calculated for a subset of the models for the 90 N-m knee extension torque task. Results from the five musculoskeletal models that could be directly used to compute the knee joint contact force are presented in Fig. 5. The within model range of knee joint contact force spanned 219 N and 4204 N for the BoB and AnyBody—Leg models, respectively, over the range of knee angles evaluated. Both the Steele 2012 and the AnyBody—Leg models exhibited a substantial increase in tibiofemoral contact force as knee extension angle decreased past −50°. In contrast, the remaining three models had only slight changes in joint reaction force above and below −50° knee extension. At −100° knee extension, the knee joint reaction force ranged from 1839 to 3754 N between models, a difference of 2.6 body weights. In contrast, the knee joint reaction force ranged from 1525 to 2269 N at −20° knee extension, a difference of approximately one body weight.

DISCUSSION

The study compared knee extensor moment arms, muscle force predictions, and knee joint contact force predictions for several similarly scaled musculoskeletal models available to the biomechanics community. Substantial variation among models was observed for all aspects evaluated. The one exception was the relatively consistent (among models) within-model moment arm range spanned by the quadriceps muscle group of each model (e.g., Fig. 1). Differences between models were influenced by knee angle, with better inter-model agreement occurring at knee flexion angles in the range from 10 to 60°.

The within-model moment arm range was relatively small and always less than 1.33 cm for each of the models evaluated. A slightly smaller value was observed in data presented by Klein Horsman,[38] which showed a maximum range between individual quadriceps muscle moment arms of less than 1 cm for knee extension angles spanning −135 to 0° for a single cadaver specimen. Similarly, the maximum range of the averaged moment arms (15 cadaver specimens) for the different quadriceps muscles presented by Buford *et al.*[10] was also less than 1 cm for a similar range of knee angles. In both studies the maximum range occurred at small angles of knee extension (i.e., near full leg extension), similar to the models evaluated in this study. The average of the maximum moment arm differences sampled at each knee posture for the Buford *et al.*[10] and Klein Horsman[38] cadaver studies were 0.47 and 0.44 cm, respectively, similar to the average of 0.44 cm of the eight evaluated musculoskeletal models. The results suggest the musculoskeletal models appear to be reasonably consistent, with each other and previous cadaveric studies, in representing the moment arm intra-specimen variability of the quadriceps muscle group.

In attempting to apply the formal concepts of verification and validation to musculoskeletal modeling, Lund *et al.*[42] states that, "…verification is a prerequisite for validation. Verification provides the evidence that the computer code correctly solves the underlying mathematical model. Absence of verification creates the risk of mixing modeling errors and errors caused by implementation." Verification is an important topic; however the study conducted here

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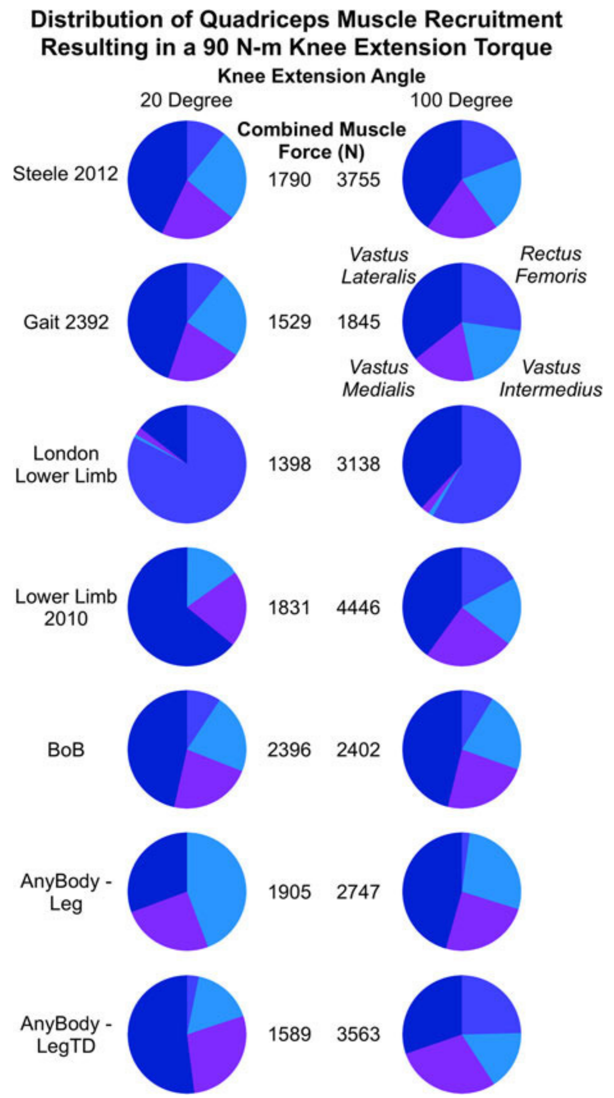

FIGURE 4. Contribution of quadriceps muscle force to produce a 90 N-m knee extension torque at 20 and 100° knee flexion.

focused on validation and not verification. For example, the results from the static optimization analyses performed in the current study were not explicitly checked and it was assumed they were consistent with the equations of motion for the defined system. Lund *et al.*[42] further defined the examination of the "correctness of variable interaction" as trend validation, a concept that has been previously used to evaluate musculoskeletal model performance and understand changes in knee loads for different walking styles.[14,47] Comparing the models tested in this study in the context of variable interaction, the majority of models (7 of 8) did exhibit smaller muscle moment arms at large angles of knee flexion compared to moderate or low knee flexion angles, a result consistent with previous studies.[10,29,68] Two of the eight models had maximum moment arm values for the vastus lateralis at +10° knee extension (hyper-extension), the maximum knee extension angle evaluated. The remaining models exhibited maximum vastus lateralis moment arms at slightly flexed knee postures, an observation more consistent with previous studies.[29,68] The general consistency of these variables between musculoskeletal models is encouraging and suggests similar interpretations from a trend type analysis may be achieved when using the majority of the available models.

The results suggest a lack of absolute consistency in the tested musculoskeletal models and that model differences are not simply an artifact of naturally occurring inter-individual differences. Although the data used to develop the individual musculoskeletal models in this study were not from a consistent or nominal population, the expectation of the generically scaled models evaluated in this study is that they each represent the mean anatomy of a male individual with 50th percentile stature. It is currently difficult to evaluate whether a single musculoskeletal model accurately represents such mean anatomy, potentially explaining the differences between models observed here, as there is limited data available and differences due to inter-individual variation are unknown. The available data quantifying the variability for the vastus lateralis moment arm is not consistent. Using data from previous studies, moment arm standard deviations (averaged across the available knee angles) for subjects include values of: 3.02 cm,[10] 2.30 cm,[19] 0.43 cm,[68] and 0.38 cm.[29] In comparing a musculoskeletal model to literature values, Klein Horsman[38] assumed absolute differences smaller than 2 cm could be attributed to inter-individual differences. In contrast to the previous literature, the average standard deviation (over all knee angles) of the vastus lateralis moment arm for the models evaluated in this study was 0.95 cm, with the maximum inter-model difference ranging from 2.0 to 6.0 cm.

Isometric scaling was applied to scale the off-axis skeletal dimensions using the same scaling factors applied to define the limb lengths in an effort to generate consistent musculoskeletal models. However, width and breadth anthropometric dimensions are not as well correlated with stature as limb length dimensions[51] suggesting that advanced scaling methods may be necessary to improve model consistency. As the quadriceps muscle moment arms have been shown to be well correlated with femoral condyle width,[39] consistent scaling between models along that dimension may reduce model differences. The AnyBody Modeling System has body-scaling functions that incorporate body mass and percent fat, which are used to influence the mediolateral and anteroposterior skeletal dimensions. As those same functions were not available in the other modeling programs, they were not investigated in this study. Previous studies have investigated the use of patient-

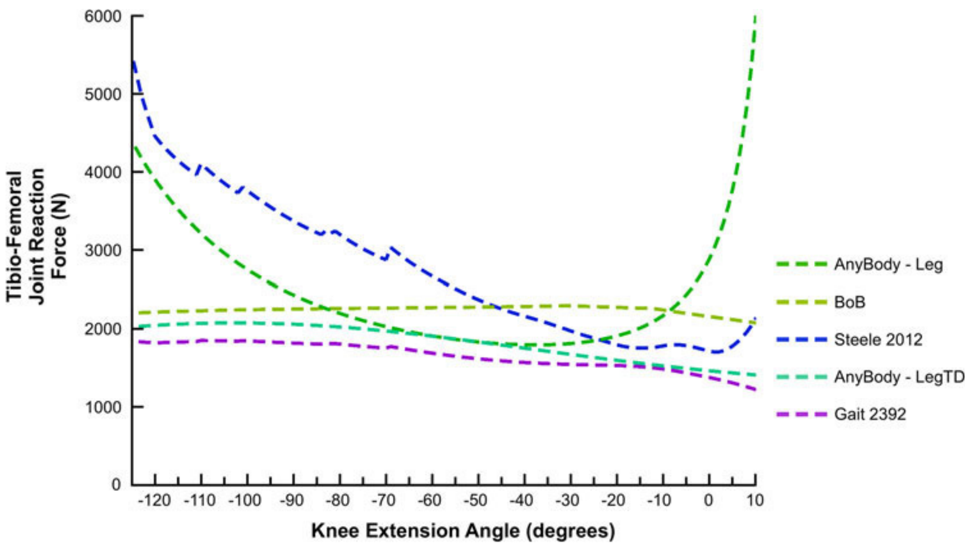

FIGURE 5. Tibiofemoral joint reaction force from quadriceps muscles resisting a 90 N-m flexion torque.

specific anatomy (derived from MR or CT imaging) to dimension and develop musculoskeletal models.[6,12,53,59] However, custom scaling and definition of the muscle path based on imaging was not performed here, as the intent of this study was to compare differences in available generic models that could be used without the need for data from a specific subject.

The vastus lateralis was the largest contributor to the overall quadriceps muscle force in the majority of models and postures evaluated. This result was expected considering that the vastus lateralis had the largest maximum isometric strength at optimal fiber length in all the models, a result consistent with previous studies which have shown the vastus lateralis to have the largest physiologic cross-sectional area of the four quadriceps muscles.[66] One notable exception was observed for the muscle recruitment results associated with the London Lower Limb model, which recruited the majority of the quadriceps force from the rectus femoris despite the vastus lateralis having a larger effective moment arm, maximum isometric strength, and maximum torque producing capability at 20 and 100° knee flexion. Upon further investigation, the relatively large rectus femoris force can be primarily explained by the large force contribution from the passive element of those muscle fascicles. At low muscle activation, comparatively high forces can be transmitted to the muscle tendon. The large passive force contribution suggests the London Lower Limb model may not have appropriately scaled muscle fiber lengths for the muscles investigated here. Upon further inspection, the normalized fiber length for one of the rectus femoris muscle fascicles ranged from 3.3 to 4.5 for the corresponding knee flexion angles of 0 and 125°, respectively, suggesting a potential modeling error with the defined optimal fiber length. Similar normalized fiber length values were also observed for the un-scaled model.

In a study analyzing the sensitivity of individual muscle parameters on computed muscle force from a static optimization procedure, Raikova and Prilutsky[48] concluded that the non-zero optimal force of each muscle was non-linearly related to the moments at all the joints, the muscle moment arms, and the physiological cross sectional areas of all the muscles, which were used to normalize the predicted forces to compute muscle activity in the static optimization objective function. The differences between the models analyzed here support those conclusions and further identify that the parameters of the muscle model (although not directly analyzed here), particularly those that define the force–length curve and the contribution between the passive and active elements, also substantially influence the subsequently recruited muscle force. Further research, methods, and protocols for reliably producing consistent muscle forces between musculoskeletal models under the same boundary conditions are necessary.

The London Lower Limb and the Lower Limb 2010 models, which did have muscle forces computed to resist the simulated 90 N-m flexion torque, could not be used to calculate accurate tibiofemoral reaction forces using the Joint Reaction analysis tool in OpenSim.[58] Both models utilized a kinematic constraint to define the position of the patella as a function of knee angle. This constraint acts in place of the patellar tendon force such that the force exerted by the quadriceps muscles acting through the patella and patella tendon are not transmitted to the proximal tibia. For both models, the tibiofemoral reaction force computed using the Joint Reaction analysis tool was zero, as the weight of the lower limb was neglected (gravity was set to zero) and a pure torque was applied about the knee joint axis of rotation. It should be highlighted that if a force vector were applied distally to the knee (e.g., a ground reaction force) for either of these models, the Joint Reaction analysis tool would report a tibiofemoral joint reaction force that would be consistent with the inter-segmental forces of the linkage. If the user were unaware of the modeling implications of the kinematic constraint and the assumptions associated with the Joint Reaction analysis tool, reported reaction forces may be misinterpreted as true joint reaction forces.

The results suggest that although musculoskeletal models can fairly easily be scaled to have the same limb lengths and use the same muscle recruitment algorithm, those are not sufficient conditions to produce consistent muscle or joint contact forces (globally or by trends), even for simplified models with idealized boundary conditions and with no potential of co-contraction. However, between −10 and −50° knee extension, joint contact forces from all models were fairly consistent and ranged between 2.0 and 3.3 body weights (BW). Two models exhibited increased joint reaction forces as knee flexion angle increased, a result consistent with the observation that "tibial forces peaked at increasing knee flexion angle" from three subjects with instrumented endoprosthesis during a knee extension task.[18] The remaining three musculoskeletal models did not exhibit that same trend, but did have joint contact forces that converged between 2.5 and 3.0 BW at 125° knee flexion. Trepczynski *et al.*[59] identified considerable subject-specific variation in peak tibiofemoral joint loads during a variety of activities (e.g., walking and stair climbing), particularly those involving large knee flexion like squatting, a result consistent with the increased variation observed here between models as knee flexion increased.

FUTURE RESEARCH

The results presented here raise several questions and potential topics for future research including: What scaling, model parameters, and underlying model constructs must be matched to produce consistent results between musculoskeletal models? Is this possible? At what level should a musculoskeletal modeling user be expected to adapt a generic model to achieve "average" population results? These questions can be complicated to address considering the difficulties in identifying appropriate methods for model validation (e.g., what is the expected average behavior the models should be matching?). Additionally, differences that do exist between models can be difficult to interpret, as differences resulting from natural inter-individual variation remain unknown. Although this study focused on the differences between generic musculoskeletal models and did not investigate models scaled to match patient-specific data, the answer to many of the questions above may rely on additional patient-specific data being made available to the musculoskeletal simulation community.[62] For example, average and inter-subject variation may have to be defined based on analysis from patient-specific models (e.g., Scheys *et al.*[52]) with the accuracy of those models being further evaluated using additional data available from instrumented endoprostheses.[27,40] For musculoskeletal simulation to be widely adopted and incorporated as an engineering discipline, verification and validation methods that are common to other computer aided engineering modalities must be more widely incorporated.[42] Consistent results between generic musculoskeletal models is one step toward accomplishing that goal such that a biomechanical analysis performed by one investigator at one location with one piece of software produces the same reliable and repeatable results as the same analysis performed by another individual, at another location, with another musculoskeletal simulation software package.

APPENDIX

See Table 7.

TABLE 7. Musculoskeletal software and model download locations.

Model name	Model accessibility	Software	Software accessibility
AnyBody—Leg	http://forge.anyscript.org/gf/project/ammr/	AnyBody Modeling System	www.anybodytech.com
AnyBody—LegTD	http://forge.anyscript.org/gf/project/ammr/		
Biomechanics of Bodies	http://www.marlbrook.com/download	Matlab	www.matlab.com
Delp 1990	https://simtk.org/home/low-ext-model	Opensim	https://simtk.org/home/opensim
Steele 2012	https://simtk.org/home/mattdemersstuff		
Gait 2392	https://simtk.org/home/torso_legs		
London Lower Limb	https://simtk.org/home/low_limb_london		
Lower Limb 2010	https://simtk.org/home/lowlimbmodel09		

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CONFLICT OF INTEREST

Matthew S. DeMers collaborated on the development of the Steele 2012 model evaluated in this manuscript. James M. Shippen is the primary developer of the Biomechanics of Bodies software used in this manuscript.

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