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American Society of Biomechanics Clinical Biomechanics Award 2017: Non-Anatomic Graft Geometry is Linked with Asymmetric Tibiofemoral Kinematics and Cartilage Contact Following Anterior Cruciate Ligament Reconstruction

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Abstract

Background—Abnormal knee mechanics may contribute to early cartilage degeneration following anterior cruciate ligament reconstruction. Anterior cruciate ligament graft geometry has previously been linked to abnormal tibiofemoral kinematics, suggesting this parameter may be important in restoring normative cartilage loading. However, the relationship between graft geometry and cartilage contact is unknown.

Methods—Static MR images were collected and segmented for eighteen subjects to obtain bone, cartilage, and anterior cruciate ligament geometries for their reconstructed and contralateral knees. The footprint locations and orientation of the anterior cruciate ligament were calculated. Volumetric, dynamic MR imaging was also performed to measure tibiofemoral kinematics, cartilage contact location, and contact sliding velocity while subjects performed loaded knee flexion-extension. Multiple linear regression was used to determine the relationship between non-anatomic graft geometry and asymmetric knee mechanics.

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Conflict of Interest Statement

The authors have no conflicts of interest to disclose.

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Findings—Non-anatomic graft geometry was related to asymmetric knee mechanics, with the sagittal plane graft angle being the best predictor of asymmetry. A more vertical sagittal graft angle was associated with greater anterior tibial translation ($\beta=0.11\text{ mm/deg}$, $P=0.049$, $R^2=0.22$), internal tibial rotation ($\beta=0.27\text{ deg/deg}$, $P=0.042$, $R^2=0.23$), and adduction angle ($\beta=0.15\text{ deg/deg}$, $P=0.013$, $R^2=0.44$) at peak knee flexion. A non-anatomic sagittal graft orientation was also linked to asymmetries in tibial contact location and sliding velocity on the medial ($\beta = -4.2\text{ mm/s/deg}$, $P=0.002$, $R^2=0.58$) and lateral tibial plateaus ($\beta = 5.7\text{ mm/s/deg}$, $P=0.006$, $R^2=0.54$).

Interpretation—This study provides evidence that non-anatomic graft geometry is linked to asymmetric knee mechanics, suggesting that restoring native anterior cruciate ligament geometry may be important to mitigate the risk of early cartilage degeneration in these patients.

Keywords

knee; anterior cruciate ligament reconstruction; cartilage; anterior cruciate ligament graft; MRI; osteoarthritis

1. Introduction

Anterior cruciate ligament reconstruction (ACLR) procedures are common, with over 150,000 occurring in the United States each year (Sanders et al., 2016). The primary goal of ACLR is to restore knee stability such that patients can return to previous levels of activity. ACLR is generally successful at meeting this short-term goal, with good or excellent outcomes reported in 90% or more of patients (Corry et al., 1999; O'Neill, 2001; Otero and Hutcheson, 1993). However, the long-term prognosis after ACLR is not as favorable, with greater than 50% of patients exhibiting radiographic evidence of post-traumatic osteoarthritis (PTOA) at 10 to 15 year follow-up (Lohmander et al., 2007).

It is theorized that osteoarthritis in ACLR patients may arise, in part, from abnormal cartilage tissue loading after surgery (Chaudhari et al., 2008; Stergiou et al., 2007). Indeed, prior studies have identified asymmetries in ACLR knee kinematics that would affect the location of cartilage contact (Decker et al., 2011; Hofbauer et al., 2014; Kaiser et al., 2017; Scanlan et al., 2010; Tashman et al., 2006). Additionally, *in vitro* studies have shown that cartilage damage can be initiated by altering the load on cartilage tissue from its homeostatic condition (Griffin and Guilak, 2005). Animal studies have also shown that abnormal cartilage loading, specifically increased tibiofemoral sliding velocity, can initiate cartilage degeneration patterns seen in PTOA (Anderst and Tashman, 2009; Beveridge et al., 2013). Thus, identifying surgical factors that can be modified to restore normal knee mechanics may help improve long-term outcomes of ACLR.

ACL graft tunnel location has been linked to abnormal post-operative tibiofemoral kinematics, suggesting that it may be an important surgical factor in ACLR. Cadaveric studies have shown that variations in femoral tunnel location lead to altered anterior tibial translation (Musahl et al., 2011; Zavras et al., 2005). Additionally, a more vertical graft increases the passive anterior and rotational laxities of the knee (Brophy and Pearle, 2009; Loh et al., 2003). *In vivo* investigations have found that graft tunnel location also effects

tibiofemoral mechanics during functional movement (Abebe et al., 2011b; Ristanis et al., 2009; Zampeli et al., 2012). For example, variations in femoral tunnel placement have been associated with greater anterior tibial translation, medial tibial translation, and internal tibial rotation in ACL reconstructed knees during a quasi-static lunge (Abebe et al., 2011a, 2011b). However, the effects of these kinematic asymmetries on cartilage loading patterns are less understood, though important to understand implications for PTOA after ACLR.

The objective of this study was to investigate the relationship between ACL graft tunnel location, graft orientation, and tibiofemoral kinematics and cartilage contact in ACLR subjects. To do this, we utilized a combination of static and dynamic magnetic resonance imaging to obtain *in vivo* measurements of ACL graft geometry and post-operative knee mechanics during loaded knee flexion-extension. We first investigated whether there was evidence of systematic bias in ACL graft geometry, tibiofemoral kinematics, or tibial cartilage contact in ACLR knees when compared to the contralateral knees. We then tested the hypothesis that a more vertically oriented ACL graft would be related to greater anterior tibial translation and internal tibial rotation and would also be related to abnormal cartilage contact location and sliding velocity on the tibial plateau.

2. Methods

2.1 Subjects

The bilateral knees of 18 subjects that underwent a primary unilateral, isolated ACLR were tested (9 male, 9 female, mean: 24.8 (SD: 5.7) yrs, 78.9 (SD: 16.5) kg, 20.2 (SD: 8.7) months post-op). The surgeon performing the procedure and the graft type used in the ACLR were not controlled (9 bone-patellar tendon-bone grafts, 9 hamstrings tendon grafts, 1 subject with small, stable medial and lateral meniscal tears, 2 subjects with small partial, lateral meniscectomies). Subjects provided informed consent according to a University of Wisconsin-Madison Institutional Review Board approved protocol. The subjects' reconstructed knees had no history of septic, inflammatory, or crystalline-induced arthritis, and no post-operative complications. The contralateral knees had no history of pain, injury, or surgery and no history of septic, inflammatory, or crystalline-induced arthritis. At the time of testing, all subjects had been released to return to full participation in sporting activities based on successful completion of post-operative physical therapy.

2.2 Static and Dynamic MR Imaging

The subjects underwent a bilateral static magnetic resonance (MR) imaging protocol consisting of a three-dimensional intermediate-weighted fast spin-echo sequence (3D FSE Cube) and a three-dimensional spoiled gradient recall-echo sequence with iterative decomposition with echo asymmetry and least squares estimation fat-water separation sequence (IDEAL-SPGR) (Fig. 1) (Kaiser et al., 2013). MR scans were performed in a 3.0 T clinical scanner (Discovery MR750, GE Healthcare, Waukesha, WI, USA) using an 8-channel phased array extremity coil (InVivo, Orlando, FL, USA) positioned around the knee. The 3D FSE Cube sequence was used to obtain images for characterizing ligament and cartilage morphology (in-plane sagittal resolution, 0.39×0.39 mm; slice thickness, 1.0 mm; matrix, $384 \times 384 \times 96$; repetition time, 2066.7 ms; echo time, 19.8 ms; flip angle, 90°). The

IDEAL-SPGR sequence was used to obtain images for characterizing bone morphology (in-plane axial resolution, 0.37×0.37 mm; slice thickness, 0.9 mm; matrix, $512 \times 512 \times 304$; repetition time, 10 ms; echo times, 4.5/5.5/6.1 ms; flip angle, 14°).

Following completion of static MR imaging, subjects underwent a bilateral dynamic MR imaging protocol with their lower leg secured to the lever arm of an MR-compatible knee-loading device (Westphal et al., 2013). Subjects performed cyclic knee flexion-extension at 0.5 Hz with the rate denoted by an audible metronome. During this task, the device induced active stretch-shortening quadriceps contractions by applying an inertial load to the subjects' lower leg via a set of rotating disks (Westphal et al., 2013). This quadriceps loading paradigm mimics the loading-response phase of walking (Besier et al., 2009; Whittington et al., 2008).

Subjects performed this knee flexion-extension motion for 5 minutes while a 3D SPGR sequence with vastly under-sampled isotropic projections (3D SPGR-VIPR) continuously acquired volumetric data (Fig. 1) (1.5-mm isotropic resolution; repetition time, 4 ms; echo time, 1.4 ms; flip angle, 8° ; receiver bandwidth, 32.5 kHz; unique radial lines, 93,922; field of view, 48 cm). During this motion, an MRI-compatible rotary encoder tracked the lever arm angle (Micronor Inc., Camarillo, CA, USA). This angle was used to determine the beginning of each flexion-extension cycle. The SPGR-VIPR projections were then sorted into 60 equally-spaced bins based on percent of the total cycle (Kaiser et al., 2013). The sorted projections were reconstructed into 60 volumetric image sets over the flexion-extension motion using conjugate gradient least squares minimization (Pruessmann et al., 2001). Each of these image frames consisted of an average of the MR data collected during 1.67%, or 33.3 ms, of the 2 sec flexion-extension cycle.

2.3 MR Image Processing

Distal femur and proximal tibia bone geometries were manually segmented from the IDEAL-SPGR images (Fig. 1) (MIMICS, Materialise Group, Leuven, Belgium). Femoral and tibial articular cartilage, native ACL, and ACL reconstruction graft geometries were manually segmented from the 3D FSE Cube images and registered to the bone geometries. Bone, cartilage, and ACL geometries were smoothed and converted into triangular meshes (7000 triangles/surface for bones and approximately $0.33 \text{ mm}^2/\text{triangle}$ for cartilage and ACL meshes); (Geomagic Studio, 3D Systems, Rock Hill, SC, USA and MeshLab, Visual Computing Lab-ISTI-CNR, Pisa, Italy). Anatomical reference frames were independently established for each femur and tibia using an algorithm that automatically determined the coordinate systems based on the geometric and inertial properties of the bone (Miranda et al., 2010).

The segmented bone and ACL meshes were used to characterize ACL geometry (Fig. 2A). The ACL femoral and tibial footprints were defined as the intersection of the ACL mesh with the femoral and tibial meshes, respectively. We then computed the location of the center of the femoral and tibial footprints relative to anatomical landmarks and expressed the location in the femoral and tibia; reference frames, respectively. Femoral footprint location was measured relative to the most anterior point of the trochlear groove and the most inferior point of the lateral femoral condyle. The tibial footprint location was measured relative to

the most anterior and most medial points of the tibial plateau. A cylinder was then fit to the mid-substance of the ACL and ACL graft and a plane was fit to the tibial plateau. The long axis of the cylinder was used to compute the ACL orientation in the frontal and sagittal planes relative to the best-fit plane. Non-anatomic ACL graft geometry metrics were computed as the side-to-side differences (reconstructed minus contralateral) in the femoral and tibial footprint locations, the ACL frontal plane orientation, and the ACL sagittal plane orientation.

Tibiofemoral kinematics were measured by tracking the femur and tibia in the 3D dynamic MR images (Fig. 1). To do this, the femoral and tibial bone meshes were manually placed in the first dynamic image frame. The bones' position and orientation were then determined by minimizing the sum of squared intensities at each vertex of the bone meshes (Kaiser et al., 2013; Powell, 1964). The solution for this frame was then used as the initial guess for the next frame and the optimization proceeded until the bones were tracked in all remaining dynamic MR image frames. This model-based tracking method provides kinematics with precisions of less than 0.8° and 0.5 mm (Kaiser et al., 2016a). Tibiofemoral kinematics were computed as the position and orientation of the tibia relative to the femur in each dynamic image frame (Grood and Suntay, 1983). Kinematics were low-pass filtered at a cutoff frequency of 5 Hz, which is 10 times greater than the flexion-extension cycle rate performed by subjects.

Cartilage contact was characterized based on the proximity between the femoral and tibial cartilage meshes at each frame of the cyclic motion. For each triangle in the tibial mesh, proximity was computed by projecting along the triangle normal, and then computing the distance between the triangle center and the point of intersection on the femoral mesh (Smith et al., 2016). A positive proximity was indicative of mesh overlap and, thus, cartilage contact (Fig. 1). The proximity of the tibial cartilage mesh was re-zeroed such that at least one triangle was in contact at each frame of the flexion-extension motion in both the medial and lateral compartments (Borotikar and Sheehan, 2013; Kaiser et al., 2016b). The center of contact locations on the medial and lateral tibial plateaus and the medial and lateral femoral condyles were computed as the weighted-average position of the contact region, with the position of each triangle weighted by its proximity. This method of measuring the center of contact location from the SPGR-VIPR images has precisions less than 0.25 mm and 0.49 mm in the anterior-posterior and medial-lateral directions, respectively (Kaiser et al., 2016a).

A first-order central finite difference approximation was used to compute the center of contact velocities for the medial and lateral compartments of the femur and tibia (Beveridge et al., 2013). The femoral center of contact velocities were then subtracted from the corresponding velocities of the tibia to determine the sliding velocity vectors. The sliding velocity vectors were then projected onto a tangent plane at the center of contact, and the magnitudes of the projected velocity vectors were computed. This metric has previously been used as a measure of cartilage contact sliding velocity (Anderst and Tashman, 2009; Beveridge et al., 2013).

Kinematic and cartilage contact metrics were computed for the reconstructed and contralateral knees. Kinematic metrics were defined as the tibiofemoral kinematics (i.e.

internal tibial rotation, adduction angle, and anterior, lateral, and superior tibial translations) at peak knee flexion and the range in tibiofemoral kinematics during the extension phase of the flexion-extension motion (Fig. 2B). Cartilage contact location metrics were defined as the anterior and lateral location of the center of contact on the medial and lateral tibial plateaus at peak knee flexion (Fig. 2C). Cartilage contact sliding velocity metrics were computed as the sliding velocity at peak knee flexion and the mean absolute sliding velocity during knee extension for the medial and lateral compartments. Asymmetric kinematics and cartilage contact were computed as the side-to-side differences (reconstructed minus contralateral) in these metrics.

2.4 Statistical Analysis

Paired t-tests were used to determine if there was a bias in the ACL graft footprint locations or orientations relative to the native ACLs ($\alpha = 0.05$). Paired t-tests were also used to test for differences in the kinematics and contact metrics between reconstructed and contralateral knees ($\alpha = 0.05$). A multiple linear regression analysis was used to determine the relationship between non-anatomic ACL graft geometry and asymmetric tibiofemoral kinematics and cartilage contact. In this linear regression analysis, the independent variables were the non-anatomic ACL graft geometry metrics and the dependent variables were the asymmetric tibiofemoral kinematics, cartilage contact location, and cartilage contact sliding velocity metrics. When a significant relationship was found ($\alpha = 0.05$), the coefficient of the linear regression model (β) was used to determine the sensitivity of kinematics and cartilage contact to non-anatomic graft geometry.

3. Results

The tibial footprints of the reconstructed knees were significantly more posterior than that of the contralateral knees (reconstructed mean: 21.6 (SD: 3.3) mm, contralateral mean: 19.3 (SD: 2.5) mm, $P=0.001$, Cohen's $d=0.92$) (Table 1). No other ACL geometry metrics were significantly different between the reconstructed and contralateral knees.

Reconstructed knees exhibited altered kinematics and cartilage contact relative to the contralateral knees. Specifically, reconstructed knees were more externally rotated (side-to-side difference: 2.9° (SD: 5.4°), $P=0.04$, Cohen's $d=0.53$) and medially translated (1.6 (SD: 2.5) mm, $P=0.01$, Cohen's $d=0.65$) at peak knee flexion. Additionally, reconstructed knees exhibited a greater range in knee adduction during extension (0.81° (SD: 1.4°), $P=0.03$, Cohen's $d=0.56$). The center of contact was more posterior in reconstructed knees on both plateaus at peak knee flexion (side-to-side difference medial: 2.0 (SD: 3.4) mm, $P=0.02$, Cohen's $d=0.60$; side-to-side difference lateral: 1.2 (SD: 2.2) mm, $P=0.04$, Cohen's $d=0.53$).

Non-anatomic ACL graft geometry was associated with asymmetric tibiofemoral kinematics, with the sagittal plane orientation of the ACL being the best predictor of asymmetric motion (Table 2). Specifically, a more vertical graft in the sagittal plane (i.e. a greater sagittal plane angle) was significantly linked with greater anterior tibial translation ($\beta=0.11 \text{ mm/deg}$, $P=0.049$, $R^2=0.22$), internal tibial rotation ($\beta=0.27 \text{ deg/deg}$, $P=0.042$, $R^2=0.23$), and adduction angle ($\beta=0.15 \text{ deg/deg}$, $P=0.013$, $R^2=0.44$) at peak knee flexion (Fig.

3). Additionally, a more vertical graft in the frontal plane (i.e. a greater frontal plane angle) was related to a reduced range in lateral tibial translation during extension ($\beta = -0.089 \text{ mm/deg}$, $P=0.046$, $R^2=0.23$). The tibial footprint location was also related to asymmetric kinematics, with a more lateral tibial footprint associated with a greater range in anterior translation ($\beta = 0.49 \text{ mm/mm}$, $P=0.0089$, $R^2=0.36$) and a more posterior tibial footprint related to a less adducted knee at peak knee flexion ($\beta = -0.44 \text{ deg/mm}$, $P=0.032$, $R^2=0.44$). A more superior femoral footprint was linked to greater lateral tibial translation ($\beta = 0.26 \text{ mm/mm}$, $P=0.021$, $R^2=0.29$).

Non-anatomic ACL graft geometry was also associated with asymmetric tibiofemoral cartilage contact (Table 3). Similar to kinematics, the graft sagittal plane orientation was most often significantly related to asymmetries in the center of contact location and cartilage contact sliding velocity. A greater sagittal plane angle was associated with a more medial center of contact on both the medial ($\beta = -0.32 \text{ mm/deg}$, $P=0.003$, $R^2=0.43$) and lateral ($\beta = -0.27 \text{ mm/deg}$, $P=0.002$, $R^2=0.45$) tibial plateaus and a more posterior center of contact on the lateral plateau at peak knee flexion ($\beta = -0.11 \text{ mm/deg}$, $P=0.030$, $R^2=0.45$) (Fig. 4A). During extension, a greater sagittal plane angle was related to lower mean absolute contact sliding velocity in the medial compartment ($\beta = -4.2 \text{ mm/s/deg}$, $P=0.002$, $R^2=0.58$), but greater mean absolute sliding velocity in the lateral compartment ($\beta = 5.7 \text{ mm/s/deg}$, $P=0.006$, $R^2=0.54$) (Fig. 4B).

The tibial and femoral footprint locations were also frequently associated with asymmetric cartilage contact (Table 3). A more lateral tibial footprint was related to a more posterior center of contact on the lateral plateau ($\beta = -0.51 \text{ mm/mm}$, $P=0.018$, $R^2=0.45$). Additionally, a more posterior tibial footprint was related to lower mean absolute contact sliding velocity during extension ($\beta = -12 \text{ mm/s/mm}$, $P=0.008$, $R^2=0.58$) and at peak knee flexion ($\beta = -37 \text{ mm/s/mm}$, $P=0.020$, $R^2=0.29$) in the medial compartment, but greater sliding velocity in the lateral compartment ($\beta = 20 \text{ mm/s/mm}$, $P=0.006$, $R^2=0.54$). A more posterior femoral footprint was associated with reduced sliding velocity in the lateral compartment at peak knee flexion ($\beta = -33 \text{ mm/s/mm}$, $P=0.049$, $R^2=0.22$).

4. Discussion

Current ACLR surgical techniques do not substantially mitigate the risk of early cartilage degeneration, with greater than 50% of patients exhibiting radiographic evidence of PTOA at 10 to 15 years post-surgery (Lohmander et al., 2007). Based on the link between altered loading and cartilage damage (Griffin and Guilak, 2005), it is theorized that abnormal knee mechanics that remain after ACLR play a role in the pathogenesis of PTOA (Chaudhari et al., 2008). Thus, identifying surgical factors to better restore normative cartilage loading may reduce the risk of PTOA in these patients. In this study, we found that there was a posterior bias in positioning of the tibial tunnel, relative to the native ACL footprint. We also observed abnormal tibiofemoral kinematics and an altered center of contact location in ACL

reconstructed knees, relative to the contralateral knee. Further, we found that non-anatomic graft geometry was linked to side-to-side differences in tibiofemoral kinematics, cartilage contact location, and cartilage contact sliding velocity.

Our kinematic observations are generally consistent with previously observed links between ACL graft geometry and kinematics. Prior studies found that a more vertical graft was associated with increased anterior and rotational laxities (Brophy and Pearle, 2009; Loh et al., 2003) and greater anterior tibial translation, medial tibial translation, and internal tibial rotation during a quasi-static lunge (Abebe et al., 2011b). Given that the ACL provides the primary restraint to anterior tibial translation and a secondary restraint to internal tibial rotation (Andersen and Dyhre-Poulsen, 1997; Fukubayashi et al., 1982; Gabriel et al., 2004; Sakane et al., 1997), these results suggest that the functionality of an ACL graft is potentially related to its sagittal plane orientation. Additionally, the similarities between our findings and previous studies suggest that graft geometry can affect knee behavior across a range of open- and closed-chain tasks.

We have shown that side-to-side differences in kinematics give rise to altered cartilage contact patterns, which are also linked to ACL graft geometry. Specifically, the graft sagittal plane orientation and the posterior location of the tibial footprint were linked to asymmetries in cartilage contact location and sliding velocity (Table 3). The observed trends suggest a subject with a graft sagittal plane angle that is greater than that of the contralateral ACL is likely to exhibit more anterior tibial translation, internal tibial rotation, and adduction in their reconstructed knee (Fig. 5). On the medial plateau, this same subject would exhibit a more medial center of contact and a decreased contact sliding velocity. On the lateral plateau, this subject would exhibit a more medial and posterior center of contact and an increased contact sliding velocity. These effects are important because a shift in contact location and contact sliding velocity can load cartilage in a manner that the composition and microstructure may not be well adapted for, potentially initiating a degenerative pathway that leads to PTOA (Beveridge et al., 2013; Chaudhari et al., 2008).

Femoral and tibial tunnel placement are primary determinants of ACL graft orientation. However, surgical placement of ACL graft tunnels remains technically challenging. Using a transtibial drilling technique, experienced surgeons placed the femoral tunnel an average of 8 to 9 mm from the native ACL footprint center (Abebe et al., 2009; Kaseta et al., 2008; Kopf et al., 2010). These tunnels were typically placed anterior and superior to the femoral footprint center, resulting in a more vertical graft in the sagittal plane. Femoral tunnel placement accuracy improved to 2 to 3 mm when drilling the femoral and tibial tunnels independently. While this improvement in accuracy would result in knee mechanics significantly closer to that of the contralateral knee, our results suggest that small, residual abnormalities in kinematics and contact location and a relatively large difference in contact sliding velocity would still remain in the reconstructed knee (Tables 2 and 3). We also observed a posterior bias in the tibial footprint location (Table 1). This bias towards a posterior tibial tunnel position may be intentional to prevent graft impingement against the superior notch during knee extension (Amis and Jakob, 1998; Bedi et al., 2011). While further work is needed to assess the threshold for an acceptable difference in tunnel location relative to the native ACL footprints, the accuracy of current techniques may limit surgeons'

ability to adequately restore normative knee mechanics. Advancements in the treatment of ACL injury through computer-assisted reconstruction (Dessenne et al., 1995; Jalliard et al., 1998; Kaseta et al., 2008; Picard et al., 2001) and a more thorough understanding of the link between slight alterations in tunnel position and knee kinematics may allow surgeons to accurately recreate the native ACL geometry needed to restore normative knee mechanics and improve long-term outcomes of ACLR.

There are three primary limitations to consider in this study. First, knee mechanics were studied during an open-chain task, in contrast to the closed-chain loading that occurs during locomotion. However, in our previous work, we found that small, subtle shifts in tibiofemoral kinematics detected during this open-chain task are similar to those observed during downhill running (Kaiser et al., 2017). Additionally, strain in the ACL is similar between open-chain and closed-chain knee flexion (Beynnon et al., 1997), which suggests the experimentally measured relationships in this study may extend to locomotion. Using a simple task may also reduce the variability in neuromuscular coordination between ACLR subjects that exists during functional motions (Ciccotti et al., 1994; Gokeler et al., 2010) and can influence knee mechanics (Chmielewski et al., 2005; Zeller et al., 2003). The open-chain motion used in this study may then more readily isolate the effect of graft geometry on post-operative knee mechanics. Second, the kinematics and cartilage contact data throughout the flexion-extension motion were reduced to metrics at peak knee flexion and during knee extension. Peak knee flexion corresponds with peak quadriceps loading during this cyclic motion (Westphal et al., 2013). The quadriceps loading during the extension phase of this motion mimics the loading-response phase of walking, in which the quadriceps brake knee flexion and then induce knee extension (Besier et al., 2009; Whittington et al., 2008). Thus, these portions of the flexion-extension motion were selected based on their potential functional relevance. Third, we did not control graft type or initial graft tension in these subjects. Previous work has shown that graft type is related to differences in knee kinetics during walking (Webster et al., 2005) and initial graft tension is related to differences in passive knee mechanics under simple loading conditions (Boylan et al., 2003; Brady et al., 2007; Nicholas et al., 2004). Further work is needed to assess the effect of these surgical parameters on cartilage loading during active knee motion and to assess their interaction with graft geometry when attempting to restore normative knee mechanics with ACLR.

In conclusion, the findings of this study provide evidence that non-anatomic ACL graft geometry is related to asymmetric tibiofemoral kinematics and cartilage contact. Given that abnormal cartilage loading may precipitate cartilage degeneration, these findings suggest that replicating native ACL geometry may be critical for normalizing mechanics and mitigating the risk of PTOA following ACLR. Future longitudinal studies are needed to determine whether the changes in cartilage contact and contact sliding velocity due to non-anatomic graft geometry are directly related to early cartilage degeneration within abnormally loaded regions of the knee.

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Highlights

- Used MRI to determine link between graft geometry and post-operative knee mechanics
- Observed a posterior bias in the graft tibial tunnel location relative to native
- Non-anatomic graft geometry was linked to abnormal kinematics and cartilage contact
- Graft sagittal plane orientation was the best predictor of abnormal knee mechanics
- Provides support for restoring native anterior cruciate ligament geometry

Fig. 1. Workflow for MR Image Analysis

(Top row) Subject-specific bone, cartilage, and ACL geometries were created by segmenting IDEAL-SPGR and 3D FSE Cube MR images of both knees of each subject. (Bottom row) Subjects performed an active flexion-extension motion against an inertial load while volumetric, dynamic MR images were collected with an SPGR-VIPR sequence. The bones were tracked in the dynamic images using the bone geometries. Tibiofemoral kinematics were then computed using the position and orientation of the tibia relative to the femur (Grood and Suntay, 1983). Cartilage contact was computed by prescribing the tibiofemoral kinematics to the femoral and tibial cartilage geometries and measuring the overlap of the cartilage surfaces.

Fig. 2. ACL Geometry, Kinematics, and Cartilage Contact Metrics

(A) The orientation of the ACL relative to the tibial plateau in the sagittal and frontal planes, the location of the tibial footprint in the axial plane, and the location of the femoral footprint in the sagittal plane were computed for both knees of each subject. (B) Representative plot showing internal tibial rotation throughout the flexion-extension motion for both knees of one subject. The extension phase of the motion is denoted with an arrow. Similar plots were created for the other five degrees of freedom of the tibiofemoral joint. Kinematics metrics were then computed as the kinematics at peak knee flexion and the range in kinematics during knee extension. (C) Top row shows the proximity of the tibial cartilage to the femoral cartilage at peak knee flexion for the contralateral knee of a representative subject. Red is indicative of cartilage contact. Similar maps were generated for both knees of each subject and used to compute the center of contact location on the tibial plateaus. Bottom row shows the cartilage surface sliding velocity on the medial and lateral tibial plateaus during knee extension for the contralateral knee of a representative subject. The mean absolute sliding velocity during extension and the sliding velocity at peak knee flexion were computed for both knees of each subject.

Fig. 3. Relationship between Kinematics and ACL Sagittal Plane Angle

Scatter plots show the relationship between asymmetric kinematics and non-anatomic graft sagittal plane angles for anterior tibial translation, internal tibial rotation, and adduction angle at peak knee flexion. The coefficient of the linear regression model (β) is shown for each relationship.

Fig. 4. Relationship between Cartilage Contact and ACL Sagittal Plane Angle

Scatter plots show the relationships between asymmetric cartilage contact and non-anatomic graft sagittal plane angles for (A) the center of contact location and (B) the mean absolute contact sliding velocity for the medial and lateral tibial plateaus. The coefficient of the linear regression model (β) is shown for each relationship.

Fig. 5. Representative subject with a vertical ACL graft

Graphic shows the anterior tibial translation, internal tibial rotation, and center of contact location at peak knee flexion and the side-to-side difference in contact sliding velocity during knee extension for a subject with a more vertical ACL graft in the sagittal plane, relative to the native ACL. The asymmetries in kinematics and cartilage contact measured in this subject are representative of the relationship between asymmetric knee mechanics and non-anatomic ACL graft sagittal plane orientation observed across all subjects.

Table 1

Side-to-side comparisons of ACL graft geometry

The location of the femoral footprint in the sagittal plane, location of the tibial footprint in the axial plane, and orientation of the ACL in the sagittal and frontal planes for the reconstructed and contralateral knees. *P*-values are the result of paired t-tests. Effect size computed as Cohen's *d* coefficient.

		Reconstructed mean (SD)	Contralateral mean (SD)	<i>P</i> -value	Effect Size
Femoral Footprint Location (mm)	Superior	21.1 (5.1)	21.6 (2.4)	0.75	-0.075
	Posterior	49.1 (4.4)	49.8 (3.8)	0.37	-0.22
Tibial Footprint Location (mm)	Lateral	40.3 (2.6)	40.5 (2.9)	0.77	-0.070
	Posterior	21.6 (3.3)	19.3 (2.5)	0.001	0.92
ACL Orientation (deg)	Sagittal Plane	51.2 (6.0)	53.8 (7.4)	0.22	-0.30
	Frontal Plane	73.5 (5.7)	72.8 (5.7)	0.65	0.11

Table 2

Relationships between Kinematics and ACL Graft Geometry.

		ACL Orientation (deg)		Tibial Footprint Location (mm)		Femoral Footprint Location (mm)	
		Sagittal Plane	Frontal Plane	Lateral	Posterior	Superior	Posterior
Anterior Translation (mm)	At Peak Flexion Range	0.11 mm/deg^*	-	-	-	-	-
		-	-	0.49 mm/mm^{**}	-	-	-
Lateral Translation (mm)	At Peak Flexion Range	-	-	-	-	0.26 mm/mm^*	-
		-	-0.087 mm/deg^*	-	-	-	-
Internal Rotation (deg)	At Peak Flexion	0.27 deg/deg^*	-	-	-	-	-
Adduction Angle (deg)	At Peak Flexion	0.15 deg/deg^*	-	-	-0.44 deg/mm^*	-	-

Coefficients of the linear regression model (β) for those relationships between asymmetric kinematics and non-anatomic ACL graft geometry that were statistically significant (* $P < 0.05$, ** $P < 0.01$).

Table 3

Relationships between Cartilage Contact and ACL Graft Geometry.

	ACL Orientation (deg)		Tibial Footprint Location (mm)		Femoral Footprint Location (mm)	
	Sagittal Plane	Frontal Plane	Lateral	Posterior	Superior	Posterior
Center of Contact Location (mm)	Anterior	–	–	–	–	–
	Lateral	$-0.32 \text{ mm/deg}^{**}$	–	–	–	–
	Anterior	-0.11 mm/deg^*	-0.51 mm/mm^*	–	–	–
	Lateral	$-0.27 \text{ mm/deg}^{**}$	–	–	–	–
Sliding Velocity (mm/s)	Mean Absolute	$-4.2 \text{ mm/s/deg}^{**}$	–	$-11.9 \text{ mm/s/mm}^{**}$	–	–
	At Peak Flexion	–	–	-36.9 mm/s/mm^*	–	–
	Mean Absolute	$5.7 \text{ mm/s/deg}^{**}$	–	$20.2 \text{ mm/s/mm}^{**}$	–	–
	At Peak Flexion	–	–	–	–	-33.0 mm/s/mm^*

Coefficients of the linear regression model (β) for those relationships between asymmetric cartilage contact and non-anatomic ACL graft geometry that were statistically significant (* $P < 0.05$, ** $P < 0.01$).