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The effect of ACL deficiency on the end-to-end distances of the tibiofemoral ACL attachment during in vivo dynamic activity☆

Willem A. Kernkamp [a,b,c,d], Nathan H. Varady [a,b], Jing-Sheng Li [a], Peter D. Asnis [b], Ewoud R.A. van Arkel [c], Rob G.H.H. Nelissen [d], Samuel K. Van de Velde [e], Guoan Li [a,*]

[a] Bioengineering Laboratory, Newton-Wellesley Hospital, Newton, MA, United States of America
[b] Sports Medicine Center, Massachusetts General Hospital/Harvard Medical School, Boston, MA, United States of America
[c] Focus Clinic Orthopedic Surgery, Haaglanden Medical Center, The Hague, the Netherlands
[d] Orthopedic Surgery, Leiden University Medical Center, Leiden, the Netherlands
[e] Division of Pediatric Orthopaedic Surgery, The Hospital for Sick Children, Toronto, Ontario, Canada

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A B S T R A C T

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.

1. 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

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* Corresponding author at: Bioengineering Laboratory, Newton-Wellesley Hospital, Newton, MA 02462, United States of America.
E-mail address: gli1@partners.org. (G. Li).

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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 [1–5] and in vivo studies [6,7] have assessed the length changes of the ACL. Yoo et al. [6] 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. [3] recently examined the differences between intact and ACL-deficient knees in a cadaveric setting. In our recent work [8], 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 [9,10].

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.

2. Methods

2.1. 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 [11].

2.2. Imaging procedure

The MRI and dual fluoroscopic imaging techniques for the measurement of ligament kinematics have been described in detail previously [12]. 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) [13]. 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 <0.1 mm and 0.3° in measuring tibiofemoral joint translations and rotations, respectively [12–14]. The matching procedure was then repeated, providing the in vivo knee kinematics of the step-up motion.

2.3. 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 [15]. 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. [16] and Ferretti et al. [17]. The geometrical centers of the native ACL, AM and PL attachment areas were determined and used as three distinct tibial attachment points (Figure 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. [18] was applied to the 3D models.