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Dependence of muscle moment arms on in-vivo three-dimensional kinematics of the knee

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Abstract

Quantification of muscle moment arms is important for clinical evaluation of muscle pathology and treatment, and for estimating muscle and joint forces in musculoskeletal models. Moment arms estimated with musculoskeletal models often assume a default motion of the knee derived from measurements of passive cadaveric flexion. However, knee kinematics are unique to each person and activity. The objective of this study was to estimate moment arms of the knee muscles with *in vivo* subject- and activity-specific kinematics from seven healthy subjects performing seated knee extension and single-leg lunge to show changes between subjects and activities. 3D knee motion was measured with a high-speed stereo-radiography system. Moment arms of ten muscles were estimated in OpenSim by replacing the default knee motion with *in vivo* measurements. Estimated inter-subject moment arm variability was similar to previously reported *in vitro* measurements. RMS deviations up to 9.0 mm (35.2% of peak value) were observed between moment arms estimated with subject-specific knee extension and passive cadaveric motion. The degrees of freedom that most impacted inter-activity differences were superior/inferior and anterior/posterior translations. Musculoskeletal simulations used to estimate *in vivo* muscle forces and joint loads may provide significantly different results when subject- and activity-specific kinematics are implemented.

Key terms

subject-specific; musculoskeletal modeling; fluoroscopy; muscle force

INTRODUCTION

An accurate quantification of muscle moment arms is necessary to evaluate muscle function and estimate forces needed to actuate movement. Moment arm is defined as the instantaneous measure of the “effectiveness” with which the contraction force of a given muscle can generate a moment at a “joint of interest”, while in a given configuration³¹. The standard calculation of moment arm of a force as minimum distance between its line of action and the axis of rotation of a joint cannot always be readily achieved in biomechanics

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DISCLOSURE OF POTENTIAL CONFLICTS

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because of the complexity of the involved structures. The line of action of most muscles cannot be simply defined as a straight line between origin and insertion because of wrapping around bones and other muscles¹¹, and complex biological joints like the knee present coupled rotations and translations that complicate the identification of a simple joint rotational axis^{28, 30, 36}.

Muscle moment arms have been measured and estimated through experiments (both *in vivo* and *in vitro*) and computational models. Each one of these three approaches presents peculiar advantages and limitations. Historically, *in vitro* measurements performed on a number of specimens^{6, 33} have been the gold-standard for the quantification of moment arms. Notably, moment arms of the muscles spanning the knee were measured *in vitro* by Buford et al. using the tendon excursion method¹. This technique enables comprehensive measurement of moment arms using cadaver specimens because of the direct access to tendon length measurements and precise control of the motion of the joint.

Although *in vitro* measurements are currently the gold-standard, they are an approximation of *in vivo* moment arms since they are obtained in unloaded conditions, during passive motion of the joints, and with resting muscles. In reality, the kinematics of the knee are affected by the muscle and external load demands of different activities, and these changes will affect the moment arms of the muscles. For example, internal/external rotation and anterior/posterior translation of the knee during seated knee extension can differ from extension of the knee in a weight bearing activity^{19, 24}. Since the 6 dof rotations and translations at the knee determine the location of the instantaneous axis of rotation of the joint, they also affect the muscle moment arm calculation. For this reason, researchers have developed techniques for measuring moment arm *in vivo* using imaging⁴ like static MR images^{17, 38}, computed tomography²⁶ and ultrasound¹⁴. These studies revealed subject and activity-specific variations in moment arm measures. Although these works have significantly enhanced the knowledge of *in vivo* muscle function, in most cases moment arm calculation relied on measurements of the orthogonal distance between the tendon and an estimate of the joint center on 2D images, and most MR-based measures are based on static acquisitions, precluding their use in assessment of moment arms for normal activities of daily living.

Musculoskeletal models have also been used to estimate moment arms of the lower limb through representation of the musculoskeletal geometry^{2, 3, 8, 27}, and prescription of the complex motion of the knee as a function of knee flexion angle^{3, 8, 21, 29}. However, the motion prescribed to the joint is commonly derived from the mean passive motion of cadaveric knees^{30, 36}. Since the *in vivo* kinematics of individuals differs from passive cadaveric motion^{13, 19}, moment arms estimated with musculoskeletal models may not be representative of muscle effectiveness *in vivo* during daily living activities. The influence of *in vivo* subject-specific and activity-specific kinematics on moment arms of muscles spanning the knee has never been investigated.

The overall goal of this study was to investigate the dependence of muscle moment arms on joint kinematics by combining musculoskeletal modeling and 3D knee kinematics measured *in vivo* to overcome some of the limitations of current approaches. Specifically, our purpose

was to: (1) analyze moment arms of the muscles spanning the knee estimated with *in vivo* subject-specific knee kinematics and compare these results to those obtained with passive cadaveric motion frequently used in musculoskeletal models; (2) investigate differences in moment arms estimated with *in vivo* subject-specific knee kinematics from a non-weight bearing activity (seated knee extension) and a weight bearing activity (single-leg lunge); (3) identify the dof of the knee that affect moment arm estimates. We hypothesized that substantial differences would be found between moment arms predicted from *in vivo* kinematics and those reported from *in vitro* motion, and further, moment arm estimates would change between weight bearing and non-weight bearing activity.

MATERIALS AND METHODS

Experimental session

Seven healthy subjects (age 63 ± 8 , 170 ± 9 cm, 72 ± 9 kg) performed knee extension and deep lunge activities. This study was approved by the University of Denver Institutional Review Board and all participants provided informed consent. The volunteers had no history of injuries or surgeries to the lower limbs. While performing the knee extension task, the subjects were in a seated position and were asked to slowly extend their knee from high flexion to full extension. No resistance was applied at the leg. The subjects were then asked to perform a single-leg lunge. Dynamic stereo radiographic images were collected using two matching custom radiography systems with 40 cm image intensifiers positioned at a relative angle of 60° . High-speed, high-definition cameras interfaced with the image intensifiers captured the motion at 50 frames/sec in a 'low-dose pulsed' x-ray mode¹⁵. A custom calibration cube with 52 beads was positioned inside the capture volume to determine the precise position and orientation of the two image planes. Image distortion, caused by the image intensifiers, was corrected by imaging a perforated steel plate positioned in front of each image plane. Tracking of bone motion was performed by optimizing the position of bone geometry models to the 2D stereo radiographs using Autoscooper (XROMM, Brown University, RI) and relative tibiofemoral (TF) and patellofemoral (PF) positions were obtained²⁰. Bone geometry (femur, tibia, fibula and patella) was semi-automatically reconstructed from static CT images with slice thickness of 1.0 mm. The coordinate system of the femur was placed at the midpoint of the centerline of a cylinder fitted to the medial and lateral posterior condyles (Fig. 1)²³. The medial/lateral axis was defined as the axial direction of the cylinder, whereas the superior/inferior axis was aligned to the posterior line of the femur. The anterior-posterior axis was defined according to the right-handed coordinate system notation. The femoral coordinate system was assigned to the tibia and patella at full extension during the knee extension activity. Tracking accuracy was previously validated and errors were sub-mm and sub-degree¹⁵.

Musculoskeletal model

Subject-specific full body musculoskeletal models with subject-specific representations of TF and PF kinematics were created in OpenSim⁷ (Fig. 2). The models incorporated the knee kinematics from knee extension and the lunge activity. Each model included a total of 12 segments (torso, pelvis, femurs, tibiae, tali, calcanei, toes). Coordinate systems of the segments were defined as in Arnold et al.³. In order to compare moment arm estimates with

different knee kinematics, a distal coordinate system was defined for the femur, coincident with the coordinate system described above. Its origin was located at the midpoint of the femoral condyles and its axes were parallel to the main femoral system at the femoral head. The coordinate system of the tibia and patella were defined coincident to this second femoral system at full extension. The lower limb included a ball-and-socket hip joint, a revolute ankle joint and a knee joint with prescribed TF and PF relative motion calculated from stereo radiography. TF and PF kinematics were decomposed into a sequence of intrinsic Euler angles and a translation vector. The sequence of rotations was the following: z (medial/lateral axis), y' (superior/inferior axis, after the first rotation), and x'' (anterior/posterior axis, after the second rotation)³⁹. All TF dof were prescribed, whereas the dof prescribed to the PF joint were flexion/extension, superior/inferior translation and anterior/posterior translation.

Ten musculotendon units per limb^{35, 41} were included: rectus femoris, vastus medialis, vastus intermedius, vastus lateralis, semimembranosus, semitendinosus, biceps femoris long head and short head, gastrocnemius medialis and lateralis. Musculotendon geometry (origin and insertion sites) was based on a previous model⁸. The patellar tendon was also modeled as a point-to-point muscle component that connects the lower part of the patella to the tibial tuberosity. Wrapping surfaces were included in the model and were used to represent muscle paths over other muscles and underlying bones. Semimembranosus and gastrocnemius (medialis and lateralis) wrap on a cylindrical surface at the femoral condyles and the quadriceps wraps on a similar surface that approximates the anterior face of the femoral cartilage. Subject-specific muscle attachment sites on the patella (quadriceps and patellar tendon) were selected for each subject since proximal and distal portions of the bone were easily identifiable anatomical landmarks.

Moment arms calculation

Moment arms of the muscles included in the model were calculated with respect to knee flexion angle for knee extension, lunge and passive cadaveric motion. Moment arms were calculated using OpenSim, which employs the “generalized force method” described in Sherman et al.³¹. Since the patellar tendon was described by a point-to-point segment and was assumed inextensible, its moment arm could not be estimated using the ‘generalized force method’. Therefore, it was calculated as the minimum distance between its line of action and the TF instantaneous helical axis³⁴. The consistency of the estimates obtained with the two methods was verified.

Data analysis

In vivo knee motion for the two activities was described as a function of flexion, and compared inside the minimum flexion range available across subjects. Specifically, paired Student’s t-tests were performed at every flexion angle (1° intervals) to compare knee extension and lunge kinematics. Knee motion collected *in vivo* was also compared to translations and rotations measured by Walker et al.³⁶ and used in previous musculoskeletal models^{3, 8, 40}.

Differences between moment arm estimates were calculated by means of root mean square deviations (RMSD) and Pearson Product-Moment Correlations (R). RMSD is a quantification of the absolute difference, whereas R measures the trend difference between two moment arm curves. For example, a relatively large RMSD (greater than 25% of the peak value) and a correlation R close to 1 indicate that the compared moment arms have similar trends but a relatively constant shift occurs between them. Specifically, moment arms estimated with activity-specific knee motion were compared for each subject to quantify the intra-subject variability. In addition, subject-specific and activity-specific moment arms were compared to estimates obtained with the same musculoskeletal model but with prescribed knee kinematics from a previously published model³. Moment arms measured in cadaveric studies⁶ were also included in the comparison. Moment arms were compared between the two activities with paired Student's t -tests at every flexion angle (1° intervals) inside the common flexion range. Average σ were also compared between muscles to identify which muscles presented the largest inter-subject variability.

To identify which TF dofs had the largest impact on differences between moment arms with different kinematics, five additional conditions were simulated. Specifically, each subject-specific knee extension TF motion was replaced by the corresponding motion from Walker et al.³⁶ (TFW) (e.g. the anterior/posterior translation for knee extension of subject #2 was replaced by the same dof from TFW). Moment arms were estimated for the five conditions, and RMSD and R with respect to moment arms with baseline knee extension kinematics were calculated. Low R correlation and large RMSD indicate that the dof changed in the model strongly influences the differences between moment arms.

Robustness analysis

A Monte Carlo analysis that models the uncertainty in muscle attachment sites was performed for each subject and activity in order to evaluate the robustness of our results. Attachments of every muscle in the model were perturbed in all directions according to a normal distribution (μ = baseline location, σ = 5 mm). A 5 mm σ is comparable to the range of landmark location errors reported in the literature^{18, 25, 37}. Only attachments on the patella (quadriceps and patellar tendon) were not perturbed, since the correct anatomical landmark was easily identified on the subject-specific geometry. RMSD and R between moment arms for different activities were calculated for every instance of the Monte Carlo analysis and compared to the same results obtained with the baseline model. In addition, 5–95% confidence bounds (the region in which the true result would lie with a confidence of 90%) were calculated for each Monte Carlo analysis. One thousand instances ensured a sufficient convergence of the results: μ and σ for RMSD and R over the last 100 instances lay within 5% of the final results for every subject and muscle, except for biceps femoris short head of subject #6 who's mean R was within 8.5% of its final mean R .

RESULTS

Kinematics

RMSD of the kinematics between activities averaged across subjects were under 3.7° and 2.5 mm for rotations and translations, respectively (Fig. 3). Statistically significant

differences ($p < 0.05$) were observed for TF internal/external rotation, anterior/posterior translation, and PF flexion/extension when kinematics for the two activities were compared (Fig. 3). The greatest difference between TFW and average *in vivo* knee extension kinematics was found for superior/inferior translation (RMSD: 4.6 mm). The largest differences between the PF motion adopted by Arnold et al.³ (PFA) and *in vivo* PF kinematics during knee extension were observed for flexion/extension (RMSD: 6.7°) and anterior/posterior translation (RMSD: 4.4 mm).

Moment arms

The muscles with greatest RMSD between moment arms estimated when *in vivo* knee extension and passive cadaveric motion were prescribed to the model were the biceps femoris long and short head (8.8 mm and 9.0 mm on average, respectively), whereas gastrocnemius lateralis presented the smallest RMSD (2.9 mm on average) (Fig. 4). Biceps femoris long and short head moment arms also showed the smallest correlations with moment arms calculated with passive cadaveric motion (−0.17 and 0.18 on average, respectively).

When moment arms were compared between activities, similar trends were observed (Fig. 4). Knee extension showed larger mean σ for all the muscles (+ 15% on average). The largest mean σ (indicating the greatest inter-subject variability) was observed for biceps femoris long head moment arm (6.7 mm). Statistically significant differences ($p < 0.05$) between moment arms for the two activities were found for the medial hamstrings, the biceps femoris short head and the gastrocnemius medialis and lateralis, for a maximum flexion range of 13° (see horizontal bars in Fig. 4).

RMSD and R between moment arms for different activities of single subjects were calculated and compared to the same results averaged across Monte Carlo instances (Fig. 5). Largest RMSD and smallest R were observed for biceps femoris short and long head (RMSD: 7.8 and 8.2 mm, respectively), and gastrocnemius medialis (R : 0.19). Patellar tendon showed smallest RMSD and largest R (2.6 mm and 0.96). RMSD averaged across Monte Carlo instances was within 2.4% of baseline values for every muscle (Fig. 5). Averaged Monte Carlo R was within 2.0% for every muscle except for gastrocnemius medialis (Fig. 5).

The muscle that showed the largest 5–95% confidence bounds from the Monte Carlo analyses was semimembranosus (17.2 mm during lunge), whereas the smallest bounds were observed for rectus femoris (0.33 mm during knee extension) (Fig. 6a).

Muscle groups were affected differently by the change of single dof from knee extension to TFW motion (Fig. 6b). RMSD and R between baseline models and models with changed dof were calculated and compared (Fig. 7). Large RMSD and small R indicate that the altered dof has a large influence on the difference between moment arms estimated for different kinematics. All the muscles presented relatively larger RMSD and smaller R when superior/inferior translation was changed to the spline from TFW (RMSD averaged across subjects peaked at 7.2 mm for semitendinosus) (Fig. 7). Biceps femoris was also affected by changes in internal/external rotation (peak RMSD: 4.9 mm for biceps femoris short head), whereas

gastrocnemius was influenced by changes in anterior/posterior translation (peak RMSD: 4.3 for gastrocnemius lateralis).

DISCUSSION

Moment arms estimated with *in vivo* knee kinematics from two activities showed several differences in comparison to those obtained with cadaveric passive motion and also in moment arm estimations between activities. These results are relevant because they show that muscle forces estimated with non-subject-specific joint kinematics can over or under estimate muscle forces generated by the subject. In addition, the dof that most influenced moment arm estimates were identified as superior/inferior translation, internal/external rotation, and anterior/posterior translation. This result shows which dof of the knee are crucial for more accurate estimates of moment arms and, consequently, muscle forces.

Moment arms calculated in the present study were activity dependent. A comparison between *in vivo* activities performed by the same subject reveals RMSD averaged across subjects up to 8.2 mm and R values down to 0.19. The hamstrings muscles, particularly the short and long heads of biceps femoris, were the most affected by changes in TF kinematics (Fig. 5). Moment arms changed because, at corresponding flexion angles, the tibia assumed a more anterior and more internally rotated position during the lunge (weight bearing). These changes in kinematics were not unexpected as the *in vivo* kinematics measured in the present study upholds previously reported data that compared TF motion in weight bearing and non-weight bearing activities^{16, 24}.

Moment arms estimated with subject and activity-specific kinematics also showed differences when compared to moment arms calculated with passive cadaveric motion. The largest difference in terms of RMSD was found for biceps femoris short and long head (8.8 mm and 9.0 mm on average, respectively), moreover correlation R was negative for biceps femoris short head. Nonetheless, moment arms estimated with passive cadaveric knee motion from Walker et al.³⁶ closely resembled previously reported *in vitro* measurements⁶ (dashed red line vs grey region in Fig. 4). Model estimates lay inside $\mu \pm \sigma$ areas from Buford et al.⁵ throughout the analyzed flexion range (0–120°) for most of the muscles and for the patellar tendon. Some disagreement was present in quadriceps moment arms estimated with the musculoskeletal model that exceeded those measured by Buford et al.⁵ in early flexion. However, peak quadriceps moment arms similar to our predictions (40–50 mm) have been reported using either the tendon excursion method^{9, 33} or the definition of moment arm (torque over force)¹². Implementing the same type of motion used in the experiment (passive cadaveric motion) in the knee model isolates the dependence of moment arm estimates on muscle geometry. Therefore, the close resemblance of estimates with passive cadaveric motion (TFW and PFA) prescribed to the model and *in vitro* measurements provided confidence in the modeled muscle geometry (combination of attachment locations and wrapping surfaces).

The moment arms estimated in this study represent the “effectiveness” of muscles to generate torque at the knee³¹. However, given the results of this study, the torque generated by the muscles at the joint can be substantially different depending on the knee kinematics

implemented in the musculoskeletal model. For example, since average biceps femoris moment arms estimated with *in vivo* knee extension kinematics peak at 25.3 and 30.1 mm, calculated RMSD (8.8 mm and 9.0 mm) represent 30% of the peak. A 30% increase in moment arm during a musculoskeletal simulation to estimates muscle force from inverse dynamics joint torques would approximately produce a 30% decrease in the force prediction for that same muscle. Consequently, substantial changes in TF contact loads would also be estimated. Therefore, simplifying knee kinematics by using normative data, such as provided by Walker et al.³⁶, will not provide the same estimates of moment arms and muscle forces as subject-specific knee kinematics. Similarly, depending on the kinematic assumptions implemented in the knee model, in a forward dynamics simulation the same muscle activations will produce different muscle forces and moments at the knee, producing different motions of the model.

The dof that most influenced the moment arm estimates were superior/inferior translation, internal/external rotation, and anterior/posterior translation. This finding is relevant because it provides the information needed to focus on estimating the motion of the knee that is crucial to muscle force estimation. While internal/external rotation of the knee may be measured *in vivo* with conventional motion capture methods and incorporated into the model knee¹⁰, capturing activity-specific differences in knee translations is more challenging without techniques such as stereo radiography. However, it may be possible to use a principal component approach to leverage relationships between specimen-specific TF/PF kinematics and anatomy³². Given subject-specific geometries of femur and tibia, such a technique may allow inference of TF translations in anterior/posterior and superior/inferior directions for a given knee flexion task.

The results of the present work suggest that, although muscle geometry and path in a model should be validated by comparing moment arms estimated with passive motion of the knee to cadaveric measurements, more accurate muscle force estimates for *in vivo* activities would be achieved if subject- and activity-specific knee kinematics were implemented. A previous study has shown similar dependencies of moment arm estimates on joint kinematics. Arnold et al.² compared estimates from an MRI-based specimen-specific musculoskeletal model to *in vitro* measurements from the same specimen. The study showed that using specimen-specific kinematics measured during the experiment provided substantially more accurate moment arm estimates (see Fig. 8 in Arnold et al.²).

There were two primary limitations associated with this investigation. Muscle attachments in the musculoskeletal models were not subject-specific. Moment arms mainly depend on muscle path (attachment sites, via points and wrapping surfaces). Average muscle attachment locations from cadaveric investigations were used⁸. Prior work has shown that variations in muscle geometry can affect estimates of muscle force and joint load^{22, 25}. This uncertainty in muscle geometry may propagate to moment arm estimates²⁷. To address this limitation a Monte Carlo analysis that modeled the uncertainty in muscle attachment location was performed for each subject and activity (a total of 14 Monte Carlo simulations were performed). Although large 5–95% confidence bounds were predicted for each muscle, RMSD and *R* between activities averaged across Monte Carlo instances agreed with baseline results (Fig. 5). The explanation for this result is that perturbing attachment locations

generates a shift in moment arm estimates, but the same shift happens for both the knee extension and lunge activities. Therefore, the difference between activities for each instance remains similar to the baseline difference. However, this limitation does not allow an accurate estimate of inter-subject variability because the 5–95% confidence bounds calculated for each muscle were comparable in size to the inter-subject variability estimated with baseline muscle geometry. Therefore, without knowing the actual location of muscle attachment sites, an estimate of the moment arm variability among subjects would be biased by the geometry chosen for the baseline model.

The second limitation was that some uncertainty in the *in vivo* kinematics due to tracking errors may be present. If the errors were comparable in scale to RMSD between moment arm predictions, the results showing the effect of knee kinematics on moment arms would be invalidated. However, the accuracy of the stereo radiography system was previously validated¹⁵ and the translational and rotational tracking errors are 0.151 ± 0.133 mm and $0.41^\circ \pm 0.30^\circ$. Therefore, the authors have confidence that differences observed in moment arms due to different kinematics were not an artifact of tracking errors.

In conclusion, the influence of *in vivo* subject- and activity-specific knee kinematics measured with a stereo radiography system on muscle moment arms was evaluated. The comparison with moment arms evaluated with kinematics from *in vitro* measurements of knee passive motion revealed that differences in superior/inferior translation, internal/external rotation and anterior/posterior translation had a relevant impact on moment arm trends throughout a 0–120° flexion range. In addition, intra-subject activity-specific kinematics also influenced moment arms estimates. Musculoskeletal models that lack subject-specific knee kinematics may over or under estimate muscle forces utilized by the subject.

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Figure 1.

The coordinate system of the femur was defined for all the subjects by fitting a cylinder to the medial and lateral posterior condyles³. The origin was placed at the midpoint of the centerline of the cylinder. The medial/lateral axis was defined as the axial direction of the cylinder, whereas the superior/inferior axis was aligned to the posterior line of the femur. The anterior-posterior axis was defined according to the right-handed coordinate system notation. The femoral coordinate system was assigned to tibia and patella at full extension during the knee extension activity.

Figure 2.

Workflow of the study. TF and PF kinematics from seven subjects performing seated non-weight bearing knee extension and single-leg lunge activities were collected using a high-speed stereo radiography system. Subject-specific and activity-specific knee kinematics were incorporated in a musculoskeletal model in OpenSim. TF and PF kinematics from the model presented by Arnold et al.¹ were also incorporated in the same musculoskeletal model. Moment arms were estimated in OpenSim and compared between subjects and activities. Estimates were also compared to moment arms measured in vitro and estimated with knee

passive motion (kinematics from Arnold et al.¹). Monte Carlo analyses in which uncertainty in muscle attachment locations was modeled as normal distributions with a $5mm\sigma$ were performed to evaluate the robustness of the results. Knee extension moment arms were also estimated with models in which each TF dof was replaced (one at a time) with corresponding passive motion to identify the dof that had the greatest impact on moment arms.

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Figure 3.

$\mu \pm \sigma$ TF and PF Kinematics from in vivo measurements for knee extension (dark grey area) and lunge (light grey area) as function of knee flexion, compared to passive cadaveric knee kinematics (red dashed line: TFW and PFA, used in model presented in Arnold et al.¹). $\mu \pm \sigma$ in vivo kinematics was estimated only within the minimum flexion range common to all the subjects. Statistically significant differences ($p < 0.05$) between knee extension and lunge kinematics were observed for TF internal/external rotation, anterior/posterior translation and PF flexion/extension (*). Differences between in vivo motion and kinematics

used in previous models are in TF superior/inferior and anterior/posterior translations, and in PF anterior/posterior translation.

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Figure 4.

Moment Arms estimated with in vivo kinematics and passive motion from Arnold et al.¹ compared to in vitro measurements². Blue and green areas represent $\mu \pm \sigma$ moment arms across subjects for knee extension and lunge activities, respectively. Differences between moments arms estimated with in vivo kinematics and with passive motion can be seen in most of the muscles. Statistically significant differences ($p < 0.05$) between mean moment arms for knee extension and lunge are shown (*).

Figure 5.

Average RMSD and correlation R between moment arms of the same subject during knee extension and lunge activities. RMSD and R values averaged across Monte Carlo instances in which muscle attachment locations were perturbed (light blue bars) were within 1.6% and 8.5% of baseline values for every muscle except R for biceps femoris long head.

Figure 6.

Examples of (a) Monte Carlo 5–95% confidence bounds; (c) dof changed one at a time from knee extension to TFW⁴ kinematics.

Figure 7.

RMSD and R between moment arms with knee extension kinematics and moment arms with models in which each dof was in turn changed to the corresponding dof from TFW⁴ kinematics. Large RMSD and small R indicate that the altered dof has a large influence on the difference between moment arms estimated for different kinematics. The dof that had the greatest impact on moment arms were superior/inferior translation for all the muscles, internal/external rotation for the hamstrings, and anterior/posterior translation for gastrocnemius and patellar tendon.