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Mapping isometry and length changes in ligament reconstructions of the knee

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Chapter 3

The effect of ACL deficiency on ACL end-to-end distance during in-vivo dynamic activity

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ABSTRACT

Purpose: To evaluate the effect of ACL deficiency on the in vivo changes in end-to-end distances and to determine appropriate graft fixation angles for commonly used tunnel positions in contemporary ACL reconstruction techniques.

Methods: Twenty-one patients with unilateral ACL-deficient and intact contralateral knees were included. Each knee was studied using a combined magnetic resonance and dual fluoroscopic imaging technique while the patients performed a dynamic step-up motion (~50° of flexion to extension). The end-to-end distances of the centers of the anatomic anteromedial (AM), posterolateral (PL) and single-bundle ACL reconstruction (SB-anatomic) tunnel positions were simulated and analyzed. Comparisons were made between the elongation patterns between the intact and ACL-deficient knees. Additionally, a maximum graft length change of 6% was used to calculate the deepest flexion fixation angle.

Results: ACL-deficient knees had significantly longer graft lengths when compared with the intact knees for all studied tunnel positions ($P = 0.01$). The end-to-end distances for the AM, PL and SB-anatomic grafts were significantly longer between 0-30° of flexion when compared with the intact knee by $P = 0.05$ for all. Six percent length change occurred with fixation of the AM bundle at 30° of flexion, PL bundle at 10° and the SB-anatomic graft at 20°.

Conclusions: ACL-deficient knees had significantly longer in vivo end-to-end distances between 0°-30° of flexion for grafts at the AM, PL and SB-anatomic tunnel positions when compared with the intact knees. Graft fixation angles of <30° for the AM, <10° for the PL, and <20° for the SB-anatomic grafts may prevent permanent graft stretch.

INTRODUCTION

Anterior cruciate ligament (ACL) reconstruction is technically demanding. Tibiofemoral tunnel positioning is a critical determinant to achieve successful ACL reconstruction. If the distance between the tunnels increases substantially during flexion or extension of the knee, the graft tightens and either the motion of the knee is restricted or the graft stretches ultimately causing graft failure. Alternatively, if the tunnels' distance substantially decreases, the graft slackens and is not supportive. Furthermore, tunnel positioning determines the graft length change pattern, which is a crucial variable to decide upon an appropriate knee fixation angle for graft fixation.

Previous cadaveric [3, 19, 22, 26, 36] and in vivo studies [23, 29] have assessed the length changes of the ACL. Yoo et al.[39] examined the in vivo end-to-end distances of the ACL during a non-weight-bearing, static, range-of-motion in intact knees, while Jang et al.[19] recently examined the differences between intact and ACL-deficient knees in a cadaveric setting. In our recent work,[20] in vivo ACL isometry was mapped and the strains of the anatomic and classical transtibial tunnel position were examined in intact knees. However, no prior study has assessed the differences in end-to-end distances of the ACL between intact and ACL-deficient knees during dynamic in vivo weight-bearing (i.e., physiological) activity. Improved understanding of graft length changes is important for surgeons and could help to determine the knee flexion angle for fixation and tensioning which may reduce graft failure rates. In addition, differences in end-to-end distances between the intact and ACL-deficient knee during functional activity could have critical importance in the development of proper ACL rehabilitation programs.[7, 9]

The purpose of this study was to evaluate the effect of ACL deficiency on the in vivo changes in end-to-end distances and to determine appropriate graft fixation angles of grafts at commonly used tunnel positions in contemporary ACL reconstruction techniques: the anatomic anteromedial (AM), posterolateral (PL) and single-bundle ACL reconstruction (SB-anatomic) during dynamic, physiological weight-bearing motion. We hypothesized that the end-to-end distances of the AM, PL and SB-anatomic tunnel positions would be longer in the ACL-deficient knees when compared with the intact knees, and that the differences in end-to-end distances between the intact and ACL-deficient knees would be most pronounced at lower flexion angles, i.e. where the ACL is most active in restraining anterior tibial translation and internal tibial rotation.

METHODS

Patient selection

This study was approved by our Institutional Review Board. Written consent was obtained from all patients prior to participation in this study. This study included 21 patients (13 men, eight women; age range 18–59 years; length 160–193 cm; active on a moderate athletic level before injury) with a diagnosed unilateral ACL tear. The ACL tear was confirmed by clinical examination and magnetic resonance imaging (MRI) performed by a specialized orthopedic sports surgeon and specialized musculoskeletal radiologist respectively. Patients with injury to other ligaments, noticeable cartilage lesions, and injury to the underlying bone were excluded from the study. Five patients had no significant damage to the menisci, eight had a medial meniscal tear and eight had a lateral meniscal tear which required partial meniscectomy (<30% removal) during surgery. There was no evidence or history of injury, surgery or disease in the contralateral knees. These patients were included in our previous study on meniscus injuries and knee kinematics.[18]

Imaging procedure

The MRI and dual fluoroscopic imaging techniques for the measurement of ligament kinematics have been described in detail previously.[24] MRI scans of the knee joints were done in the sagittal plane using a three-Tesla MRI scanner (MAGNETOM Trio, Siemens, Malvern, PA) with a double-echo water-excitation sequence (thickness of one millimeter; resolution of 512 × 512 pixels).[11] The images were then imported into solid modeling software (Rhinoceros; Robert McNeel and Associates, Seattle, WA, USA) to construct three-dimensional (3D) surface models of the tibia, fibula and femur.

The knee of each subject was simultaneously imaged using two fluoroscopes (BV Pulsera, Philips, the Netherlands). The fluoroscopes took 30 evenly distributed snapshot images per second as the patient performed the step-up motion. Next, the fluoroscopic images were imported into solid modeling software and placed in the imaging planes based on the projection geometry of the fluoroscopes during imaging of the patient. Finally, the MRI-based knee model of each subject was imported into the software, viewed from the directions corresponding to the fluoroscopic X-ray source used to acquire the images, and independently manipulated in six-degrees-of-freedom inside the software until the projections of the model matched with the outlines of the fluoroscopic images. When the projections best matched the outlines of the images taken during in vivo knee motion, the positions of the models were considered to be reproductions of the in vivo 3D positions of the knees. This system has an error of <0.1mm and 0.3° in measuring tibiofemoral joint translations and rotations, respectively.[11, 24, 25] The matching procedure was then repeated, providing the in vivo knee kinematics of the step-up motion.

Tibial and femoral attachment points

To determine the in vivo changes in end-to-end distances of the grafts during motion, various tibial and femoral attachment sites were used. The tibial attachment areas of the ACL were determined by the MR images in both sagittal and coronal planes.37 The anatomic ACL attachment area was directly mapped onto the 3D MRI-based tibia model. The attachment area was then subdivided into an AM and PL portions guided by the meticulously performed anatomic descriptions of Edwards et al.[12] and Ferretti et al.[13] The geometrical centers of the native ACL, AM and PL attachment areas were determined and used as three distinct tibial attachment points (Fig. 1).

A true medial view of the femur was established (perpendicular to the medial–lateral femoral axis). To account for the geometric variations between knees, a quadrant method (4 × 4 grid) developed by Bernard et al.[6] was applied to the 3D models. As described previously by Forsythe et al.[14], no Blumensaat line is present on the 3D models; therefore, the most anterior edge of the femoral notch roof was chosen as the reference for the grid alignment (line h). The femoral tunnel locations were based upon the review article by Parkar et al.[33], summarizing the available literature using Bernard's quadrant method to describe the femoral AM (21 × 25, i.e. h × t), PL (49 × 33), and SB-anatomic location (35 × 29) (Fig. 2). The deficient knees were mirrored with respect to the sagittal plane to match the intact knee. Then, the mirrored 3D models of the deficient knee were aligned to find the best-fit position with respect to the intact knee using a surface-to-surface registration method.[11]

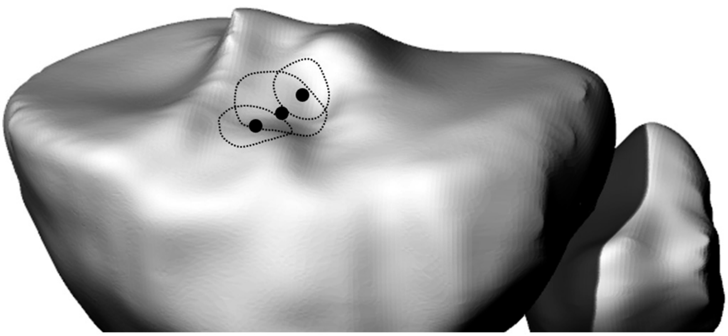

Fig. 1 Three-dimensional tibia model showing the distribution of the anteromedial, central and posterolateral tibial attachment points.

Length change measurements

The changes in end-to-end distances for each theoretical graft were measured as a function of knee flexion. To simulate the path of a true, massive ligament, the direct line connecting the femoral and tibial attachment point was projected on the bony surfaces to create a curved line avoiding penetration of the connecting line through bone, i.e. a wrapping path (Fig. 3). An optimization procedure was implemented to determine the projection angle to find the shortest 3D wrapping path at each flexion angle of the knee. This technique has been described in previous studies for measurements of ligament kinematics.[35] The length of the projected line (i.e. curved around the bony surfaces) was measured as the length of the graft.

Graft peak strains greater than six percent [1, 8] have been shown to cause permanent graft stretch/damage. Therefore, the greatest observed end-to-end distance of the AM, SB-anatomic and PL tunnel positions was used to calculate the maximum graft length resulting in the threshold of six percent length change: greatest length bundle / 1.06 = *maximum graft length*. The flexion angles corresponding to the maximum graft length without exceeding the six percent threshold were then suggested as the critical margin for flexion fixation angles.

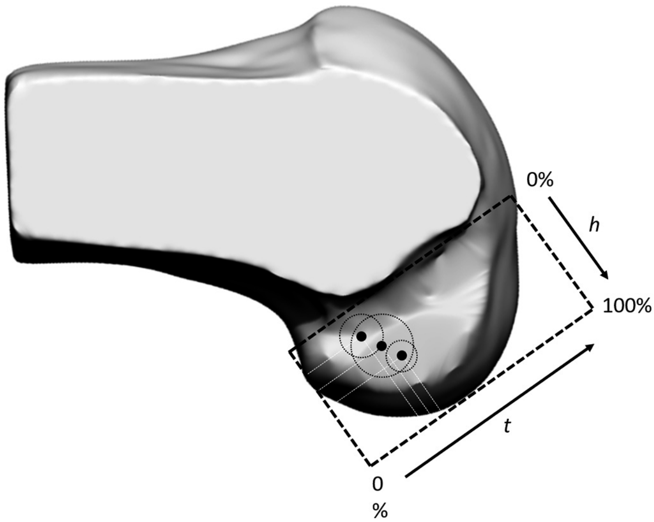

Fig. 2 Medial view of a 3D femur model in 90° of flexion. Bernard et al's [6] quadrant method was applied to the medial aspect of the lateral femoral condyle. A line extending along the Blumensaat line was used as a landmark for the anterior border of the grid (line t). Parallel to line t, a line was drawn to the posterior edge of the lateral condyle to form the posterior border. The proximal and distal borders were formed by two lines perpendicular to the Blumensaat line (line h) originating from the proximal and distal bony borders of the lateral femoral condyle. The locations of the studied grafts were based upon the review article of Parkar et al.,[33] anteromedial (21 × 25, i.e. h × t), posterolateral graft (49 × 33), and single-bundle anatomic graft (35 × 29).

Table 1. Percentage length change of the AM, SB-anatomic and PL bundle during the dynamic step-up motion

Knee Flexion Angle	**AM length change, %**	**Accumulated length change, %**	**SB-anatomic length change, %**	**Accumulated length change, %**	**PL length change, %**	**Accumulated length change, %**
0 – 5°	-1.2	-1.2 (-0.6 to -1.7)	-1.7	-1.7 (-1.4 to -2.0)	-2.4	-2.4 (-1.8 to -3.1)
5 – 10°	-1.0	-2.2 (-1.3 to -3.1)	-1.6	-3.3 (-2.8 to -3.7)	-2.2	-4.6 (-3.4 to -5.8)
10 – 15°	-0.5	-2.7 (-1.5 to -3.9)	-1.3	-4.6 (-4.0 to -5.2)	-2.2	-6.9 (-5.3 to -8.4)
15 – 20°	-0.5	-3.2 (-1.8 to -4.6)	-1.2	-5.8 (-5.1 to -6.5)	-2.0	-8.8 (-7.1 to -10.6)
20 – 25°	-0.7	-3.9 (-2.3 to -5.5)	-1.5	-7.2 (-6.4 to -8.1)	-2.3	-11.1 (-9.0 to -13.2)
25 – 30°	-0.9	-4.8 (-2.9 to -6.6)	-1.5	-8.7 (-7.8 to -9.7)	-2.2	-13.4(-10.9 to -15.8)
30 – 35°	-0.5	-5.3 (-3.7 to -6.8)	-1.2	-9.9 (-9.1 to -10.8)	-1.7	-15.1 (-13.1 to -17.1)
35 – 40°	-1.0	-6.3 (-4.6 to -7.9)	-1.6	-11.5 (-10.7 to -12.3)	-2.4	-17.4 (-15.5 to -19.4)
40 – 45°	-1.7	-8.0 (-6.0 to -10.0)	-2.2	-13.7 (-12.8 to -14.7)	-3.2	-20.6 (-18.3 to -22.9)
45 – 50°	-1.7	-9.7 (-7.6 to -11.7)	-2.3	-16.1 (-15.1 to -17.0)	-3.1	-23.7 (-21.3 to -26.0)

NOTE. Values are expressed as a percentage of the length as normalized to the ligament length at 0° of knee flexion. Values are presented as mean ± 95% confidence interval; negative values indicate shortening between the distance of the attachment points. AM, anteromedial bundle; SB-anatomic, single-bundle anatomic reconstruction bundle; PL, posterolateral bundle.

Statistical analyses

A two-way analysis of variance (ANOVA) was first used to examine the effect of flexion angle and ACL intact/deficiency on length changes for each individual bundle (i.e., AM, central, PL). Paired Student's t-tests were then used to compare the healthy and deficient knees at corresponding flexion angles (e.g., AM healthy at 0° vs. AM deficient at 0°). Finally, a one-way ANOVA test was used to examine differences between the three healthy bundles. If significant, Tukey's Honest Significant Difference tests were employed to compare the various pairs of three bundles (AM vs. SB-anatomic, SB-anatomic vs. PL, AM vs. PL). The same procedure was then completed for the deficient bundles. Stats were performed in R version 3.3.2 and P values less than 0.05 were considered significant.

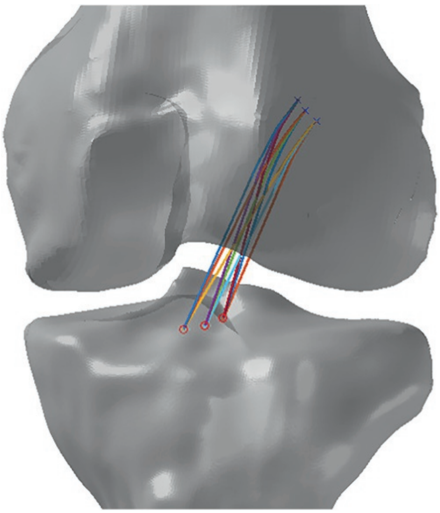

Fig. 3 Anterior–posterior view of a 3D knee model illustrating the lines curving over the bony geometry of the femur and tibia, that is, the "wrapping effect." At each flexion angle, an optimization procedure was implemented to determine the graft projection angle to find the shortest 3D wrapping path, mimicking the path of least resistance for the ACL graft.

RESULTS

The mean maximum flexion angles during the dynamic step-up motion for the intact and ACL-deficient knees were 55 ± 5° and 52 ± 5° respectively (mean ± standard deviation). The AM, PL and SB-anatomic grafts were longest in length at 0° of flexion for both the intact and ACL-deficient knee. ACL-deficient knees had significantly longer end-to-end distances for the AM (P = 0.01), PL (P = 0.01) and SB-anatomic grafts (P = 0.01) when compared with the intact knees. When comparing the intact and ACL-deficient knees at each flexion angle, longer end-to-end distances in the ACL-deficient knee were found for the AM, PL and SB-anatomic grafts at 0°, five degrees, 10°, 15°, 20°, 25° and 30° of flexion (P = 0.05 for all) (Fig. 4).

In the intact knee, all three grafts showed a significant decrease in length with increasing flexion from 42.2 ± 4.1 mm at 0° to 38.1 ± 3.5 mm at 50° for the AM graft (P = 0.001); 33.2 ± 3.4 mm at 0° to 25.3 ± 2.7mm at 50° for the PL graft (P = 0.001); and 37.5 ± 3.9 mm at 0° to 31.4 ± 3.2mm at 50° for the SB-anatomic graft (P = 0.001) (Fig. 4, Table 1). These accounted to decreases of approximately 10%, 24%, and 16% over the 50° of flexion respectively. The mean maximum lengths for AM, PL and SB-anatomic grafts were found at 0° of flexion; therefore, a mean of 2.4 mm, 1.9 mm and 2.1 mm, respectively, represents the theoretical maximum allowed length increase of six percent. The maximum allowed length changes corresponded to flexion angles of approximately 30°, 20° and 10° for the AL, SB-anatomic and PL grafts respectively.

In both the intact and ACL-deficient knees, significantly longer end-to-end distances were found for the AM graft than for SB-anatomic (P = 0.01) and the PL (P = 0.001) grafts, and for the SB-anatomic graft compared with the PL graft at all flexion angles (P = 0.01).

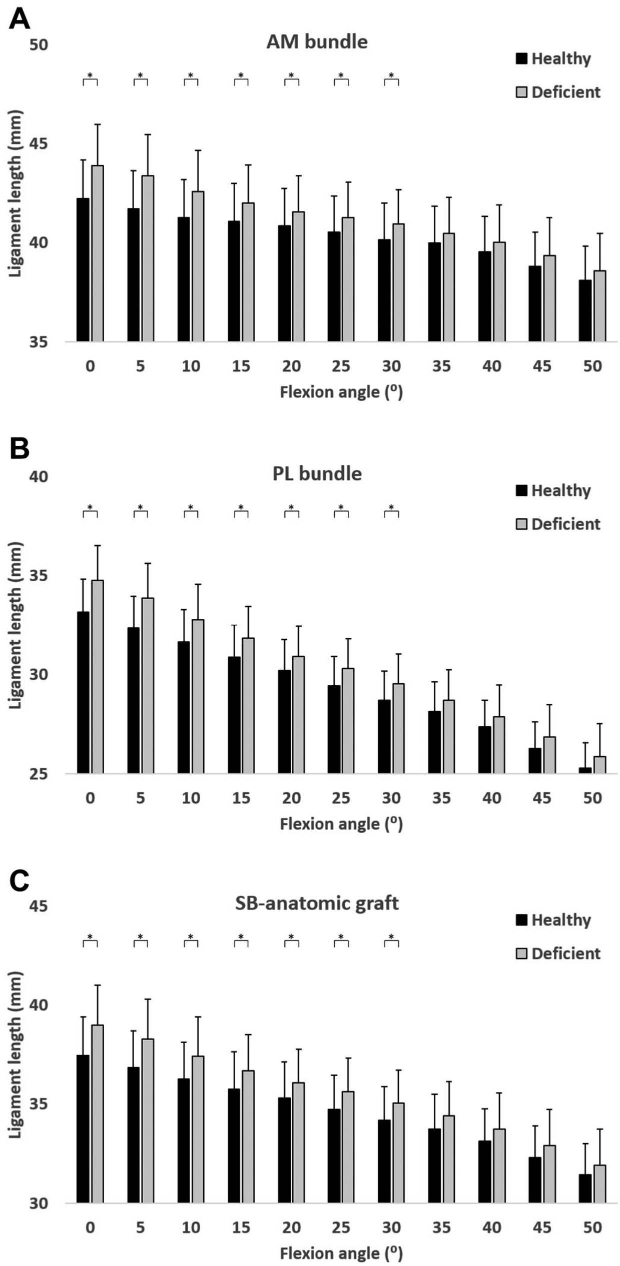

Fig. 4 Absolute length changes for the anteromedial bundle, posterolateral bundle and single-bundle anatomic graft for the ACL-deficient (gray) and intact knees (black) during the dynamic step-up motion. Mean values are shown, with the shaded area indicating the 95% confidence interval (CI).

DISCUSSION

The most important finding of this study was that ACL-deficient knees had significantly longer end-to-end distances when compared with the intact knees of all three tunnel positions during the dynamic step-up motion. The graft lengths in the ACL-deficient knees were significantly longer at lower flexion angles (<30°), corresponding to the area in which the ACL is most active in restraining anterior tibial translation and internal tibial rotation.[31] For both the intact and ACL-deficient contralateral knees, the three grafts had their longest length at 0° of flexion and consistently decreased with increasing flexion angles.

This study expands on recent cadaveric work, providing in vivo length change data of intact and ACL-deficient knees during functional activity. Specifically, Jang et al.[19] examined 10 cadaveric knees with and without axial load (1000 N) in ACL-intact and -deficient knee state between 0 and 60° of flexion. They found no changes in end-to-end distances of the ACL in the intact knees during flexion with and without axial loading, while the ACL-deficient knees yielded significantly longer end-to-end distances with increasing flexion angles only during axial loading. Based on these findings, the authors concluded that the end-to-end distances of the ACL-deficient knees increase with increasing flexion angles due to excessive femoral rollback.[19] Similar to Jang et al.,[3] in the current study, ACL deficiency yielded significantly longer end-to-end distances. In contrast to the in vitro results, however, our data demonstrated that the end-to-end differences between intact and deficient knees were relatively constant and did not increase with increasing flexion angles. In fact, the differences were most pronounced at lower flexion angles. The increased end-to-end distances observed in the ACL-deficient knees when compared with the ACL-intact knees are the result of the increased anterior tibial translation and internal tibial rotation caused by the ACL deficiency, the lower flexion angles correspond to the area where the ACL is most active in restraining anterior tibial translation and internal tibial rotation.[31] Next, these results may highlight the significant role of muscle action in restraining knee motion, as the lack of muscle action was described as a major limitation of the cadaveric study.[19] Next, we also found significant length change of the ACL during knee flexion in the intact knees, consistent with previous in vivo works.[23, 39] Given the prominent role of muscle action in knee restraint and other kinematic differences caused by dynamic in vivo movement, care may need to be taken when extrapolating cadaveric results of end-to-end distances to the in vivo physiologic setting.

The importance of understanding the mechanical properties of ACL-deficient knees is highlighted by the fact that less than one out of every four patients who sustain an ACL rupture undergoes ACL reconstruction within three years.[10] Improved understanding may be important when designing physical therapy and rehabilitation protocols. Our results demonstrated significant differences in mechanical characteristics between the intact and ACL-deficient knees during in vivo functional activity. Moreover, our study highlights the possible role muscle functioning may play in knee restraint, as compared with results of in vitro ACL-deficient knees.[19] Rehabiliation programs may need to differ based on whether or not the ACL has been repaired, and proper muscle functioning may aid in the restoration of knee stability.

Our study also builds upon previous work examining in vivo length changes of the intact knee and their potential insight into ACL reconstruction graft fixation angle. It has been proposed that length changes over four to six percent will lead to permanent ACL graft stretch.[1, 8] To prevent such irreversible graft elongation, graft fixation at full extension has been advocated,[4, 26, 30] however, biomechanical studies demonstrated improved kinematics with fixation angles deeper than full extension.[15, 16, 28] Ultimately, there is no consensus on graft fixation angles, and fixation angles between 10° and 90° for the AM graft and 0° and 45° for the PL graft have been used/proposed.[2, 5, 21, 26, 27, 32, 34, 38] Given the importance of length change in causing irreversible graft stretch, previous studies have used end-to-end distances of the ACL to provide insight into optimal graft fixation angles. For example, Yoo et al.[39] analyzed the in vivo ACL length changes in 10 subjects during non-weight-bearing range of motion at fixed angles between 0 and 135° using computer tomography scans. Their results suggested that fixation of both AM and PL grafts should occur near full extension, although they were limited by angular resolution (45° increments). Our data of additional flexion angles may extend these results and provide insight into how different fixation angles could perform in a dynamic setting. Based upon these data, appropriate fixation angles, i.e. not exceeding the critical threshold of six percent, would be <30° for the AM graft, <20° for the SB-anatomic graft, and <10° of flexion for the PL graft. Future studies should compare the kinematics of the knee joint and clinical outcomes after ACL reconstruction using graft fixation angles less than 30° of flexion.

Limitations

There are several limitations to this study. Only one functional activity, a step-up motion, was studied. Other in vivo studies should consider more demanding motions, such as lunging, running and pivoting to assess the effect of excessive rotational moments. No ACL forces were measured. However, no effective ways exist to measure true ACL forces during in vivo activities.[17] It is not possible to identify the exact AM and PL grafts on the 3.0-T MRIs, therefore the highly detailed anatomic descriptions were used.[12, 29] Since the current study was limited to length change measurements only, we cannot assess if the recommended flexion angles are sufficient to prevent excessive femoral roll-back. This study also does not assess temporality; mechanical properties may change over time.

Conclusions

ACL-deficient knees had significantly longer in vivo end-to-end distances between 0° to 30° of flexion for grafts at the AM, PL and SB-anatomic tunnel positions when compared with the intact knees. Graft fixation angles of <30° for the AM, <10° for the PL, and <20° for the SB-anatomic grafts may prevent permanent graft stretch.

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