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In Vivo Tibiofemoral Kinematics During 4 Functional Tasks of Increasing Demand Using Biplane Fluoroscopy

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Background: The anterior cruciate ligament (ACL) has been well defined as the main passive restraint to anterior tibial translation (ATT) in the knee and plays an important role in rotational stability. However, it is unknown how closely the ACL and other passive and active structures of the knee constrain translations and rotations across a set of functional activities of increasing demand on the quadriceps.

Hypothesis: Anterior tibial translation and internal rotation of the tibia relative to the femur would increase as the demand on the quadriceps increased.

Study Design: Controlled laboratory study.

Methods: The in vivo 3-dimensional knee kinematics of 10 adult female patients (height, 167.8 ± 7.1 cm; body mass, 57 ± 4 kg; body mass index [BMI], 24.8 ± 1.7 kg/m^2; age, 29.7 ± 7.9 years) was measured using biplane fluoroscopy while patients completed 4 functional tasks. The tasks included an unloaded knee extension in which the patient slowly extended the knee from 90° to 0° of flexion in 2 seconds; walking at a constant pace of 90 steps per minute; a maximum effort isometric knee extension with the knee at 70° of flexion; and landing from a height of 40 cm in which the patient stepped off a box, landed, and immediately performed a maximum effort vertical jump.

Results: Landing (5.6 ± 1.9 mm) produced significantly greater peak ATT than walking (3.1 ± 2.2 mm) and unweighted full extension (2.6 ± 2.1 mm) ($P < .01$), but there was no difference between landing and a maximum isometric contraction (5.0 ± 1.9 mm). While there was no significant difference in peak internal rotation between landing (19.4° ± 5.7°), maximum isometric contraction (15.9° ± 6.7°), and unweighted full knee extension (14.5° ± 7.7°), each produced significantly greater internal rotation than walking (3.9° ± 4.2°) ($P < .001$). Knee extension torque significantly increased for each task ($P < .01$): unweighted knee extension (4.7 ± 1.2 N·m), walking (36.5 ± 7.9 N·m), maximum isometric knee extension (105.1 ± 8.2 N·m), and landing (140.2 ± 26.2 N·m).

Conclusion: Anterior tibial translations significantly increased as demand on the quadriceps and external loading increased. Internal rotation was not significantly different between landing, isometric contraction, and unweighted knee extension. Additionally, ATT and internal rotation from each motion were within the normal range, and no excessive amounts of translation or rotation were observed.

Clinical Relevance: This study demonstrated that while ATT will increase as demand on the quadriceps and external loading increases, the knee is able to effectively constrain ATT and internal rotation. This suggests that the healthy knee has a safe envelope of function that is tightly controlled even though task demand is elevated.

Keywords: ACL; rehabilitation; quadriceps; knee kinematics

The anterior cruciate ligament (ACL) is the main passive restraint to anterior tibial translation (ATT) in the knee and has an important role in rotational stability.[1,2] However, its role in maintaining knee joint stability during functional tasks while dynamic muscle forces are present is not completely understood. The kinematics of the knee during functional tasks is influenced by external forces, joint position, and the balance of active and passive contributory forces across the knee. Typical protocols for strengthening the quadriceps after an ACL injury use progressive loading of the quadriceps during both weightbearing and nonweightbearing exercises.[17,45,51] The way different exercises are used in a rehabilitation protocol depends on the level of strain that will be anticipated on the ACL and the other passive knee restraining structures through both translation and rotation.[7,8]

Cadaveric studies have reported that quadriceps contraction applies an anterior shear force on the proximal end of the tibia through the patellar tendon. Moreover, it has been reported that as quadriceps force increases, so does anterior shear force, ATT, and ACL force.[18] Indeed, when the knee is in an extended position, and in the absence of hamstring coactivation, the quadriceps can

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reportedly produce enough ATT to tear the ACL.[18,32,37] A modeling study of ACL function during drop landing suggested that quadriceps force and compressive force acting at the tibiofemoral joint contribute greatly to the total load at the ACL.[41] However, in simulated knee extensions, both cadaveric and modeling studies have reported significant reductions in ATT as well as ACL force when a hamstring load was added in the presence of a quadriceps load.[39,40,48,55] There is a renewed clinical interest in internal rotation with the introduction of double-bundle ACL reconstruction, which is reported to improve rotational stability.[54] Internal rotation of the tibia relative to the femur can also significantly strain the ACL, particularly in the presence of ATT.[7,36]

In vivo, the low demand task of walking has been reported to result in minimal translation and rotation of the knee.[31] Accurate quantification of the 6 degrees of freedom knee kinematics during tasks of differing demand in healthy individuals is essential to identify abnormal or pathologic motions associated with clinically relevant questions such as injury mechanisms or factors leading to joint degeneration. How closely the active and passive structures of the knee constrain translations and rotations across a set of functional activities of increasing demand has not been demonstrated in vivo. Biplane fluoroscopy systems enable the precise measurement of 6 degrees of freedom kinematics of the knee during dynamic activities.[15,22,29,31,34]

The purpose of this study was to use biplane fluoroscopy to measure, describe, and compare tibiofemoral translations and rotations of healthy individuals measured in vivo during 4 functional tasks of increasing demand on the quadriceps commonly included in rehabilitation protocols. We hypothesized that anterior translation and internal rotation of the tibia would increase as the quadriceps demand of the task increased.

MATERIALS AND METHODS

Study Protocol

This study was approved by the institutional review board, and all participants signed an informed consent. All had a history of participation in jumping and cutting sports, and no participant had any history of injuries or surgeries to the lower limbs. The in vivo 3-dimensional knee kinematics of 10 adult female patients (means ± standard deviations: height, 167.8 ± 7.1 cm; body mass, 57 ± 4 kg; body mass index [BMI], 24.8 ± 1.7 kg/m^2; age, 29.7 ± 7.9 years) was measured using biplane fluoroscopy while the patients completed 4 tasks that commonly appear in different stages of ACL rehabilitation programs and produce increasing demand.[50] The tasks included an unloaded knee extension in which the patient was in a seated position and extended the knee from 90° to 0° of flexion over 2 seconds; walking at a constant pace of 90 steps per minute; a maximum effort isometric knee extension with the knee in 70° of flexion; and landing from a height of 40 cm in which the patient stepped off a box, landed, and then immediately performed a maximum effort vertical jump (Figure 1). The demand on the quadriceps was quantified by the knee extension torque necessary to perform each task. The knee was placed at 70° for the maximum effort isometric knee extension, because patients normally produce peak extensor torque near this angle as demonstrated by computer modeling[46] and our own pilot work. In addition, this is supported by the measurements of Defrate et al,[16] who found that ATT increased with knee flexion during a lunge activity. The tasks were designed to produce an increasing amount of torque on the quadriceps while being relevant to the progression of rehabilitation programs following an ACL injury, going from early rehabilitation with the slow knee extension to the functional movement of walking to the end stage of rehabilitation with the isometric knee extension and finally return to sport with the landing. Each of the tasks appears in different stages of common ACL reconstruction rehabilitation programs.[46]

Surface Marker Motion Capture

Traditional surface marker motion capture techniques were used to collect kinematic and kinetic data of the lower extremities simultaneously with the fluoroscopy data. Ten cameras (Motion Analysis Corp, Santa Rosa, California) were positioned at approximately 36° intervals around the testing area and sampled at 240 Hz. Retroreflective spherical markers (10-mm diameter) were secured to anatomic landmarks on each participant to produce a modified "Helen Hayes" marker set with a 3-marker-per-segment configuration as reported previously.[13] All participants landed with their dominant foot on a force plate (Bertec, Columbus, Ohio).

Biplane Fluoroscopy System

The biplane fluoroscopy system was composed of 2 commercially available BV Pulsera C-arms with 30-cm image intensifiers (Philips Medical Systems, Best, The Netherlands), which were modified under appropriate US Food and Drug Administration guidelines. Two coupled, high-speed, high-resolution (1024 × 1024) digital cameras

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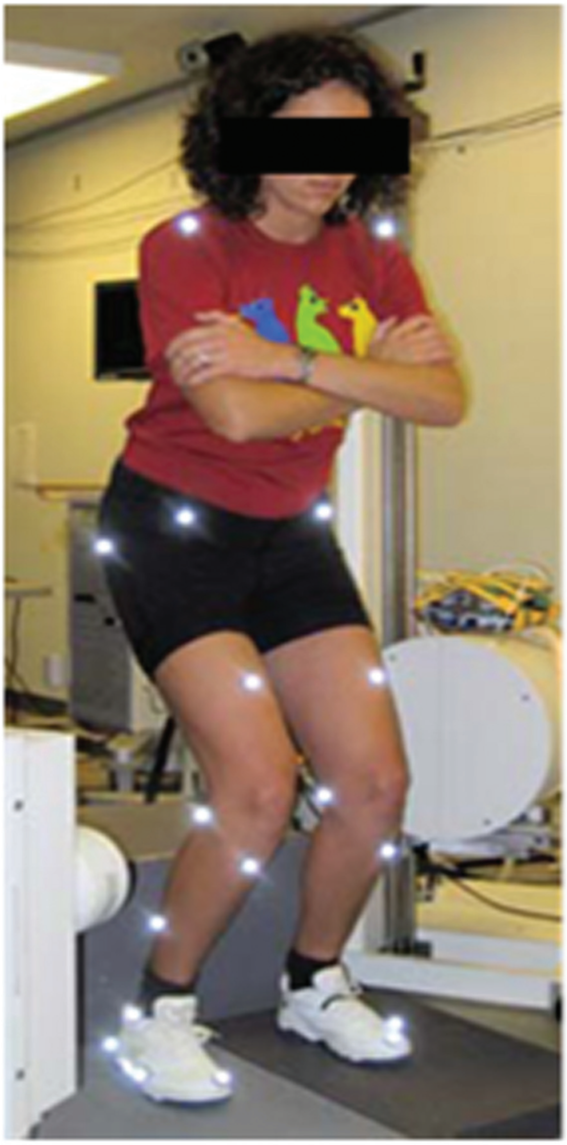

Figure 1. Participant performing the landing activity from a height of 40 cm within the biplane fluoroscopy system.

(Phantom V5.1, Vision Research, Wayne, New Jersey) were interfaced with the image intensifiers of the fluoroscopy systems using a custom built interface. The biplane fluoroscopy system was corrected for image distortion and calibrated using standard techniques that have been previously described.[10,19,49]

Biplane Fluoroscopy System Accuracy

The biplane fluoroscopy system was validated using standard validation techniques.[6,25,27] Tantalum beads were placed in 3 cadaveric knee specimens (5 beads per bone), and the specimens were dropped from a height of 40 cm within the biplane fluoroscopy system. This was done during a morning testing session and then repeated 4 hours later in an afternoon testing session. A total of 24 frames were selected from the image sequences during the morning testing session and afternoon testing session. The beads and bones were tracked independently using the same bone tracking methods as described in this study. The bead coordinate systems were registered to the femur and tibia/fibula coordinate systems, and subsequently the knee kinematics based on the bead data was determined. Bias and precision[4] representing the mean and standard deviation of the difference between the bone tracking and the bead tracking was calculated from the 24 frames.[4] The average bias and precision was $0.3° \pm 06°$, $0.4° \pm 0.5°$, and $0.3° \pm 0.6°$ for flexion-extension, varus-valgus, and internal-external rotations, respectively. The average bias and precision was 0.3 ± 0.4 mm, 0.4 ± 0.4 mm, and 0.3 ± 0.4 mm for anterior-posterior, medial-lateral, and distraction-compression translations, respectively. These values were consistent with previously reported studies using similar biplane fluoroscopy technology.[3,6,33,49]

In Vivo Biplane Fluoroscopy Data Collection

The in vivo biplane fluoroscopy data collection for the 4 motions consisted of obtaining a static computed tomography (CT) scan of the knee joint and collecting biplane fluoroscopy data during the motion trials simultaneously with the standard surface-marker based motion capture system. To obtain accurate 3-dimensional geometric descriptions of the bones in the knee, a high-resolution, static CT scan of the knee (approximately 12 cm above and below the joint line) using an Aquilion 64 scanner (Toshiba America Medical Systems, Tustin, California) was obtained. The sequence of images from the scan, representing slices of 0.5-mm thickness with a resolution of 512×512 pixels (ie, voxel size of approximately $0.7 \times 0.7 \times 0.5$ mm), was obtained using a standard 120-kVp and 200-mA bone reconstruction technique. Commercial software (Mimics, Materialize Inc, Ann Arbor, Michigan) used to extract the bone contours from the CT images and to reconstruct the 3-dimensional geometry of the femur, tibia, and fibula. These 3-dimensional geometries were then ready to be used by our tracking software[27] (Model-Based RSA, Medis Specials, Leiden, The Netherlands) for the 3-dimensional position and orientation (pose) estimation of the knee in the dynamic biplane fluoroscopy images.

Biplane fluoroscopy data were collected at 500 frames per second with a shutter speed of 1/2000 of a second. The pose of the femur and tibia/fibula during the motions was determined for each frame by semi-automatically extracting (automatically detected, manually assigned) both the inner and outer bony contours of the femur and tibia/fibula from the biplane fluoroscopy images. Subsequently, a fully automatic 6 degrees of freedom optimization algorithm determined the pose that optimally matched the detected contours with the projected contours from the imported bone geometries (Figure 2).

To determine the origin of the femoral coordinate system, a cylinder was fitted to the medial and lateral posterior condyles. The center of the coordinate system was placed on the midpoint of the cylinder center line. The

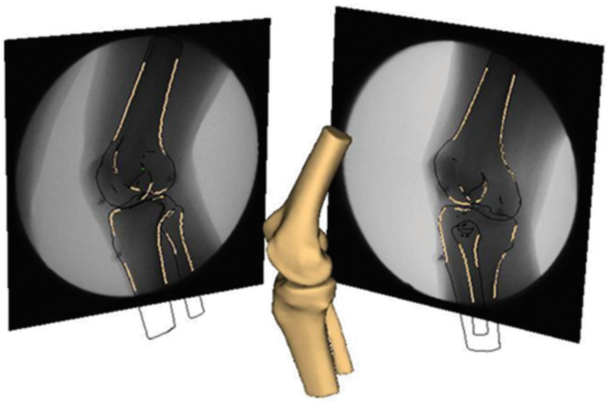

Figure 2. An image of the reconstructed tibia/fibula and femur in the biplane fluoroscopy system. Contours of bony landmarks were detected in each radiographic image (pale lines), and detected contours were optimally matched with the projected contours (black lines) from the imported bone geometries.

mediolateral (ML) axis of the femur was determined as the line through the long axis of this cylinder. The superior-inferior (SI) axis was aligned to the posterior line of the femur, and an anterior-posterior (AP) axis was determined as the cross product of the ML and SI axes. The tibial coordinate system was defined by the position of the tibia at full extension in the knee extension trial. At full knee extension, the femoral coordinate system was copied, with the duplicate coordinate system representing the tibia.[49] Using these coordinate systems, knee kinematics was calculated using methods described by Grood and Suntay.[21]

In addition, fluoroscopy kinematic data obtained during the isometric knee extension, walking, and landing trials were referenced to the unloaded knee extension trial data by subtracting the knee extension data from the data of each other task at the same flexion angles to obtain "relative" kinematic data. For example, ATT data from the unloaded knee extension trial was subtracted from ATT data during a landing at the same knee flexion angle to yield relative ATT measures.

Referencing to the passive motion of the knee collected during a low load activity, such as knee extension, in which the ACL is generally taut but not overly strained,[1,7] is important for studying the change in kinematics when loads are applied. Referencing to a low load task has been used previously during cadaver and in vivo investigations, has been reported to reduce intraparticipant variability, and produces more consistent comparisons of the changes in knee kinematics that occur during activity.[26,49] Specifically, internal-external rotation and anterior-posterior translation of the tibia relative to the femur are coupled to knee flexion angle because of the geometry of the articular surfaces and the soft tissue restraints in the knee. By referencing data to kinematics measured at each flexion angle of the low load knee extension, we removed this bias. We present here the relative kinematics associated specifically with the forces applied to the lower limb during the 4 tasks.

Statistical Analysis

Differences between knee translations and rotations across the 4 activities were compared using a repeated-measures analysis of variance (ANOVA) (α = .05). If significant results were obtained, a Bonferroni adjusted post hoc comparison was performed. All statistical analyses were completed with SPSS (IBM Inc, Chicago, Illinois). A power analysis of the rotation and translation data collected in pilot work revealed that 10 participants would detect a difference of 1.75 mm/6.0° using the statistical analysis described here with a β of .8 and an α of .05.

RESULTS

Landing produced significantly greater peak ATT (5.6 ± 1.9 mm) than unweighted full knee extension (2.6 ± 2.1 mm) ($P < .006$) and walking (3.1 ± 2.2 mm) ($P < .02$), but there was no difference between landing and the maximum isometric contraction (5.0 ± 1.9 mm). Additionally, the maximum isometric contraction produced significantly greater peak ATT than unweighted full knee extension ($P < .05$). There was no significant difference in the amount of medial tibial translation (MTT) between any of the 4 tasks (Table 1) (Figure 3).

During landing, peak ATT occurred at a knee flexion angle of 56.5° ± 4.8°. However, during walking, the knee stays in a relatively extended position throughout the motion from foot strike to toe off, and peak ATT occurred at a significantly lower knee flexion angle of 14.8° ± 5.5° ($P < .0001$). When the knee was moved from 90° of flexion to 0° during the extension trial, ATT was greatest at the more flexed position and steadily decreased as the knee got closer to full extension. Figure 4 demonstrates the ATT for the 4 tasks as a function of knee flexion angle. Anterior tibial translation peaked during landing at an average of 45 ± 21 milliseconds after ground contact. During walking, ATT did not peak until an average of 98 ± 53 milliseconds after ground contact. During the unweighted knee extension, ATT was greatest early in the motion when participants were in greater amounts of knee flexion, and ATT tended to decrease as the knee was extended (Figure 5).

Using the unweighted knee extension for reference, we found that relative ATT exhibited a similar relationship to the absolute ATT with increasing demand. Landing produced significantly greater relative ATT than isometric extension, and isometric extension produced significantly greater relative ATT than walking ($P < .05$) (Table 2).

While there was no significant difference in peak internal rotation between landing (19.4° ± 5.7°), maximum isometric contraction (15.9° ± 6.7°), and unweighted full knee extension (14.5° ± 7.7°), each produced significantly greater internal rotation than walking (3.9° ± 4.2°) ($P < .001$) (Table 1, Figure 6). This same relationship was observed in the relative amount of internal rotation for the 3 tasks analyzed. Walking produced on average relative external rotation

TABLE 1
Maximum, Minimum, Range, and Average Knee Kinematics for Landing, Maximum Isometric Knee Extension, Walking, and Unweighted Knee Extension[a]

	Flexion Angle, deg	Int Rot, deg	Valgus Angle, deg	ATT, mm	MTT, mm
Landing					
Maximum	89.7 (11.8)	19.9 (5.9)	2.3 (3.6)	5.6 (1.9)	0.4 (2.6)
Minimum	33.6 (8.3)	10.9 (6.1)	−2.5 (2.9)	2.7 (2.2)	3.5 (2.2)
Range	56.2 (8.6)	8.9 (2.5)	4.8 (1.3)	2.9 (1.7)	3.1 (1.4)
Average	64.4 (10.8)	16.2 (6.6)	−0.3 (3.2)	4.3 (1.8)	2.0 (2.5)
Iso ext					
Maximum	70.7 (5.8)	16.6 (7.2)	0.2 (3.7)	5.0 (2.8)	1.8 (2.7)
Walking					
Maximum	15.2 (5.2)	4.0 (4.2)	−0.2 (3.1)	3.1 (2.4)	3.1 (3.6)
Minimum	6.3 (3.8)	−0.2 (6.4)	−3.1 (2.8)	−0.8 (2.7)	−0.5 (2.0)
Range	8.9 (4.9)	4.2 (4.0)	2.9 (1.6)	3.9 (2.6)	3.6 (1.9)
Average	11.4 (3.9)	2.4 (4.5)	−1.9 (2.9)	1.3 (1.5)	−0.9 (2.3)
Unweighted ext					
Maximum	85.0 (9.7)	15.1 (8.0)	1.8 (2.2)	2.6 (2.1)	2.3 (1.9)
Minimum	−0.5 (0.5)	−0.9 (1.6)	−2.9 (3.8)	−0.7 (0.9)	−1.8 (1.4)
Range	84.6 (9.7)	16.1 (6.8)	4.7 (2.5)	3.3 (1.8)	4.1 (1.7)
Average	25.9 (4.4)	6.2 (5.1)	−0.9 (3.2)	0.8 (1.4)	−0.5 (2.0)

[a]Data in parentheses indicate ±1 standard deviation. Only maximum values are presented for isometric knee extension (iso ext) because only 1 position was recorded. Unweighted ext, unweighted knee extension; int rot, internal rotation; ATT, anterior tibial translation; MTT, medial tibial translation.

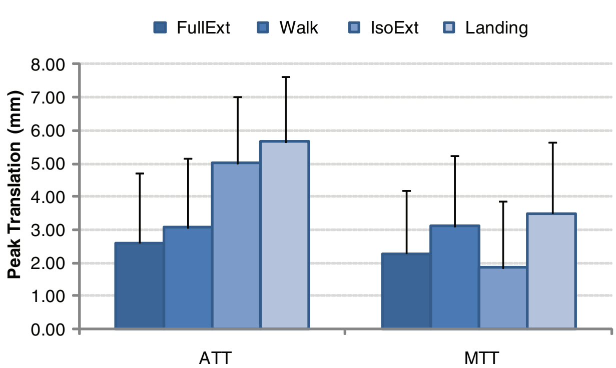

Figure 3. Peak anterior tibial translation (ATT) and medial tibial translation (MTT) for landing, maximum isometric knee extension (Iso Ext), walking, and unweighted knee extension.

(maximum, 6.2° ± 3.6°; average, 3.5° ± 2.8°), whereas both landing (maximum, 3.8° ± 4.5°; average, 2.2° ± 4.2°) and isometric contraction resulted in internal rotation (maximum, 3.9° ± 2.4°) (Table 2).

Landing (2.8° ± 3.0°) produced a significantly greater peak valgus angle than walking (0.1° ± 2.7°) ($P < .05$) and maximum isometric contraction (0.1° ± 2.3°) ($P < .05$). There were no other significant differences in peak valgus angle (Table 1).

Each movement produced significantly different knee extension torque from the other 3 movements ($P < .001$). The unweighted knee extension generated the least amount of torque (4.7 ± 1.2 N·m), and the landing generated the greatest (140.2 ± 26.2 N·m). Additionally, landing produced a significantly greater peak resultant ground-reaction force

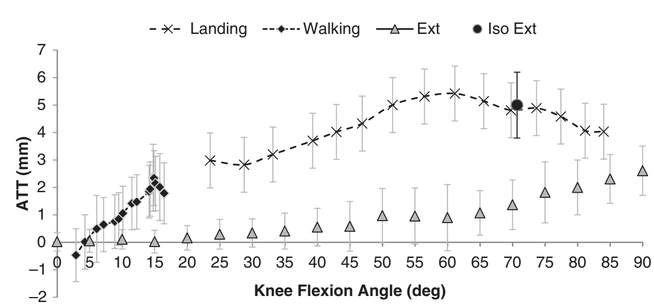

Figure 4. Anterior tibial translation versus knee flexion angle for 4 tasks of increasing demand.

than walking (2.60 ± 2.78 body weight [BW] vs 1.07 ± 0.07 BW) ($P < .0001$) (Figure 7).

DISCUSSION

We found that ATT significantly increased as demand on the quadriceps increased. This supported our hypothesis that increased demand on the quadriceps would lead to increased ATT for activities such as landing compared with walking and an unweighted knee extension compared with a maximum isometric contraction. However, a maximum voluntary isometric quadriceps contraction produced the same amount of increased ATT as the large eccentric quadriceps loads that were present during landing.

While there has not been an in vivo study that compared knee kinematics during dynamic tasks of increasing demand, the effect of increasing quadriceps demand has been investigated during static tasks. Yack et al[53]

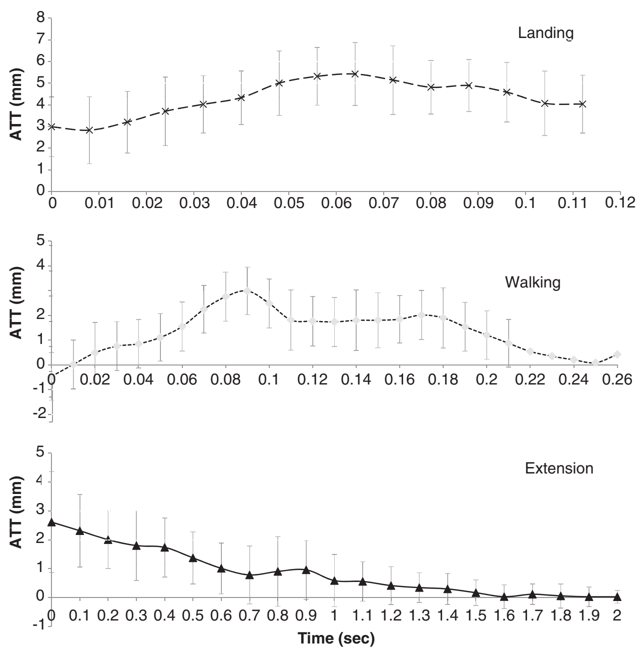

Figure 5. Anterior tibial translation versus time for landing, walking, and full extension, with the isometric extension omitted because data were captured from a single instant.

measured ATT during weightbearing and nonweightbearing isometric extensions of increasing torque with the knee in 20° of flexion. They reported that ATT increased by an average of 2.75 mm from a 25% body weight load to a 100% body weight load but only in the nonweightbearing condition. They reported that ATT was not different for the 4 loads applied in the weightbearing condition.[53] During dynamic weightbearing tasks such as walking and landing, we did not witness a similar relationship. A maximum isometric contraction resulted in greater ATT than walking and only slightly less ATT than the landing trial. This may be because the isometric contraction that was assessed in this study was done at 70° of knee flexion under maximum effort and resulted in greater torque than what was assessed in the Yack et al[53] study. Schmitz et al[44] reported ATT under increasing amounts of axial load during the transition from nonweightbearing to weightbearing. They measured ATT under 3 axial loads of 20% BW, 40% BW, and 60% BW and found that ATT increased after each increase in axial load with a minimum ATT of 4.7 ± 1.7 mm under 20% BW to 8.8 ± 2.1 mm of ATT under 60% BW.[44] The magnitudes of the translations reported by Yack et al[53] and Schmitz et al[44] were much larger than what was observed in the current study. This is most likely because both used an arthrometer to measure ATT and measured ATT with the knee in 20° of flexion. However, their findings support the relationship of increased ATT under increasing demand that we observed during functional tasks.

Eccentric quadriceps contraction during landing and the resulting shear force on the proximal end of the tibia has been reported to be a risk factor for ACL injury during competitive sports situations,[23] while maximum isometric knee extensions are frequently used in knee rehabilitation programs by those who advocate the use of open chain exercises.[9,20] Our results indicate that there were nonsignificant differences in ATT between a controlled landing in the laboratory and a maximum isometric contraction, despite landing producing significantly greater knee extension torque. When the absolute data were referenced to the unweighted knee extension to produce relative ATT, that same relationship was observed. Landing and isometric contraction produced significantly greater ATT than walking but were similar to each other. This would support the use of closed chain exercises in knee rehabilitation given that significantly greater amounts of knee extension torque can be generated in a closed chain activity, such as landing, without producing significantly greater ATT than a maximum effort open chain task at the same flexion angle.

Anterior tibial translation for each motion was within the normal range, and no excessive amounts of translation or rotation were observed. Our results are similar to the average of 5.2 mm of ATT reported from KT-1000 measurements when the quadriceps are active.[14] In vitro studies have reported that normal knees can experience up to 13.6 ± 0.8 mm of increased ATT before the ACL ruptures.[52] An in vivo study of ATT during walking reported a maximum ATT of 4.8 ± 2.3 mm occurring at 18° ± 12° of knee flexion during the stance phase.[30] This study also reported a mean increase in ATT of 2.8 ± 1.2 mm from heel strike to the first half of stance phase. We found that the peak ATT during walking was only slightly less than what was reported at 3.1 ± 2.2 mm, and we observed a 1.1 ± 1.9 mm increase in ATT from heel strike to 50% of stance phase. However, peak ATT occurred at the slightly lower knee flexion angle of 14.8° ± 5.5°, which may help explain the smaller values of ATT that were observed here along with differences in the methodology used to measure ATT.

Our measurements of internal rotation were between 2° and 5° during walking, similar to that reported by others.[29,35,56] Internal rotation during walking can be caused by the action of the quadriceps, especially the vastus medialis, extending the knee during midstance.[43,47] Greater activation of this muscle group with the knee in a flexed position during the isometric contraction could also cause greater internal rotation. For this reason, we expected internal rotation of the knee to mirror the relationship of ATT with increasing demand resulting in increased internal rotation, particularly because female patients have been reported to exhibit large amounts of internal rotation during landing.[23] However, while internal rotation was greatest during landing, it was not significantly different between landing, isometric contraction, and unweighted knee extension, and each of these 3 movements produced significantly greater internal rotation than walking. Additionally, isometric contraction produced significantly greater relative internal rotation than walking. It appears that complex interactions are taking place between the thigh musculature and knee geometry, resulting in internal rotation of the knee during high demand dynamic tasks.

TABLE 2
Maximum, Minimum, Range, and Average Relative Knee Kinematics for Landing, Maximum Isometric Knee Extension, and Walking, With Unweighted Knee Extension Used for Reference at the Same Flexion Angles[a]

	Int Rot, deg	Valgus Angle, deg	ATT, mm	MTT, mm
Landing				
Maximum	3.8 (4.5)	1.8 (1.5)	4.3 (1.3)	2.3 (1.1)
Minimum	−0.1 (4.4)	−0.6 (1.6)	1.2 (1.0)	−0.2 (1.2)
Range	3.9 (2.0)	2.4 (1.5)	3.1 (1.6)	2.5 (1.3)
Average	2.2 (4.2)	0.4 (1.4)	2.9 (1.0)	1.0 (0.9)
Iso ext				
Maximum	3.9 (2.4)	1.8 (1.3)	3.5 (1.4)	1.0 (1.3)
Walking				
Maximum	−1.3 (2.9)	0.4 (3.1)	2.3 (2.1)	2.6 (2.1)
Minimum	−6.2 (3.6)	−2.5 (2.0)	−1.0 (1.1)	−1.0 (1.7)
Range	4.9 (3.6)	2.9 (2.2)	3.3 (2.0)	3.6 (1.7)
Average	−3.5 (2.8)	−1.3 (2.1)	0.7 (1.1)	0.4 (1.3)

[a]Data in parentheses indicate ±1 standard deviation. Only maximum values are presented for isometric knee extension (iso ext) because only 1 position was recorded. Unweighted ext, unweighted knee extension; int rot, internal rotation; ATT, anterior tibial translation; MTT, medial tibial translation.

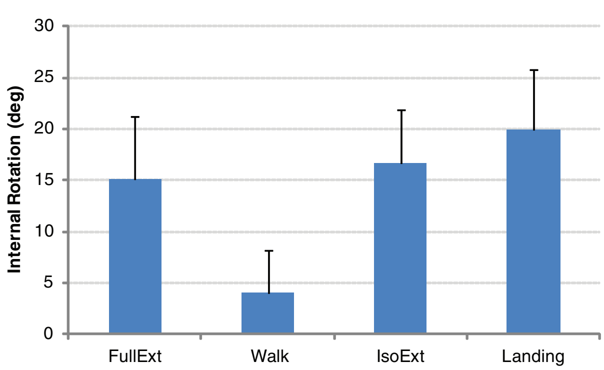

Figure 6. Peak internal rotation angle for the 4 tasks of increasing demand.

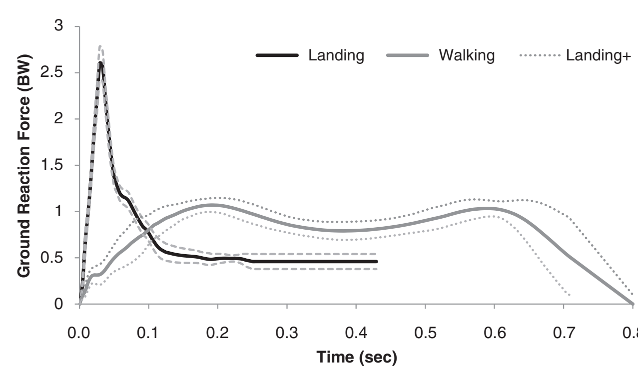

Figure 7. Ground-reaction force ± 1 standard deviation versus time for landing and walking.

It has been reported that during tasks of increased demand on the thigh musculature, women cannot generate a sufficient reaction moment against the externally applied internal rotation moment of the shank relative to the femur.[28] The internal structures of the knee that are responsible for this moment are contraction of the biceps femoris muscle, or ligaments in tension such as the medial and lateral collateral ligaments and ACL. Women have been reported to have lower torsional knee joint stiffness than men in response to combined rotational loads.[11,24] We hypothesized that our population of female volunteers would demonstrate the greatest amount of internal rotation during landing compared with all other activities. When internal rotation during walking, isometric contraction, and landing was referenced to internal rotation during the unweighted knee extension at the same flexion angle, only walking was different from the other 2 conditions. The high demand tasks of isometric contraction and landing produced only approximately 4° of peak relative internal rotation as opposed to approximately 15° to 20° of absolute rotation that was observed in this and other studies.[12,38] Given that the peak knee flexion angle recorded during walking was of 14.8° ± 5.5°, it is suggested that the large absolute internal rotation angles that have been reported during landing and other activities performed with greater knee flexion angles may be a result of the natural geometry and kinematics of the knee as it moves through the flexion range of motion.

The knee mechanically regulates stability by the inherent geometry of the bones, ligaments, and muscles, and the body balances these factors in different manners as task demand increases. It has been reported that muscle preactivation occurs before foot strike during dynamic activities.[30] Both preparatory and reflexive muscle activations of quadriceps and hamstrings are integral during activities such as walking and landing, in providing knee joint stability to control translations and rotations.[5,42] It is unknown how exactly this pre-activation modulates ATT during normal dynamics tasks, but our results indicate that the thigh musculature is able to regulate translation and rotation of the tibia within a small safe range despite increases in demand. This may also help explain the larger ATT and internal rotation values reported in previous studies where efforts have been made to minimize thigh musculature preactivation as well as hamstring coactivation during trials.[7,44]

We recognize limitations of the current study. This study used a group of healthy female participants in a controlled setting for experimentation and, as such, is a limitation in understanding knee motion in patients with joint pathology. This population was selected to better understand the normal knee envelope of function in healthy women who had a history of participation in jumping and cutting sports. Follow-up studies should test men and women who would be identified as at risk for ACL injury or who currently demonstrate knee pathology. Furthermore, participants performed each of the activities in this study in a controlled laboratory setting and did not experience the intensity of a competitive setting or unanticipated events that may occur when the activities are performed in a natural setting.

In conclusion, we found that while ATT will increase as demand on the quadriceps increases, internal rotation was not significantly different between landing, isometric contraction, and unweighted knee extension. The knee is able to effectively constrain ATT and internal rotation. This suggests that the healthy knee has a safe envelope of function that is tightly controlled even when task demand is elevated.

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