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External Loads Associated with Anterior Cruciate Ligament Injuries Increase the Correlation Between Tibial Slope and Ligament Strain During *In Vitro* Simulations of *In Vivo* Landings

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Abstract

Background: The aim of the present study was to evaluate the relationship between tibial slope angle and ligament strain during *in vitro* landing simulations that induce ACL failure through the application of variable external loading at the knee. The hypothesis tested was that steeper posterior tibial slope angle would be associated with higher ACL strain during a simulated landing task across all external loading conditions.

Methods: Kinetics previously derived from an *in vivo* cohort performing drop landings were reproduced on 45 cadaveric knees via the mechanical impact simulator. MRIs were taken of each specimen and used to calculate medial compartment posterior tibial slope, lateral compartment posterior tibial slope, and coronal plane tibial slope. Linear regression analyses were performed between these angles and ACL strain to determine whether tibial slope was a predictive factor for ACL strain.

Findings: Medial and lateral posterior tibial slope were predictive factors for ACL strain during some landings with higher combined loads. Medial posterior slope was more predictive of ACL strain in most landings for male specimens, while lateral posterior and coronal slope were more predictive in female specimens, but primarily when high abduction moments were applied.

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Interpretation: Tibial slope has the potential to influence ACL strain during landing, especially when large abduction moments are present at the knee. Deleterious external loads to the ACL increase the correlation between tibial slope and ACL strain, which indicates that tibial slope angles are an additive factor for athletes apt to generate large out-of-plane knee moments during landing tasks.

Keywords

Anterior cruciate ligament; tibial slope; knee biomechanics; injury risk factors; landing

1.0 INTRODUCTION

Anterior cruciate ligament (ACL) tears are one of the most common injuries in sports that involve jumping, landing, pivoting and cutting, such as basketball, football, and soccer.^{13, 50} More than 125,000 ACL reconstructions are performed annually in the United States.³⁵ ACL injury can result in the loss of an athletic season, the loss of scholarship funding, and is associated with a high risk of developing osteoarthritis.^{28, 39, 40} In addition, ACL injuries incur a significant financial cost to repair and rehabilitate, with the approximate cost of an ACL reconstruction and rehabilitation in excess of \$17,000.²⁷ However, despite this cost, ACL reconstruction fails to fully restore native knee mechanics³ and following reconstruction only 81% of players return to sports, of which 65% return to their preinjury level and 55% return to competitive sport.^{50, 62} With these modest outcomes, it is important to consider the factors that lead to ACL injuries and work towards injury prevention.

Research has identified factors associated with increased risk of ACL injury.^{13, 29} Identified risk factors include extrinsic variables such as athletic task performed, visual perturbations, bracing, physical contact, and shoe-surface interaction as well as intrinsic variables, including hormonal response, neuromuscular control, and anatomical geometry.²⁸ Studies that investigate anatomical risk factors tend to focus on geometric measurements such as lower extremity bone lengths, femoral notch width, and tibial slope angle.^{28, 67} Greater medial compartment posterior tibial slope angle (MPTS) and lateral compartment posterior tibial slope angle (LPTS) have been associated with increased ACL injury risk,^{60, 61} and graft failure following ACL reconstruction.¹⁵ Steeper tibial slope angles can increase anterior tibial translation during direct compressive loading of the knee as the loading forces drive the femur down the tibial slope and shift the tibiofemoral contact locations in a posterior direction on the tibial plateau.^{16, 21} As the ACL is the primary passive restraint against anterior tibial translation,^{14, 17, 24} increased translation has been shown to subsequently induce greater mechanical demand on the ACL throughout flexion.^{14, 49, 65, 66} Thus, steeper tibial slope angle has been hypothesized to increase ACL strain during athletic task performance in response to the increased compressive loads and anterior tibial translation relative to gait.⁶⁴ A previous study reported that a 10° increase in tibial slope angle resulted in 6 mm of increased anterior tibial translation.⁶¹ Therefore, individuals with steeper tibial slopes may be predisposed to greater ACL strain and, thus, would be more vulnerable to injury.

In vivo experiments have previously investigated tibial slope in ACL-injured versus control populations.^{12, 59} However, these experiments were limited to association of tibial slope with injury incidence, as ligament strain could not be directly assessed *in vivo*. Accordingly, there are disagreements pertaining to the magnitude of influence that tibial slope angle exhibits in relation to ACL injury risk.⁶⁴ Alternatively, *in vitro* models can reliably reproduce joint kinetics and kinematics with respect to *in vivo* recorded athletic tasks.^{4, 9} Moreover, *in vivo* loads can now be modeled using *in vitro* testing configurations to simulate ligament injury events in a clinically representative presentation.^{9, 10} Robotic-controlled, kinematic-constrained simulations of jumping, landing, and cutting activities on cadaveric knees revealed no correlation between tibial slope angle and ACL strain.⁵ However, these simulations lacked the variability of motion²⁰ and impulsive forces associated with *in vivo* landing and were not designed to incur damage to the joint structures.^{4, 20, 53} Therefore, it is possible that in a more dynamic simulation environment representative of an injury scenario, tibial slope may exhibit greater influence on ACL strain. The impulsive load delivery of the mechanical impact simulator enacts greater perturbation and excursion on a cadaveric knee in a more rapid timeframe than does the robotic manipulation of a tibia around a femur.^{4, 9} Additionally, when athletes land from a drop vertical jump *in vivo* they are typically classified into three injury-risk categories (low, moderate, high) based on their biomechanics.² Those athletes who exceed the high-risk classification threshold exhibit poor neuromuscular control and are most likely to experience non-contact ACL injury. The mechanical impact simulator is kinetically driven and delivers external knee abduction torques to each specimen across a range of magnitudes that is inclusive of individuals who expressed the highest potential injury risk in an *in vivo* cohort.⁹ Previous robotic simulations of landing were kinematically-driven, and degrees of rotation do not universally represent a kinetic output across all specimens.⁴ The mean peak knee abduction moment in robotic simulations achieved the threshold of high-risk classification for non-injurious *in vivo* landings.^{2, 29, 43} Impact simulations well exceed this value and consistently produce clinical ACL failures in cadaveric specimens.¹⁰ Accordingly, the larger, faster loading environment of the mechanical impact simulator may induce greater excursion of the femoral condyles than in robotic simulations. This excursion would be partially constrained by the plane of the tibial slope and potentially allow subtle changes in tibial slope angle to effect greater influence on ACL strain.

The aim of the present study was to evaluate the relationship between tibial slope angle and ligament strain measured during *in vitro* landing simulations that induce ACL failure through the application of variable external loading at the knee. Direct quantification of this relationship between tibial slope and ACL strain during landing tasks could provide justification in favor of or against the use of high tibial osteotomies in ACL reconstruction cases where the patient exhibits excessive tibial slope angle. Alternatively, this quantification could potentially lead practitioners to alter return to sport recommendations in patients with excessive tibial slope angles. The hypothesis tested was that steeper tibial slope angle would account for enough variance in ACL ligament strain during a simulated landing tasks to be included as a predictive factor in ACL injury risk assessments.

2.0. METHODS

2.1. Overview of Experimental Design

Forty-five lower extremity cadaveric specimens were obtained from an anatomical donations program (Anatomy Gifts Registry, Hanover, MD). Inclusion criteria were that the specimen was between 14 and 50 years of age and those with previous history of knee trauma, bone cancer or knee surgery were excluded.⁹ Six specimens were excluded from analysis due to failure induced by the preparation process, substandard stiffness of bony tissue, and inconsistency in testing execution, which left 39 specimens for analysis (19 males, 20 females, age = 41.5 (8.4) years, mass = 85.7 (25.6) kg, height = 173 (11) cm, BMI = 28 (8)).

Magnetic resonance images (MRI) were obtained for each specimen prior to mechanical impact testing. After MRI acquisition, a novel mechanical simulator was used to simulate the impulse ground reaction force generated during an *in vivo* drop vertical jump (DVJ) landing.⁹ External knee abduction moment (KAM), anterior tibial shear force (ATS), and internal tibial rotation moments (ITR) loads were applied to the specimen via pneumatic actuators. These external loads were selected because they are known to add strain to the ACL and potentially induce damage under the appropriate simulation conditions.^{3, 38, 45} Strain gauges were implanted on the ACL to record ligament strains.^{9, 11, 19, 38, 45} MPTS, LPTS, and coronal plane tibial slope angle (CPTS) were calculated for each specimen using their MRI images. These tibial slope measurements were then evaluated for reliability using interclass correlation coefficients (ICC). Pearson correlations coefficients and corresponding r-squared values were then used to assess relationships between tibial slope angle and ligament strain.

2.2. Tibial slope

Magnetic resonance imaging (MRI) images were captured of the tibiofemoral joint for each specimen. Images were captured on a 3.0T MRI (Discovery MR750w, General Electric, Boston, MA, USA) across a 254 mm length centered around the patella. Tibial slope calculations were made from MRI images collected from each specimen and followed the protocol established by Hashemi, et al.²² Two novice researchers were trained by a senior investigator, then independently followed the published protocol to measure MPTS, LPTS, and CPTS three separate times for each specimen. Measurement iterations were performed no less than four days apart in RadiAnt DICOM Viewer (Medixant, Poznan, Poland). Using built-in software tools to define the anatomical landmarks designated by Hashemi, et al., MPTS and LPTS were calculated as the angle between a line perpendicular to the diaphysis axis and a secondary line that passed through the anterior and posterior osseous peaks of tibial plateau (Fig. 1).²² CPTS was calculated as the angle between a line perpendicular to the tibial diaphysis axis and a secondary line that passed through the medial and lateral osseous peaks of the tibial plateau.²²

2.3. Specimen Preparation

The specimen preparation techniques used in this investigation were previously described in the literature where impact testing was performed.⁹ Briefly, each specimen was kept frozen at -20 °C until 24 h prior to use. The thigh skin was removed and all the thigh muscles were

individually isolated. Muscle tissue was then rasped away leaving the tendons intact. Each tendon was organized into one of three different cable clamps (medial hamstrings, lateral hamstrings, and quadriceps) and secured with a U-bolt. The femur was then resected 20 cm proximal to the superior aspect of the patella and the distal end was potted with fiberglass resin. Carriage bolts were placed through pre-drilled holes in the tibia and used to mount a custom mechanical fixture to the specimen. Two differential variable strain transducers (DVRT, LORD MicroStrain, Inc., Williston, VT) were implanted on the distal third of the anteromedial bundle of the ACL.^{9, 11, 19, 38, 45}

Once prepared, the specimen was inverted and secured in the testing simulator. The femur was fixed at 25° flexion with respect to the vertically oriented tibia. A 25° flexion angle was selected as this represents the average knee flexion angle achieved by high school athletes when landing from a 31 cm drop.¹ Cable clamps were affixed to the hamstrings and quadriceps tendon bundles and attached to pneumatic actuators (CG5LN40SV-100 and CG5LN50SV-100, SMC Corporation, Tokyo, Japan) that maintain a constant 1:1 force ratio between the quadriceps and hamstrings muscles. Five additional pneumatic actuators (MQQTL40TN-100DM and CG5LN80SV-100, SMC Corporation, Tokyo, Japan) attached to designated points on the custom tibial fixture and allowed for the application of KAM, ATS, and ITR to the knee (Fig. 2).^{9, 10} Specifically, two actuators were mounted perpendicular to the long axis of the tibia in the sagittal plane and positioned anterior to the specimen on the medial and lateral sides of the knee. The medial and lateral cylinder were equidistant from the center of the knee in the frontal plane. The carbon fiber rope that extended from these actuators affixed to posts that extruded medially and lateral from the custom tibial frame perpendicular to the long axis of the tibia. By applying equal loads to the tibial frame both medial and lateral to the knee, these actuators generated ATS on the specimen. Two additional pneumatic actuators were mounted anterior-medial and posterior-lateral with respect to the knee. The carbon fiber rope from these actuators attached to the same posts on the tibial fixture that were used to generate ATS. As they were mounted in opposite orientations, when these two actuators retracted with equivalent force, they applied ITR to the joint. Finally, two pneumatic actuators were mounted perpendicular to the long axis of the tibia in the frontal plane on both the medial and lateral aspects of the simulator. Using a bar-linkage with pulleys on either end, these actuators were able to simultaneously route carbon fiber rope to rotary mounts positioned on the anterior and posterior aspects of the custom tibial fixture. As with the ITR actuators, these pneumatics were arranged in opposite orientations, such that their equally applied force of contraction would work in a complimentary fashion to generate KAM at the knee. Each of these externally-applied, pneumatically-actuated loads were initiated at the start of each trial and achieved peak magnitude by 1.0 sec, which preceded the timing of impulse delivery in each simulation. All external loads were relaxed 3.0 sec after they were initiated.

2.4. Impact Simulations

Simulations were repeated on each specimen through a randomized protocol of impacts with a constant impact weight (34 kg) and drop height (31 cm), but varied magnitudes of KAM, ATS and ITR. This was referred to as sub-threshold testing as the external loading magnitudes were determined relative to previous *in vivo* motion analysis data collected on a

cohort of 44 young athletes (age = 23.3 (4.1) years; mass = 72.6 (13.9) kg; height = 172 (10) cm) performing DVJ tasks. Inverse dynamics were calculated from these motion data using an established biomechanical model in Visual3D (version 5, C-motion Inc., Germantown, MD, USA).⁹ The output joint kinematics were then categorized into groups. The magnitude of KAM, ATS, and ITR represented by the 0th percentile rank-order of this *in vivo* cohort was selected as the baseline load for simulation trials as these values represented the minimum physiologic kinetics expressed by the *in vivo* cohort. The remainder of the *in vivo* kinetics were separated to represent low-risk, moderate-risk, and high-risk landing biomechanics.^{2, 44} The KAM, ATS, and ITR loading magnitudes were selected from between the 0th and 200th percentile of this *in vivo* cohort (Table 1) and applied to the impactor simulations in a block-randomized order, where the conditions with the highest KAM, ATS, and ITR loads were introduced after all other simulations were completed.^{9, 10} Previous literature on the mechanical impact simulator has indicated that the externally applied KAM, ATS, and ITR loads were separated into tertiles based on joint load estimates calculated from the aforementioned *in vivo* cohort.^{9, 10, 42, 47, 48} However, data from uniaxial load cells (MLP-300, MLP-1K, & SWO-2K, Transducer Techniques, LLC, Temecula, CA, USA) mounted on each pneumatic piston revealed that these estimated inputs were realized in the magnitudes indicated in Table 1. Both the anticipated-loading and realized-loading stratifications represent a diversity of low, moderate, and high-risk landing simulations enacted on the cadaveric specimens.⁴³ Testing was terminated once the specimen suffered a hard or soft tissue disruption of any kind on the macroscopic level.⁹ Before and after the impact simulations, each specimen was clinically and arthroscopically evaluated, via an arthroscopic lens (Stryker Corporation, Kalamazoo, MI, USA), by a board certified orthopedic surgeon to guarantee the integrity of and consequential damage to each structure within the knee. Twenty-one (21) out of 39 tested specimens experienced failure during sub-threshold testing. Female specimens were especially predisposed to experience failure prior to the completion of sub-threshold testing. As the trial in which a specimen will fail is not predictable, the number of specimens that survived each simulation condition are indicated in Table 2. Throughout the impact simulations the DVRT implanted in the ACL recorded ligament strain. Ligamentous structural damage was identified when a significant change in or complete absence of elastic behavior was recorded by these sensors.⁹

2.5. Data analysis

Intra-rater (ICC (2, k)) and inter-rater (ICC (2, 1)) correlations coefficients were calculated in order to determine within and between rater reliability of the tibial slope measurement protocol established by Hashemi, et al.²² ICC Classifications were defined as: ICC < 0.4 poor, 0.4 < ICC < 0.75 fair-to-good, and ICC > 0.75 excellent.¹⁸ Student's t-tests were used to examine statistical significance in tibial slope angle between compartments with a significance threshold of $\alpha < 0.05$. Pearson correlations were used to determine associations between tibial slope angle and measurements of ligament strain within each simulated loading condition. An R-squared threshold of 0.2 was used to determine whether each tibial slope variable accounted for enough of the variance in ligament strain to be incorporated as a component in multivariate regression models related to injury risk.^{6, 29, 31} This threshold has been established in injury prevention as a linear regression cutoff criterion in predictive models²⁹ and adheres to the premise that, in any predictive model, a

maximum of five variables will contribute to risk factor analysis in a meaningful manner.³¹ Statistical analyses were performed in MATLAB (version 2016a, The MathWorks, Inc., Natick, MA, USA) using built in ICC and corr functions. Correlations were calculated by both raters, and the mean of those values were presented in the results. Correlation between tibial slope and ligament strain was separately evaluated in three formats: slope vs. absolute peak strain, slope vs. change in strain from the point of initial contact (IC), and slope vs. change in strain from baseline. For each of these relationships, the linear-regression slope was also calculated. Peak absolute strain represents a measure of how close each specimen came to achieving the average failure strain for ACLs (15.3 (8.7)%)¹⁰ during each simulation. However, there is large inter-specimen variability in ACL strain to failure;³⁸ thus, normalization of strain to the change that occurred throughout simulation can provide more consistent values for comparison. Strain was normalized from IC because this represents change induced by ground contact and ACL injuries are projected to occur within the first 67 msec after IC in response to the ground reaction forces propagating through the lower extremity.^{30, 36, 37} Strain was further normalized from baseline because this will evaluate the change that was induced on the ligaments both from the applied external loading and ground impulse. As external loads during landing are known to stress the ACL,^{3, 38, 45} are a modifiable factor that can be reduced through interventional neuromuscular training,^{25, 26, 56, 57} and predispose athletes to injury,²⁹ it is important to consider the relationship they have with ACL strain in addition to ground contact response. IC was defined as the time point when the vertical impact force imparted on the specimen exceeded 25 N, while baseline was defined as the condition where the specimen was at rest with no external KAM, ATS, ITR, or impulse force being enacted. Correlations were calculated individually for each iteration of tibial slope measurements performed by both raters. Within-rater and between-rater averages were then calculated from these individual trial calculations. Comparisons were not made between simulation conditions as the number of specimens that completed each condition were not consistent.

3.0. RESULTS

3.1. Tibial Slope Reliability

Intra-tester reliability in tibial slope calculation was excellent for each compartment and across the coronal plane (Table 3). Inter-tester reliability was excellent for the MPTS, fair-to-good for the LPTS, and poor for the CPTS.

3.2. Tibial Slope Correlations

Across the subject population, MPTS was 6.5 (3.4)°, LPTS was 6.6 (2.6)°, and CPTS was 1.4 (1.7)° (Fig. 3). There was no difference between MPTS and LPTS ($P = 0.936$). However, the CPTS was significantly more shallow than both the MPTS ($P < 0.001$) and LPTS ($P < 0.001$). Across the entire study cohort, MPTS, LPTS, and CPTS did not account for sufficient variance to be a significant factor for modeling peak ACL strain prior to failure (Table 4). However, within some individual loading conditions, both MPTS and LPTS accounted for sufficient variance in peak ACL strain, change in ACL strain from IC, and change in ACL strain from baseline to be considered a predictive factor (Table 5). R-squared values of 0.2 were most prevalent for simulations where the KAM magnitude was at the

200th percentile. CPTS was not a predictive factor for any measure of ACL strain. For most all simulated loading conditions, the regression slope between ACL strain and posterior tibial slope angle was less than 1% per 1 ° of increased tibial slope steepness.

3.3. Sex Differences

There were no differences between female and male specimens in MPTS angle ($P = 0.339$; Table 6), LPTS angle ($P = 0.107$), or CPTS angle ($P = 0.119$), but females did express greater variation than males. For the female specimens, MPTS was a predictive factor for peak ACL strain primarily when KAM was at the 99th percentile and ATS was at the 90th percentile (Table 7). LPTS was frequently identified as a predictive factor of both peak ACL strain and change in ACL strain for conditions where two or more external loads met the following parameters: KAM 68th percentile, ITR = 100th percentile, ATS = 90th percentile. CPTS was above the predictive threshold for peak ACL strain during a few trials with 90th percentile ATS loading, and for change in ACL loading during trials with 200th percentile KAM loading or during trials when either 90th percentile ATS or ITR were combined with any magnitude of KAM greater than the 2nd percentile.

For male specimens, the MPTS was a predictive factor for change in ACL strain for the majority of simulated impact conditions (Table 8). Relative to peak ACL strain, MPTS only occurred as a predictive factor when all three external loads were applied with at least one load stimulus applied at or above the 90th percentile. LPTS was a predictive factor for change in ACL strain in male specimens in select simulations where KAM was at the 200th percentile. CPTS did not demonstrate correlation with ACL strain in the male specimens. In male specimens, there were a greater number of impact conditions where MPTS was predictive than females, while LPTS and CPTS were above the threshold of prediction more frequently in female specimens. In male specimens, predictive correlations occurred most frequently in simulations with the highest applied KAM exhibited the (21 out of 54 comparisons, 39%) as compared to the highest applied ATS (21 out of 81 comparisons, 26%) and the highest applied ITR (15 out of 54 comparisons, 28%). In female specimens, predictive correlations occurred most frequently in simulations with the highest applied KAM (37 out of 54 comparisons, 68%), then the highest applied ATS (49 out of 81 comparisons, 60%), then the highest applied ITR (28 out of 54 comparisons, 52%).

Female specimens were much less likely to survive the full sub-threshold simulation protocol as 16 of 20 (80%) of female specimens failed prior to protocol completion. For male specimens only 4 of 19 (21%) failed prior to sub-threshold protocol completion.

4.0. DISCUSSION

The objective of the present study was to evaluate the associations between tibial slope and ligament strain in the ACL during *in vitro* simulations of ACL failure. The tested hypothesis that steeper tibial slope angle would correlate with ACL ligament strain was partially supported. Across the whole specimen cohort, there was no significant correlation between tibial slope and peak ACL strain prior to failure. However, when broken down by individual simulation conditions, the MPTS angle did account for enough variance ($R^2 > 0.2$) to be considered an important factor^{29, 31} for modeling peak ACL strain and change in ACL

strain, but primarily during simulations that incorporated high KAM loading. This MPTS influence was much more readily noted in male than female specimens; therefore, it is likely that male specimens were driving the significance of MPTS during high KAM simulations for the whole cohort. Similarly, LPTS angle was predictive of change in ACL strain during some simulated impacts that incorporated high KAM, but was not as consistent as the MPTS. Tibial slope in both compartments has been associated with increased risk of ACL injury for *in vivo* populations.^{60, 64} In addition, greater MPTS and LPTS angles during a simulated controlled athletic task has been correlated with increased peak KAM⁶ which is a known risk factor for ACL injury.²⁹ However, limited magnitudes of isolated rotational stimuli applied to a knee joint have failed to produce statistically significant changes in ACL strain as compared to the baseline knee orientation.⁸ Accordingly, the present findings that MPTS and LPTS were potentially predictive factors for ACL strain, but only in simulated conditions with high KAM loading, corroborated the existing literature.

The magnitude of external forces and moments enacted on specimens in the present study did influence the how well tibial slope correlated with ACL strain and change in ACL strain. The largest R-squared values reported presently were typically found when the combinatorial loads of KAM, ATS, and ITR were at or above the 90th percentile recorded from the *in vivo*. These three external loading mechanisms couple together straining the ACL as ITR predisposes valgus orientation through slight superior and inferior translation of the medial and lateral femoral condyles, respectively,³⁴ which is compounded by KAM that effects greater change in strain on the ACL than any other isolated knee rotation,⁸ and is further exacerbated by anterior tibial translation introduced from ATS. Individually, very high KAM (200th percentile) demonstrated the most consistent association with tibial slope r-squared values being above the predictive threshold of 0.2. This was expected as KAM is the only loading factor to be prospectively directly associated with increased risk of ACL injury within a cohort of *in vivo* athletes.²⁹ In addition, as the ACL accounts for 85% of the passive restraint of anterior tibial translation at the knee,¹⁴ it was unsurprising that high ATS (90th percentile) load demonstrated the next most consistent correlation between tibial slope and ACL strain. The greater prevalence of correlations observed during high-load simulations in this investigation indicated that tibial slope angles may significantly contribute to the magnitude of ligament strain achieved during the performance of highly dynamic athletic tasks, as these tasks generate large external knee loads outside of the sagittal plane (such as KAM, ATS, and ITR) in athletes who exhibit poor neuromuscular control.²⁶

The absence of predictive association between MPTS and LPTS and peak ACL strain in the trial prior to failure was unsurprising. If such a correlation were observed, then it would have indicated that tibial slope somehow altered the ultimate failure strain of the ligament, which is a material property inherent to the tissue. Thus, the presence of such a correlation would have indicated that tibial slope angle influences or determines the mechanical properties of the ACL. On average, the ACL ultimate failure strain is expected to occur when the ligament has been stretched 15-20% beyond its neutral loading length^{9, 10, 14, 38} and has not previously been correlated with tibial slope magnitude. Analysis of each individual simulation condition that occurred prior to specimen failure allowed the present investigation to address how tibial slope correlated ligament strain behavior during landing

and not ligament failure mechanics. The regression slope between ACL strain and tibial slope angle was less consistent in female specimens than in male specimens and the overall population cohort. This may be due to the limited number of female specimens that completed the higher load simulations prior to failure, as this may have left an insufficient number of congruent data points to derive a curve representative of the whole cohort in these cases.

The absence of correlation of CPTS with ACL strain across the specimen population is supportive of previous literature that identified no increased risk of ACL injury relative to CPTS.⁶⁰ However, regression models created from MRI tibial slope measurements coupled with three-dimensional motion analysis have predicted that in females during landing a 1° increase in CPTS correlates with an increase of 1.8° in peak hip adduction, 1.2° in peak knee abduction, and 2.2° in peak ITR.⁵¹ These predicted alterations to landing mechanics add to the justification of why CPTS correlated with change in ACL strain when the present specimens were parsed out by sex.¹⁵

The present findings indicate that LPTS was more significant to ACL strain biomechanics in the female knee than MPTS or in the male knee. This finding corroborates previous data that found lateral tibial slope, along with femoral notch width, were the anatomical factors most closely associated with ACL injury risk and graft failure in female athletes, but not male athletes.^{15, 55} ACL injury incidence is a sex-specific event with female athletes being up to 7 times more likely to experience rupture than their male counterparts.⁵² As such, it should be expected that there is potential for male and female limbs to vary in their response to identical mechanical stimuli due to geometrical differences.⁴⁷ Previously, an absence of differences in intra-articular mechanics at the knee have been shown between male and female specimens simulated through athletic task kinematics.⁷ However, these prior simulations were conducted in a controlled, non-injury environment that also removed the impulse force delivery associated with ground impact; whereas, the current simulations were driven by impact and designed to recreate injury events.^{4, 9} Accordingly, sex-based differences may be more readily identified during injury-event simulations, but further work is necessary to validate this hypothesis.

In conjunction with previous literature, the present study found that tibial slope in the male knee is associated with ACL biomechanics.⁴⁶ However, the present study identified that increased MPTS correlated more significantly with increased ACL strain than LPTS; whereas, LPTS was previously identified as the sole independent predictive factor for ACL injury risk in male football athletes.⁴⁶ A second study found no correlation between MPTS or LPTS and ACL injury risk in male athletes.⁵⁵ These discrepancies may exist because the present study directly quantified mechanical behavior of the ligament within the knee, while the previous investigations used a dichotomous indication of ACL injury events to draw their associations. The quantified mechanical response of the ACL relative to an *in vivo* population remains unknown.

The presence of sex differences in geometry for MPTS, LPTS, and CPTS varies throughout literature.^{60, 63} In the present study sex differences were not observed in MPTS, LPTS, or CPTS, which agrees with a recent literature review of tibial slope angles.⁶⁰ Previous

investigation between the MPTS and LPTS revealed poor correlations and significant within-subject differences.²² Therefore, in order to fully characterize the tibial plateau, it is important to utilize slope measurements from more than one compartment. Further, understanding differences in tibial slope between compartments is important for surgical planning in order to identify the optimal location of the knee joint hinge as well as to understand how changes in tibial slope contribute to tibiofemoral arthrokinematics⁴¹

As with all simulations, this investigation faced limitations. Intra-rater reliability was in accordance with previous literature as each rater exhibited excellent ICC scores.¹⁷ Inter-rater reliability for the LPTS and CPTS was below excellent, and thus, differed from the literature where inter-rater reliability for LPTS and MPTS measurements ranged from 0.78-0.91. The reduced reliability in the present study may be related to the MRI images collected. Specimens were only imaged from 20 cm below the tibiofemoral joint line to 20 cm above the tibiofemoral joint line. This limited window may have affected the rater ability to accurately align the diaphysis of the tibia. Additionally, not all specimens completed all simulated impact conditions as testing was aborted immediately once hard or soft tissue damage occurred within each specimen.^{9, 10} Accordingly, values for each condition were averaged from each specimen that successfully complete that particular simulation as indicated in Table 2. In the case of female specimens, this left some simulation conditions under-powered for statistical analysis as only 6 or fewer female specimens completed each high-load simulation. Further, the externally-applied joint loads were derived from a mixed-sex *in vivo* cohort of athletes. As such, the higher loads may have exceeded the physiologic range of some smaller specimens. While joint motion along the articular geometry would be expected, it is plausible that very high loads could have disarticulated the knee against the natural tibiofemoral geometry in smaller or less-stiff specimens. This limited data likely resulted in the exaggerated and inverse-relationship regression slopes observed between ACL strain and tibial slope angle in some high-load female simulations for the MPTS and LPTS.

One of the exclusionary criteria for specimens in the current study was that they exhibited no prior history of ACL injury.^{9, 10} Literature has previously demonstrated that patients who experience ACL failure exhibit larger magnitude of posterior tibial slope than healthy controls.^{60, 64} Therefore, it is possible that the exclusionary criteria could have inadvertently reduced the variance and magnitude of tibial slope angle observed within the present population as ACL-rupture specimens with potentially steeper tibial slope angles would have been rejected. In studies where tibial slope was measured via MRI, mean MPTS ranged from 4.1-9.5° in healthy controls and 4.7-10.6° in ACL injured subjects.^{12, 23, 32, 33, 54, 58, 64} Similarly, mean LPTS ranged from 2.6-9.0° in controls and 4.6-11.5° in ACL injured subjects. The injured subjects had a steeper MPTS by 0.7 (1.1)° and steeper LPTS by 2.0 (1.0)°. The mean MPTS of 6.5° and LPTS of 6.6° in the present study fall within the midrange of healthy controls. Therefore, inclusion of ACL injured specimens may have increased the overall steepness of tibial slope in the specimen cohort.

5.0. CONCLUSIONS

During a simulated drop landing, tibial slope angle can influence ACL strain, but the magnitude of that influence is likely dependent on the external loading that occurs about the knee. MPTS and LPTS exhibited the greatest sex-specific correlations with change in ACL strain during a landing and that association was greatest during simulations with high KAM loading. Under the various simulation conditions tested, MPTS in male specimens exhibited a high prevalence of predictive correlations with ACL strain and ACL change in strain, while LPTS and CPTS in female specimens exhibited a higher prevalence of predictive correlations with ACL strain and ACL change in strain. Deleterious external loads to the ACL increase the correlation between tibial slope and ACL strain indicates that tibial slope angles are an additive risk factor for athletes with poor neuromuscular control who are apt to generate large out-of-plane knee moments during landing tasks.

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ABBREVIATIONS

ACL	Anterior cruciate ligament
ATS	Anterior tibial shear
CPTS	Coronal plane tibial slope
DVJ	Drop vertical jump
IC	Initial contact
ICC	Interclass correlation coefficient
ITR	Internal tibial rotation
KAM	Knee abduction moment
LPTS	Lateral compartment posterior tibial slope
MPTS	Medial compartment posterior tibial slope
MRI	Magnetic resonance image

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HIGHLIGHTS

- Tibial slope angle is an accepted risk factor for ACL injury risk.
- This study directly quantifies ACL strain relative to tibial slope during landing. The influence of tibial slope angle on ACL strain was sex-specific.
- Medial plateau tibial slope angle had greater influence on strain in males. Lateral plateau tibial slope angle had greater influence on strain in females.

Fig. 1:
Anatomical landmarks used for tibial slope measurements. a) MPTS, b) LPTS, and c) CPTS.

Fig. 2:
(A) Meta-view of custom designed mechanical impact simulator for creation of ACL ruptures, (B) cable pulley system used to deliver pneumatically actuated loads to the quadriceps and hamstrings tendons, (C) external fixation frame attached to the tibia and used to deliver pneumatically actuated KAM, ATS, and ITR loads to each specimen. This figure has been reproduced from Bates, et al. 2018. *Am J Sports Med.*^{10, 34}

Fig. 3:
Box plot of the mean tibial slope angles across all specimens for the MPTS, LPTS, and CPTS. + indicates outlier of the data. * indicates significant difference between slope angles.

Table 1:

External loading parameters for pneumatic actuators in the mechanical impact simulator.

KAM		ATS		ITR	
Percentile of <i>in vivo</i> Population Cohort	Load (Nm)	Percentile of <i>in vivo</i> Population Cohort	Load (N)	Percentile of <i>in vivo</i> Population Cohort	Load (Nm)
2%	2.4	0%	40	0%	1.0
68%	27.0	90%	98	33%	9.7
99%	53.6			67%	18.6
200%	114.6			100%	53.7

Table 2:

Number of specimens that survived each simulation condition.

Full Cohort					Female Specimens					Male Specimens									
		KAM						KAM						KAM					
		2%	68%	99%	200%			2%	68%	99%	200%			2%	68%	99%	200%		
ATS & ITR	0%, 0%	34	33	33	NaN			0%, 0%	15	15	14	NaN			0%, 0%	19	18	19	200%
	0%, 33%	34	34	33	19			0%, 33%	15	15	15	5			0%, 33%	19	19	18	14
	0%, 67%	34	34	34	20			0%, 67%	15	15	15	5			0%, 67%	19	19	19	15
	0%, 100%	NaN	24	20	20			0%, 100%	NaN	9	5	4			0%, 100%	NaN	15	15	16
	90%, 33%	NaN	20	21	19			90%, 33%	NaN	6	6	5			90%, 33%	NaN	14	15	14
	90%, 67%	NaN	19	20	19			90%, 67%	NaN	4	5	4			90%, 67%	NaN	15	15	15
	90%, 100%	NaN	19	21	20			90%, 100%	NaN	5	5	4			90%, 100%	NaN	14	16	16

Table 3:

Intra- and inter-tester reliability for tibial slope measurements calculated by novice testers using the method established by Hashemi, et al.²²

	Intra-tester ICC (2-k)			Inter-tester ICC (2-1)
	Tester 1	Tester 2	Mean	
MPTS	0.950	0.917	0.931	0.840
LPTS	0.895	0.792	0.844	0.670
CPTS	0.835	0.762	0.798	0.196

Table 4:

R-squared values between tibial slope angle and ligament strain values from the trials where peak strain was recorded prior to failure.

	MPTS	LPTS	CPTS
max ACL strain	0.060	0.048	0.004
ACL strain from IC	0.025	0.090	0.000
ACL strain from Baseline	0.080	0.043	0.013

TS = tibial slope, IC = initial contact, = change

Table 5:

Linear regression slope (change in ACL strain % relative to 1 ° increase in posterior tibial slope angle) and R-squared correlations between tibial slope angle and ACL strains for each simulated impact condition. Columns are organized by magnitude of applied KAM, rows are organized by magnitude of applied ATS and ITR. Simulation conditions where R-squared was 0.2 are highlighted³¹

MPTS vs Peak Absolute ACL Strain					MPTS vs ACL Strain from IC					MPTS vs ACL Strain from Baseline						
	KAM					KAM					KAM					
	2%	68%	99%	200%		2%	68%	99%	200%		2%	68%	99%	200%		
ATS & ITR	0%, 0%	0.431, 0.04	0.223, 0.02	0.218, 0.02	NaN, NaN	ATS & ITR	0%, 0%	0.483, 0.08	0.250, 0.03	0.308, 0.05	NaN, NaN	0%, 0%	0.483, 0.08	0.233, 0.02	0.232, 0.02	NaN, NaN
	0%, 33%	0.298, 0.02	0.289, 0.03	0.224, 0.02	0.896, 0.20		0%, 33%	0.407, 0.06	0.377, 0.05	0.417, 0.11	0.396, 0.16	0%, 33%	0.408, 0.06	0.371, 0.05	0.325, 0.06	0.637, 0.37
	0%, 67%	0.285, 0.02	0.319, 0.03	0.181, 0.02	1.035, 0.20		0%, 67%	0.267, 0.03	0.386, 0.07	0.422, 0.12	0.839, 0.37	0%, 67%	0.293, 0.04	0.360, 0.05	0.277, 0.04	1.139, 0.38
	0%, 100%	NaN, NaN	0.996, 0.14	0.787, 0.15	0.569, 0.08		0%, 100%	NaN, NaN	0.890, 0.18	0.226, 0.08	0.284, 0.17	0%, 100%	NaN, NaN	0.862, 0.15	0.488, 0.18	0.487, 0.22
	90%, 33%	NaN, NaN	0.523, 0.08	1.443, 0.34	0.519, 0.09	90%, 33%	NaN, NaN	0.238, 0.11	1.086, 0.31	0.376, 0.25	90%, 33%	NaN, NaN	0.389, 0.18	1.206, 0.38	0.412, 0.21	
	90%, 67%	NaN, NaN	0.442, 0.07	0.707, 0.13	0.578, 0.08	90%, 67%	NaN, NaN	0.133, 0.04	0.110, 0.03	0.364, 0.24	90%, 67%	NaN, NaN	0.267, 0.08	0.303, 0.06	0.517, 0.31	
	90%, 100%	NaN, NaN	1.254, 0.27	0.318, 0.04	0.701, 0.13	90%, 100%	NaN, NaN	0.973, 0.23	0.201, 0.06	0.242, 0.13	90%, 100%	NaN, NaN	1.136, 0.29	0.457, 0.14	0.481, 0.23	

LPTS vs Peak Absolute ACL Strain					LPTS vs ACL Strain from IC					LPTS vs ACL Strain from Baseline						
	KAM					KAM					KAM					
	2%	68%	99%	200%		2%	68%	99%	200%		2%	68%	99%	200%		
ATS & ITR	0%, 0%	0.515, 0.05	0.353, 0.03	0.380, 0.04	NaN, NaN	ATS & ITR	0%, 0%	0.527, 0.07	0.256, 0.03	0.236, 0.03	NaN, NaN	0%, 0%	0.527, 0.07	0.269, 0.03	0.336, 0.04	NaN, NaN
	0%, 33%	0.455, 0.04	0.372, 0.04	0.302, 0.02	0.633, 0.09		0%, 33%	0.447, 0.06	0.330, 0.04	0.273, 0.04	0.235, 0.06	0%, 33%	0.453, 0.06	0.366, 0.04	0.300, 0.04	0.527, 0.21
	0%, 67%	0.286, 0.02	0.387, 0.04	0.375, 0.04	1.043, 0.17		0%, 67%	0.185, 0.02	0.371, 0.05	0.312, 0.06	0.769, 0.26	0%, 67%	0.222, 0.02	0.426, 0.06	0.313, 0.04	1.113, 0.31
	0%, 100%	NaN, NaN	1.095, 0.15	0.380, 0.05	0.471, 0.06		0%, 100%	NaN, NaN	0.881, 0.15	0.176, 0.06	0.195, 0.09	0%, 100%	NaN, NaN	0.946, 0.16	0.372, 0.10	0.303, 0.09
	90%, 33%	NaN, NaN	0.377, 0.05	1.185, 0.19	0.637, 0.10	90%, 33%	NaN, NaN	0.237, 0.09	1.068, 0.25	0.185, 0.06	90%, 33%	NaN, NaN	0.331, 0.10	1.169, 0.29	0.260, 0.08	
	90%, 67%	NaN, NaN	0.304, 0.04	0.390, 0.05	0.593, 0.07	90%, 67%	NaN, NaN	-0.013, 0.04	-0.082, 0.03	0.240, 0.10	90%, 67%	NaN, NaN	0.057, 0.04	0.144, 0.04	0.429, 0.18	
	90%, 100%	NaN, NaN	1.101, 0.19	0.405, 0.04	0.571, 0.07	90%, 100%	NaN, NaN	0.934, 0.19	0.031, 0.04	0.238, 0.11	90%, 100%	NaN, NaN	1.012, 0.22	0.206, 0.04	0.254, 0.07	

CPTS vs Peak Absolute ACL Strain					CPTS vs ACL Strain from IC					CPTS vs ACL Strain from Baseline						
	KAM					KAM					KAM					
	2%	68%	99%	200%		2%	68%	99%	200%		2%	68%	99%	200%		
ATS & ITR	0%, 0%	-0.196, 0.02	0.028, 0.02	-0.192, 0.02	NaN, NaN	ATS & ITR	0%, 0%	-0.065, 0.02	-0.005, 0.02	-0.135, 0.02	NaN, NaN	0%, 0%	-0.070, 0.02	-0.029, 0.02	-0.008, 0.02	NaN
	0%, 33%	-0.124, 0.01	-0.248, 0.02	0.005, 0.02	-0.147, 0.01		0%, 33%	-0.067, 0.02	-0.097, 0.01	-0.143, 0.03	-0.115, 0.01	0%, 33%	-0.069, 0.02	-0.135, 0.01	-0.163, 0.03	-0.1
	0%, 67%	-0.071, 0.01	-0.138, 0.02	-0.266, 0.02	-0.811, 0.05		0%, 67%	-0.045, 0.01	-0.150, 0.02	-0.196, 0.03	-0.492, 0.07	0%, 67%	-0.065, 0.01	-0.119, 0.02	-0.166, 0.02	-0.7
	0%, 100%	NaN, NaN	-0.526, 0.03	-0.521, 0.03	-0.495, 0.02		0%, 100%	NaN, NaN	-0.830, 0.08	-0.345, 0.07	-0.063, 0.02	0%, 100%	NaN, NaN	-0.649, 0.05	-0.325, 0.04	-0.3

CPTS vs Peak Absolute ACL Strain					CPTS vs ACL Strain from IC					CPTS vs ACL Strain from Baseline							
KAM					KAM					KAM							
		2%	68%	99%	200%			2%	68%	99%	200%			2%	68%	99%	200%
	90%, 33%	NaN, NaN	-0.230,0.03	-0.768,0.05	-0.313, 0.01			NaN, NaN	-0.126,0.06	-0.782,0.09	-0.110,0.02			NaN, NaN	-0.127,0.06	-0.753,0.09	-0.292,0.05
	90%, 67%	NaN, NaN	-0.440,0.02	-0.642,0.04	-0.398, 0.02			NaN, NaN	-0.476,0.10	-0.512,0.08	-0.143,0.03			NaN, NaN	-0.442,0.08	-0.581,0.08	-0.165,0.03
	90%, 100%	NaN, NaN	-0.743,0.05	-0.538,0.04	-0.284, 0.01			NaN, NaN	-1.045,0.12	-0.448,0.10	-0.079,0.02			NaN, NaN	-0.915,0.10	-0.522,0.08	-0.377,0.07

NaN = that combination of external loading magnitudes was not utilized during impact simulations

Table 6:

Mean male and female tibial slope angle measurements by compartment.

	MPTS	LPTS	CPTS
Males	7.1 (2.5)°	5.9 (2.1)°	1.9 (1.5)°
Females	6.0 (4.1)°	7.3 (2.9)°	1.0 (1.8)°
P-value	0.339	0.107	0.119

Table 7:

Linear regression slope (change in ACL strain % relative to 1 ° increase in posterior tibial slope angle) and R-squared values between tibial slope angle in female specimens and ACL strains for each simulated impact condition. Columns are organized by magnitude of applied KAM, rows are organized by magnitude of applied ATS and ITR. Simulation conditions where R-squared was 0.2 are highlighted.³¹

MPTS vs Peak Absolute ACL Strain					MPTS vs ACL Strain from IC					MPTS vs ACL Strain from Baseline				
KAM					KAM					KAM				
	2%	68%	99%	200%		2%	68%	99%	200%		2%	68%	99%	200%
ATS & ITR	0%, 0%	-0.459, 0.02	-0.300, 0.00	-0.170,0.01	NaN, NaN		-1.064, 0.07	-0.909, 0.01	-0.942, 0.02	NaN, NaN		-1.063, 0.07	-0.898, 0.01	-0.729, 0.00
	0%, 33%	-0.276, 0.01	-0.413, 0.01	-0.209, 0.01	-0.348,0.42		-0.926, 0.05	-0.933, 0.03	-0.822, 0.08	0.151, 0.07		-0.928, 0.05	-0.952, 0.03	-0.933, 0.03
	0%, 67%	-0.369, 0.00	-0.350, 0.01	-0.225,0.01	-2.351, 0.36		-0.848, 0.02	-0.822, 0.05	-0.645, 0.08	-2.226, 0.28		-0.872, 0.03	-0.780, 0.03	-0.768, 0.01
	0%, 100%	NaN, NaN	-0.613, 0.04	0.662,0.05	0.904,0.05	ATS & ITR	NaN, NaN	-1.469, 0.16	0.406, 0.27	0.113, 0.02	ATS & ITR	NaN, NaN	-1.233, 0.09	0.020, 0.01
	90%, 33%	NaN, NaN	0.783, 0.04	-2.727, 0.40	1.024,0.10		NaN, NaN	0.120, 0.01	-3.252, 0.35	-0.070, 0.09		NaN, NaN	0.037, 0.03	-3.412, 0.43
	90%, 67%	NaN,NaN	1.054, 0.11	1.135, 0.28	0.800,0.10		NaN,NaN	-0.308, 0.11	0.557, 0.26	0.025, 0.04		NaN, NaN	-0.325, 0.23	0.445, 0.09
	90%, 100%	NaN,NaN	-2.357, 0.21	1.021, 0.44	0.937,0.04		NaN,NaN	-3.844, 0.24	0.029, 0.13	0.076, 0.02		NaN, NaN	-3.947, 0.32	0.225, 0.04
														-0.825, 0.55

LPTS vs Peak Absolute ACL Strain					LPTS vs ACL Strain from IC					LPTS vs ACL Strain from Baseline				
KAM					KAM					KAM				
	2%	68%	99%	200%		2%	68%	99%	200%		2%	68%	99%	200%
ATS & ITR	0%, 0%	-0.261, 0.02	-0.302, 0.02	-0.063,0.01	NaN, NaN		-0.853, 0.03	-0.959, 0.01	-1.042, 0.01	NaN, NaN		-0.844, 0.03	-0.892, 0.01	-0.562, 0.01
	0%, 33%	-0.142, 0.01	-0.339, 0.01	-0.077,0.01	-0.166,0.11		-0.879, 0.02	-1.019, 0.01	-0.912, 0.06	0.322, 0.54		-0.861, 0.02	-1.001, 0.01	-0.875, 0.03
	0%, 67%	-0.122, 0.01	-0.290, 0.01	-0.151, 0.01	-0.026,0.34	ATS & ITR	-0.655, 0.01	-0.959, 0.02	-0.800, 0.05	-0.359, 0.31		-0.676, 0.01	-0.759, 0.02	-0.734, 0.01
	0%, 100%	NaN, NaN	0.067, 0.08	0.712,0.20	0.925,0.33		NaN,NaN	-1.238, 0.21	0.31, 0.18	0.144, 0.71	ATS & ITR	NaN,NaN	-0.721, 0.15	-0.178, 0.09
	90%, 33%	NaN, NaN	0.703, 0.14	-1.060,0.45	1.075, 0.17		NaN,NaN	-0.187, 0.16	-1.565, 0.51	-0.032, 0.41		NaN,NaN	-0.288,0.19	-1.920, 0.58
	90%, 67%	NaN,NaN	1.023, 0.42	1.144,0.39	0.841,0.30		NaN,NaN	-0.322, 0.22	0.518, 0.28	0.086, 0.57		NaN,NaN	-0.313, 0.53	0.205, 0.10
	90%, 100%	NaN,NaN	0.460, 0.31	1.350, 0.52	0.967, 0.49		NaN,NaN	-0.977, 0.49	-0.181, 0.08	0.123, 0.48		NaN,NaN	-1.019, 0.52	0.034, 0.07
														-0.733, 0.23

CPTS vs Peak Absolute ACL Strain					CPTS vs ACL Strain from IC					CPTS vs ACL Strain from Baseline				
KAM					KAM					KAM				
	2%	68%	99%	200%		2%	68%	99%	200%		2%	68%	99%	200%
ATS & ITR	0%, 0%	0.059, 0.06	-0.458, 0.15	-0.094,0.12	NaN, NaN		0.047, 0.05	-0.273, 0.09	0.148, 0.04	NaN, NaN		0.039, 0.05	-0.288, 0.09	-0.238, 0.10
	0%, 33%	-0.001, 0.11	0.125, 0.06	0.089, 0.09	0.240, 0.16	ATS & ITR	0.043, 0.06	0.266, 0.04	0.394,0.03	-0.542, 0.44		0.027, 0.06	0.260, 0.04	0.326,0.03
	0%, 67%	0.145, 0.10	0.086, 0.10	0.095, 0.08	1.985, 0.17		0.25, 0.07	0.278, 0.05	0.297,0.04	1.344, 0.21		0.254, 0.07	0.104, 0.08	0.083,0.06
	0%, 100%	NaN, NaN	0.192, 0.10	0.027,0.16	0.367,0.13		NaN, NaN	1.386, 0.07	-0.400,0.13	-0.014, 0.47		NaN, NaN	1.010, 0.08	0.109, 0.23
														0.094, 0.44

CPTS vs Peak Absolute ACL Strain						CPTS vs ACL Strain from IC						CPTS vs ACL Strain from Baseline									
KAM						KAM						KAM									
			2%	68%	99%	200%				2%	68%	99%	200%				2%	68%	99%	200%	
	90%, 33%	NaN, NaN	NaN, NaN	-0.490, 0.17	1.998, 0.17	0.619, 0.06			90%, 33%	NaN, NaN	0.002, 0.42	1.970, 0.22	-0.455, 0.27				90%, 33%	NaN, NaN	0.053, 0.43	2.288, 0.24	0.127, 0.32
	90%, 67%	NaN, NaN	NaN, NaN	0.220, 0.21	-0.357, 0.22	0.390, 0.12			90%, 67%	NaN, NaN	-0.173, 0.14	-0.650, 0.14	-0.069, 0.49				90%, 67%	NaN, NaN	-0.010, 0.39	-0.280, 0.13	0.119, 0.44
	90%, 100%	NaN, NaN	NaN, NaN	2.377, 0.16	-1.171, 0.30	0.298, 0.16			90%, 100%	NaN, NaN	2.310, 0.22	-0.216, 0.17	-0.144, 0.50				90%, 100%	NaN, NaN	2.530, 0.22	-0.492, 0.56	-0.027, 0.35

NaN= that combination of external loading magnitudes was not utilized during impact simulations.

Table 8:

Linear regression slope (change in ACL strain % relative to 1° increase in posterior tibial slope angle) and R-squared values between tibial slope angle in male specimens and ACL strains for each simulated impact condition. Columns are organized by magnitude of applied KAM, rows are organized by magnitude of applied ATS and ITR. Simulation conditions where R-squared was > 0.2 are highlighted³¹

MPTS vs Peak Absolute ACL Strain					MPTS vs ACL Strain from IC					MPTS vs ACL Strain from Baseline					
KAM					KAM					KAM					
	2%	68%	99%	200%		2%	68%	99%	200%		2%	68%	99%	200%	
ATS & ITR	0%, 0%	0.540, 0.12	0.578, 0.13	0.531, 0.14	NaN, NaN	0%, 0%	0.214, 0.22	0.213, 0.22	0.208, 0.21	NaN, NaN	0%, 0%	0.213, 0.22	0.201, 0.22	0.196,0.20	NaN,NaN
	0%, 33%	0.695, 0.11	0.500, 0.12	0.531, 0.11	0.367, 0.24	0%, 33%	0.283, 0.26	0.103, 0.20	0.146, 0.22	0.117, 0.37	0%, 33%	0.283, 0.25	0.128, 0.21	0.126, 0.22	0.132, 0.36
	0%, 67%	0.412, 0.19	0.495, 0.13	0.534, 0.14	0.726, 0.11	0%, 67%	0.222, 0.21	0.246, 0.30	0.083, 0.28	0.086, 0.40	0%, 67%	0.262, 0.21	0.213, 0.3	0.091, 0.27	0.052, 0.26
	0%, 100%	NaN, NaN	0.545, 0.28	0.503, 0.25	0.843, 0.11	ATS & ITR	NaN, NaN	0.004, 0.22	0.055, 0.31	0.134, 0.37	ATS & ITR	NaN, NaN	1 0.119,0.27	0.156, 0.32	0.129, 0.21
	90%, 33%	NaN, NaN	0.399, 0.23	0.500, 0.28	0.485, 0.23	90%, 33%	NaN, NaN	0.041, 0.28	0.028, 0.30	0.097, 0.40	90%, 33%	NaN, NaN	0.113,0.33	0.115, 0.31	0.060, 0.25
	90%, 67%	NaN\NaN	0.632, 0.14	0.489, 0.27	0.804, 0.11	90%, 67%	NaN\NaN	-0.011, 0.08	0.034, 0.11	0.104, 0.41	90%, 67%	NaN, NaN	0.021, 0.12	0.124, 0.17	0.119, 0.20
	90%, 100%	NaN\NaN	0.426, 0.25	0.778, 0.13	0.503, 0.17	90%, 100%	NaN, NaN	-0.091, 0.13	-0.007, 0.15	0.142, 0.26	90%, 100%	NaN, NaN	0.011,0.21	0.116, 0.28	0.231, 0.22

LPTS vs Peak Absolute ACL Strain					LPTS vs ACL Strain from IC					LPTS vs ACL Strain from Baseline					
KAM					KAM					KAM					
	2%	68%	99%	200%		2%	68%	99%	200%		2%	68%	99%	200%	
ATS & ITR	0%,0%	1.048, 0.04	0.990, 0.07	1.053, 0.05	NaN, NaN	0%, 0%	0.345, 0.09	0.352, 0.10	0.408, 0.08	NaN, NaN	0%, 0%	0.344, 0.09	0.343, 0.10	0.359, 0.08	NaN,NaN
	0%, 33%	1.253, 0.05	1.021, 0.04	1.072, 0.03	0.876, 0.14	0%, 33%	0.497, 0.09	0.306, 0.05	0.346, 0.05	0.338, 0.18	0%, 33%	0.496, 0.09	0.329, 0.05	0.287,0.05	0.398, 0.24
	0%, 67%	0.946, 0.04	0.944, 0.03	0.964, 0.05	1.302, 0.12	0%, 67%	0.452, 0.04	0.345, 0.09	0.363, 0.06	0.354, 0.22	0%, 67%	0.482, 0.04	0.325, 0.09	0.325, 0.07	0.372, 0.14
	0%, 100%	NaN, NaN	0.981, 0.10	1.039, 0.07	1.263, 0.09	ATS & ITR	NaN, NaN	0.372, 0.06	0.264, 0.11	0.326, 0.24	ATS & ITR	NaN, NaN	0.484, 0.07	0.402, 0.11	0.445, 0.09
	90%, 33%	NaN, NaN	0.823, 0.11	0.931, 0.09	0.819, 0.15	90%, 33%	NaN, NaN	0.226, 0.15	0.288, 0.09	0.359, 0.25	90%, 33%	NaN, NaN	0.376, 0.17	0.328, 0.10	0.409, 0.12
	90%, 67%	NaN\NaN	1.251, 0.10	0.954, 0.07	1.337, 0.13	90%, 67%	NaN\NaN	0.301, 0.06	0.193, 0.04	0.298, 0.26	90%, 67%	NaN, NaN	0.390, 0.06	0.371,0.04	0.346, 0.20
	90%, 100%	NaN\NaN	0.802, 0.12	1.384, 0.11	0.586, 0.11	90%, 100%	NaN, NaN	0.302, 0.07	0.395, 0.07	0.139, 0.28	90%, 100%	NaN, NaN	0.373, 0.07	0.574, 0.08	0.398, 0.09

CPTS vs Peak Absolute ACL Strain					CPTS vs ACL Strain from IC					CPTS vs ACL Strain from Baseline					
KAM					KAM					KAM					
	2%	68%	99%	200%		2%	68%	99%	200%		2%	68%	99%	200%	
ATS & ITR	0%,0%	0.769, 0.06	0.956, 0.04	0.642, 0.06	NaN, NaN	0%, 0%	0.257, 0.04	0.241, 0.04	0.260, 0.04	NaN, NaN	0%, 0%	0.251, 0.04	0.251, 0.04	0.260, 0.04	NaN,NaN
	0%, 33%	0.549, 0.06	0.719, 0.06	0.963, 0.05	1.183, 0.01	ATS & ITR	0%, 33%	0.138, 0.05	0.389, 0.03	0.208, 0.05	0.300, 0.01	0%, 33%	0.135, 0.05	0.363, 0.03	0.241, 0.05
	0%, 67%	0.692, 0.04	0.797, 0.05	0.415, 0.06	0.993, 0.03		0%, 67%	0.034, 0.07	-0.012, 0.05	0.109, 0.04	0.081, 0.04	0%, 67%	0.003, 0.07	0.023, 0.05	0.107, 0.05
	0%, 100%	NaN, NaN	0.617, 0.03	0.553, 0.04	0.825, 0.03		0%, 100%	NaN, NaN	-0.127, 0.19	-0.086, 0.09	0.122, 0.04	0%, 100%	NaN, NaN	-0.151, 0.14	-0.133, 0.06

CPTS vs Peak Absolute ACL Strain						CPTS vs ACL Strain from IC						CPTS vs ACL Strain from Baseline				
KAM						KAM						KAM				
	2%	68%	99%	200%		2%	68%	99%	200%			2%	68%	99%	200%	
	90%, 33%	NaN, NaN	1.029, 0.03	0.526, 0.03	1.196, 0.01	90%, 33%	NaN, NaN	-0.043, 0.11	-0.118, 0.14	0.125, 0.04		90%, 33%	NaN, NaN	-0.025, 0.09	-0.189, 0.11	0.096, 0.08
	90%, 67%	NaN, NaN	0.834, 0.05	0.471, 0.04	0.885, 0.03	90%, 67%	NaN, NaN	-0.013, 0.16	-0.248, 0.12	0.098, 0.04		90%, 67%	NaN, NaN	0.033, 0.15	-0.269, 0.11	0.054, 0.03
	90%, 100%	NaN, NaN	0.932, 0.03	0.694, 0.04	0.978, 0.01	90%, 100%	NaN, NaN	-0.111, 0.20	-0.095, 0.16	0.273, 0.02		90%, 100%	NaN, NaN	-0.212, 0.14	-0.304, 0.12	-0.062, 0.10

NaN = that combination of external loading magnitudes was not utilized during impact simulations