Effect of axial load on anterior tibial translation when transitioning from non-weight bearing to weight bearing

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Abstract:

Background

While the application of compressive joint loads and thigh muscle activity are associated with anterior tibial translation in vitro, less is known during early load acceptance in vivo. We investigated the effects of increasing axial loads on anterior tibial translation and thigh muscle activity in healthy knees during transition from non-weight bearing to early weight bearing.

Methods

Participants (11 males, 11 females) underwent 20%, 40%, and 60% body weight acceptance trials at 20° knee flexion while electromagnetic sensors measured anterior tibial translation (mm), and surface electromyography recorded quadriceps and hamstring muscle onset times (ms) and amplitudes (% maximal voluntary isometric contraction). Repeated measures ANOVA compared values across loads. Pearson correlations examined relationships between anterior tibial translation and muscle onset times and amplitudes within each load.

Findings

As load increased, anterior tibial translation (Mean (standard deviation)) (20% = 4.7 (1.7) mm < 40% = 7.1 (1.9) mm < 60% = 8.8 (2.1) mm), and quadriceps (20% = 23.6 (14.9)% maximal voluntary isometric contraction <40% = 32.7 (11.8)% maximal voluntary isometric contraction <60% = 41.1 (13.5)% maximal voluntary isometric contraction) and hamstring (20% = 15.5 (15.7)% maximal voluntary isometric contraction <40% = 23.0 (16.4)% maximal voluntary isometric contraction <60% = 27.6 (19.1)% maximal voluntary isometric contraction) activation increased, while quadriceps (20% = 96.7 (28.4) ms > 60% 80.2 (21.8) ms) and hamstring (20% = 141.5 (65.0) ms and 40% = 126.3 (68.8) > 60% 107.6 (28.4) ms) onset times decreased ($P \leqslant 0.05$). There were no relationships between anterior tibial translation and muscle activation amplitudes ($R = 0.033$–0.294) or onset times ($R = -0.031$–0.374) ($P > 0.09$).

Interpretation

Greater axial loads near full knee extension during early weight acceptance result in greater anterior tibial translation, regardless of faster and stronger activation amplitudes. These findings support injury prevention programs aimed to reduce impact forces as they may in turn reduce anterior tibial translation and corresponding ligamentous strain during dynamic activity.

Keywords: Knee; ACL; Loading rate

Article:

1. Introduction

Injury to the anterior cruciate ligament (ACL) is often observed in non-contact mechanisms that occur during landing, cutting, deceleration, and foot strike with near full knee extension (Olsen et al., 2004). Further, it is thought that actual injury occurs very early in the loading of the limb (Olsen et al., 2004), with documented increases in ACL strain prior to foot contact during a single leg hop task (Cerulli et al., 2003). Thus it may be important to better understand tibiofemoral mechanics during the transitions from non-weight bearing to early weight bearing (Shultz et al., 2009). The integrity and resulting stability of the knee joint is maintained by both passive and active structures around the knee joint. As the ACL is the primary restraint to anterior tibial translation (ATT) (Butler et al., 1980), it is thought that factors contributing to excessive ATT may put the ACL at risk of injury (Shultz et al., 2006).

Although it is well understood that a weight bearing posture creates an increase in knee stiffness and greater anterior–posterior stability, ([Markolf et al., 1981] and Markolf et al., 1978 K.L. Markolf, A. Graff-Radford and H.C. Amstutz, In vivo knee stability. A quantitative assessment using an instrumented clinical testing apparatus, J. Bone Joint Surg. Am. 60 (1978), pp. 664–674. (137)[Markolf et al., 1978]) much less is understood about anterior–posterior tibiofemoral mechanics during the weight acceptance process. A study of robotic loading of cadaver knees suggested that axial loads create greater in situ forces in the ACL which implied that excessive axial loads may contribute to ACL injury (Li et al., 1998). Further it has been demonstrated that high axial loads are capable of rupturing the ACL (Meyer and Haut, 2008). Several cadaveric studies have demonstrated ATT in response to axial loads ([Giffin et al., 2004] and [Torzilli et al., 1994]). A modeling study of ACL function during drop function suggested that quadriceps force and compressive force acting at the tibiofemoral joint contribute greatly to the total load at the ACL (Pflum et al., 2004). Further, in vivo weight acceptance activities have been shown to create ATT ([Yack et al., 1994] and [Kvist, 2006]) as well as create ACL elongation (Hosseini et al., 2009). Taken together these studies demonstrate the need to better understand in vivo knee mechanics of the weight acceptance phase of activity.

It is well accepted that the knee extensor and flexor muscle groups contribute to changes in ACL strain ([Boden et al., 2000], [Renstrom et al., 1986], [Durselen et al., 1995], [Li et al., 1999] and [Markolf et al., 2004]). Cadaveric studies report that unopposed quadriceps muscle contraction near full knee extension is considered a contributing factor of ATT ([Li et al., 1999] and [Markolf et al., 2004]). Additionally, cadaveric studies have also demonstrated that quadriceps contraction applies an anterior shear force on the tibia through the patellar tendon ([Markolf et al., 1995] and [Li et al., 1999]). This shear force in turn may lead to an ACL injury when knee flexion angle is less than 30° when the hamstring muscle is not sufficient to provide posterior shear force ([Boden et al., 2000], [Hewett et al., 1996] and [Huston and Wojtys, 1996]). The specific role of thigh musculature in controlling or creating ATT as the limb transitions from non-weight bearing to weight bearing in vivo is poorly understood.

To our knowledge, only one study (Yack et al., 1994) has previously examined the effect of increasing axial compressive load on ATT when transitioning from a non-weight bearing to weight bearing condition in vivo. No difference in ATT was noted during progression from 25% to 100% BW loads (Yack et al., 1994). While compressive joint loads and thigh muscle activity are associated with the amount of ATT in vitro, less is known about these characteristics when transitioning from non-weight bearing to weight bearing in vivo. Therefore, the purpose of this study was to investigate the effect of increasing axial loads on ATT and quadriceps and hamstring muscle activation (onset timing and amplitude) in healthy knees during the transition from non-weight bearing to weight bearing. Our expectation was that increased axial loads would result in greater ATT during the transition.

2. Methods

Twenty-two subjects (11 females, 11 males; M (SD) 24.9 (3.2) years, 169.9 (7.9) cm, 68.3 (12.2) kg) were recruited from a university population. Participants had no history of knee ligament injuries. Before participation, all subjects reviewed and signed a consent form approved by the University's Institutional Review Board. We defined the dominant limb as the stance leg when kicking a ball and all data was collected from the dominant leg. Prior to testing, we collected demographic information such as sex, height, and weight.
Surface electromyographic (sEMG) data were taken from four muscle locations in the dominant limb: the medial and lateral quadriceps, and the medial and lateral hamstring muscles. Electrode placement was confirmed with manual muscle testing. Before attaching electrodes, we shaved and cleaned the skin with the 70% isopropyl alcohol preparation pad to reduce skin impedance. 10-mm bipolar Ag–AgCl surface electrodes (Blue sensor N-00-S; Ambu Products, Olstykke, Denmark; 44.8 × 22 mm diameter; skin contact size 30 × 22 mm) were placed over the prepared skin sites spaced 2 cm apart. The reference electrode was placed

over the proximal shaft of the tibia. Good signal fidelity and absence of cross talk for each electrode site was manually checked for each muscle group using the oscilloscope of the sEMG system. After attaching electrodes, subjects completed a 5-min warm-up on a stationary bike before data collection. We recorded sEMG signals via 16-channel Myopac system (Run Technologies, Mission Viejo, CA, USA) (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 to normalize sEMG data obtained during the transition from non-weight bearing to weight bearing. Subjects were 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°. The medial epicondyle of the dominant leg was aligned to the axis of rotation of the dynamometer. All straps for chest, waist, thigh, and ankle were used to prevent unwanted movement during MVIC tests. In preparation for obtaining the MVIC signals for the quadriceps and hamstring muscles, subjects first performed isometric muscle contractions at 25%, 50%, 75% and 100% of maximal effort. After 30-s rest, subjects performed three, 3-s maximal isometric contractions of quadriceps and hamstring muscles while collecting sEMG data. Consistent verbal encouragement was provided with each contraction to insure maximal effort.

Using explicit methods described elsewhere we assessed ATT with the Vermont Knee Laxity Device (VKLD) (Shultz et al., 2006). 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 (Uh et al., 2001). 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 (Shultz et al., 2009).

Subjects were placed in the VKLD (Fig. 1) and the foot was strapped to the foot cradle connected to a calibrated six degree-of-freedom 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. We used the centroid method for estimating the center of rotation of the ankle, knee, and hip joints and then digitized each joint center. After digitization of joint centers, the ankle and knee were flexed to 90° and 20° respectively and subjects were asked to relax their leg muscles. We checked the knee flexion angle (20°) both manually with a goniometer and with the electromagnetic position sensors. Once properly positioned in the VKLD, we applied three anterior to posterior forces 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, USA).

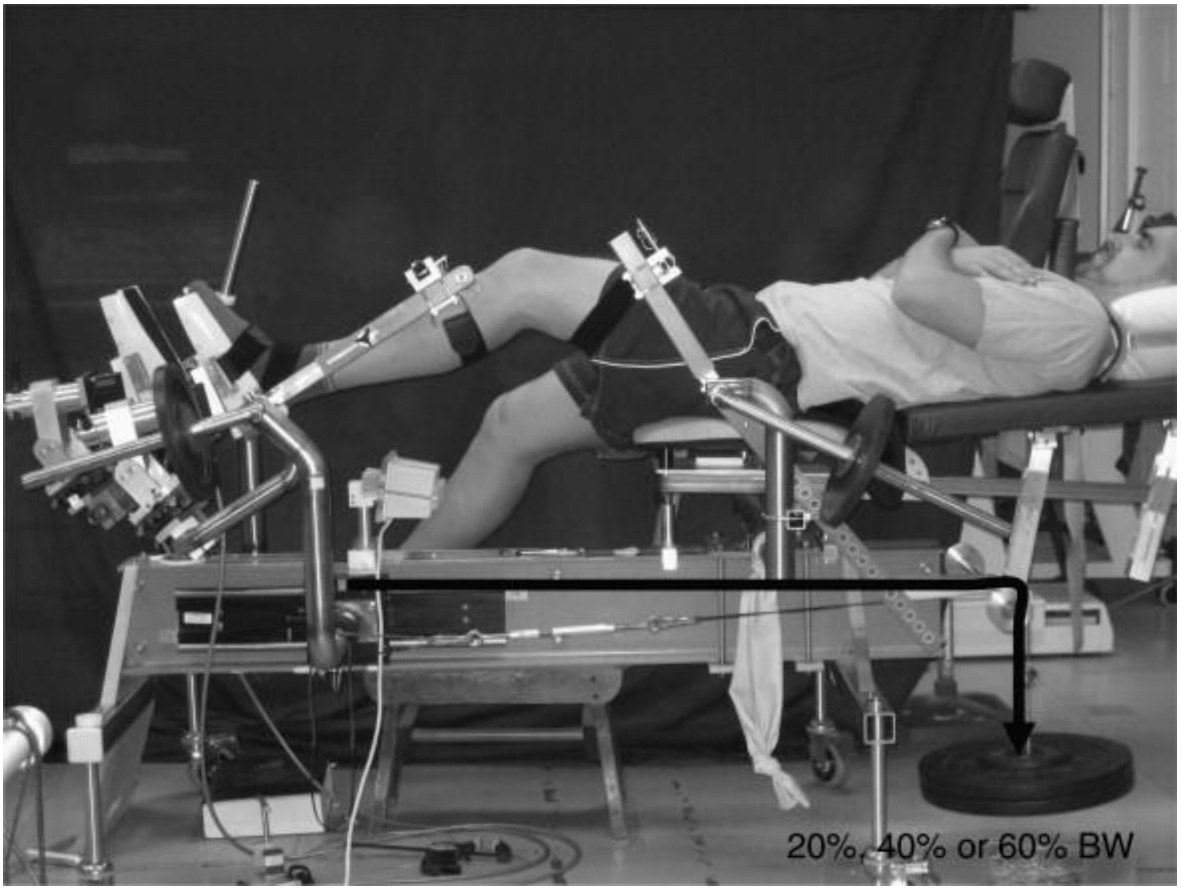

Fig. 1.
Vermont Knee Laxity Device (VKLD).

Prior to actual data collection, we performed 3–5 practice trials to familiarize the subject with the weight acceptance trials. Once the zero compressive and shear load were obtained, compressive loads equal to 20%, 40% and 60% of body weight (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 three magnitudes of the axial load were counterbalanced across subjects, with three trials performed at each load. Subjects were instructed to respond to the axial force as quickly as possible after the release of the weight and to try and maintain the initial knee position (20° knee flexion). We have previously established the between day measurement consistency (intraclass correlation coefficient (ICC) = 0.88) and precision (standard error of measurement (SEM) = 0.84 mm) of measuring ATT using this instrumentation and procedure (Shultz et al., 2006).

During each weight acceptance trial, sEMG signals were acquired at 1000 Hz from the medial and lateral quadriceps and hamstrings during the first 250 ms of the transition from non-weight bearing to weight bearing for each compressive load using the trigger sweep acquisition mode. Position data of the tibiofemoral joint were collected at 100 Hz from Electromagnetic position sensors attached to the thigh and shank (Mini Birds, Ascension Technologies, Colchester, VT, USA) and Motion Monitor software (Innovative Sports Training Inc., Chicago, IL, USA).

2.1. Data reduction and analysis

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″. ATT was then defined as the amount of anterior displacement of the tibia with respect to the patellar sensor from the initial position non-weight bearing to the peak axial compression force (Fig. 2). The average ATT value of the three trials at each load was used for analysis.

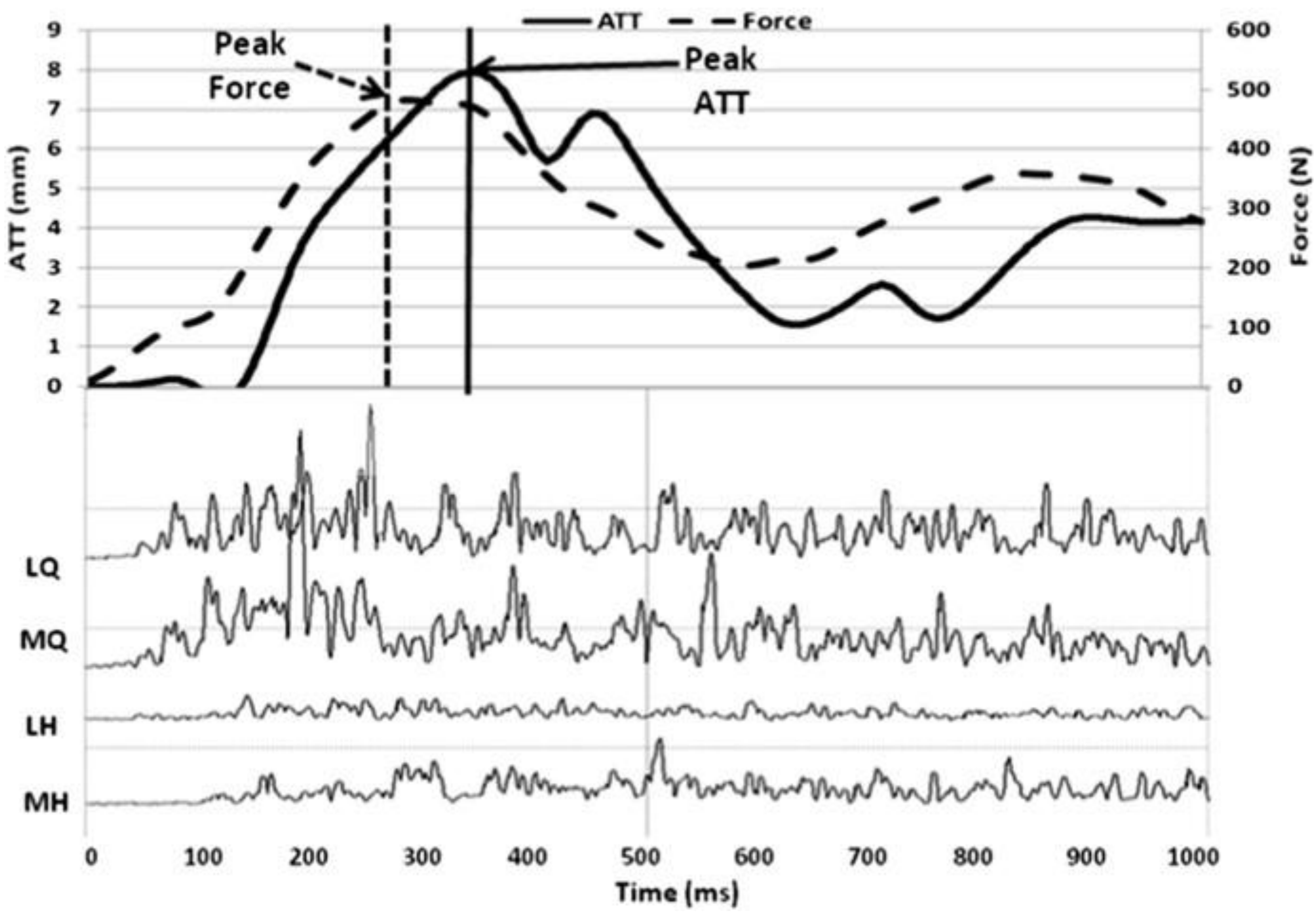

Fig. 2.
Representative graph showing anterior tibial translation (ATT), weight bearing force and quadriceps (LQ, MQ) and hamstring (LH, MH) muscle activation.

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) or 5 ms (weight acceptance) time constant (Shultz et al., 2009). Muscle onset (ms) was defined as the time when muscle activation exceeded 5 SD of the baseline sEMG signal for 10 ms. The mean RMS amplitude over the first 250 ms of weight acceptance was ensemble averaged over the three trials for each load and then normalized to the subject's MVIC peak RMS value and reported as a percentage of the MVIC (%MVIC). The medial quadriceps (MQ) and lateral quadriceps (LQ) values and medial hamstrings (MH) and lateral hamstrings (LH) were then averaged to give composite quadriceps and hamstrings values, respectfully.

2.2. Statistical analyses

One-way repeated measures analyses of variance (ANOVA) determined the effect of compressive load (20%, 40%, and 60% BW) on ATT and quadriceps and hamstring muscle activations (onset timing and amplitude) when transitioning from non-weight bearing to weight bearing. Significant differences were further examined using post hoc pair-wise comparisons with Bonferroni correction. Bivariate correlations examined the relationships between muscle onset timing and amplitude and ATT within each load. All statistical comparison was performed with the level of significance set at $P \leqslant 0.05$. Statistical analyses were conducted in SPSS for Window, version 15.0 version (SPSS Inc., Chicago, IL, USA).

3. Results

All data are reported as Mean (SD). The 20%, 40%, and 60% BW loading conditions resulted in peak loads of 29.0 (4.9)%, 55.5 (5.6)%, and 82.5 (7.5)% BW with loading rates of 1.2 (0.3), 2.0 (0.4), and 2.7 (0.4) BW s^{-1} and time to peak loads of 245.4 (23.6) ms, 282.0 (31.5) ms, and 305.1 (34.3) ms, respectively. Table 1 lists the means and standard deviations for each dependent variable across the three axial loads. Significant increases in ATT and quadriceps and hamstring muscle activation amplitude were observed across each of the three axial loading conditions ($P < 0.001$) and significant decreases in quadriceps and hamstring onset times were observed as load increased ($P \leqslant 0.05$) (Table 1). No significant correlations were noted between ATT and thigh muscle onset timing ($P \geqslant 0.09$, Pearson $R = -0.031$–0.374) or thigh muscle activation amplitude ($P \geqslant 0.185$, Pearson $R = 0.033$–0.294) within each load.

Table 1. Means (SD) of anterior tibial translation (ATT), quadriceps EMG, and hamstrings EMG for the 20%, 40%, and 60% BW loading conditions.

	20% BW load	40% BW load	60% BW load
ATT (mm) *	4.7 (1.7)	7.1 (1.9)	8.8 (2.1)
Quadriceps EMG onset (ms) **	96.7 (28.4)	89.4 (20.6)	80.2 (21.8)
Hamstrings EMG onset (ms) ***	141.5 (65.0)	126.3 (68.8)	107.6 (28.4)
Quadriceps EMG (%MVIC) *	23.6 (14.9)	32.7 (11.8)	41.1 (13.5)
Hamstrings EMG (%MVIC) *	15.5 (15.7)	23.0 (16.4)	27.6 (19.1)

* Significant ($P \leqslant 0.001$) increase across loading conditions (20% BW < 40% BW < 60% BW).
** Significant ($P \leqslant 0.05$) decrease in time to quadriceps muscle onset (20% BW > 60% BW).
*** Significant ($P \leqslant 0.05$) decrease in time to hamstrings muscle onset (20% BW and 40% BW > 60% BW).

4. Discussion

The primary finding was that during transition from non-weight bearing to weight bearing condition in vivo, ATT increased as the applied axial load increased from 20% to 40% to 60% BW. As would be expected, both quadriceps and hamstring activation amplitudes significantly increased while onset timing significantly decreased with an increase in axial compressive loads. However, there was no relationship between quadriceps and hamstring activation timing or amplitude with the amount of ATT within each load.

It is well established that applied axial loads are effective in increasing joint stability when anteriorly directed forces are applied to the knee ([Li et al., 1998] and [Markolf et al., 1981]). However, little is known of tibiofemoral kinematics during the weight acceptance phase. To our knowledge, only one study has previously examined the effect of increasing axial compressive load on ATT when transitioning from non-weight bearing to weight bearing condition in vivo reporting no significant increase in ATT from 25% BW (6.4 mm) to 100% BW (6.9 mm) (Yack et al., 1994). Our study measured ATT during the transition from non-weight bearing to weight bearing by simulating initial foot strike with the zero shear force knee positioned in slight knee flexion, as non-contact ACL injuries are often thought to occur during the early phase of landing, cutting, and jumping.

Additionally participants were instructed to relax prior to weight release to minimize the effect of pre-activation, thus allowing us to better understand the role of passive restraints. The current finding of increased ATT in response to increased axial load may differ from the Yack et al. as the previous work occurred in an upright posture with the subject allowed to preactivate the thigh musculature.

The result of axial loading causing ATT at shallow knee angles in this in vivo model is consistent with previous work on cadaver knees. In situ testing with a 200 N axial load resulted in a 2.7 mm ATT with the knee at 30° flexion (Giffin et al., 2004). Anterior loading of porcine knees in 30° of flexion with the addition of a 200 N axial load increased in situ forces of the ACL (Li et al., 1998). Additionally, it has been reported that a 444 N axial load combined with a 133 N quadriceps load resulted in substantial ATT (2.9–5.2 mm) in cadaver testing at shallow knee flexion angles (Torzilli et al., 1994). These values increased when the ACL was sectioned, suggesting that the ACL is of critical importance in controlling joint motion caused by compressive loads (Torzilli et al., 1994). Further, a cadaver study of ATT as axial load was increased to 1600 N demonstrated increasing ATT as load was applied (Liu-Barba et al., 2007). Although not statistically analyzed in their study, axial loads similar in magnitude to our current 20% and 60% BW loading conditions created increases in ATT from $\sim$3 mm to $\sim$5 mm with the knee in 30o flexion (Liu-Barba et al., 2007). Although we reported a $\sim$4 mm increase in ATT from 20% to 60% BW loading conditions, the larger increases in vivo may be due to the addition of quadriceps muscle forces acting across the tibiofemoral joint. Taken together it appears as though the ACL plays a significant role in restraining ATT during axial loading and that increased axial loading results in greater ATT and potentially greater ACL loading.

Although not a primary research hypotheses, we did perform secondary analyses to better understand the reactive loading of the joint and determine if increasing the load from 20% to 40% to 60% BW would cause greater amounts of knee flexion excursion, time to peak load, and resultant loading rates during the weight acceptance process in our experimental model. Previously using the same experimental model, a 10° increase in knee flexion excursion has been documented to underestimate ATT by approximately 0.5 mm (Shultz et al., 2006). Analysis of the current data revealed statistically greater ($P \leqslant 0.001$) knee flexion excursion, time to peak load, and loading rates as loading increased (Table 2). Given the small amount of knee flexion excursion that occurred, it is unlikely that the 2–3° total difference between loading conditions impacted the observed ATT differences. Although we found that muscle onset times decreased in response to increasing load (Table 1), the corresponding increase in time to peak load as loading magnitude increased is suggestive that participants took a longer time to generate adequate muscular forces to control the load as instructed. This in turn may have resulted in the 2–3° increase in knee flexion excursion across loading conditions. As increased loading rates have been suggested to be indicative of increased stress application to the limbs during a short time (Hargrave et al., 2003), the observed increase in loading rates provides further support that the observed increase in ATT was a function of the load applied to the lower extremity and not a function of the observed differences in sagittal plane rotations across loading conditions.

Table 2. Means (SD) of knee flexion excursion and loading rate for the 20%, 40%, and 60% BW loading conditions.

	20% BW load	40% BW load	60% BW load
Knee flexion excursion (°) *	3.6 (1.2)	5.0 (1.4)	6.0 (1.6)
Loading rate (BW s^{-1}) *	1.2 (0.3)	2.0 (0.4)	2.7 (0.4)
Time to peak load (ms) *	245.4 (23.6)	282.0 (31.5)	305.1 (34.3)

* Significant ($P \leqslant 0.001$) increase across loading conditions (20% BW < 40% BW < 60% BW).

As would be expected to control increasing axial load during weight bearing, there was a corresponding increase in both quadriceps and hamstrings activation. However, this increased activation did not result in increased restraint of ATT. It has been reported that quadriceps muscle force alone induces substantial ATT with the knee near full extension (less than 30° knee flexion), resulting in strains in the ACL in vitro ([Pandy and Shelburne, 1997] and [Torzilli et al., 1994]). Further, the addition of hamstring muscle force has been shown to significantly reduce ATT and ACL strain ([Draganich and Vahey, 1990], [Li et al., 1999] and [More et al., 1993]). Given the reactive nature of our task, it is unlikely that the observed increases in ATT as load increased were solely a function of our findings of more rapid onset of muscle activation and increased muscle activation amplitude as load increased. Additionally within each load, muscle onset timing and muscle activation amplitude were not significantly correlated to ATT. Thus other mechanical factors may likely be responsible for increased ATT with increased axial load.

One mechanical factor that may cause ATT is tibial slope geometry. For example, in cadaveric knees it has been noted that an increase in the anterior to posterior inferior tibial slope along with axial compression resulted in greater ATT (Giffin et al., 2004). Further, it has been documented that increased posterior tibial slope is associated with an increase risk of ACL injury (Brandon et al., 2006). Taken together it appears that posterior tibial slope geometry and magnitude of applied axial load may be contributing factors to ATT. However the current study did not perform MRI imaging to ascertain posterior tibial slope.

Findings from this study may provide further support for the rationale for ACL injury prevention programs. Many of the injury prevention programs focus on landing mechanics that stress “soft” landings ([Mandelbaum et al., 2005] and [Hewett et al., 1999]) that are represented by landings associated with increased knee flexion angles at initial contact as well as increased flexion displacement during the landing process which in turn reduces peak impact forces ([Onate et al., 2001] and [Onate et al., 2005]). Results of this study suggest that reduced impact forces may in turn create less ATT and corresponding reduced ACL strain in actual dynamic gait. Further, in comparing our results to Yack et al. (1994) it is suggestive that muscle pre-activation is important, as reactive muscle control does not appear to control the amount of ATT.

Comparisons of our current results to actual human gait should be made cautiously as our design included instruction to the subject to relax as much as possible before weight acceptance began, resulting in a reactive mechanism. We fully acknowledge that muscle pre-activation occurs before foot strike during actual dynamic activity (Kvist and Gillquist, 2001). Both preparatory and reflexive muscle activations of quadriceps and hamstring are important during sports activities such as landing, cutting and jumping because it provides knee joint stability to prevent collapse ([Baratta et al., 1988] and [Riemann and Lephart, 2002]). It is unknown how this pre-activation modulates ATT during weight acceptance. An in vivo study of ATT during walking gait reported maximal ATT of 4.8 (2.3) mm occurring at 18 (12)° of knee flexion during the stance phase (Kvist and Gillquist, 2001). The absence of pre-activation in the current investigation may explain the approximately 4 mm greater ATT observed in the current study. Studies are underway to examine the extent to which pre-activation may act to control ATT.

Additionally there are mechanical differences between our very constrained reactive loading and actual dynamic activity. The peak loads in the current study range from 29% to 82% BW. However, previous single limb landing research reports loading magnitudes over 300% BW ([Madigan and Pidcoe, 2003] and [Schmitz et al., 2007]). Thus the data presented in the current investigation is representative of only the very early loading period. Further it is understood that sagittal trunk position can impact lower extremity loading patterns (Kulas et al., 2008), thus we chose to standardize the trunk position. It is unknown if altering trunk sagittal position (as typically occurs during functional activity) may affect our reported data.

In conclusion, greater axial compressive loads at initial foot contact with the knee near full extension produced greater ATT, regardless of increasing quadriceps and hamstring activation levels and reduced onset times. Increasing axial compressive loads with shallow knee flexion angle may cause substantial ATT which may result in increased ACL loading during functional, weight acceptance activities. The effects of pre-activation, tibial geometry, knee flexion angle and center of mass position on the relationship between ATT and quadriceps and hamstring muscle activation during early weight bearing activities in vivo should be investigated further.

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References

Baratta et al., 1988 R. Baratta, M. Solomonow, B.H. Zhou, D. Letson, R. Chuinard and R. D'Ambrosia, Muscular coactivation. The role of the antagonist musculature in maintaining knee stability, Am. J. Sports Med. 16 (1988), pp. 113–122.
Boden et al., 2000 B.P. Boden, G.S. Dean, J.A. Feagin Jr. and W.E. Garrett Jr., Mechanisms of anterior cruciate ligament injury, Orthopedics 23 (2000), pp. 573–578.
Brandon et al., 2006 M.L. Brandon, P.T. Haynes, J.R. Bonamo, M.I. Flynn, G.R. Barrett and M.F. Sherman, The association between posterior–inferior tibial slope and anterior cruciate ligament insufficiency, Arthroscopy 22 (2006), pp. 894–899. Abstract Order Document
Butler et al., 1980 D.L. Butler, F.R. Noyes and E.S. Grood, Ligamentous restraints to anterior–posterior drawer in the human knee. A biomechanical study, J. Bone Joint Surg. Am. 62 (1980), pp. 259–270.
Cerulli et al., 2003 G. Cerulli, D.L. Benoit, M. Lamontagne, A. Caraffa and A. Liti, In vivo anterior cruciate ligament strain behaviour during a rapid deceleration movement: case report, Knee Surg. Sports Traumatol. Arthrosc. 11 (2003), pp. 307–311.
Draganich and Vahey, 1990 L.F. Draganich and J.W. Vahey, An in vitro study of anterior cruciate ligament strain induced by quadriceps and hamstrings forces, J. Orthop. Res. 8 (1990), pp. 57–63.
Durselen et al., 1995 L. Durselen, L. Claes and H. Kiefer, The influence of muscle forces and external loads on cruciate ligament strain, Am. J. Sports Med. 23 (1995), pp. 129–136.
Giffin et al., 2004 J.R. Giffin, T.M. Vogrin, T. Zantop, S.L. Woo and C.D. Harner, Effects of increasing tibial slope on the biomechanics of the knee, Am. J. Sports Med. 32 (2004), pp. 376–382.
Hargrave et al., 2003 M.D. Hargrave, C.R. Carcia, B.M. Gansneder and S.J. Shultz, Subtalar pronation does not influence impact forces or rate of loading during a single-leg landing, J. Athl. Train. 38 (2003), pp. 18–23.
Hewett et al., 1996 T.E. Hewett, A.L. Stroupe, T.A. Nance and F.R. Noyes, Plyometric training in female athletes. Decreased impact forces and increased hamstring torques, Am. J. Sports Med. 24 (1996), pp. 765–773.
Hewett et al., 1999 T.E. Hewett, T.N. Lindenfeld, J.V. Riccobene and F.R. Noyes, The effect of neuromuscular training on the incidence of knee injury in female athletes. A prospective study, Am. J. Sports Med. 27 (1999), pp. 699–706.
Hosseini et al., 2009 A. Hosseini, T.J. Gill and G. Li, In vivo anterior cruciate ligament elongation in response to axial tibial loads, J. Orthop. Sci. 14 (2009), pp. 298–306.
Huston and Wojtys, 1996 L.J. Huston and E.M. Wojtys, Neuromuscular performance characteristics in elite female athletes, Am. J. Sports Med. 24 (1996), pp. 427–436.
Kulas et al., 2008 A. Kulas, P. Zalewski, T. Hortobagyi and P. Devita, Effects of added trunk load and corresponding trunk position adaptations on lower extremity biomechanics during drop-landings, J. Biomech. 41 (2008), pp. 180–185.
Kvist, 2006 J. Kvist, Tibial translation in exercises used early in rehabilitation after anterior cruciate ligament reconstruction exercises to achieve weight-bearing, Knee 13 (2006), pp. 460–463.
Kvist and Gillquist, 2001 J. Kvist and J. Gillquist, Anterior positioning of tibia during motion after anterior cruciate ligament injury, Med. Sci. Sports Exerc. 33 (2001), pp. 1063–1072.
Li et al., 1998 G. Li, T.W. Rudy, C. Allen, M. Sakane and S.L. Woo, Effect of combined axial compressive and anterior tibial loads on in situ forces in the anterior cruciate ligament: a porcine study, J. Orthop. Res. 16 (1998), pp. 122–127. (56)
Li et al., 1999 G. Li, T.W. Rudy, M. Sakane, A. Kanamori, C.B. Ma and S.L. Woo, The importance of quadriceps and hamstring muscle loading on knee kinematics and in-situ forces in the ACL, J. Biomech. 32 (1999), pp. 395–400.
Liu-Barba et al., 2007 D. Liu-Barba, M.L. Hull and S.M. Howell, Coupled motions under compressive load in intact and ACL-deficient knees: a cadaveric study, J. Biomech. Eng. 129 (2007), pp. 818–824.

Madigan and Pidcoe, 2003 M.L. Madigan and P.E. Pidcoe, Changes in landing biomechanics during a fatiguing landing activity, J. Electromyogr. Kinesiol. 13 (2003), pp. 491–498.
Mandelbaum et al., 2005 B.R. Mandelbaum, H.J. Silvers, D.S. Watanabe, J.F. Knarr, S.D. Thomas, L.Y. Griffin, D.T. Kirkendall and W. Garrett Jr., Effectiveness of a neuromuscular and proprioceptive training program in preventing anterior cruciate ligament injuries in female athletes: 2-year follow-up, Am. J. Sports Med. 33 (2005), pp. 1003–1010.
Markolf et al., 1978 K.L. Markolf, A. Graff-Radford and H.C. Amstutz, In vivo knee stability. A quantitative assessment using an instrumented clinical testing apparatus, J. Bone Joint Surg. Am. 60 (1978), pp. 664–674.
Markolf et al., 1981 K.L. Markolf, W.L. Bargar, S.C. Shoemaker and H.C. Amstutz, The role of joint load in knee stability, J. Bone Joint Surg. Am. 63 (1981), pp. 570–585.
Markolf et al., 1995 K.L. Markolf, D.M. Burchfield, M.M. Shapiro, M.F. Shepard, G.A.M. Finerman and J.L. Slauterbeck, Combined knee loading states that generate high anterior cruciate ligament forces, J. Orthop. Res. 13 (1995), pp. 930–935.
Markolf et al., 2004 K.L. Markolf, G. O'Neill, S.R. Jackson and D.R. Mcallister, Effects of applied quadriceps and hamstrings muscle loads on forces in the anterior and posterior cruciate ligaments, Am. J. Sports Med. 32 (2004), pp. 1144–1149.
Meyer and Haut, 2008 E.G. Meyer and R.C. Haut, Anterior cruciate ligament injury induced by internal tibial torsion or tibiofemoral compression, J. Biomech. 41 (2008), pp. 3377–3383.
More et al., 1993 R.C. More, B.T. Karras, R. Neiman, D. Fritschy, S.L. Woo and D.M. Daniel, Hamstrings – an anterior cruciate ligament protagonist. An in vitro study, Am. J. Sports Med. 21 (1993), pp. 231–237.
Olsen et al., 2004 O.E. Olsen, G. Myklebust, L. Engebretsen and R. Bahr, Injury mechanisms for anterior cruciate ligament injuries in team handball: a systematic video analysis, Am. J. Sports Med. 32 (2004), pp. 1002–1012.
Onate et al., 2001 J.A. Onate, K.M. Guskiewicz and R.J. Sullivan, Augmented feedback reduces jump landing forces, J. Orthop. Sports Phys. Ther. 31 (2001), pp. 511–517.
Onate et al., 2005 J.A. Onate, K.M. Guskiewicz, S.W. Marshall, C. Giuliani, B. Yu and W.E. Garrett, Instruction of jump-landing technique using videotape feedback: altering lower extremity motion patterns, Am. J. Sports Med. 33 (2005), pp. 831–842.
Pandy and Shelburne, 1997 M.G. Pandy and K.B. Shelburne, Dependence of cruciate-ligament loading on muscle forces and external load, J. Biomech. 30 (1997), pp. 1015–1024.
Pflum et al., 2004 M.A. Pflum, K.B. Shelburne, M.R. Torry, M.J. Decker and M.G. Pandy, Model prediction of anterior cruciate ligament force during drop-landings, Med. Sci. Sports Exerc. 36 (2004), pp. 1949–1958.
Renstrom et al., 1986 P. Renstrom, S.W. Arms, T.S. Stanwyck, R.J. Johnson and M.H. Pope, Strain within the anterior cruciate ligament during hamstring and quadriceps activity, Am. J. Sports Med. 14 (1986), pp. 83–87.
Riemann and Lephart, 2002 B.L. Riemann and S.M. Lephart, The sensorimotor system, part I: the physiologic basis of functional joint stability, J. Athl. Train. 37 (2002), pp. 71–79.
Schmitz et al., 2007 R.J. Schmitz, A.S. Kulas, D.H. Perrin, B.L. Riemann and S.J. Shultz, Sex differences in lower extremity biomechanics during single leg landings, Clin. Biomech. 22 (2007), pp. 681–688.
Shultz et al., 2006 S.J. Shultz, Y. Shimokochi, A.D. Nguyen, J.P. Ambegaonkar, R.J. Schmitz, B.D. Beynnon and D.H. Perrin, Nonweight-bearing anterior knee laxity is related to anterior tibial translation during transition from nonweight bearing to weight bearing, J. Orthop. Res. 24 (2006), pp. 516–523.
Shultz et al., 2009 S.J. Shultz, B.D. Beynnon and R.J. Schmitz, Sex differences in coupled knee motions during the transition from non-weight bearing to weight bearing, J. Orthop. Res. 27 (2009), pp. 717–723.
Torzilli et al., 1994 P.A. Torzilli, X. Deng and R.F. Warren, The effect of joint-compressive load and quadriceps muscle force on knee motion in the intact and anterior cruciate ligament-sectioned knee, Am. J. Sports Med. 22 (1994), pp. 105–112.
Uh et al., 2001 B.S. Uh, B.D. Beynnon, D.L. Churchill, L.D. Haugh, M.A. Risberg and B.C. Fleming, A new device to measure knee laxity during weightbearing and non-weightbearing conditions, J. Orthop. Res. 19 (2001), pp. 1185–1191.
Yack et al., 1994 H.J. Yack, L.M. Riley and T.R. Whieldon, Anterior tibial translation during progressive loading of the ACL-deficient knee during weight-bearing and nonweight-bearing isometric exercise, J. Orthop. Sports Phys. Ther. 20 (1994), pp. 247–253.