Anterior tibiofemoral intersegmental forces during landing are predicted by passive restraint measures in women

By: Randy J. Schmitz, Jerome J. Sauret, Sandra J. Shultz

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Abstract:

Background: Passive restraint capabilities may influence sagittal plane knee joint mechanics during activity. This study aimed to determine if measures associated with passive restraint of anterior translation of the tibia are predictive of peak anterior knee shear force during landing.

Methods: Passive restraint measures were assessed via joint arthrometry and during 40% body weight simulated weight acceptance using recreationally active students (73 F, 42 M; 21.8 ± 2.9 yr, 1.69 ± 0.1 m, 68.9 ± 14.1 kg). Anterior knee laxity (mm) at 133 N and initial (0–20 N) and terminal (100–130 N) anterior stiffnesses (N/mm) were calculated from arthrometer data. Peak anterior tibial acceleration ($m \cdot s^{-2}$) relative to the femur was assessed via electromagnetic position sensors during 40% body weight acceptance trials. Peak knee shear force was assessed during double-leg drop jumps.

Results: Sex specific linear stepwise regressions revealed that in females, increasing peak tibial acceleration (5.1 ± 1.8 $m \cdot s^{-2}$) ($R^2\Delta$ = 7.3%, PΔ = 0.021), increasing initial anterior stiffness (31.0 ± 14.0 N/mm) ($R^2\Delta$ = 5.9%, PΔ = 0.032), and decreasing terminal anterior stiffness (43.4 ± 17.4 N/mm) ($R^2\Delta$ = 4.9%, PΔ = 0.046) collectively predicted greater peak knee shear forces (66.6 ± 12.03% BW) (multiple R^2 = 18.1%). No male regressions were significant.

Conclusions: Sagittal laxity measures are associated with anterior knee shear loads during landing in females. Greater tibial acceleration during early axial load along with greater initial and lesser terminal anterior stiffnesses predicted increasing anterior knee shear forces. Future work should investigate the combined contribution of passive and active restraints to high-risk ACL biomechanics.

Keywords: Tibial acceleration | Laxity | ACL

Article:

1. INTRODUCTION

Sagittal plane mechanisms are commonly thought to contribute in part to high risk biomechanics associated with ACL injury [1]. Specifically, upright landing styles associated with decreased knee [2] and [3] and hip flexion [4] angles are thought to increase sagittal plane knee loading [4], likely through increased reliance on passive restraints (ligamentous structures) for knee stability. This upright posture is thought to result in increased landing forces [5] and subsequently increase anterior knee shear forces and ACL loading [6]. Proximal anterior tibial shear force is considered to be a major contributor to the loading of the ACL [7] and [8] with the ACL acting as the primary restraint system to anterior displacement of the tibia with respect to the femur [9]. Previous work has demonstrated that a combination of biomechanical factors including posterior ground reaction force, external knee flexion moment, knee flexion angle, integrated EMG activity of the vastus lateralis, and sex predicted a large amount of the variance (~ 86%) in anterior tibial shear force during a deceleration task [10]. While it is understood that internal and external loads may influence anterior shear forces, it is not understood how passive restraints responsible for controlling anterior tibial motion (which is dominated by ACL function [9]) may affect sagittal knee plane joint biomechanics during functional activity. As greater anterior knee laxity has been associated with greater anterior tibial translation [11], and greater translations are ultimately a function of anteriorly directed forces, it can be theorized that anterior knee laxity may be related to intersegmental shear forces.

It has been suggested that risk of ACL injury may increase in the absence of sufficiently taut passive restraints [12]. While multiple factors likely contribute to greater ACL injury risk in females, both retrospective [13], [14], [15], [16] and [17] and prospective [12] and [18] studies identify a relationship between ACL injury and increased sagittal knee joint laxity. Although it is acknowledged that ACL injury is likely multiplanar in nature and that non-sagittal plane laxity may also contribute to high-risk mechanics [19], other work has focused on the relation of anterior knee laxity to total anterior tibial translation during weight acceptance, demonstrating a positive association between increased anterior knee laxity and greater anterior tibial translation upon weight acceptance [11]. Further, a combination of sagittal plane laxity measures has been associated with a landing strategy that resulted in greater workload about the knee [20]. This was postulated to be an attempt to stabilize the knee and reduce the loads applied to the ligamentous tissue; however also potentially rendering the knee less able to resist injurious forces [20]. Collectively, these studies suggest that passive restraints may be an important contributor to dynamic knee stability and impact ACL injury risk [12].

A deconstruction of previously reported sagittal plane measures of anterior knee laxity and anterior tibial translation upon weight acceptance [21] may allow even greater insight into passive restraint mechanics of anterior knee loading during functional activity (e.g. landing from a jump). Characteristics of the load–displacement curve (stiffness) of anterior knee laxity are thought to be important with respect to the clinical functioning of the knee joint [22] with alterations in loading range specific anterior knee stiffness postulated to be a factor in ACL injury risk [23]. The terminal phase of arthrometer loading is thought to be where the ACL fully engages in restraint [24]. Decreases in incremental stiffness at higher arthrometer loadings may be associated with ligamentous restraint behavior that may potentially result in altered arthrokinematics during functional activity [25]. The initial loading phase is thought to largely represent the resistance provided by the weight of the limb; however, the impact of early loading range stiffness on functional biomechanics is unknown.

While anterior knee laxity is predictive of anterior tibial translation during weight acceptance [11], ACL strain has been reported to be proportional to anterior tibial acceleration during weight acceptance [26]. Thus, it would be of benefit to further understand how passive (ligamentous) knee joint behavior translates to functional behavior of the knee during activity commonly associated with the ACL injury mechanism of landing from a jump. Thus the purpose of our investigation was to determine if measures associated with passive restraint of anterior tibial translation at the knee joint are predictive of peak anterior knee shear force during a drop jump landing in females and males. Given the widely reported sex differences in landing biomechanics, we chose to include sex-specific analyses to account for these sex differences in landing mechanics. It was hypothesized that a combination of measures associated with passive restraint of anterior translation of the tibia (anterior knee laxity, initial and terminal knee stiffness during anterior loading, and tibial acceleration during simulated weight acceptance) would collectively predict greater proximal anterior knee shear forces during a landing maneuver.

2. METHODS

For purposes of a larger study examining the effects of hormone mediated knee joint laxity on weight bearing knee joint neuromechanics [21], 73 females and 42 males (21.8 ± 2.9 yr, 1.69 ± 0.1 m, 68.9 ± 14.1 kg) between 18 and 30 years of age volunteered to participate. Eight fewer males than the original study [21] were included due to technical problems with continuous anterior load–displacement data acquisition The study was approved by the University Institutional Review Board and all participants gave written informed consent to participate. Participants were recreationally active (2.5–10 h/week) for the past 3 months and non-smokers, and had a body mass index (weight/height^2) ≤ 30 and no history of ligament or cartilage injury to the knee. Females were tested during the first 6 days of menses to control for any potential acute hormone effects on joint laxity [27] and subsequently, neuromuscular control [21]. To ensure inclusion of a broad range of knee laxity values, participants were prescreened on anterior knee laxity and needed to fall within a predetermined anterior knee laxity distribution matrix. All testing was performed on the dominant leg (preferred stance leg when kicking a ball). Subjects were familiarized to all study procedures approximately 2 weeks prior to testing, and were asked to refrain from any physical activity on the day of testing until all measurements were obtained. Subjects completed a 5-minute warm-up on a stationary bike before data collection.

To determine anterior knee laxity (AKL) and initial and terminal knee stiffness, instrumented joint arthrometer testing was performed. Procedures for obtaining anterior knee laxity (AKL) data and its measurement and consistency have been previously reported [27]. Following common clinical practice and previously established methods of instrumented knee laxity testing, AKL represented the anterior displacement (mm) of the tibia relative to the femur produced by an anterior load of 133 N applied to the posterior tibia with the knee flexed to 25 ± 5° using the KT-2000TM knee arthrometer (Medmetric Corp, San Diego, CA) [21] and [27] Real-time load and displacement data were collected from the three AKL trials and were exported to a spreadsheet for later calculation of incremental stiffness values [25]. Using methods previously described [25], we extracted the initial (0–20 N load) and terminal (100–130 N load) stiffnesses (N/mm). The average of the three trials was used for analysis. Measurement consistency and prediction were previously assessed on 38 males tested 2 weeks apart (Initial Stiffness — ICC2,3 = 0.87, SEM = 4.2 N/mm & Terminal Stiffness — $ICC_{2,3}$ = 0.90, SEM = 5.3 N/mm) [25].

To measure anterior tibial acceleration during the initial phase of weight acceptance, tibiofemoral kinematics during 40% weight bearing acceptance was assessed with the Vermont Knee Laxity Device (VKLD) as described previously in detail [11]. The VKLD measures displacement of the tibia relative to the femur as the knee transitions from non-weight bearing to weight bearing, and characterizes the anterior–posterior load–displacement behavior of the knee [28]. Features of the VKLD include the capability to apply quantifiable loads to the tibiofemoral joint under the control of gravity, by first creating an absolute zero shear load condition across the tibiofemoral joint while it is un-weighted to establish a reproducible neutral initial position of the tibia relative to the femur, and then to apply standardized compressive loads through the ankle and hip axes of rotation of the limb to simulate weight-bearing [29].

Subjects were placed in the VKLD and the foot was strapped to the foot cradle connected to a calibrated six-degree force transducer. The second metatarsal was visually aligned to the anterior superior iliac spine (ASIS) and the greater trochanter and the lateral malleolus were aligned to the axes of the hip and ankle counter-weight systems respectively. These counter-weight systems were applied to the shank and thigh to eliminate gravity forces caused by the shank and thigh segments and created zero shear forces across the knee joint. Three electromagnetic position sensors (Mini Birds, Ascension Technologies, Colchester, VT USA) were attached on the midpoint of the lateral thigh, the center of the patellar and the midpoint of the shaft of the tibia. The centroid method estimated the center of rotation of the ankle, knee, and hip joints. After determination of joint centers, the ankle and knee were flexed to 90° and 20° respectively and subjects were asked to relax their leg muscles. Knee flexion angle (20°) was confirmed manually and with the electromagnetic position sensors. Once properly positioned in the VKLD, three anterior to posterior forces were applied to the tibia just below the knee joint line to standardize the neutral position of the knee joint at the beginning of every trial. An initial zero compressive load to the tibia was also confirmed prior to each trial with a six degree-of-freedom load transducer (Model MC3A, Advanced Medical Technology, Inc; Watertown, MA).

Prior to actual data collection, we performed 3–5 practice trials to further familiarize the subject with the weight acceptance trials. Once the zero compressive and shear load were obtained, compressive loads equal to 40% BW were applied by the release of the prescribed weight via a pulley system, which acted through the ankle and hip joint axes to simulate the transition from non-weight bearing to weight bearing (Fig. 1). The 40% BW load is consistent with what would be experienced during double-leg stance assuming 50% of BW applied to each leg and 10% of body weight distributed below the knee [11] and is intended to represent the early weight bearing phase [29]. Starting from a relaxed neuromuscular state, participants were instructed to respond to the axial force as quickly as possible after the release of the 40% BW and to try and maintain the initial knee position (20° knee flexion). During each weight acceptance trial, sEMG signals were acquired at 1000 Hz from the medial and lateral quadriceps and hamstrings while electromagnetic sensors collected kinematic data at 100 Hz.

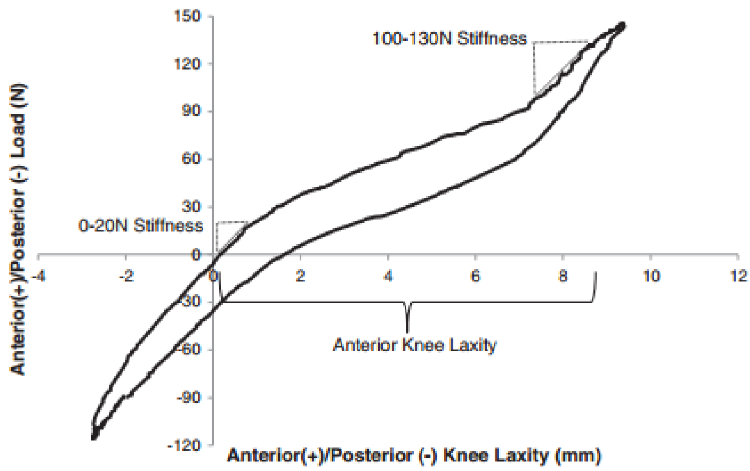

Fig. 1. Anterior knee stiffness – exemplar load–displacement curve during the application of 90–133 N load with resultant anterior knee laxity and 0–20 N and 100–130 N stiffnesses graphically represented.

For VKLD testing, raw position data were low-pass filtered at 10 Hz using a 4th order zero lag Butterworth filter. A segmental reference system quantified the three dimensional kinematics of the knee during the transition from NWB to WB. For each segment the + Z axis was directed laterally, the + Y axis was directed superiorly, and the + X axis was directed anteriorly. Euler's equations described joint motion about the knee with a rotational sequence of Z Y′ X″. Anterior tibial translation was then defined as anterior displacement of the tibia with respect to the patellar sensor from the initial position non-weight bearing to the peak axial compression force. The second derivative of the anterior translation data was computed to assess anterior tibial acceleration (Fig. 2). The average of the 3 trials was used for analysis. Between day consistency and precision of this measure were assessed on the 42 males who had returned to the lab approximately 2 weeks later for an identical testing session. ($ICC_{2,3}$ = 0.82, SEM = 0.74 $m \cdot s^{-2}$).

To determine the time of muscle onset relative to the resultant anterior tibial translations during 40% weight bearing trials, muscle activation was recorded with surface electromyography (sEMG) using methods previously described in detail [11]. Briefly, sEMG data were obtained from the medial and lateral quadriceps, and the medial and lateral hamstring muscles of the dominant limb. For normalization purposes, sEMG signals were first recorded via a 16-channel Myopac system (Run Technologies, Mission Viejo, CA) (differential detection; CMRR = 90 dB min. @ 60 Hz; input impedance = 1 MΩ; amplification = 1000 ×) during maximal voluntary isometric contractions (MVIC) of the quadriceps and hamstring muscle groups while seated in the Biodex System 3 Isokinetic dynamometer (Biodex Medical Systems, Inc, Shirley, NY USA) with hip flexion angle of 85°and knee flexion angle of 20°. After warm-up, subjects performed three, 3-second maximal isometric contractions of quadriceps and hamstring muscles while collecting sEMG data. Consistent verbal encouragement was provided with each contraction to insure maximal effort. The sEMG signals obtained during both the MVIC and the weight acceptance trials were band pass filtered from 10 Hz to 350 Hz, using a 4th order, zero lag Butterworth filter, and processed with a centered root mean square (RMS) algorithm with 100 ms (MVIC data) or 5 ms (weight acceptance data) time constant [29]. Starting from the resting, relaxed position prior to weight acceptance, muscle onset (ms) was defined as the time when muscle activation exceeded 5 SD of the baseline sEMG signal for 10 ms [30] and [29]. The MQ and LQ values and MH and LH were then averaged to give composite quadriceps and hamstring values, respectfully.

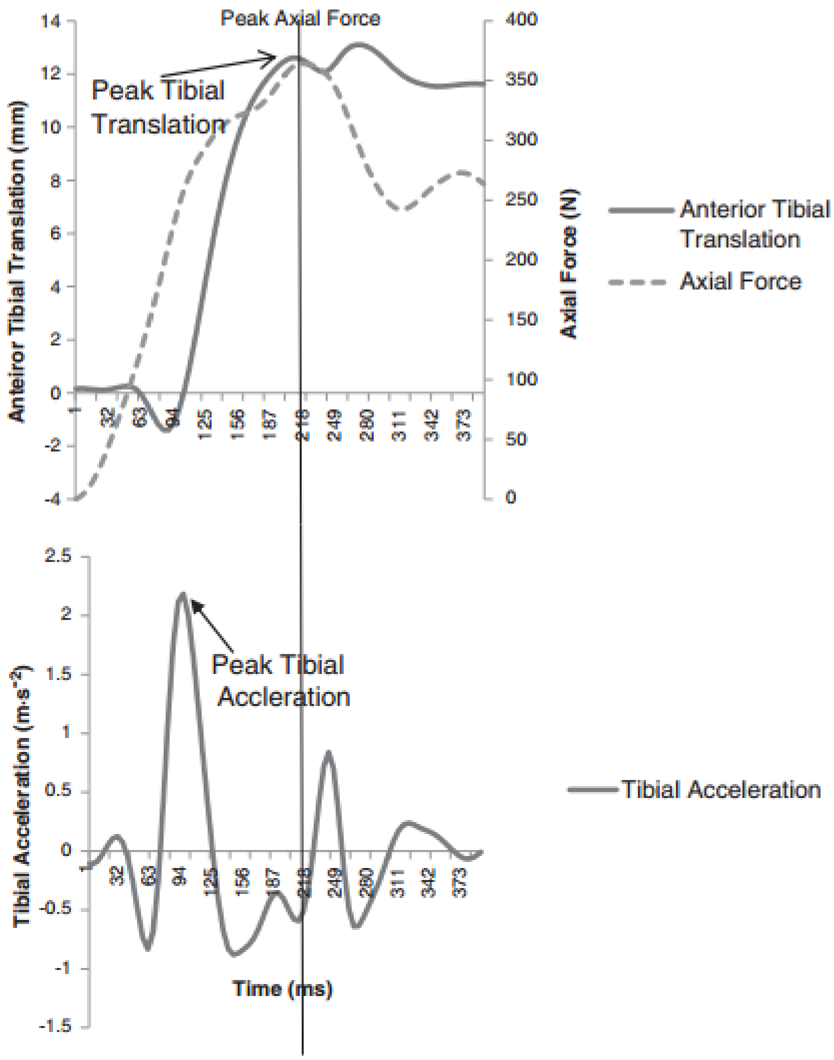

Fig. 2. Exemplar data demonstrating the time dependency of axial load (N), tibial translation (mm) and tibial acceleration ($m \cdot s^{-2}$).

To assess proximal anterior knee shear forces during a landing maneuver, drop jumps were performed barefoot as described previously [31]. Subjects were instrumented with six degrees of freedom electromagnetic sensors (Ascension Technologies, Burlington, VT) over the anterior mid-shaft of the third metatarsal, the mid-shaft of the medial tibia, the lateral aspect of the mid-shaft of the femur, the sacrum and the spinous process of C7. Joint centers were determined using rotation [32] (hip) and centroid [33] (knee and ankle) methods. Kinematic (100 Hz) and kinetic data (1000 Hz) were simultaneously acquired (10 N foot contact threshold trigger) during 5 trials from a 0.45 m platform positioned 0.1 m behind the rear edge of the force plate (Type 4060-nonconducting; Bertec Corporation, Columbus, OH) [19] and [31]. For both tasks, the hands were held at ear level and toes were aligned over the leading edge of the platform. Subjects were instructed to fall off of the box, without jumping or stepping, and land with each foot contacting a forceplate at the same time and immediately performing a maximum double leg vertical jump. No instructions were provided on landing mechanics to prevent experimenter bias. Participants were allowed sufficient practice to insure they were comfortable with the task prior to testing. A trial was discarded if the subject stepped or jumped off the box, contacted the ground with asynchronous foot contact, or landed with the foot off the intended force plate.

For the drop jump, three dimensional hip, knee, and ankle flexion angles were calculated using Euler angle definitions with a rotational sequence of Z Y′ X″. Kinematic data were low pass filtered at 12 Hz and linearly interpolated to kinetic data. Ground reaction force data were also low pass filtered at 12 Hz using a fourth order, zero-lag Butterworth filter (to minimize impact artifact and avoid inaccuracies in joint kinetics [34] and [35]), and normalized to body weight (% BW). Knee intersegmental anterior force (shear force) was calculated via inverse dynamics (Motion Monitor Software; InnSport, Chicago, IL), averaged across 5 trials and normalized to each subject's weight (% BW). Between day consistency and precision were assessed on the 42 males who returned for an identical testing session 2 weeks later ($ICC_{2,5}$ = 0.87; SEM = 9% BW). All data represent the average value across the five trials for each condition.

2.1 Analyses

Although not a direct hypothesis of the study, we first conducted a sex by item (quadriceps, hamstring, and peak tibial acceleration) ANOVA to compare the timing of peak acceleration to time of muscle onset to insure that peak tibial acceleration measures from the 40% BW loading trials occurred prior to quadriceps and hamstrings EMG onsets. To directly address our stated purpose, sex-specific linear stepwise regressions ($P \leq 0.490$ and ≥ 0.510 entry and removal criteria, respectively) determined the extent to which measures associated with passive restraint of anterior tibial translation at the knee joint (anterior knee laxity, initial anterior stiffness, terminal anterior stiffness, & peak anterior tibial acceleration) predicted peak anterior shear knee joint forces during landings.

3. RESULTS

Times to peak acceleration (88.5 ± 28.6 ms) and quadriceps muscle onset (94.2 ± 19.8 ms) were not significantly different ($P > 0.05$) while both were significantly less ($P < 0.05$) than hamstring muscle onset time (124.1 ± 28.8 ms) with no sex main effect ($P > 0.05$) or interaction ($P > 0.05$).

Mean data for all variables are reported by sex in Table 1. The female regression analysis (Table 2) revealed that peak tibial acceleration entered the model first, explaining 7.3% of the variance ($R^2\Delta$ = 7.3%, $P\Delta$ = 0.021), followed by 0–20 N anterior stiffness ($R^2\Delta$ = 5.9%, $P\Delta$ = 0.032), and 100–130 N anterior stiffness ($R^2\Delta$ = 4.9%, $P\Delta$ = 0.046) to collectively predict greater peak knee shear forces (multiple R^2 = 18.1%). Direction of the regression coefficients indicated that higher peak tibial acceleration, increased initial stiffness and decreased terminal stiffness predicted greater peak knee shear forces. None of the independent variables was significant predictors of peak knee shear forces in the male regression model ($P > 0.05$) (Table 3).

Table 1
Descriptive statistics by sex.

	Females		Males	
	Mean ± SD	Min–max	Mean ± SD	Min–max
Time to peak acceleration (ms)	90.2 ± 26.0	56.7–156.7	84.7 ± 32.2	60.0–253.3
Quadriceps onset time (ms)	95.1 ± 21.1	31.0–143.0	92.6 ± 17.9	46.5–124.5
Hamstrings onset time (ms)	126.9 ± 31.7	59.0–203.5	120.5 ± 22.1	78.0–163.0
Peak tibial shear force (% BW)	66.6 ± 12.0	44.0–101.0	62.5 ± 13.7	40.0–94.0
Peak Tibial Acceleration ($m{\cdot}s^{-2}$)	5.1 ± 1.8	1.2–10.3	5.2 ± 1.8	1.8–9.0
0–20 N anterior knee stiffness (N/mm)	31.0 ± 14.0	13.2–121.4	31.6 ± 10.3	11.1–49.9
100–130 N anterior knee stiffness (N/mm)	43.4 ± 17.4	18.0–118.4	36.1 ± 14.9[a]	13.8–71.1
Anterior knee laxity (mm)	6.8 ± 1.9	3.0–13.0	6.3 ± 1.7	4.0–12.0
N = 73 F, 42 M				

[a] Females greater than males ($P < 0.05$).

Table 2
Female stepwise regression model summary.

Stepwise model (with constant)	Change statistics			
	R^2	R^2 change	F change	Sig. F change
Peak tibial acceleration	0.073	0.073	5.555	0.021
Peak tibial acceleration, 0–20 N stiffness	0.132	0.059	4.789	0.032
Peak tibial acceleration, 0–20 N stiffness, 100–130 N stiffness	0.181	0.049	4.139	0.046
Final significant regression equation: Anterior knee shear = 0.585 + 0.016(peak acceleration) + 0.002(0–20 N stiffness) − 0.002(100–130 N stiffness)				

Table 3
Male stepwise regression model summary.

Stepwise model (with constant)	Change statistics			
	R^2	R^2 change	F change	Sig. F change
0–20 N stiffness	0.018	0.018	.744	0.394

4. DISCUSSION

The primary finding of this paper is that measures associated with passive restraint of anterior tibial translation at the knee joint collectively predicted anterior tibial shear forces during landing in females but not males. Specifically, increased tibial acceleration, increased initial stiffness, and decreased terminal stiffness collectively predicted greater anterior tibial shear. Although it is recognized that ACL loading is multiplanar in nature we chose to study anterior shear force as it represents a direct loading mechanism to the ACL; [36] while also acknowledging that the intersegmental forces calculated represent the sum of all forces acting at the knee, not a singular shear force transmitted to the ACL. While ~ 81% of the variance in anterior tibial shear force was not explained by measures associated with passive restraint of anterior tibial translation at the knee joint in the current study, it appears as though that these passive measures do play a role in expression of anterior tibial shear forces. Further, EMG onset time data results suggest voluntary muscle force generation did not influence peak tibial acceleration values, especially after accounting for a typical electromechanical delay of ~ 85 ms [37].

Tibiofemoral axial loading acts through the posterior-inferiorly directed slope of the tibial plateau to induce an anterior directed shear force [38] that has been demonstrated to increase anterior translation and subsequent ACL strain [39]. Mechanistically it has been demonstrated that there is a "slack-taut" position in which the ACL becomes loaded [40]. It is possible that this "slack-taut" position could be adversely altered in those with reduced passive restraint abilities that could lead to a knee position where the tibia may have more time to move anteriorly from an unloaded position before the ACL becomes taut. In turn, this could lead to greater peak anterior tibial acceleration from the axial load thus contributing to anterior shear force.

Given that the ACL injury mechanism involves anterior tibial motion relative to the femur, we chose to study tibia relative to femur linear acceleration. Better understanding of acceleration of the tibia with respect to the femur may be beneficial to understanding dynamic knee stability as past work on the pivot shift has suggested that anterior tibial acceleration is a more important factor than absolute joint motion in pivot shift grading in ACL injury [41]. Whereas tibial acceleration with respect to the world has been reported in the literature in the context of shear mechanics [42], we have been able to locate only one other work reporting acceleration as calculated in the current investigation as the linear anterior acceleration of the tibia with respect to the femur during impacts [26]. Using a cadaver approach, it was demonstrated that peak anterior tibial acceleration was significantly correlated to anterior medial bundle ACL strain during axial loading. Further, it was demonstrated that segmental accelerations were influenced by tibiofemoral joint geometry, as larger posterior tibial slope predicted increased anterior tibial accelerations during simulated single leg landings [26]. We fully acknowledge that our approach to determine acceleration is not completely representative of knee joint mechanics during functional activity as there was no pre-activation. However, this was purposefully done to best ascertain the mechanical characteristics of the sagittal passive restraint system. Given that the hamstrings were activated 124.4.2 ± 28.8 ms following load onset and peak acceleration occurred at 88.5 ± 28.6 ms following load onset, it is likely that hamstring activity had minimal influence on obtained peak acceleration values.

How anterior translation of the tibia to the femur is "dampened" by the passive restraints of the knee (and their respective stiffness properties) may potentially affect joint loading. If the ACL, as the primary passive restraint to anterior tibial shear, had increased dampening properties, it may result in lesser tibial acceleration. Given that $F = ma$, this would functionally result in lesser forces acting on the system. As decreased terminal stiffness has been associated with ACL deficiency [43], those with decreased terminal stiffness, likely have less restraint capability in response to an anteriorly directed load. Further, partial transection of ACLs in animal models suggests that observed decreases in stiffness are accompanied by decreases in ultimate tensile force [44]. Collectively, it appears that anterior tibial acceleration and decreased anterior knee stiffness may be associated with lesser knee stability.

Contrary to our hypothesis, an increase in initial stiffness predicted increase anterior shear load. The 0–20 N (initial) stiffness has been attributed to soft tissue compression during a period in which the arthrometer is matching the weight of the lower extremity [43] and [24]. However, a secondary correlation analysis of body weight and 0–20 N stiffness of current participants revealed no relationship ($r = .09$). We originally hypothesized that lesser initial stiffness would be associated with greater shear due to decreased resistance to anterior tibial translation. However it was greater initial stiffness that was included in the final regression equation to predict greater shear forces. An explanation for this greater initial stiffness may be due to short-range stiffness (the increased stiffness associated with the initial elongation of muscle [45]) as there is likely some level of resting muscle tone in the conscious participant during laxity testing (supported by greater anterior knee laxity in the unconscious versus conscious ACL patient [46]). Thus, increased 0–20 N stiffness may be present in those with increased short-range stiffness, in turn this may be indicative of heighten quadricep force production that has been associated with greater anterior shear forces [47].

Although anterior tibial translation measures were not included in the current analyses, it was previously reported that there was no relationship between peak anterior shear forces and peak anterior translations in healthy females during drop landings [48]. Given that peak accelerations in the current study were calculated as the second derivative of anterior tibial translation, this suggests that translation and acceleration may be unique sources of information with regard to knee shear forces experienced during weight acceptance. It is acknowledged that the current study obtained translation and the resultant tibial acceleration data during simulated weight bearing whereas the previously mentioned study utilized biplane fluoroscopy to assess translations during actual landings [48]. It is possible that in the previous study that the role of active stabilizers [49] and larger impact forces [20] may have affected relationship of shear force to tibial translation, mitigating the influence of the passive restraint system.

Whereas the focus of the current investigation was the ability of combined measures associated with passive restraint of anterior tibial translation at the knee joint to predict knee shear force, 81.9% of the variance in peak knee shear force was still unaccounted for in our female model. It is accepted that other biomechanical variables are highly related to knee shear forces as a model including quadriceps activation, peak posterior ground reaction force, knee flexion (external) moment, knee flexion angle and sex combined to accounted for 86% of the variance of anterior shear force in stop jumps [10]. Similarly, knee flexion excursion, hip flexion excursion, knee extensor moment and quadricep activation post ground contact collectively predicted 53.8% of the variance in anterior shear force in drop landings [31]. Although a multitude of biomechanical factors can better predict the variance in peak shear forces, the current investigation demonstrates that measures associated with passive restraint of anterior tibial translation at the knee joint do play a role with regard to the magnitude of shear force experienced. Thus, future work that intends to better understand contributing factors to high risk biomechanics should include the investigations that include both active and passive restraint mechanics.

It was somewhat surprising as to the discrepancy in the results of the sex specific shear prediction models. The ability of measures associated with passive restraint of anterior tibial translation at the knee joint to significantly predict landing shear forces in females only potentially suggests that females may rely more on the passive restraint to help stabilize the knee during dynamic activity. Such an idea of increased ligament dependence is supported by the finding of greater ACL strain in female than male knees from height and age matched donors that was attributed to joint morphology that included smaller female ACL cross-sectional area and greater lateral posterior tibial slope [50]. Collectively this suggests that it may be especially important for females to adopt landing strategies to decrease shear forces and protect the passive restraints.

Limitations certainly exist for the current investigation. Although skin based markers were used in the current investigation, it is acknowledged that such methods are prone to movement artifact [51] and [52]. However, our reported consistency and precision of variables derived from skin mounted kinematic markers (tibial acceleration and knee shear) demonstrate our ability to reliably and carefully acquire this data. Additionally the use of a convenience population of healthy participants can be considered a limitation in understanding ACL injury mechanism [48]. Finally, while we attempted to ensure that the subject was as relaxed as possible prior to the 40% WB trials and demonstrated muscle onset times at or after the time of peak anterior tibial acceleration, it is impossible to fully separate the contributions of the active and passive structures to the observed tibial accelerations as low level hamstring activity has the potential to control anterior tibial translation [53].

5. CONCLUSIONS

This study demonstrated measures associated with passive restraint of anterior tibial translation at the knee joint are predictive of anterior knee shear loads during landing in females but not males. Specifically, the combination of greater tibial acceleration during early axial load along with greater initial and lesser terminal anterior stiffnesses predicted increasing anterior knee shear forces. While these passive restraint measures are often considered non-modifiable, recent data suggests that the surrounding muscle mass may contribute to these passive restraint capabilities [54], which in turn suggests that they may be modifiable to some extent. This is further supported by the fact that males (who have greater muscle mass) did not demonstrate these relationships. Thus, while the current findings suggest that screening for measures of passive restraint of anterior tibial translation might be an additional tool useful in indentifying individuals who display at-risk biomechanics, they also provide a focus for ACL prevention strategies. To further the work in understanding ACL injury mechanisms along with optimizing injury prevention programs, future work should investigate the combined contribution of passive and active restraints to high-risk mechanics associated with ACL injury.

6. CONFLICT OF INTEREST STATEMENT

The authors have no commercial or proprietary interest in any device, equipment, instrument, or drug that is the subject of the article in question.

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