Published in final edited form as:
Clin Biomech (Bristol, Avon). 2011 December ; 26(10): 1019–1024. doi:10.1016/j.clinbiomech.2011.06.010.

Relationship of Knee Shear Force and Extensor Moment on Knee Translations in Females Performing Drop Landings: A Biplane Fluoroscopy Study

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Abstract

Background—Research has linked knee extensor moment and knee shear force to the non-contact anterior cruciate ligament injury during the landing motion. However, how these biomechanical performance factors relate to knee translations in vivo it is not known as knee translations cannot be obtained with traditional motion capture techniques. The purpose of this study was to combine traditional motion capture with high-speed, biplane fluoroscopy imaging to determine relationships between knee extensor moment and knee shear force profiles with anterior and lateral tibial translations occurring during drop landing in females athletes.

Methods—15 females performed drop landings from a height of 40 cm while being recorded using a high speed, biplane fluoroscopy system and simultaneously being recorded using surface marker motion capture techniques to estimate knee joint angle, reaction force and moment profiles.

Findings—No significant statistical relationships were observed between peak anterior or posterior knee shear force and peak anterior and lateral tibial translations; or, between peak knee extensor moment and peak anterior and lateral tibial translations. Although differences were noted in peak shear force ($P = 0.02$) and peak knee extensor moment ($P < 0.001$) after stratification into low and high shear force and moment cohorts, no differences were noted in anterior and lateral tibial translations (all $P \geq 0.18$).

Interpretation—Females exhibiting high knee extensor moment and knee shear force during drop landings do not yield correspondingly high anterior and lateral tibial translations.

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Keywords

Biomechanics; ACL injury risk factors; Neuromuscular Training; ACL Injury Prevention

Introduction

The annual rate of injury to the anterior cruciate ligament (ACL) is reported to be above 1 in 3000 in the United States with a large proportion (~70%) of ACL injuries non-contact in nature (Boden et al., 2000, Miyasaka et al., 1991).

Considerable research has demonstrated that loading of the ACL is partly determined by the force in the muscles that cross the knee. *In vitro* research has shown that the force in the muscles that produce knee extension moment affect ACL loading and under certain conditions can cause ACL damage (Renstrom et al., 1986, Durselen et al., 1995, Demorat et al., 2004). Recent modeling and simulation studies have also shown that the ACL is loaded when the shear force applied to the tibia is in the anterior direction; indicating that the quadriceps muscles, via the patellar tendon, are contributors to ACL loading during dynamic activities such as walking (Shelburne et al., 2004, Shelburne et al., 2005).

Human performance studies utilizing traditional surface marker motion capture techniques have also attempted to isolate and identify biomechanical performance factors, including knee shear forces and extensor moments, which may be related to non-contact ACL injury (Chappell et al., 2007, Chappell et al., 2002, Decker et al., 2003b, Huston et al., 2000, Kernozek et al., 2005, Yu et al., 2006, Yu et al., 2005). Chappell et al. (2002) showed that female recreational athletes had increased peak anterior knee shear force and peak knee extension moment and decreased knee flexion angle during landings in comparison to their male counterparts. Yu et al. (2005) also showed that the peak knee extension moments and peak anterior knee shear forces are significantly correlated to each other, supporting their view that that anterior knee shear force and knee extension moment are indicators of ACL loading during dynamic tasks such as the drop landing which evokes large eccentric quadriceps forces. Thus, based on the collective summary of many of these *in vivo* performance studies, as well as epidemiological data that shows relatively few combined anterior cruciate ligament/medial collateral ligament injuries are observed across the non-contact anterior cruciate ligament injured athlete populations (Majewski and Kentsch, 2002, Miyasaka et al., 1991), Yu and Garret (2007) and others have reasoned that a sagittal plane, non-contact anterior cruciate ligament injury mechanism is plausible. However, the results of these articles should be treated with caution. First, the anterior shear force described in these articles represents only a portion of the shear forces acting at the knee and does not include the contribution of joint reaction, muscle, and ligament. Second, it has been questioned whether the anterior shear force quoted in many of these articles is not in fact the reaction load due to the posterior shear force applied to the tibia and thus would represent the net force applied *by* muscles, ligaments and joint contact, and not the joint shear force applied to the tibia (Van Den Bogert and Mclean, 2006, Shin et al., 2007).

While estimates of external knee shear forces and moments can be calculated *in vivo* via inverse dynamics, the determination of actual tibio-femoral translations *in vivo* using traditional motion analysis techniques is difficult due to the soft tissue surrounding the knee joint (Leardini et al., 2005). Thus, investigators have been unable to directly determine what affect external knee shear forces and extensor moments have on knee translations *in vivo* during dynamic activities such as drop landing. Understanding these relationships *in vivo* is critical to understanding how the knee functions under high demand activities; and, could provide further insight into the mechanism of the non-contact ACL injury.

The objectives of this study were to: 1) utilize traditional human motion analysis techniques in combination with high-speed, biplane fluoroscopic imaging to describe the relationships between knee shear force and knee extensor moment profiles with anterior and lateral tibial translations (ATT and LTT) during the drop landings; and, 2) to measure the ATT and LTT kinematic differences between subjects exhibiting higher versus lower knee shear force and knee extensor moments during drop landings. Based on the "sagittal plane, quadriceps load induced injury" theory (Yu and Garrett, 2007) of the non-contact ACL injury, we hypothesized individuals exhibiting high external knee shear forces and/or knee extensor moments would exhibit larger ATT values during the drop landing.

Methods

Subjects and Landing Protocol

Fifteen female recreational athletes (Age 26.1 years (SD = 6.3); Height 167.9 cm (SD = 6.3), Body Mass 58.2 Kg (SD = 5.2) volunteered to participate in this study. These participants had no history of lower extremity injury. All subjects provided written informed consent approved by the Vail Valley Medical Center in accordance with the National Institutes of Health's guidelines.

The subjects were instructed to perform a drop landing maneuver by stepping off a 40 cm high platform onto a force plate and then immediately performing a peak height vertical jump (Myer et al., 2007). No other verbal or visual instruction as to the landing style was given prior to testing. All subjects were provided tight-fitting shorts and top and standardized court shoes (Turntec, model no. TM08061). After a 5-minute warm-up period, the subjects completed 10 landing trials in which a single trial for each subject was collected by fluoroscopy for analysis. The foot of the dominant limb landed directly onto a force plate (Bertec Corp., Columbus, Ohio, USA) fixed to the laboratory floor. The force plate data was sampled at 1200Hz and synchronized with the fluoroscopy video images such that the specific fluoroscopy frame of ground contact could be determined.

The subjects then completed a slow (2 sec), unloaded knee extension motion from a seated position (hip angle at 90 degrees) starting at a knee flexion of 90 degrees to the fully extended position. The unloaded knee extension trial was recorded only with fluoroscopy and used as a reference for the translations and rotations of the tibia relative to the femur measured during the landing trials. Also, the location of the tibia and femur at full extension was used to define the zero reference position of the bones of the knee (Blankevoort et al., 1990).

Surface Marker Motion Analysis Data Capture

Lower extremity, three-dimensional kinematics and kinetics were collected using a 10-camera system (Motion Analysis Corporation, Santa Rosa, CA, USA) sampling at 240 Hz. The 3D kinematics of each trial were captured by securing 37 retroreflective spherical (10 mm diameter) markers to anatomical landmarks on each subject to produce a standard, 3-marker-per-segment configuration (Kernozek et al., 2005). Marker trajectories and force data were filtered at the same frequency (20Hz) using a fourth order Butterworth filter (Bisseling and Hof, 2006). An initial standing "neutral" (erect) standing position of the subject in the anatomical position was also collected to establish relative angle adjustments for the landing data set. Joint angular position, velocities, and accelerations were calculated from the filtered 3D marker coordinate data and estimates of the (external) joint forces and moments were calculated using The Motion Monitor software (The Motion Monitor, Version 7.79, Chicago, IL, USA) by combining the kinematic and force plate data with anthropometric data using an inverse dynamics solution.

Description and calibration of the biplane fluoroscopy system

The biplane fluoroscopy system consists of two commercially available BV Pulsera c-arms with 30 cm image intensifiers (Philips Medical Systems, Best, Holland) coupled with two synchronized, high-speed, high-resolution (1024 × 1024) digital cameras (Phantom V5.1, Vision Research, Wayne, NJ, USA) which were interfaced with the image intensifiers of the fluoroscopy systems.

Image distortion was corrected and the focus position of each fluoroscopy system and the 3D relationship between the two fluoroscopy systems were calculated with previously described techniques (Torry et al., 2010, Garling et al., 2005). The transformation from the distorted image to the corrected image was determined and applied to all subsequent images using Model-Based Roentgen Stereophotogrammetric Analysis (MBRSA) software (Medis Specials, BV, Leiden, The Netherlands)(Hurschler et al., 2009).

Collection of in vivo landing data using biplane fluoroscopy

For each subject, a supine, high-resolution (voxels: 0.7×0.7×0.5mm), static bone CT scan of the knee (12cm above and below the joint line) utilizing an Aquilion 64 (Toshiba America Medical Systems, Tustin, CA, USA) was obtained. For each landing, biplane fluoroscopy data was collected for 1.0 s at 500 frames/s with a shutter speed of 1/2000 of a second. The x-ray generators were operated in radiographic mode at 60mA and approximately 60kV.

Fluoroscopy data reduction

Three-dimensional bone geometry reconstruction of the femur and tibia/fibula from CT data were extracted using commercial software (Mimics, Materialize, Inc, Ann Harbor, MI, USA). The origin of the femoral coordinate system was placed at the midpoint between the medial and lateral femoral condyles on the center line of a cylinder fitted to the medial and lateral posterior condyles. The medial-lateral axis was placed on the axis of the cylinder, and the inferior/superior axis aligned with the long axis of the femoral shaft (Blankevoort et al., 1990). The origin and alignment of the tibial coordinate system was assigned to the femoral coordinate system at full extension (i.e., 0 deg) as recorded by the fluoroscopy system in the knee extension trial (Torry et al., 2011, Torry et al., 2010). ATT was measured anterior to the tibia by the anterior distance between the femoral and tibial coordinate frames. Similarly, LTT was measured lateral to the tibia by the lateral distance between the femoral and tibial coordinate frames (Torry et al., 2011, Torry et al., 2010).

Determination of the bone poses from the biplane fluoroscopy data were performed using MBRSA (Medis Specials, Leiden, The Netherlands). For each frame, both inner and outer contours of the femur and tibia/fibula were semi-automatically extracted from the biplane fluoroscopy images. A fully-automatic 6 degree-of-freedom optimization algorithm was used to determine the pose, which optimally matched the detected contours with the projected contours from the imported bone geometries. From these bone pose sequences, knee kinematics were calculated using methods described by Grood and Suntay (Grood and Suntay, 1983). Following the examples of precision motion measurements taken from cadaver knees (Ishibashi et al., 1997), tibial translations and rotations measured from each subject during landing were referenced to the unloaded knee extension motion data.

Kinematic accuracies for tracking tantalum beads and bones using this biplane fluoroscopy system were determined. Our accuracy and precision of tracking 1.0 mm diameter tantalum markers were -5.5 μm and 32.5μm, respectively (Braun et al., 2010). However, in the present study implanting tantalum beads in the volunteers was not practical and only bones were tracked. Based on the high accuracy of tracking tantalum beads, bead tracking can be used as the gold standard for bone tracking methods. Therefore, tantalum beads were placed in a cadaveric knee specimen (3 beads per bone) and the specimen was manually ranged and cycled through full flexion-extension, varus-valgus and internal-external rotation. The mean ± 1 standard deviation of the differences in joint kinematics between those determined by tantalum beads and those by bone tracking were 0.2 mm (SD = 0.3), -0.1 mm (SD = 0.1), -0.05 mm (SD = 0.1) in translations; and, 0.1° (SD = 0.1), 0.3° (SD = 0.2), 0.1° (SD = 0.3) in rotations (Torry et al., 2010). To validate methods specific to this high impact study protocol a cadaver drop test was conducted and the data has been previously described (Torry et al., 2011). In brief, the mean and 1 standard deviation of the differences in joint kinematics between those determined by tracking tantalum beads and those by bone tracking of a cadaveric knee dropped from a 40 cm height were 0.07 mm (SD = 0.7), 0.1 mm (SD = 0.6) in medial-lateral and anterior-posterior translations; and, 0.09° (SD = 0.4), 0.1° (SD = 0.05), 0.1° (SD = 0.8) in flexion, varus-valgus, internal-external rotations, respectively. These values are consistent with previous work using similar technology (Anderst et al., 2009, Bey et al., 2008).

Statistical Analysis

Linear regression analyses were conducted to determine if peak anterior knee shear force and/or peak knee extensor moment are significant predictors of peak ATT and/or LTT during the drop landing. In order investigate differences in ATT or LTT in individuals possessing higher and lower peak anterior knee shear force and/or peak knee extensor moment profiles, the subject pool (N = 15) was subdivided into upper (n=5) and lower (n=5) cohorts according to each of these variables. The peak ATT and LTT between these upper and lower subdivisions were then compared utilizing an independent t-test ($\alpha = 0.05$).

Results

Regression Analyses

Means and 1 standard deviation (SD) for peak knee shear force and peak knee extensor moment as measured by motion analysis system as well as peak ATT and LTT as measured by the biplane fluoroscopy system are presented in Table 2.

The subjects landed with resultant ground reaction forces of 1315.2 N (SD = 629.3 N) (22.3 N/Kg, SD = 9.4 N/Kg); and as measured by the motion capture system, exhibited knee (ground contact) flexion angles of 14.6° (SD = 4.6°), and experienced peak knee flexion angles of 83.6° (SD = 9.2°).

The knee shear force exhibited a positive value (anterior shear force) just after ground contact and then transitioned to a negative value (posterior shear force). There was no correlation of the peak ATT value and the peak anterior knee shear force occurring during the first 0.0-0.018 sec after ground contact ($P = 0.73$, Figure 1A). No significant relationship was observed between peak posterior knee shear force measured by the motion capture system and peak ATT or LTT measured by the fluoroscopy system ($P = 0.40$, $P = 0.53$, respectively, Table 1, Figure 1A).

No significant relationship was observed between peak knee extensor moment measured by the motion capture system and peak ATT or LTT measured by the fluoroscopy system ($P = 0.89$, $P = 0.71$, respectively; Table 1, Figure 1B).

Low versus High Risk Group Comparisons

When the group (N = 15) was subdivided into upper (n=5) and lower (n=5) cohorts based on each individual's peak knee shear force value, significant differences in the peak knee shear force as determined by the motion capture system were noted ($P = 0.02$, Table 2). However, no differences in peak ATT or LTT were observed (*P* = 0.18, *P* = 0.66, respectively). This non-significant finding remained after the knee shear force profiles were normalized to subject body mass (N*m/Kg).

When the group was subdivided into upper and lower cohorts based on each individual's peak knee extensor moment value, significant differences in the peak knee extensor moment as determined by the motion capture system were noted (*P* < 0.0001; Table 2). However, no differences in peak ATT or LTT were observed (*P* = 0.34, *P* = 0.99, respectively). This non-significant finding remained even after the knee extensor moment profiles were normalized to subject body mass (N*m/Kg).

Discussion

Yu and Garrett (2007) and others have reasoned that sagittal plane biomechanics play a key role in the non-contact ACL injury mechanism (Chappell et al., 2007, Chappell et al., 2002). Specifically, these authors have suggested that the peak external knee shear force and/or the peak knee extensor moment during landing or a deceleration motion may be the mechanical factors that incite the non-contact ACL injury. The present study found no relationship between peak knee shear (anterior or posterior) force or peak knee extensor moment with peak ATT or LTT during the drop landings. These non-significant findings in the knee translations remained even though significant differences in peak knee shear force and knee extensor moments were identified using the motion capture system to stratify the individuals into upper and lower bounds based on higher versus lower peak knee shear force and peak knee extensor moment values.

The sagittal plane injury theory is based largely on the premise that the ACL provides ~80-85% of the total restraint to ATT when the knee is at low (20-30°) flexion angles(Markolf et al., 1976, Butler et al., 1980); and, anterior knee shear forces are expected to place loads on the ACL during deceleration motions such as landing from a jump *in vivo* (Sell et al., 2007, Chappell et al., 2002, Yu et al., 2002). Moreover, substantial cadaveric and computer modeling evidence has shown that the patellar tendon can cause ATT, increased ACL strains and even damage when the quadriceps produce high loads (Markolf et al., 1995, Renstrom et al., 1986, Demorat et al., 2004). The lack of association in ATT or LTT to knee extensor moment and/or external knee shear forces was in contrast to what was expected to occur at the knee during landings based on previous *in vivo* landing studies (Sell et al., 2007, Chappell et al., 2002, Yu et al., 2002); but, these results are not entirely unexpected based on modeling efforts (Pflum et al., 2004, Shin et al., 2007). In 2004, Pflum et al, showed that the pattern of ACL force in landing cannot be explained by the mechanism of quadriceps force alone. The maximum force transmitted to the model's ACL resulted from a complex interaction between the patellar tendon force, the compressive force acting at the tibiofemoral joint, and the ground reaction force. The external shear force calculated here and by others using inverse dynamics does not include the force in the muscles, ligaments, and bony contact. We believe that this incomplete summation of shear force explains why no correlation was found between ATT and LTT and the external knee shear force of the subjects. In addition, knee shear force was primarily directed posterior for much of the landing motion (Figure 1). Shin et al. (2007) (Shin et al., 2007) noted that when posterior forces that replicated in vivo conditions were applied to their model, the peak ACL strain was reduced. Likewise, utilizing cadavers limbs dropped from a known height, Withrow et al., (2006) (Withrow et al., 2006) reported ACL strain was larger for the impulsive compression loading in valgus *and* flexion compared to loading in isolated flexion.

Soft landings are thought to allow for less ground reaction force and greater energy dissipation through greater ranges of motion at the knee (Decker et al., 2003a) and are thus taught in many ACL prevention programs. If the knee is more fully extended during ground contact, the patellar tendon will be more anteriorly inclined relative to the long axis of the tibia. This, combined with a large quadriceps force developed in the eccentric contraction, causes a large anterior shear force to be applied to the lower leg (Decker et al., 2003b, Sell et al., 2007). An increase in anterior shear force is suggested to increase ATT, thereby causing an increase in ACL strain (Yu and Garrett, 2007). In the present study, no verbal cues were given to encourage either a soft or stiff landing. Instead, the females were allowed to land in a self-selected manner and to complete a peak vertical jump as described by previous ACL injury screening studies (Hewett et al., 2005, Myer et al., 2007). The females in this study landed with knee (contact) flexion angles that were ~10° less flexed than reported by previous studies (Decker et al., 2003a, Sell et al., 2007, Yu et al., 2006); indicating a more erect landing style which should have increased our ability to observe ATT in these subjects. Thus, how soft versus stiff landings actually affect knee translations need to be more fully understood before recommendations regarding "how to land safely" are promoted.

The magnitudes of ATT and lateral translation found in this study are also reasonably close to the values in previous dynamic, *in vivo* reports (Li et al., 2009, Tashman et al., 2004). Li et al. reported peak ATT ranging from 4.23 mm - 4.89 mm and peak medio-lateral translations ranging from 5.58 mm - 5.80 mm during gait (speeds of 1.5 - 3.0 m/s). Extrapolating data published by Tashman et al. 2004 revealed that the healthy limbs experienced approximately 8 mm of A/P translation and approximately 1 mm of lateral translation when jogging downhill (10° slope) at 2.5 m/s. Deneweth et al. 2010, (Deneweth et al., 2010) using similar technology recently reported 12 mm of anterior tibial translation in the healthy limb of individuals with contralateral ACL repair.

Current knowledge regarding ACL stiffness and load mitigation are based primarily on *in vitro* investigations which are limited in physiologic load applications (Woo et al., 1991, Woo et al., 1992). Biplane fluoroscopy offers methods to overcome the inherent errors associated with traditional motion capture studies in understanding knee translations in vivo during dynamic tasks; as well bridge the insufficient physiologic load deficits gap employed by current in vitro technology in assessing joint stability. In such, biplane radiographic techniques in combination with traditional motion capture methods show promise in investigating how the underlying neuromuscular control mechanics are employed and are altered and/or related to knee joint function as the functional demand of the knee increases through both internal and external forces and moments.

Within the scope of this study, limitations are acknowledged. This study utilized a group of healthy females in a controlled setting for experimentation and as such, is a limitation in understanding knee motion during the actual non-contact ACL injury mechanism. Moreover, only double-legged landings were evaluated; single-limb landings may increase or otherwise alter the external loads and provide different results. Additionally, the subjects were not instructed to land softer or stiffer but rather in self selected manner. Enforcing the subjects to land with a more "stiff" landing profile emphasizing less knee flexion may elicit greater ATT and LTT values. The subjects in this study exhibited peak knee shear forces at 8 ± 3 N/Kg; which are similar to previous studies (Kernozek et al., 2008, Sell et al., 2007, Yu et al., 2006) but it is possible that these values did not reach sufficiently high thresholds to elicit significant ATT. The review process raised issues with our selected filtering methodology and we acknowledge that the inherent signal-to-noise error in the motion capture technique and our selected filtering parameters will have caused dampening of the peak shear force and moment data reported; and, the time series data for traditional motion capture (240 Hz) and fluoroscopy data (500Hz) were synchronized beginning at the point of ground contact.

After this point, however, the data do not always match time point for time point with the data synchronize every 50 msec with a maximum offset of 1 msec. A better choice of motion capture collection frequency would have precisely synchronized the data collected. Even so, both collection frequencies were high to characterize the changes that occur during landing. This was our goal given the fact that the focus was primarily on assessing relationships that exist between peak values of ATT and peak knee moment/joint reaction force.

Prospective cohort studies commonly use epidemiological research designs for determining injury and risk factors. These studies, however, are descriptive in nature and do not examine the cause-and-effect relationship between identified risk factors and the injury. The result of this study suggest that this may also be true for performance based studies where inverse dynamic analyses are utilized to isolate and label biomechanical factors as predictive or injurious. Without a good understanding of the injury mechanisms, the risk factors for sustaining non-contact ACL injuries identified from epidemiological or biomechanical performance studies could be misinterpreted and lead to the selection of non-optimal injury prevention programs. Thus, although high knee extensor moments and anterior knee shear forces have been associated with ACL loads in modeling and cadaveric studies, these parameters did not produce correspondingly high ATT values *in vivo* in the present study.

Conclusion

Although several studies have implicated high anterior knee shear forces and extensor quadriceps moment in causing strain on the ACL during landing motions, the present study found no association between peak anterior knee shear force or peak knee extensor moment with peak ATT or LTT during the drop landing in females athletes.

Acknowledgments

This study was funded by the National Institutes of Health (AR39683 to PI: Savio L-Y. Woo) and the Steadman Philippon Research Institute. The authors wish to thank Andrea North, Elizabeth Hageman and Nicole Pinwell for help with data collection and the reduction process. Medis Specials is acknowledged for use of MBRSA software.

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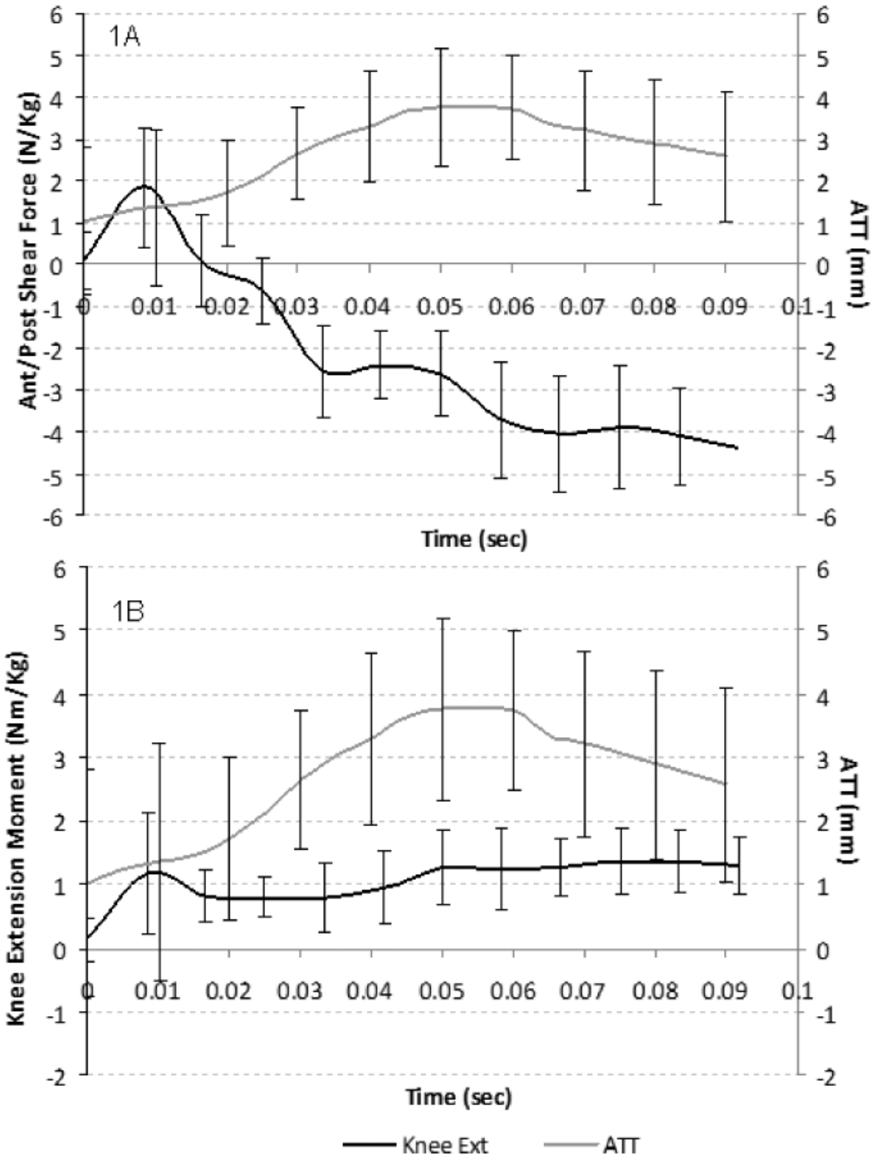

FIGURE 1.

(1A): Time series plot of ATT (grey line) derived from fluoroscopy and knee shear force estimated from the motion capture system and the inverse dynamics process (black line). Positive values represent an anteriorly directed shear force. (1B) Time series plot of ATT derived from fluoroscopy (grey line) and the external knee extensor torque curve estimated from the motion capture system and the inverse dynamics process (black line). Positive torque values represent external knee extensor torque values. Deviations depicted in both time series data sets represent +/- 1 SD from the mean.

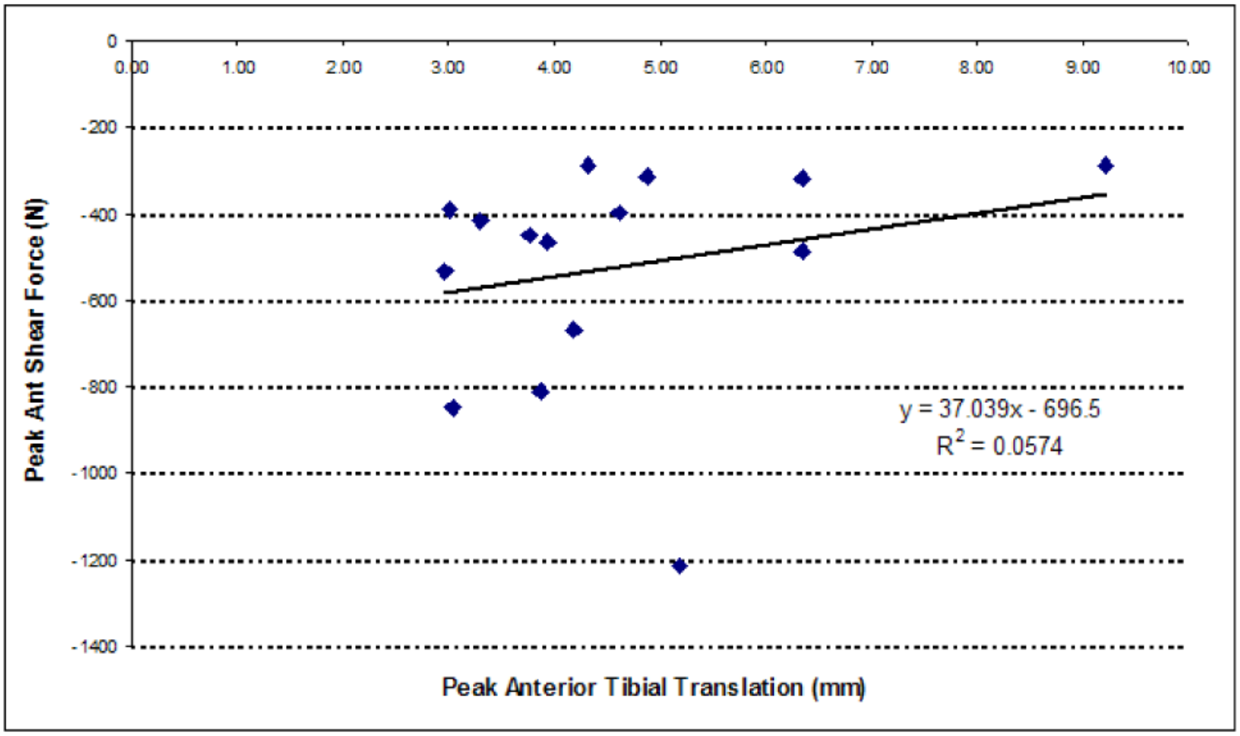

FIGURE 2.
Regression plot and equation of peak knee shear force versus anterior tibial translation.

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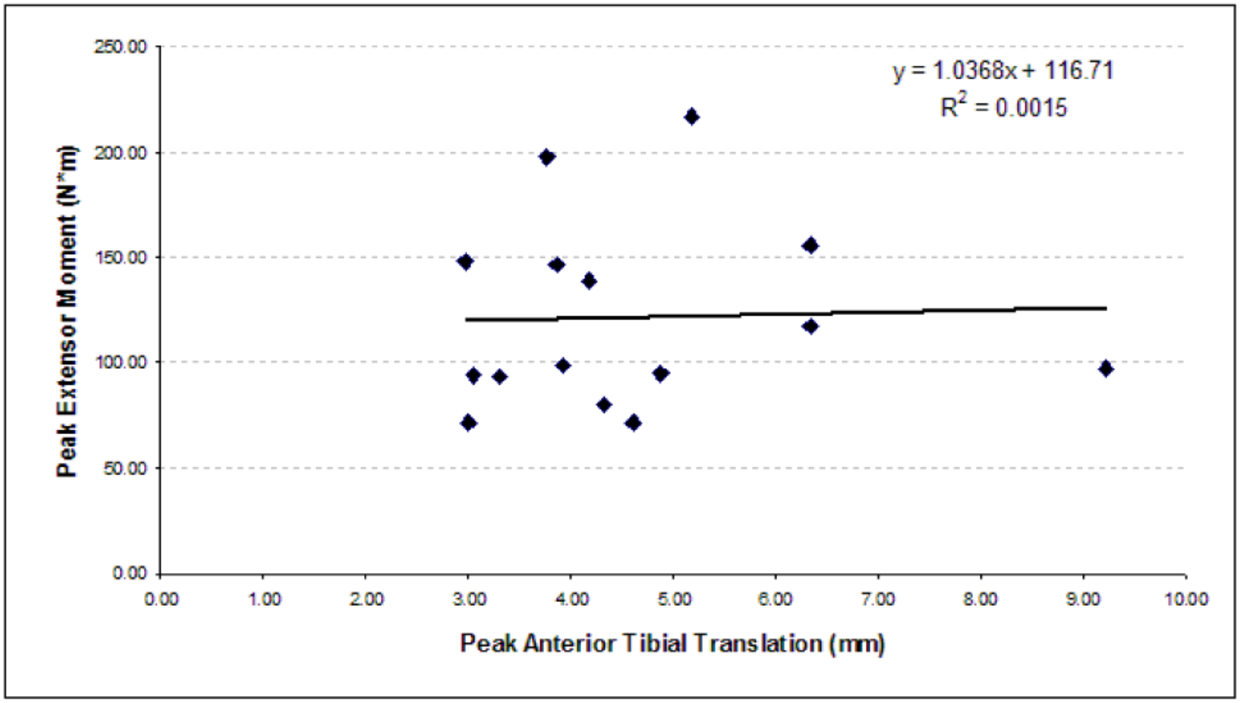

FIGURE 3.
Regression plot and equation of peak knee extensor moment versus anterior tibial translation.

Table 1

R-square values for peak shear force, peak knee extensor moment, peak anterior tibial translation and peak lateral translation.

Variables	r^2	p-value
Peak ATT (mm), Peak Shear (N/Kg)	0.06	0.4
Peak ATT (mm), Peak Ext Torque (N*m/Kg)	0.005	0.79
Peak ATT (mm), Peak Ext Torque (N*m)	0.002	0.89
Peak ATT (mm), Peak Shear Force (N)	0.06	0.4
Peak LTT (mm), Peak Shear Force (N/Kg)	-0.03	0.53
Peak LTT (mm), Peak Ext Torque (N*m/Kg)	-0.02	0.64
Peak LTT (mm), Peak Ext Torque (N*m)	-0.01	0.71
Peak LTT (mm), Peak Shear Force (N)	-0.03	0.51

Table 2

Group means (±1SD) for peak shear force, peak knee extensor moment, anterior tibial translation and lateral tibial translations. Group means (±1SD) for the low and high stratification also provided within each column.

	Peak Shear F (N)	**Peak ATT (mm)**	**Peak LTT (mm)**
Group			
Mean (±1SD)	-526.2 (258.3)	4.6 (1.7)	0.4 (1.0)
Low Knee Shear F Group (n = 5)			
Mean (±1SD)	-320.4 (258.3)	5.6 (2.4)	0.4 (1.0)
High Knee Shear F Group (n = 5)			
Mean (±1SD)	-814.9 (254.4)	3.9 (0.9)	0.7 (1.1)
Comparison of Low vs High Knee Shear F Groups	$p = .02$	$p = .18$, $\beta = .14$	$p = .66$, $\beta = .07$
	Peak Ext Moment (Nm)	Peak ATT (mm)	Peak LTT (mm)
Group			
Mean (±1SD)	121.5 (44.4)	4.6 (1.7)	0.4 (1.0)
Low Knee Ext. Moment Group (n = 5)			
Mean (±1SD)	82.3 (11.3)	3.7 (0.8)	0.2 (1.3)
High Knee Ext. Moment Group (n = 5)			
Mean (±1SD)	172.8 (32.0)	4.4 (1.3)	0.2 (1.0)
Comparison of Low vs High Knee Ext Moment Groups	$p < .0001$	$p = .34$, $\beta = .17$	$p = .99$, $\beta = .05$

p-values represent t-tests between low and high group stratifications (peak shear, peak knee extensor moment) and within each column heading. When not significant, post-hoc power analysis were conducted (β = values).