MRI Analysis of In Vivo Meniscal and Tibiofemoral Kinematics in ACL-Deficient and Normal Knees

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ABSTRACT: The objectives of this study were to analyze simultaneously meniscal and tibiofemoral kinematics in healthy volunteers and anterior cruciate ligament (ACL)-deficient patients under axial load-bearing conditions using magnetic resonance imaging (MRI). Ten healthy volunteers and eight ACL-deficient patients were examined with a high-field, closed MRI system. For each group, both knees were imaged at full extension and partial flexion ($\sim 45^\circ$) with a 125N compressive load applied to the foot. Anteroposterior and medial/lateral femoral and meniscal translations were analyzed following three-dimensional, landmark-matching registration. Interobserver and intraobserver reproducibilities were less than 0.8 mm for femoral translation for image processing and data analysis. The position of the femur relative to the tibia in the ACL-deficient knee was 2.6 mm posterior to that of the contralateral, normal knee at extension. During flexion from 0° to 45°, the femur in ACL-deficient knees translated 4.3 mm anteriorly, whereas no significant translation occurred in uninjured knees. The contact area centroid on the tibia in ACL-deficient knees at extension was posterior to that of uninjured knees. Consequently, significantly less posterior translation of the contact centroid occurred in the medial tibial condyle in ACL-deficient knees during flexion. Meniscal translation, however, was nearly the same in both groups. Axial load-bearing MRI is a noninvasive and reproducible method for evaluating tibiofemoral and meniscal kinematics. The results demonstrated that ACL deficiency led to significant changes in bone kinematics, but negligible changes in the movement of the menisci. These results help explain the increased risk of meniscal tears and osteoarthritis in chronic ACL deficient knees. J Orthop Res 24:1208–1217, 2006
Keywords: knee; anterior cruciate ligament; meniscus; magnetic resonance imaging; in vivo kinematics

INTRODUCTION

The anterior cruciate ligament (ACL) is a commonly injured ligament in the knee. In 1991, it was estimated that this injury affected roughly 1 in 3,000 people per year in the United States alone.[1] The incidence is even higher now with a more active and healthy population involved in sporting and high-risk activities.[2] ACL ruptures, which mostly affect the young and active population, can cause severe pain and instability leading to subsequent damage to other soft tissue structures such as the meniscus and articular cartilage, which in turn may lead to premature onset of osteoarthritis.

The most common clinical assessment of ACL injury is a knee joint laxity examination (e.g., using a KT-1000), which measures anteroposterior (AP) displacement of the tibia relative to the femur.[3] Gait analysis can also be used to determine dynamic changes in the knee by calculating changes in joint moments during gait.[4–6] Both of these assessment techniques, however, measure the relative movement between the bones and do not evaluate soft tissue movement. The meniscus is a critical component of the knee joint, distributing axial load from the femoral condyles on the tibial plateau. Alterations in meniscal kinematics due to ACL injury, therefore, may be associated with further meniscal tears, which in turn may also predispose the knee to the early onset of osteoarthritis.

Magnetic resonance imaging (MRI) has significantly changed the diagnosis of soft tissue pathology. Recently, advances in MRI have demonstrated good image quality of both bone and soft tissue,

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enabling noninvasive, in vivo analysis of knee kinematics.[7–12] Von Eisenhart-Rothe et al.[12] used an open MRI system with flexion/extension torques applied to the tibia at 30° and 90°. They found an increase in posterior translation of the medial femoral condyle relative to the tibial plateau centroid in ACL-deficient patients compared to uninjured knees. However, the meniscal movement was not different between ACL-deficient knees and controls. Logan et al.[8] examined tibiofemoral kinematics in ACL-deficient knees using an open MRI system under weight-bearing conditions (squatting) and found that the total anterior translation of the lateral femoral condyle was the same in ACL-deficient and normal subjects during flexion. The position of the lateral femoral condyle, however, was subluxed posteriorly in ACL-deficient knees at all flexion angles compared to normals. Scarvell et al.[13] measured the movement of the center of the contact area in ACL-deficient knees in a closed MR scanner under axial loading conditions. They also found the center of contact area on the lateral tibial plateau to be more posterior in ACL-deficient knees than in normals during flexion from 0° to 90°.

To investigate how ACL injury affects the meniscus, a simultaneous analysis of tibiofemoral and meniscal kinematics under load-bearing conditions is necessary. The objectives of this study were to establish an in vivo analytical method to determine meniscal and tibiofemoral kinematics using high-field MRI, and to compare tibiofemoral and meniscal kinematics between right and left legs of subjects with uninjured, normal knees and between injured and contralateral uninjured knees in subjects with ACL deficiency.

METHODS

Eight ACL-deficient patients (ages: 30.8 ± 7.7 years), and 10 normal volunteers with no major knee injuries (age: 28.6 ± 3.1 years) were included in the study. The inclusion criteria for patient recruitment were isolated ACL deficiency with no other knee ligament injuries, no meniscal tears, no diffuse or focal cartilage injuries, and no injuries in the contralateral knee. None of the volunteers had a history of knee pain, injury, or arthritis. Informed consent was obtained from the patient after the nature of the exams had been fully explained. All examinations were approved by and performed in accordance with the rules and regulations of the local human research committee.

Magnetic resonance imaging scans were acquired with a SIGNA 1.5T echo-speed system (GE Medical Systems, Waukesha, WI) and a dual phased-array coil (USA Instruments, Cleveland, OH). For each subject, sagittal images of both the ACL-deficient and contralateral intact knees were obtained using a fast spin echo sequence (FSE) with TR of 3000 ms, TE of 9.1 ms, 16 echo train length, 512×512 pixels over 16-cm field of view (in-plane resolution of 0.3 mm), and 1.5-mm thick sections with zero spacing (out-of-plane resolution of 1.5 mm). The receiver bandwidth was 64 kHz. The three-dimensional scan was composed of 66 sagittal slices, resulting in a scan time of approximately 10 min. Images were obtained in full extension ($-9.9° \pm 4.7°$) and flexion ($42.7° \pm 8.3°$), while the patient lay supine on a custom-made, MR-compatible, load-bearing apparatus, which applied a 125 N axial load to the foot (Fig. 1). To reduce knee motion during the scan, a separate knee holder was designed to guide the knee laterally and axially with the foot positioned in neutral alignment.

To assess the accuracy of using the landmark-matching registration software, two image volumes were randomly selected. Each of the images was manipulated with three known transformations: pure translation, in-plane rotation, and out-of-plane rotation. Each transformation was performed three times by two independent observers (K. L. and P. P.). Pure translation was randomly selected in which the x and y components (in the sagittal plane) were between ± 0.31 mm and the z component (in the slice direction) was between ± 1.5 mm. In-plane rotation about the z axis was randomly selected between 0° and 30°, and out-of-plane rotation about the y axis was randomly selected between 0° and 10°. Each transformed image was registered with its corresponding original image three times by each of two observers with 6, 8, 10, 12, and 14 landmarks. The landmarks were often insertions of tendons or ligaments, bony prominences, or in some cases irregularities in bone structure that could be found on both scans. The registration results were then compared with the known solutions. Residual distances between corresponding landmarks, and a translational error and a rotational error of the image volume were recorded. The residual landmark distances were defined as the averaged Euclidean distance between corresponding landmarks after registration. The translational error was defined as the Euclidean distance between the centers of the image volumes after registration. The rotational error was the angle between the diagonals of the image volumes after registration.

To assess reproducibility, registrations between both knees of the same subject were randomly selected from the patient group. Each registration was repeated three times by three independent observers (K. L., M. A. S., and P. P.) with 6, 8, 10, 12, and 14 landmarks. Following the procedure described below for analysis of tibiofemoral kinematics, kinematic parameters were obtained. A reproducibility coefficient (RMSE) was defined as the root mean square of the standard deviations in all cases. Additionally, three sets of images were randomly selected from the patient group; segmentation of each meniscus was repeated three times by each of three observers. Following the procedure described below for analysis of meniscal kinematics, all meniscal positions were calculated and RMSEs were obtained.

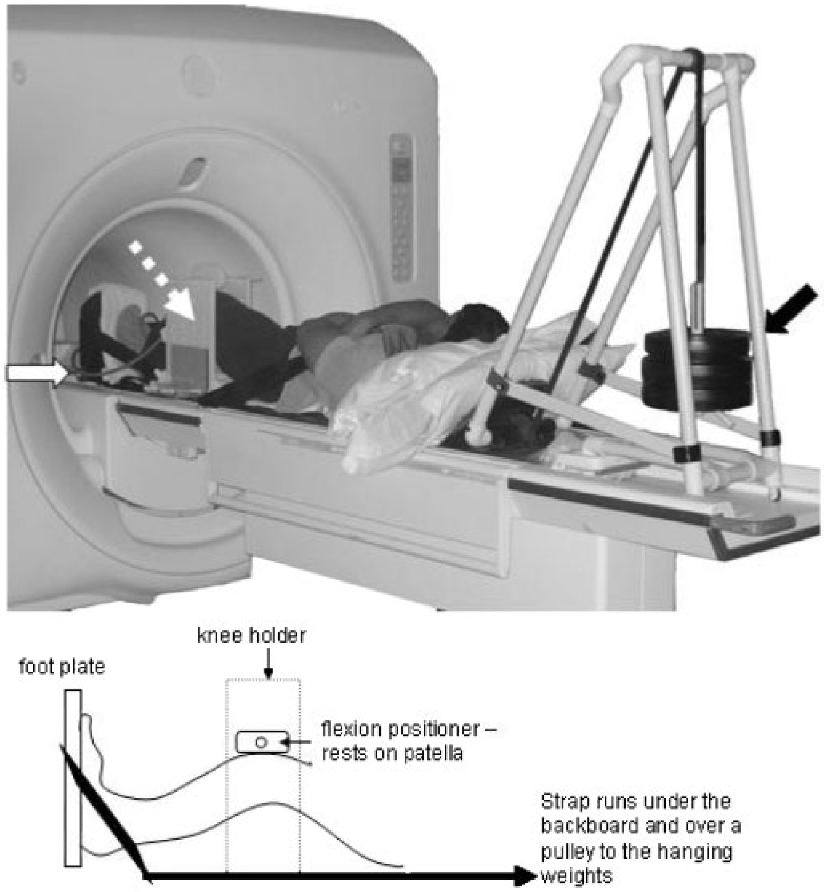

Figure 1. A custom-made, load-bearing apparatus. The patient lay supine on the table, and the knee is supported between two plates of a custom-made knee holder (dashed white arrow). The axial load is applied through the foot via a plate (white arrow) connected to the weights hanging behind the patient (black arrow).

For analysis of kinematics, we first determined alignment of the ACL-injured knee relative to the contralateral uninjured side at full extension and 45° flexion (Fig. 2, Comparison 1 and 2). Then we examined the bone (tibofemoral kinematics) and meniscal (meniscal kinematics) movements from extension to 45° flexion (Comparison 3 and 4) by assessing two static positions under axial load. In particular, we measured the AP femoral translation (A/PT), medial/lateral (ML), femoral translation (M/LT), femoral rotation (measured as translation of the medial and lateral femoral condyles), meniscal displacement, and contact area relative to a reference position. The reference position was the uninjured side when determining changes in alignment (Comparison 1 and 2) between injured and uninjured knees. The reference position was the extended knee

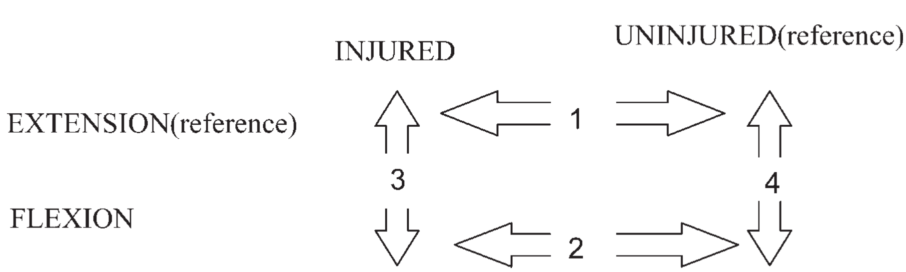

Figure 2. Alignment of the injured knee was compared to the uninjured knee at both extension and 45° flexion (1, 2). Kinematics (translation) of the bones and menisci were compared between extension and flexion (3, 4). In the normal group, the reference was always the left knee.

when determining the tibiofemoral and meniscal kinematics from full extension to 45° flexion (Comparison 3 and 4). In the volunteer group with normal knees, the reference knee was assumed to be the left knee for Comparisons 1 and 2.

We used two coordinate systems fixed in the femur and tibia. The mediolateral femoral axis (Z_f) corresponded to the optimal flexion axis and passed through the centers of the posterior femoral condyles.[14] Spheres were fit to the condyles from points selected on the sagittal bone profiles (VGStudioMax 1.2.1, Heidelberg, Germany) (Fig. 3A); the center of the sphere defined the center of the condyle. The line joining the center of the spheres was the Z_f axis, and the midpoint was the origin of the femoral coordinate system. A longitudinal line through the middle of the diaphysis was drawn in a single image slice at the center of the knee. The AP (X_f) axis was defined by the cross product of the Z_f axis with this longitudinal diaphyseal line, which was not necessarily perpendicular to the Z_f axis. The inferior-superior axis (Y_f) was the cross product of the Z_f and X_f axes (Fig. 3B). The ML tibial axis (Z_t) was determined by the line connecting the posterior tibial condyles. The midpoint was the origin of the tibial reference frame. The AP (X_t) axis was defined by the cross product of the Z_t axis with a line through the tibial shaft. The inferior-superior axis (Y_t) was the cross product of the Z_t and X_t axes.

A 3D image in-house registration program was used to align images based on an anatomical landmark-matching algorithm. The algorithm translated, rotated, and scaled images, matching corresponding anatomical landmarks by minimizing the least squared errors of the residual distances between corresponding landmarks. We used 12 tibial anatomical landmarks to align the source and reference tibias, in effect aligning the coordinate system of the tibias.

Following registration, three measures of tibiofemoral alignment were determined for all comparisons (Fig. 2): A/PT, M/LT, and AP movement of the femoral condyles. Translations and rotations of the femoral coordinate system were determined relative to the tibial reference frame. A/PT was defined as movement of the femoral reference frame origin in the X_t direction (+ = posterior). MTT was defined as movement of the femoral reference frame in the Z_t direction (+ = lateral). AP translations of the femoral condyles (represented by the center of the spheres fitted to the condyles) were also determined.

For each scan, menisci and tibiofemoral contact areas were manually segmented. Contact area was defined as the area of the tibial and femoral cartilage surfaces in contact with each other. Meniscal positions were evaluated in two sagittal slices, 20% and 80% of the ML tibial width for the medial and lateral meniscus, respectively. Because the true slices did not usually fall at these exact widths, meniscal position was interpolated from the surrounding slices (Matlab 7.0.1., Mathworks, Natick, MA). Four positions were recorded in both menisci (Fig. 4A) for a total of eight positions. By using the previously obtained registrations, the menisci and contact areas at extension and flexion were reconstructed on the tibial plateau (Fig. 4B, C). Meniscal positions were determined as percentage of the AP length of the tibial plateau in that particular sagittal slice. AP movement of the meniscus between extension and flexion was calculated as the translation of each of the eight points along the X_t axis.

The total area (mm^2) and centroid of the contact areas were determined for the medial and lateral sides. The

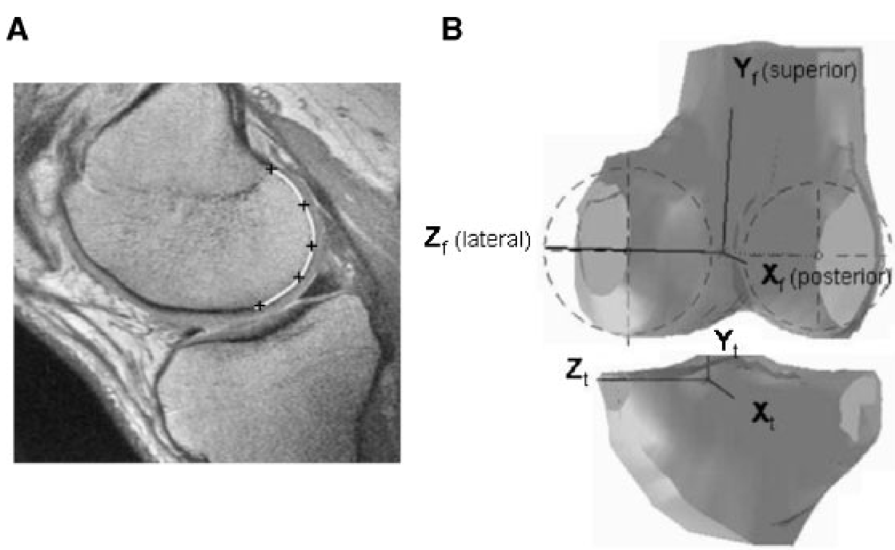

Figure 3. Tibial and femoral coordinate systems. (A) The femoral condylar axis was defined by fitting spheres to the condyles from selected points in multiple sagittal images, shown here in a representative slice. (B) Posterior view of the femur with spheres fit to condyles. The +X axis was directed posteriorly, +Y, superiorly, and +Z, laterally.

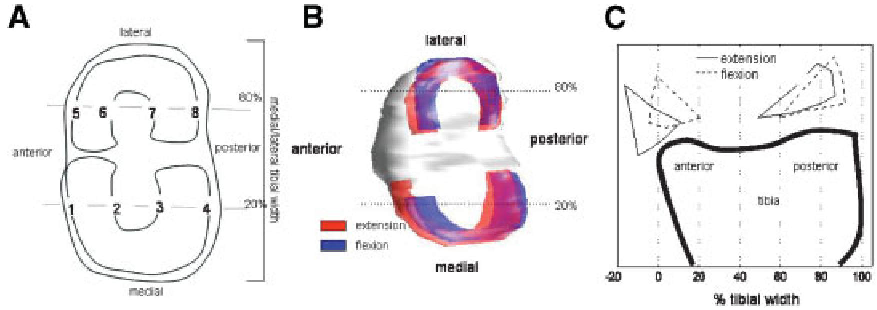

Figure 4. (A) Positions where meniscal translations were measured. (B) A representative reconstruction for an ACL-deficient knee. The menisci areas were reconstructed on the tibial plateau. The meniscus positions are shown in red for extension, and in blue for flexion. (C) A sagittal section of the medial compartment. The horns of the medial meniscus are shown on the tibial plateau. The anterior and posterior points of the tibia are given 0% and 100%, and the positions of the meniscus are reported as percentages with respect to the tibia.

midpoint (X_m, Y_m, Z_m) of the contact line in each sagittal slice was determined. The centroid was then calculated by:

$$(X_{cent}, Y_{cent}, Z_{cent}) = \frac{\sum_{i=1\ldots n} width_i(X_{mi}, Y_{mi}, Z_{mi})}{\sum_{i=1\ldots n} width_i}$$

where i was the i^{th} sagittal slice and $width_i$ was the length of the contact line in that slice. AP movement of the centroid was calculated as the translation of the centroid along the X_t axis.

One-way ANOVA was performed on the tibiofemoral, meniscal, and contact kinematics data that accounted for the side of the injury and determined significant differences between knee alignment and translations. Significance was set at $p < 0.05$.

RESULTS

The average residual landmark distances were 0.56, 0.35, and 0.87 mm for pure translation, and in-plane and out-of-plane rotations, respectively. The translational errors were 0.66, 0.26, and 1.20 mm for pure translation, and in-plane rotations (in the sagittal plane, i.e., flexion/extension) and out-of-plane rotations (i.e., axial rotation), respectively. The accuracy for pure translation was approximately twice as large as the pixel size (in-plane resolution), about one pixel size for in-plane rotation, and smaller than the slice thickness for out-of-plane rotation. Moreover, comparing the three types of transformations, out-of-plane rotations produced the largest errors and in-plane rotations had the highest accuracy.

As the number of landmarks increased, intraobserver and interobserver RMSEs decreased for all three kinematics measures. The intraobserver reproducibility coefficient for A/PT was 1.19/1.04/0.57/0.64/0.54 mm for 6/8/10/12/14 landmarks, respectively. The interobserver coefficient for A/PT was 1.24/1.25/0.76/0.76/0.71 mm for 6/8/10/12/14 landmarks, respectively. Using 10 landmarks or more, the intraobserver and interobserver RMSEs were all <0.80 mm for A/PT and M/LT. Given the in-plane resolution of 0.3 mm (field of view = 160 mm, encoding steps = 512×512), the reproducibility for tibial translation ranged between two to three pixels. Registration using 12 landmarks was chosen as our standard and used for data processing. In meniscal kinematics, the intraobserver RMSEs were $\leq$0.65 mm, and the interobserver RMSEs were all <1.15 mm for the translations of all medial meniscal positions. The intraobserver and interobserver RMSEs were all $\leq$0.55 mm for the translations of all lateral meniscal positions.

The right and left legs of the normal group at extension (Comparison 1, Fig. 2) and flexion (Comparison 2) showed no difference in AP alignment (Table 1A). In the ACL-deficient group, however, the femur on the injured side was posterior to the femur on the uninjured side at extension. At 45° flexion, no significant difference was found in the tibiofemoral alignment. There was no significant difference in ML alignment of the tibia and femur between extension and flexion for both normal and ACL-deficient groups (Table 1B). The lateral femoral condyle of the ACL-deficient knee was posteriorly subluxed at extension when compared to the contralateral uninjured knee (Table 1C). This difference was not seen in flexion. The alignment of the medial femoral condyle in the injured knee was not different than the uninjured knee in flexion or extension.

No difference was found in the femoral translation relative to the tibia in the right and left normal

Table 1. Femoral Alignment (Comparisons 1 and 2)[a]

	Normal Group (Difference between Right and Left)		ACL-Deficient Group (Difference between Deficient and Uninjured)	
A. AP alignment of femur with respect to tibia				
Extension	-0.0 ± 1.1 mm		2.6 ± 1.7 mm*	
Flexion	0.5 ± 1.5 mm		0.1 ± 1.6 mm	
B. ML alignment of femur with respect to tibia				
Extension	0.7 ± 2.4 mm		0.1 ± 1.4 mm	
Flexion	-0.1 ± 1.9 mm		$+0.1 \pm 2.2$ mm	
C. AP alignment of femoral condyles				
	medial condyle	lateral condyle	medial condyle	lateral condyle
Extension	1.4 ± 1.8 mm	-1.4 ± 2.4 mm	1.4 ± 2.9 mm	3.8 ± 2.0 mm*
Flexion	0.9 ± 2.3 mm	0.3 ± 3.3 mm	-0.1 ± 1.8 mm	0.3 ± 1.9 mm

[a]Positive values indicate posterior (or lateral) position of the femur relative to the tibia.
*Indicates significant difference ($p < 0.05$) between right/left or deficient/uninjured legs.

knees produced by movement from extension to 45° flexion. All normal data were therefore pooled (right and left knees from the normal group and contralateral uninjured knees from the ACL-deficient group). In ACL-deficient knees, significant anterior femoral translation of the injured knee was produced by movement from extension to flexion (Fig. 2, Comparison 3; Table 2). Movement from extension to flexion in uninjured knees (Comparison 4), however, produced no significant A/PT. Significant lateral translation occurred in both injured and uninjured knees, though the amount was not significantly different, implying that ACL deficiency did not affect ML translation from extension to flexion. A larger sample size, however, is needed to validate this finding. The medial femoral condyle translated anteriorly during flexion in all knees, but translated significantly more in injured knees. The lateral femoral condyle translated posteriorly during flexion in uninjured knees, but did not significantly translate in injured knees.

The contact area centroid of ACL-deficient knees was significantly posterior to the centroid of the contralateral uninjured knees in both condyles at extension (Table 3A, Fig. 5A). Significant posterior translation of the contact area centroid occurred in both medial and lateral tibial condyles in all knees during extension to flexion, except for lateral tibial condyles of injured knees (Table 3B, Fig. 5B). The medial contact area centroid in uninjured knees translated more posteriorly than injured knees. The contact centroid in ACL-deficient knees might have translated less posteriorly because they started more posteriorly at extension (Table 3A), and, therefore, did not have as far to translate during flexion.

Significant differences in meniscal position were observed between ACL deficient and uninjured knees at extension in regions 3, 6, and 7 (Table 4). No significant differences were found in meniscal position at 45° flexion between injured and uninjured knees. In general, the meniscus translated posteriorly during flexion with significant translation at positions 1, 2, 3, 5, and 6 in the uninjured knee, and 1, 2, 3, and 5 in the injured knee. In the normal group, no difference occurred in the amount of meniscal translation during flexion between

Table 2. Femoral Translations Produced by Movement of the Knee from Extension to Flexion (Comparisons 3 and 4)[a]

	A/PT	M/LT	A/P Medial Condyle Translation	A/P Lateral Condyle Translation
Injured	-4.3 ± 2.7 mm*	1.9 ± 1.8 mm*	-6.8 ± 2.3 mm*	-1.9 ± 3.5 mm
Uninjured	-0.2 ± 3.5 mm	2.3 ± 1.6 mm*	-2.8 ± 4.6 mm*	2.2 ± 3.9 mm*

[a]Positive values indicate posterior (or lateral) translation of the femur relative to the tibia.
*Indicates statistically significant translation ($p < 0.05$).

Table 3. Contact Area Centroid (A) Alignment and (B) Translation

	Normal Group (Difference between Right and Left)		ACL-Deficient Group (Difference between Deficient and Uninjured)	
A. Alignment of contact area centroid in an injured knee relative to that in an uninjured knee. Positive values indicate position posterior to reference.				
	Medial centroid	lateral centroid	medial centroid	lateral centroid
Extension	-0.6 ± 1.7 mm	-0.5 ± 0.5 mm	1.4 ± 0.7 mm*	1.4 ± 1.0 mm*
Flexion	0.1 ± 1.5 mm	-0.3 ± 2.2 mm	-0.4 ± 3.0 mm	-0.6 ± 2.3 mm
B. Translations of contact area centroid produced by movement of the knee from extension to flexion. Positive values indicate posterior translation.				
	medial centroid		lateral centroid	
Injured	3.6 ± 3.0 mm**		1.1 ± 1.9 mm	
Uninjured	6.3 ± 3.4 mm**		3.4 ± 2.1 mm**	

*Indicates statistically significant differences in alignment between deficient/uninjured legs ($p < 0.05$).
**Indicates statistically significant translation ($p < 0.05$) between extension and flexion.

right and left knees. Comparing ACL-deficient and uninjured knees during movement from extension to flexion, the only significant difference in meniscal translation occurred at the posterior border of the anterior horn of the lateral meniscus (position 6).

DISCUSSION

We developed a 3D, MR-based technique to obtain in vivo meniscal and tibiofemoral kinematics in the presence of axial load applied to the foot. Three-dimensional reconstructions of the tibia, femur, menisci, and contact area were used to determine differences in knee kinematics with ACL deficiency. Other studies analyzed only bone or meniscal kinematics using a single slice image (radiograph or sagittal MRI slice), while our method used full 3D representation, making it less prone to alignment errors. This is the first study to determine effects of ACL injury on meniscal and bone kinematics with an axial load applied to the knee using a 3D approach.

Many studies have measured tibiofemoral kinematics in vivo and in vitro using a variety of imaging modalities and loading conditions. Blankevoort et al. used stereo radiography of cadaver knees to demonstrate that axial loading induced the “screw-home” mechanism in which flexion was accompanied by internal tibial rotation.[15] Others have confirmed internal tibial rotation during flexion using digitized points[14,16] and in vivo analysis using single sagittal MRI slices.[10,17–19] Our results also indicate internal tibial rotation during flexion.

Though most studies agree that internal tibial rotation accompanies flexion, inconsistent results exist as to how the medial and lateral

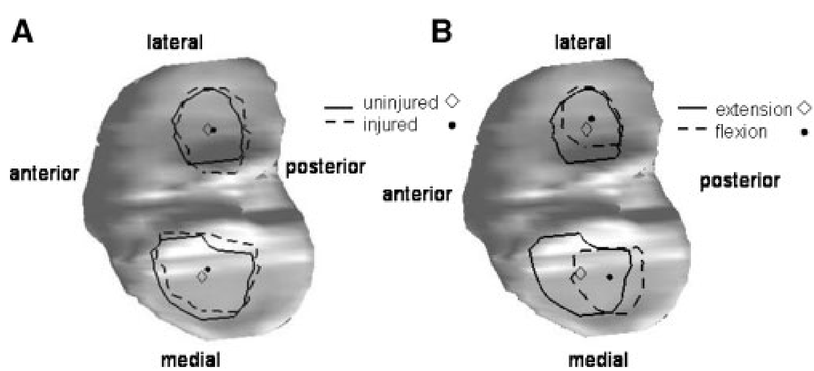

Figure 5. (A) Slight posterior alignment of the contact area centroid in an injured compared to an uninjured knee at extension for a representative subject. (B) Posterior translation of contact area centroid from extension to flexion in an uninjured knee for a representative subject.

Table 4. Meniscal Positions at Extension and Flexion for the Normal, ACL-Deficient, and Contralateral Uninjured Knees[a]

	At Extension			At Flexion		
Position	Uninjured	Deficient	Normal	Uninjured	Deficient	Normal
1	-9.7 ± 8.1	-7.0 ± 4.2	-2.1 ± 5.8	-2.4 ± 4.7	-2.1 ± 5.8	-9.3 ± 2.2
2	13.1 ± 2.8	14.8 ± 2.5	22.3 ± 5.0	20.9 ± 6.2	22.3 ± 5.0	9.1 ± 4.1
3	55.6 ± 10.1	60.4 ± 9.1*	64.4 ± 10.0	65.8 ± 7.5	64.4 ± 10.0	57.2 ± 6.4
4	94.2 ± 6.2	93.4 ± 5.7	93.8 ± 8.0	96.8 ± 5.0	93.8 ± 8.0	90.6 ± 4.1
5	11.6 ± 3.4	15.4 ± 5.0	26.0 ± 4.6	21.5 ± 2.7	26.0 ± 4.6	12.3 ± 6.0
6	39.7 ± 2.7	46.1 ± 3.6*,**	45.9 ± 7.0	44.5 ± 2.9	45.8 ± 7.0	39.4 ± 4.3
7	70.2 ± 5.7	74.1 ± 5.5*	76.1 ± 3.6	74.5 ± 3.5	76.1 ± 3.6	66.6 ± 6.3
8	95.4 ± 4.1	98.8 ± 5.8	95.1 ± 3.2	95.9 ± 2.9	95.1 ± 3.2	91.3 ± 4.2

[a]All numbers are in % with respect to the tibia AP width.
*Meniscal position in the ACL-deficient knee was significantly different when compared with the uninjured knee using paired t-tests ($p < 0.05$).
**Meniscal translation during flexion was significantly different when compared with the uninjured knee using paired t-tests ($p < 0.05$).

compartments translate to produce the relative rotation.[18,19] Differences between studies may be attributed to different coordinate systems and different measured parameters. Blankevoort et al. showed that small rotation errors of the coordinate axes around the inferior/superior (y) or AP (x) axis caused significant errors in the measured rotation.[15] Piazza et al. confirmed in a mechanical model that "cross-talk" errors arise when coordinate axes are not aligned with the flexion-extension axis.[20] Kinematics results also depend on the measured parameter. The kinematics of the condyle center, epicondyle, and contact area centroid are not the same.[10,16] Therefore, care must be taken in comparing results with other studies. Similar to other studies, we found that movement from extension to small flexion angles (45°) in a normal knee resulted in anterior movement of the medial femoral condyle center relative to the tibia and posterior movement of the lateral condyle center (Table 2).[10,16] The contact area centroid, on the other hand, moved posteriorly in both condyles with more movement in the medial condyle. It was suggested that during movement from extension to small flexion angles, the flexion axis is not through the condylar centers, and therefore movement of the condyles may have limited application.[10] Because we examined small angles, where the femur rolls on the tibial plateau, a line through the contact area centroids is a more accurate representation of the axis than the center of the condyles.

In extension, the femur in an ACL-deficient knee was aligned slightly posteriorly to that of an uninjured knee (Table 1A). No difference occurred, however, at flexion, agreeing with other results[10,17] and indicating that the ACL may aid in guiding AP position of the femur relative to the tibia, particularly at extension. This difference in alignment at extension between ACL-deficient and uninjured knees is supported by the contact kinematics. The position of the medial and lateral tibial contact centroid at extension in ACL-deficient knees is posterior to that of uninjured knees in extension but does not differ in flexion (Table 3A).

During flexion in an uninjured knee, the femur rotates externally (internal tibial rotation), which is known as the screw-home mechanism. This motion is shown in our study by the movement of the femoral condyles in the normal group (Table 2). In uninjured knees the lateral femoral condyle moved posteriorly while the medial condyle moved anteriorly, resulting in external femoral rotation. In the ACL-deficient knees, the medial condyle moved even more anteriorly while the lateral condyle did not move. This creates external femoral rotation through different means. The alteration in motion is not captured if only the rotation of the femoral (or tibial) coordinate system is examined. Tracking the motion of the condyles reveals the difference in *how* the rotation is produced.

Our meniscal kinematics results match those of others who found the meniscus translates posteriorly with flexion.[11,12,21–23] This corresponds with the movement of the contact centroid during flexion. The amount of movement of the meniscus and the contact centroid was similar, indicating that our results have likely captured the effects of posterior roll of the femur on the tibial plateau during flexion. We also found the anterior horn moved significantly more than the posterior horn, in agreement with other studies.[11,12,21–23] Von Eisenhart-Rothe et al. found that meniscal movement in ACL-deficient knees is similar to that in normal knees even though bone kinematics are different.[12] We only found a difference in meniscal translation between injured and uninjured knees at one point (#6), even though the contact centroid translation was significantly different in ACL-deficient knees. This may explain the high prevalence of meniscal tears in the posterior horns in chronic ACL-deficient knees.[24] The contact centroids of ACL-deficient knees were posterior with no significant movements of the posterior horn of the medial meniscus. Changes in meniscal kinematics may be different with other loading conditions. We are currently investigating the effects of rotation about the long axis of the tibia on tibiofemoral and meniscal kinematics following ACL injury.

We could only examine low flexion angles because the size of the magnet bore limited the possible amount of flexion. Others examined kinematics using an open magnet where higher flexion angles and weight-bearing (squatting) are possible.[11,12,17,24] Image resolution in these machines, however, is not as high as in closed bore magnets. Previous studies showed that ACL deficiency most affects kinematics at small joint angles.[25,26] In addition, the clinical pivot-shift test for ACL deficiency is performed at small angles (15°–30° flexion).[27] Our simple axial loading condition does not simulate normal gait, but it is useful to determine the effects of simple loading conditions to find which most influence meniscal translation.

Similar to previous studies of meniscal kinematics, we modeled the menisci as rigid bodies. In reality, the menisci deform, particularly with loading. Future studies will use finite element modeling of the meniscus to investigate strains under various loading conditions.[28] Future studies will also use this method to examine the effects of internal/external rotation and varus/valgus moment, a simulated pivot-shift load, on meniscal kinematics. In this study, we examined both acute and chronic ACL-deficient knees with full range of motion, assuming the kinematics were the same in acute and chronic patients. We will examine possible differences between acute and chronic ACL injury knee kinematics in subsequent studies.

In conclusion, we applied a 3D, MR-based method to analyze in vivo knee kinematics under axial load-bearing conditions. This noninvasive method allowed us to evaluate meniscal and tibiofemoral kinematics. We are currently evaluating patients with symptomatic cruciate ligament insufficiencies pre- and post-ligament reconstruction to determine whether ligament reconstructions can restore normal tibiofemoral and meniscus kinematics. We believe that the development of a more accurate 3D in vivo analysis of knee kinematics is vital to evaluate current treatment and subsequently improve outcomes following ligament injuries to the knee.

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