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What is This?
Strain Response of the Anterior Cruciate Ligament to Uniplanar and Multiplanar Loads During Simulated Landings

Implications for Injury Mechanism

Ata M. Kiapour,†‡ PhD, Constantine K. Demetropoulos,§ PhD, Ali Kiapour,‡ PhD, Carmen E. Quatman,¶¶ MD, PhD, Samuel C. Wordeman,¶¶# PhD, Vijay K. Goel,‡ PhD, and Timothy E. Hewett,**††† PhD

Investigation performed at the Engineering Center for Orthopaedic Research Excellence (ECORE), The University of Toledo, Toledo, Ohio, USA

Background: Despite basic characterization of the loading factors that strain the anterior cruciate ligament (ACL), the interrelationship(s) and additive nature of these loads that occur during noncontact ACL injuries remain incompletely characterized.

Hypothesis: In the presence of an impulsive axial compression, simulating vertical ground-reaction force during landing (1) both knee abduction and internal tibial rotation moments would result in increased peak ACL strain, and (2) a combined multiplanar loading condition, including both knee abduction and internal tibial rotation moments, would increase the peak ACL strain to levels greater than those under uniplanar loading modes alone.

Study Design: Controlled laboratory study.

Methods: A cadaveric model of landing was used to simulate dynamic landings during a jump in 17 cadaveric lower extremities (age, 45 ± 7 years; 9 female and 8 male). Peak ACL strain was measured in situ and characterized under impulsive axial compression and simulated muscle forces (baseline) followed by addition of anterior tibial shear, knee abduction, and internal tibial rotation loads in both uni- and multiplanar modes, simulating a broad range of landing conditions. The associations between knee rotational kinematics and peak ACL strain levels were further investigated to determine the potential noncontact injury mechanism.

Results: Externally applied loads, under both uni- and multiplanar conditions, resulted in consistent increases in peak ACL strain compared with the baseline during simulated landings (by up to 3.5-fold; P < .032). Combined multiplanar loading resulted in the greatest increases in peak ACL strain (P < .001). Degrees of knee abduction rotation (R² = 0.45; β = 0.42) and internal tibial rotation (R² = 0.32; β = 0.23) were both significantly correlated with peak ACL strain (P < .001). However, changes in knee abduction rotation had a significantly greater effect size on peak ACL strain levels than did internal tibial rotation (by ~2-fold; P < .001).

Conclusion: In the presence of impulsive axial compression, the combination of anterior tibial shear force, knee abduction, and internal tibial rotation moments significantly increases ACL strain, which could result in ACL failure. These findings support multiplanar knee valgus collapse as one of the primary mechanisms of noncontact ACL injuries during landing.

Clinical Relevance: Intervention programs that address multiple planes of loading may decrease the risk of ACL injury and the devastating consequences of posttraumatic knee osteoarthritis.

Keywords: anterior cruciate ligament; ACL; knee; injury mechanism; landing; multiplanar valgus collapse

Noncontact injuries (those without a direct blow to the knee joint) are reported to be the predominant mechanism of anterior cruciate ligament (ACL) injury (>70% of all ACL injuries). They occur during landing from a jump and lateral cutting maneuvers that may occur in athletic activities such as basketball and soccer. Injury prevention strategies such as neuromuscular training are an appealing option to avoid short- and long-term complications associated with these injuries. Identification of high-risk maneuvers that result in increased risk of ACL injury is a major step in the development of new, as well as optimization of existing, neuromuscular training programs in an effort to prevent these devastating injuries from occurring.

Neuromuscular control deficits during dynamic movements are postulated to be the primary causal factors for...
Both primary and secondary ACL injuries (reinjury after ACL reconstruction).\textsuperscript{18} Deficits in dynamic active neuromuscular control manifest as excessive joint loads and ultimately lead to detrimental ACL stress/strains and ultimate failure. Over the past 2 decades, considerable efforts to characterize ACL injury mechanisms utilizing in vivo, ex vivo, and in silico techniques have been made.\textsuperscript{22} Among all studied loading scenarios, anterior tibial shear force,\textsuperscript{42,46,11,25,29,55} aggressive quadriceps contraction,\textsuperscript{8,17,50,52} axial compression,\textsuperscript{24,30,34,50,54} knee abduction,\textsuperscript{§§} and internal tibial rotation moments\textsuperscript{7,11,21,29,34,37,45} have been reported to be associated with increased ACL loading and potential risk of injury at shallow knee flexion angles, using ex vivo and in silico approaches. These findings were further supported by in vivo biomechanical and video analysis studies that showed increased risk and/or frequency of noncontact ACL injuries in the presence of high ground-reaction forces, quadriceps contractions, knee abduction, and internal tibial rotations.\textsuperscript{52,19,26,27,39}

Recent systematic reviews of the literature that focused on determination of probable noncontact ACL injury mechanisms have concluded that ACL injuries are likely multiplanar in nature.\textsuperscript{42,44} Despite general agreement on the effect of aforementioned uniplanar loading factors on ACL loading and risk of injury, the interrelationship and additive nature of these loading conditions that result in high strains in the ACL and lead to noncontact ACL injuries have been a subject of controversy and considerable debate. The potential contribution of knee frontal plane loading in multiplanar mechanism of noncontact ACL injuries has become one of the main topics of this debate, with some studies showing increased ACL strain in the presence of knee abduction moment,\textsuperscript{3} while others report no difference.\textsuperscript{3,11,31,38} This discrepancy is mainly associated with lack of a comprehensive, detailed parametric analysis of all loading conditions shown to increase ACL loading and risk of injury under injurious conditions in a uni- and multiplanar fashion using valid experimental approaches. Therefore, the current study was designed to investigate the changes in peak ACL strain, as a surrogate for injury risk, during physiologic simulation of high-risk dynamic landings under a wide range of uni- and multiplanar loading conditions. We hypothesized that in the presence of an impulsive axial compression, simulating vertical ground-reaction force during landing (1) both knee abduction and internal tibial rotation moments would result in increased peak ACL strain, and (2) combined multiplanar loading condition, including both knee abduction and internal tibial rotation moments, would increase the peak ACL strain to levels greater than under each of these separate loading modes alone.

**METHODS**

Multiple dynamic landing conditions after a jump were simulated in an established cadaveric model.\textsuperscript{24} This model has been previously validated against in vivo data with regard to various landing kinetics and kinematics factors\textsuperscript{24} and has been shown to be consistent in generating ACL injuries with similar patterns as observed in real cases of ACL injuries.\textsuperscript{20} The ACL strain and tibiofemoral rotations were measured and compared between the loading groups.

**Specimen Preparation**

Seventeen normal unembalmed fresh-frozen cadaveric lower limbs (age, 45 \( \pm \) 7 years; 9 female and 8 male), free from any signs of soft or hard tissue were acquired. Specimens were sectioned at the mid-femoral diaphysis (30 cm above the joint line) and potted in polyester resin after removal of all soft tissues up to 15 cm proximal to the joint line. The quadriceps and hamstrings (medial and lateral) tendons were then isolated and clamped to allow for simulation of muscle loads across the knee. The remaining musculature along with the skin was maintained intact. The foot and ankle were also maintained intact to provide a realistic load transfer interface. Specimens were stored at \(-20^\circ\text{C}\) until 24 hours before testing, when they were slowly thawed to room temperature. Exposed tissue about the knee joint was kept moist with 0.9% buffered saline solution at all times during testing. Data for the distribution of injury patterns to the ACL and surrounding structures, in addition to the ACL to medial collateral ligament (MCL) strain ratios, during simulated landings on these specimens have been previously published by our group.\textsuperscript{28,40}

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\textsuperscript{4}Kiapour et al The American Journal of Sports Medicine

\textsuperscript{§§}References 4, 9, 13, 14, 21, 28, 32, 40, 46, 51.

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Cadaveric Simulation of Dynamic Landing

A physiologic, custom-designed drop-stand was used to simulate dynamic landings (Figure 1A). The unconstrained nature of this experimental setup allows for a broad range of externally applied loading conditions (ie, anterior shear force, knee abduction, and tibial axial rotation) to be applied during simulated landing. Specimens were positioned inverted with the tibia orientated vertically and the foot positioned above the tibia while rigidly fixed at the proximal femur to a fixture. All specimens were tested with the knee at 25° of flexion, as this flexion angle has been reported during video analysis of ACL injury events. Muscle forces across the knee were applied with multiple cable-pulley systems along with static weights to apply constant forces to the quadriceps (1200 N) and hamstrings (800 N; 400 each to medial and lateral group). The simulated muscle forces have been derived from our previous unpublished work to provide a baseline joint stability under applied external loads, with magnitudes used in this study, without overprotecting the knee joint or causing any soft tissue injuries.

Adjustable pulley systems were used to maintain the physiologic line of action of each muscle (Figure 1B). An external fixation frame with an integrated pulley system was rigidly attached to the tibia to apply external loads, such that different landing postures could be simulated (Figure 1C). A combination of static weights and cable-pulley system produced forces to generate an anterior tibial shear force and force couples to generate pure knee abduction and internal tibial rotation moments (Figure 1C). As unconstrained force couples (free to move with the specimens) were used to generate pure moments in this study, their action was independent of the point of application. The distal extremity (lower leg and foot) was not constrained until after loading and repositioning of the lower limb. After the application of all external loads, the tibia was vertically realigned by rotating and translating the femoral fixture to be in line with the axis of the impactor, while maintaining 25° of knee flexion (Figure 1A). An athletic shoe was placed on the foot to simulate realistic load transfer during initial contact. Floor contact was simulated by an inverted floor plate set upon the shoe that was...
maintained in a horizontal position by a series of linear bearings. Landing was simulated by release of a specific weight from a specific height utilizing a hemispherical impactor with an integrated weight stack. The drop weight exerted an impulsive axial compression to simulate vertical ground-reaction force during landing from a jump.24,28

Instrumentation

ACL strain was calculated based on the measurements of a differential variable reluctance transducer (DVRT; MicroStrain Inc) that was arthroscopically placed on the distal third of the anteromedial (AM) bundle through 2 parapatellar incisions. To calculate absolute strain values, ACL reference length was calculated based on established methods as the distinct inflection point in force versus DVRT displacement data collected before the testing.10 It was assumed that the average strain across the ACL AM bundle was equal to the change in length of the measured segment divided by the reference length obtained from DVRT measurements using the following equation:

\[
\text{Strain} \% = \frac{L - L_0}{L_0} \times 100
\]

where \( L \) is the instantaneous length measured across the DVRT, and \( L_0 \) is the length measured across the DVRT at the reference length of the ligament. The rigid body motion of the femur and tibia were tracked using an Optotrak 3020 3D motion capture system (Northern Digital) with arrays of 3 infrared-LED markers rigidly attached to each bone. Generated impulsive axial impact loads were captured using a 6-axis load cell (R.A. Denton Corp) embedded within the floor pad (Figure 1A). Data collection from all data acquisition units were synchronized utilizing a simultaneous trigger. Analog data (load cell and the DVRT) were collected at 4 kHz, while motion data were collected at 400 Hz.

Loading Protocol

Bipedal landings after a jump were simulated by the release of one-half (350 N) body weight from a height of 30 cm.12 To simulate high-risk injurious conditions, multiple combinations of knee abduction and internal tibial rotation moments with or without anterior tibial shear force were used to establish 2 loading groups (Table 1). Both loading protocols begin and end with similar loading conditions to facilitate comparisons between specimens while evaluating a broad range of injurious conditions. Shin et al45 have reported knee abduction moments up to 51 N\( \cdot \)m and internal tibial rotation moments up to 30 N\( \cdot \)m during noninjurious landings. However, higher magnitudes (up to 3 times) of the aforementioned loads were applied to simulate high-risk injurious conditions as reported in prior studies.5,32 Specimens were randomly assigned to each loading group and sequentially tested from loading 1 through 16, or until failure was observed. Observations of the mechanical response during testing were monitored as indicators of ACL failure. Ultimately, manual evaluation of joint laxity and arthroscopic inspection were employed to confirm catastrophic tissue failure.28

Statistical Analysis

Analysis of variance (ANOVA) was used to investigate the overall effect of tested loadings (type and magnitude) on generated peak ACL strain. Measurements obtained under loading conditions that led to an ACL failure were excluded from the analysis to avoid measurement errors associated

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<th>Knee Abduction Group</th>
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with joint injury. Dunnett post hoc analysis was used to conduct pair-wise comparisons, adjusted for multiple testing, between each loading group and baseline. Linear regression analysis was used to investigate the relationship between knee abduction and internal tibial rotation degrees on peak ACL strain. The effect of knee abduction and internal tibial rotations on peak ACL strain levels were measured based on slope of the fitted linear model ($\beta$). Results are reported as mean $\pm$ standard error of the mean (SEM); $P < .05$ was considered statistically significant. Analyses were conducted using SPSS statistical package (version 22.0; IBM Corp).

RESULTS

ACL failure was generated in 15 of 17 specimens (88%), with a clinically relevant distribution of soft tissue (ie, meniscus) and bony (ie, bone bruises of femur and tibia) injuries commonly reported to occur in patients who have sustained ACL injuries. Details of the peak ACL strain at failure, associated loading mechanisms, and observed injury patterns for these specimens have been previously presented by Levine et al. $^{28}$ Average values for all quantified outcomes, including peak axial impact load, peak ACL strain, and peak knee rotation in frontal and axial planes under each tested modes of loading, are presented in Table 2. Simulated landing under baseline, with greater strain levels observed under higher magnitudes of applied moments ($P = .011$ for overall effect)

The externally applied internal tibial rotation moments resulted in increased peak ACL strain compared with the baseline, with greater strain levels observed under higher magnitudes of applied moments ($P = .011$ for overall effect)

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The externally applied internal tibial rotation moments resulted in increased peak ACL strain compared with the baseline, with greater strain levels observed under higher magnitudes of applied moments ($P = .011$ for overall effect)
of internal tibial rotation moment on peak ACL strain) (Figure 2C). The post hoc analysis showed significant increases in peak ACL strain by 4.1% ± 1.7% (P = .049) and 5.8% ± 1.7% (P = .005) under applied 60- and 80-N·m internal rotation moments, respectively. However, no significant differences were observed between peak ACL strain under baseline (6.8% ± 1.0%) and those under 10 N·m (8.2% ± 1.9%; P = .388), 20 N·m (8.8% ± 2.1%; P = .231), or 40 N·m (9.7% ± 2.4%; P = .145) of internal rotation moments.

Simulated landings under combined multiplanar loading (anterior tibial shear load, knee abduction, and internal tibial rotation moments) resulted in highest increases in peak ACL strain compared with the baseline, with greater strains levels observed under higher magnitudes of applied loads (P < .001) (Figure 2D). The post hoc analysis showed significant increases in peak ACL strain compared with baseline, by 10.2% ± 1.2%, 10.3% ± 1.3%, 13.9% ± 1.3%, and 16.8% ± 1.3% under combined loading with 75, 100, 125, and 150 N·m of knee abduction moment, respectively (P < .001 for all comparisons).

Figure 3 shows the distribution of peak ACL strain during simulated landings under all 5 modes of loading conditions tested in this study. The group-wise comparisons showed significantly higher peak ACL strain during landings under applied knee abduction moment (P = .001), internal tibial rotation moment (P = .040), anterior tibial shear load (P = .046), and combined multiplanar loading (P < .001) compared with the baseline (Figure 3). The analysis further showed that simulated landings in the presence of combined multiplanar loading resulted in significantly greater peak ACL strain than observed under applied anterior tibial shear loads, knee abduction, and internal rotation moments alone (P < .001 for all comparisons).

As shown in Figure 4, knee abduction and internal tibial rotations were both significantly correlated (P < .001) with quantified peak ACL strain levels. Knee abduction rotation showed higher correlation with peak ACL strain (r = 0.67, explaining 45% of variation in peak ACL strain) than internal tibial rotation (r = 0.57, explaining 32% of variation in peak ACL strain). Greater peak ACL strain levels were observed under higher degrees of knee abduction and internal tibial rotations. However, the linear regression analysis showed that knee abduction rotation has a significantly greater (P < .001) effect on increased peak ACL strain (β = 0.42 ± 0.04) than does internal tibial rotation (β = 0.23 ± 0.03).

DISCUSSION

Characterization of the complex mechanisms associated with ACL injury can help prevent and develop a more efficient treatment for this injury. Although the loading factors that affect the ACL strain/force and potential risk of injury have been studied extensively, it remains unclear how these loads interact together in a multiplanar manner and lead to noncontact ACL injuries. A comprehensive parametric analysis using a validated cadaveric model of landing was used to fully characterize the ACL strain under a wide range of high-risk injurious landing scenarios. Results supported our hypotheses and showed that in the
presence of an impulsive axial compression, externally applied uni- and multiplanar loads resulted in consistent increases in peak ACL strain levels. It was further shown that combined multiplanar loading result in the greatest increases in peak ACL strain, to levels beyond those observed under uniplanar loading conditions tested in this study.

Effect of Uniplanar Loading on Peak ACL Strain During Landing

In this study, increases in peak ACL strains were observed under increased anterior tibial shear force (0-268 N), knee abduction (0-75 Nm), and internal tibial rotation (0-80 Nm) moments, in the presence of impulsive axial compression and simulated muscle forces (Figure 2, A-C). However, only the highest magnitude of each loading condition significantly increased the peak ACL strains compared with baseline. This could be due to the small sample size and high intrasubject variability in tissue properties between the tested knees.

Externally applied anterior tibial shear force resulted in elevated peak ACL strains by up to 1.5-fold (<268 N) compared with the baseline at 25° of knee flexion during simulated landings (Figure 2A). The applied anterior tibial shear force in the current study was used to simulate the effects of different factors such as excessive quadriceps contraction,5,53 deceleration,45 or other factors and should not be mistaken for the shear force that was generated across the knee as a result of the ground-reaction force.24,35 Similarly, simulated landings with externally applied knee abduction (Figure 2B) and internal tibial rotation (Figure 2C) moments resulted in significant increases in peak ACL strain levels by up to 1.8-fold (<75 Nm of knee abduction or 80 Nm of internal tibial rotation moment).

Previous ex vivo and in silico studies have reported similar findings under both static and dynamic loading conditions. However, the majority of these investigations studied these effects within a limited loading range, primarily under static loading conditions simulating the noninjurious events, and/or without consideration of role of muscle loads and impulsive impact. The current parametric analysis builds on those previous findings and provides improved insight into how anterior tibial shear force, knee abduction, and internal rotation moments regulate ACL strain and potential risk of injury under a wider loading range (up to 3 times greater loading magnitudes than those reported during noninjurious landings) during a physiologic simulation of high-risk

Figure 3. Peak anterior cruciate ligament (ACL) strain levels under all tested uni- and multiplanar loading conditions. Data include the strain values under all magnitudes of loading within each loading group. All the comparisons have been made after adjusting for the effect of loading magnitude. P values are for comparisons between all loading conditions and the baseline loading. The P value for comparisons between combined multiplanar loading and other 3 modes of loading (anterior shear, knee abduction, and internal tibial rotation) are indicated with asterisk (*). ABD, abduction moment; AS, anterior shear load; ITR, internal rotation moment.

Figure 4. Univariate regression analysis showing relationships between the peak anterior cruciate ligament (ACL) strain and (A) knee abduction rotation and (B) internal tibial rotation.
Effect of Combined Multiplanar Loading on Peak ACL Strain During Landing

Despite substantial effort in characterization of the effect of external loadings (plane and magnitude) on ACL strain/force, only a small portion of these studies have investigated the additive nature of these loadings and how they affect ACL strain/force under a multiplanar mechanism.4,7,28,29,36,38,40,45 There are even a more limited number of studies that have investigated the ACL strain/force under dynamic, high-risk multiplanar loading conditions, which better resemble the injurious events.28,38,40,45

Markolf and colleagues29 were among the first who investigated the effect of combined multiplanar loading on ACL force, measured in situ, during static cadaveric experiments. They showed significant increases in ACL force under combined loadings (anterior shear + abduction, anterior shear + internal rotation, and abduction + internal tibial rotation) compared with isolate uniplanar loadings.29 More recently, they showed that the application of an axial compressive force to the tibia will generate combined anterior tibial translation, knee abduction, and internal tibial rotation, which result in elevated ACL force at shallow knee flexion angles.30 Later, Shin et al45 used a computer modeling approach to study the effect of increasing knee abduction and internal rotation moments in uni- and multiplanar modes on peak ACL strain during simulated landings. They showed increased peak ACL strain under combined knee abduction and internal tibial rotation moments to levels greater than those seen under each of the uniplanar loadings alone.45 More recently, Oh and colleagues38 showed significantly greater peak ACL strain generated under combined knee abduction and internal tibial rotation moments than under baseline (no frontal and axial plane moments) during a cadaveric simulation of landing.

In the current study, simulated landings under combined multiplanar loading conditions significantly increased the peak ACL strain levels by up to 3.5-fold (Figure 2D) and resulted in maximum peak ACL strain among all simulated loading conditions (Figure 3). The data further showed that the combined multiplanar loading elevated the peak ACL strain to levels beyond those observed under each modes of uniplanar loading conditions (Figure 3). These observations correspond to the fact that 47% of ACL failures occurred under combined multiplanar loading, whereas only 20%, 27%, and 7% of ACL failures were generated under anterior shear force, knee abduction, and internal tibial rotation moments, respectively.28 While the findings of this study are in agreement with those reported by other investigators,28,38,45 the extensive parametric loading conditions (including the injurious magnitudes of external loads) and the physiologic simulation of landing (as a well-established high-risk task) have helped us to mitigate some of the limitation associated with previous studies.

It was also noted that the additive nature of combined multiplanar loading on peak ACL strain is disproportional and nonlinear, with generated peak ACL strain values under combined multiplanar loading being greater than peak ACL strain under each uniplanar loading and smaller than the linear summation of ACL strain values under each uniplanar loading alone (Table 2). For instance, simulated landings with applied 268 N of anterior shear force, 75 N-m of knee abduction, and 60 N-m of internal tibial rotation moments resulted in peak ACL strain levels of 10.6% ± 1.1%, 12.4% ± 1.7%, and 10.9% ± 2.9%, respectively. However, simulated landing under combined loading (268 N anterior shear + 75 N-m abduction + 60 N-m internal rotation) generated a peak ACL strain of 16.9% ± 1.7%, which is greater than strain levels under individual uniplanar loadings but less than 34% (~summation of average strain values under uniplanar loadings).

Multiplanar Valgus Collapse Mechanism of ACL Injury

Simulated landings in this study demonstrated that both externally applied knee abduction and internal tibial rotation moments have similar increasing effects on peak ACL strain levels (Figure 2, B and C). However, knee abduction rotation showed a greater degree of correlation with peak ACL strain and was shown to explain a higher percentage of variability in peak ACL strain than did internal tibial rotation (Figure 4). Most importantly, changes in knee abduction rotation were shown to have a significantly greater effect size on peak ACL strain levels than did the quantified effect size of the internal tibial rotation. This larger effect size (by ~2-fold) indicates greater sensitivity of peak ACL strain to changes in knee abduction rotation compared with changes in internal tibial rotation. This is in agreement with the in vivo biomechanical and video analysis studies that show increased risk and/or frequency of noncontact ACL injuries in the presence of knee abduction.19,26,27,39 Considering the fact that both knee abduction and internal tibial torques lead to knee valgus collapse,21,34,47 the current findings support multiplanar knee valgus collapse as one of the primary mechanisms of noncontact ACL injuries in the presence of axial impact loads observed during landing. Given the fact that the MCL has been reported to be the primary restraint against knee valgus motion and abduction loads,16 a high incidence of combined ACL/MCL injuries would be expected under valgus collapse mechanism. However, concomitant ACL/MCL injuries make up only a small percentage (4%-17%) of total ACL injuries.40,42 This can be explained by our recent findings demonstrating that although both the ACL and MCL resist knee valgus/abduction during landing, physiological magnitudes of the applied loads leading to high ACL strain levels and injury are not sufficient to compromise MCL integrity.40 This proposed injury mechanism is also supported by the bone bruise patterns across the lateral femoral condyle and tibial plateau reported in more than 80% of the noncontact ACL injuries.48,49 Valgus collapse increases the compression across the lateral
and develop new prevention and rehabilitation strategies, thus limiting the risk of ACL and secondary injuries. Prevention strategies that neglect to address the multplanar contributions to ACL injury may seriously hamper injury prevention efforts. Intervention programs that address multiple planes of loading including knee abduction are needed to effectively mitigate the risk of ACL injury. Such measures may in turn minimize associated long-term clinical sequelae associated with these injuries, in particular, the devastating consequences of posttraumatic knee osteoarthritis.

ACKNOWLEDGMENT

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