Abnormal Rotational Knee Motion During Running After Anterior Cruciate Ligament Reconstruction

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Background: The effectiveness of anterior cruciate ligament reconstruction for restoring normal knee kinematics is largely unknown, particularly during sports movements generating large, rapidly applied forces.

Hypothesis: Under dynamic in vivo loading, significant differences in 3-dimensional kinematics exist between anterior cruciate ligament–reconstructed knees and the contralateral, uninjured knees.

Study Design: Prospective, in vivo laboratory study.

Methods: Kinematics of anterior cruciate ligament–reconstructed and contralateral (uninjured) knees were evaluated for 6 subjects during downhill running 4 to 12 months after anterior cruciate ligament reconstruction, using a 250 frame/s stereoradiographic system. Anatomical reference axes were determined from computed tomography scans. Kinematic differences between the uninjured and reconstructed limbs were evaluated with a repeated-measures analysis of variance.

Results: Anterior tibial translation was similar for the reconstructed and uninjured limbs. However, reconstructed knees were more externally rotated on average by 3.8 ± 2.3° across all subjects and time points (P = .0011). Reconstructed knees were also more adducted, by an average of 2.8 ± 1.6° (P = .0091). Although differences were small, they were consistent in all subjects.

Conclusions: Anterior cruciate ligament reconstruction failed to restore normal rotational knee kinematics during dynamic loading.

Clinical Relevance: Although further study is required, these abnormal motions may contribute to long-term joint degeneration associated with anterior cruciate ligament injury/reconstruction.

Keywords: knee; anterior cruciate ligament (ACL); kinematics; reconstruction; running

The primary goals of treatment after ACL injury are the reduction or elimination of knee instability, the restoration of lost function, and the prevention of long-term joint degeneration. Although modern ACL reconstruction procedures are arguably successful for meeting the first 2 goals, long-term joint health after ACL injury remains a concern.

The ACL injury has long been associated with a high incidence of degenerative knee osteoarthritis (OA).[14,25,50] Animal models have shown a clear connection between mechanical instability and progressive OA.[4,31,32] Thus, it is logical to assume that if ACL reconstruction surgery restores knee stability, it should reduce the risk of subsequent OA. However, a similar or higher incidence of knee OA has been reported in ACL-deficient individuals who underwent ligament reconstruction as compared to those who did not.[11,15] Despite criticisms of these studies (retrospective, uncontrolled, possible selection bias between patients who choose surgical reconstruction vs those who do not), to date there is no clear evidence of a long-term protective effect from the procedure. Although other factors may contribute to OA development in the reconstructed joint (such as possible damage to other tissues at the time of injury), failure of the reconstruction to restore normal knee kinematics could be an important contributor to progressive cartilage degeneration.

The effectiveness of ACL reconstruction for restoring normal knee function has been difficult to assess. A variety of radiographic and mechanical measurement methods have been employed to assess static knee stability in vivo by measuring the anterior displacement of the tibia relative to the femur resulting from a specified anterior force

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applied to the tibia with the femur held in a fixed position. These studies have shown static instability (typically 5-10 mm) following ACL loss as well as improvement to near-normal levels following ACL reconstruction.[10,21,26,42,48] Three-dimensional (3D) radiographic techniques have also shown increased rotational laxity (internal/external tibial rotation) in ACL-deficient knees.[20] However, static stability measures have not correlated well with any known measure of functional outcome for ACL-injured subjects before or after reconstruction.[3,9,17,39,43] In addition, there are reports of individuals with statically unstable knees performing asymptomatically at high levels of athletic activity,[28,30] suggesting that static stability may not be a prerequisite for good knee function.

Why is static stability not predictive of outcome? During functional activities, the knee is subjected to a combination of dynamic physical forces (gravitational, inertial, and contact) and active muscular forces.[27,37] The combination of these forces and the constraints imposed by passive structures (articular surface geometry, ligaments, etc) lead to a complex motion path, with rotations and translations in all 3 planes of movement.[23] Static, 1-dimensional testing cannot predict the behavior of the reconstructed joint under realistic loading conditions. Thus, dynamic assessment of 3D skeletal kinematics is necessary to assess the effectiveness of ACL reconstruction surgery for restoring normal knee function.

Accurate, reliable 3D knee-motion measurement is problematic with commonly used motion analysis techniques. Conventional motion-measurement methods, employing either optoelectronic or video-based systems, rely on markers attached to the skin to estimate joint motion. Studies using implanted bone pins have shown that markers affixed to the skin shift relative to underlying bone by as much as 30 mm, particularly during rapid movements or activities involving impact such as heelstrike.[18,24,35] This marker-tracking error varies with bone, marker position, activity, and joint angle and is often correlated with the movement, complicating efforts to develop algorithms for error correction.[6] Techniques have been developed combining large numbers of skin markers with algorithms to detect violations of the rigid body assumption and model soft tissue deformation.[1,2,7,44] A single-subject validation study demonstrated that this approach can track tibial motion well during slow, nonimpact movements (low-speed 10 cm step-up; average error = 0.8 mm, peak error = approximately 2.6 mm).[1] However, the performance for tracking the femur, where errors are likely to be higher because of greater soft tissue thickness between markers and bone, has not been reported. Validation studies during faster movements including impact events (eg, gait, running, jumping), where larger magnitude and more complex skin deformation occurs, are not available.

It is possible to avoid skin motion artifacts by directly measuring bone movement, either by physically attaching measurement devices to the bone or via medical-imaging techniques. Accurate 3D in vivo kinematics studies have been performed using external marker arrays rigidly fixed to bone.[23] However, the risks and discomfort associated with this technique have prevented widespread use and made serial studies difficult. Dynamic MRI and CT methods show promise[36,41] but are limited by low frame rates and environments too restrictive for most dynamic, weightbearing activities.

Radiographic imaging offers a minimally invasive alternative for knee-motion assessment. Conventional fluoroscopy permits direct visualization of bone motion but is limited to 2-dimensional assessment and is prone to errors due to parallax and motion blur. Biplane or stereo radiographic imaging enables accurate quantitative 3D motion assessment as well as direct visualization of bone motion. Use of biplane radiographic film methods (radiographic stereophotogrammetric analysis [RSA]) for 3D studies of static bone position has been well established,[21,38] with precision reported in the ±10 to 250 μm range.[19] We have developed a dynamic RSA system for accurate assessment of dynamic joint motion. This system (described briefly in the Methods section below and in detail elsewhere[45]) is well suited for dynamic knee-motion measurement, with 3D accuracy in the order of ±0.1 mm and rates up to 1000 frames/s.

The goal of this study was to use this high-speed biplane radiography system to assess the 3D kinematics of healthy and ACL-reconstructed knees during dynamic, functional activities. It was hypothesized that accurate 3D measurement would reveal differences in dynamic joint motion between reconstructed and healthy joints. If true, these findings might help to explain why reconstruction is not protective against long-term joint degeneration associated with ACL injury. With this information, it may become possible to design improved reconstruction techniques that restore more normal joint motion.

MATERIALS AND METHODS

Subjects undergoing unilateral, primary arthroscopic ACL reconstruction and who were between the ages of 16 and 50 years were recruited for the study. Exclusion criteria included any prior significant injury to the contralateral limb as well as significant damage to other knee structures in the ACL-injured limb (subjects with minor meniscal tears, requiring removal of no more than one third of the radial width of the meniscus, were not excluded). All patients who were not excluded by these criteria and who underwent ACL reconstruction by 1 of the 3 participating surgeons were asked to participate in the study. Informed consent was obtained from all enrolled subjects, and the protocol was approved by the appropriate institutional review board for human subject research. During the ACL reconstruction procedure, tantalum spheres (1.6-mm diameter) were inserted into the distal femur and proximal tibia of both limbs using a cannulated drill. These markers provided high-accuracy radiographic targets for RSA and have been used similarly in many human studies with no adverse reactions reported.[19] Three noncollinear markers were inserted into each bone to enable full 6 degrees of freedom tracking. The goal of marker placement was to ensure a minimum 2-cm separation between markers and to avoid overlap between markers or with any implanted hardware in the radiographic views. No effort was made to place the markers at specific anatomical landmarks.

Kinematic testing was performed as soon as possible after the subject completed his or her rehabilitation program and was cleared by the surgeon for return to light sports activities. This time period was typically 4 to 6 months but was 12 months for 1 subject (the first in the study; data collected at 4 months for this subject were unusable due to a technical problem). Within 2 weeks of the kinematic testing, static knee laxity was assessed (using a KT-1000 arthrometer at 89 N anterior force) for comparison with the dynamic measurements.

Subjective knee function was assessed with the Cincinnati Knee Ligament Rating System.[29] The 4 reported values in the Cincinnati Knee Ligament Rating System are (1) current sports activity level, (2) ability to participate in sports, (3) functional assessment rating scale, and (4) rate of pain, swelling, and giving way. All were based on a 0% to 100% scoring system, with 100% being maximum function/minimum pain. In a comparison of several systems (applied to 65 patients at a mean of 35 months after ACL reconstruction), the Cincinnati system was recommended because it was “the most elaborate and detailed assessment of activity-related symptoms and function,” and it “most precisely defines outcome in athletically active patients.”[40]

Moderate-speed downhill running was selected as the movement activity for study. This task was selected because it is more stressful on the ACL than level-ground running.[22] In addition to placing greater mechanical demands on the knee than walking, running also eliminates the double-support phase and reduces the effects of compensation from the contralateral limb. It can be performed in a controlled, repeatable fashion within the laboratory environment and is unlikely to put the individual at significant risk for injury. Downhill running has been used previously to assess performance in ACL-deficient individuals.[47] A moderately slow running speed (“jog”) was selected to ensure that all subjects would be able to perform the task. Testing was performed at 2.5 m/s on a standard treadmill (Landice L8, 46 × 152 cm belt, Landice Inc, Randolph, NJ) with the rear supports elevated 25 cm to provide a 10° downward slope (Figure 1). For each trial, kinematic data were collected from shortly before footstrike through midstance for 1 step of the test leg (approximately 0.5 seconds duration), using an electronic timer system and an accelerometer strapped to the shank to detect footstrike.

Knee kinematics were assessed with dynamic RSA. The RSA is a technique for determining 3D kinematic information from stereo-pair radiographic images of musculoskeletal tissue with implanted high-contrast markers.[38] This method has been successfully employed to determine static or slow-moving 3D behavior of a variety of human joints and joint structures, using pairs of plain film radiographs.[19,38] Similar techniques were employed for the study of dynamic knee movement, replacing radiographic

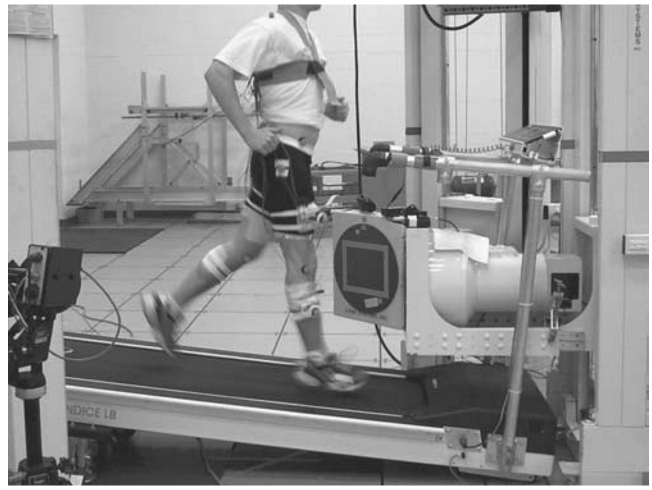

Figure 1. High-speed biplane radiographic system, configured for downhill treadmill running. Images are acquired simultaneously at 250 frames/s for the 2 views (60° separation).

film with high-speed digital imaging. The dynamic RSA system consisted of 2 gantries (each containing a 150 kW x-ray source, 30-cm image intensifier, and 250 frame/s digital video system) configured to provide 2 beams parallel to the ground, with an interbeam angle of 60°. This provided sufficient open space to incorporate a standard treadmill and a 3D-imaging area (where the beams intersect) approximately 45 cm wide by 30 cm long by 25 cm high (Figure 1). This system is capable of tracking implanted markers with an accuracy of approximately ±0.1 mm, as previously described.[45]

Two-dimensional marker positions were automatically determined for each pair of digital images and passed to a commercial 3D-tracking software package (Eva, Motion Analysis Corp, Santa Rosa, Calif) for calibration, tracking, and 3D reconstruction. The resulting 3D marker coordinates were smoothed using a fourth-order, zero-lag Butterworth low-pass filter with a cutoff frequency of 25 Hz (selected by residual analysis of several trials[49]).

Transformations between implanted marker-based coordinates and anatomical axes/landmarks were determined from CT (Figure 2), as previously described.[45] Rotations of the tibia relative to the femur were calculated using body-fixed axes in the order (flexion/extension, adduction/abduction, internal/external rotation) corresponding to the rotational component of the joint coordinate system originally described by Grood and Suntay.[16] Translations for the reconstructed joint were determined relative to ACL graft origin/insertion, estimated by identifying the center of the tibial and femoral tunnels at the joint surface of 3D bone models generated from subject-specific CT scans. To minimize side-to-side variability, the same locations were used for the uninjured limb by using a mirror image of the reconstructed limb CT and coregistering it with the uninjured limb CT. Displacements of the tibia relative to the femur were measured from ACL origin to insertion and

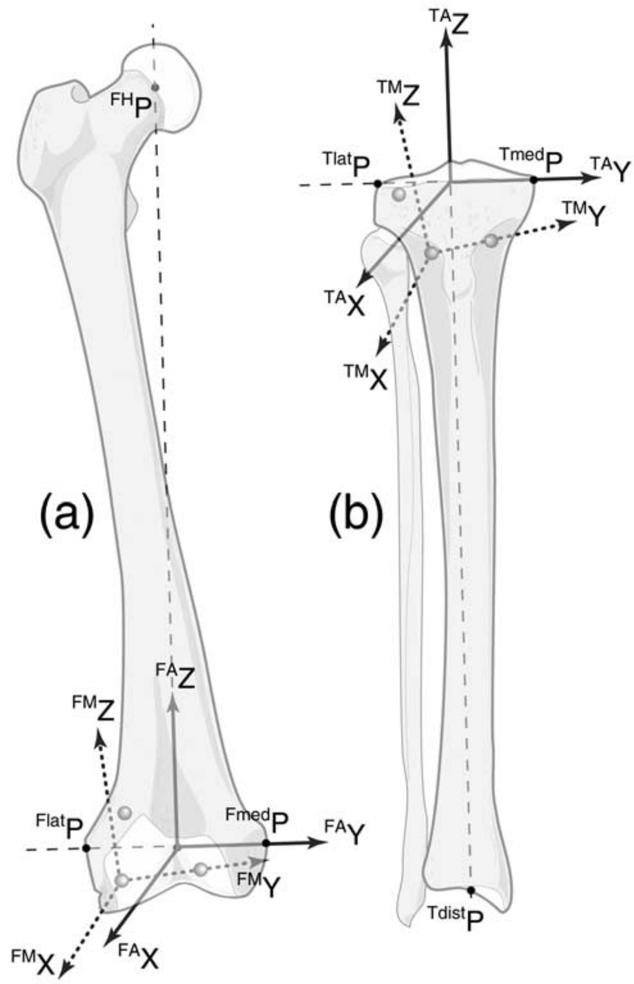

Figure 2. Coordinate system definitions. Anatomical landmarks were identified using surface reconstructions from subject-specific CT scans. Fixed transformations between marker-based ("M" superscripts) and anatomical ("A" superscripts) coordinate systems were also determined from CT. Translations were expressed in the anatomical coordinate system fixed to the tibia.

expressed in an orthogonal anatomical coordinate system fixed to the tibia. Functional ACL-graft length was estimated as the magnitude of the origin-to-insertion vector.

Rotation and displacement curves were ensemble averaged over all valid trials (typically 3) for each test session. Kinematic variables (3 rotations, 3 translations, ACL-graft length) were extracted at 0.0, 0.025, 0.05, 0.075, and 0.10 seconds after footstrike for statistical analysis. Within-subject differences between the intact and reconstructed limbs were evaluated via repeated-measures analysis of variance (significance set at $P = .05$).

RESULTS

A summary of subject characteristics is provided in Table 1. The subjects represented a heterogeneous mix of ages, sex, graft type, time from injury, and meniscal health. Static stability was restored in all subjects, with a mean side-to-side difference in KT-1000 arthrometer measurements of +1.2 mm (range, –1 to 2.5 mm). All subjects were able to perform the downhill-running task without difficulty, with no obvious limping or asymmetry. There were no statistically significant correlations between age, sex, graft type, injury/surgery timing, static laxity, or functional scores and any of the kinematic measures.

Joint kinematics for a typical subject are shown in Figures 3 and 4; group ensemble averages are provided in Figures 5 and 6. Knee flexion at footstrike ranged from mild hyperextension to nearly 20° of flexion in the uninjured (intact) limbs. Although differences between intact and reconstructed limbs were noted in several subjects (eg, Figure 3A) and there was a slight trend toward loss of extension in reconstructed knees (Figure 5A), there were no statistically significant differences in flexion at any time point ($P = .71$). Variability was highest in this plane, with mean across-subject standard deviations of 13° in the intact limbs and 8° in the reconstructed limbs.

Significant differences between limbs were identified in the other 2 rotational axes (Figures 5B and C). Reconstructed knees were more externally rotated by 3.8 ± 2.3° (mean ± SD) across all subjects and time points ($P = .0011$). Reconstructed knees were also more adducted (by a mean of 2.8 ± 1.6°; $P = .0091$). Although these differences were small, they were observed consistently in all 6 subjects. As shown in Figure 5, the shapes of the motion curves were similar for both limbs, but there were consistent baseline shifts toward external rotation and adduction on the reconstructed side. Across-subject standard deviations were less than 3° at all time points.

No significant differences were found in any of the joint translations. Mean anterior tibial translation curves were nearly identical for the intact and reconstructed limbs (see Figure 6C), with a mean difference across all subjects and time points of only 0.5 mm (reconstructed less than intact; $P = .75$). Three of the subjects had consistently greater anterior tibial translation in the reconstructed limb (eg, Figure 4C), whereas 3 had less. The estimated ACL-graft-length data were more consistent, with the functional graft length shorter than the functional ACL length at 1 or more time points for all subjects (and at all time points for 4 subjects), with a mean decrease of –0.9 ± 1.48 mm across all time points and subjects. This trend is visible in the mean curve (Figure 6D) but was not statistically significant ($P = .20$).

Four of the 6 subjects had greater lateral tibial translation in the reconstructed limbs. Although this trend was apparent in the group mean curve (Figure 6A), it was not significant for this sample size (mean difference = 1.12 mm, $P = .12$). No trends were evident in the proximal/distal translations (Figure 6B).

Translation magnitudes varied considerably across subjects, with between-subject standard deviations ranging from 2 to 6 mm. This was due in part to geometric differences (eg, size) between the joints, creating baseline shifts in the translation curves. Because bone shape and size were consistent across the 2 limbs of each subject, these

TABLE 1
Subject Characteristics[a]

Subject	Age	Sex	Graft Type	Additional Procedures	Injury→ Surgery, mo	Surgery→ Test, mm	KT-1000 Difference, mo	Cincinnati Knee Score (4 scales)
1	48	Female	Hamstring	Partial lateral meniscectomy	18	6	1	40/40/60/40
2	44	Female	BPTB		24	5	2	80/80/80/70
3	42	Female	BPTB		24	4	1	90/60/80/80
4	40	Male	Hamstring		24	12	–1	55/55/80/40
5	40	Female	Hamstring	Medial meniscal repair	2	4	2.5	75/60/80/60
6	24	Male	Hamstring	Partial medial meniscectomy	12	4	1.5	60/80/100/80

[a]Grafts were either quadrupled hamstring tendon with interference screw fixation or bone–patellar tendon–bone with interference screw fixation. KT-1000 arthrometer measurements were displacement of reconstructed side minus displacement of uninjured side, at 89 N force. The 4 reported values in the Cincinnati Knee Ligament Rating System (as described by Noyes et al[29]) are (1) sports activity level, (2) ability to participate in sports, (3) functional assessment rating scale, and (4) rate of pain, swelling, giving way; are all based on a 0% to 100% scoring system (with 100% being maximum function/minimum pain). KT-1000 arthrometer scoring and functional assessments were performed within 2 weeks of kinematic testing.

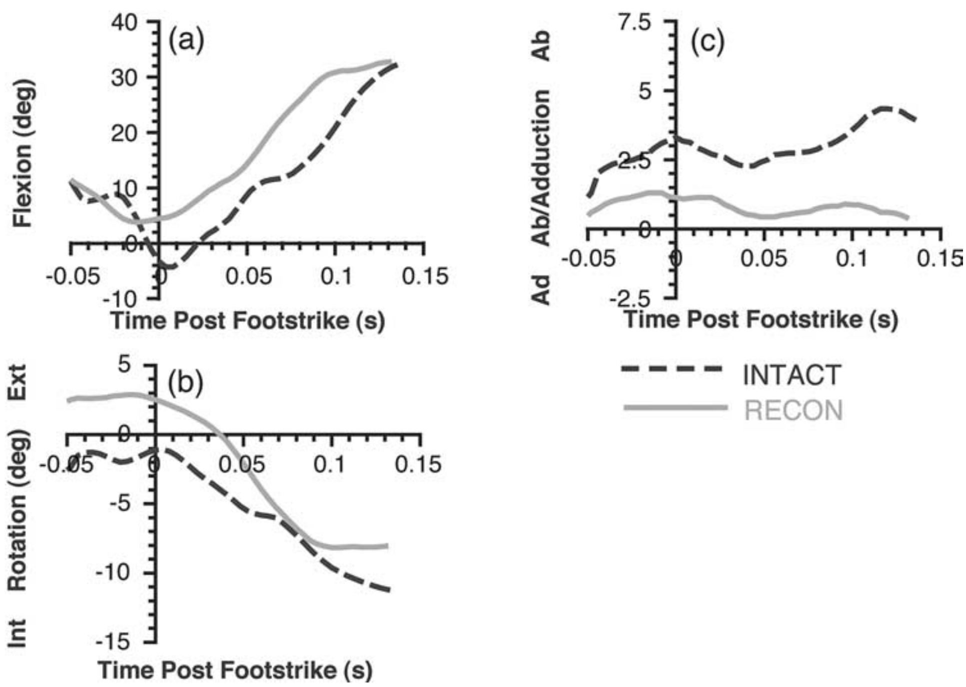

Figure 3. Joint rotational kinematics, typical subject. The dashed line is the intact (uninjured) limb; the solid line is the ACL-reconstructed limb.

geometric differences did not affect the paired statistical analysis.

DISCUSSION

The primary clinical goals for ACL reconstruction surgery are to restore stability and regain function lost due to ACL injury. The primary instability traditionally associated with ACL loss is excessive anterior translation of the tibia relative to the femur. Thus, reconstructive surgery is generally considered successful if anterior tibial translation in the reconstructed knee is similar to that in the contralateral, uninjured knee and the patient is eventually able to resume a preinjury activity level. Based on these evaluation criteria, ACL reconstruction is generally successful. The above results show that anterior instability is eliminated by the procedure under dynamic, stressful loading as well as during static (KT-1000 arthrometer) testing.

The results of this study show, however, that restoration of normal anterior/posterior motion patterns does not necessarily imply normal knee function. Consistent, statistically significant differences were found between uninjured and reconstructed knee motion for both internal/external

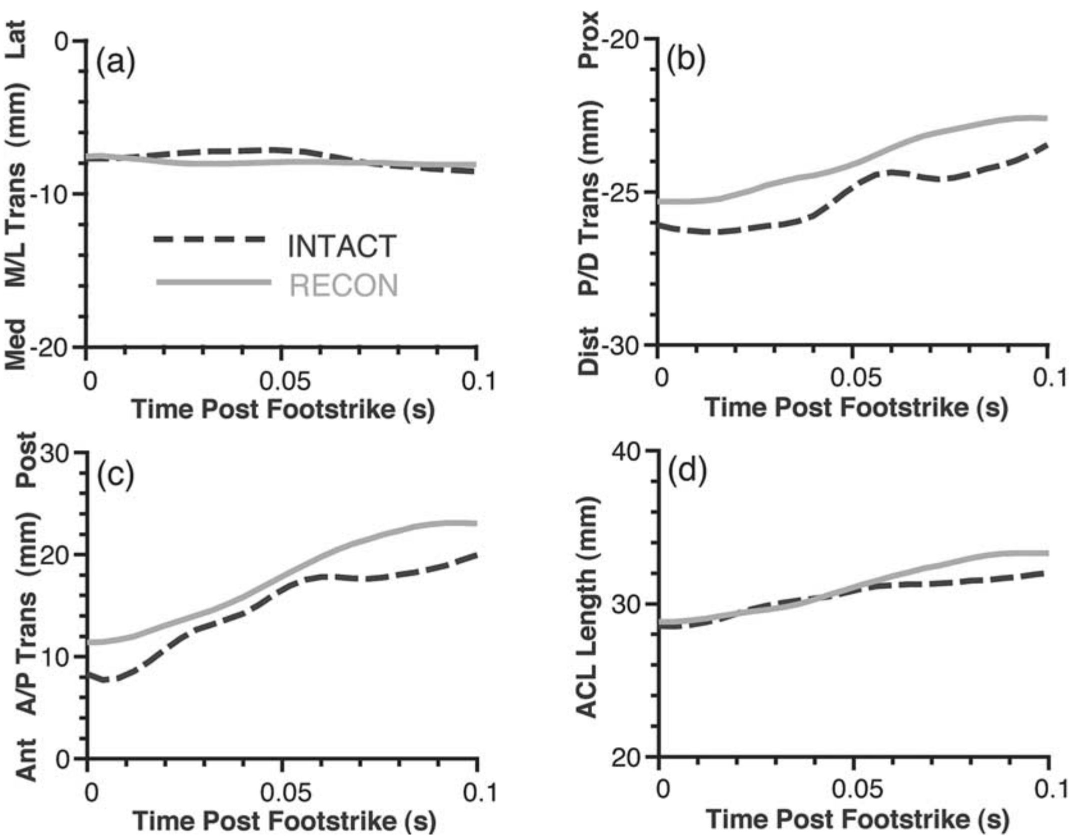

Figure 4. Joint translational kinematics, typical subject. The dashed line is the intact (uninjured) limb; the solid line is the ACL-reconstructed limb.

rotation and abduction/adduction. These differences were small in absolute terms, averaging only 4° external rotation and 3° adduction. However, these differences are large percentages of the range of observed motion about these axes (51% for internal/external rotation and 270% for abduction /adduction relative to the average range of motion in the uninjured limbs).

Three-dimensional animations of recorded tibiofemoral kinematics (using bone surfaces rendered from subject-specific CT data) provided a qualitative assessment of these rotational changes on motion at the articulating surfaces of the tibiofemoral joint. The shift toward adduction typically resulted from an increase in lateral compartment separation as well as a decrease in medial compartment separation, whereas the external tibial rotation shifted the tibial contact area primarily in the lateral compartment. If the medial compartment contact point was unchanged, a simple geometric analysis showed that a 4° increase in external rotation would shift the lateral compartment tibial plateau contact area anteriorly by approximately 3.5 mm in a typical-size joint. Similarly, assuming equal effects in both compartments, a 3° adduction would create a 1.3-mm loss of medial compartment separation and a 1.3-mm increase in lateral compartment separation. Thus, these movement pattern shifts could significantly alter the location, pattern, and magnitude of stresses applied to both cartilage and menisci.

Two other kinematic parameters—medial/lateral translation and ACL-graft length—showed trends suggesting differences between the uninjured and reconstructed limbs. Although these trends did not reach statistical significance, the possibility of true differences cannot be ruled out until more subjects are included in the analysis. The medial/lateral translation data (which were closest to significance at $P = .12$) had a power of only 0.32 to detect the observed difference of 1.12 mm. A post hoc power analysis suggested that a sample size of 25 would be required to detect a difference of this magnitude with a power of 0.8.

There is considerable support in the literature to suggest that mechanical abnormalities are associated with OA development and progression, based on both human and animal studies.[5,12,13,33,34] Knee adduction has been specifically linked to both higher incidence and faster progression of knee OA.[8] Development of OA in the ACL-deficient dog has been shown[4,32] and was thought to be driven primarily by the large anterior tibial shift created by ACL loss. However, recent high-speed radiographic studies have shown that changes in the patterns of joint contact were better predictors of OA development in this model than the magnitude of anterior/posterior instability.[46] Although the clinical significance cannot be assessed from the data presented, possible links between the rotational abnormalities detected in ACL-reconstructed knees and increased OA incidence in this population should be investigated further.

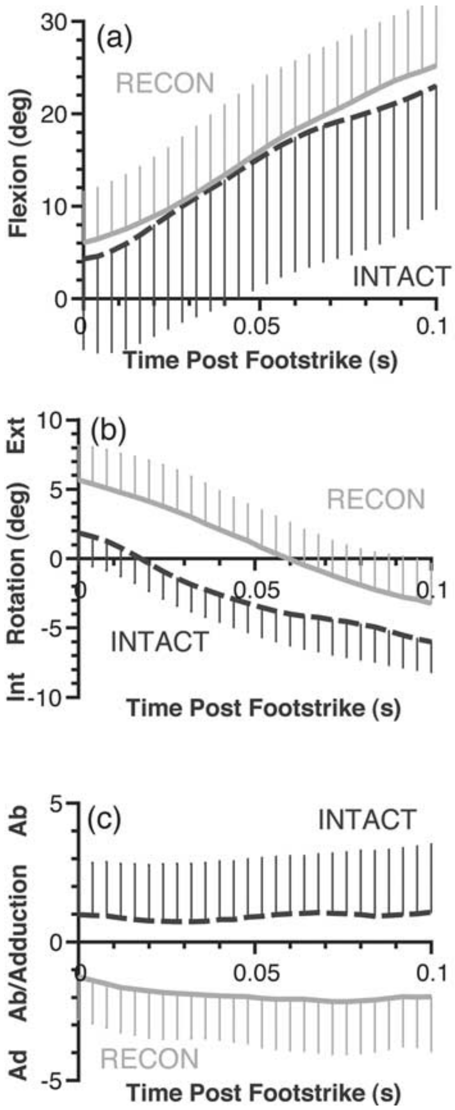

Figure 5. Joint rotational kinematics, average of 6 subjects. The dashed line is the intact (uninjured) limb; the solid line is the ACL-reconstructed limb. Vertical lines are ±1 SD.

These rotational abnormalities in ACL-reconstructed knees have not been previously reported. However, as discussed in the introduction above, the techniques previously employed to measure dynamic in vivo knee kinematics (eg, skin markers) may not have been sufficiently reliable to identify the small (3°-4°) differences found in this study. Static studies of knee stability were typically designed to test the behavior of the knee only near the limits of the passive range of motion of the joint. The results above show shifts in the "operating point" of the joint but no significant increase in the range of motion or any evidence that the joint is operating at or near the limits of passive stability. This may explain why traditional static stability measures have not been predictive of outcome.[3,9,17,39,43]

The combination of a somewhat stressful task, high-accuracy skeletal kinematics (free of skin motion artifact), subject-specific 3D bone models, and within-subject side-to-side comparisons was sufficient to detect subtle differences in dynamic joint behavior. The low across-trial standard deviations (<3° across subject; typically <1° within subject) are evidence to the reliability of this technique. Although higher across-subject variability was observed in the translational measurements (SDs ranging from 2 to 6 mm), it was due primarily to anatomical differences between the subjects. Because translations were measured relative to specific anatomical locations (estimated ACL origin and insertion), larger joints would naturally tend to have larger translation baselines (while rotations would be unaffected by joint size). Normalization of translations to joint size might have reduced this variability but would not have changed the statistical outcome (which was based on paired within-subject comparisons). Within-subject translation variability was much lower (typically less than 1 mm).

A significant limitation of this study was the heterogeneity of the subject population. Although no effects of age, graft type, meniscal injury, surgery/testing timing, or functional scoring were detected, the sample size and statistical power were probably insufficient to detect such relationships. Graft positioning was also not tightly controlled (although a coronal angle of 30° from vertical was targeted). These factors could have a significant influence on some of the results reported here, especially considering the relatively small sample size. For example, even though no significant group differences were found, translational differences were identified in some subjects. This suggests that there may be additional trends within the data, but there were insufficient subjects to link these observations to any specific subgroup. Data on all these elements have been collected (including graft positioning), from CT. As this study is ongoing, it should be possible to investigate the influence of these confounding factors as data from more subjects become available.

It is important to note, however, that rotational abnormalities reported above were consistent in both direction and magnitude even across this relatively diverse subject group. In all subjects, reconstruction failed to restore normal 3D rotational tibiofemoral motion under dynamic, stressful loading. Further research is required to determine the relationship between these residual abnormalities and early-onset knee OA and to investigate modifications to the reconstruction procedure that could reduce harmful movement abnormalities and improve long-term outcome after ACL reconstruction (eg, changes in tunnel placement, anatomical/double-bundle grafts).

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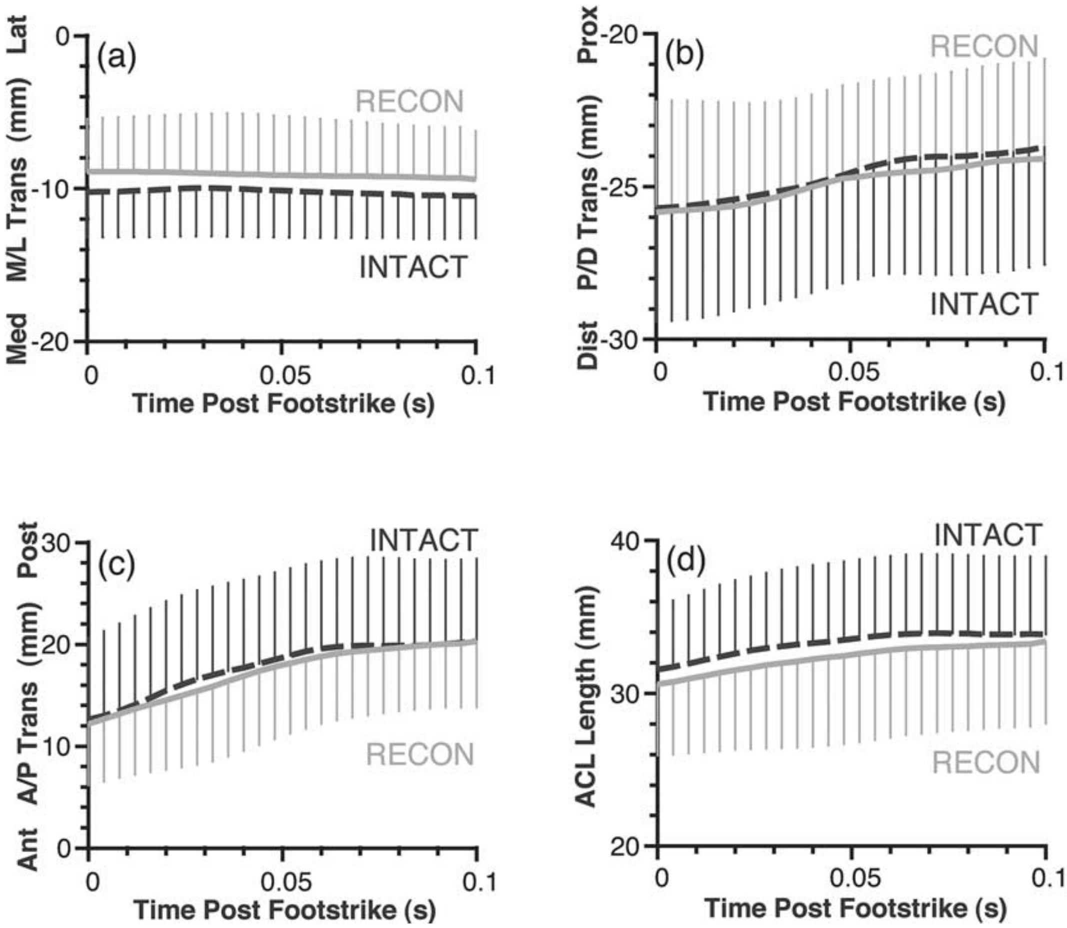

Figure 6. Joint translational kinematics, average of 6 subjects. The dashed line is the intact (uninjured) limb; the solid line is the ACL-reconstructed limb. Vertical lines are ±1 SD.

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