

Hamstrings Stiffness and Landing Biomechanics Linked to Anterior Cruciate Ligament Loading

J. Troy Blackburn, PhD, ATC*; Marc F. Norcross, PhD, ATC†; Lindsey N. Cannon, BA‡; Steven M. Zinder, PhD, ATC§

*Neuromuscular Research Laboratory, University of North Carolina at Chapel Hill; †Department of Nutrition and Exercise Sciences, Oregon State University, Corvallis; ‡Department of Physical Therapy, East Carolina University, Greenville, NC; §Department of Orthopaedics, University of South Florida, Tampa

Context: Greater hamstrings stiffness is associated with less anterior tibial translation during controlled perturbations. However, it is unclear how hamstrings stiffness influences anterior cruciate ligament (ACL) loading mechanisms during dynamic tasks.

Objective: To evaluate the influence of hamstrings stiffness on landing biomechanics related to ACL injury.

Design: Cross-sectional study.

Setting: Research laboratory.

Patients or Other Participants: A total of 36 healthy, physically active volunteers (18 men, 18 women; age = 23 ± 3 years, height = 1.8 ± 0.1 m, mass = 73.1 ± 16.6 kg).

Intervention(s): Hamstrings stiffness was quantified via the damped oscillatory technique. Three-dimensional lower extremity kinematics and kinetics were captured during a double-legged jump-landing task via a 3-dimensional motion-capture system interfaced with a force plate. Landing biomechanics were compared between groups displaying high and low hamstrings stiffness via independent-samples *t* tests.

Main Outcome Measure(s): Hamstrings stiffness was normalized to body mass ($\text{N/m}\cdot\text{kg}^{-1}$). Peak knee-flexion and -valgus angles, vertical and posterior ground reaction forces,

anterior tibial shear force, internal knee-extension and -varus moments, and knee-flexion angles at the instants of each peak kinetic variable were identified during the landing task. Forces were normalized to body weight, whereas moments were normalized to the product of weight and height.

Results: Internal knee-varus moment was 3.6 times smaller in the high-stiffness group ($t_{22} = 2.221$, $P = .02$). A trend in the data also indicated that peak anterior tibial shear force was 1.1 times smaller in the high-stiffness group ($t_{22} = 1.537$, $P = .07$). The high-stiffness group also demonstrated greater knee flexion at the instants of peak anterior tibial shear force and internal knee-extension and -varus moments (t_{22} range = 1.729–2.224, $P < .05$).

Conclusions: Greater hamstrings stiffness was associated with landing biomechanics consistent with less ACL loading and injury risk. Musculotendinous stiffness is a modifiable characteristic; thus exercises that enhance hamstrings stiffness may be important additions to ACL injury-prevention programs.

Key Words: viscoelastic, musculotendinous, valgus, anterior tibial shear force

Key Points

- Individuals with greater hamstrings stiffness displayed more favorable landing biomechanics for anterior cruciate ligament (ACL) loading and injury risk than individuals with less hamstrings stiffness as evidenced by smaller frontal-plane knee moments and a more-flexed knee at the instants of critical biomechanical knee events.
- Greater hamstrings stiffness was associated with smaller anterior tibial shear forces.
- A high level of hamstrings stiffness may limit ACL injury risk by limiting frontal- and sagittal-plane ACL-loading mechanisms.

Anterior cruciate ligament (ACL) injury commonly occurs during landing,¹ and researchers have suggested that a landing biomechanics profile consisting of large ground reaction forces, anterior tibial shear force, knee-valgus angle, and external knee-flexion and -valgus moments increases ACL loading.^{2–4} A more-extended knee during landing exacerbates this profile, whereas a more-flexed knee decreases these variables,^{4–6} likely limiting ACL loading and injury risk. For example, Blackburn and Padua⁵ demonstrated that increasing knee-flexion angle during landing reduced ground reaction forces. Similarly, Pollard et al⁶ categorized participants into high- and low-flexion groups based on performance of

a landing task and reported smaller knee-valgus angles and moments in the high-flexion group.

Stiffness quantifies the resistance of the musculotendinous unit to lengthening, and hamstrings stiffness may have important implications for ACL loading and injury risk. Greater hamstrings stiffness is associated with greater function in ACL-deficient individuals.⁷ During controlled perturbations, healthy individuals with greater hamstrings stiffness also display less anterior tibial translation, which is an arthrokinematic motion that directly loads the ACL.⁸ Given that anterior tibial translation results from anterior tibial shear force, greater hamstrings stiffness seemingly would resist anterior tibial shear force during landing more effectively than less hamstrings stiffness. Greater ham-

strings stiffness also is correlated with less hamstrings flexibility.⁹ This heightened resistance to knee extension may lead to a more flexed knee during landing, producing more favorable landing biomechanics for ACL loading and injury risk. This notion is supported by Boden et al,¹ who reported that participants with ACL injuries displayed greater hamstrings flexibility than an uninjured cohort, suggesting that “above-average” hamstrings flexibility and, therefore, less hamstrings stiffness may increase ACL injury risk.

Musculotendinous stiffness is a modifiable neuromuscular property^{10,11} that could be targeted in ACL injury-prevention programs. Whereas greater hamstrings stiffness appears to limit ACL loading during controlled perturbations,⁸ it is unclear how hamstrings stiffness influences biomechanical ACL-loading mechanisms during dynamic tasks in which ACL injury commonly occurs. Therefore, the purpose of our investigation was to evaluate the influence of hamstrings musculotendinous stiffness on lower extremity kinematics and kinetics during landing. We hypothesized that individuals with greater hamstrings stiffness would display greater knee flexion during landing, resulting in smaller peak ground reaction forces, anterior tibial shear forces, internal knee-extension and -varus moments (ie, the internal/muscular responses to external moments), and knee-valgus angles. We also hypothesized that individuals with greater hamstrings stiffness would display greater knee-flexion angles at the instants of peak kinetics.

METHODS

Participants

A total of 36 healthy individuals (18 men, 18 women; age = 23 ± 3 years, height = 1.8 ± 0.1 m, mass = 73.1 ± 16.6 kg) volunteered to participate. Participants had no history of ACL injury, lower extremity surgery, neurologic disorder, or lower extremity musculoskeletal injury within the 6 months before data collection and were physically active, involved in at least 20 minutes of physical activity 3 times per week. All participants provided written informed consent, and the study was approved by the University of North Carolina at Chapel Hill Biomedical Institutional Review Board.

Procedures

We collected data during a single testing session in which we assessed hamstrings stiffness and landing biomechanics in a counterbalanced order. Because hamstrings stiffness does not differ between limbs in healthy individuals,¹² all data were sampled from the right lower extremity only. Hamstrings maximal voluntary isometric contractions (MVICs) were performed to determine standardized loading conditions for the stiffness assessment. Participants were positioned prone with the right hip supported in 30° of flexion and the foot fixed to a loading device so the knee was maintained in 30° of flexion and the calcaneus was in contact with a load cell (model 41; Honeywell Sensotec, Columbus, OH) (Figure 1A). Participants performed a submaximal warm-up contraction, then a 5-second maximal knee-flexion effort during which we sampled load cell data.

We assessed hamstrings stiffness by modeling the knee as a single-degree-of-freedom mass-spring system and observing the damping effect imposed by the hamstrings on oscillatory knee flexion and extension. We positioned participants identically as for the hamstrings MVICs except the foot was free to move (Figure 1B). A splint was secured over the plantar aspect of the foot and posterior shank to standardize ankle position and gastrocnemius length, and weights representing 45% MVIC were secured near the ankle. The investigator (J.T.B. or M.F.N.) positioned the participant's shank parallel to the floor, and the participant contracted the hamstrings isometrically to maintain this position. Within 5 seconds after this contraction, the investigator applied a downward manual perturbation to the calcaneus, extending the knee and initiating oscillatory knee flexion and extension. Participants were instructed not to intervene with the perturbation^{7,13} and to attempt to keep the hamstrings active only to the level necessary to support the shank in the testing position. This oscillatory motion was recorded via the tangential acceleration of the shank segment obtained from an accelerometer (model 356A32; PCB Piezotronics, Inc, Depew, NY) attached to the splint on the posterior shank and foot. The time interval between the first 2 oscillatory peaks (t_1 and t_2) was used to calculate the damped frequency of oscillation (Figure 2). Next, we used this value to calculate stiffness via the equation $k = 4\pi^2mf^2$, where k is stiffness, m is the system mass (shank and foot segment¹⁴ + 45% MVIC), and f is the damped frequency of oscillation. Participants performed 5 trials separated by 30-second rests to reduce the likelihood of fatigue. Given that stiffness increases as a function of mass,^{15,16} we normalized stiffness values to participant mass before statistical analyses. This method has been used extensively to assess lower extremity muscle stiffness^{7,8,12,16} and demonstrated excellent intrasession reliability in this investigation (intraclass correlation coefficient [2,1] = 0.82, standard error of the mean = 2.90 N/m·kg⁻¹).

The musculotendinous unit is viscoelastic/loading-rate sensitive, so the magnitude of perturbation influences its stiffness. Therefore, fluctuations in the magnitude of the manual perturbation applied to the shank may have influenced the stiffness values. To address this potential limitation, we calculated the magnitude of perturbation as the product of the peak downward acceleration immediately after perturbation onset (Figure 2) and the system mass (shank and foot segment¹⁴ + 45% MVIC). The average perturbation magnitude was 56 ± 11 N, and the average coefficient of variation across trials (standard deviation / mean) was $12\% \pm 5\%$. Furthermore, the regression of normalized stiffness on perturbation magnitude ($r^2 = 0.03$) demonstrated that variance in stiffness attributable to variance in perturbation magnitude was negligible.

The landing task involved a double-legged jump landing from a 30-cm height located at 50% of the participant's height from 2 force plates (model 4060; Bertec Corporation, Columbus, OH). Participants were instructed to minimize upward displacement upon leaving the box, to land with each foot centered on a single force plate, and to perform a maximal vertical leap immediately after ground contact. Electromagnetic motion-capture sensors (mini-BIRD 800; Ascension Technology Corporation, Burlington, VT) were placed on the pelvis over the sacrum, midthigh,

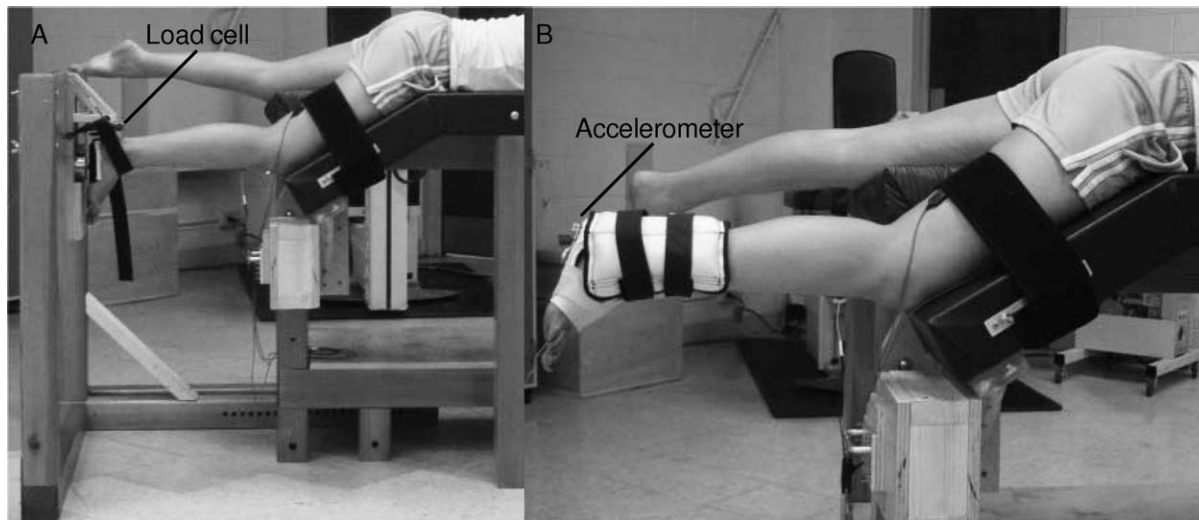


Figure 1. Participant positioning for assessment of A, hamstrings maximal voluntary isometric contraction, and B, hamstrings stiffness.

and proximal anteromedial shank to measure lower extremity kinematics (Figure 3). Three-dimensional coordinate data and ground reaction forces were sampled during the landing task. Participants performed 3 practice trials to familiarize themselves with the task, then performed 5 recorded trials separated by 30-second rests to reduce the likelihood of fatigue. The foot segment was not included in the biomechanical model of the lower extremity due to the potential for substantial motion artifact in foot-segment kinematics near ground contact.¹⁷ Given that peak kinetic ACL-loading mechanisms and ACL injury likely occur shortly after initial ground contact,^{18,19} these errors in ankle-joint motion can substantially influence knee-joint kinetics.¹⁷ Instead, we used a static foot contribution in the model in which angular acceleration of the foot segment was assumed to be zero. Authors^{4,20–23} of previous investigations of ACL-loading mechanisms have used similar models. Unpublished data from our research group have indicated that excluding the foot segment has a negligible effect on knee-joint kinetics during this time interval (Appendix).

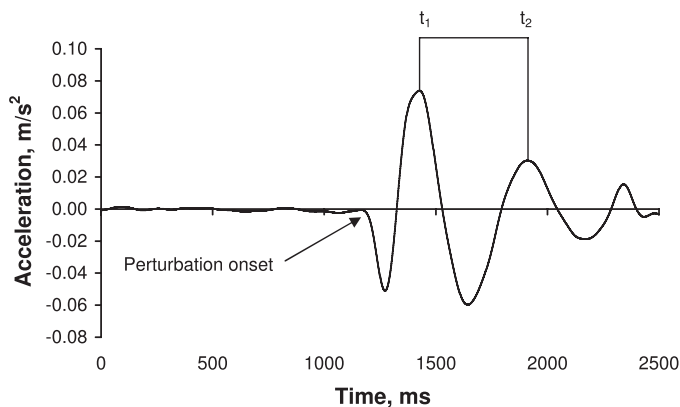


Figure 2. Tangential shank acceleration during assessment of hamstrings stiffness. Stiffness was calculated via the equation $k = 4\pi^2mf^2$, where k is stiffness, m is the system mass, and f is the damped frequency of oscillation ($f = 1 / [t_2 - t_1]$).

Data Sampling and Reduction

Electromagnetic sensor coordinate data were sampled at 100 Hz, and force-plate, accelerometer, and load-cell data were sampled at 1000 Hz. Coordinate data were low-pass filtered at 10 Hz with a fourth-order Butterworth filter, time synchronized to the analog data, and resampled to 1000 Hz via linear interpolation. The world, segment, and force-plate axis systems were established so that positive values for the x-axis, y-axis, and z-axis were defined as forward/anterior, leftward/medial, and upward/superior, respectively. We created a segment linkage model of the lower extremity by digitizing the medial and lateral femoral epicondyles and malleoli and the left and right anterosuperior iliac spines. Locations of the knee- and ankle-joint centers were defined as the midpoints of the digitized medial and lateral femoral epicondyles and malleoli, respectively, whereas the hip-joint center was estimated as a function of the 3-dimensional distance between the digitized left and right anterosuperior iliac spines.²⁴ We calculated knee-joint angles as Euler angles (YXZ sequence) defined as motion of the shank reference frame relative to the thigh reference frame such that flexion, varus, and internal rotation represented positive values. Force-plate data were low-pass filtered at 50 Hz with a fourth-order Butterworth filter and combined with kinematic and anthropometric data to derive the net forces acting on the shank segment and internal knee moments via an inverse dynamics solution.²⁵ Load-cell and accelerometer data were low-pass filtered at 10 Hz with a fourth-order Butterworth filter. Biomechanical variables were derived via commercial motion-capture software (The Motion Monitor; Innovative Sports Training, Inc, Chicago, IL).

For MVICs, we calculated a consecutive 100-millisecond moving average (ie, all 100-millisecond intervals) from which we used the largest hamstrings force value to represent the MVIC. Peak kinematics (knee flexion and valgus angles) and kinetics (vertical and posterior ground reaction forces, anterior tibial shear force, and internal knee-extension and -varus moments) were identified during the *loading phase of the landing*, which was defined as the interval from initial ground contact (vertical ground reaction force > 10 N) to peak knee-flexion angle. Ground

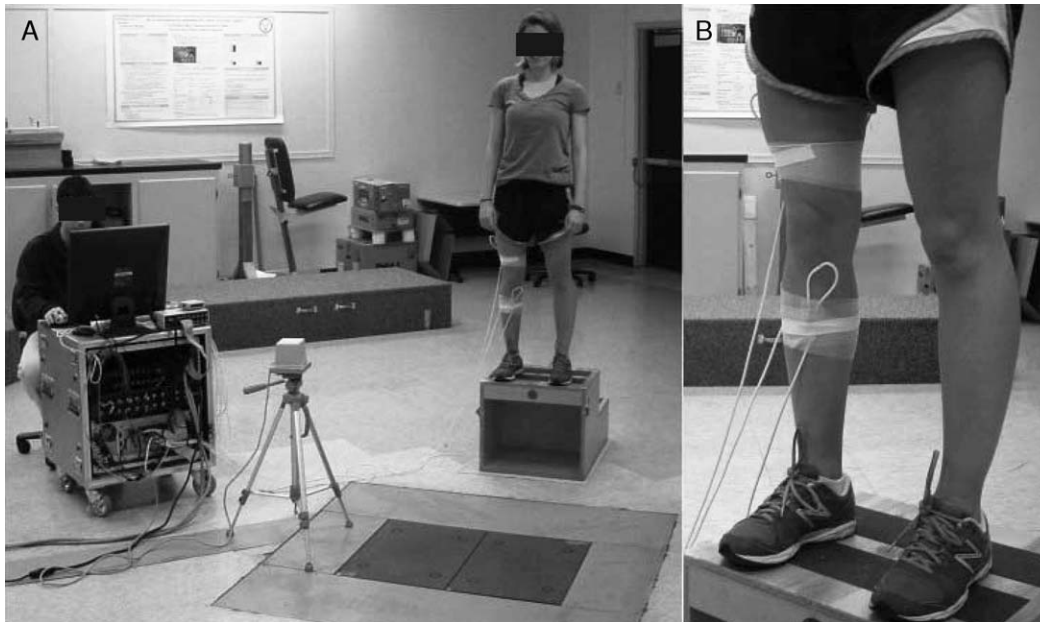


Figure 3. A, Participant positioning and B, electromagnetic sensor locations for the jump-landing task.

reaction forces and anterior tibial shear force were normalized to weight (N), and joint moments were normalized to the product of weight and height (N·m).⁴ We also identified knee-flexion angles at the instant of each peak kinetic variable.

Statistical Analyses

Mean values for each dependent variable were calculated across the 5 trials for each task. We arranged stiffness data into tertiles ($n = 12$ per tertile) and performed independent-samples t tests to compare stiffness and landing biomechanics between the highest and lowest tertiles (ie, high-stiffness and low-stiffness groups). The use of tertiles to define the high-stiffness and low-stiffness groups is arbitrary and not based on meaningful thresholds for stiffness. However, this grouping of participants eliminated individuals who overlapped the extremes of the distribution (ie, the middle tertile), creating 2 distinct groups with different mean stiffness values. All analyses evaluated 1-tailed hypotheses with the α level set a priori at equal to or less than .05. We used SPSS (version 19; IBM Corporation, Armonk, NY) for all analyses.

RESULTS

Hamstrings stiffness was greater in the high-stiffness (25.81 ± 4.61 N/m·kg⁻¹) than low-stiffness (14.52 ± 1.51 N/m·kg⁻¹) group ($t_{22} = 8.074$, $P < .001$), confirming the presence of 2 groups with differing hamstrings stiffness. These groups roughly were stratified equally for sex (low = 7 men, 5 women; high = 5 men, 7 women). The results for high-stiffness and low-stiffness group comparisons are described for kinematics and kinetics.

Kinematics

Peak knee-flexion ($t_{22} = 0.970$, $P = .17$) and -valgus ($t_{22} = 0.971$, $P = .28$) angles did not differ between the high-stiffness and low-stiffness groups. However, knee-flexion

angle was greater in the high-stiffness group at the instants of peak internal knee-varus moment ($t_{22} = 2.224$, $P = .02$), peak internal knee-extension moment ($t_{22} = 1.743$, $P = .048$), and peak anterior tibial shear force ($t_{22} = 1.729$, $P = .049$). We found no other group differences in kinematics. Values for each kinematic variable for the high-stiffness and low-stiffness groups are detailed in Figure 4.

Kinetics

Peak internal knee-extension moment, internal knee-varus moment, and anterior tibial shear force occurred 77 ± 42 milliseconds, 72 ± 54 milliseconds, and 119 ± 74 milliseconds, respectively, after initial ground contact. Peak internal knee-varus moment was smaller in the high-stiffness than low-stiffness group ($t_{22} = 2.221$, $P = .02$) (Figure 4B). Given that the net internal joint moment reflects the musculoskeletal response to external loading, these data indicate less knee-valgus loading (ie, a smaller internal knee-varus moment) in individuals with greater hamstrings stiffness. A statistical trend indicated that anterior tibial shear force was smaller in the high-stiffness than low-stiffness group ($t_{22} = 1.537$, $P = .07$, observed power = 0.44) (Figure 4A). This difference represents a moderate-to-large effect size (0.63), and post hoc power analyses²⁶ indicated that 32 participants per group would have been necessary to provide an a priori power of 0.80. We found no other group differences or statistical trends (ie, $P < .10$ for kinetics). Values for each kinetic variable for the high- and low-stiffness groups are detailed in Figure 5.

DISCUSSION

We designed our investigation to evaluate the influence of hamstrings stiffness on landing biomechanics associated with ACL loading and injury risk. The results indicated that individuals with greater hamstrings stiffness displayed smaller peak frontal-plane knee moments and greater knee flexion at the instants of peak anterior tibial shear force,

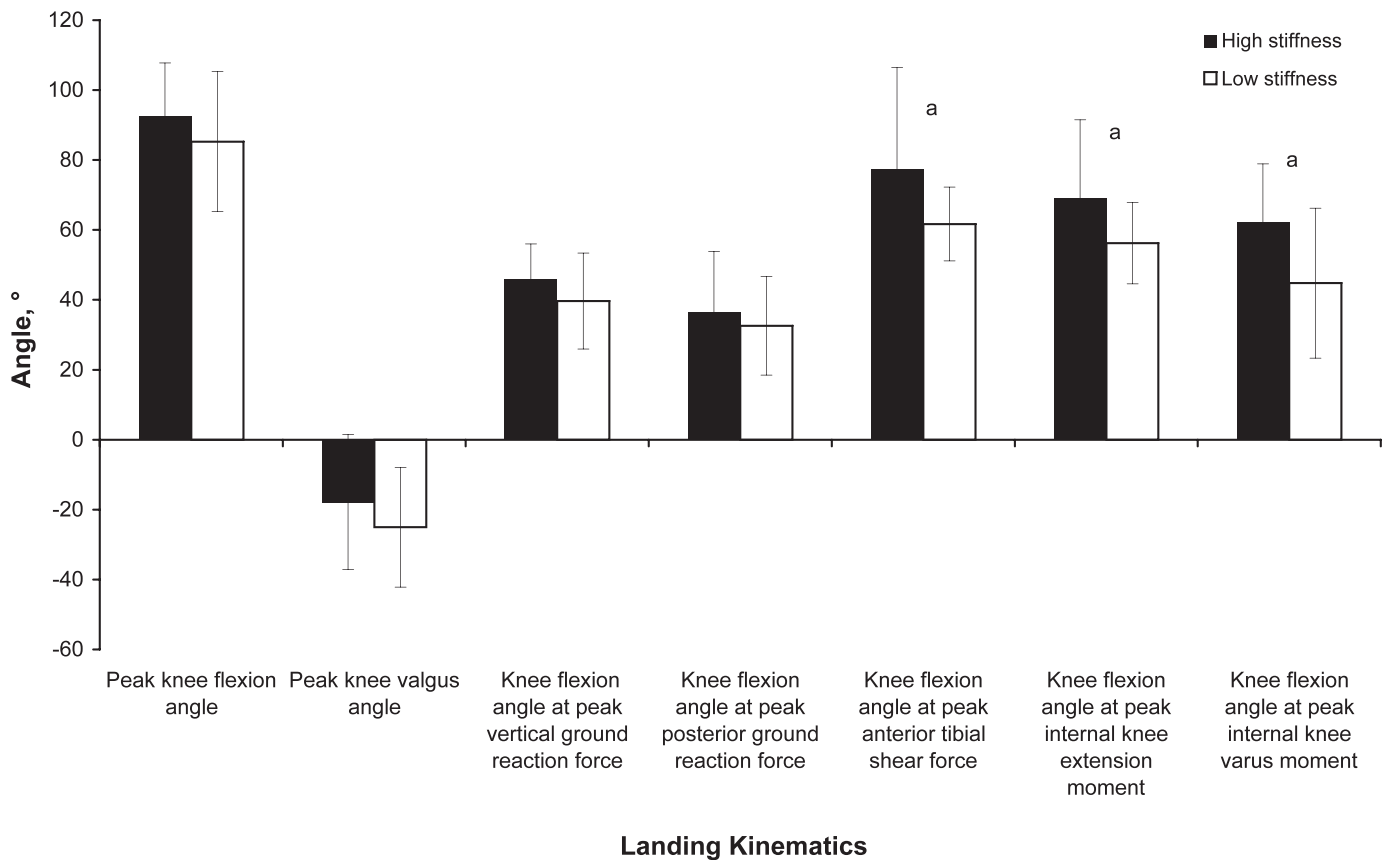


Figure 4. Comparison of landing kinematics between high-stiffness and low-stiffness groups (mean \pm SD). Negative values reflect knee-valgus motion. ^a Indicates difference between high- and low-stiffness groups ($P < .05$).

internal knee-varus moment, and internal knee-extension moment during landing than individuals with less hamstrings stiffness. Peak anterior tibial shear force was also smaller in individuals with greater hamstrings stiffness, which was a finding that approached statistical significance ($P = .07$).

Hewett et al² demonstrated prospectively that peak external knee-valgus moment predicted ACL injury risk with 78% sensitivity and 73% specificity. In addition, individuals who sustained an ACL injury displayed peak external knee-valgus moments that were 2.5 times greater on average than the corresponding values in uninjured individuals. The low-stiffness group in our study displayed peak internal knee-varus moments (ie, the internal musculoskeletal response to an external valgus moment) that were 3.6 times greater than in the high-stiffness group, differences that are comparable with those Hewett et al² reported for injured versus uninjured individuals. Therefore, a high level of hamstrings stiffness may limit ACL injury risk by limiting frontal-plane knee loading. Numerous types of training, including isometric,^{11,27-29} isotonic,³⁰ eccentric,³¹ plyometric,^{10,30} and endurance,^{32,33} have been demonstrated to enhance musculotendinous stiffness. Hewett et al³⁴ noted that plyometric training decreased the external knee-valgus moment during a double-legged landing task. Given that plyometric training increases musculotendinous stiffness, an increase in hamstrings stiffness may explain why the plyometric training that Hewett et al³⁴ used reduced frontal-plane knee loading.

The ACL is loaded via sagittal-plane, frontal-plane, and transverse-plane mechanisms.³⁵ Given that the hamstrings attach posteriorly on the medial and lateral aspects of the shank segment, they can influence 3-dimensional knee-joint loading.^{8,36-38} Hamstrings activity can limit frontal-plane knee loading during controlled loading conditions^{39,40} and during static⁴¹ and dynamic tasks.⁴² Cadaveric data also have indicated that hamstrings force can limit ACL loading attributable to anterior tibial shear force.^{38,43} Our data suggest that hamstrings stiffness influences both sagittal-plane (anterior tibial shear force) and frontal-plane (internal knee-varus moment) ACL-loading mechanisms. In addition, individuals with greater hamstrings stiffness displayed a more-flexed knee at the instants of peak anterior tibial shear force, internal knee-extension moment, and internal knee-varus moment. The ACL elevation angle decreases with knee flexion, orienting the ACL less vertically in the knee-joint space.^{44,45} Ligament is optimized morphologically to resist tensile rather than shear loading⁴⁶; thus greater knee flexion orients the ACL more favorably for resisting sagittal-plane loading. Therefore, the stress imparted to the ACL via sagittal-plane loading decreases as knee flexion increases.^{35,38,45} In this manner, the same magnitudes of anterior tibial shear force and internal knee-extension moment likely are associated with less ACL loading at a greater knee-flexion angle than at a more-extended knee angle. Internal knee-extension moment primarily is derived from quadriceps activity, and quadriceps force can load and even rupture the cadaveric ACL.^{47,48} The ability of the quadriceps to generate anterior

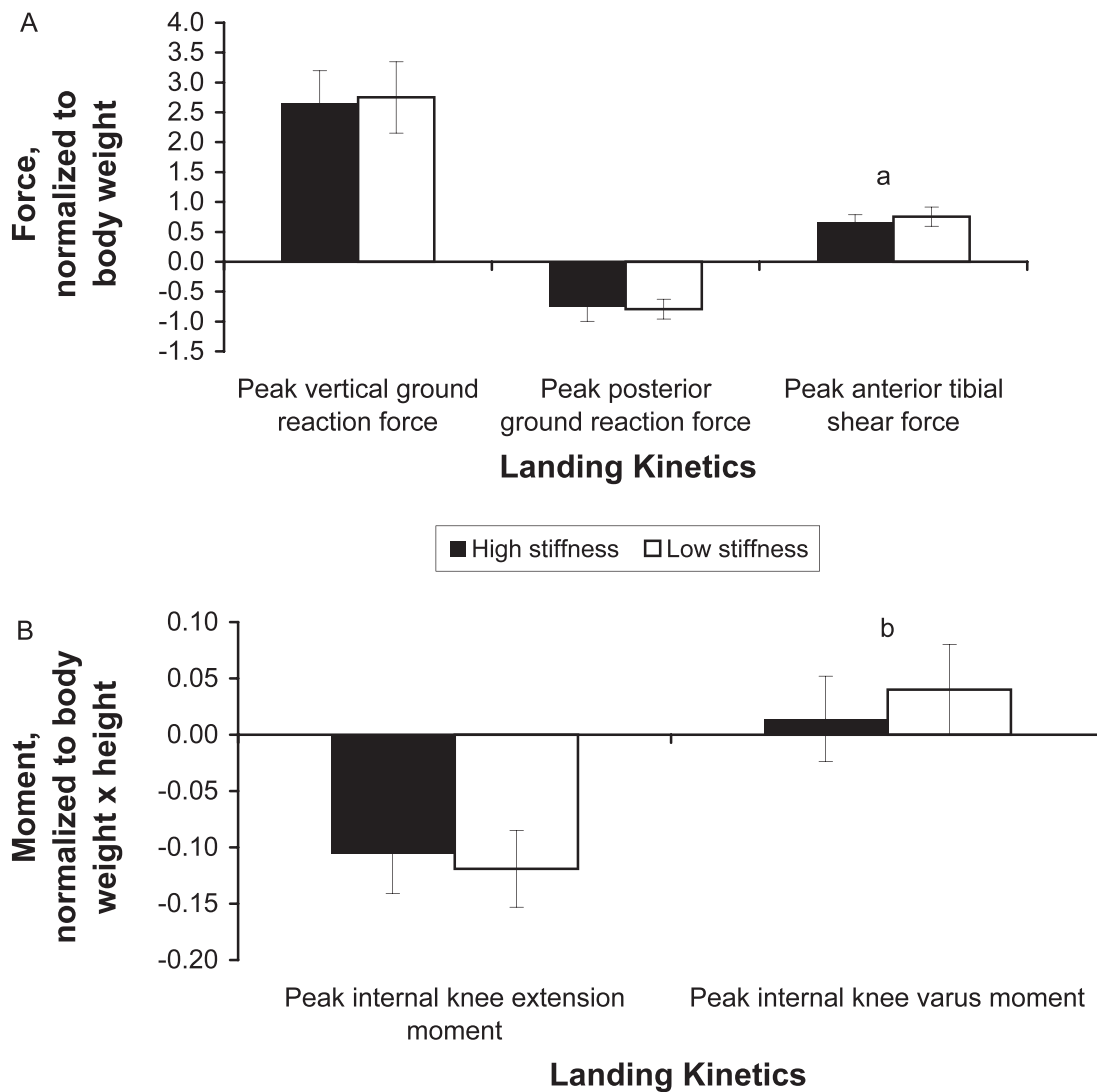


Figure 5. Comparison of landing A, forces and B, moments between high-stiffness and low-stiffness groups (mean \pm SD). Negative values reflect posterior forces and extension moments. ^a Indicates difference between high-stiffness and low-stiffness groups ($P < .05$). ^b Indicates statistical trend for difference between high-stiffness and low-stiffness groups ($P < .10$).

tibial shear force is diminished with knee flexion due to a decrease in the insertion angle of the patellar tendon on the tibia.⁴⁸ Therefore, ACL loading attributable to a given internal knee-extension moment is smaller at greater knee-flexion angles. Finally, Markolf et al³⁵ demonstrated that ACL loading attributable to isolated valgus loading and combined valgus and anterior shear loading decreases with knee flexion. The finding that individuals with greater hamstrings stiffness exhibited greater knee flexion at the instants of peak anterior tibial shear force, internal knee-extension moment, and internal knee-valgus moment during landing suggests that these individuals experienced less ACL loading than those with less hamstrings stiffness.

Whereas the finding that greater hamstrings stiffness was associated with smaller ACL-loading mechanisms agrees with our hypotheses, we anticipated that this influence would be a consequence of greater knee flexion. Our hypothesis regarding the effect of hamstrings stiffness on knee-flexion angle was based on the work of Blackburn et al.⁹ They demonstrated a correlation between hamstrings stiffness and extensibility, which they assessed by placing

the hip in 90° of flexion and instructing participants to actively extend the knee as far as possible. During landing, the hamstrings simultaneously are lengthened proximally due to hip flexion but shortened distally due to knee flexion. As such, their passive tension and extensibility likely have little influence on sagittal-plane knee kinematics during landing because hamstrings length remains in the midrange rather than approaching its extremes, as it does during the assessment of extensibility.⁴⁹ Our hypotheses regarding ground reaction forces and internal knee-extension moment were predicated on previous research⁴⁻⁶ in which investigators associated greater knee flexion with smaller values for these variables. Therefore, the lack of association between hamstrings stiffness and peak knee-flexion angle likely explains the lack of association with these kinetic variables.

Limitations

The biomechanical data in this investigation were derived from a double-legged landing in a laboratory setting. Whereas similar tasks have been demonstrated to predict ACL injury risk,² we do not know if this task represents

those during which ACL injury occurs or if the data are comparable with those of other more challenging dynamic tasks, such as single-legged landings and cutting tasks. Furthermore, the segment-linkage model of the lower extremity we used to calculate knee kinetics did not include the foot segment. We excluded the foot segment to avoid the kinematic noise caused by ground contact and the associated errors introduced to the calculations of knee-joint kinetics.¹⁷ Whereas this model produces kinetic values that are slightly larger than those resulting from models that include the foot segment, unpublished data from our research team indicated that these errors are small and likely had a negligible influence on our results (Appendix). These results may be specific to the task, however, because we calculated knee-joint kinetics soon after ground contact, when ankle-joint contributions to inverse dynamics were small. The participants also constituted a convenience sample of recreational athletes, and our definition of *minimal physical activity* was somewhat liberal; therefore, they may not represent individuals in whom ACL injury risk is greatest.⁵⁰ Last, the oscillatory technique for measuring musculotendinous stiffness reflects contributions from the knee-flexor musculature. However, using identical methods, Blackburn et al¹³ demonstrated that the contributions from the gastrocnemii were negligible. We attempted to standardize any contributions from the gastrocnemii by restricting sagittal-plane ankle motion via application of the rigid splint. Furthermore, the gastrocnemii can introduce ACL loading by producing posterior translation of the femur relative to the tibia,⁵¹ thus it is unlikely that they contributed to the “better” biomechanics profile in the high-stiffness group.

Summary and Clinical Application

Individuals with greater hamstrings stiffness displayed more favorable landing biomechanics for ACL loading and injury risk, as evidenced by smaller frontal-plane knee moments and a more-flexed knee at the instants of critical kinetic events. A trend in the data also suggested greater hamstrings stiffness is associated with less ACL loading attributable to anterior tibial shear force. These findings suggest a high level of hamstrings stiffness may limit ACL injury risk.

Whereas more research is certainly necessary to validate this notion objectively, increasing hamstrings stiffness may have negative consequences for hamstrings musculotendinous injury. Watsford et al⁵² demonstrated prospectively that individuals who sustained acute hamstrings strain injuries had greater hamstrings stiffness than an uninjured cohort. However, whereas the bilateral average hamstrings stiffness differed between groups, that of the injured limb was not different from the uninjured group. Although the implications of these seemingly contradictory findings are unclear, we suggest the existence of an optimal level of hamstrings stiffness at which the risks of ACL injury and musculotendinous injury may be minimized. Future research is necessary to evaluate the injury-specific influences of hamstrings stiffness.

Researchers should attempt to extrapolate these findings to more dynamic and challenging tasks that are more representative of scenarios during which ACL injury occurs and to the populations at heightened risk of ACL injury. Given that muscle stiffness can be enhanced via various

training mechanisms, targeting this neuromuscular property may be beneficial for future ACL injury-prevention programs and for rehabilitation of patients with ACL deficiency or reconstruction. However, research is necessary to determine the most effective types of training to enhance hamstrings stiffness and the effects of enhancing hamstrings stiffness on ACL-loading mechanisms and ACL injury risk.

ACKNOWLEDGMENTS

This study was funded by a grant from the Injury Prevention Research Center at the University of North Carolina at Chapel Hill.

REFERENCES

1. Boden BP, Dean GS, Feagin JA Jr, Garrett WE Jr. Mechanisms of anterior cruciate ligament injury. *Orthopedics*. 2000;23(6):573–578.
2. Hewett TE, Myer GD, Ford KR, et al. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes. *Am J Sports Med*. 2005;33(4):492–501.
3. Kernozek TW, Torry MR, Van Hoof H, Cowley H, Tanner S. Gender differences in frontal and sagittal plane biomechanics during drop landings. *Med Sci Sports Exerc*. 2005;37(6):1003–1012.
4. Yu B, Lin CF, Garrett WE. Lower extremity biomechanics during the landing of a stop-jump task. *Clin Biomech (Bristol, Avon)*. 2006; 21(3):297–305.
5. Blackburn JT, Padua DA. Sagittal-plane trunk position, landing forces, and quadriceps electromyographic activity. *J Athl Train*. 2009;44(2):174–179.
6. Pollard CD, Sigward SM, Powers CM. Limited hip and knee flexion during landing is associated with increased frontal plane knee motion and moments. *Clin Biomech (Bristol, Avon)*. 2010;25(2):142–146.
7. McNair PJ, Wood GA, Marshall RN. Stiffness of the hamstring muscles and its relationship to function in anterior cruciate ligament deficient individuals. *Clin Biomech (Bristol, Avon)*. 1992;7(3):131–137.
8. Blackburn JT, Norcross MF, Padua DA. Influences of hamstring stiffness and strength on anterior knee joint stability. *Clin Biomech (Bristol, Avon)*. 2011;26(3):278–283.
9. Blackburn JT, Padua DA, Riemann BL, Guskiewicz KM. The relationships between active extensibility, and passive and active stiffness of the knee flexors. *J Electromyogr Kinesiol*. 2004;14(6): 683–691.
10. Burgess KE, Connick MJ, Graham-Smith P, Pearson SJ. Plyometric vs. isometric training influences on tendon properties and muscle output. *J Strength Cond Res*. 2007;21(3):986–989.
11. Kubo K, Ikebukuro T, Yaeshima K, Yata H, Tsunoda N, Kanehisa H. Effects of static and dynamic training on the stiffness and blood volume of tendon in vivo. *J Appl Physiol*. 2009;106(2):412–417.
12. Jennings AG, Seedhom BB. The measurement of muscle stiffness in anterior cruciate injuries: an experiment revisited. *Clin Biomech (Bristol, Avon)*. 1998;13(2):138–140.
13. Blackburn JT, Riemann BL, Padua DA, Guskiewicz KM. Sex comparison of extensibility, passive, and active stiffness of the knee flexors. *Clin Biomech (Bristol, Avon)*. 2004;19(1):36–43.
14. Dempster WT, Gabel WC, Felts WJ. The anthropometry of manual work space for the seated subject. *Am J Phys Anthropol*. 1959;17: 289–317.
15. Blackburn JT, Bell DR, Norcross MF, Hudson JD, Kimsey MH. Sex comparison of hamstring structural and material properties. *Clin Biomech (Bristol, Avon)*. 2009;24(1):65–70.
16. Granata KP, Wilson SE, Padua DA. Gender differences in active musculoskeletal stiffness: part I. Quantification in controlled

- measurements of knee joint dynamics. *J Electromyogr Kinesiol.* 2002;12(2):119–126.
17. Bisseling RW, Hof AL. Handling of impact forces in inverse dynamics. *J Biomech.* 2006;39(13):2438–2444.
 18. Cerulli G, Benoit DL, Lamontagne M, Caraffa A, Liti A. In vivo anterior cruciate ligament strain behaviour during a rapid deceleration movement: case report. *Knee Surg Sports Traumatol Arthrosc.* 2003;11(5):307–311.
 19. Koga H, Nakamae A, Shima Y, et al. Mechanisms for noncontact anterior cruciate ligament injuries: knee joint kinematics in 10 injury situations from female team handball and basketball. *Am J Sports Med.* 2010;38(11):2218–2225.
 20. Boling MC, Padua DA, Marshall SW, Guskiewicz K, Pyne S, Beutler A. A prospective investigation of biomechanical risk factors for patellofemoral pain syndrome: the Joint Undertaking to Monitor and Prevent ACL Injury (JUMP-ACL) cohort. *Am J Sports Med.* 2009;37(11):2108–2116.
 21. Chappell JD, Yu B, Kirkendall DT, Garrett WE. A comparison of knee kinetics between male and female recreational athletes in stop-jump tasks. *Am J Sports Med.* 2002;30(2):261–267.
 22. Herman DC, Onate JA, Weinhold PS, et al. The effects of feedback with and without strength training on lower extremity biomechanics. *Am J Sports Med.* 2009;37(7):1301–1308.
 23. Padua DA, Marshall SW, Boling MC, Thigpen CA, Garrett WE Jr, Beutler AI. The Landing Error Scoring System (LESS) is a valid and reliable clinical assessment tool of jump-landing biomechanics: the JUMP-ACL study. *Am J Sports Med.* 2009;37(10):1996–2002.
 24. Bell AL, Pedersen DR, Brand RA. A comparison of the accuracy of several hip center location prediction methods. *J Biomech.* 1990;23(6):617–621.
 25. Gagnon D, Gagnon M. The influence of dynamic factors on triaxial net muscular moments at the L5/S1 joint during asymmetrical lifting and lowering. *J Biomech.* 1992;25(8):891–901.
 26. Portney LG, Watkins MP. *Foundations of Clinical Research: Applications to Practice.* 3rd ed. Upper Saddle River, NJ: Prentice-Hall Inc; 2009:832–836.
 27. Kubo K, Kanehisa H, Fukunaga T. Effects of different duration isometric contractions on tendon elasticity in human quadriceps muscles. *J Physiol.* 2001;536(pt 2):649–655.
 28. Kubo K, Kanehisa H, Ito M, Fukunaga T. Effects of isometric training on the elasticity of human tendon structures in vivo. *J Appl Physiol.* 2001;91(1):26–32.
 29. Kubo K, Ohgo K, Takeishi R, et al. Effects of isometric training at different knee angles on the muscle-tendon complex in vivo. *Scand J Med Sci Sports.* 2006;16(3):159–167.
 30. Kubo K, Morimoto M, Komuro T, et al. Effects of plyometric and weight training on muscle-tendon complex and jump performance. *Med Sci Sports Exerc.* 2007;39(10):1801–1810.
 31. Pousson M, Van Hoecke J, Goubel F. Changes in elastic characteristics of human muscle induced by eccentric exercise. *J Biomech.* 1990;23(4):343–348.
 32. Grosset JF, Piscione J, Lambert D, Perot C. Paired changes in electromechanical delay and musculo-tendinous stiffness after endurance or plyometric training. *Eur J Appl Physiol.* 2009;105(1):131–139.
 33. Kubo K, Kanehisa H, Kawakami Y, Fukunaga T. Elastic properties of muscle-tendon complexes in long-distance runners. *Eur J Appl Physiol.* 2000;81(3):181–187.
 34. Hewett TE, Stroupe AL, Nance TA, Noyes FR. Plyometric training in female athletes: decreased impact forces and increased hamstring torques. *Am J Sports Med.* 1996;24(6):765–773.
 35. Markolf KL, Burchfield DM, Shapiro MM, Shepard MF, Finerman GA, Slauterbeck JL. Combined knee loading states that generate high anterior cruciate ligament forces. *J Orthop Res.* 1995;13(6):930–935.
 36. Lloyd DG, Buchanan TS. Strategies of muscular support of varus and valgus isometric loads at the human knee. *J Biomech.* 2001;34(10):1257–1267.
 37. Nyland J, Caborn DN, Shapiro R, Johnson DL, Fang H. Hamstring extensibility and transverse plane knee control relationship in athletic women. *Knee Surg Sports Traumatol Arthrosc.* 1999;7(4):257–261.
 38. Li G, Rudy TW, Sakane M, Kanamori A, Ma CB, Woo SL. The importance of quadriceps and hamstring muscle loading on knee kinematics and in-situ forces in the ACL. *J Biomech.* 1999;32(4):395–400.
 39. Dhaher YY, Tsoumanis AD, Houle TT, Rymer WZ. Neuromuscular reflexes contribute to knee stiffness during valgus loading. *J Neurophysiol.* 2005;93(5):2698–2709.
 40. Zhang LQ, Wang G. Dynamic and static control of the human knee joint in abduction-adduction. *J Biomech.* 2001;34(9):1107–1115.
 41. Claiborne TL, Armstrong CW, Gandhi V, Pincivero DM. Relationship between hip and knee strength and knee valgus during a single leg squat. *J Appl Biomech.* 2006;22(1):41–50.
 42. Besier TF, Lloyd DG, Ackland TR. Muscle activation strategies at the knee during running and cutting maneuvers. *Med Sci Sports Exerc.* 2003;35(1):119–127.
 43. Withrow TJ, Huston LJ, Wojtys EM, Ashton-Miller JA. Effect of varying hamstring tension on anterior cruciate ligament strain during in vitro impulsive knee flexion and compression loading. *J Bone Joint Surg Am.* 2008;90(4):815–823.
 44. Li G, Papannagari R, DeFrate LE, Yoo JD, Park SE, Gill TJ. Comparison of the ACL and ACL graft forces before and after ACL reconstruction: an in-vitro robotic investigation. *Acta Orthop.* 2006;77(2):267–274.
 45. Wu C, Noorani S, Vercillo F, Woo SL. Tension patterns of the anteromedial and posterolateral grafts in a double-bundle anterior cruciate ligament reconstruction. *J Orthop Res.* 2009;27(7):879–884.
 46. Woo SL, Hollis JM, Adams DJ, Lyon RM, Takai S. Tensile properties of the human femur-anterior cruciate ligament-tibia complex: the effects of specimen age and orientation. *Am J Sports Med.* 1991;19(3):217–225.
 47. DeMorat G, Weinhold P, Blackburn T, Chudik S, Garrett W. Aggressive quadriceps loading can induce noncontact anterior cruciate ligament injury. *Am J Sports Med.* 2004;32(2):477–483.
 48. Withrow TJ, Huston LJ, Wojtys EM, Ashton-Miller JA. The relationship between quadriceps muscle force, knee flexion, and anterior cruciate ligament strain in an in vitro simulated jump landing. *Am J Sports Med.* 2006;34(2):269–274.
 49. Alonso J, McHugh MP, Mullaney MJ, Tyler TF. Effect of hamstring flexibility on isometric knee flexion angle-torque relationship. *Scand J Med Sci Sports.* 2009;19(2):252–256.
 50. Arendt EA, Agel J, Dick RW. Anterior cruciate ligament injury patterns among collegiate men and women. *J Athl Train.* 1999;34(2):86–92.
 51. Fleming BC, Renstrom PA, Ohlen G, et al. The gastrocnemius muscle is an antagonist of the anterior cruciate ligament. *J Orthop Res.* 2001;19(6):1178–1184.
 52. Watsford ML, Murphy AJ, McLachlan KA, et al. A prospective study of the relationship between lower body stiffness and hamstring injury in professional Australian rules footballers. *Am J Sports Med.* 2010;38(10):2058–2064.

Appendix

We provide justification for excluding the foot segment from our biomechanical model of the lower extremity. The Table contains unpublished data from our research group comparing knee-joint kinetics derived from biomechanical models with (full model) and without (modified model) foot-segment data. We obtained these data during 5 trials of a landing task identical to the one in the current investigation in a similar sample of 18 participants (10 men, 8 women; height = 1.7 ± 0.1 m, mass = 73.0 ± 11.8 kg) using an optical motion-capture system (Vicon,

Table. Kinetic Comparison of Biomechanical Models

Variable	Full Model, Mean \pm SD	Modified Model, Mean \pm SD	Model Difference ^a	R^{2b}
Internal knee-extension moment, N·m	-160.1 \pm 36.9	-167.7 \pm 38.2	-7.6	0.98
Internal knee-varus moment, N·m	24.6 \pm 11.6	26.5 \pm 12.3	2.0	0.97
Anterior tibial shear force, N	537.9 \pm 106.9	537.0 \pm 107.1	-1.0	0.99

^a Model difference = modified model mean – full model mean.

^b R^2 is the coefficient of determination for the relationship between kinetic values derived from each model.

Centennial, CO) with markers placed on the foot, shank, thigh, and pelvis segments. The data were sampled and processed identically to those in the current investigation with one exception. For the full model, kinematic data from all lower extremity segments were included in the inverse dynamics calculations, whereas in the modified model, the foot-segment data were excluded from these calculations (ie, identical to the current investigation).

The data provided in the Table indicate that the differences in peak knee-joint kinetics between the 2 models are likely negligible because the model difference was substantially smaller than the standard deviation for each kinetic variable. Furthermore, the coefficients of determination (R^2) for the relationships between the models for each kinetic variable indicate extremely high model agreement, because 97% to 99% of the variance in one model was attributable to variance in the other.

Peak internal knee-extension moment, internal knee varus moment, and anterior tibial shear force in the full model occurred at 103 ± 83 milliseconds, 36 ± 31 milliseconds, and 153 ± 80 milliseconds, respectively, after initial ground contact. Although these values differ

somewhat from the data in this manuscript, likely due to several sample-specific factors (eg, energy-absorption strategies, extensor activity, lower extremity strength), they are within range of those in our data (77 ± 42 milliseconds, 72 ± 54 milliseconds, and 119 ± 74 milliseconds, respectively) and correspond with the interval during which peak anterior cruciate ligament loading and injury likely occur.^{18,19} Data from the full model indicated that the ankle-joint moments during this time interval are relatively small, thus the error introduced to knee kinetics via exclusion of the foot segment is also minimal. For example, the mean ankle-extension moment at the instant of peak knee extension moment was 48 N·m, which is a value 3.3 times smaller than the knee-extension moment. Similarly, the frontal-plane ankle moment at peak knee-varus moment was 4 N·m (positive = inversion), or 6.2 times smaller than the peak knee-varus moment. These relatively small values indicate the ankle joint contribution to knee-joint kinetics over the interval immediately after ground contact is small, leading to minimal errors in knee-joint kinetics attributable to exclusion of the foot segment.

Address correspondence to J. Troy Blackburn, PhD, ATC, Neuromuscular Research Laboratory, University of North Carolina at Chapel Hill, 124 Fetzer Hall, CB #8700, Chapel Hill, NC 27599-8700. Address e-mail to troyb@email.unc.edu.