Effect of Increased Quadriceps Tensile Stiffness on Peak Anterior Cruciate Ligament Strain during a Simulated Pivot Landing

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ABSTRACT: ACL injury prevention programs often involve strengthening the knee muscles. We posit that an unrecognized benefit of such training is the associated increase in the tensile stiffness of the hypertrophied muscle. We tested the hypothesis that an increased quadriceps tensile stiffness would reduce peak anteromedial bundle (AM-)ACL relative strain in female knees. Twelve female cadaver knees were subjected to compound impulsive two-times body weight loads in compression, flexion, and internal tibial torque beginning at 15° flexion. Knees were equipped with modifiable custom springs to represent the nonlinear rapid stretch behavior of a normal and increased stiffness female quadriceps (i.e., 33% greater stiffness). Peak AM-ACL relative strain was measured using an in situ transducer while muscle forces and tibiofemoral kinematics and kinetics were recorded. A 3D ADAMS™ dynamic biomechanical knee model was used in silico to interpret the experimental results which were analyzed using a repeated-measures Wilcoxon test. Female knees exhibited a 16% reduction in peak AM-ACL relative strain and 21% reduction in change in flexion when quadriceps tensile stiffness was increased by 33% (mean (SD) difference: 0.97% (0.65%), $p = 0.003$). We conclude that increased quadriceps tensile stiffness reduces peak ACL strain during a controlled study simulating a pivot landing. © 2013 Orthopaedic Research Society. Published by Wiley Periodicals, Inc. J Orthop Res 32:423–430, 2014.

Keywords: anterior cruciate ligament; quadriceps; muscle stiffness; training; computational model

There are 300,000 anterior cruciate ligament (ACL) injuries in the United States each year,[1] with a disproportionate number of these injuries occurring in female athletes.[2,3] ACL injuries are a major public health concern because they are immediately disabling with high associated treatment costs, lost time, and the increased risk of developing early onset knee osteoarthritis.[4] ACL injury prevention programs aimed at reducing the number of injuries in female athletes often include exercises to strengthen the knee extensor muscles, including leg press, squatting, and lunge exercises.[5–9] However, the benefits of such exercises remain uncertain given limited success.[10]

The muscles of female athletes may provide less resistance to knee joint shear and torsional loads than for male athletes.[11,12] Since females have 45% less active quadriceps stiffness than males,[13] peak ACL strain may be routinely higher in females. Indeed, Lipps et al.[14] showed that female knees equipped with a female quadriceps tensile stiffness value had 95% greater peak ACL strain than height- and weight-matched male knees equipped with 25% greater quadriceps tensile stiffness in simulated pivot landings.

Muscle tensile stiffness can be increased by both muscle hypertrophy[15] as well as by increased muscle activation,[16] both of which can be affected by training targeting the innervation and size of the fibers within a muscle through expression of the AKT pathway[17] or inhibition of transforming growth factor-β signaling pathways.[18] Knee extensor strength training in humans will produce an increase in type II muscle fiber cross-sectional area within the vastus lateralis[19,20] as well as an increase in quadriceps rectus femoris volume.[21,22] The associated increase in vastus lateralis musculotendon stiffness, ranging from 16% to 58% after 12 weeks of training, was more strongly associated with gains in strength than in muscle activation.[15,23,24] While previous studies have shown that a large quadriceps force can increase the risk of ACL injury,[25] a knowledge gap exists regarding whether achievable training-related increases in quadriceps tensile stiffness in females via prevention programs could reduce peak ACL strain during a given pivot landing.[26] We therefore tested the primary hypothesis in vitro that increasing quadriceps tensile stiffness would reduce peak AM-ACL relative strain in female knees during a simulated pivot landing, and interpreted our results using an in silico model.

METHODS

Twelve female unembalmed cadaver limbs with no visual signs of surgery and deformities were acquired from the University of Michigan Anatomical Donations Program and the Anatomy Gifts Registry (Hanover, MD). The specimens were of height and weight (mean (SD) age: 58 (14) years; height: 164.5 (6.8) cm; mass: 70.2 (5.3) kg). As previously described,[14] the limbs were dissected, leaving the ligamentous structures of the knee joint intact along with the tendons of the quadriceps, medial and lateral hamstrings, and medial and lateral gastrocnemius. The proximal femur and distal tibia and fibula were cut 20 cm from the knee joint line and potted in a PVC cylinder filled with polymethylmethacrylate (PMMA). The specimens were stored at −20°C until being thawed at room temperature 12 h prior to testing.

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The cadaver limbs were placed in a modified Withrow testing apparatus[27] capable of simulating a single-leg 2*BW pivot landing[14,28] (Fig. 1). The methods for this testing apparatus have been previously outlined.[14] Briefly, each trial begins by dropping an impact weight from 7 cm onto an impact rod in series with a torsional device, delivering an impulsive compound load to the knee joint (Fig. 1). The strut angle within the torsional device controls the gain between internal tibial torque and the compression force, and a cylinder can be keyed into the center of the torsional device to only allow a compressive load. The applied impulsive load was standardized to 2*BW of the specimen and consisted of a compression force, knee flexion moment, and internal tibial torque, peaking at 60 ms. The 2*BW loading allows for large ACL strains while maintaining ACL integrity during testing.[14,26,28] The initial knee flexion angle was 15°, consistent with the knee flexion angle at the time of ACL injury.[29] This will deliver the impulsive force 4.5 cm posterior to the knee joint since the universal joints were concentric in the axial plane and the distance between the knee joint line and each universal joint was 35 cm.

Tibiofemoral kinematics were recorded at 400 Hz with two sets of three infrared emitting diodes (IREDs) using an Optotrak Certus camera system (Northern Digital, Inc., Waterloo, Ontario, Canada). A digitizer related bony anatomic landmarks to the IRED location in order to follow the absolute and relative 3D translations and rotations of the femoral condyles[30] relative to the tibial plateau, throughout each trial. The input and reaction three-dimensional forces and torques applied to the distal tibia and fibula were measured at 2 kHz with 6-axis load cells (MC3A-1000, AMTI, Watertown, MA). Relative strain was measured on the distal 3rd of the anteromedial bundle of the ACL using a differential variable reluctance transducer (DVRT) (3-mm stroke length, MicroStrain, Inc., Burlington, VT).

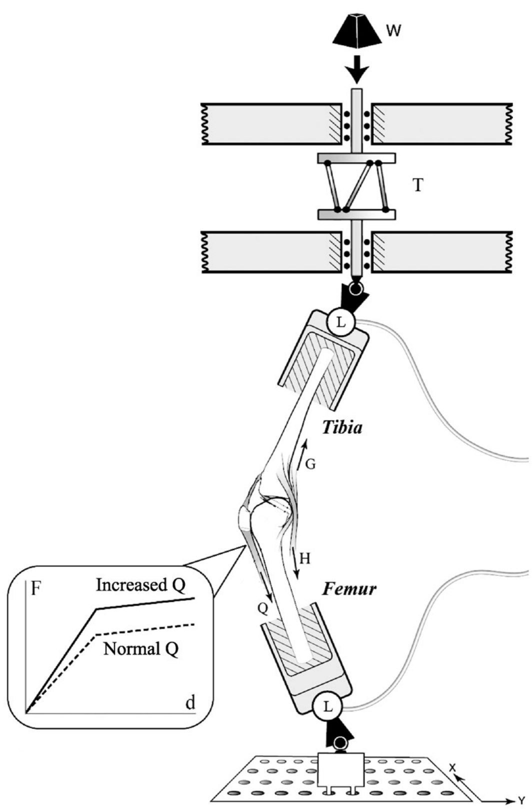

Figure 1. Diagram of the modified Withrow-Oh testing apparatus. Inset: Force-displacement relationship for a nonlinear quadriceps spring representing the normal and increased stiffness female quadriceps values. Abbreviations: W = drop weight; T = torsional device; L = 6-axis load cell; Q = quadriceps tendon; H = hamstrings tendon; G = gastrocnemius tendon. Figure modified from Lipps et al.[14] and Oh et al.[26,28]

Cryoclamps were attached to the quadriceps, hamstrings, and gastrocnemius muscle tendons. Muscle tension was recorded at 2 kHz with in-series uniaxial load cells (TLL-1K and TLL-500, Transducer Techniques, Inc., Temecula, CA). Each muscle tendon and its corresponding load cell were in series with a spring acting as a muscle equivalent. Woven nylon cord (stiffness: 200 N/mm) pretensioned to 70 N served as the muscle equivalent for the medial and lateral hamstrings and medial and lateral gastrocnemius muscles, consistent with prior in vitro studies using this testing apparatus.[14,26,28] The quadriceps tendon was pre-tensioned to 180 N and utilized a nonlinear spring to model the muscle's bi-linear stiffness response to rapid stretch.[14] Throughout this study, we intend to use quadriceps tensile stiffness to refer to the musculotendinous stiffness of the quadriceps femoris muscles and the quadriceps tendon. This study utilized two quadriceps tensile stiffness values: a stiffness values similar to those seen in a typical females[13] (initial stiffness: 156 N/mm; final stiffness: 35 N/mm) as well as a 33% increase in quadriceps tensile stiffness to simulate training increases in musculotendon unit stiffness (initial stiffness: 208 N/mm; final stiffness: 47 N/mm). The freezing of the rectus femoris tendon made the tendon very stiff; therefore, any changes in the stiffness within our in vitro quadriceps musculotendon unit would be caused by changes in stiffness within our nonlinear quadriceps spring.

A repeated measures design (A-B-C-A) was used to measure peak AM-ACL relative strain in female knees undergoing a simulated pivot landing with a normal and increased stiffness female quadriceps spring. Since the impact mass and drop height remained constant between trial blocks, the knee received the same energy input in each trial block. The A trial blocks represented pre-baseline and post-baseline non-pivot landings consisting of 2*BW compressive force with a resultant knee flexion moment. No significant difference in peak AM-ACL relative strain between the baseline blocks indicated that ACL integrity was maintained. The torsional device was activated for the B and C trial blocks to load the knee joint with 2*BW simulated pivot landing consisting of compression force, knee flexion moment, and a 20-Nm internal tibial torque, which produces large ACL strains.[28] The B trial block tested female knees with female quadriceps tensile stiffness under a simulated pivot landing, while the C trial block increased the short range stiffness of the quadriceps spring by 33% to represent a female muscle with increased stiffness. The order of the B and C trial blocks was randomized. Five preconditioning trials were performed prior to data collection to remove hysteresis effects. Following the preconditioning trials, 5 impact trials were performed for each trial block, with 1 preconditioning trial between testing blocks, resulting in 28 total trials.

Biomechanical Model

A published three-dimensional lower-limb dynamic biomechanical model was used to replicate the knee loading within the Withrow-Oh in vitro testing apparatus in order to interpret the experiment results.[28] The model was previously validated in the sagittal plane against experimental data from the testing apparatus,[28] and was updated (Supplemental Text 1) to use bone morphology obtained from magnetic resonance imaging scans of a human knee (3T Phillips Scanner [Best, the Netherlands], 3D-PD sequence, 160 mm field of view, 0.7 mm slice thickness with no inter-slice gap, $0.35\,mm \times 0.35\,mm$ pixel resolution) by segmenting the distal femur, proximal tibia and fibula, and patella (SolidWorks 2010, Dassault Systems SolidWorks Corp., Concord, MA; Rhinoceros, McNeel North America, Seattle, WA). The model was then imported into a dynamic motion simulation software package (MD Adams R3, MSC Software, Inc., Santa Ana, CA), where the viscoelastic muscles, ligaments, and capsule structures were modeled using Kelvin–Voigt elements.[28] For the dynamic analysis, the forces and moments from experimental data were applied to the distal tibia. The quadriceps tensile stiffness was varied in 5% increments from 70% to 130%, with 70% and 100% quadriceps tensile stiffness representing the normal and increased stiffness female experimental values. Peak AM-ACL relative strain and change in knee flexion angle were monitored. Additional information on the biomechanical model, including validation and limitations, is provided in Supplemental Text 1.

Statistical Analysis

Statistical analyses were performed in SPSS 19 (IBM, Inc., Armonk, NY), with the five trials from each testing block averaged to yield a representative data set. The primary hypothesis was tested with Wilcoxon signed-rank tests to analyze the differences in peak AM-ACL relative strain between the A-B-C-A testing blocks. Two-sided paired t-tests were used to analyze dynamic inputs and kinematic outputs between the B and C testing blocks. Pearson correlations were utilized for secondary analyses to compare knee flexion angle and knee abduction angle, along with anterior tibial translation and quadriceps force. A p-value < 0.05 and a correlation coefficient, $|r| < 0.575$ (which is equivalent to $p < 0.05$ for $n = 12$) was considered statistically significant. The sample size ($n = 12$) was based on a two-tailed dependent means power analysis of peak AM-ACL relative strain with varying quadriceps tensile stiffness for six female knees with alpha $= 0.05$ and power $= 0.8$. The a priori effect size was 1.29, requiring a sample size of seven female knees. Additional knees were included in the study to improve power, resulting in a post hoc power of 0.997 and a post hoc effect size of 1.49.

RESULTS

The primary hypothesis was supported in that peak AM-ACL relative strain was reduced in female knees with increased quadriceps tensile stiffness in response to a 2*BW simulated pivot landing (mean (SD) difference: 0.97 (0.65) %, $p = 0.003$, Fig. 2). On average, this resulted in a 16% decrease in peak AM-ACL relative strain with a 33% increase in the short-range quadriceps tensile stiffness. When compared to pre-baseline and post-baseline trials, peak AM-ACL relative strain was greater for simulated pivot landings with both normal ($p = 0.002$ and $p = 0.002$, respectively) and increased female quadriceps tensile stiffness ($p = 0.002$ and $p = 0.002$, respectively). There was no difference between pre-baseline and post-baseline testing

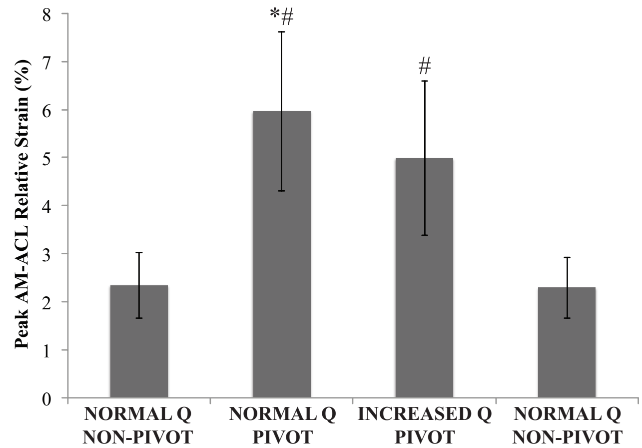

Figure 2. Peak AM-ACL relative strain for 12 female knees tested under the four repeated measures testing conditions. Pre-baseline and post-baseline 2*BW non-pivoting landings (NON-PIVOT) are simulated with compression + flexion knee loads are compared to 2*BW pivot landings (PIVOT) under compression, flexion, and internal tibial torque knee loads. The pivot landings were tested with the knee protected by normal and increased stiffness female quadriceps (Q) muscle springs. Error bars indicate 95% confidence intervals. *Significant stiffness difference ($p = 0.003$); #significant greater than baseline testing ($p < 0.01$)

($p = 0.875$), indicating that ACL integrity was unchanged by the high dynamic impulsive loads. Therefore, the observed changes in ACL strain were the results of changes in muscle stiffness.

Sample temporal behavior for a single specimen and trial for normal and increased quadriceps stiffness are shown in Figure 3. The loads applied to the knee and the resultant kinematics and muscles loads for the four trial blocks are shown in Table 1. The change in knee flexion angle and the knee abduction angle were correlated ($r = 0.67$) with the lesser quadriceps tensile stiffness spring, while both the change in knee flexion angle and knee abduction angle were reduced using the greater quadriceps tensile stiffness ($p < 0.001$ and $p < 0.001$, respectively). The greater quadriceps tensile stiffness did produce a small increase in anterior tibial translation over the lesser quadriceps spring ($p = 0.009$). However, this likely occurs due to the significantly greater dynamic quadriceps force with the greater quadriceps tensile stiffness ($p < 0.001$). Quadriceps force and anterior tibial translation were highly correlated ($r = 0.595$).

Experimentally, the increase in quadriceps tensile stiffness from the normal to the greater quadriceps stiffness resulted in a large reduction in knee flexion in each knee, and a corresponding reduction in peak AM-ACL relative strain (Fig. 4). The effect of quadriceps tensile stiffness on peak AM-ACL relative strain was further investigated with a dynamic biomechanical model, which confirmed the experimental results (Fig. 5). The in silico model demonstrated that an increase in quadriceps tensile stiffness reduced the peak change in knee flexion angle, as seen

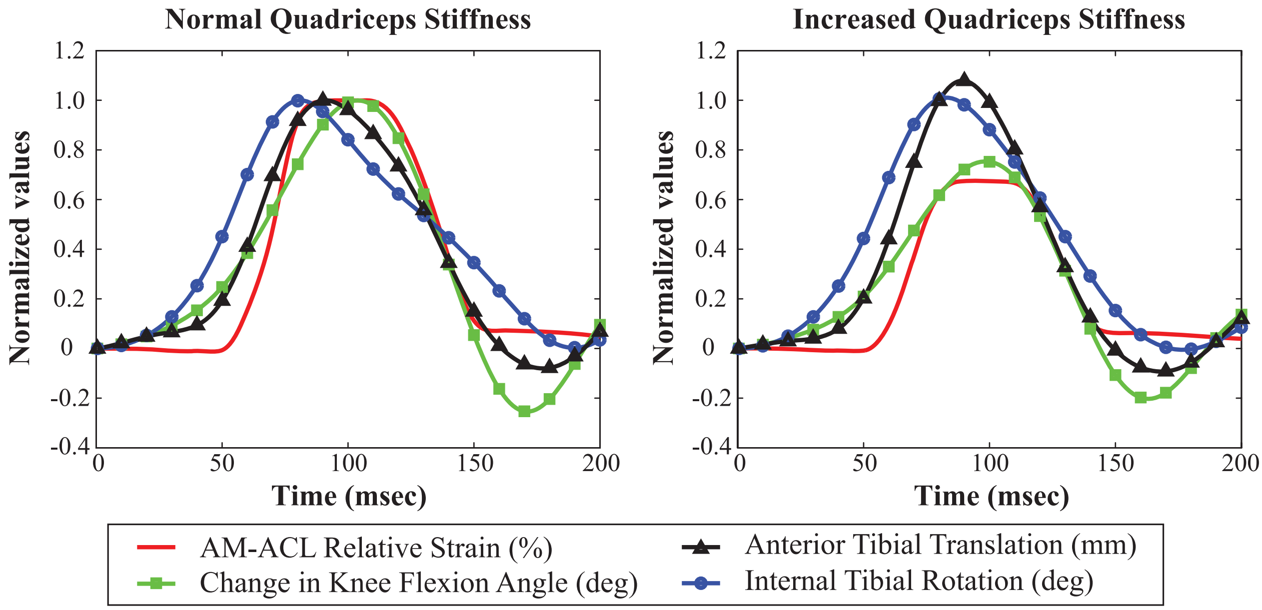

Figure 3. Sample temporal behavior of outcome measurements for normal and increased quadriceps stiffness: AM-ACL relative strain, change in knee flexion angle, anterior tibial translation, and internal tibial rotation (Specimen ID: F33434L). Measurements are normalized to the peak values using the normal quadriceps stiffness to ease comparisons (peak AM-ACL relative strain: 5.2%; peak change in knee flexion angle: 9.8°; peak anterior tibial translation: 5.2 mm; peak internal tibial rotation: 19.6°).

experimentally, thereby reducing the amount the femur rolled posteriorly on the tibial plateau.

DISCUSSION

By resisting rapid stretch, the quadriceps is the primary muscle resisting knee flexion during the first 100 ms of a jump landing.[31] This study demonstrates for the first time that increased quadriceps muscle resistance to rapid stretch significantly reduces peak AM-ACL strain during a 2*BW pivot landing in female knees. The in vitro experimental results showed a 33% increase in quadriceps tensile stiffness led to a 16% reduction in peak AM-ACL relative strain during a simulated pivot landing; the in silico model predicted a 19% reduction in peak AM-ACL relative strain with a 30% increase in quadriceps tensile stiffness. These reductions in peak AM-ACL relative strain were associated with a smaller increase in knee flexion and knee abduction during the simulated pivot landing. Hence, increasing the tensile stiffness of the quadriceps, either via muscle hypertrophy or by increased neural drive, may reduce peak ACL strains.

ACL injury prevention programs have shown limited success,[10] and even when they are successful, the

Table 1. Input Forces and Torques and Outcome Strains, Forces, and Kinematics Under Four Loading Conditions for 12 Female Knees (Q = Quadriceps)

	Non-Pivot Landing	Pivot Landing		Non-Pivot Landing
	Prebaseline	Normal Q Stiffness	Increased Q Stiffness	Postbaseline
Loading inputs				
Peak impact force (*N*)	1158 (117)	622 (102)	652 (90)	1143 (126)
Peak internal tibial torque (Nm)	1.8 (0.5)	20.0 (2.1)*	21.2 (2.2)	1.8 (0.4)
Primary outcomes				
Peak AM-ACL strain (%)	2.34 (1.22)	5.96 (2.92)*	4.99 (2.85)	2.29 (1.12)
Normalized peak AM-ACL Strain	0.99 (0.12)	2.66 (0.93)*	2.13 (0.75)	0.99 (0.10)
Secondary outcomes				
Quadriceps force (*N*)	766 (61)	924 (61)**	1023 (88)	759 (75)
Change in knee flexion (deg)	7.0 (1.1)	10.1 (1.7)**	8.0 (1.6)	6.8 (1.0)
Change in knee abduction (deg)	1.8 (1.0)	4.8 (1.7)**	3.4 (1.1)	1.7 (1.0)
Anterior tibial translation (mm)	2.5 (0.9)	4.9 (1.3)*	5.4 (1.3)	2.4 (0.9)
Internal tibial rotation (deg)	4.5 (1.0)	17.3 (2.3)	17.7 (2.2)	4.6 (1.2)

Note. Non-pivot landings were only performed with the normal female quadriceps tensile stiffness. *Significant stiffness effect $p < 0.05$. **Significant stiffness effect $p < 0.001$.

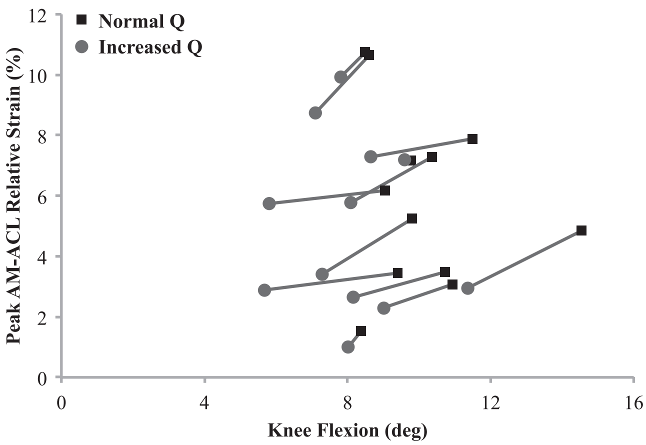

Figure 4. Scatterplot of peak AM-ACL relative versus peak change in knee flexion angle during the experimental studies of a simulated pivot landing. For each female knee, the effect of the normal and increased female quadriceps tensile stiffness is shown by the square and circle symbols, respectively, which are connected by a straight line for ease of identification.

mechanism by which they reduce ACL injury risk is unclear. A common component of ACL injury prevention programs involves knee extensor strength training.[5–9] Training will produce gains in muscle strength and muscle hypertrophy,[21,22] and these will increase muscle tensile stiffness,[50] principally through muscle fiber hypertrophy and therefore more muscle tissue being placed in parallel. For example, young females enrolled in a 16-week, 3 days/week progressive resistance training program developed a 38% increase in knee extensor strength.[20] The corresponding increase in muscle size was 6–8% in quadriceps femoris muscle volume during a 9-week, 3 days/week strength training program.[21,22] Similar gains in knee extensor

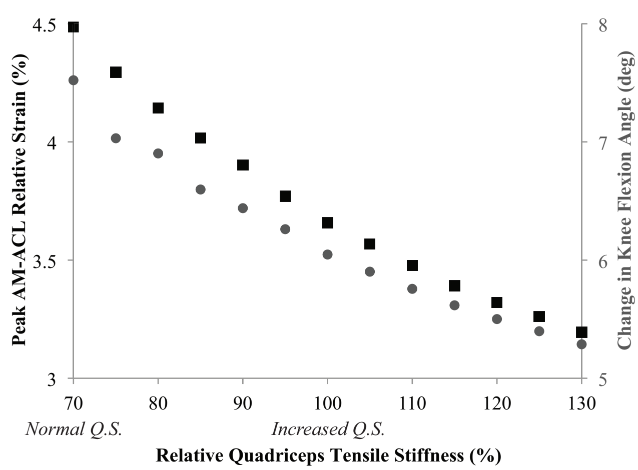

Figure 5. In silico results showing the predicted effect of varying quadriceps tensile stiffness on peak AM-ACL relative strain (black squares) and the peak change in knee flexion angle (gray circles) during a pivot landing. The normal and increased female quadriceps tensile stiffness values used during the experimental model are labeled as *Normal Q* and *Increased Q*, respectively.

strength (32–34%), muscle volume (6–8%), and 30–58% increase in vastus lateralis musculotendon stiffness were found in young males after a 12-week knee extensor strength training program.[15,23] It is also possible to achieve a 34% gain in musculotendon stiffness in 4 weeks by increasing muscle activation, rather than muscle strength.[32] We conclude that the 33% increase in female quadriceps tensile stiffness used in this study is achievable by in vivo resistance training, but it nears the maximum that can be expected with a 3-month training program.

The finding that peak AM-ACL relative strain increases with knee flexion may seem counterintuitive, since ACL strain under passive knee flexion[33] and active, weight-bearing squatting[34] decreases as the knee is flexed past 15°. However, these scenarios are fundamentally different from a high-impact jump landing, for which there is limited knowledge on the effect of dynamic knee flexion on in vivo ACL strain during a jump or pivot landing. Only one study has directly measured ACL strain during a jump landing and that was in a single individual,[35] while bi-plane radiographic measurements of ACL strain during a jump landing[36] assume the ACL is a straight line between the tibial origin and femoral insertion, ignoring the ACL's natural twist. The rolling-sliding behavior of the femoral condyle on the tibial plateau during knee flexion may help explain the increase in ACL strain with dynamic knee flexion (Fig. 6). The stiffer quadriceps spring exhibited greater anterior tibial translation despite lower AM-ACL relative strain; this may be an artifact of the greater in vitro quadriceps force. The anterior tibial translations and internal tibial rotations measured during our simulated 2*BW pivot landings (Table 1) were comparable to those seen during in vivo landings with bi-plane fluoroscopy.[37,38]

The in vitro (Fig. 4) and in silico (Fig. 5) results are consistent: the less-stiff quadriceps allowed a greater increase in dynamic knee flexion, leading to greater peak AM-ACL relative strain than when the quadriceps had greater stiffness. Dynamic knee flexion under a given load will cause the femoral condyles to roll posteriorly relative to the tibial plateau,[39–41] thereby loading the ACL and increasing ACL strain as pure femoral rolling occurs (Fig. 6), as seen in this current study. However, this pure rolling mechanism will eventually cause the initial femoral contact point to slide anteriorly relative to the initial tibial contact point, unloading the ACL at deeper flexion angles. Reducing quadriceps activation (i.e., less quadriceps tensile stiffness) will cause femoral rolling to dominate sliding in weight-bearing knees near full extension.[42] The sliding may occur as the knee flexes because the condyle has a smaller radius of curvature[43] or alternatively, due to changes in the tibial and meniscal geometry.

Strengths of this study include the use of a novel nonlinear quadriceps spring to investigate in vivo differences in muscle stiffness with an in vitro testing

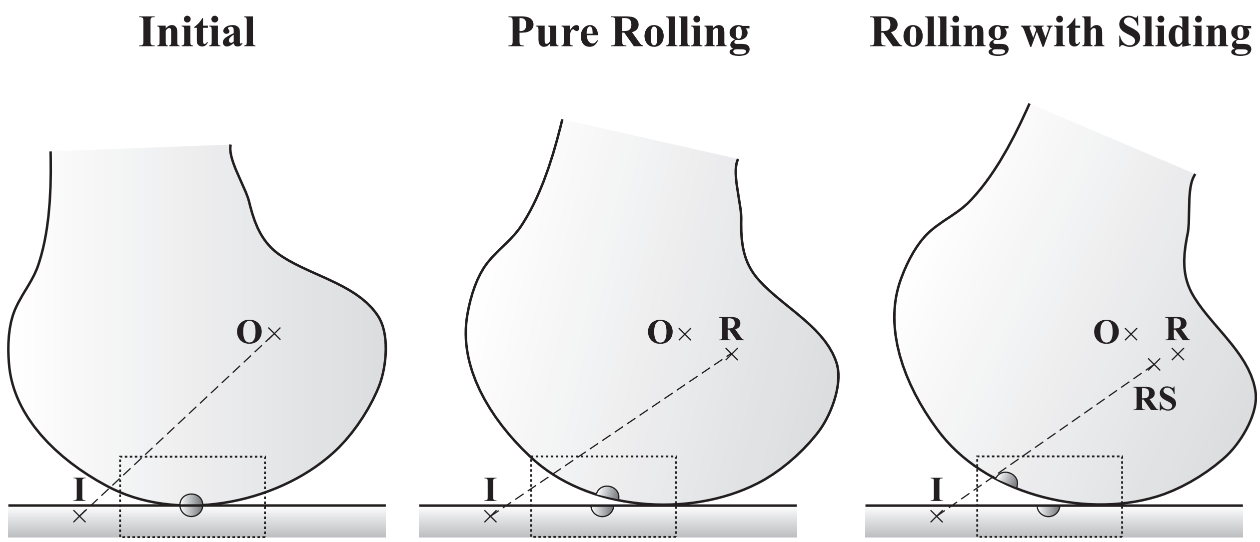

Figure 6. Schematic drawing (lateral view) showing the effect of femoral condyle rolling (top) on the tibial plateau (bottom). The ACL tibial insertion (I) and femoral origin (O) are marked with x symbols. (a) Initial tibiofemoral contact prior to the application of knee flexion. (b) To show pure rolling, the femur rolls 15° posteriorly down the tibial plateau, increasing ACL strain as indicated by the new ACL femoral origin location (R). (c) As the femur continues to roll, simultaneous sliding will develop, as shown by the initial femur contact point being located more anterior than the initial tibial contact point in the dotted box. This sliding mechanism will begin to unload the ACL, as shown by the new location of the ACL femoral origin (RS).

apparatus, a repeated measures design with randomized testing blocks, and the use of a dynamic biomechanical model to interpret in silico the in vitro testing results. This study is limited by the inability of cadaveric studies to model the full range of physiological loading conditions acting during an in vivo landing. While muscle hypertrophy can be modeled by increasing the spring stiffness of the quadriceps muscle equivalent, we did not alter the muscle line-of-action to reflect another feature of muscle hypertrophy; however, that could be done in future experiments. The current study investigated a narrow set of loading combinations in these cadaver knees in order to preserve their integrity. Therefore, the findings of this study may not apply to other loading combinations and initial knee flexion angles. While the peak quadriceps force developed in the current study is the right order of magnitude it is lower than previous studies that ruptured the ACL near full knee extension.[25,44] In addition, prior studies may have utilized quadriceps forces and/or axial compression forces that match or exceed physiological levels.[45–47] We only utilized the rectus femoris tendon for the quadriceps muscle equivalent; altering the stiffness of the other quadriceps muscles could affect frontal and axial plane movements. The apparatus can only represent the monoarticular actions of the major knee muscles and does not represent the biarticular actions of these muscles. Lastly, each knee used the same quadriceps stiffness values, rather than a cadaver-specific stiffness value scaled to physiologic muscle size.

While the current study has focused on modulation of quadriceps stiffness, the results are conservative because we have not accounted for the likely stiffening of the hamstrings during in vivo training. Although we did not explore the effect of hamstring stiffening in the current study, we have shown in prior work that lengthening the hamstrings will have an even greater effect on reducing ACL strain than seen in the current study.[48] Based on our results, there may be a nonlinear relationship between quadriceps stiffness and AM-ACL relative strain, but these findings only pertain to the set of stiffness values studied here. It is a limitation that ACL strain was only measured in the anteromedial bundle to prevent the femoral notch impinging on the DVRT had it been placed on the posterolateral bundle. The DVRT only measures relative strain, which likely underestimates the absolute strain on the ligament due to the pretensioned muscle forces. However, AM-ACL relative strain is highly correlated with overall ACL force.[49] Finally, it is a weakness that the specimens utilized in this experiment were older than the adolescents and young adult populations at highest risk for ACL injuries. Limitations of the dynamic biomechanical model are explained in the Supplemental Text 1.

We conclude that the knee extensor strength training present in many ACL injury prevention programs may be beneficial by reducing peak ACL strain during a dynamic maneuver. An increase in quadriceps tensile stiffness of 33%, which can be achieved in vivo with muscle hypertrophy and/or increased muscle activation, can reduce peak AM-ACL relative strain by 16% during a 2^*BW simulated pivot landing.

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