

A Numerical Simulation Approach to Studying Anterior Cruciate Ligament Strains and Internal Forces Among Young Recreational Women Performing Valgus Inducing Stop-Jump Activities

Julia Kar & Peter M. Quesada

Annals of Biomedical Engineering

The Journal of the Biomedical Engineering Society

ISSN 0090-6964

Ann Biomed Eng

DOI 10.1007/s10439-012-0572-x



Your article is protected by copyright and all rights are held exclusively by Biomedical Engineering Society. This e-offprint is for personal use only and shall not be self-archived in electronic repositories. If you wish to self-archive your work, please use the accepted author's version for posting to your own website or your institution's repository. You may further deposit the accepted author's version on a funder's repository at a funder's request, provided it is not made publicly available until 12 months after publication.

A Numerical Simulation Approach to Studying Anterior Cruciate Ligament Strains and Internal Forces Among Young Recreational Women Performing Valgus Inducing Stop-Jump Activities

JULIA KAR¹ and PETER M. QUESADA²

¹Thayer School of Engineering, Dartmouth College, 8000 Cummings Hall, Hanover, NH 03755, USA; and ²Department of Mechanical Engineering, Speed School of Engineering, University of Louisville, 200 Sackett Hall, Louisville, KY 40292, USA

(Received 29 December 2011; accepted 10 April 2012)

Associate Editor Joel D. Stitzel oversaw the review of this article.

Abstract—Anterior cruciate ligament (ACL) injuries are commonly incurred by recreational and professional women athletes during non-contact jumping maneuvers in sports like basketball and volleyball, where incidences of ACL injury is more frequent to females compared to males. What remains a numerical challenge is *in vivo* calculation of ACL strain and internal force. This study investigated effects of increasing stop-jump height on neuromuscular and bio-mechanical properties of knee and ACL, when performed by young female recreational athletes. The underlying hypothesis is increasing stop-jump (platform) height increases knee valgus angles and external moments which also increases ACL strain and internal force. Using numerical analysis tools comprised of Inverse Kinematics, Computed Muscle Control and Forward Dynamics, a novel approach is presented for computing ACL strain and internal force based on (1) knee joint kinematics and (2) optimization of muscle activation, with ACL insertion into musculoskeletal model. Results showed increases in knee valgus external moments and angles with increasing stop-jump height. Increase in stop-jump height from 30 to 50 cm lead to increase in average peak valgus external moment from 40.5 ± 3.2 to 43.2 ± 3.7 Nm which was co-incident with increase in average peak ACL strain, from 9.3 ± 3.1 to $13.7 \pm 1.1\%$, and average peak ACL internal force, from 1056.1 ± 71.4 to 1165.4 ± 123.8 N for the right side with comparable increases in the left. In effect this study demonstrates a technique for estimating dynamic changes to knee and ACL variables by conducting musculoskeletal simulation on motion analysis data, collected from actual stop-jump tasks performed by young recreational women athletes.

Keywords—Anterior cruciate ligament, Biomechanics, Musculoskeletal simulation, Optimization, Inverse Kinematics, Forward Dynamics.

Address correspondence to Julia Kar, Thayer School of Engineering, Dartmouth College, 8000 Cummings Hall, Hanover, NH 03755, USA. Electronic mail: julia.kar@dartmouth.edu, pmques01@louisville.edu

INTRODUCTION

Young female athletes (professional and recreational) suffer anterior cruciate ligament (ACL) injuries 4–6 times greater than male athletes in high impact activities like jumping and side-cutting.¹ Most ACL injuries, occurring in both female and male athletes, are classified as non-contact and generally occur during high impact maneuvers like landing phase of jumps, cross cutting and pivoting.^{5,10,13,23,25} Impact from these activities can make knee joint's supporting muscles and ligaments subject to very high forces and torques.

Although the precise patho-physiology of ACL injuries remains unclear, several neuromuscular and biomechanical factors are involved in the process. Research groups have investigated variables such as high valgus external moments in the knee,^{15,20} low knee flexion angles⁴⁰ and low hamstring to quadriceps activation (H:Q) ratio^{9,18} as major contributors to ACL injury. What the above studies could not show was the relation between the variables identified to cause injury with ACL strain and internal force. In addition, a variety of preventive measures have also been tried, which claim that lack of either proprioception or lower extremity neuromuscular training^{6,15} can lead to lower extremity movements with risk of ACL injury. While a number of these preventive measures have been known to reduce hip, knee and ankle loads and improve H:Q ratio, these studies could not provide quantitative assessments of any reduction in ACL strain and internal force. Similarly, in a study where female athletes practiced drop-jumps using bilateral medial posts,²⁰ decreases were seen in knee

valgus angle and ankle pronation-eversion but scientific evidence on reduced ACL strain could not be provided. One study group³⁴ showed increased knee valgus load leading to increased ACL strain rates in an assembled knee model. However, this model could not provide lower extremity structural variability found in *in vivo* cases. Hence, the remaining challenge is to find methods by which the ACL strain and internal force can be quantified.

Primary focus of this study was to investigate ACL strain and internal force developed during real time stop-jump activities, under combined flexion-extension and valgus external moments and by integrating right and left ACLs into the human musculoskeletal model. Numerical simulations then estimated kinematic and kinetic variables and provided a basis for comparing ACL strains with knee valgus and flexion external moments. Motivation behind this study was the limited amount of research^{12,23} done toward directly quantifying ACL strain and internal force during athletic activities, particularly in comparison to volume of work done on measuring peripheral musculoskeletal variables identified as contributors to ACL injury.

MATERIALS AND METHODS

General Outline

This study involved two distinct parts; the first part being motion analysis trials conducted on 11 young recreational female athletes performing stop-jumps from various heights. Second part of this study was numerical simulation of stop-jumps using Inverse Kinematics (IK), Computed Muscle Control (CMC) and Forward Dynamics.^{8,37} Simulation steps enabled estimation of knee joint kinetics and kinematics as well as ACL strain and internal force, with increasing stop-jump height as the parameter for varying intensity of activity. Results were then (statistically) analyzed and compared to findings of other similar studies.^{9,14,23,24,33,40}

Trial Procedure

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. Parental consent was not required since all subjects were 18 years of age or above.

Eleven female subjects with average body weight of 59.7 ± 7.7 (1 SD) kg, height of 164.4 ± 12.7 cm and

median age of 20 participated in stop-jump activities from platforms of height 30, 40 and 50 cm. The subjects were classified as recreational athletes since they participated in group or individual athletic activities on a regular basis. Each subject was fitted with 24 reflective markers placed at strategic anatomical locations as shown in Fig. 1a. Prior to stop-jump trials, each subject performed three static trials for a second each to scale and align her body segments and joint locations with each subject standing in an erect posture (EP) with feet slightly apart. Body segment and scaling of kinematic constraints at joints for the musculoskeletal model (Appendix A in Supplementary material 1) of each subject were conducted with static marker positions. Figures 1b and 1c show details of knee joint in the musculoskeletal model, in sagittal and frontal planes, respectively. For stop-jumps, the subjects jumped from platforms of 30, 40 and 50 cm height onto a pair of force plates (Bertec, Columbus, OH), one foot on each, and stabilized according to their natural abilities without rebound. All consenting subjects attended a brief pre-trial session where they were instructed on drop-jumping technique and practiced to make sure they were comfortable with drop-jumps. On the day of recorded trials, each subject performed three jumps at each height to allow averaging of variables for data analysis and simulation.

Equipment

Marker positions and force plate signals were recorded with Evart 5.0 (Motion Analysis Corp, Santa Rosa, CA) motion capture software, which ran in a data acquisition computer. The motion capture system was equipped with 8 digital Hawk cameras (Motion Analysis Corp, Santa Rosa, CA) and sampling rate was set at 100 Hz. Force plate output was acquired at 1000 Hz frequency and amplified prior to capture. An electromyography (EMG) data collection system, Myomonitor (Delsys Inc., Boston, MA), was used for collecting EMG data from a number of muscle sites for further investigation. Data from the multi-camera setup, force plates and EMG equipment were all collected through a SCB-100 connector block (National Instruments, Austin, TX) and NI 6071E DAQ adapter card (National Instruments, Austin, TX). Temporal alignment was ensured through use of a single dedicated comprehensive system for simultaneous collection, display and recording of all data.

Motion Capture

During trials subjects were instructed to stand on platforms with toes and inside of heels apart by

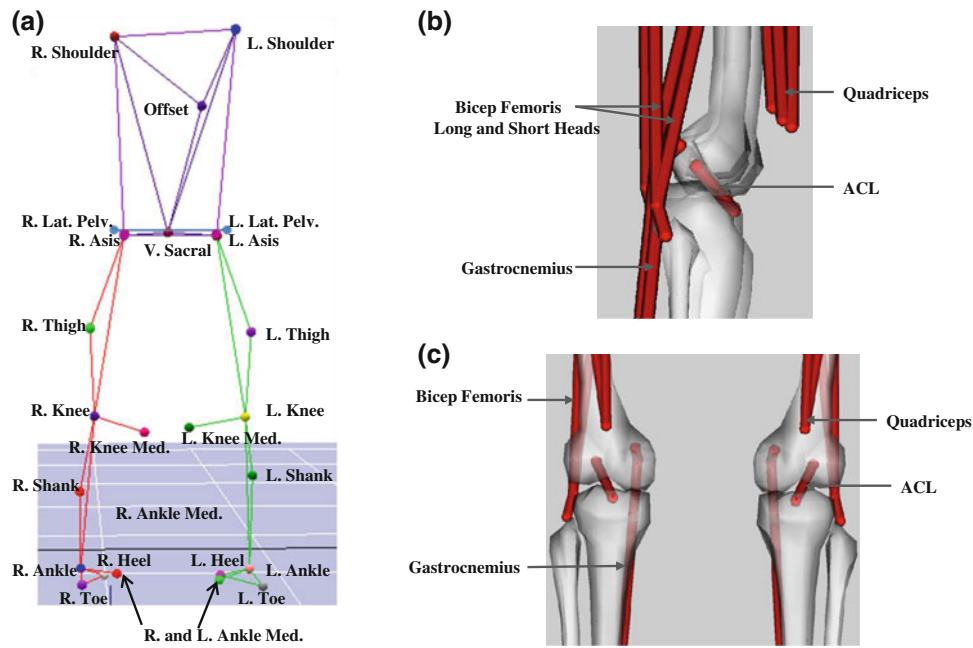


FIGURE 1. (a) Location of markers used for the drop jump trials. Insertion points of the ACL in the inter-condyloid eminence of the femur and front meniscus of the tibia shown in (b): frontal, and (c) sagittal views.

<7.5 cm, and drop directly down on to the force plates. Subjects regained their upright stance after initial contact and braking on the plates, following which they were asked to remain in upright stance for 2 s or more before walking away. Figure 2 shows some

of the main stages of a typical stop-jump trajectory. Motion analysis was conducted between initial touch-down (TD) (Fig. 2c) and subject's regaining an EP (Fig. 2e). Duration of trial was defined as time between TD at initial contact with force plates and recovery to

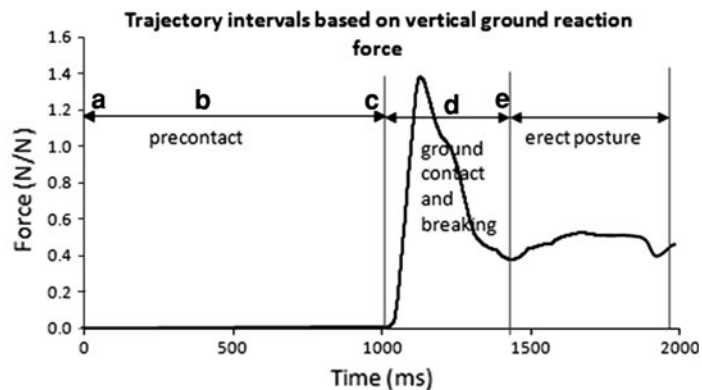
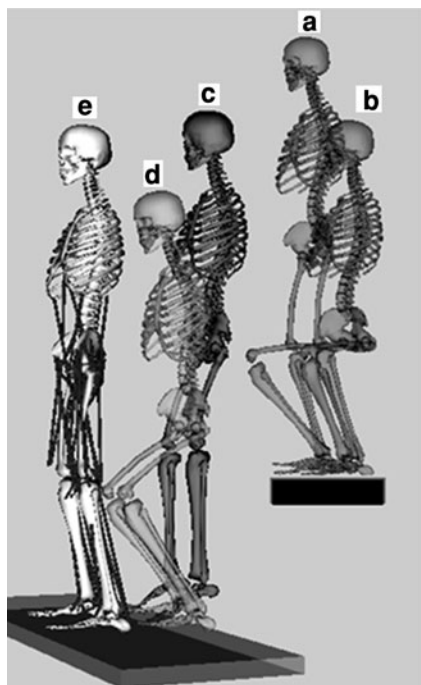


FIGURE 2. Sequence of events for the entire duration of the drop jump created using OpenSim 2.20 graphical interface. In chronological sequence, (a): upright position before jump, (b): crouch before takeoff, (c): initial contact at landing, (d): braking on force plates, and (e): return to upright position on ground.

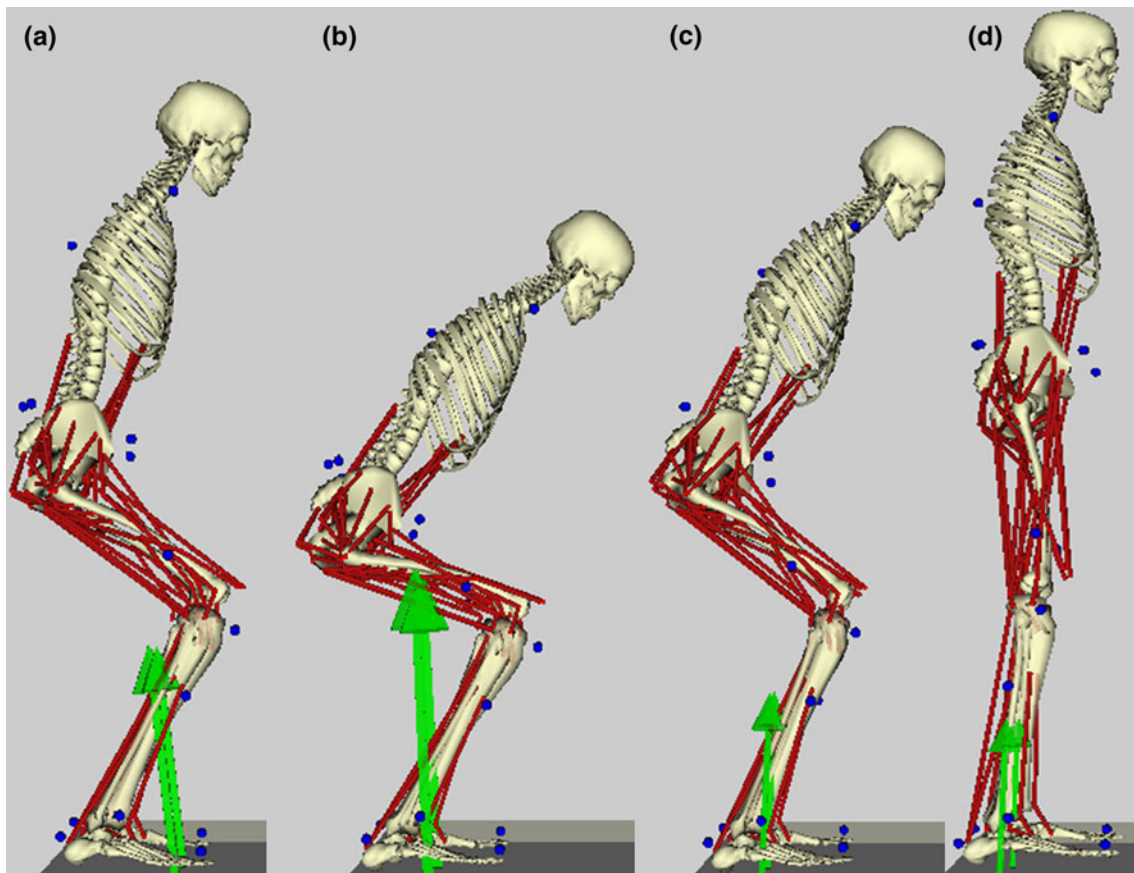


FIGURE 3. Force plate contact dynamics of the jump trajectory for Forward Dynamics simulation created using OpenSim 2.20. Muscles are shown in red, body segments in gray, markers in blue and force plate in black. The green arrows indicate net ground reaction forces. In chronological sequence, (a): TD, (b): maximum crouch at braking, (c): recovering to upright stance, and (d): upright stance.

an EP as shown in Fig. 3. Recovery or EP refers to point in time when each subject had regained an upright stance and combined ground reaction forces (from two force plates) were equal to subject's weight.

Kinematic Analysis

Marker trajectories captured from stop-jump trials were processed with the IK tool of OpenSim version 2.20 (Stanford University, Stanford, CA), an open-source research software. Joint motion data for twenty-three rotational DOFs, three translational DOFs and four dependent translational DOFs at the knee joint were obtained. Knees were assigned two independent rotational DOFs in flexion and valgus directions. Convention for determining joint angles (and external moments) was that extension and varus (adduction) were both positive. Flexion was included as knee DOF due to its inherent role in stop-jumps. Valgus was chosen as the second DOF since several studies have shown that valgus knee motion and valgus external moments are directly associated with ACL

injury.^{15,24,30} One particular cadaveric study³⁸ showed that ACL strain increased by 30% due to a combined flexion-valgus torque than a flexion torque by itself. The knee also had two dependent translational DOFs which were functions of the knee flexion angle. These dependent DOFs were in the sagittal plane, in directions normal to frontal (coronal) and transverse planes, an existing feature of base OpenSim model.³² Normal directions to frontal, sagittal, and transverse planes are shown in Appendix A in Supplementary material 1. The choice of knee DOFs were in conjunction with a six DOF pelvis where quadriceps (attached to pelvis) supported majority of vertical and horizontal forces arising from body weight and braking.³²

Computed Muscle Control and Forward Dynamics Simulation—Existing Framework

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 that obtained experimentally.^{3,4,36} Pri-

mary target of CMC algorithm is muscle activation optimization to resolve redundancies in dynamic force generating capacity of muscle actuators.⁸

Musculotendon dimensions of individual subjects, scaled from anatomical segments determined from pre-trial marker positions, formed part of input to simulation. Muscles are modeled as homogenous collection of motor units which were similar in structural, functional and excitation properties.^{41,42} Simulation started with solving fundamental equation of state that yields accelerations of generalized coordinates at any point in time, followed by solving of a closed loop equation that minimized error between model and experimental kinematics.^{8,36} Integrated into closed loop equation was optimization of a dimensionless performance criterion, J that minimizes square of muscle activations.³⁷ These activations, along with passive components defined steady-state muscle forces. CMC also includes use of reserve actuators located at joint DOFs that make up for muscle strength deficiencies.³⁷ In the final stage, neural excitations (computed from optimized muscle activations) were input to Forward Dynamics. Forward Dynamics is a numerical integration scheme which uses of muscle forces to provide accelerations of joint coordinates to advance motion one step ahead in time.⁸

ACL Modeling and Properties

Simulation application used was OpenSim (detailed earlier) where a baseline musculoskeletal model (given in Appendix A in Supplementary material 1) already exists. ACL was modeled as a passive soft tissue with two fixed ends tunnel inserted into femur and tibia

(Figs. 1b, 1c). Top part of each ACL went into depths of inter-condyles of femur while lower part was attached to front meniscles of tibia. Actual adhesion mechanism was not relevant to this study as the goal was to compute overall strains, assuming fixed ends. The main two fiber bundles of ACL, anteromedial (AM) and posterolateral (PL) bundles were not modeled separately, as they were assumed to have the similar characteristic. Average length of ACL was 32.3 ± 3 mm across the 11 subjects which is slightly less than average length of AM bundle in sagittal plane as reported in other studies.⁷ Width or cross section of ACL (assumed at 10 mm) was irrelevant to this study as computing stresses were not part of this study.

In general the internal force generated within a Hill type muscle¹⁷ is given by

$$f_m^* = \{a_m^* f_{lv}(l_m^*, \dot{l}_m^*) + f_{psv}(l_m^*)\} \cos(\alpha_m^*) \quad (1)$$

where l_m^* is muscle length, \dot{l}_m^* is muscle tendon velocity in fiber direction, a_m^* is muscle activation, f_{lv} is active force from force-length-velocity curve of Hill's model, f_{psv} is passive fiber force and α_m^* is muscle's pennation angle. To make ACL a passive tissue, contractile element activation was fully suppressed. Hence, only the second force term in Eq. (1) is applicable to ACL and due to its zero activation ACL was not part of the optimization process.

ACL extensions during IK were restricted by limits of the knee joint's DOFs. Hence ACL strain, due to knee valgus, flexion and two translations dependent on flexion (Figs. 4a, 4b), was strictly a property of knee joint motion and ACL orientation (Figs. 1b, 1c).

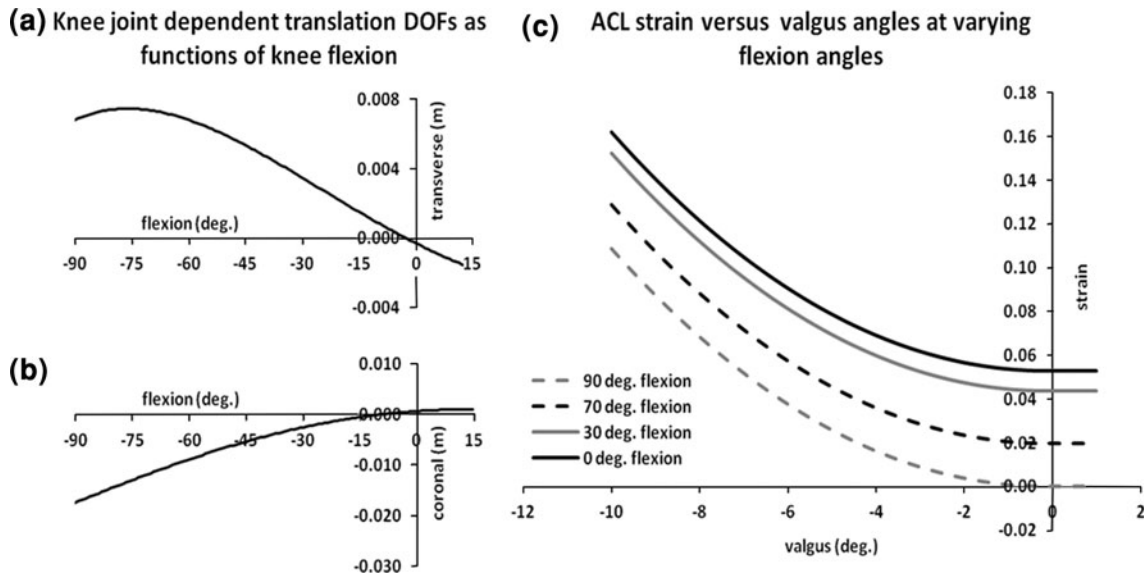


FIGURE 4. The two translational (right) knee DOFs for one subject, dependent on (functions of) the knee's flexion DOF. (a) The first direction was anterior–posterior, normal to the frontal (coronal) plane and (b) second to the transverse plane. (c) The variations in ACL strain due to valgus and flexion DOFs and two (flexion) dependent translational DOFs for the same subject.

Figure 4c shows the variation in ACL strain with valgus angles at several flexion angles for one subject. For example, if combined knee joint kinematics and ACL insertion locations caused a knee flexion of 70° and valgus of 10° for the 50 cm stop-jump, then the resulting ACL strain would be 12% (Fig. 4c). With results of Forward Dynamics simulation following CMC, the ACL strains still remained a function of knee joint kinematics but knee joint kinematics themselves were functions of the muscle optimization process.

The ACL material property for passive fiber strain at maximum isometric force⁸ was adjusted so that passive ACL strain versus flexion–extension and varus–valgus angles would be limited to 15% approximately (Fig. 4). The above passive fiber strain property was assigned as a very low percentage (<1%) for the passive ACL, a property that may need adjusting in individual models in order to keep finer control over strain limits. 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%.^{29,34} Similarly, maximum strain was limited to <2.5–3.0% in the osteoligamentous junction, contact region between ligament and bone, starting at a very low value of 0.2% or less at isometric force. The idealized strains established above are then used during CMC for checking ligament (muscle) and osteoligamentous (tendon) lengths as the whole musculotendon unit deforms. CMC optimization fails under extreme circumstances when the musculotendon length is too small.⁸ The ACL's linear elastic stiffness was equal to 240 N/mm, a stiffness assigned from cadaveric studies³⁹ of humans aged 22–35. It should be noted that the constant stiffness was characteristic only in the region of lower strains (8–10%) giving away to higher stiffness for higher strains according to Gaussian law,⁸ whose parameters can be assigned as material property in OpenSim.

Data Processing and Statistical Analysis

Ensemble averages (from 11 subjects) at percentage points in time (between TD and EP) were calculated to observe trends in ACL strain and internal force, knee variables and quadriceps and hamstring activation levels. Additionally, peak values for ACL and knee variables, averaged from individual peaks of 11 subjects, were computed for all three jump heights. Statistical one-way ANOVA tests were conducted to observe variations in several variables, like knee valgus and flexion angles and external moments, hamstring and quadriceps activation and ACL strain and internal force, between stop-jump heights. Pearson's correlation, r , coefficient of determination, r^2 , and associated

p -values were calculated to confirm that kinematic results from Forward Dynamics simulations strongly correlated with experimentally obtained kinematics computed with IK, thus providing validation for the Forward Dynamics simulations.

RESULTS

Results presented in this section focus on mechanical properties of ACL and knee and activation of hamstring and quadriceps (both knee joint) muscles from simulations. Ensemble averages of knee flexion and valgus angles, obtained versus percentage time, increased with increased stop-jump height as shown in Fig. 5. Ensemble averages of right and left flexion and valgus knee external moments (scaled to individual body mass) are shown in Fig. 6. Ensemble averages of ACL strain and internal force (scaled to individual body weight) with increasing stop-jump height are given in Fig. 7. It can be seen that peak forces and strains for 50 cm stop-jump occurred within 40% of landing phase. Results of hamstring and quadriceps activation at three stop-jump heights are given in Fig. 8. Quadriceps activations (maximum peak at 50 cm stop-jump) were found to be higher than that of hamstring (peaking at approximately 45% for 50 cm stop-jump) for all three stop-jump heights. Averages of reserve actuator contribution to balancing knee valgus and flexion external moments, calculated as a fraction of knee's valgus and flexion external moments, respectively, are given in Appendix B in Supplementary material 2. A sharp drop off in reserve actuators usage occurs after first 10% of total time of landing phase (Appendix B in Supplementary material 2).

Average peak knee flexion and valgus angles increased as stop-jump height increased. Knee flexion angle increased from $77.3 \pm 9.3^\circ$ (30 cm stop-jump) to $83.8 \pm 7.4^\circ$ (50 cm stop-jump) for right and from $76.6 \pm 8.9^\circ$ (30 cm) to $84.9 \pm 8.9^\circ$ (50 cm) for left. Similarly, average peak knee valgus angle increased from $8.8 \pm 1.6^\circ$ (30 cm) to $9.9 \pm 1.5^\circ$ (50 cm) for right and from $10.0 \pm 1.5^\circ$ (30 cm) to $10.4 \pm 1.0^\circ$ (50 cm) for left. Average peak knee flexion external moment increased from 76.9 ± 8.9 (30 cm) to 95.0 ± 10.7 Nm (50 cm) for right and from 70.7 ± 8.5 (30 cm) to 92.4 ± 11.1 Nm (50 cm) for left. And finally, average peak knee valgus external moment increased from 40.5 ± 3.2 (30 cm) to 43.2 ± 3.7 Nm (50 cm) for right and from 37.7 ± 3.0 (30 cm) to 42.6 ± 3.9 Nm (50 cm) for left. Hence increased peak variables were found for both ACL and knee. Average peak strains showed an increase from 9.3 ± 3.1 to $13.7 \pm 1.1\%$ for right and from 8.0 ± 2.8 to $13.4 \pm 2.0\%$ for left ACL, as stop-jump height increased from 30 to 50 cm.

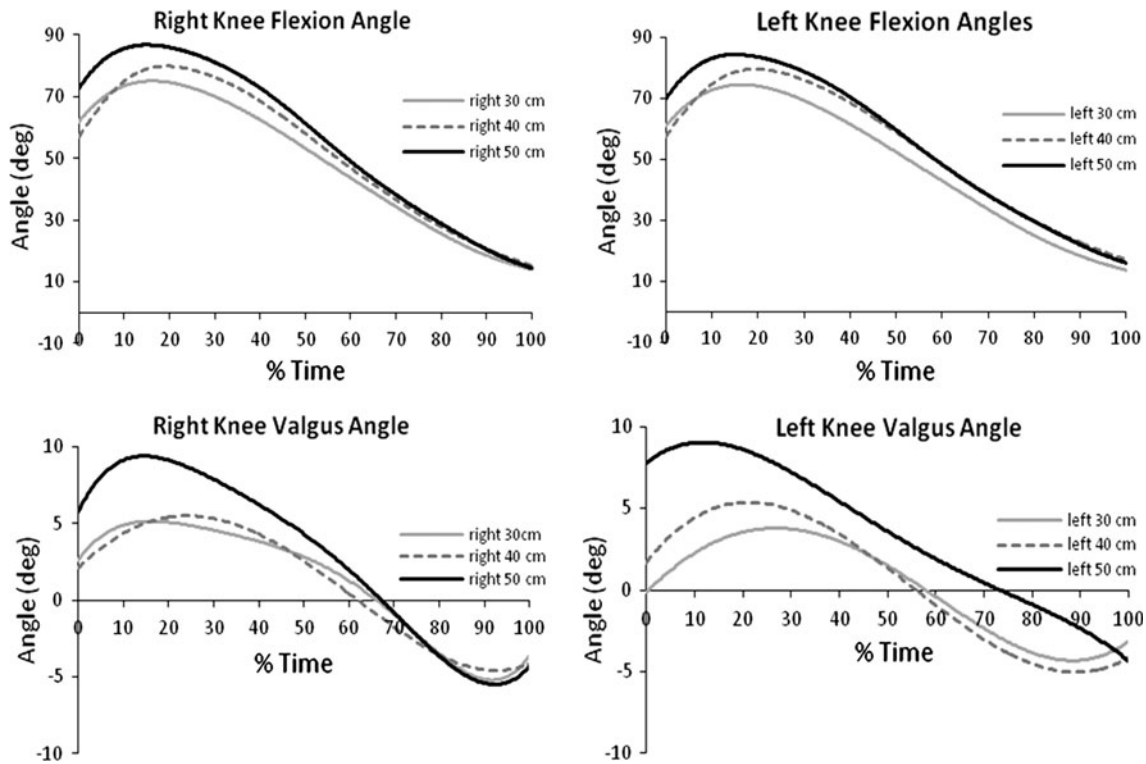


FIGURE 5. Ensemble average of right and left knee flexion and valgus angles against percentage time obtained from 11 subjects participating in stop-jumps from heights of 30, 40 and 50 cm.

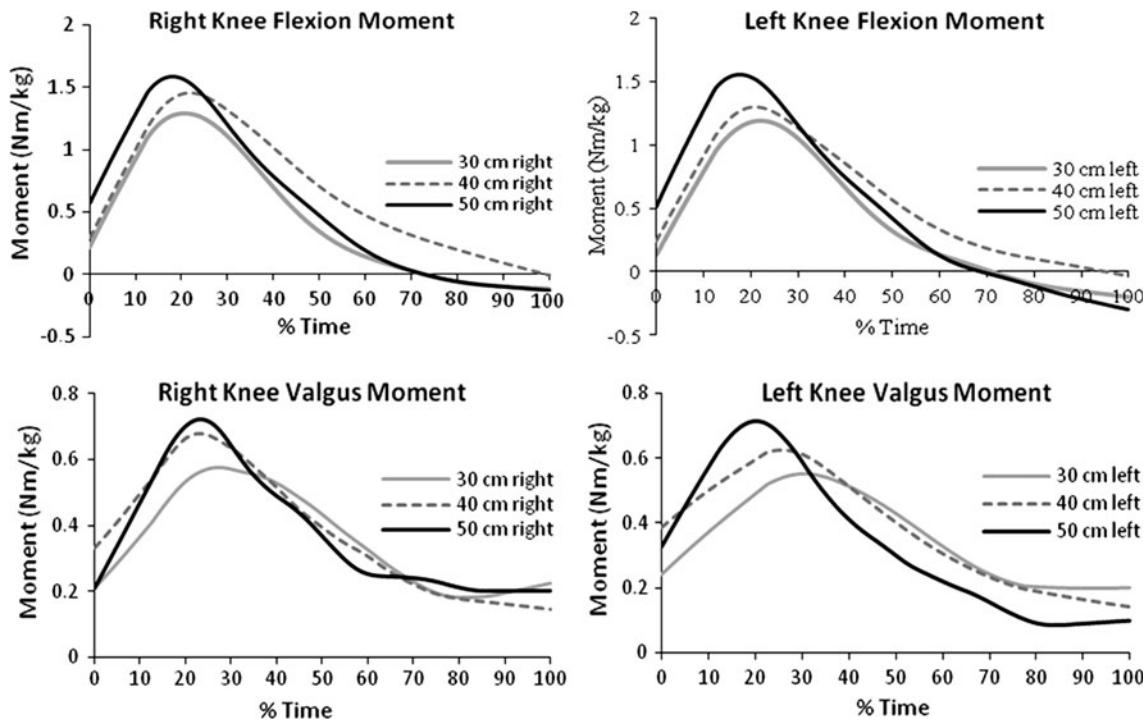


FIGURE 6. Ensemble average of right and left knee flexion and valgus external moments against percentage time obtained from 11 subjects participating in stop-jumps from heights of 30, 40 and 50 cm.

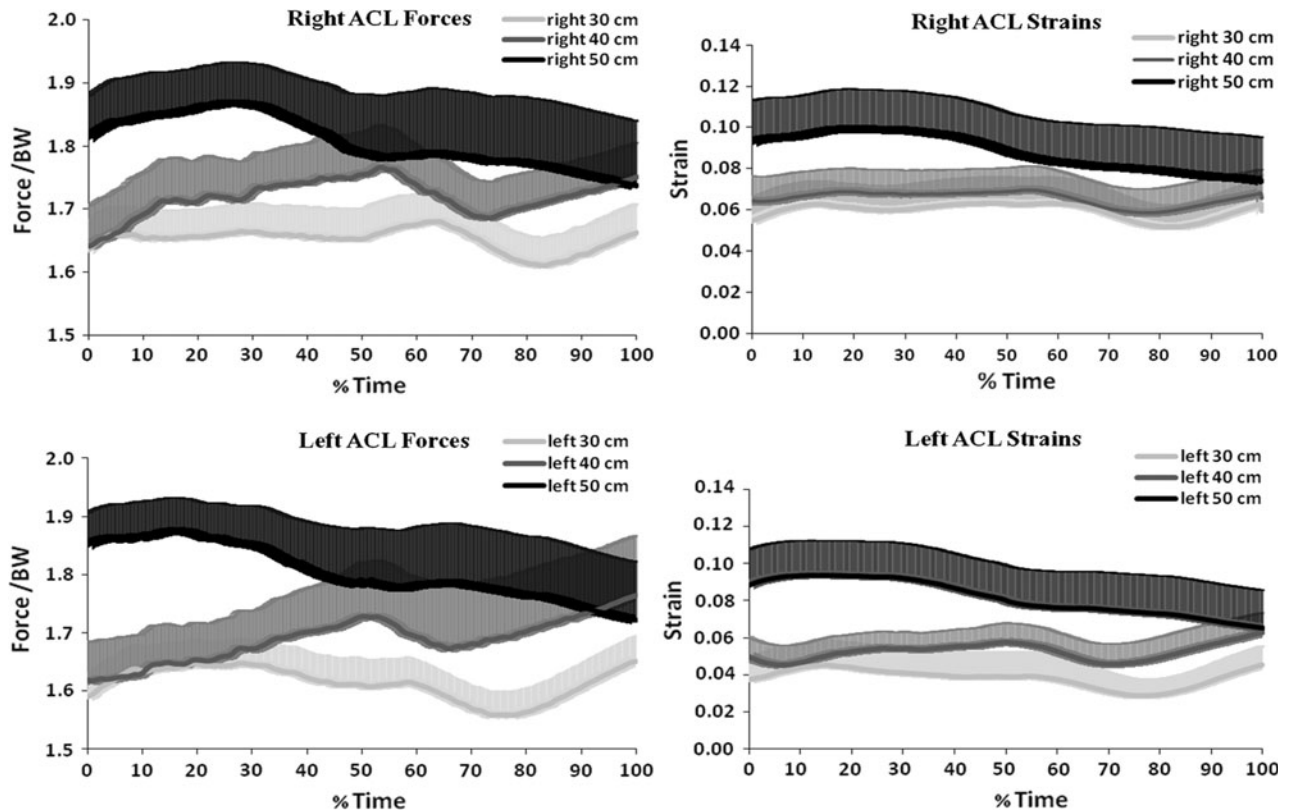


FIGURE 7. Ensemble average of right and left right and left ACL fiber (internal) forces and strains + 1 SD against percentage time obtained from 11 subjects participating in stop-jumps from heights of 30, 40 and 50 cm.

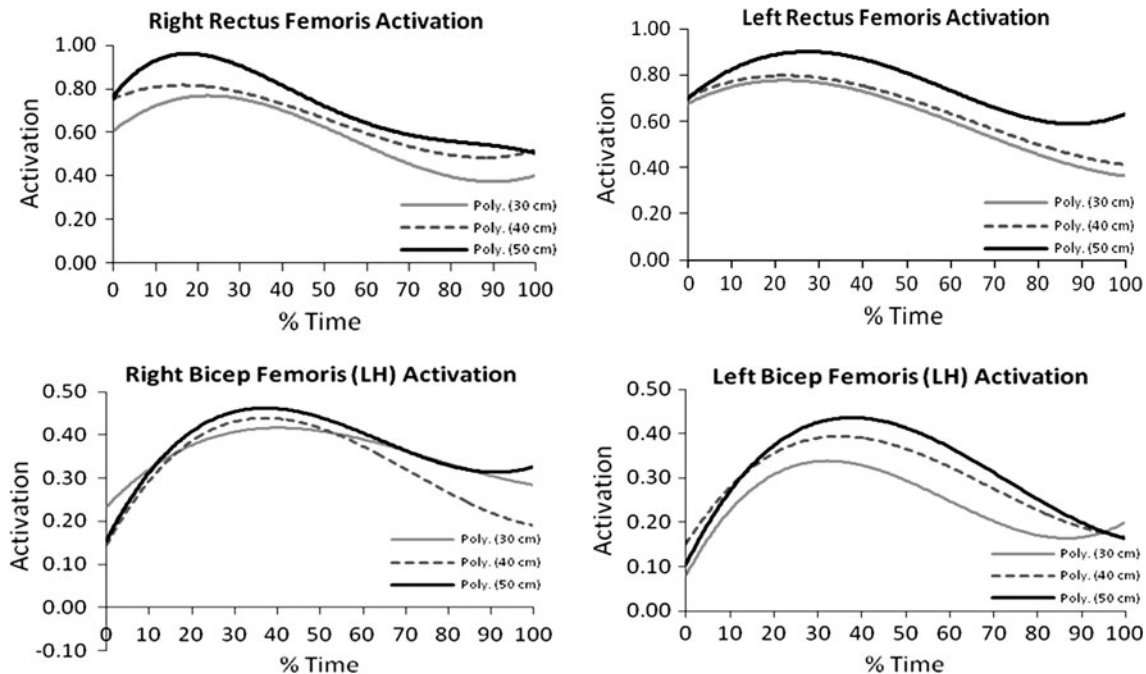


FIGURE 8. Ensemble average of right and left right hamstring, long head (LH) and quadriceps activations against percentage time obtained from 11 subjects participating in stop-jumps from heights of 30, 40 and 50 cm.

Average of 11 peak forces increased from 1056.0 ± 71.4 N to 1165.4 ± 123.8 for right and 1043.1 ± 61.3 N to 1160.6 ± 121.3 N for left, as stop-jump height increased from 30 to 50 cm.

The $p < 0.05$ results of one-way ANOVA tests in Table 1 indicate that most of the variables had significantly differing means with increasing stop-jump height. Table 2 shows that statistically significant values for r and r^2 were obtained, validating the results of Forward Dynamics simulation.

DISCUSSION

Initial portion of this study gave us experimental kinematic data of recreational female athletes

performing stop-jumps ranging from 30 to 50 cm in height. Higher stop-jump heights were not considered since ethical considerations necessitated that subjects' exposure to actual injury risks during trials remain minimal. However kinematic changes incurred with increase in stop-jump heights used in this study were sufficient to facilitate investigation of effects of stop-jump height on predicted ACL strain and internal force. In the following results of this study are discussed, where parts of it are compared to results from similar studies.

First, both knee flexion and valgus angles measured during this study are similar to findings in other studies.^{16,24} Second, substantially increased knee valgus external moments were found with increased stop-jump height and were accompanied by increased ACL

TABLE 1. One-way ANOVA results for knee and ACL variables computed at 30, 40 and 50 cm of stop-jump in this study.

Variable	Mean \pm 1 SD @ 30 cm	Mean \pm 1 SD @ 40 cm	Mean \pm 1 SD @ 50 cm	p -value
Strain (mm/mm)				
Right ACL	$0.06 \pm 3.62 \times 10^{-3}$	$0.07 \pm 3.7 \times 10^{-3}$	$0.09 \pm 8.70 \times 10^{-3}$	$<10^{-12}$
Left ACL	$0.04 \pm 4.71 \times 10^{-3}$	$0.05 \pm 4.30 \times 10^{-3}$	$0.08 \pm 9.44 \times 10^{-3}$	$<10^{-12}$
Force (N)				
Right ACL	967.51 ± 9.97	1007.28 ± 16.04	1058.11 ± 23.31	$<10^{-12}$
Left ACL	944.85 ± 15.92	988.42 ± 20.22	1056.71 ± 26.89	$<10^{-12}$
Angle ($^\circ$)				
Right knee abd.	1.22 ± 10.84	1.15 ± 10.73	2.96 ± 14.76	$<10^{-12}$
Left knee abd.	0.49 ± 10.73	0.15 ± 10.82	3.50 ± 13.27	$<10^{-12}$
Right knee flex.	49.53 ± 3.66	52.76 ± 3.72	56.87 ± 5.29	$<10^{-12}$
Left knee flex.	48.89 ± 3.90	53.65 ± 2.97	55.89 ± 4.25	$<10^{-12}$
Moment (Nm)				
Right knee abd.	26.66 ± 29.49	40.37 ± 28.32	33.80 ± 35.51	$<10^{-6}$
Left knee abd.	23.20 ± 28.09	33.36 ± 26.23	30.37 ± 36.73	$<10^{-3}$
Right knee flex.	22.78 ± 8.51	21.41 ± 10.67	22.70 ± 10.61	0.051*
Left knee flex.	22.59 ± 7.73	21.21 ± 9.89	20.09 ± 12.85	$<10^{-3}$
Activation				
Right rectus Fem.	0.58 ± 0.17	0.66 ± 0.15	0.73 ± 0.17	$<10^{-12}$
Left rectus Fem.	0.62 ± 0.16	0.65 ± 0.15	0.75 ± 0.13	$<10^{-12}$
Right bicep (LH) Fem.	0.35 ± 0.10	0.33 ± 0.13	0.37 ± 0.11	$<10^{-12}$
Left bicep (LH) Fem.	0.24 ± 0.10	0.30 ± 0.11	0.32 ± 0.13	$<10^{-12}$
Time (s)	0.35 ± 0.21	0.37 ± 0.17	0.42 ± 0.27	$<10^{-3}$

abd. = abduction, flex = flexion, Fem. = Femoris.

The p -value > 0.05 indicates lack of variation for that variable among the jump heights.

* $p > 0.05$ indicates insignificant changes in variable mean value with increasing stop-jump height.

TABLE 2. Correlation coefficients between kinematics from experimental (calculated with IK) and Forward Dynamics simulation data for one subject computed between TD and recovery to EP.

Jump height (cm)	Direction of rotation	Correlation coeff. (r)		Coeff. of determination (r^2)	
		Right	Left	Right	Left
30	Flexion	0.9986*	0.9973*	0.9984	0.9968
40	Flexion	0.9821*	0.9646*	0.9818	0.9639
50	Flexion	0.9499*	0.9022*	0.9544	0.9110
30	Valgus	0.9731*	0.9468*	0.9844	0.9690
40	Valgus	0.8902*	0.7925*	0.8684	0.7541
50	Valgus	0.9244*	0.8546*	0.8944	0.7999

Correlation r values with * indicates $p < 1E-6$.

strains. Additionally, peak valgus external moment and peak ACL strain happened about the same point in time and within the first 30% of total time of landing phase. These results are similar to findings in cadaveric studies^{12,21} which show a direct relationship between increased valgus external moment and increased ACL strain. Several non-cadaveric studies have also shown that increased valgus loading leads to high ACL strains.^{10,15,20} Furthermore, due to stress-strain behavior of passively elastic ligaments,⁴¹ deduction can be made that valgus external moment increased both ACL strain and internal force. The increase in ACL internal force with strain was an important finding of this study.

Some previous studies²⁷ have claimed that knee flexion or extension external moment does not play a key role in ACL injury. In contrast, increased flexion external moments were found with increasing stop-jump height (Fig. 6) implying that knee flexion external moment does have sensitivity to enhanced intensity of activity and therefore may play a key role in ACL injury. Other studies⁴⁰ have claimed that it is knee flexion angle (not external moment) that should be associated with ACL injury, particularly when such flexion angles are low. Assuming the above claim to be true, results of this study that showed increasing flexion angles with stop-jump height (Fig. 5) meant that participating subjects were not prone to ACL injury due to low flexion angles.

A large amount of research has been done with muscle co-activation in female athletes performing jump landing tasks (with or without rebound), and in particular on estimating hamstring:quadriceps (H:Q) ratios. Studies show that post contact H:Q ratios can be as low as 0.38 for drop-jumps from a height of 31 cm.⁹ Other studies conducted for a range of drop-jump heights reported H:Q ratios to be approximately between 0.35 and 0.52.²² Low H:Q ratios found in this study are not just comparable to two studies mentioned above but also compare to low H:Q ratios found by a host of past studies^{5,9,18,28,38}. The conclusion drawn from results of low H:Q ratio is that ACL plays a dominant role in compensating for reduced hamstring muscle activation leading to higher strains. An accompanying finding was that both hamstring and quadriceps muscles undergo peak contractions just after initial contact. Furthermore, it can be concluded that Forward Dynamics simulation can well predict low H:Q ratios found in young recreational female athletes when they perform stop-jump activities.

Constrained DOFs restrict knee and ACL motion and so do dependent translational DOFs despite these dependencies being created to represent more physiological tibiofemoral motion in the condyloidic knee joint.⁸ A brief discussion on how the knee's

constrained (and unconstrained) and dependent DOFs effect ACL strain and internal forces is given below. In the lower flexion range, 0°–30°, the anterior horns of the medial and lateral menisci prevent the femur from anterior translation, and it is at this stage that the ACL is fully extended and has maximum tension.^{2,11} The higher strains and internal forces at maximum extension can be seen for the 30 and 40 cm stop-jump heights in Fig. 7, at the end between TD and EP. It is noted that the 50 cm stop-jump height strains and internal forces are however not the highest at EP. With further flexion, tibiofemoral contact is concentrated on the tibial plateau when both femoral condyles move on tibial articulating surface.^{11,19} However, as the knee flexes, differing vertical levels of the tibial surface in contact with medial and lateral condyles can lead to valgus.¹¹ The valgus can worsen if there is lack of contact area or even lift off between the medial femoral condyle and tibial surface. Similar results were found in this study which caused expected higher strains and internal forces in the ACL during such valgus (Figs. 5, 7). This was also the reason why ACL strains and internal forces at higher angles of flexion surpassed those at maximum extension for the 50 cm stop-jump height (Figs. 5, 7).

However, there were several limitations associated with restricting condyloidic (knee) joint motion to two independent rotational DOFs and two dependent translational DOFs. One limitation was restricting knee sagittal plane translations as functions of flexion only. As mentioned above, tibiofemoral translation in the vertical direction is interrelated with valgus where valgus induced tibiofemoral separation would increase ACL strain and internal force. Hence, relative vertical tibiofemoral translation should more accurately be a function of both flexion and valgus. An additional limitation was not using internal-external (IE) rotational DOF and it possibly restricted knee transverse plane translation, particularly in the anterior-posterior direction perpendicular to the frontal plane. IE rotation, particularly occurring in the mid-flexion range, 30°–60°,³³ which tries to cause larger tibiofemoral separation in the anterior-posterior direction is primarily opposed by the ACL. A shorter anterior-posterior separation would result in less strains and internal forces at the ACL. Hence, for better accuracy the IE rotational DOF needs to be unconstrained and the knee anterior-posterior translational DOF made a function of both flexion and IE rotational DOF.

Finally, arguments in support of restricting sagittal plane translational DOFs as functions of flexion only and constraining translational and rotational DOFs are given below. As mentioned under Kinematic Analysis, most of the weight and braking forces were supported by the quadriceps, which made remaining

residual forces in the direction of the knee's three translational DOFs expectedly low. Given the above situation, any small changes to ACL strain and internal force from making each translational DOF a function of multiple rotational DOFs could be ignored. IE rotational DOF was not included to meet the study goal of estimating ACL strain and internal force solely due to valgus and flexion. Valgus external moment coupled with external tibial rotation^{2,35} can reduce ACL strain and *vice versa*. Hence, including knee IE rotational DOF would have been at cross purposes with goals of this investigation. However, a follow-up investigation with added IE rotational DOF would be beneficial for understanding mechanisms of ACL strain with or without it. The reason for not including the mediolateral DOF was that shear in this direction has negligible effect on the ACL.³¹

There were several other limitations to this study related to both experiment and simulation other than those imposed by the DOFs. First, there was lack in modeling important biomechanical elements that contribute to knee joint motion and orientation, mainly three other knee ligaments: Posterior Cruciate Ligament (PCL), Medial Cruciate Ligament (MCL) and Lateral Cruciate Ligament (LCL), which when integrated into knee joint will improve accuracy of musculoskeletal simulation. The PCL in particular needs to be included since it shares anterior-posterior load bearing function with ACL.²⁶ Secondly, representing ACL as a single bundle of fibers (not with separate AM and PL bundles) can be considered a limitation, given current trends in practicing double bundle ACL reconstruction surgeries.⁴³ A third limitation was limiting study population to female only subjects. Studies on gender based outcome of stop-jump activities have not been conducted, although similar studies have been done on gender differences in muscle activation²⁸ and neuromuscular properties³⁰ with drop-jump activities. A comparison of knee and ACL variables between female recreational athletes with their male counterparts, who are expected to have more "neutral"³⁰ alignments in their lower extremities would further expose any abnormalities in these variables and should be considered for future studies.

The specific conclusion here is that increased stop-jump height lead to increased valgus external moment which in turn is a major contributor to increasing ACL strain and internal force. Estimating ACL strain and internal force answered our primary research objective, a deduction validated by correlating experimental knee angles (calculated with IK) with knee angles from Forward Dynamics simulation. Similar simulation studies on other sports such as hockey where maneuvers such as side cutting can cause ACL injury need to be conducted, with ACL insertion in musculoskeletal

model as was done for this study. It will be highly advantageous to establish some form of musculoskeletal based screening program that help to precondition athletes against ACL injury, especially if such screening programs included ACL and other ligaments that are prone to injury during such activities. Special attention should be paid to perturbation of knee joint and its surrounding structure that cause ACL strain and internal force to increase. The mechanisms of perturbation that cause ACL rupture need to be investigated in details without enhancing risk of injury for trial participants. For example, laboratory trials for this study were conducted under strict restrictions of human subject safety protocol and risk of ACL injury was minimal; in all practicality, injury inducing activities cannot be conducted during experimental trials. In contrast, real life sporting events involve much higher risk of ACL injury. However, effects of perturbation, for example effect of increasing valgus external moment, can be investigated within safety of a simulation environment. The strain obtained from such perturbed simulation could then be compared to ACL rupture strain found in scientific (cadaveric) literature as done during this study, all without risk of injury to participants.

ELECTRONIC SUPPLEMENTARY MATERIAL

The online version of this article (doi:[10.1007/s10439-012-0572-x](https://doi.org/10.1007/s10439-012-0572-x)) contains supplementary material, which is available to authorized users.

ACKNOWLEDGMENTS

We are thankful to the Department of Mechanical Engineering, University of Louisville, Louisville, KY for funding the stop-jump laboratory trials. We express our special thanks to Dr. A. Swank, Department of Sports Physiology, University of Louisville, Louisville, KY for helping recruit young female participants for the stop-jump trials.

REFERENCES

- ¹Agel, J., E. A. Arendt, and B. Bershadsky. Anterior cruciate ligament injury in national collegiate athletic association basketball and soccer: a 13-year review. *Am. J. Sports Med.* 33:524–530, 2005.
- ²Amiri, S., D. Cooke, I. Y. Kim, and U. Wyss. Mechanics of the passive knee joint. Part 2: interaction between the ligaments and the articular surfaces in guiding the joint motion. *Proc. Inst. Mech. Eng. H J. Eng. Med.* 221:821–832, 2007.

- ³Anderson, F. C., and M. G. Pandy. A dynamic optimization solution for vertical jumping in three dimensions. *Comput. Methods Biomech. Biomed. Eng.* 2:201–231, 1999.
- ⁴Anderson, F. C., M. G. Pandy, and D. G. Hull. A parameter optimization approach for the optimal control of large-scale musculoskeletal systems. *J. Biomech. Eng.* 114:450–460, 1992.
- ⁵Besier, T. F., D. G. Lloyd, J. L. Cochrane, and T. R. Ackland. External loading of the knee joint during running and cutting maneuvers. *Med. Sci. Sports Exerc.* 33:1168–1175, 2001.
- ⁶Caraffa, A., G. Cerulli, M. Proietti, G. Aisa, and A. Rizzo. Prevention of anterior cruciate ligament injuries in soccer: a prospective controlled study of proprioceptive training. *Knee Surg. Sports Trauma Arthrosc.* 4:19–21, 1996.
- ⁷Cohen, S. B., C. VanBeek, J. S. Starman, D. Armfield, J. J. Irrgang, and F. H. Fu. MRI measurement of the two bundles of the normal anterior cruciate ligament. *Orthopedics* 32(9), 2009. doi:10.3928/01477447-20090728-35.
- ⁸Delp, S. L., F. C. Anderson, A. S. Arnold, P. Loan, A. Habib, and C. T. John. OpenSim: open-source software to create and analyze dynamic simulations of movement. *IEEE Trans. Biomed. Eng.* 54:1940–1950, 2007.
- ⁹Ebben, W. P., M. L. Fauth, E. J. Petushek, L. R. Garceau, E. H. Brittni, B. N. Lutsch, and C. R. Feldman. Gender-based analysis of hamstring and quadriceps muscle activation during jumping and cutting. *J. Strength Cond. Res.* 24(2):408–415, 2010.
- ¹⁰Ford, K. R., G. D. Myer, and T. E. Hewett. Valgus knee motion during landing in high school female and male basketball players. *Med. Sci. Sports Exerc.* 35:1745–1750, 2003.
- ¹¹Freeman, M. A., and V. Pinskerova. The movement of the normal tibiofemoral joint. *J. Biomech.* 38(2):197–208, 2005.
- ¹²Fukuda, Y., S. L. Woo, and J. C. Loh. A quantitative analysis of valgus torque on the ACL: a human cadaveric study. *J. Orthop. Res.* 21:1107–1112, 2003.
- ¹³Herrington, L., and A. Munro. Drop jump landing knee valgus angle; normative data in a physically active population. *Phys. Ther.* 11:56–59, 2010.
- ¹⁴Hewett, T. E., G. D. Myer, and K. R. Ford. Puberty decreases dynamic knee stability in female athletes: a potential mechanism for increased ACL injury risk. *J. Bone Joint Surg. Am.* 86:1601–1608, 2004.
- ¹⁵Hewett, T. E., G. D. Myer, K. R. Ford, S. Robert, R. S. Heidt, Jr., A. J. Colosimo, and S. G. McLean. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. *Am. J. Sports Med.* 34(3):445–455, 2006.
- ¹⁶Hewett, T. E., M. V. Paterno, and G. D. Myer. Strategies for enhancing proprioception and neuromuscular control of the knee. *Clin. Orthop. Relat. Res.* 402:76–94, 2002.
- ¹⁷Hill, A. V. The heat of shortening and the dynamic constants of muscle. *Proc. R. Soc. Lond. Ser. B Biol. Sci.* 126(843):136–195, 1938.
- ¹⁸Hosokawa, T., K. Sato, S. Mitsueda, H. Umehara, K. Hidume, T. Okada, I. Kanisawa, A. Tsuchiya, K. Takahashi, and H. Sakai. Effects of anterior cruciate ligament injury prevention program on lower extremity alignment, isokinetic muscle strength and electromyographic activity. *Br. J. Sports Med.* 45(4):353–360, 2011.
- ¹⁹Iwaki, H., V. Pinskerova, and M. A. Freeman. Tibiofemoral movement 1: the shapes and relative movements of the femur and tibia in the unloaded cadaver knee. *J. Bone Jt. Surg. Br.* 82(8):1189–1195, 2000.
- ²⁰Joseph, M., D. Tiberio, J. L. Baird, T. H. Trojian, J. M. Anderson, W. J. Kraemer, and C. Marsh. Knee valgus during drop jumps in National Collegiate Athletic Association Division I female athletes: the effects of a medial post. *Am. J. Sports Med.* 36:285–289, 2008.
- ²¹Kanamori, A., S. L. Woo, and C. B. Ma. The forces in the anterior cruciate ligament and knee kinematics during a simulated pivot shift test: a human cadaveric study using robotic technology. *Arthroscopy* 16:633–639, 2000.
- ²²Kellis, E., F. Arabatzis, and C. Papadopoulos. Muscle co-activation around the knee in drop jumping using the co-contraction index. *J. Electromyogr. Kinesiol.* 13:229–238, 2003.
- ²³Kernozek, T. W., and R. J. Ragan. Estimation of anterior cruciate ligament tension from inverse dynamics data and electromyography in females during drop landing. *Clin. Biomech.* 23:279–286, 2008.
- ²⁴Krosshaug, T., J. R. Slaughterbeck, L. Engebretsen, and R. Bahr. Biomechanical analysis of anterior cruciate ligament injury mechanisms: three-dimensional motion reconstruction from video sequences. *Scand. J. Med. Sci. Sports* 17(5):508–519, 2007.
- ²⁵Landry, S. C., K. A. McKean, C. A. Hubley-Kozey, W. D. Stanish, and K. J. Deluzio. Neuromuscular and lower limb biomechanical differences exist between male and female elite adolescent soccer players during an unanticipated run and crosscut maneuver. *Am. J. Sports Med.* 35(11):1901–1911, 2007.
- ²⁶Li, G., T. J. Gill, L. E. DeFrate, S. Zayontz, V. Glatt, and B. Zarins. Biomechanical consequences of PCL deficiency in the knee under simulated muscle loads—an in vitro experimental study. *J. Orthop. Res.* 20(4):887–892, 2002.
- ²⁷McLean, S. G., X. Huang, A. Su, and A. J. van den Bogert. Sagittal plane biomechanics cannot injure the ACL during sidestep cutting. *Clin. Biomech.* 19:828–838, 2004.
- ²⁸Nagano, Y., H. Ida, M. Akai, and T. Fukuyabashi. Gender differences in knee kinematics and muscle activity during single knee drop landing. *Knee* 14:218–223, 2007.
- ²⁹Noyes, F. R. Functional properties of knee ligaments and alterations induced by knee immobilization. *Clin. Orthop. Rel. Res.* 123:210–242, 1977.
- ³⁰Noyes, F. R., S. D. Barber-Westin, C. Fleckenstein, C. Walsh, and J. West. The drop jump screening test: difference in lower limb control by gender and effect of neuromuscular training in female athletes. *Am. J. Sports Med.* 33(2):197–207, 2005.
- ³¹Pflum, M. A., K. B. Shelburne, M. R. Torry, M. J. Decker, and M. G. Pandy. Model prediction of anterior cruciate ligament force during drop-landing. *Med. Sci. Sport Exerc.* 36(11):1948–1949, 2004.
- ³²Seth, A., M. A. Sherman, J. A. Reinbolt, and S. L. Delp. OpenSim: a musculoskeletal modeling and simulation for in silico investigation and exchange. *Procedia IUTAM* 2:212–232, 2011.
- ³³Shimokochi, Y., and S. J. Shultz. Mechanisms of noncontact anterior cruciate ligament injury. *J. Athl. Train.* 43(4):396–408, 2008.
- ³⁴Shin, C. S., A. M. Chaudhari, and T. P. Andriacchi. The effect of isolated valgus moments on ACL strain during single-leg landing: a simulation study. *J. Biomech.* 42(3):280–285, 2009.
- ³⁵Shin, C. S., A. M. Chaudhari, and T. P. Andriacchi. Valgus plus internal rotation moments increase anterior cruciate

- ligament strain more than either alone. *Med. Sci. Sport Exerc.* 43(8):1484–1491, 2011.
- ³⁶Thelen, D. G., and F. C. Anderson. Using computed muscle control to generate forward dynamic simulations of human walking from experimental data. *J. Biomech.* 39:1107–1115, 2006.
- ³⁷Thelen, D. G., S. L. Delp, and F. C. Anderson. Generating dynamic simulations of movement using computed muscle control. *J. Biomech.* 36:321–328, 2003.
- ³⁸Withrow, T. J., L. J. Huston, E. M. Wojtys, and J. A. Ashton-Miller. Effect of varying hamstring tension on anterior cruciate ligament strain during in vitro impulsive knee flexion and compression loading. *J. Bone Jt. Surg. Am.* 90:815–823, 2008.
- ³⁹Woo, S. L., R. E. Debski, J. D. Withrow, and M. A. Janaushek. Tensile properties of the human femur-anterior cruciate ligament–tibia complex. *Am. J. Sports Med.* 27(4):533–543, 1999.
- ⁴⁰Yu, B., and W. E. Garrett. Mechanisms of non-contact ACL injuries. *Br. J. Sports Med.* 41:47–51, 2007.
- ⁴¹Zajac, F. E. Muscle and tendon: properties, models, scaling, and application to biomechanics and motor control. *Crit. Rev. Biomed. Eng.* 17(4):359–411, 1989.
- ⁴²Zajac, F. E. Muscle coordination of movement: a perspective. *J. Biomech.* 26:109–124, 1993.
- ⁴³Zelle, B. A., A. F. Vidal, P. U. Brucker, and F. H. Fu. Double-bundle reconstruction of the anterior cruciate ligament: anatomic and biomechanical rationale. *J. Am. Acad. Orthop. Surg.* 15(2):87–96, 2007.