Coupled Motions Under Compressive Load in Intact and ACL-Deficient Knees: A Cadaveric Study

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Knowledge of the coupled motions, which develop under compressive loading of the knee, is useful to determine which degrees of freedom should be included in the study of tibiofemoral contact and also to understand the role of the anterior cruciate ligament (ACL) in coupled motions. The objectives of this study were to measure the coupled motions of the intact knee and ACL-deficient knee under compression and to compare the coupled motions of the ACL-deficient knee with those of the intact knee. Ten intact cadaveric knees were tested by applying a 1600 N *compressive load and measuring coupled internal-external and varus-valgus rotations and anterior-posterior and medial-lateral translations at* 0 deg*,* 15 deg*, and* 30 deg *of flexion. Compressive loads were applied along the functional axis of axial rotation, which coincides approximately with the mechanical axis of the tibia. The ACL was excised and the knees were tested again. In the intact knee, the peak coupled motions were* 3.8 deg *internal rotation at* 0 deg *flexion changing to* −4.9 deg *external rotation at* 30 deg *of flexion,* 1.4 deg *of varus rotation at* 0 deg *flexion changing to* −1.9 deg *valgus rotation at* 30 deg *of flexion,* 1.4 mm *of medial translation at* 0 deg *flexion increasing to* 2.3 mm *at* 30 deg *of flexion, and* 5.3 mm *of anterior translation at* 0 deg *flexion increasing to* 10.2 mm *at* 30 deg *of flexion. All changes in the peak coupled motions from* 0 deg to 30 deg *flexion were statistically significant* ($p < 0.05$)*. In ACL-deficient knees, there was a strong trend (marginally not significant,* $p = 0.07$*) toward greater anterior translation* (12.7 mm) *than that in intact knees* (8.0 mm)*, whereas coupled motions in the other degrees of freedom were comparable. Because the coupled motions in all four degrees of freedom in the intact knee and ACL-deficient knee are sufficiently large to substantially affect the tibiofemoral contact area, all degrees of freedom should be included when either developing mathematical models or designing mechanical testing equipment for study of tibiofemoral contact. The increase in coupled anterior translation in ACL-deficient knees indicates the important role played by the ACL in constraining anterior translation during compressive loading.* [DOI: 10.1115/1.2800762]

Introduction

Osteoarthritis (OA) affects more than 21 million adults in the US and that number will keep growing so that nearly 40% of US citizens will have OA by 2030 [1]. Because of its load bearing role and because of the large number of injuries to soft tissues involved in load transmission, the articular surfaces of the knee joint are particularly susceptible to OA. OA of the knee is debilitating in the advanced stages and often must be treated by total joint replacement so that the etiology of OA is a subject of intense research. One factor that has been implicated in the etiology of OA is the tibiofemoral contact, which is created by compressive loads transmitted by the knee. Increases in the contact pressure, decreases in the contact area, and shifts of the contact area are all believed to be related to the genesis of the disease [2]. Hence, the study of tibiofemoral contact is a subject of importance in understanding the etiology of OA.

Tibiofemoral contact has been studied extensively both computationally (for example, Refs. 3–5) and experimentally in cadaveric knees (for example, Refs. 6–8). Collectively, these studies have provided much valuable information important to the prevention of OA such as the biomechanical criteria that meniscal replacements must satisfy to restore the tibiofemoral contact pressure to normal, the influence of rotational abnormalities of the lower limbs on the knee joint at short and long terms, and the role of the ligaments and menisci in controlling tibiofemoral contact when loads other than compression are transmitted.

Assuming that flexion-extension rotation is constrained, compression of the knee may develop coupled motions in the remaining degrees of freedom, which include internal-external (I-E) rotation, varus-valgus (V-V) rotation, medial-lateral (M-L) translation, and anterior-posterior (A-P) translation. Coupled motions in response to various applied loads have been well documented (for example, Refs. 9–15) but coupled motions in response to compression loading have not been studied to the knowledge of the authors.

Knowledge of these coupled motions would be useful for several reasons. Two reasons for studying coupled motions under compression is to determine which degrees of freedom should not be constrained when either developing mathematical models of tibiofemoral contact or designing test equipment to apply compressive loads to study tibiofemoral contact in whole knee joints. If warranted, then constraining degrees of freedom is advantageous because obtaining the finite element solution is simplified and because designing the test equipment is simplified. Two previously developed finite element models have imposed constraints on coupled motions [3,4]. However, constraining rotations af-

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fected contact variables by as much as 19% compared to unconstrained rotation [5]. Thus, inappropriately constraining degrees of freedom, which develop coupled motions, can greatly affect the results.

Another reason to study coupled motions under compression is to determine how the absence of the anterior cruciate ligament (ACL) affects coupled motions. Many studies have observed changes in tibiofemoral kinematics in ACL-deficient knees compared to intact knees during weight bearing [16], leg presses [17], and walking [18,19]. Because the coupled motions in the ACL-deficient knee would be expected to differ from those of the intact knee, knowledge of the coupled motions in ACL-deficient knees is important to computational and experimental studies of tibiofemoral contact in these knees because the required degrees of freedom may differ from those of the intact knee. Also, the risk of developing OA following ACL injury is high [20]. The increased risk may be due to a shift in the tibiofemoral contact pattern resulting in overloading of certain areas of the contact surface [2]. Accordingly, comparing coupled motions of the intact and ACL-deficient knees may provide insight into the role of the ACL in determining the pattern of tibiofemoral contact.

One objective of this study was to measure the coupled motions of the intact knee in I-E rotation, V-V rotation, M-L translation, and A-P translation under compression. A second objective was to measure the coupled motions of the ACL-deficient knee. A final objective was to compare the coupled motions of the ACL-deficient knee with those of the intact knee to determine the role of the ACL in coupled motions.

Materials and Methods

Specimen Preparation. Knees from elderly humans ($n=10$, average 63 years, range 46–77 years) were harvested and stored at $-20°C$. Any specimen with evidence of either degenerative arthritis or chondrocalcinosis by radiographic analysis was excluded from the study. The fibula was cut approximately 10 cm from the joint line and a screw was driven through the fibula near the cut end into the tibia to keep the fibula in place. The intermedullary canals of the femur and tibia were reamed to 9.5 mm in diameter and steel rods were inserted and potted in the intermedullary canals using polymethylmethacrylate such that the ends of the rods closest to the knee were 11 cm from the joint line [21].

A six degree-of-freedom load application system (Fig. 1) was used to align, precondition, and test each knee [21]. With the load application system, loads can be applied in any degree of freedom while the resulting coupled motions are measured. The flexion angle, adjustable over the range of physiologic motion, is fixed at the desired angle while the remaining degrees of freedom are unconstrained. The degrees of freedom are unconstrained through the use of low-friction linear bearings for the three translations, a low-friction ball bearing for axial rotation, and air bearings for varus-valgus rotation. Loads are applied using servo-controlled pneumatic actuators and motions are measured using either linear variable or rotary variable differential transformers with an accuracy of 0.25 mm in translations or 0.075 deg in rotations. The coordinate system used for application of loads and measurement of motions follows that described by Grood and Suntay [22].

Each specimen was aligned using the functional axes method [21,23], in which the natural axes of joint motion in flexion extension and axial rotation are aligned with those of the load application system. The goal of this method is to position the knee specimen so that the coupled translations in the plane perpendicular to the axis of interest are eliminated when the knee is rotated passively. For example, as the knee is moved in flexion-extension rotation, coupled A-P and compression-distraction (C-D) translations are virtually eliminated [21]. Using this method of alignment in conjunction with the coordinate system of Grood and Suntay [22], tibial rotation occurs about the tibial-fixed axis and compression loads are applied along this same axis, which is invariant with changes in knee flexion angle.

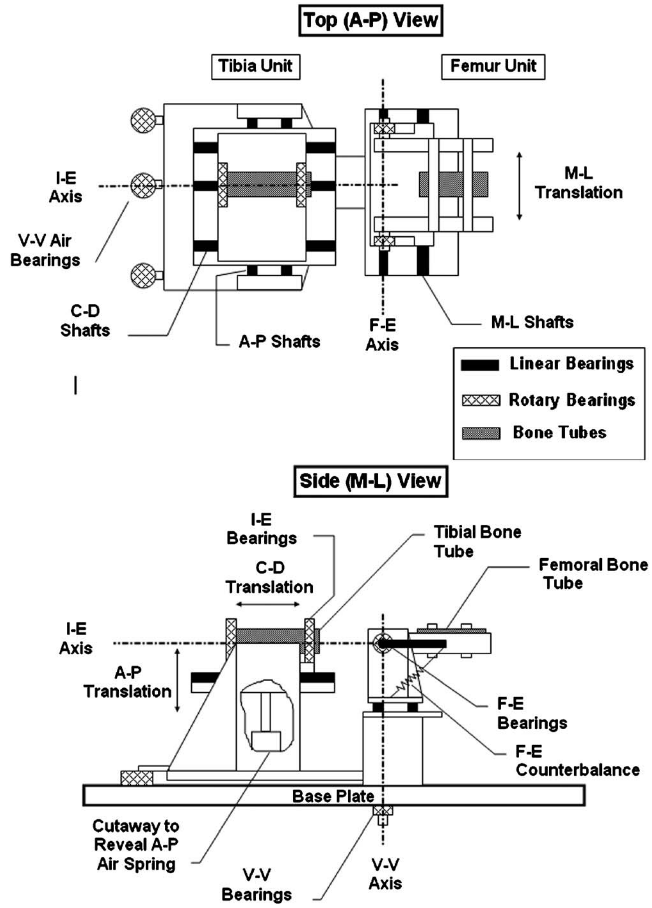

Fig. 1 Diagram of six degree-of-freedom knee loading apparatus. The degrees of freedom follow the coordinate system of Grood and Suntay [22] so that the flexion-extension axis is fixed to the femur and the axial rotation axis is fixed to the tibia. Accordingly, the femur unit provides two degrees of freedom, flexion-extension (F-E) rotation and medial-lateral (M-L) translation. The tibia unit provides the remaining four degrees of freedom. Pneumatic actuators (omitted for clarity) develop loads in all degrees of freedom except F-E where loads are developed by a stepper motor. The loads developed by the actuators are measured with strain gage load cells. Motions in all degrees of freedom are enabled through the use of low-friction bearings. The motions are measured with linear variable differential transformers (LVDTs) for translations and rotational variable differential transformers (RVDTs) for rotations. Flexion angle is adjustable over the full physiologic range and the F-E counterbalance ensures that the weight of the specimen does not create an unwanted F-E moment. In the present study, compression loading was applied with the knee at a preset flexion angle and the coupled motions in the remaining degrees of freedom were measured.

The steel rods are used to interface the specimen with alignment fixtures, which in turn are attached to the load application system. Using alignment fixtures each of which offers adjustability in six degrees of freedom, the specimen is moved relative to the load application system to establish alignment as described above. This alignment method offers good repeatability [23]. Once aligned, the tibia and femur are potted in square aluminum tubes and clamped to the load application system for testing.

Measurement of Coupled Motions. The intact knee was preconditioned by applying four load cycles at 0 deg and 30 deg of flexion each. Each load cycle consisted of applying and removing a 1600 N compressive load (approximately 2.1 times body weight) in 100 N increments [24]. Compressive loads were applied along the functional axial rotation axis, which coincides approximately with the mechanical axis of the tibia. The order of flexion angle was selected randomly. Data from a pilot study indicated that the peak coupled translations differed by less than 0.1 mm and that the peak coupled rotations differed by less than 0.3 deg between the third and fourth load cycles. With the tibial bone tube fixed in a horizontal position, the 0 deg flexion position was defined by positioning the femoral bone tube horizontally using a bubble level.

After the knee was preconditioned, the coupled motions of the intact knee were measured by applying a compression-distraction-compression load cycle at 0 deg, 15 deg, and 30 deg of flexion in random order. This range of flexion angles encompasses the normal range of flexion during the weight acceptance phase of the gait cycle [19]. A 50 N compressive load was applied and removed and a 50 N distractive load was applied and removed. The position of the tibia relative to the femur after removal of the 50 N distractive load defined the unloaded position of the tibia relative to the femur. Then, a 1600 N compressive load was applied in 100 N increments while coupled motions in I-E, V-V, M-L, and A-P degrees of freedom were recorded. The 1600 N compressive load (approximately 2.1 times body weight) was selected to approximate the tibiofemoral contact force during normal walking [25]. In these tests on the intact knee, the joint capsule was intact and no forces were applied to any muscles crossing the knee.

The intact knee was removed from the load application system. Medial and lateral parapatellar incisions were made and the patella and the patellar tendon were reflected distally. These incisions have no significant effect but allowed ready access to the joint space [26,27]. The joint was visually inspected and specimens with evidence of degenerative arthritis, chondrocalcinosis, or other soft tissue damage (e.g., meniscal tear as evidenced by partial meniscectomy) were excluded from the study at this time. The ACL was excised and the ACL-deficient knee was replaced in the load application system. The ACL-deficient knee was preconditioned using the same protocol that was used for the intact knee. The coupled motions for the ACL-deficient knee were measured using the same loading protocol described for the intact knee. As for the intact knee, no forces were applied to any muscles crossing the knee while testing the ACL-deficient knee.

Data Analysis. The average peak translations and rotations, corresponding standard deviations, and corresponding ranges at each flexion angle were computed. The peak translations and rotations were the greatest absolute translations and rotations developed between the unloaded intact knee and maximum load. A single-factor, repeated-measures analysis of variance (ANOVA) was performed for each degree of freedom to determine the effect of flexion angle on peak translation and rotation values. The analyses were performed separately for the intact knee and ACL-deficient knee. Significant effects were further analyzed with the Tukey's test.

A two-factor ANOVA with a split-plot design blocked by subjects was performed to determine the effect of knee condition and flexion angle on peak translations (M-L translation and A-P translation) and rotations (I-E rotation and V-V rotation). The main factor was knee condition at two levels (intact and ACL deficient) and the subfactor was flexion angle at three levels (0 deg, 15 deg, and 30 deg). Although the order of flexion angle was randomized, it was not possible to randomize the order of knee condition. Because the two factors were not completely randomized, a split-plot design is more appropriate than a repeated-measures design. Significance was set at $p < 0.05$.

Results

Typical Load-Displacement Curves. As the knee was loaded with a compressive force, motions in the four coupled degrees of freedom were observed. In the example data shown at 30 deg of flexion for an intact knee specimen, the tibia exhibited external and valgus rotations as well as medial and anterior translations (Fig. 2). The peak external rotation of 9 deg was greater than the peak valgus rotation of −4 deg. Also, the peak anterior translation of 12 mm was greater than the peak medial translation of 4 mm.

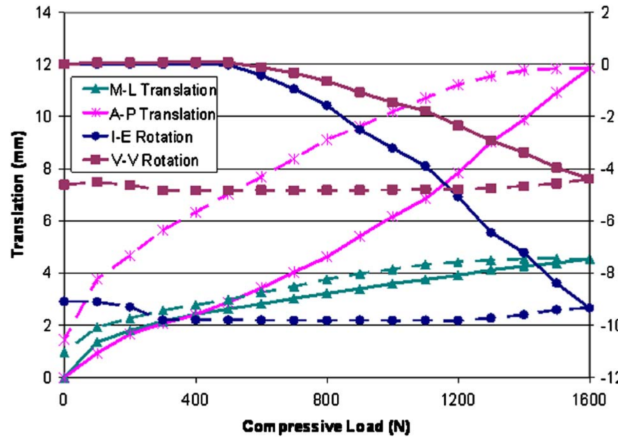

Fig. 2 Example load-displacement curves for coupled motions in four degrees of freedom under compressive load for one intact knee specimen at 30 deg of flexion. Positive numbers represent movement in the first-listed direction of the respective degree of freedom (e.g., "internal" for I-E rotation). For V-V rotation, varus is the first rotation. Solid lines represent loading; dashed lines represent unloading. The knee was loaded from 0 N to 1600 N (2.1×BW) and unloaded in 100 N increments.

Coupled Motions in the Intact Knee. Flexion angle significantly affected coupled motions in the intact knee (Table 1). I-E rotation in the intact knee was internal at 0 deg of flexion but gradually became external as the flexion angle increased to 30 deg (Table 1 and Fig. 3). V-V rotation in the intact knee was varus at 0 deg of flexion but gradually became valgus as the flexion angle increased to 30 deg (Table 1 and Fig. 4). In M-L translation, medial translation increased with increasing flexion angle (Table 1 and Fig. 5). In A-P translation, anterior translation increased with increasing flexion angle (Table 1 and Fig. 6).

Table 1 Peak translations and rotations in each degree of freedom at three flexion angles under a 1600 N compressive load for the intact knee. Numbers shown are average peak values±standard deviation (range). Rotations are expressed in degrees; translations are expressed in millimeters. Positive numbers represent movement in the first-listed direction of the respective degree of freedom (e.g., "internal" for I-E rotation). For V-V rotation, varus is the first rotation. Degrees of freedom where the flexion angle effect was statistically significant are denoted by * (p<0.05). Within each degree of freedom, different letters (i.e., "A" versus "B") indicate a statistically significant difference between the corresponding flexion angles in that pair. The same letters indicate no statistically significant difference.

Intact knee				
Flexion angle (deg)	I-E rotation* (p=0.0006)	V-V rotation* (p=0.0083)	M-L translation* (p=0.0210)	A-P translation* (p<0.0001)
0	3.8±4.3 (12.4)^{A}	1.4±2.0 (5.7)^{A}	1.4±1.3 (4.6)^{A}	5.3±2.6 (8.5)^{A}
15	1.7±5.5 (14.9)^{A}	0.2±1.9 (6.0)^{A,B}	2.3±1.1 (3.4)^{B}	8.5±4.2 (12.4)^{B}
30	−4.9±8.0 (26.6)^{B}	−1.9±3.6 (12.3)^{B}	2.3±1.4 (4.1)^{B}	10.2±4.7 (14.0)^{B}

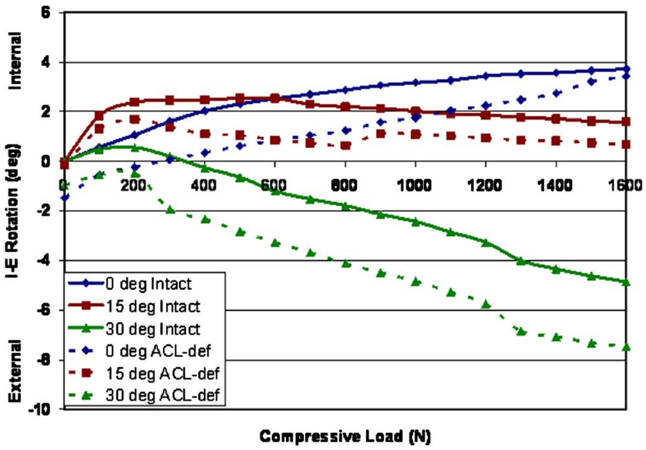

Fig. 3 Average I-E rotations for the intact knee and ACL-deficient knee for each flexion angle during loading referenced to the unloaded position of the intact knee. Positive numbers represent movement in the first-listed direction of the respective degree of freedom (e.g., "internal" for I-E rotation). For V-V rotation, varus is the first rotation. Error bars are not shown for clarity. The average rotations were comparable for the intact knee and ACL-deficient knee.

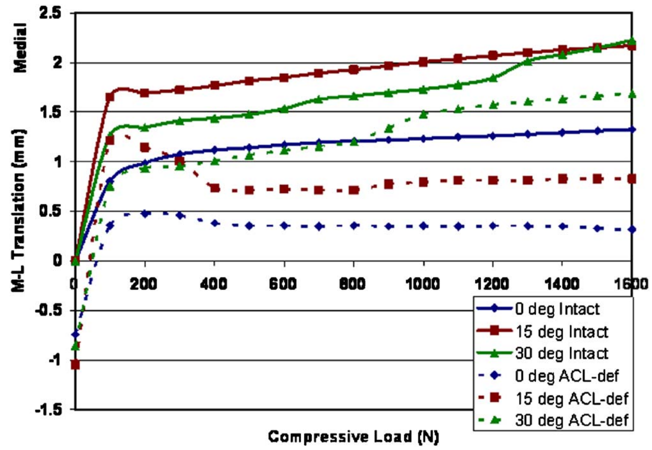

Fig. 5 Average M-L translations for the intact knee and ACL-deficient knee for each flexion angle during loading referenced to the unloaded position of the intact knee. Positive numbers represent movement in the first-listed direction of the respective degree of freedom (e.g., "internal" for I-E rotation). For V-V rotation, varus is the first rotation. Error bars are not shown for clarity. The average translations were comparable for the intact knee and ACL-deficient knee.

Coupled Motions in the Anterior Cruciate Ligament Deficient Knee. Coupled motions in the ACL-deficient knee under a compressive load were observed (Figs. 3–6). Two degrees of freedom (I-E rotation and A-P translation) in the ACL-deficient knee exhibited significant differences between flexion angles (Table 2). Coupled motions in I-E rotation and A-P translation in the ACL-deficient knee exhibited similar trends to those of the intact knee. In M-L translation, medial translation increased with increasing flexion angle, although not significantly.

Comparison of Intact and Anterior Cruciate Ligament Deficient Knees. Anterior translation showed the greatest difference between the intact knee and ACL-deficient knee with little difference in coupled motions for the other degrees of freedom. Coupled anterior translation in the ACL-deficient knee (12.7 ± 11.8 mm) was marginally not statistically greater than that in the intact knee (8.0 ± 4.3 mm) (Table 3). There was no significant interaction effect between knee condition and flexion angle in any degree of freedom ($p \geqslant 0.2236$).

Discussion

Because knowledge of coupled motions of the knee under compressive load is useful to the study of tibiofemoral contact, two objectives of our study were to measure the coupled motions of the intact knee in I-E rotation, V-V rotation, M-L translation, and A-P translation under compression of the intact knee and ACL-deficient knee. A third objective was to compare the coupled motions of the ACL-deficient knee with those of the intact knee to determine the role of the ACL in coupled motions. We determined the envelope of coupled motions in the intact knee and ACL-deficient knee (Figs. 3–6) and found that flexion angle signifi-

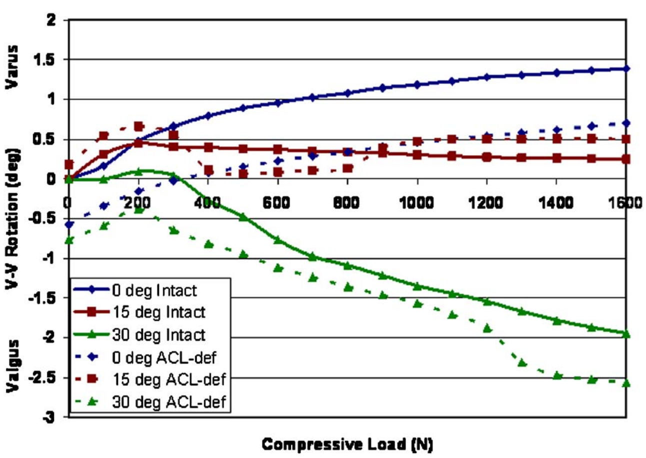

Fig. 4 Average V-V rotations for the intact knee and ACL-deficient knee for each flexion angle during loading referenced to the unloaded position of the intact knee. Positive numbers represent movement in the first-listed direction of the respective degree of freedom (e.g., "internal" for I-E rotation). For V-V rotation, varus is the first rotation. Error bars are not shown for clarity. The average rotations were comparable for the intact knee and ACL-deficient knee.

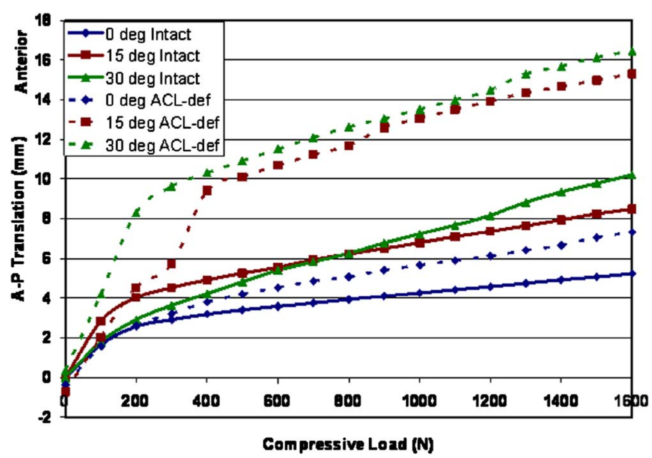

Fig. 6 Average A-P translations for the intact knee and ACL-deficient knee for each flexion angle during loading referenced to the unloaded position of the intact knee. Positive numbers represent movement in the first-listed direction of the respective degree of freedom (e.g., "internal" for I-E rotation). For V-V rotation, varus is the first rotation. Error bars are not shown for clarity. The average translations significantly increased for the ACL-deficient knee compared to the intact knee.

Table 2 Peak translations and rotations in each degree of freedom at three flexion angles under a 1600 N compressive load for the ACL-deficient knee referenced to the unloaded position of the intact knee. Numbers shown are average peak values±standard deviation (range). Rotations are expressed in degrees; translations are expressed in millimeters. Positive numbers represent movement in the first-listed direction of the respective degree of freedom (e.g., "internal" for I-E rotation). For V-V rotation, varus is the first rotation. Degrees of freedom where the flexion angle effect was statistically significant are denoted by * ($p<0.05$). Within each degree of freedom, different letters (i.e. "C" versus "D") indicate a statistically significant difference between the corresponding flexion angles in that pair. Either no letter or the same letters indicate no statistically significant difference.

ACL-deficient knee				
Flexion angle (deg)	I-E rotation* ($p=0.0092$)	V-V rotation ($p=0.0971$)	M-L translation ($p=0.3755$)	A-P translation* ($p=0.0077$)
0	−0.3±6.3 (19.3)[C]	−0.5±1.5 (4.3)	−0.5±2.2 (5.7)	7.2±6.5 (21.9)[C]
15	−0.1±8.2 (24.3)[C]	0.2±3.3 (11.1)	0.4±3.7 (11.0)	15.5±14.8 (44.0)[D]
30	−9.9±6.3 (26.4)[D]	−3.3±4.4 (11.8)	0.7±2.8 (7.5)	17.7±12.4 (36.0)[D]

cantly affected peak translations and rotations in all degrees of freedom in the intact knee and two degrees of freedom (I-E rotation and A-P translation) in the ACL-deficient knee. We also found that anterior translation in the ACL-deficient knee was greater than that in the intact knee by 4.7 mm on average (marginally not significant) but that coupled motions in the other three degrees of freedom were comparable.

To determine whether the coupled motions observed in the present study are large enough to affect the tibiofemoral contact pressure distribution, the magnitude of the coupled motions can be compared with the tibiofemoral contact area. The tibiofemoral contact area has been measured to be approximately 3.5 cm^2 in the medial compartment and approximately 2.25 cm^2 in the lateral compartment at 30 deg of flexion [28]. If the shape of the tibiofemoral contact area in each compartment is assumed elliptical in the horizontal plane with an A-P depth (major axis of the ellipse) twice as long as the M-L width (minor axis of the ellipse) [29,30], then the corresponding dimensions of the tibiofemoral contact area are approximately 30 mm × 15 mm in the medial compartment and 24 mm × 12 mm in the lateral compartment.

Considering the translational degrees of freedom, the coupled anterior translation (10.2 mm at 30 deg of flexion) is 34% of the A-P depth of the medial tibiofemoral contact area and 43% of the depth of the lateral tibiofemoral contact area. Coupled medial translation (2.3 mm at 30 deg of flexion) is 15% of the M-L width of the medial tibiofemoral contact area and 19% of the width of the lateral tibiofemoral contact area. The magnitudes of the observed coupled translations are substantial relative to the dimensions of the tibiofemoral contact area.

Considering next the rotational degrees of freedom, I-E rotation occurs about a longitudinal axis that approximately passes through the spherical center of the medial femoral condyle [30,31]. In addition, the spherical centers of the medial and lateral femoral condyles are approximately 46 mm apart [30,32]. Therefore, coupled internal rotation (3.8 deg at 0 deg of flexion) corresponds to an anterior shift in the lateral compartment relative to the medial compartment of 3.0 mm (sin 3.8 deg × 46 mm) or 13% of the A-P depth of the lateral tibiofemoral contact area. The magnitude of the resulting displacement from I-E rotation is substantial relative to the tibiofemoral contact area.

V-V rotation occurs about an A-P axis that runs through the medial flexion facet center [31]. Therefore, coupled varus rotation of 1.4 deg corresponds to 1.1 mm (sin 1.4 deg × 46 mm), which translates into less compression in the lateral compartment.

Because the coupled motions are substantial for both the intact knee and ACL-deficient knee compared to the dimensions of the tibiofemoral contact areas, these degrees of freedom should not be constrained in mathematical models for studying tibiofemoral contact under compressive load. A-P and M-L translations can affect tibiofemoral contact by translating the contact pressure distribution in the transverse plane. I-E rotation can affect tibiofemoral contact by rotating the contact pressure distribution in the transverse plane. V-V rotation can directly affect compressive forces in the medial and lateral tibiofemoral contact areas by increasing compression in one compartment while decreasing compression in the other. Donahue et al. demonstrated that constraining V-V rotation at 800 N of compressive load in a finite element model increased contact force in the lateral compartment by 14% and decreased contact force in the medial compartment by 15% [5]. Similarly, mean pressure increased 7% in the lateral compartment and decreased 11% in the medial compartment.

These degrees of freedom also should not be constrained when designing mechanical testing equipment. We are currently designing an apparatus to be placed in the bore of a magnetic resonance imaging (MRI) scanner that will cyclically load knee specimens to determine the deformation of articular cartilage [33]. Ideally, the apparatus should be as simple as possible because of the constraint on size and materials. However, based on the results of this study, constraining certain degrees of freedom to reduce the complexity of the apparatus may lead to inaccurate results. Thus, as many degrees of freedom should be included in the testing apparatus as practical.

The coupled motions observed in the intact knee can be explained to some extent by the shapes of the tibial and femoral articular surfaces. Coupled anterior translation was most likely caused by the slope of the tibial plateau. In the sagittal view, the anterior portion of the medial tibial plateau slopes proximal and anterior approximately 11 deg relative to the transverse plane while the surface of the medial femoral condyle is a circular arc [30]. Due to the slope of the medial tibial plateau, the vector of the normal force on the tibia during compression is slightly angled with respect to the transverse plane. Thus, an anterior component of this vector would develop leading to anterior translation of the tibia.

The increase in coupled anterior translation when the flexion angle is changed from 0 deg to 15 deg and 30 deg (Table 1) might be due to the increase in anterior laxity that occurs in the intact knee as the knee is flexed from 0 deg to 30 deg [34]. This increase in laxity occurs because the posterorlateral bundle of the ACL, which is taut in extension thus loading the ligament, increases in length as the knee is moved from full extension into flexion [35], hence unloading the ligament to near zero force at 15 deg and 30 deg [36].

Table 3 Peak translations and rotations in each degree of freedom for the intact knee and ACL-deficient knee referenced to the unloaded position of the intact knee and averaged over all flexion angles under a 1600 N compressive load. Numbers shown are average peak values±standard deviation. Positive numbers represent movement in the first-listed direction of the respective degree of freedom (e.g., "internal" for I-E rotation). For V-V rotation, varus is the first rotation. Rotations are expressed in degrees; translations are expressed in millimeters.

Knee condition	I-E rotation ($p=0.4671$)	V-V rotation ($p=0.7161$)	M-L translation ($p=0.1713$)	A-P translation ($p=0.0696$)
Intact	0.2±7.0	−0.1±2.9	2.0±1.3	8.0±4.3
ACL deficient	−1.4±10.4	−0.5±4.0	0.8±3.3	12.7±11.8

It was noted that the coupled motions observed during compression did not return to zero after removing the compressive load, particularly in rotational degrees of freedom (Fig. 1). The most likely explanation is that there was some free play in the coupled degrees of freedom, particularly the I-E and A-P degrees of freedom. It has been shown that I-E rotation [37,38] and A-P translation [39] each have a region of free play where the rotatory stiffness and translational stiffness, respectively, are essentially zero. Also, there was a small amount of friction in the loading apparatus when the knee was not loaded [21].

The results of the present experimental study are consistent with those of a previous computational study, which determined coupled motions of the intact knee under compressive load. With the knee in full extension, posterior translation of the femur of approximately 3 mm occurred in the intact knee under a compressive load of 1000 N [4]. In the present study, we found anterior translation of the tibia (equivalent to posterior translation of the femur) of approximately 4 mm at 1000 N of compressive load with the intact knee in full extension (Table 1 and Fig. 6).

In comparing coupled motions between intact and ACL-deficient knees, we found a strong trend where anterior translation under a compressive load was 4.7 mm greater on average in ACL-deficient knees than in intact knees. This trend was marginally not significant because of the high variability in the peak value for the ACL-deficient knee (Table 3). Accordingly, if the sample size were increased, then it is likely that this difference would become significant statistically. If so, then this difference indicates that a compressive load generates a coupled anterior force applied to the tibia that the ACL resists. A previous computational study of tibiofemoral contact found that tension in the ACL increased with greater compressive load due to femoral posterior displacement (or tibial anterior displacement) and reached 117 N at 1000 N of compression [4].

Recognizing that any comparison of our results to those of human gait should be made with caution because muscle forces were not applied in our tests, the result that coupled anterior translation was greater in the ACL-deficient knee than that in the intact knee is consistent with a previous in vivo experimental study of human gait. The maximum anterior translation was significantly greater in the injured knee (8.3 mm) than in the noninjured contralateral knee (6.2 mm) and uninjured controls during the stance phase of walking (4.8 mm) [40]. Because compressive load is a major component of walking, the increased anterior translation found in the present study for the ACL-deficient knee could be a primary cause of the increased anterior translation found by Kvist and Gillquist for injured knees. The increased anterior translation of ACL-deficient knees in walking may translate into a shift in the tibiofemoral contact area and a corresponding higher risk of developing OA [2], which, in fact, has been documented clinically for people with an ACL injury [20].

In summary, our study showed that coupled motions in I-E rotation, V-V rotation, M-L translation, and A-P translation under a compressive load are substantial relative to the tibiofemoral contact area for both the intact knee and ACL-deficient knee. Accordingly, these degrees of freedom should be included when either developing mathematical models or designing mechanical testing equipment for studying tibiofemoral contact in the intact knee and ACL-deficient knee. In addition, the present study showed that coupled anterior translation increased by almost 5 mm in the ACL-deficient knee, suggesting that the ACL plays an important role in restraining coupled anterior translation.

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