

A Musculoskeletal Modeling Approach for Estimating Anterior Cruciate Ligament Strains and Knee Anterior–Posterior Shear Forces in Stop-Jumps Performed by Young Recreational Female Athletes

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Abstract—The central goal of this study was to contribute to the advancements being made in determining the underlying causes of anterior cruciate ligament (ACL) injuries. ACL injuries are frequently incurred by recreational and professional young female athletes during non-contact impact activities in sports like volleyball and basketball. This musculoskeletal-neuromuscular study investigated stop-jumps and factors related to ACL injury like knee valgus and internal–external moment loads, knee anterior–posterior (AP) shear forces, ACL strains and internal forces. Motion capture data was obtained from the landing phase of stop-jumps performed by eleven young recreational female athletes and electromyography (EMG) data collected from quadriceps, hamstring and gastrocnemius muscles which were then compared to numerically estimated activations. Numerical simulation tools used were Inverse Kinematics, Computed Muscle Control and Forward Dynamics and the knee modeled as a six degree of freedom joint. Results showed averaged peak strains of $12.2 \pm 4.1\%$ in the right and $11.9 \pm 3.0\%$ in the left ACL. Averaged peak knee AP shear forces were 482.3 ± 65.7 N for the right and 430.0 ± 52.4 N for the left knees, approximately equal to 0.7–0.8 times body weight across both knees. A lack of symmetry was observed between the knees for valgus angles ($p < 0.04$), valgus moments ($p < 0.001$) and muscle activations ($p < 0.001$), all of which can be detrimental to ACL stability during impact activities. Comparisons between recorded EMG data and estimated muscle activations show the relation between electrical signal and muscle depolarization. In summary, this study outlines a musculoskeletal simulation approach that provides numerical estimations for a number of variables associated with ACL injuries in female athletes performing stop-jumps.

Keywords—Anterior cruciate ligament, Musculoskeletal simulation, Electromyography, Computed muscle control, Inverse kinematics, Forward dynamics.

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INTRODUCTION

There is continued effort to prevent anterior cruciate ligament (ACL) injury in young female athletes during high impact maneuvers like jumping, cross cutting and pivoting.^{18,22,32,44} The topmost concern during performances of impact activities is preventing knee joint muscles and ligaments from undergoing high impact forces and torques. Several research groups have been investigating the pathophysiology of ACL injury and their dedication to identifying neuromuscular and musculoskeletal factors influencing noncontact ACL injury have borne substantial rewards over the last two decades. Milestone studies done to date have delineated general differences in knee kinematics and kinetics between genders,^{5,35} showed large valgus knee motion⁹ and valgus torque,^{11,19} and low hamstring:quadriceps (H:Q) muscle activation ratios^{8,26} as factors closely related to ACL injury in young female athletes. A video analysis study²⁴ of female athletes injuring their ACL during basketball showed that the chances of female athletes incurring valgus collapse is 5.3 times more than males. A preceding study²¹ by authors of this current one showed increases in ACL strain closely related to increased valgus external moments. As researchers continue to identify causes of ACL injury and find preventive measures, there is rising demand to gain more comprehensive understanding of ACL injury pathologies.

The primary motivation behind this study was the prediction of ACL strains and internal forces as well as knee external loads in a six degree of freedom (DOF) knee (vs. our previously modeled four DOF knee) during the landing phase of stop-jumps performed by young recreational female athletes. However, an equally important goal was to study the excitation-activation relationship in

a group of knee muscles that provide the important balance of flexor vs. extensor muscle strengths in the dynamic knee. Disproportionately low ratios of flexor to extensor recruitments has been associated with ACL injuries in female athletes in previous studies.^{16,18,35,36} Of the excitation-activation pair, the excitations or signal from the central nervous system (CNS) was collected using a standard electromyography (EMG) device and activation or the actual depolarization within muscle fibers was computed using a numerical optimization schemes detailed in following sections.^{2,49} The targeted muscles were eight lower extremity knee muscles (four in each leg) including the right and left rectus femoris (RF), vastus lateralis (VL), bicep femoris (BF) and gastrocnimius (GAS). Several previous musculoskeletal studies have also made comparisons between EMG data and muscle activations in a range of 7–13 muscles. The dynamic tasks undertaken in the past studies include walking^{43,48} and landing from a jump (Laughlin *et al.*²⁷; Spagele *et al.*⁴⁷; Pflum *et al.*³⁸). One of these studies³⁸ also estimated internal forces borne by the ACL during the landing phase of drop jumps performed by a single male subject.

To attain the goals of this study, a number of musculoskeletal simulation tools incorporating aspects of full-body and segment kinematics, muscle and ligament force predictions and optimization of muscle excitations were used. The goal was to provide an idealized view of the musculoskeletal system working to accomplish a common athletic activity, with both *in vitro* and *in vivo* data. Importantly, the musculoskeletal simulation approach in this study provides information that is not directly obtainable from *in vivo* studies of human movement,^{7,42,48} for example, ACL lengthening during athletic activities with high risks of injury. Two musculoskeletal simulation tools used in this study were Inverse Kinematics (IK) and Inverse Dynamics (ID), which are often used to find joint kinematics and joint loads, respectively.⁷ Another tool used was Computed Muscle Control (CMC) which is a dynamic optimization process used to find muscle contractions and forces that produce motion at minimum metabolic cost.⁴⁸ A fourth tool was Forward Dynamics used to find joint and segment motion from a given set of muscle excitations.⁷ Results of ACL and knee biomechanics obtained using all the above tools are presented later in the study.

MATERIALS AND METHODS

Experimental Procedure: Trial Preparation

Prior to laboratory trials, permission was obtained from University of Louisville's Institutional Review Board (IRB) to conduct research trials on healthy

adult female human subjects. In accordance with IRB protocols, all subjects signed forms indicating their consent to participate in these trials. Eleven female recreational athletes with average body weight of 59.7 ± 7.7 kg (1 SD), height of 164.4 ± 12.7 cm and median age of 20 participated in stop-jump activities. Each subject was fitted with an arrangement of modified Helen Hayes system of markers (24 in total) and EMG sensors in eight muscle bellies including the right and left RF, VL, BF and GAS. To begin, each subject stood for static trials, with erect posture and feet slightly apart, to scale body segments and align joints. In addition, isometric maximum voluntary contractions (MVC) were collected to normalize EMG data from the trials, for all eight muscles. The reason behind preferring an isometric method of obtaining MVCs over a dynamic task specific one was the low reliability of widely ranging peaks from the stop-jump (dynamic) trials. These task specific dynamic peaks also caused significant drops in inter-subject variability of the EMG data which is a trait reported in other studies.^{3,40} In contrast, the isometric peaks from the MVCs consistently showed similar magnitudes. Furthermore, the use of isometric peaks for obtaining MVCs in athletes who train regularly is a recommended method under the guidelines of the International Society of Electromyography and Kinesiology (ISEK) and the Journal of Electromyography and Kinesiology (JEK). Three sets of MVC data were collected from the right and left BF, RF, VL and GAS before subjects performed the stop-jumps and three sets after. For collecting MVC on the quadriceps (RF and VL) muscles, the subjects performed maximal concentric contraction with knees slightly flexed and both hands pulling on a fixed to ground dynaband, at mid shank level. Hamstring (BF) MVCs were measured with the subject seated on the edge of a bench and contracting the hamstring while pulling back the dynaband wound around the ankles and a post in front of the subject. For the GAS muscle, each subject stood on toes, erect with hips and knees extended, and pulled on a similar strength longer dynaband. While no visual feedback was used, subjects were instructed to consciously try isolating specific muscle groups when performing MVCs.

Experimental Procedure: Equipment for Motion Capture and EMG

Marker positions, EMG and ground reaction force (GRF) signals were recorded with Evart 5.0 (Motion Analysis Corp, Santa Rosa, CA) motion capture software running in a dedicated computer. The motion capture system was equipped with eight digital Hawk cameras (Motion Analysis Corp, Santa Rosa, CA) at a

sampling rate of 100 Hz. Two force plates (Bertec, Columbus, Ohio) were used for capturing ground reaction forces, acquired at 1000 Hz frequency and preamplified before recording. A surface mounted EMG data collection system, Myomonitor (Delsys Inc., Boston, MA), was used to collect EMG data from the muscle sites mentioned above. The Myomonitor communicated with the Evart motion capture system via a wireless router (D-Link, Fountain Valley, USA) and Delsys preamplifier system (Delsys Inc., Boston, MA) at a frequency of 1000 Hz. Data from multi-camera setup, force plates and EMG were all synchronously collected through a SCB-100 connector block (National Instruments, Austin, TX) and NI 6071E DAQ adapter card (National Instruments, Austin, TX).

Experimental Procedure: Stop-Jump Trials

Subjects were instructed to perform stop-jumps by standing on a 50 cm high platform, with inside of heels apart by less than 7.5 cm. They jumped onto the force plates with one foot on each plate. Subjects were instructed not to rebound from the jump and stabilize naturally and to stay in their upright position until a hand-held sound trigger signaled completion. Landing phase duration for each trial was defined as time between feet initially touching down (TD) on force plates and recovery to an erect posture (EP). EP refers to ending time of the landing phase when the (upright) subject's weight was approximately equal to total GRF.²¹

Simulation Procedure: Inverse Kinematics Analysis

Marker trajectories captured from stop-jump trials were processed with the Inverse Kinematics tool of OpenSim 2.4 (Stanford University, Stanford, CA), an open source research software. A full body musculoskeletal model (54 muscles in 12 body segments, 23 independent DOFs and four dependent DOFs) sans arms was used. Additional DOFs added for the purposes of the current study were mediolateral translations, varus-valgus (adduction-abduction) rotation and internal-external (IE) rotation at both knees. Joint motion data for twenty-four rotational DOFs and nine translational DOFs (pelvis and two knees) were obtained. Each knee had a set of three translational and three rotational DOFs as shown in Fig. 1. All knee DOFs were independent except vertical translation which was a dependent DOF and a function of knee flexion angle to keep knee motion more physiological.⁷ Convention for determining joint angles (and external moments) was that extension, internal and varus rotations were all positive.

Simulation Procedure: Computing Muscle Activation and Forward Dynamics

One of the main goals of this study was to compare recorded EMG signals to activations resulting from CMC optimization. EMG data were filtered with a high-pass sixth-order Butterworth filter (30 Hz) to remove motion artifacts, following which the filtered EMG was full wave rectified.²⁸ EMG signals for MVC were similarly processed. Peak MVC excitation was equal to the average of the isometric MVC peaks for each muscle. Stop-jump EMG data for each muscle was then normalized using their respective MVC peak. Prior to obtaining activations from simulation, the accuracy of finding the joint external loads was improved by using a residual reduction algorithm (RRA) which used inverse dynamics to find the external loads at joints and reduced residual torques by adjusting the torso mass center at each time step.^{1,7,25} Previous studies investigating ACL internal forces^{38,43} had used a full body musculoskeletal model (sans arms) with three knee DOFs for estimating some of the joint loads and then transferred those to a lower extremity model where joint loads for DOFs not calculated earlier were found. However, this study used the full body model with modified six DOF knees for all simulations.

Computed Muscle Control (CMC) is a computational tool for muscle control optimization where feed forward and feedback controls are used to drive the kinematic trajectory of a dynamic model toward experimentally obtained ones.^{7,48} Muscle, tendon and ligament dimensions are scaled in the same manner as body segments, from the static trial marker positions. CMC computation in this study involved a static optimization scheme based on a minimizing criterion that reduced the sum of squared excitations.^{48,49} While a number of optimization technique can be used to drive a model's acceleration close to experimentally obtained ones, the static optimization technique used in this study was characterized by (a) A second order feedback control system for estimating the generalized coordinates^{7,48} and (b) Reducing a sum of the squared muscle excitations.⁷ A schematic for the algorithm is given in Fig. 2. Required for this optimization is a differential equation that relates neuronal excitation with muscle activation, given by

$$\dot{a} = \begin{cases} (u - a) \left(\frac{1}{\tau_{\text{act}}} \right) & u \geq a \\ (u - a) \left(\frac{1}{\tau_{\text{deact}}} \right) & u < a \end{cases} \quad (1)$$

where u is the excitations and comparable to EMG, a and \dot{a} represent activation and its first derivative, τ_{act} is the activation time constant equal to 40 ms and τ_{deact} is the deactivation time constant equal to 10 ms.

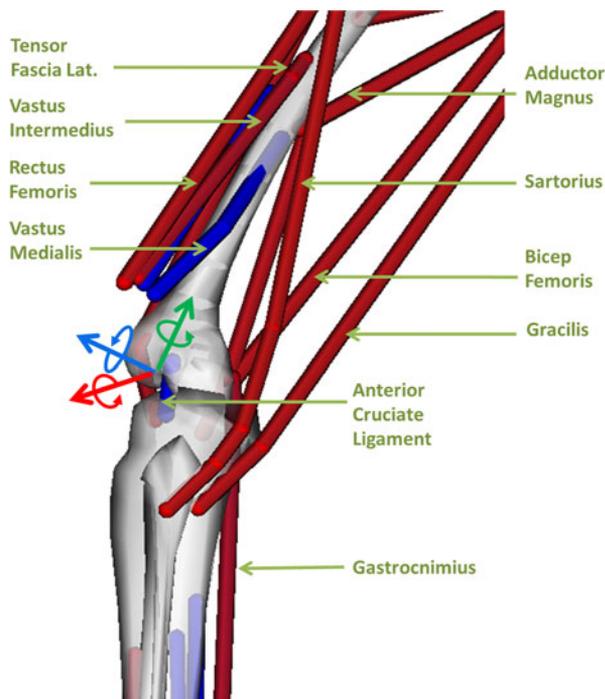


FIGURE 1. Local (joint) coordinate system for the six degrees of freedom knee. Directions key: red: frontal (x), green: transverse (y) and blue: sagittal (z). Also shown are a number of knee muscles and anterior cruciate ligaments in the right and left tibiofemoral joints.

With the completion of CMC optimization, the estimated activations, symbolizing depolarization within muscle fibers, can be compared to available EMG data or the electrical signal sent from the body's main control system, the CNS. In some cases direct techniques for estimating muscle activations from experimental EMG data can be used but such simulation methods require excitations from a majority of muscles in the model, the unavailability of which can restricts modeling to a particular segment or extremity as done in previous studies.²⁸ In the final stage, muscle activations computed from optimized excitations, were used as input to Forward Dynamics that computed kinematics in all joints and advanced motion one step ahead in time.⁷ Forward Dynamics simulation was performed in multiple steps when results were seen to diverge due to considerable perturbation.

Simulation Procedure: ACL Modeling and Properties

The ACL was modeled as a non-linearly elastic passive soft tissue with two fixed ends tunnel inserted into the femur and tibia. Top part of each ACL went into depths of inter-condyles and the lower parts were attached to front meniscules of tibia. The main two fiber bundles of ACL, anteromedial (AM) and posterolateral (PL) bundles were modeled as a single

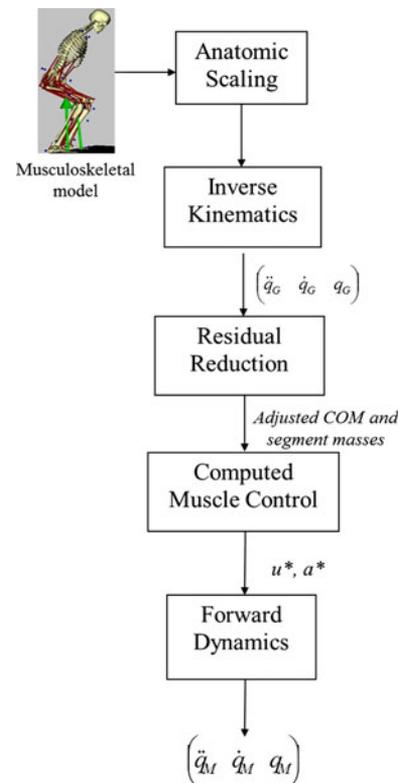


FIGURE 2. Flow chart of neuromuscular-musculoskeletal simulation scheme consisting of anatomical scaling, Inverse Kinematics (IK), Residual Reduction (RRA), Computed Muscle Control (CMC) and Forward Dynamics. Strategic outcomes of each step before progressing into the next are given next to arrows. q_G , \dot{q}_G , \ddot{q}_G represent generalized coordinates and their derivatives from IK. q_M , \dot{q}_M , \ddot{q}_M are model coordinates and their derivatives estimated from Forward Dynamics. a^* is muscle activation and u^* is muscle excitation.

entity, as they are assumed to have similar characteristics and parallel orientations. Average length of ACL was 32.3 ± 3 mm across the 11 subjects, a length slightly less than average length of AM bundle in sagittal plane as previously reported.⁶ Adopted from a Hill type muscle,¹⁷ the ACL was modeled as a passive tissue, with fully suppressed activation. ACL extensions during IK were restricted by limits prescribed to the knee joint DOFs and during CMC and Forward Dynamics simulation, the ACL strains remained primarily a function of knee joint kinematics. The ACL material property for passive fiber strain at maximum isometric force⁷ was adjusted so that passive ACL strain vs. extension-flexion, internal-external and varus-valgus angles would be limited to less than 15%, approximately.²¹ The limit of 15% strain was based on previous studies where percentage value judged sufficient for micro-fiber damage and ACL rupture to occur is between 9 and 15%.^{34,45} The ACL had elastic stiffness equal to 240 N/mm in the linear region,⁵⁰ with nonlinear stiffness characteristics for higher strains

according to Gaussian law.⁷ Further details on ACL modeling can be found in the preceding study.²¹

Post-Processing: Data and Statistical Analysis

Plots of ensemble averages (from 11 subjects) vs. percentage time of landing phase (between TD and EP) were R generated to observe trends in ACL strain and internal forces, muscle activation levels, EMG recordings and other knee variables. Statistical one-way ANOVA tests were conducted to observe variations in symmetry between left and right sides for the above variables. Experimentally obtained values of knee flexion, IE and valgus angles as well as ground reaction force (GRF) values for the vertical and forward forces were compared to those obtained from Forward Dynamics to provide validation for simulated results.

RESULTS

Results from simulation presented focus on properties of ACL, knee, muscle activations in right and left RF, VL, BF and GAS and experimental EMG. The average duration of the landing phase, between TD and EP and from 11 subjects, was 352 ± 21 ms. Ensemble averages of three sets of right and left knee angles and external moments scaled to individual body mass are given in Fig. 3a and knee AP shear forces normalized by individual body weight (BW) are given in Fig. 3b, vs. percentage time of landing phase. ACL strain and internal force scaled to BW are given in Fig. 4 where it can be seen that peak ACL strains and internal forces occurred quite early in the landing phase. Average activations from the targeted eight muscles along with original EMG signals are given in Figs. 5a–5d. Averages of reserve actuator contribution to balancing knee external moments were 23.4 ± 12.6 Nm for right and 13.4 ± 7.3 Nm for left flexion moments, 5.9 ± 3.6 Nm for right and 6.8 ± 2.0 Nm for left valgus moments, 3.6 ± 0.8 Nm for right and 2.3 ± 1.7 Nm for left IE moments. Table 1 shows the minimum, maximum and averages with ± 1 SD for the variables mentioned above. The $p \approx 0$ (<0.001) results of one-way ANOVA tests in Table 1 indicate that most of the knee variables investigated lacked symmetry between the two sides, with the exceptions of knee flexion and IE angles and moments and ACL strain and internal force. Figure 6 shows average knee joint angles and vertical and forward GRF forces calculated with Forward Dynamics to validate results of simulation⁴¹ and how closely they match the experimental traces. Coefficient of determination, r^2 calculated between experimental and Forward Dynamics knee angles for flexion, IE and valgus angles

were $r^2 > 0.92$. Similarly r^2 calculated between experimental and Forward Dynamics vertical and forward GRF forces were $r^2 > 0.88$. A movie of the landing phase is given in Appendix A in Electronic Supplementary Material. Plots of the internal moments produced by BF, GAS, VL and RF muscles for the subject in the movie are also given in Appendix A in Electronic Supplementary Material.

DISCUSSION

This section discusses the major findings from this study, including knee kinematics and kinetics, muscle excitations and activations, ACL strains and internal forces and issues of symmetry. Finally, a discussion on validation process, study limitations and a brief conclusion is presented.

Knee Kinematics and External Loads

Peak flexion and valgus angles measured during the landing phase are similar to findings in other studies^{16,24,32} on landing phase. Peak valgus external moment occurring within the first 30% of total time of landing phase is comparable to a previous study¹³ and also to findings in cadaveric studies.^{11,20} Another aspect common between this and both cadaveric and non-cadaveric studies,^{14,19,24} is the relationship between increased valgus external moment and increased ACL strain. Previous simulation studies⁴⁶ on combined valgus and IE loading has shown ACL strain of 10.5%, a magnitude which along with peak valgus and IE loads are similar to those found in this study. The peak AP shear force at the knee (only tibiofemoral since the patella was not modeled) due to medial and lateral condyles contacting the tibial articulating surface¹⁰ was approximately 0.7–0.8 BW and occurred within the first 25% of the landing phase (Fig. 3b). This peak is comparable to peaks ranging 0.24–1.0 BW reported in other studies.^{5,38,51} Additionally, the tibiofemoral shear force remained anterior during the entire landing phase in comparison to the ground reaction force which was always posterior. In the preceding study²¹ the net external knee moment was vector resolved into flexion and valgus directions only. However, this study resolved external knee moment in the directions of three rotational DOFs and some differences in magnitude of moments can be seen between the two studies.

Muscle Excitation and Activations

Comparisons of EMG signals with activations in Fig. 5 show fairly good conformity to the theory of muscle activation following excitation,^{49,52} particularly

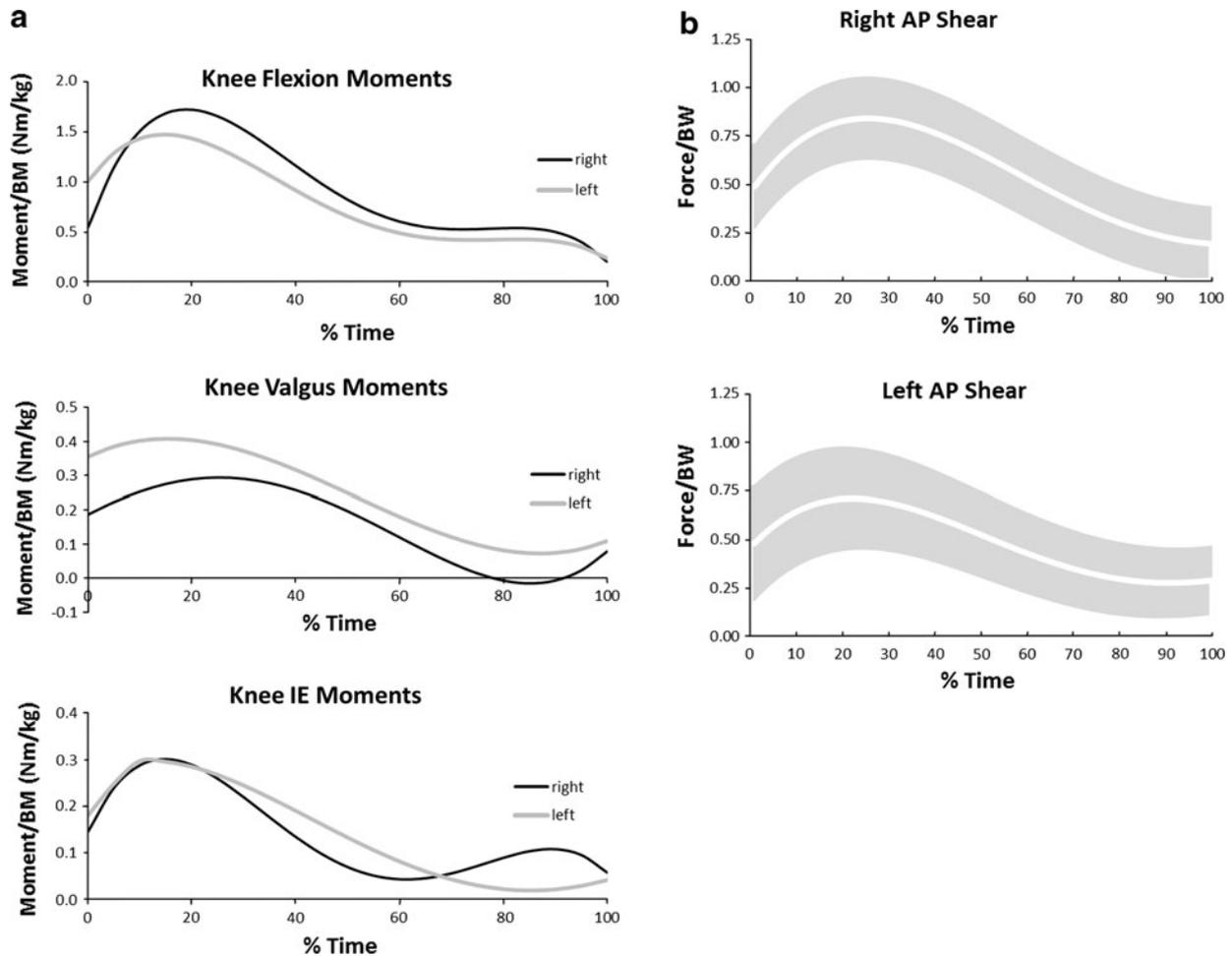


FIGURE 3. Ensemble average of right and left knee. (a) flexion moments (± 1 SD), valgus moments (± 1 SD) and internal–external (IE) moments (± 1 SD) and (b) anterior–posterior (AP) shear force (± 1 SD) against percentage time of landing phase obtained from 11 subjects participating in stop-jumps.

under the circumstances of activation not being computed from real time EMG. The BF activation peaked within the first 30–40% of landing phase where the main role of BF is production of flexion moments as the knee flexes and similar increases occurred in GAS activation. Increased activity in RF and VL, the extensor muscles that provide breaking forces also happened early in the landing phase. The above trends in activation of the eight muscles are similar to activations reported in previous studies (Sasaki and Neptune⁴¹; Palmieri-Smith *et al.*³⁷; Thelen and Anderson⁴⁸). Of particular note is the increase in muscle activations in the RF, VL, BF and GAS around the time of maximum ACL strain and internal force. Increased activation in the RF, VL and BF muscles around peak valgus load indicate that these muscles play a major role in balancing (frontal) valgus load in the knee.²⁹ It can be seen from the minimum H:Q ratios in Table 1 as well as timing of RF and BF activation peaks (Fig. 6), that the subjects were exposed to

some complications of low H:Q ratios known to cause ACL injury (Ebben *et al.*⁸; Nagano *et al.*³³) and in particular for the right side with a prolonged rectus femoris excitation peak. Comparison of RF and BF activations between the current and preceding study²¹ shows some differences in activation peaks and timing. Although caution must be exercised in comparing activations between the two studies since muscle lengths and contraction levels are sensitive to the differences in number of knee DOFs.

ACL Strain and Internal Force

Peak ACL strain and internal forces occurred within the first 20% of landing phase, at a slightly high flexion angle of 70°. The flexion angle at which high valgus and accompanying high ACL strains occur is quite a debated issue. A number of studies,^{24,36} claim high valgus and ACL strains occur at 30–60° of flexion. Other studies,^{16,30,39} have shown that a fixed range of

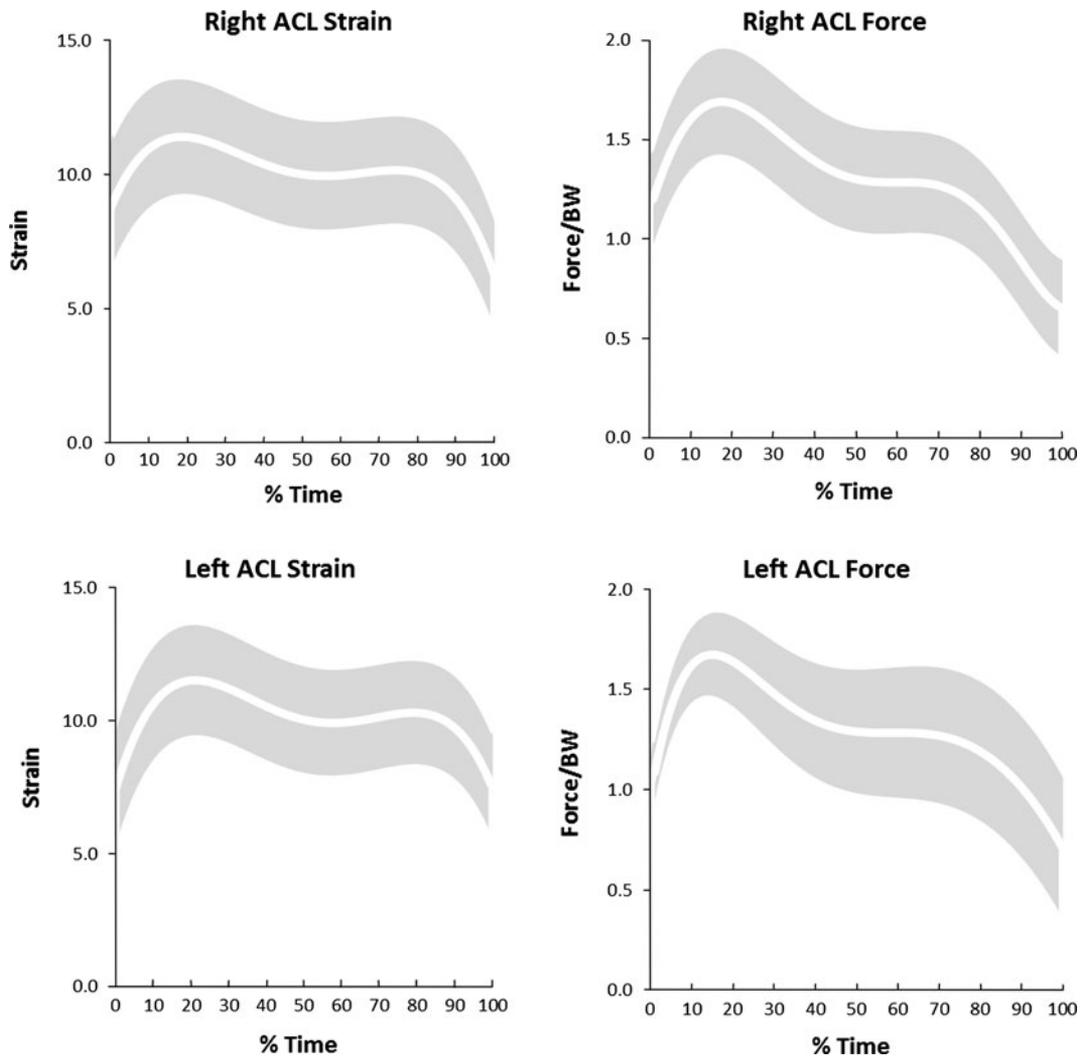


FIGURE 4. Ensemble averages of right and left anterior cruciate ligament fiber (internal) forces and strains shown with ± 1 SD from 11 subjects performing the stop-jumps.

knee flexion angles at landing cannot be the primary criterion predicting ACL injury risks. A more recent study³² showed that excessive mediolateral translation that brings the knees closer in a knock-knee position can cause high valgus with higher risks of ACL injury even at high flexion angles. The current study also showed a slightly lower ACL strain at EP than the preceding study,²¹ which could be due to decreases in both valgus and IE angular displacements in the knee near EP, compared to only valgus decrease in the preceding study.

Symmetry

Lack of symmetry has been associated with ACL injury in young female athletes where complications such as quadriceps dominance³¹ or even leg strength dominance^{15,31} in one leg over the other have been

reported. Quadriceps dominance was not a potential problem during the current study, since H:Q ratio (Table 1) which is often used as a measure of quadriceps dominance was not found to be severely low. However, the difference found in external loads between the knees (Table 1), particularly in valgus moments, could imply leg dominance in which case the dominant knee could be susceptible to ACL injury risks.³¹

Study Validation, Limitations and Conclusion

Comparisons between averaged experimental and Forward Dynamics knee angles and vertical and horizontal GRF forces yielded fairly good correlations for validating results from simulations (Fig. 6). Moreover, the validation technique used is a scientifically accepted one and according to conventions followed

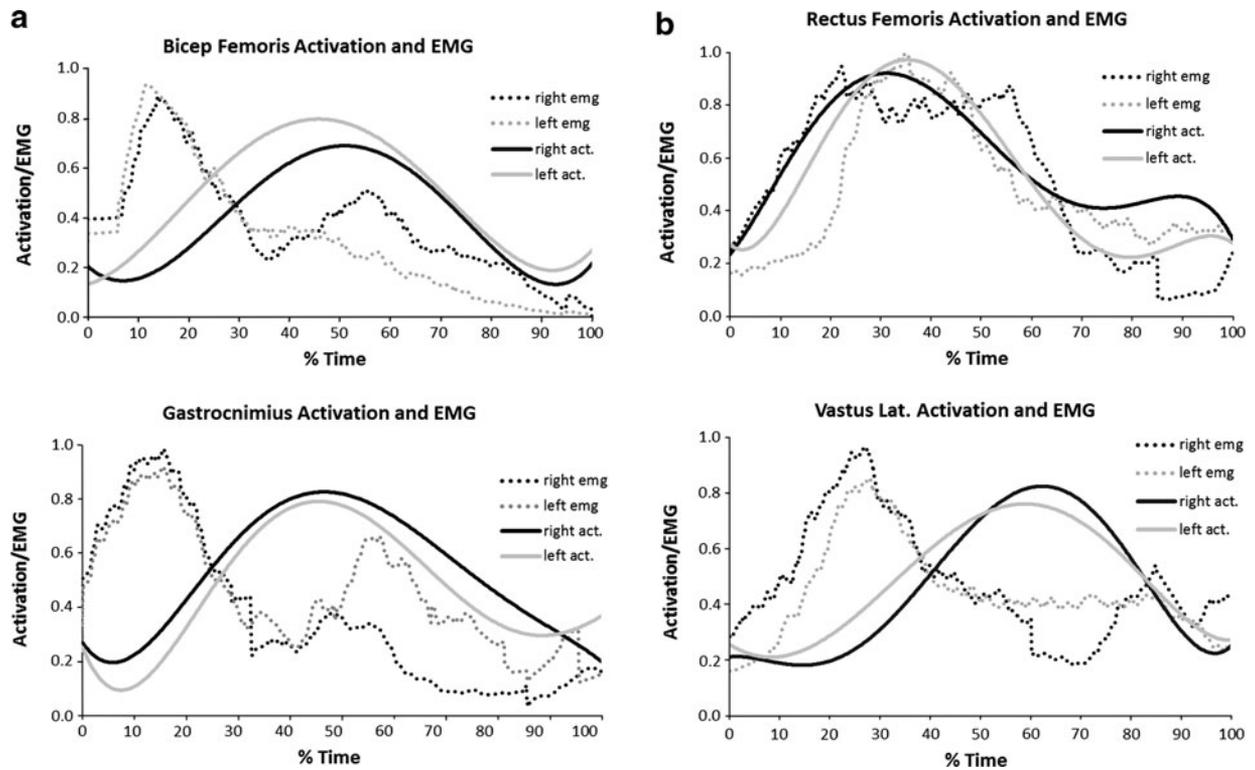


FIGURE 5. Comparisons between ensemble averages of experimentally obtained electromyography representing central nervous system (CNS) signals and activations estimated with computed muscle control (CMC) in simulation against percentage time of landing phase obtained from 11 subjects participating in stop-jumps.

TABLE 1. Statistics of knee joint, ligament (ACL) and muscle variables from the landing phase of stop-jump trials performed by 11 subjects.

Variables Joint/Ligament/Musc.	Right Max	Min	Avg.	SD	Left Max	Min	Avg.	SD	Symmetric p -value
Knee									
Ext-flex angle (°)	11.5	-91.2	-55.8	28.9	15.2	-89.6	-57.5	25.3	0.80
IE angle (°)	5.4	-0.9	3.5	3.7	5.2	-1.5	3.5	3.4	0.80
Varus-valgus angle (°)	8.4	-9.9	-0.5	8.3	4.8	-8.6	-3.9	7.2	0.00
Ext-flex moment (Nm)	11.0	-84.3	-45.4	13.5	15.0	89.5	-46.6	15.7	0.80
IE moment (Nm)	18.9	2.9	8.6	3.1	18.5	-0.7	10.6	2.3	0.40
Varus-valgus moment (Nm)	2.9	-17.5	-9.7	1.0	5.2	-20.3	-11.2	2.2	0.00
AP shear force (N)	496.0	79.3	283.5	64.5	451.0	88.0	239.8	82.8	0.00
ACL									
Fiber force (N)	1149.8	630.4	973.0	131.5	1135.6	623.9	983.3	112.9	0.70
Fiber strain (%)	12.7	7.9	10.2	3.3	12.0	5.8	10.3	2.4	0.10
Musc. activation/EMG									
Bicep femoris (EMG)	0.98	0.03	0.37	0.17	0.99	0.01	0.27	0.15	0.00
Bicep femoris (Sim)	0.87	0.16	0.49	0.05	0.80	0.17	0.38	0.02	0.00
Rectus femoris (EMG)	0.99	0.07	0.53	0.09	1.00	0.16	0.48	0.09	0.00
Rectus femoris (Sim)	0.92	0.09	0.50	0.06	0.93	0.10	0.39	0.09	0.00
Vastus lateralis (EMG)	0.96	0.18	0.50	0.07	0.71	0.13	0.37	0.12	0.00
Vastus lateralis (Sim)	0.87	0.09	0.41	0.15	0.85	0.14	0.48	0.04	0.00
Gastrocnimius (EMG)	0.98	0.04	0.36	0.14	0.96	0.12	0.47	0.15	0.00
Gastrocnimius (Sim)	0.93	0.10	0.49	0.06	0.99	0.18	0.50	0.16	0.00
H:Q ratio (EMG)	1.44	0.32	0.92	0.43	1.87	0.36	1.17	0.61	0.00

The p values in the last column depict the symmetry between the right and left of each variable. Sign convention followed was positive for extension, varus and internal rotation.

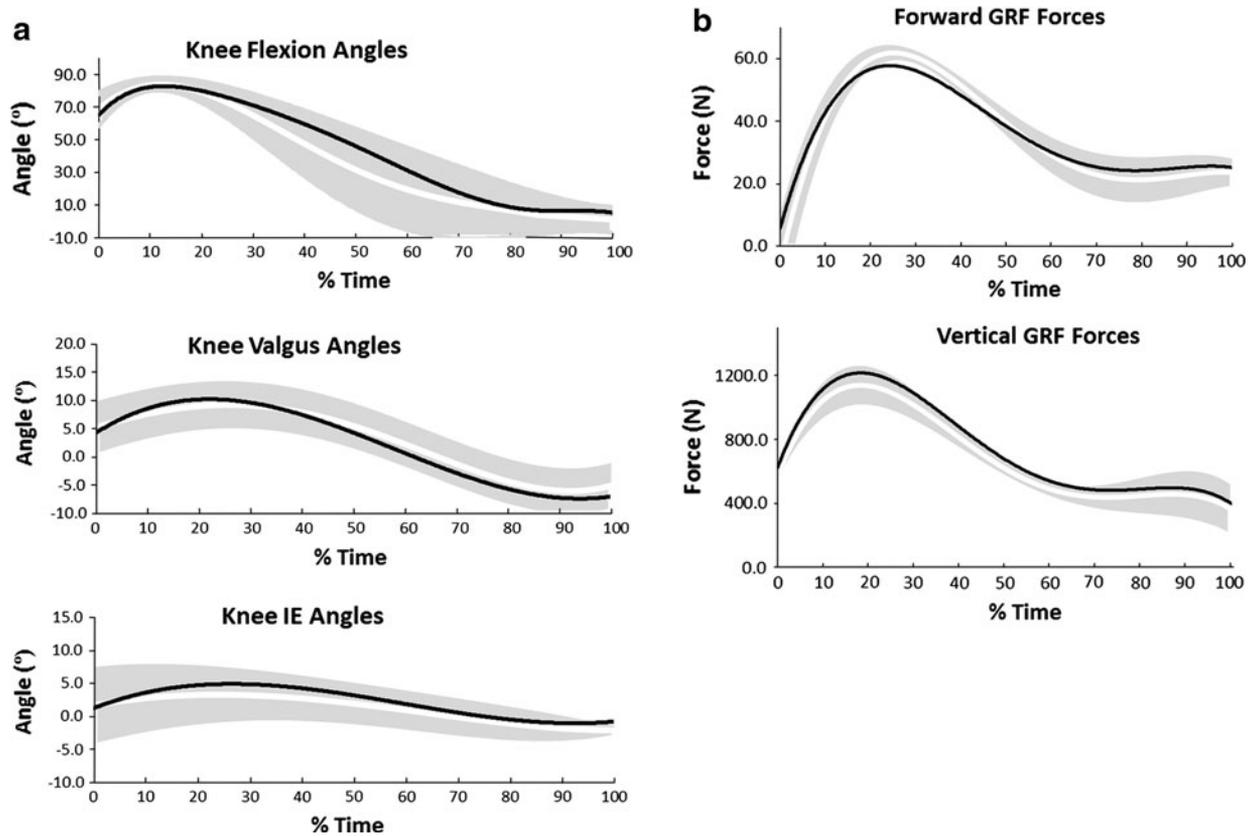


FIGURE 6. Comparison between ensemble averages of right knee angles including flexion, valgus and internal rotation angles and force plate forces in the transverse and forward directions obtained from experimental results and Forward Dynamics versus percentage time of landing phase. All averaged from 11 subjects participating in stop-jump trials. The solid white lines represent averages from Forward Dynamics analysis with ± 1 SD. The solid black lines show the averages of experimental results.

by a previous study (Sasaki and Neptune⁴¹). However, other robust methods for validation such as comparing the computed (external) joint loads to results of inverse dynamics have been used in previous studies.²⁸ Although, the latter form of validation is in-built within CMC which can be programed to keep usage of reserve actuators at very low levels.

The foremost limitation of this study was lack of EMG data from the remaining muscles which could have been used to simulate an EMG driven stop-jump task. An alternative hybrid modeling approach could have been used with part EMG driven activation with remaining excitation-activation calculated with the CMC optimization scheme. However, such methodology requires multiple iterations until activation data estimated from EMG signals converge to acceptable levels. Hence, the possibility of estimated muscle forces not equaling actual ones for muscles for which there is no comparable EMG data remain strong. However, in the case of erroneously calculated muscle forces, the use of reserve actuators during CMC to compensate for lacking muscle strength would have been disproportionately high and Forward Dynamics estimated joint kinematics would have little resemblance to the actual stop-jump tasks.⁷ A second

limitation was finding muscle activations using the excitation-activation relationship given in Eq. (1) where other methods for decoupling joint loads exist. For example, use of physiological data from EMG-dynamometer recordings²⁸ and muscle work history dependent activation estimation schemes.¹² A third limitation was not attempting to estimate physiological tibiofemoral compressive joint reaction force by replacing the fictitious patellofemoral joint in the current model with a patellofemoral one as done in a previous study.²³ The isometric method of obtaining MVCs can be considered a fourth limitation where dynamic stop-jump task specific maximum contractions could have been used. Reasons for preferring the isometric method for purposes of reliability and increasing inter-individual variability (Burden³; Chapman *et al.*⁴) has been mentioned in the “Materials and Methods” section. Another major concern was whether maximum activations could be produced by other dynamic activities such as sprinting, squat-jumping and cycling, none of which were scheduled for this study. Additionally, studies⁴⁰ that have compared isometric vs. dynamic methods of obtaining MVCs have found small differences in peak magnitudes. A fifth limitation was that muscle activation for the non-EMG muscles could

not be compared to patterns of EMG (excitation) from other studies, mainly due to differing trial conditions. For example, it would be beneficial to observe excitation-activation in muscles like the gluteus whose lack of activation has been linked with potential ACL injury.¹⁶

This musculoskeletal-neuromuscular modeling study estimated several physiological variables in the individual athlete's knee and ACL that are significant to understanding ACL injuries in women's stop-jump activities. A previous full body musculoskeletal model with the ACLs incorporated in it²¹ and OpenSim simulation tools were used to compute ACL and knee dynamic variables. A minimizing criterion that reduced the sum of squared excitations was used to predict joint kinematics and kinetics. Instead of traditional inverse dynamics that estimates only joint loads and kinematics, this study used a CMC-Forward Dynamics methodology to estimate details of muscle activations and states of soft tissues like the ACL. Comparison between *in vivo* EMG data and predicted muscle activation showed fairly good conformation to the scientifically established rule of muscle activation following CNS signaling. However, it is not always possible to obtain surface EMGs, particularly in muscles that are difficult to access like the vastus intermedius. Filling the gaps for difficult to obtain physiological data in musculoskeletal-neuromuscular modeling can be challenging. To avoid the problem, a mixed approach to muscle control optimization with part EMG driven and part predicted activation should be considered. Future studies for assessing the risk of ACL injury during impact activities in women's sports will benefit from building individual musculoskeletal models that also incorporates a wide range of physiological data. It will help identify both common physiological weaknesses characteristic of a particular activity like stop-jumps as well as weaknesses that are unique to the individual athlete.

ELECTRONIC SUPPLEMENTARY MATERIAL

The online version of this article (doi:[10.1007/s10439-012-0644-y](https://doi.org/10.1007/s10439-012-0644-y)) contains supplementary material, which is available to authorized users.

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