Mapping Isometry and Length Changes in Ligament Reconstructions of the Knee

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	prof. dr. J.A.N. Verhaar	EMC
	prof. dr. B. Reider	UChicago

I dedicate this thesis to my grandfather, prof. em. Otto Vos

LIST OF FREQUENTLY USED ABBREVIATIONS

ACL	Anterior cruciate ligament
ALC	Anterolateral complex
ALL	Anterolateral ligament
CT	Computerized tomography
LER	Lateral extra-articular reconstruction
LET	Lateral extra-articular tenodesis
MPFL	Medial patellofemoral ligament
MRI	Magnetic Resonance Imaging
PCL	Posterior cruciate ligament
OA	Osteoarthritis
OTT	Over the top
3D	Three-dimensional

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Chapter 1

General introduction

Injuries are a constant threat at all levels of sports participation. Lower extremity injuries make up more than 66% of sports injuries,[25] with 50% of these injuries involving the knee joint.[6, 9, 25] Due to the considerable forces and large moment arms that occur around the knee during trauma, knee injuries generally result in complete ruptures – rather than sprains – of one or more of its stabilizing ligaments.

The most frequently ruptured ligament is the anterior cruciate ligament (ACL), but other stabilizing structures such as the posterior cruciate ligament (PCL), extra-articular structures including the medial collateral ligament (MCL) anterolateral ligament (ALL)/anterolateral complex (ALC), and the medial patellofemoral ligament (MPFL) in the setting of a patella dislocation are often compromised as well.[20-22, 32, 38, 42, 47, 56, 58] Some patients manage to cope with a ruptured ligament, particularly when only one structure is injured, and may be treated successfully with a post-trauma rehabilitation program without the need for surgery.[7, 11, 17, 19, 36, 44, 53] However, prolonged increased knee laxity caused by the rupture of one of the cruciate ligaments has been shown to result in an increased incidence of joint swelling, pain, instability and meniscal tears in a large subset of patients.[2, 14, 23, 37, 39] Similarly, an untreated MPFL tear could result in recurrent patellofemoral luxation in up to 50% of patients.[3, 8, 12, 15, 16, 35, 64] In the end, the post traumatic knee with a torn ligament is associated with an increased incidence and more rapid progression of osteoarthritis (OA).[22, 23, 27, 43, 46] Interestingly though, this seemingly obvious association has not been corroborated by recent studies.[48] Several reasons for this are present, such as adapting to a different activity level.

Many patients will opt for surgical reconstruction of the ruptured ligament because of the desire to resume sports participation at their pre-injury level, and the hope that future OA may be averted.[60] Different approaches in ACL treatment may be distinguished, such as a more conservative or more surgical approach, while neither of these approaches has shown to be the holy grail. Shared decision making in patients with an ACL injury is an option to overcome this dilemma, but an information lag is always present in patients. For that matter, despite the high patient-reported satisfaction and clinical outcome rates of treatment of a ruptured ligament,[5, 49] less than approximately 80% of the patients return to play and only 50% return to their preinjury level of sports participation.[1, 3, 54] In addition, failure rates for ACL, PCL and MPFL reconstructions have been reported as high as 20-30%.[26, 28, 31, 33, 34, 40, 45, 52, 59, 61, 63] Moreover, ligament reconstruction has been unable to prevent the onset of posttraumatic OA.[29]

The reasons for the low rate of return to pre-injury sports participation are multifactorial and likely include as much psychological and sociological factors as they do biomechanical factors. Likewise, the high rate of OA after knee joint injury despite surgical intervention is probably a complex interaction of the trauma on the cartilage as such, micro traumata in the past, genetic predisposition, biology and biomechanics to mention a few – much of which remains outside our current capacity to modify readily. On the other hand, the high rate of failure of ligament reconstructions is primarily due to technical errors – errors which occur at the time of surgical reconstruction. Of these technical errors, inaccurate positioning of the femoral and tibial tunnels and tensioning of the graft are the most frequently encountered problems.[10, 41, 45, 50, 51, 57, 62]

Optimal tunnel positioning is a critical determinant to achieve successful ligament reconstruction. If the distance between the tunnels increases substantially during flexion or extension of the knee, excessive graft strains emerge and either the motion of the knee is restricted or the graft fails. Alternatively, if the distance between the tunnels decreases during knee motion, the graft slackens and does not provide support. Historically, it was thought that an "isometric" graft, i.e. a graft that maintains the same length as the knee changes flexion angles and thus theoretically provides support without overconstraining the knee joint, would offer the ultimate solution.[55] However, over the past few decades, our understanding of the anatomy and biomechanics of the knee ligaments has significantly improved, and it has been shown that the native anatomy may not yield such isometric behavior. As such, a move away from the quest for isometric ligament reconstruction has occurred towards more anatomic reconstruction, especially for the cruciate ligaments.

It is believed that restoration of the native anatomy will result in restoration of native knee kinematics, and thereby result in the best patient and clinical outcomes. For example, it was recently demonstrated how knees with grafts that more closely restored normal ACL function, and thus knee kinematics, experienced less focal cartilage thinning than did those that experienced abnormal knee motion.[18] Therefore, the transtibial drilling technique traditionally used in ACL reconstruction, which pursues isometric tibiofemoral tunnel positions to minimize graft length changes, made way for tibia-independent techniques, such as anteromedial portal and outside-in retrograde drilling, that were able to restore more accurately the native anatomy and length changes of the native ligament.[24, 30] Others have tried to restore anatomy using a double-bundle reconstruction technique, trying to restore the individual anteromedial and posterolateral bundle of the ACL to better restore rotatory stability of the knee.[13] If non-isometric graft behavior is desired, i.e. elongation patterns that reflect the native knee ligament throughout the range of motion, the angle at which the graft is fixed becomes even more critical, in order to prevent either overconstraint or inability to sufficiently control joint kinematics.

Historically, the effect of tunnel location on graft elongation patterns, subsequent graft forces and kinematics of the knee have been studied using cadaveric specimens during non-physiological loading conditions. This information has been translated to the operating room in which the surgeon determines tunnel location and assesses subsequent graft fixation while ranging the knee through a series of flexion and extension motions, which will approximate but – similar to the cadaveric experiments – will not reproduce the actual complex six-degrees-of-freedom in vivo joint loading of a weightbearing patient. Due to the complexity of muscle loading patterns, the simulation of the human joint function under physiological loads remains difficult to simulate in in-vitro conditions or during non-weightbearing motion. It is therefore difficult to extrapolate the biomechanical behavior of the ligaments of the knee, measured during variable loading conditions in cadaveric studies or non-weightbearing in vivo studies, to the elongation patterns seen during in-vivo weightbearing flexion of the knee.

The ***aim of this thesis*** was to measure the isometry and length changes of the most frequently reconstructed knee ligaments, i.e. the ACL, PCL, and the extra-articular ALL/ALC and the MPFL, under in-vivo weightbearing conditions. Thus, a comprehensive database of knee ligament biomechanics will be created, which could be readily used by surgeons to optimize the desired tunnel position and angle of graft fixation to have optimal biomechanical stability. Secondly, such database will improve knowledge of the surgeon on the impact of altering the tunnel position on ligament biomechanics. The latter will decrease future graft failure rates due to tunnel malpositioning.

In ***Chapter 2***, the in vivo isometry and length changes of the ACL in the healthy knee are presented. Knowledge of behavior of the native ligaments is crucial to understand and evaluate contemporary ACL reconstruction options. In ***Chapter 3*** the length changes of the anatomic anteromedial, posterolateral and an anatomic single-bundle ACL reconstruction were simulated and compared between healthy and ACL-deficient knees. In ***Chapter 4*** the isometry and length changes of the PCL were studied in healthy knees. Following the description of the length changes of the intra-articular ligaments, our studies on extra-articular stabilizing structures are discussed. In ***Chapter 5*** we measured the length changes of the anatomic ALL, since it is thought that the ALL may play an important role in restraining internal tibial rotation and may improve postoperative knee kinematics in some patients. As it became evident that an anatomic ALL reconstruction would be unable to provide a constraint in extension and early knee flexion, as well as to maintain normal knee laxity during deeper knee flexion angles, we further evaluated the isometry and length changes of the anterolateral aspect of the knee, in order to find the most optimal location for graft placement for lateral extra-articular reconstructions (***Chapter 6)***. Thus, illustrating the quandary between a biomechanically optimal reconstruction and the reproduction of native anatomy. In ***Chapter*** 7 the MPFL was analyzed. Finally, ***Chapter 8*** summarizes the studies described in this thesis with a general discussion and final conclusions.

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Chapter 2

An in vivo prediction of anisometry and strain in anterior cruciate ligament reconstruction – A combined magnetic resonance and dual fluoroscopic imaging analysis

Willem A. Kernkamp
Nathan H. Varady
Jing-Sheng Li
Tsung-Yuan Tsai
Peter D. Asnis
Ewoud R. A. van Arkel
Rob G. H. H. Nelissen
Thomas J. Gill
Samuel K. Van de Velde
Guoan Li

Arthroscopy. 2018 Apr;34(4):1094-1103.

ABSTRACT

Purpose: To evaluate the in vivo anisometry and strain of theoretical anterior cruciate ligament (ACL) grafts in the healthy knee using various socket locations on both the femur and tibia.

Methods: Eighteen healthy knees were imaged using magnetic resonance imaging and dual fluoroscopic imaging techniques during a step-up and sit-to-stand motion. The anisometry of the medial aspect of the lateral femoral condyle was mapped using 144 theoretical socket positions connected to an anteromedial, central, and posterolateral attachment site on the tibia. The 3-dimensional wrapping paths of each theoretical graft were measured. Comparisons were made between the anatomic, over the top (OTT), and most isometric (isometric) femoral socket locations, as well as between tibial insertions.

Results: The area of least anisometry was found in the proximal-distal direction just posterior to the intercondylar notch. The most isometric attachment site was found midway on the Blumensaat line with approximately 2% and 6% strain during the step-up and sit-to-stand motion, respectively. Posterior femoral attachments resulted in decreased graft lengths with increasing flexion angles, whereas anterodistal attachments yielded increased lengths with increasing flexion angles. The anisometry of the anatomic, OTT and isometric grafts varied between tibial insertions ($P < .001$). The anatomic graft was significantly more anisometric than the OTT and isometric graft at deeper flexion angles ($P < .001$).

Conclusions: An area of least anisometry was found in the proximal-distal direction just posterior to the intercondylar notch. ACL reconstruction at the isometric and OTT location resulted in nonanatomic graft behavior, which could overconstrain the knee at deeper flexion angles. Tibial location significantly affected graft strains for the anatomic, OTT, and isometric socket location.

Clinical Relevance: This study improves the knowledge on ACL anisometry and strain and helps surgeons to better understand the consequences of socket positioning during intra-articular ACL reconstruction.

INTRODUCTION

Socket positioning is one of the most critical steps in successful anterior cruciate ligament (ACL) reconstruction. ACL socket locations yielding less favorable graft behavior could lead to permanent graft stretch and graft failure. Data from the Swedish ACL registry[27] showed that more complete anatomic reconstruction reduces the risk for revision surgery. In addition, the importance of anatomic graft placement for the longevity of articular cartilage was recently emphasized by DeFrate, demonstrating how knees with grafts that more closely restored normal ACL function, and thus knee kinematics, experienced less focal cartilage thinning than did those that experienced abnormal knee motion.[2]

Over the last decade, a transition has taken place encouraging more anatomic placement of the femoral socket. Consequently, the classical transtibial femoral drilling technique, which aims to minimize graft length changes during knee flexion, has made way for tibia-independent drilling techniques (e.g., anteromedial portal and outside-in retrograde drilling), which allow for more anatomic graft placement. These techniques are associated with greater length changes during knee flexion,[17] however. Thus, it is paramount for surgeons to have a good understanding of the relation between socket positioning and ACL graft length changes. As strains of 4% to 6% can result in permanent graft stretch and/or failure,[23, 32] correct fixation angle and tensioning may be especially important for successful clinical outcomes in anisometric ACL reconstruction. Numerous ex vivo studies have explored the isometry of the ACL.[8, 14, 17, 25, 31] However, these cadaveric studies have yielded inconsistent results. Moreover, ex vivo studies are unable to consider muscle forces that control the knee during dynamic in vivo motion. Therefore, care should be taken when translating the ex vivo biomechanical measurements to the results, which would be seen in the knee during in vivo weight-bearing motion and detailed information on the effect of various socket positions during in vivo loading of the knee is lacking. Therefore, mapping the in vivo anisometry of various theoretical ACL grafts may help improve socket placement during ACL reconstruction and surgeons' understanding of its effect on graft length. The purpose of this study was to evaluate the in vivo anisometry and strain of theoretical ACL grafts in the healthy knee using various socket locations on both the femur and tibia. The hypothesis was that grafts placed more posteriorly (on both the femur and tibia) would yield more anisometric behavior during knee flexion.

METHODS

Participants

This study was approved by our institutional review board and written consent was obtained from each participant prior to taking part in this study project. All participants were tested between November 2008 and April 2010 to study the normal in vivo knee kinematics during dynamic functional activities. In this study, 18 healthy knees were studied (12 men, 6 women; age 35.4 ± 10.9 years (mean ± standard deviation); body height 175 ± 9 cm; body weight 83.3 ± 18.0 kg; body mass index 27 ± 3.5; KT-1000 67 N, 89 N, and 134 N anterior force translations were 1.8 ± 1.1 mm, 2.9 ± 1.3 mm, and 4.4 ± 1.8 mm, respectively) to investigate the strain of various theoretical ACL grafts.

All participants meeting the inclusion and exclusion criteria were enrolled through our institutional broadcast e-mail announcements. The inclusion criteria consisted of participants 18 to 60 years old with the ability to perform daily activities independently without any assistance device and without taking pain medication. Standard knee examination was performed on the knee, including the Lachman and anterior drawer test, and participants with increased laxity were excluded. Other exclusion criteria were knee pain, previous knee injury, and previous surgery to the studied lower limb. The magnetic resonance imaging (MRI) scan of the knee of each participant was assessed for potential meniscal tears, chondral defects, and ligamentous injuries; if present, the participant was excluded from further analysis.

Imaging procedure

The MRI and dual fluoroscopic imaging techniques for the measurement of ligament kinematics have been described in detail previously.[15] MRI scans of the knee joints were done in both sagittal and coronal planes using a 3-Tesla MRI scanner (MAGNETOM Trio; Siemens, Malvern, PA) with a double-echo water excitation sequence (thickness 1 mm; resolution of 512 × 512 pixels).[3] The images were then imported into solid modeling software (Rhinoceros; Robert McNeel and Associates, Seattle, WA) to construct 3-dimensional (3D) surface models of the tibia, fibula, and femur.

The knee of each participant was simultaneously imaged using 2 fluoroscopes (BV Pulsera; Philips, Eindhoven, the Netherlands) as the participant performed a step-up (55º ± 4º) and sit-to-stand motion (88º ± 10º). 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 participant. Finally, the MRI-based knee model of each participant 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 6 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 0.3º in measuring tibiofemoral joint translations and rotations, respectively.[3, 15, 16]

Tibiofemoral attachment points

To determine the in vivo length patterns of theoretical grafts during motion, various tibial and femoral attachment sites were used. The tibial attachment areas of the ACL were determined based on the MR images in both sagittal and coronal planes.[9] The anatomic ACL attachment area was directly mapped onto the 3D MRI-based tibia model. The attachment area was then subdivided into an anteromedial and posterolateral portion guided by the meticulously performed anatomic descriptions of Edwards et al.[5] and Ferretti et al.[6] The geometrical centers of the ACL, anteromedial, and posterolateral attachment areas were determined and used as 3 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.[1] was applied to the 3D models. The most anterior edge of the femoral notch roof was chosen as the reference for the grid alignment (line t), that is, the Blumensaat line (which in fact is a derivative of the true Blumensaat line, since the latter is a radiograph finding, whereas the line used in the current study was based on 3D models).[7] The segments along line t and perpendicular to line t (line h) were divided into fourths. The medial view was used to project 144 femoral attachment points to the medial aspect of the lateral femoral condyle (Fig. 2A). The region of interest for the femoral points was determined by the bony edges of the medial aspect of the lateral femoral condyle, that is, using the cartilage as borders. The region of interest was then further dived into 16 subareas (Fig. 2B). Finally, the anatomic and transtibial over-the-top (OTT) ACL socket locations were identified based on Parkar et al.[20]

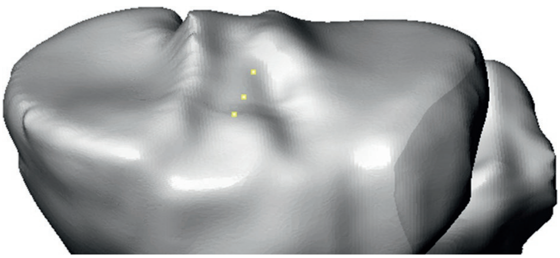

Fig. 1 Proximal-distal view of a 3D tibia and fibula model showing the distribution of the anteromedial, central, and posterolateral tibial attachment points.

Strain measurements

The length changes for each theoretical graft were measured as a function of knee flexion. The direct line connecting the femoral and tibial attachment point was projected on the bony surfaces. This allowed to create a line that avoids penetration through bone, and therefore followed bony geometry, that is, a wrapping path (Fig. 3). An optimization procedure was implemented to determine the projection angle to find the shortest 3D wrapping path (this is to mimic a path of minimal resistance) at each flexion angle of the knee. This technique has been described in previous studies for measurements of ligament kinematics.[30] The length of the projected line (i.e., curved around the bony surfaces) was measured as the length of the graft. Following the methods by Taylor et al.,[28] ACL strain was measured from the ACL length changes relative to a reference as follows: $\varepsilon = L - L_0 / L_0 \times 100\%$, where ε is relative graft strain, L is graft length, and L_0 is a reference length (defined as the length of the nonweight-bearing MR imaging position). A heat map was created to provide visual representation of the anisometry distribution over the medial aspect of the lateral femoral condyle by using the mean maximum strain - mean minimum strain of each theoretical tibiofemoral graft during both motions.

Statistics

Data were first pooled according to tibial attachment sites. A 2-way analysis of variance (ANOVA) was used to assess for differences in mean anisometry due to tibial attachment sites, flexion angle, and their interaction. Then, for each femoral attachment site, a 2-way ANOVA was used to examine differences in anisometry between the 3 studied tibial attachments. If significant, Tukey honestly significant difference post hoc tests were performed to compare between pairs of the 3 individual tibial socket positions. A similar procedure was then implemented with data pooled by femoral attachment site. A 2-way ANOVA was used to assess for differences in mean anisometry due to femoral attachment sites, flexion angle, and their interaction. Then, for each tibial attachment site, a 2-way ANOVA was used to examine differences in anisometry between the 3 studied femoral attachments. If significant, Tukey honestly significant difference post hoc tests were performed to compare between pairs of the 3 individual femoral socket positions. In contrast to the tibial pool, the interaction between femoral socket location and flexion angle was significantly associated with anisometry patterns for the femoral sockets. Therefore, Tukey honestly significant difference tests were also employed to examine differences between the femoral socket positions at each flexion angle. All analyses were performed in R version 3.3.2, and P values less than .05 were considered significant.

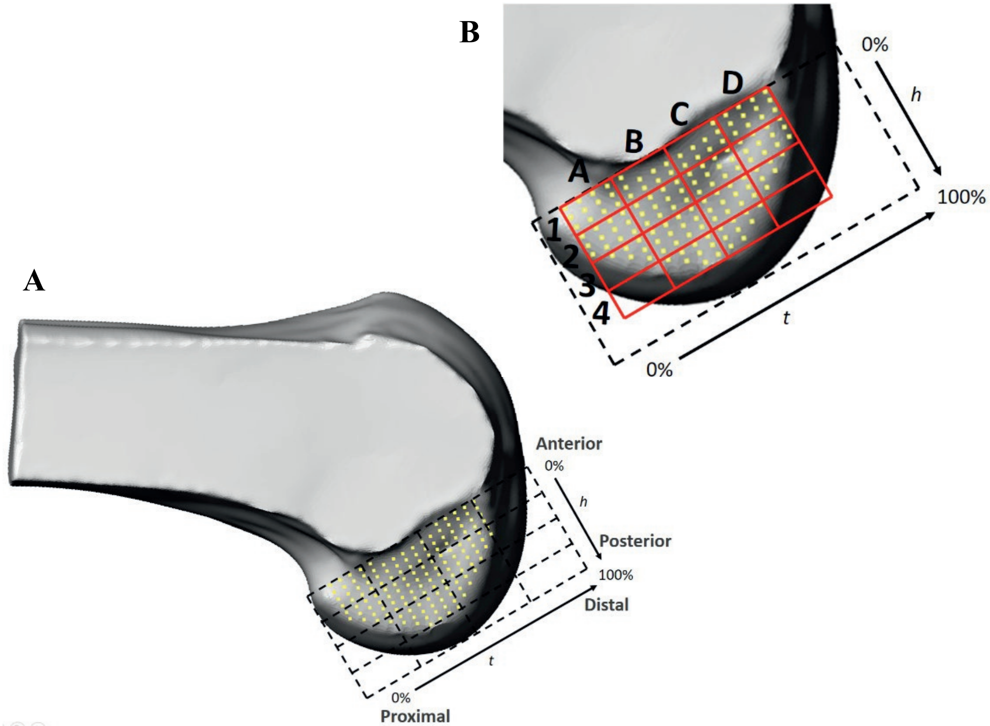

Fig. 2 (**A**) Medial view of a 3D femur model in 90º of flexion. The 4 × 4 grid as developed by Bernard et al.[1] was applied to the medial aspect of the lateral femoral condyle. A line extending along the Blumensaat line was used as a landmark for the anterior border of the grid (line t). Parallel 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 2 lines perpendicular to the Blumensaat line (line h) originating from the proximal and distal bony borders of the lateral femoral condyle. Line h: maximum distance from the proximal condylar bony border to femoral joint line. Line t: maximum distance perpendicular from the Blumensaat line to the posterior edge of the lateral condyle. (**B**) The medial view was used to project 144 femoral attachment points to the medial aspect of the lateral femoral condyle. The region of interest for the femoral points was determined by the bony edges of the medial aspect of the lateral femoral condyle, that is, using the cartilage as borders. The region of interest was then further dived into 16 subareas. Distal to proximal direction A to D; anterior to posterior direction 1 to 4.

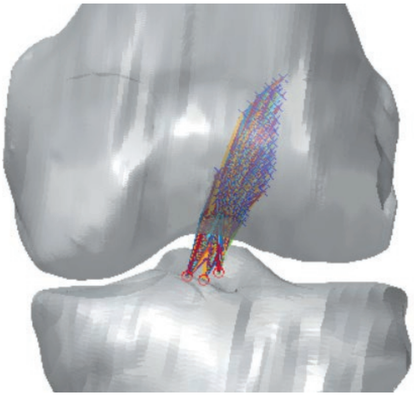

Fig. 3 Anterior-posterior view of a 3D knee model illustrating the lines curving over the bony geometry of the femur and tibia, that is, the "wrapping effect." At each flexion angle, an optimization procedure was implemented to determine the graft projection angle to find the shortest 3D wrapping path,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

RESULTS

Posterior to the femoral intercondylar notch, running in the proximal-distal direction, a zone demonstrated least anisometry during the step-up and sit-to-stand motions (i.e., the blue area on the medial aspect of the lateral femoral condyle in Figs 4 and 5). The most isometric attachment location when connected to the anteromedial, central, or posterolateral tibial attachments for each activity is described in Table 1. Attachments located posteriorly to the isometric zone resulted in decreased graft lengths with increasing flexion angles (Fig. 6), whereas distal-anterior grafts increased in length with increasing flexion angles. The anisometry heatmap during both the step-up and sit-to-stand motion is illustrated in Video 1 available on the journal's website.

Femoral comparison

During step-up and sit-to-stand motion, when the femoral bundles were connected to any of the 3 tibial locations, the isometric attachment was significantly more isometric than the anatomic ($P < .001$) and the OTT location ($P < .001$); the OTT location was significantly more isometric than the anatomic ($P < .001$) (Table 2). When connected to the central tibial location, significant differences in strain were found between the anatomic versus isometric locations from 20° to 50° of flexion ($P < .001$), anatomic versus OTT from 25° to 50° of flexion (25°, $P = .004$, 30°-50°, $P < .001$) and for the isometric versus OTT location from 30° to 50° of flexion (30°, $P = .03$, 40°-50°, $P < .001$) (Fig. 7A, Table 3). Results for the sit-to-stand motion are mentioned in Fig. 7B and Table 3.

Table 1. Most isometric graft locations.

	Step-up		**Sit-to-stand**	
	Length change (% and CI 95)	Location (t† x h‡)	Length change (% and CI 95)	Location (t† x h‡)
Anteromedial	1.7 (1.4 to 1.9)	50 x 14	2.2 (1.9 to 2.5)	43 x 8
Central	1.8 (1.5 to 2.1)	48 x 8	3.1 (2.7 to 3.5)	43 x 8
Posterolateral	2.2 (1.8 to 2.5)	48 x 8	5.2 (4.6 to 5.9)	43 x 8

†*h:* percentage along line *h* (this is perpendicular to the Blumensaat line)
‡*t:* percentage along line *t* (this is parallel to the Blumensaat line)

Tibial comparison

For the step-up motion, when connected to the isometric femoral socket, no significant differences in anisometry were found between the anteromedial and central ($P = .14$) or central and posterolateral ($P = .15$) tibial attachments; the anteromedial and posterolateral tibial attachment were significantly different ($P < .001$). When grafts were attached to the anatomic femoral socket, the anteromedial and central tibial attachments were not statistically different ($P = .08$); significant differences were found between the anteromedial and posterolateral ($P < .001$), and central and posterolateral attachments ($P = .017$). When connected to the OTT socket location, significant differences in mean isometry were found between the anteromedial and central attachment ($P = .003$), and the anteromedial and posterolateral attachment ($P < .001$), and the central and posterolateral attachment ($P < .001$) (Table 2). Results for the sit-to-stand motion are mentioned in Table 2.

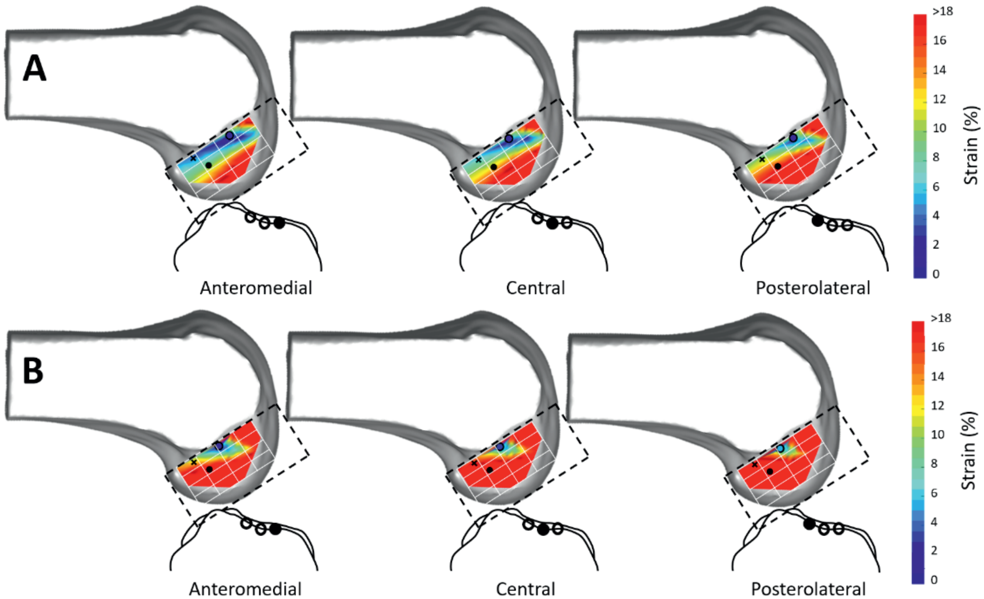

Fig. 4 Medial view of a 3D femur model in 90° of flexion. The "heat map" illustrates the isometry distribution (mean maximum strain – minimum strain) over the medial aspect of the lateral femoral condyle for single point-to-point curves when connected to the anteromedial, central, or posterolateral tibial attachment during the dynamic step-up (**A**) and sit-to-stand motion (**B**). The darkest blue area on the femur represents the most isometric attachment area, whereas the red areas highlights those with a high degree of anisometry. Specifically, the circle represents the most isometric attachment. The black cross (x) on the femur shows the "over the top" position as would be achieved by transtibial drilling; the black dot shows the center of the ACL footprint as described by Parkar et al.[18]

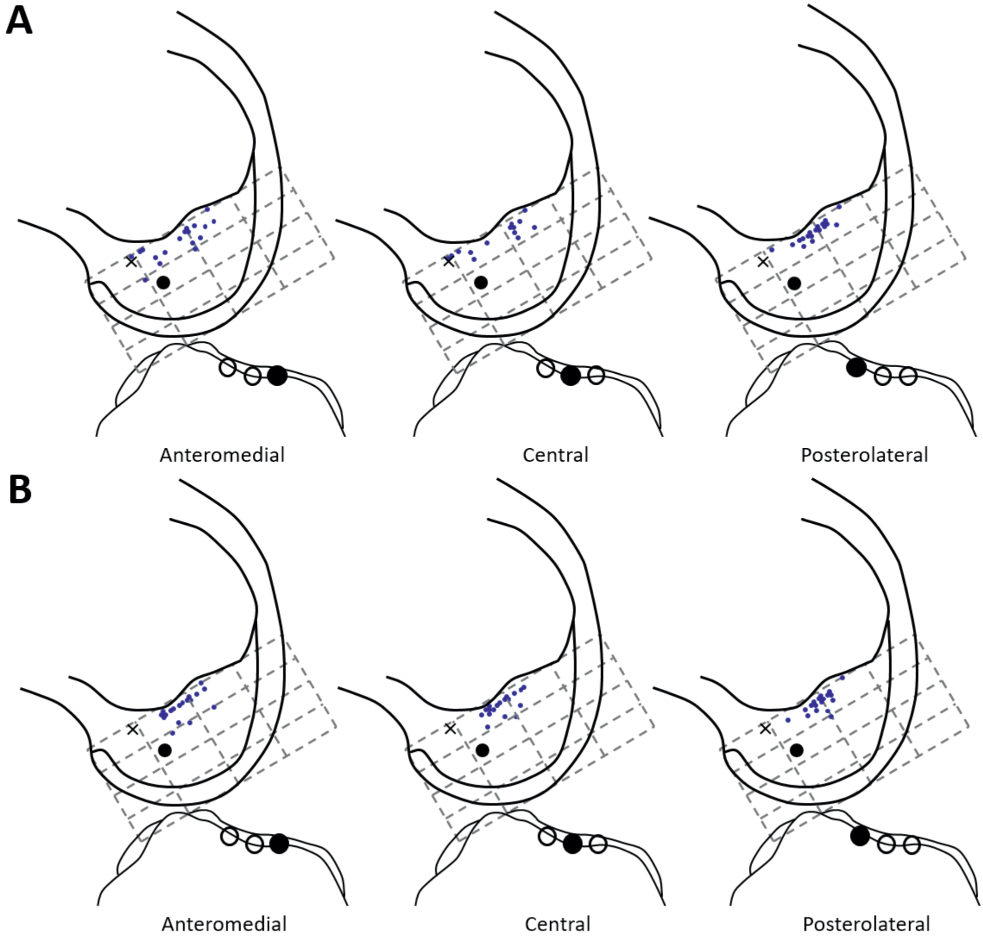

Fig. 5 Medial view of a schematic femur model in 90º of flexion. The most isometric location (mean maximum strain – minimum strain) on the medial aspect of the lateral femoral condyle per participant is illustrated when connected to the anteromedial, central, or posterolateral tibial attachment during the dynamic step-up (**A**) and sit-to-stand motion (**B**). The black cross (x) on the femur shows the "over the top" position as would be achieved by transtibial drilling; the black dot shows the center of the ACL footprint as described by Parkar et al.[18]

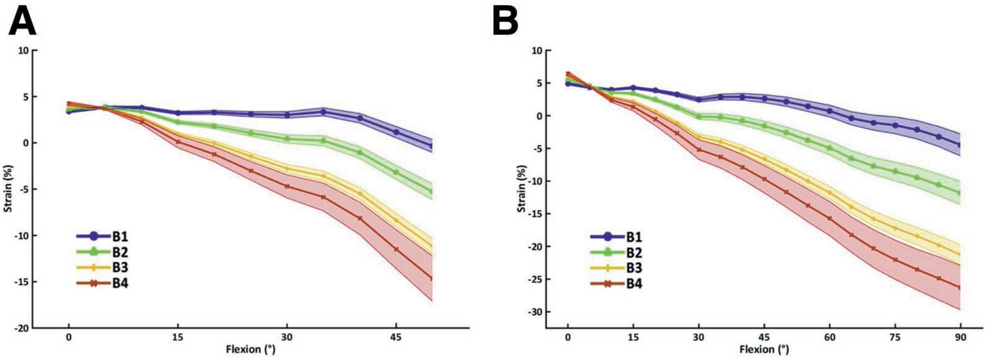

Fig. 6 Strain per area in the anterior to posterior direction, for example, B1 (anterior) to B4 (posterior) during the dynamic step-up (**A**) and sit-to-stand (**B**) motion when connected to the anteromedial tibial attachment. Values are presented as mean and 95% confidence interval.

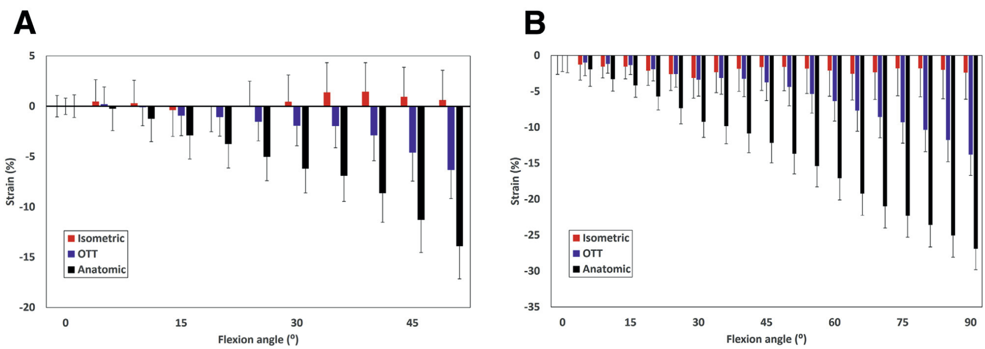

Fig. 7 Strain per area in the anterior to posterior direction, for example, B1 (anterior) to B4 (posterior) during the dynamic step-up (**A**) and sit-to-stand (**B**) motion when connected to the anteromedial tibial attachment. Values are presented as mean and 95% confidence interval.

Table 2. Statistical analysis for isometry of the various studied bundles in the step-up motion (A) and sit-to-stand motions (B). The three femoral attachments: anatomic ACL center (anatomic), over the top (OTT) and most isometric location; and three tibial locations: anteromedial, central and posterolateral.

(A) Step-up			
Femur			
Tibia	Anatomic vs Isometric	Anatomic vs OTT	OTT vs Isometric
Anteromedial	$p < 0.001$	$p = 0.01$	$p < 0.001$
Central	$p < 0.001$	$p < 0.001$	$p < 0.001$
Posterolateral	$p < 0.001$	$p < 0.001$	$p < 0.001$
Tibia			
Femur	Anteromedial vs Central	Anteromedial vs Posterolateral	Central vs Posterolateral
Anatomic	$p = 0.08$	$p < 0.001$	$p = 0.017$
OTT	$p = 0.003$	$p < 0.001$	$p < 0.001$
Isometric	$p=0.14$	$p < 0.001$	$p = 0.15$

(B) Sit-to-stand			
Femur			
Tibia	Anatomic vs Isometric	Anatomic vs OTT	OTT vs Isometric
Anteromedial	$p < 0.001$	$p < 0.001$	$p < 0.001$
Central	$p < 0.001$	$p < 0.001$	$p < 0.001$
Posterolateral	$p < 0.001$	$p < 0.001$	$p < 0.001$
Tibia			
Femur	Anteromedial vs Posterolateral	Anteromedial vs Posterolateral	Central vs Posterolateral
Anatomic	$p = 0.004$	$p < 0.001$	$p < 0.001$
OTT	$p < 0.001$	$p < 0.001$	$p < 0.001$
Isometric	$p = 0.06$	$p < 0.001$	$p = 0.004$

Note: p-values represent statistical significant differences in anisometry (*mean maximum strain – mean minimum strain*).

Table 3. Statistical analysis (results of Tukey's HSD analyses) of between-group length change by knee flexion angle in step-up (**A**) and sit-to-stand (**B**) motion; comparing the anatomic ACL center (anatomic), over the top (OTT), and most isometric bundles when connected to the central tibial location.

(A) Step-up				**(B) Sit-to-stand**			
Flexion angle (º)	**Anatomic vs Isometric**	**Anatomic vs OTT**	**Isometric vs OTT**	**Flexion angle (º)**	**Anatomic vs Isometric**	**Anatomic vs OTT**	**Isometric vs OTT**
0	$p = 0.99$	$p = 0.66$	$p = 0.63$	0	$p = 0.99$	$p = 0.83$	$p = 0.8$
5	$p = 0.93$	$p = 0.93$	$p = 0.76$	5	$p = 0.91$	$p = 0.9$	$p = 0.68$
10	$p = 0.33$	$p = 0.92$	$p = 0.57$	10	$p = 0.42$	$p = 0.94$	$p = 0.62$
15	$p = 0.07$	$p = 0.39$	$p = 0.65$	15	$p = 0.04$	$p = 0.2$	$p = 0.72$
20	$p = 0.01$	$p = 0.06$	$p = 0.35$	20	$p < 0.001$	$p = 0.075$	$p = 0.26$
25	$p = 0.010$	$p = 0.03$	$p = 0.14$	25	$p < 0.001$	$p = 0.009$	$p = 0.16$
30	$p < 0.001$	$p < 0.001$	$p = 0.03$	30	$p < 0.001$	$p = 0.002$	$p = 0.12$
35	$p < 0.001$	$p < 0.001$	$p = 0.007$	35	$p < 0.001$	$p < 0.001$	$p = 0.03$
40	$p < 0.001$	$p < 0.001$	$p < 0.001$	40	$p < 0.001$	$p < 0.001$	$p = 0.01$
45	$p < 0.001$	$p < 0.001$	$p < 0.001$	45	$p < 0.001$	$p < 0.001$	$p = 0.001$
50	$p < 0.001$	$p < 0.001$	$p < 0.001$	50	$p < 0.001$	$p < 0.001$	$p < 0.001$
				55	$p < 0.001$	$p < 0.001$	$p < 0.001$
				60	$p < 0.001$	$p < 0.001$	$p < 0.001$
				65	$p < 0.001$	$p < 0.001$	$p < 0.001$
				70	$p < 0.001$	$p < 0.001$	$p < 0.001$
				75	$p < 0.001$	$p < 0.001$	$p < 0.001$
				80	$p < 0.001$	$p < 0.001$	$p < 0.001$
				85	$p < 0.001$	$p < 0.001$	$p < 0.001$
				90	$p < 0.001$	$p < 0.001$	$p < 0.001$

Notes: p-values represent statistical significant differences in strain change change between the anatomic, OTT and most isometric bundles.

DISCUSSION

In this study, the most isometric femoral socket location was approximately midway on the Blumensaat line just posterior to the intercondylar femoral notch. This was true for the 3 studied tibial attachments (i.e., anteromedial, central, and posterolateral location) during both motions. A graft in this position underwent approximately 2% and 6% strain during the step-up and sit-to-stand motion, respectively. The theoretical ACL strains were most affected by changing the femoral socket positions in the anterior-posterior direction. Posterior femoral attachments resulted in decreased lengths with increasing flexion angles, whereas anterior-distal grafts increased in length with increasing flexion angles.

Traditional thinking in ACL reconstruction has focused on avoiding peak graft strains at full-extension, as strains greater than 4% to 6% are known to lead to undesirable graft behavior namely, overconstraint and potentially graft failure.[23, 32] Therefore, depending on the tibiofemoral socket positions, and thus the anisometry pattern, the fixation angle is a crucial variable in achieving desirable graft behavior. This is especially true for anisometric grafts, which experience greater length changes over knee range of motion. As evidenced by this study, anteriorly positioned femoral sockets show less length change, particularly pronounced during the extension to early flexion range, than more posteriorly positioned sockets, which greatly decrease in length with increasing flexion (Fig. 6). For example, graft fixation at 30º of flexion may have detrimental consequences if one prefers to place the femoral socket posteriorly (e.g., quadrants B3-4) over time because of repetitive stretch-shortening cycles from 30º to full extension. This may be especially important for the posterolateral socket during double bundle ACL reconstruction. In contrast, a surgeon may have more flexibility in fixation angle when aiming for anterior socket positioning.

Given the importance of avoiding peak strains, it may be surprising that isometric ACL reconstruction techniques are not associated with improved clinical outcomes. However, our study demonstrates that the most isometric point on the femur is located far from the anatomic ACL insertion site (Figs 4 and 5). This means that a socket drilled at the isometric location (i.e., distal and anterior to the center of the ACL footprint) will result in a nonanatomic ACL reconstruction. In fact, given their relatively constant strains, isometric and OTT grafts may experience a relatively higher strain at deeper flexion angles than an anatomic ACL reconstruction. Specifically, the isometric and OTT locations had significantly higher strains than the anatomic location (i.e., strains closer to their 0º strain, whereas the anatomic ACL decreased more in relative length) beyond approximately 20º of knee flexion. The theoretical isometric and OTT grafts yielded more isometric behavior, and are therefore relatively "longer" than an anatomic ACL reconstruction. These increased relative strains compared with the anatomic reconstruction may account for the lack of improved clinical outcomes with nonanatomic reconstructions.[2,12] Previous studies evaluating socket position in revision ACL reconstruction cases found a tendency of more anteriorly placed femoral socket and posteriorly placed tibial socket.[10, 21, 29] Although these grafts might in theory have been relatively isometric based on the anterior femoral attachment, the biomechanically inadequate orientation of the graft could have placed the reconstruction at risk of failure.

Recent anatomic studies have revealed 2 types of femoral attachment fibers of the ACL, namely, a direct type and an indirect type.[18, 24, 26] In the in vitro setting, simulated tests of uniplanar anterior and combined anterior and rotatory loads have indicated that the direct attachment serves primarily in restraining anterior tibial translation.[13, 19, 22] In addition, Nawabi et al.[19] found the direct attachment to form a key link in transmitting mechanical load to the joint (i.e., bear more force) and to be more isometric than the indirect attachment. Kawaguchi et al.[13] showed that the direct attachment (areas G and H in their study) of the ACL resisted 82% to 90% of the anterior drawer force, with most load carried by the fibers closest to the roof of the intercondylar notch (66%-84%). Interestingly, this key region for force transfer (areas G and H[13]) is located near the isometric area (dark blue zone in Fig. 4) during in vivo knee flexion as demonstrated by our study. Given DeFrate's recent work[2] demonstrating the importance of restoring functional anatomy and the concordance of isometry between recent ex vivo studies and this in vivo study, these results may encourage future research elucidating functional anatomic ACL reconstruction techniques focused on restoring the anteriorly located direct fibers of the ACL.

Another variable that is directly related to the socket position is the functional length of the graft, which is an important variable in any ligament reconstruction. Stress-strain curves consist of a nonlinear toe region and a linear region. Long grafts undergo greater elongation under the same load compared with short grafts for both nonlinear and linear regions. This means that decreasing the length of a graft, that is, a femoral socket that has close proximity to the tibial socket, linearly increases its stiffness.[4] Therefore, the socket position of the ACL graft determines the effective length and thus plays an important role in the kinematic response of the knee. In the current study, it was found that the tibial location significantly affected the mean anisometry. In the recent study by Inderhaug et al.,[11] it was shown that posterior tibial socket positioning was related to an increased rate of revision cases. Future studies may further explore the effect and its significance of the tibial socket positioning.

The present description of in vivo graft anisometry at various positions is critical information for further follow-up studies on graft behavior and clinical outcome. Independent of surgical technique, these data could help surgeons to improve the socket position and fixation angle. Moreover, these data may be useful in the setting of ACL revision; while previous studies have typically only examined the anatomic ACL insertion site, this study provides a map of the entire medial aspect of the lateral femoral condyle, which may be useful if the anatomic site is compromised.

Limitations

There are several limitations to this study. Only data from healthy knees during 2 functional activities were used. No full range-of-motion activity was studied; more specifically, no hyperextension or flexion angles beyond 90º of flexion were analyzed. Future research should consider knees with a torn ACL and more demanding in vivo functional activities (e.g., lunging, running, and jumping). No pivoting motion was performed in this study, and thus the effect of excessive rotational moments could not be assessed. In this study, strain was measured using the reference length as measured from the non-weightbearing MR imaging position. The precise reference lengths (zero-load length) are unknown because of the in vivo nature of the study. However, previously this measurement has been shown to be linearly related to the true strain.[28] Finally, no actual ACL reconstructions were performed in the present study, so no definite conclusions could be generated regarding the most optimal socket positions.

Conclusions

An area of least anisometry was found in the proximal-distal direction just posterior to the intercondylar notch. ACL reconstruction at the isometric and OTT location resulted in nonanatomic graft behavior, which could overconstrain the knee at deeper flexion angles. Tibial location significantly affected graft strains for the anatomic, OTT, and isometric socket location.

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Chapter 3

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

Willem A. Kernkamp
Nathan H. Varady
Jing-Sheng Li
Peter D. Asnis
Ewoud R. A. van Arkel
Rob G.H.H. Nelissen
Samuel K. Van de Velde
Guoan Li

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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]

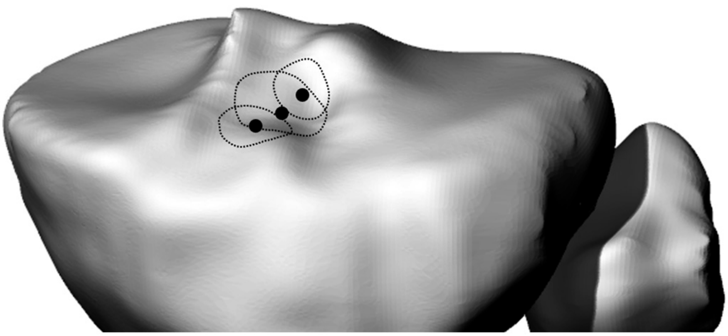

Fig. 1 Three-dimensional tibia model showing the distribution of the anteromedial, central and posterolateral tibial attachment points.

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.

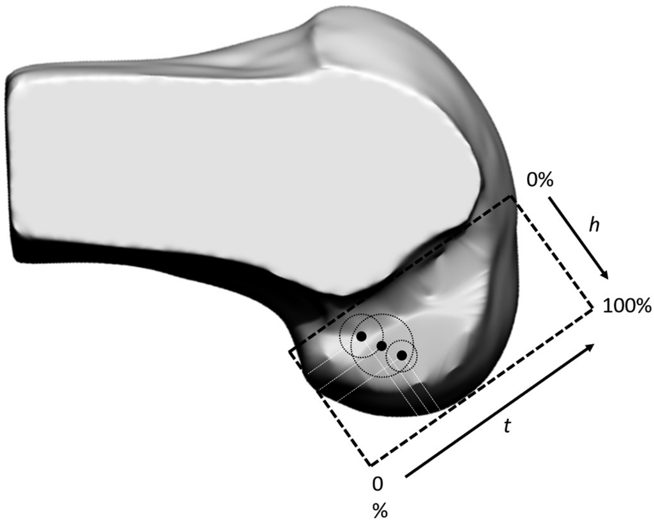

Fig. 2 Medial view of a 3D femur model in 90° of flexion. Bernard et al's [6] quadrant method was applied to the medial aspect of the lateral femoral condyle. A line extending along the Blumensaat line was used as a landmark for the anterior border of the grid (line t). Parallel to line t, a line was 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.

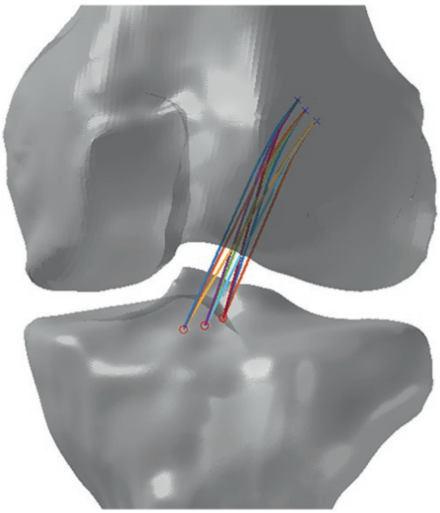

Fig. 3 Anterior–posterior view of a 3D knee model illustrating the lines curving over the bony geometry of the femur and tibia, that is, the "wrapping effect." At each flexion angle, an optimization procedure was implemented to determine the graft projection angle to find the shortest 3D wrapping path, 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 ($P = 0.01$).

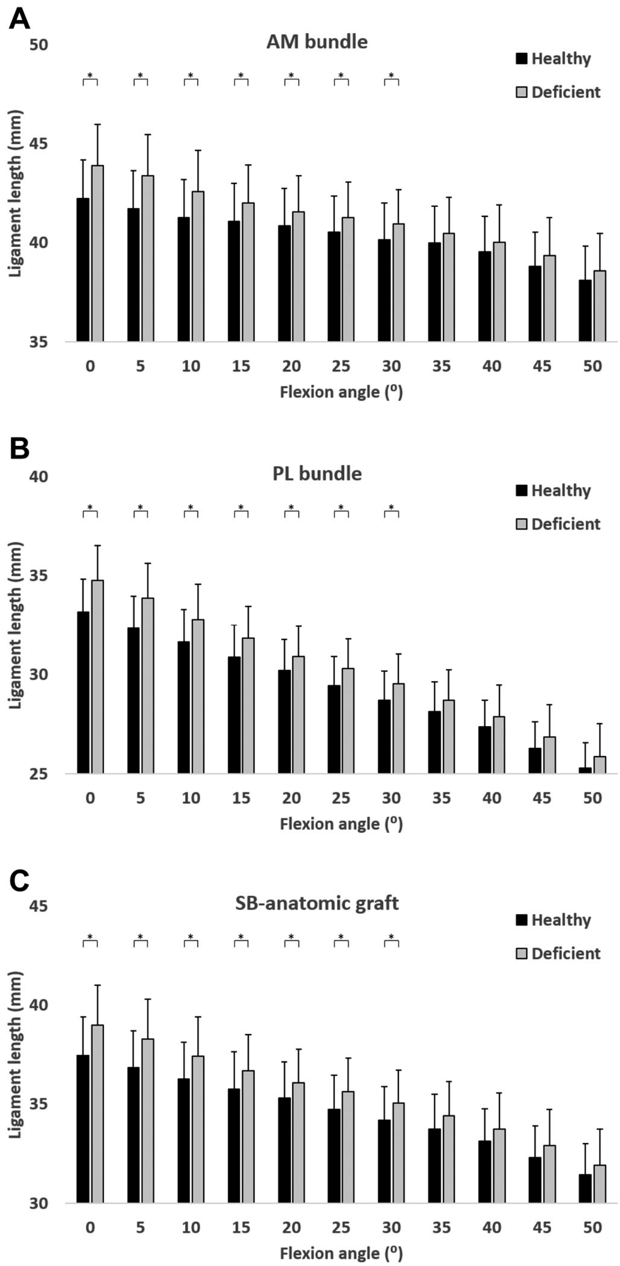

Fig. 4 Absolute length changes for the anteromedial bundle, posterolateral bundle and single-bundle anatomic graft for the ACL-deficient (gray) and intact knees (black) during the dynamic step-up motion. Mean values are shown, with the shaded area indicating the 95% confidence interval (CI).

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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Chapter 4

Anatomic is better than isometric posterior cruciate ligament tunnel placement based upon in vivo simulation

Willem A. Kernkamp
A.J.T. Jens
N.H. Varady
Ewoud R. A. van Arkel
Rob G. H. H. Nelissen
Peter D. Asnis
Robert F. LaPrade
Samuel K. Van de Velde
Guoan Li

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ABSTRACT

Purpose: To elucidate the effects of various tibial and femoral attachment locations on the theoretical length changes and isometry of PCL grafts in healthy knees during in vivo weightbearing motion.

Methods: The intact knees of 14 patients were imaged using a combined magnetic resonance and dual fluoroscopic imaging technique while the patient performed a quasi-static lunge (0°–120° of flexion). The theoretical end-to-end distances of the 3-dimensional wrapping paths between 165 femoral attachments, including the anatomic anterolateral bundle (ALB), central attachment and posteromedial bundle (PMB) of the PCL, connected to an anterolateral, central, and posteromedial tibial attachment were simulated and measured. A descriptive heatmap was created to demonstrate the length changes on the medial condyle and formal comparisons were made between the length changes of the anatomic PCL and most isometric grafts.

Results: The most isometric graft, with approximately 3% length change between 0° and 120° of flexion, was located proximal to the anatomic femoral PCL attachments. Grafts with femoral attachments proximal to the isometric zone decreased in length with increasing flexion angles, whereas grafts with more distal attachments increased in length with increasing flexion angles. The ALB and central single-bundle graft demonstrated a significant elongation from 0° to 120° of flexion ($P < 0.001$). The PMB decreased in length between 0° and 60° of flexion after which the bundle increased in length to its maximum length at 120° ($P < 0.001$). No significant differences in length changes were found between either the ALB or PMB and the central graft, and between the ALB and PMB at flexion angles ≥ 60° (n.s.).

Conclusions: The most isometric attachment was proximal to the anatomic PCL footprint and resulted in non-physiological length changes. Moving the femoral attachment locations of the PCL significantly affected length change patterns, whereas moving the tibia locations did not. The importance of anatomically positioned (i.e., distal to the isometric area) femoral PCL reconstruction locations to replicate physiological length changes is highlighted. These data can be used to optimize tunnel positioning in either single- or double-bundle and primary or revision PCL reconstruction cases.

Level of evidence: IV.

INTRODUCTION

Isolated posterior cruciate ligament (PCL) injuries have been treated non-operatively and often have good functional results.[38, 39, 49] However, the 5-year incidence of osteoarthritis in non-surgically treated PCL-deficient knees has been reported to be up to 80% and 50% on the medial femoral condyle and patella, respectively.[46] Unfortunately, PCL reconstruction has not been shown to prevent osteoarthritis, with the incidence of joint degeneration ranging from 15 to 60% after PCL reconstruction.[8, 30, 42, 45, 54, 57] A possible explanation for the persistent risk of joint degeneration might be the inability of contemporary PCL reconstruction techniques to restore normal joint biomechanics.[13, 28, 51, 53] Furthermore, high failure rates up to 30% have been indicated by several studies as soon as 4 years after primary PCL reconstruction.[15, 20, 29, 30, 33, 56, 57] Moreover, Noyes et al.[34] reported that approximately 33% of the failed PCL reconstructions had improper tunnel placement, with either too posterior femoral and/or too proximal tibial tunnels.

The PCL is particularly active in restraining posterior tibial translation at flexion angles beyond 60°, lateral tibial translation beyond 75°, and internal rotation beyond 90°.[19, 24, 25, 51] Some researchers have suggested that a double-bundle PCL reconstruction,[18, 55] an alternate graft orientation,[13] or a tibial osteotomy[12] may be needed to restore tibiofemoral kinematics to normal. In a recent meta-analysis,[22] it was shown that double-bundle PCL reconstruction was able to better restore posterior knee laxity when compared to single-bundle PCL reconstruction. However, no significant differences were found with respect to external rotation, varus rotation or coupled external rotation with posterior tibial force at any flexion angle.[22] Moreover, tunnel placement has an effect on the graft elongation patterns,[11, 14, 35, 44, 50] subsequent graft forces[6, 31, 35, 40, 41, 43] and knee kinematics.[6, 31, 40, 41, 43] Understanding the effect of adjusting tibiofemoral attachments on the length change patterns may help surgeons to optimize tunnel positioning and achieve physiological graft length changes during PCL reconstruction, reducing graft failure rates.

The purpose of this study was to elucidate the effects of various tibial and femoral attachment locations on the theoretical length changes and isometry of PCL grafts in the healthy knee during in vivo weightbearing motion. This information helps surgeons understand which areas are safe to put their tunnels and which areas should be avoided. It was hypothesized that non-anatomical attachments would be unable to reproduce anatomical graft length changes.

MATERIALS AND METHODS

Written consent was obtained from all patients prior to participating in this study. This study included 14 patients [10 men and 4 women; age 34 ± 13 years (mean ± standard deviation); height 176 ± 8 cm; body weight 82 ± 12 kg; active on a moderate athletic level before injury; no previous abnormal condition of the knee or lower limb] with diagnosed unilateral PCL injury and a healthy contralateral knee, confirmed by clinical examination and magnetic resonance imaging (MRI) performed by an orthopedic sports surgeon and musculoskeletal radiologist, respectively. The patients had no previous injuries, surgery or abnormalities of the contralateral knee or lower extremity. The average delay between injury and testing was 22 months. For this study specifically, the healthy contralateral knees were investigated. These patients were included in the previous studies of the tibiofemoral kinematics in PCL-deficient knees,[25] tibiofemoral cartilage deformation in PCL-deficient knees,[51] and posterolateral structures of the PCL-deficient knee.[21]

The MRI and dual fluoroscopic imaging techniques for the measurement of ligament kinematics have been described in detail previously.[26] MRI scans of the knee joint were obtained in the sagittal and coronal planes using a 3-Tesla MRI scanner (Siemens, Erlangen, Germany) with a double-echo water-excitation sequence (thickness 1 mm; resolution of 512 × 512 pixels).[9] The images were imported into solid modeling software (Rhinoceros; Robert McNeel and Associates, Seattle, WA) to construct three-dimensional (3D) surface models of the tibia, fibula and femur. Then, the knee of each patient was simultaneously imaged using two fluoroscopes as the patient performed a quasi-static lunge at approximately 0°, 30°, 60°, 75°, 90°, 105°, and 120° of knee flexion. Finally, the 3D-knee models of each patient were imported into the same software, and independently manipulated in 6-degrees-of-freedom inside the software until the projections of the model matched the outlines of the fluoroscopic images. When the projections matched the outlines of the images taken during in vivo knee flexion, the model reproduced the in vivo position of the knee. This system has a reported error of < 0.1 mm and 0.3° in measuring tibiofemoral joint translations and rotations, respectively.[9, 26, 27]

To determine the in vivo end-to-end distances of 3D-wrapping paths (i.e., theoretical grafts) during motion, the anatomic tibial PCL footprint was determined based on the sagittal and coronal plane MR images with guidance of anatomical descriptions.[3, 16, 36, 47] The PCL footprint was directly mapped onto the MRI-based 3D-tibia model. Since it is not possible to clearly distinguish the anterolateral bundle (ALB) and posteromedial bundle (PMB) on the MRI images, the PCL footprint was divided into an ALB and PMB portion guided by anatomic studies.[3, 16, 36, 47] The geometrical centers of the PCL, ALB and PMB were determined and used as three distinct tibial attachment locations (Fig. 1a).

Identical to the methods of previous researchers describing the PCL anatomy.[16, 36] The quadrant method as developed by Bernard et al.[5] was applied to the 3D-models to overcome the geometric variations between knees. The most anterior edge of the femoral notch roof was chosen as the reference for the grid alignment (line h). Using the medial view, 165 points were projected onto the lateral aspect of the medial femoral condyle (Fig. 1b). The attachment locations for the anatomic ALB, single-bundle PCL reconstruction (central), and PMB reconstruction were identified based upon previous anatomical descriptions.[16, 36]

The length changes for each theoretical graft were measured as a function of knee flexion using the in vivo 6-degrees-of-freedom knee joint kinematics. To create the path of a true graft, the direct line connecting the femoral and tibial attachments (i.e. direct end-to-end distance) was projected on the bony surfaces to create a curved line avoiding penetration of the connecting line through bone, i.e. a “wrapping path”. An optimization procedure was implemented to determine the projection angle to find the shortest 3D-wrapping path at each flexion angle of the knee. The length of the 3D-wrapping path was measured as the length of the theoretical graft. This technique was described in previous studies for measurements of ligament kinematics.[52] Following the methods by Taylor et al.[48] (measuring relative strain of the anterior cruciate ligament), the theoretical PCL graft length changes were normalized to a reference as follows: $Ln = L - L_0 / L_0 \times 100\%$; where Ln is normalized length change, L is graft length, and L0 is a reference length. Given that the PCL becomes taut in vivo at approximately 60° of flexion,[23, 24] the graft length at 60° was defined as the reference length for normalization.

A heat map was created to provide visual representation of the isometry distribution over the lateral aspect of the medial femoral condyle using the mean maximum percentage length change—mean minimum percentage length change of each theoretical tibiofemoral graft during quasi-static lunge. The tibiofemoral attachment combination yielding least length change was considered to be the most isometric graft. This study was approved by the institutional review board of the Massachusetts General Hospital (i.e., Partners Human Research Committees).

Statistical analysis

Changes in absolute lengths of the anatomic ALB, central, PMB and isometric grafts caused by knee flexion were examined using one-way analysis of variance (ANOVA) tests. The length changes were examined with one-way ANOVA tests to assess differences between grafts at each studied flexion angle from 60° to 120°. If significant, Tukey’s Honest Significant Difference (HSD) tests were used to assess for differences between the various pairs of bundles at each flexion angle (e.g., ALB 75° vs. central 75°, central 75° vs. PMB 75° etc.). One-way ANOVA tests were used to examine whether the tibial attachment affected length changes for each studied femoral attachment (e.g., ALB femur connected to AL, central or PM tibia), and Tukey’s HSD tests were employed when significant. Analyses were performed in R version 3.3.2 and P values less than 0.05 were considered significant.

RESULTS

Isometry and heat map

The most isometric attachment location was located proximal to the centers of the anatomic ALB, central and PMB attachments. Detailed information is shown in Fig. 2a, b and Video 1. Theoretical grafts with attachments distal to the isometric zone yielded increasing graft lengths with increasing flexion angles, whereas attachments proximal to the isometric zone resulted in decreased lengths with increasing flexion angles (Fig. 3). Moving attachments in the anterior–posterior direction had a less profound effect on the graft length changes compared to the proximal–distal direction (Fig. 3). The greater the distance of an attachment to the isometric zone, the greater the magnitude in length change as the knee was flexed.

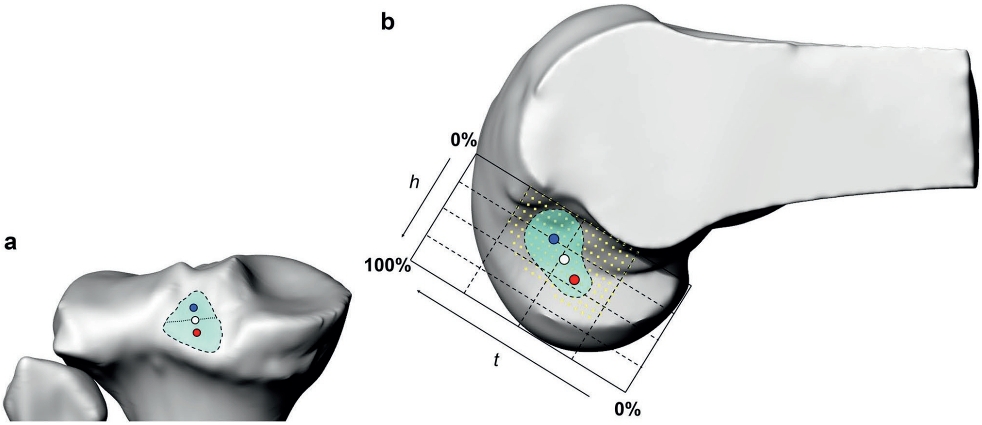

Fig. 1 Distribution of the tibial and (a) femoral (b) attachment points. Dashed lines show the outline of the anatomical posterior cruciate ligament footprint, the centers of the PCL (white dot), anterolateral (blue dot), and posteromedial bundles (red dot). The femoral grid as developed by Bernard et al.[5] was applied to the lateral aspect of the medial femoral condyle. A line extending along the intercondylar notch was used as a landmark for the anterior border of the grid (line h). Parallel to line h a line was drawn to the posterior edge of the medial condyle. The proximal and distal borders were formed by two lines perpendicular to line h originating from the proximal and distal bony borders of the medial femoral condyle (line t). Along the true mediolateral view 165 attachment points were projected to the lateral aspect of the medial condyle.

Femoral attachments

The ALB demonstrated a significant increase in length from 28.5 mm (95% confidence interval, 26.6–30.4 mm) at 0° to a maximum of 37.2 mm (35.8–38.5 mm) at 120° of flexion ($P < 0.001$). The central PCL graft significantly increased in length from 31.4 mm (28.1–32.3 mm) at 0° to 36.4 mm (35.0–37.9 mm) at 120° ($P < 0.001$). The PMB significantly decreased in length between 0–60° of flexion from 34.0 mm (32.1–35.9 mm) at 0° to its minimum length of 31.5 mm (29.5–33.6 mm) at 60° ($P < 0.001$); beyond 60° the bundle significantly increased in length to its maximum length of 35.4 mm (33.9–36.9 mm) ($P < 0.001$). The isometric graft had a length of 34.8 mm (32.7–36.9 mm) at 0° of flexion and did not significantly change during the quasi-static lunge (n.s.) (Fig. 4).

No significant differences in normalized length changes were found between either the ALB or PMB and the central PCL graft at 60°, 75°, 90°, 105° and 120° of flexion (n.s. for all). Similarly, no significant differences in normalized length changes between the ALB and PMB were found ≥ 60° of flexion (n.s.) (Fig. 5). The isometric graft was associated with significantly smaller length changes compared to the anatomic ALB, central graft and PMB at all flexion angles ($P < 0.001$ for all comparisons).

Tibial attachments

Moving the tibial attachment location (i.e., AL, central or PM attachment) had no significant effect on the normalized length changes of the ALB, central graft and PMB (n.s. for all).

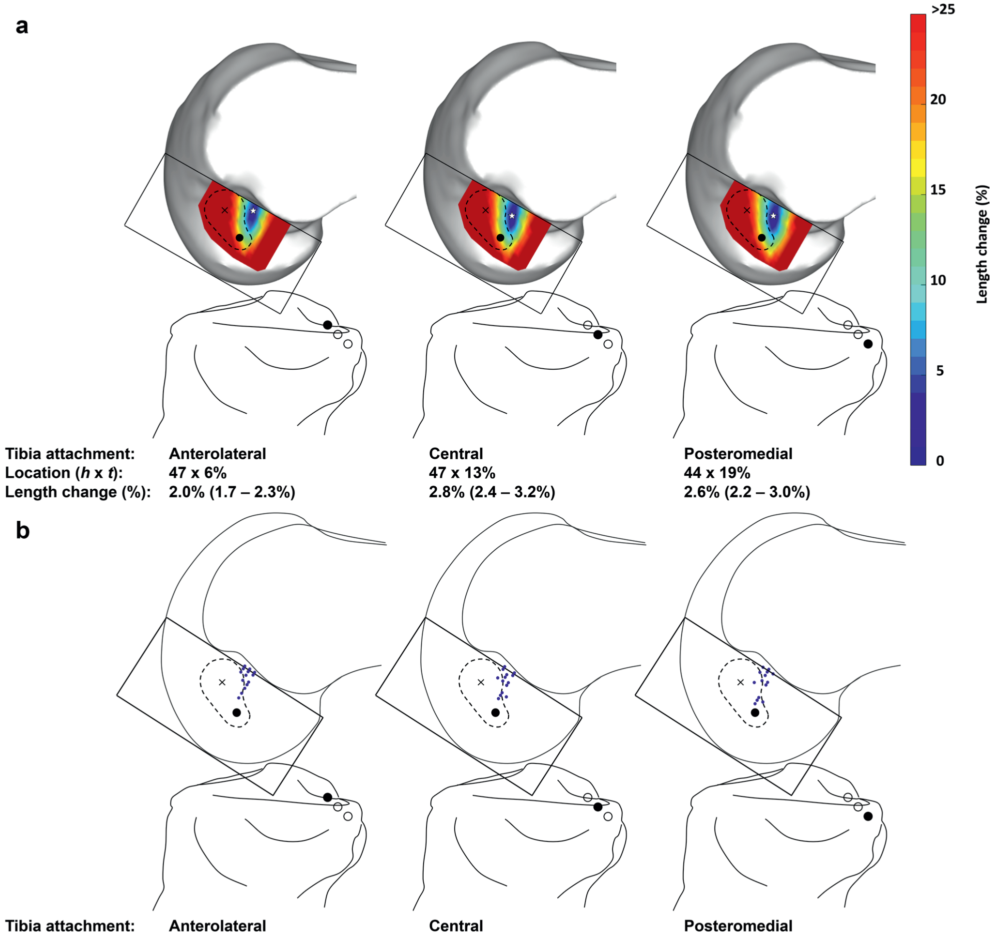

Fig. 2 Intact knees. (**A**) The "heat map" illustrates the isometry distribution (mean maximum % length change − minimum % length change) over the lateral aspect of the medial femoral condyle for single curves, wrapped around the bony contours when connected to anterolateral, central and posteromedial tibial attachment. The darkest blue area on the femur shows near isometric attachment area, while red areas highlight areas with a high degree of isometry. White star represents the most isometric attachment (values are: mean, 95% confidence interval). Dashed lines show the outline of the anatomical posterior cruciate ligament footprint. The black x on the femur shows the center of the anterolateral bundle; the black dot shows the center posteromedial bundle. (**B**) Distribution of the most isometric attachment location per patient.

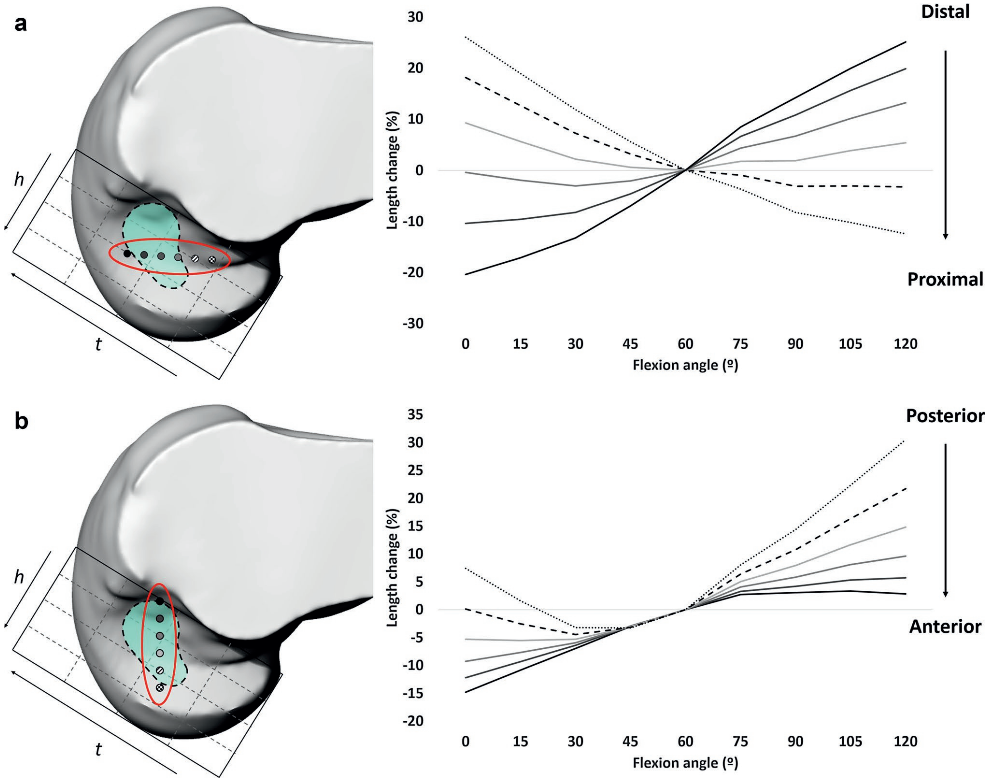

Fig. 3 On the left, a lateral view of a 3D femur model in 90° of flexion with several attachment points illustrated when moving along proximal–distal direction (**A**) or along the anterior–posterior direction (**B**). The normalized length changes for the illustrated attachments, when connected to the central tibial attachment are shown by the line graphs on the right. Distal attachments increased in length with increasing flexion angles, whereas proximal attachments decreased in length with increasing flexion angles. The greater the distance of a femoral attachment to the isometric zone, the greater the percentage length change as the knee flexes. When moving the attachments along the anterior–posterior direction, the length changes had a more similar pattern with greater magnitudes for more posterior attachments. Similar trends were found for the anterolateral and posteromedial tibial attachment.

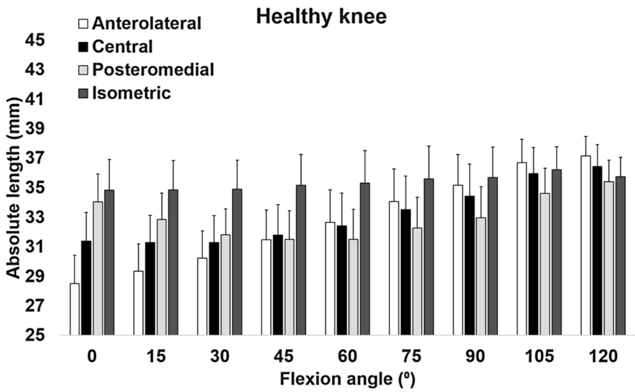

Fig. 4 Absolute length by knee flexion angle during the quasi-static lunge in intact knees, for theoretical grafts at the anatomic anterolateral, central, posteromedial and most isometric tunnel positions. Graft length increased with increasing flexion angles for the anterolateral bundle; the central graft was near isometric between 0° and 30° of flexion and increased in length thereafter; the posteromedial bundle decreased in length between 0° and 60° and increased to its maximum length at 120° of flexion; the theoretical isometric graft had about the same length during the quasi-static lunge. Values are shown as mean and 95% confidence interval.

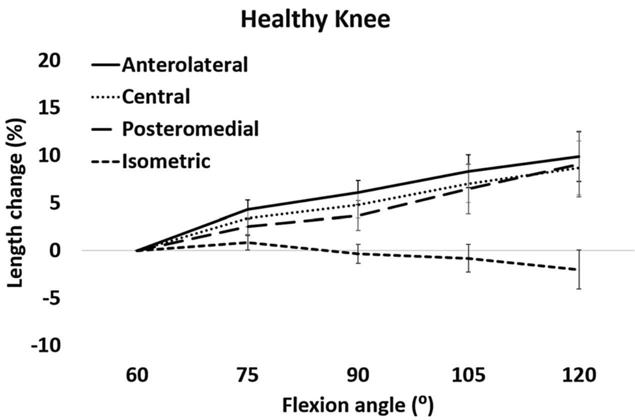

Fig. 5 Normalized length changes between 60° and 120° of flexion for theoretical grafts at the anatomic anterolateral, central, posteromedial and most isometric attachments. No significant differences in normalized length changes were found for either the anterolateral or posteromedial bundle to the central graft and the anterolateral and posteromedial bundle at 60°, 75°, 90°, 105° and 120° of flexion (n.s. for all). The most isometric graft was significantly different compared to all anatomic grafts at all flexion angles beyond 60° ($P < 0.001$ for all). Excursion represents normalized of the graft relative to its 60° of flexion length (zero point). Values are shown as mean and 95% confidence interval.

DISCUSSION

The most important finding of this study was that too proximal (i.e., non-anatomical) femoral attachments are unable to replicate anatomical graft length changes. The most isometric femoral attachment was located proximal to the anatomic ALB, central and PMB attachments and resulted in significantly smaller length changes. Thus, distal femoral locations (i.e., anatomical) may be essential to replicate anatomic graft behavior. In addition, moving the theoretical grafts in the proximal–distal direction greatly affected the length change patterns. Grafts proximal to the isometric zone decreased in length with increasing flexion whereas attachments distal to the isometric zone increased in length with increasing flexion, with greater magnitudes in length changes for more proximal or distal attachments. The PCL tibial attachment location had only a minor, non-significant effect on length change patterns for the anatomic grafts. Therefore, surgical errors on the femur in the anterior–posterior direction would be more forgiving than in the proximal–distal direction. More specifically, when performing PCL surgery, proximal femoral tunnel positioning should be avoided and distal tunnel locations (close to the articular cartilage surface) may be preferred.

Analogous to the advances made in anterior cruciate ligament reconstruction, PCL reconstruction has evolved from pursuing an isometric reconstruction[41] (aiming to prevent graft overload as a result of the excessive length changes during movement) to pursuing an anatomic reconstruction (aiming to reproduce native biomechanics).[17–19, 55] This idea is corroborated by the results of this study: only a was found to overlap the most isometric femoral zone which was located predominantly proximal to the anatomical PCL footprint. Similarly, Sidles et al.[44] found a tightly localized anterior–posterior distribution of the isometric area; however, it was slightly more distal then was found in this study. These differences may be explained not only by the kinematic difference between in vitro and in vivo loading of the knee but also by the wrapping effect of the tibiofemoral curves was not considered in their study. A graft with its femoral tunnel at the most isometric area would result in non-anatomic PCL reconstruction and could lead to nonanatomic graft behavior. Specifically, an isometric graft would increase too much in length (i.e., tight) relative to the anatomic graft at lower flexion angles and too little increase in length (i.e., slack) at deeper flexion angles, resulting in abnormal knee kinematics as was found in the cadaveric work by Race and Amis.[41] Moreover, the anterior-proximal femoral location of an isometric graft would be associated with a longer effective graft length (thus lower stiffness [10]).

In line with previous in vitro studies, these results demonstrated that cross-matching the anatomic ALB, central graft and PMB to the different tibial attachments (AL, central, and PM attachment) had only a small, non-significant effect on the length change patterns.[4, 14, 44] Although, the tibial attachments varied most in anterior–posterior in this study direction, the cadaveric study by Markolf et al.[32] has shown that errors in the medial–lateral tibial tunnel positioning do not affect the anterior–posterior laxity of the reconstructed knee.

These results may help explain the clinical results of Mariani et al.[30], who were unable to correlate improper tibial tunnel placement with clinical outcomes. Thus, given the similar biomechanical patterns between tibial tunnel positioning, a surgeon may have greater flexibility in placing these tunnels while respecting the tibial PCL footprint.

Appropriate graft fixation angles are critical because length changes over 4–6% will result in permanent graft stretch.[1, 7] Recent cadaveric experiments found that graft fixation angles between 75° and 105° of flexion equally restored knee kinematics in single-bundle PCL reconstruction.[17, 18] For double-bundle PCL reconstruction, graft fixation angles were found to be most favorable at 90° and 0° of flexion for the ALB and PMB, respectively. The mean maximum lengths of the ALB, central graft and PMB were 37.2 mm, 36.4 mm, 35.4 mm, respectively; thus, an increase of 2.10 mm for the AMB, 2.06 mm for the central graft and 2.0 mm for the PMB would yield the theoretical maximum allowed length increase of 6% required to avoid permanent graft stretch.[1, 7] Based on these measurements, the ALB, central graft and PMB may be safely fixed at ≥ 90° of flexion, while the PMB could also be fixed < 25° of flexion. These results also build on the results of Kennedy et al.[18] who showed increased PMB graft forces when the PCL graft was fixed at 15° compared to 0° of flexion. Graft length changes from 15° to 120° (4.4%) are greater than from 0° to 120° (0.7%) and hence higher graft forces would be expected (Fig. 5). However, the 15° length change of 4.4% does not exceed the critical threshold of 6%. Thus, surgeons have the choice between a tighter graft that may better restrain excessive knee laxity by fixing at 15° or a potentially looser graft that may be less prone to excessive graft stretch by fixing at 0°.

PCL reconstruction failure rates (i.e., side-to-side difference of > 5 mm) up to 30% have been reported.[15, 20, 29, 30, 33, 56, 57] Few reports are available on the etiology of failed PCL reconstructions. Noyes et al.[34] reported as most common causes for PCL failures unaddressed posterolateral corner injuries (44%) and incorrect tunnel placement (33%). All abnormal tibial tunnels were placed too proximal and abnormal femoral graft placement were too posterior.[34] Based upon the experience of the authors, albeit anecdotal, failed PCL reconstruction caused by abnormal tunnel positioning had too proximal femoral and far too anterior tibial tunnels (beyond the anatomical PCL footprint causing iatrogenic meniscal root tears). When performing PCL surgery, these data suggest that proximal femoral attachments yield non-anatomical graft length changes and should be avoided, whereas distal femoral attachments (closer to the articular cartilage surface) may be preferred.

Finally, no significant differences in normalized length changes were found between either the ALB or PMB and the central graft, nor between the ALB and PMB at ≥ 60° of flexion. These results suggest a similar function of the anatomic grafts at deeper flexion angles, further highlighting the codominant function of the PCL bundles in vivo, as had been suggested by others.[2, 19, 37]

Limitations

There are some limitations to this study. One inherit limitation of using true in vivo data is that 165 reconstructions were not performed; if a graft were to be placed in a different location it would likely slightly alter the kinematics of the knee, and therefore, small changes in graft length changes would be expected. Next, data from only one activity, a quasi-static lunge, was used; kinematics may change with more strenuous activities. In this study, graft length changes were normalized to a reference length and cannot be directly related to true ligament strains because the reference lengths of the ligaments (zero-load length) are unknown. However, previously this measurement has been shown to be linearly related to the true strain.[48] Graft thickness was not considered in this study. Finally, the effect of the different attachment locations on the graft bending angle was beyond the scope of this study.

Conclusions

The most isometric attachment was proximal to the anatomic PCL footprint and resulted in non-physiological length changes. In moving the femoral attachment locations of the PCL significantly affected length change patterns, whereas moving the tibia locations did not. The importance of anatomically positioned (i.e., distal to the isometric area) femoral PCL reconstruction locations to replicate physiological length changes is highlighted. These data can be used to optimize tunnel positioning in either single- or double-bundle and primary or revision PCL reconstruction cases.

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Chapter 5

In vivo anterolateral ligament length change in the healthy knee during functional activities – A combined magnetic resonance and dual fluoroscopic imaging analysis

Willem A. Kernkamp
Samuel K. Van de Velde
Ali Hosseini
Tsung-Yuan Tsai
Jing-Sheng Li
Ewoud R. A. van Arkel
Guoan Li

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ABSTRACT

Purpose: To measure the in vivo anterolateral ligament (ALL) length change in healthy knees during step-up and sit-to-stand motions.

Methods: Eighteen healthy knees were imaged using magnetic resonance and dual fluoroscopic imaging techniques during a step-up and sit-to-stand motion. The ALL length change was measured using the shortest three-dimensional wrapping path, with its femoral attachment located slightly anterior-distal (ALL-Claes) or posterior-proximal (ALL-Kennedy) to the fibular collateral ligament attachment. The ALL length measured from the extended knee position of the none-weight-bearing magnetic resonance scan was used as a reference to normalize the length change.

Results: During the step-up motion (approximately 55º flexion to full extension), both the ALL-Claes and ALL-Kennedy showed a significant decrease in length of 21.2% (95% confidence interval 18.0-24.4, $P < .001$) and 24.3% (20.6-28.1, $P < .001$), respectively. During the sit-to-stand motion (approximately 90º flexion to full extension), both the ALL-Claes and ALL-Kennedy showed a consistent, significant decrease in length of 35.2% (28.8-42.2, $P < .001$) and 39.2% (32.4-46.0, $P < .001$), respectively. From approximately 90º to 70º of flexion, a decrease in length of approximately 6% was seen; 70º of flexion to full extension resulted in an approximately 30% decrease in length.

Conclusions: The ALL was found to be a nonisometric structure during the step-up and sit-to-stand motion. The length of the ALL was approximately 35% longer at approximately 90º of knee flexion when compared with full extension and showed decreasing length at lower flexion angles. Similar ALL length change patterns were found with its femoral attachment located slightly anterior-distal or posterior-proximal to the fibular collateral ligament attachment.

Clinical relevance: These data suggest that, if performing anatomic ALL reconstruction, graft fixation may be performed beyond 70º flexion to reduce the chance of lateral compartment overconstraint. Anatomic ALL reconstruction may affect the knee kinematics more in high flexion than at low flexion angles.

INTRODUCTION

In recent cadaveric studies, discrepancy exists in the description of length change patterns of the anterolateral ligament (ALL) during knee flexion. This information is important when considering optimal graft fixation during ALL reconstruction.[1] Some researchers have found the ALL to be close to isometric between 0º and 60º of knee flexion angles and decrease in length from 60º to 90º of flexion.[14] These findings are directly at odds with findings by others who found the ALL to be nonisometric and gradually increase in length during 0º to 90º of flexion; its greatest length increase was noticed from 60º to 90º of flexion.[3] Similar nonisometric behavior was found in another independent study group.[32]

Possible explanations for the aforementioned differences in length change patterns might be the variability of the femoral attachment of the ALL used for ALL measurement in the cadaveric studies. The femoral insertion of the ALL has been described either together with the fibular collateral ligament (FCL),[2, 24] anterior-distal to the FCL,[1, 8, 32] posterior[6] or more posterior-proximal to the FCL.[4, 13, 19] Minor shifts in position around the rotational axis of the femur would result in contrary ligament kinematic patterns.[22] Another explanation might be the high dependence of the tibiofemoral biomechanics on the muscle loading conditions and subsequent length change patterns of the knee during in vitro testing. Even the most advanced in vitro experiments are limited by the difficulty in simulating the complex physiological loading conditions that occur during weight-bearing knee flexion.[29] Therefore, care should be taken when extrapolating the biomechanical behavior of the ALL that were measured during variable loading conditions in the in vitro setting to the length change patterns that would be seen in the healthy knee during in vivo weight-bearing flexion.

Therefore, in this study, we aimed to quantify the length change of the ALL in healthy subjects during dynamic in vivo functional activities, namely step-up and sit-to-stand weight-bearing motions of the knee to evaluate its isometric behavior. We hypothesized that during the dynamic functional activities, the ALL of the healthy knee would show nonisometric behavior with greater length at higher flexion angles.

METHODOLOGY

Patient Selection

This study was approved by our institutional review board. All subjects meeting the inclusion and exclusion criteria were enrolled from our institutional broadcast e-mail announcements. The inclusion criteria for this study were an age of 18 to 60 years, and the ability to perform daily activities independently without any assistance device and without taking pain medication. The exclusion criteria were knee pain, previous knee injury, and previous surgery to the lower limb. The magnetic resonance (MR) imaging scan of the knee of each subject was assessed for potential meniscal tears, chondral defects, and ligamentous injuries; if present, the subject was excluded for further analyses. Written consent was obtained from each subject. All subjects were tested between November 2008 and April 2010 to study the normal in vivo knee kinematics during dynamic functional activities. To address the research aim of the current study, the knees were analyzed to investigate the change in length of the anatomic ALL.

Imaging Procedure

The MR and dual fluoroscopic imaging techniques for the measurement of ligament kinematics have been described in detail previously.[15,18] The healthy knee was imaged with an MR scanner to create 3-dimensional (3D) meshed models of the knees, using a protocol established in our laboratory.[3] MR imaging was used to scan the knee joint in the sagittal plane using a 3-Tesla MR imaging scanner (MAGNETOM Trio; Siemens, Malvern, PA) with a double-echo water-excitation sequence (thickness 1 mm; resolution of 512 × 512 pixels). The images were then imported into solid modeling software (Rhinoceros; Robert McNeel and Associates, Seattle, WA) to construct 3D surface mesh models of the tibia, fibula, and femur. The attachment sites of the FCL were identified as previously described and included in the 3D knee model.[28] On these anatomical knee models the attachment sites of the ALL were presented as points. The femoral attachment sites of the ALL were positioned based on both the description by (1) Claes et al.,[1] that is, slightly anterior-distal with respect to the attachment of the FCL (ALL-Claes), and the description by (2) Kennedy et al.,[13] that is, posterior-proximal of the FCL origin (ALL-Kennedy). The tibial attachment site of the ALL was positioned midway between the center of Gerdy's tubercle and the anterior margin of the fibular head.[1, 13]

After the MR imaging-based computer models were constructed, the knee of each subject was simultaneously imaged using 2 fluoroscopes (BV Pulsera, Philips, the Netherlands) as the patient performed 2 dynamic motions: step-up and sit-to-stand motion. The motions were practiced multiple times before recording the finale motion that was used for analyses. Next, the fluoroscopic images were imported into solid modeling software and placed in planes based on the position of the fluoroscopes during imaging of the patient. Finally, the 3D MR imaging-based knee model of each subject was imported into the same software, viewed from the directions corresponding to the fluoroscopic setup used to acquire the images, and independently manipulated in 6 degrees of freedom inside the software until the projections of the model matched the outlines of the fluoroscopic images. When the projections matched the outlines of the images taken during in vivo knee flexion, the model reproduced the in vivo position of the knee. This system has an error of <0.1 mm and 0.3° in measuring tibiofemoral joint translations and rotations, respectively.[3, 17, 18]

Length Change Measurement of the ALL

The ALL length was measured as a function of knee flexion with several combinations of the tibiofemoral attachment points (Fig. 1). The direct line connecting the attachment sites was projected on the bony surfaces to create a curved ligament path to avoid penetration of the connecting line through bone. An optimization procedure was implemented for determination of the line projection angle to find the shortest 3D wrapping path of the ALL around the femoral condyles and the tibial plateau at each flexion angle of the knee. This technique has been described in previous studies for measurements of ligament kinematics.[15, 16, 22, 28] The length of this projected curve was measured as the length of the ligament. For each subject, the length change data were normalized to percentage length change by using the relaxed, non-weight-bearing MR imaging scan as a reference ([length – MR length]/MR length × 100%). The ALL is likely to be unloaded at this position and the length change is not representative of true ligament strain (i.e., change in length due to an applied force divided by the original length) but rather an increase in the distance between 2 anatomical sites.

Statistical Analysis

Changes in the length of the ALL (dependent variable), based on the descriptions by both Claes et al.[1] and Kennedy et al.,[13] caused by independent variables flexion of the knee and functional activities (step-up and sit-to-stand) were examined using a one-way analysis of variance with pairwise comparisons, having the Newman-Keuls post hoc procedure for multiple comparisons. Values are described as the mean percentage length change and 95% confidence intervals (CIs) (lower limit to upper limit). P values less than .05 were considered significant.

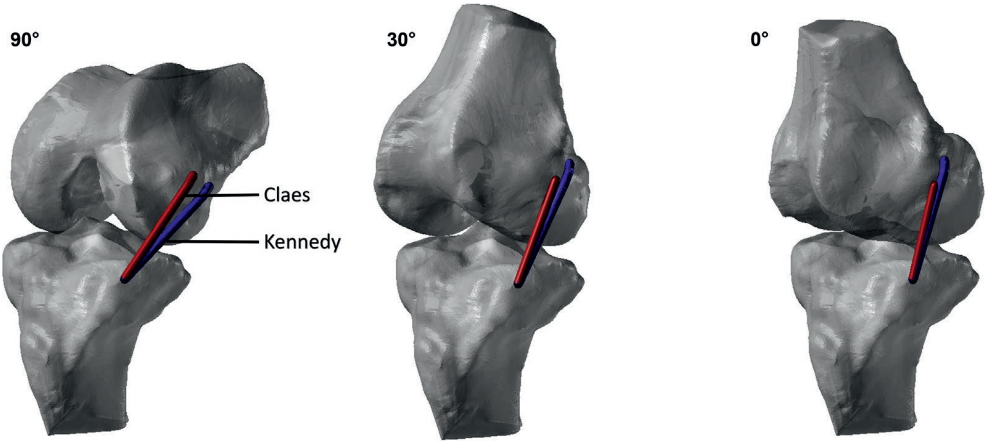

Fig. 1 Lateral view of a 3-dimensional knee model illustrating the anatomic anterolateral ligament with the femoral attachment anterior-distal (Claes et al.[1]) and posterior-proximal (Kennedy et al.[13]) with respect to the attachment of the fibular collateral ligament at 90º, 30º of knee flexion, and full extension during the sit-to-stand motion. The tibial insertion is midway between Gerdy's tubercle and the anterior margin of the fibula.

RESULTS

Eighteen healthy knees were included in this study (12 male, 6 female; age 35.4 years ± 10.9 years [mean ± standard deviation]; body height 175 ± 9 cm; body weight 83.3 ± 18.0 kg; body mass index 27 ± 3.5).

Reference Length

The mean length of the ALL-Claes (i.e., slightly anterior-distally to the FCL) as based on the non-weight-bearing MR imaging scan was 33.9 mm (95% confidence interval [CI], 32.5-35.4), and that of the ALL-Kennedy (i.e., posterior-proximal to the FCL) was 44.0 mm (95% CI, 41.8-46.2). The knees of the healthy subjects were slightly flexed during MR imaging, on average 2 ± 3.5°.

Step-Up Motion

The mean maximum flexion angle was 55 ± 4° (Fig. 2, Table 1). The ALL-Claes showed a consistent, significant decrease in length of 21.2% with decreasing flexion (95% CI, 18.0-24.4) over approximately 55° of flexion ($P < .001$) as compared with the non-weight-bearing MR reference length. The ALL-Kennedy also showed a consistent, significant decrease in length of 24.3% with decreasing flexion (95% CI, 20.6-28.1) over approximately 55° of flexion ($P < .001$) as compared with the MR reference length.

Sit-to-Stand Motion

The mean maximum observed flexion angle was 88 ± 10° (Fig. 3, Table 2). Both the ALL-Claes and ALL-Kennedy showed a consistent, significant decrease in length of 35.2% (95% CI, 28.2-42.2, $P < .001$) and 39.2% (95% CI, 32.4-46.0, $P < .001$), respectively, over approximately 90° of flexion as compared with the MR reference length. Length change from approximately 90° to 70° of flexion accounted for 5.0% (95% CI, 3.3-6.8, $P < .001$) and 6.0% (95% CI, 4.5-7.6, $P < .001$), respectively, whereas 70° of flexion to full extension resulted in 30.1% (95% CI, 23.6-36.6, $P < .001$) and 31.5% (95% CI, 25.4-37.7, $P < .001$). Likewise, from approximately 90° to 45° of flexion, the ALL showed a decrease in length of 13.1% (95% CI, 9.0- 17.2, $P < .001$) and 14.5% (95% CI, 11.0-17.9, $P < .001$); 45° of flexion to full extension resulted in an additional 22.0% (95% CI, 17.2-26.8, $P < .001$) and 23.1% (95% CI, 18.1-28.1, $P < .001$) decrease in length for the ALL-Claes and ALL-Kennedy, respectively.

Fig. 2

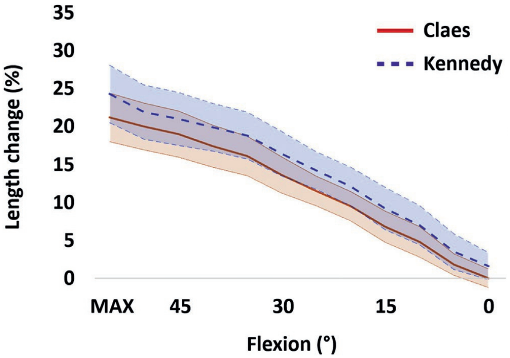

Fig. 3

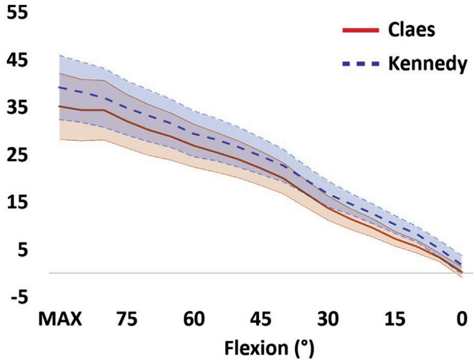

Fig. 2 and Fig. 3 The anterolateral ligament (ALL) length change (%) of intact knees as a function of the flexion (º) in 18 healthy subjects during the step-up (left) and sit-to-stand motion (right), the mean maximum flexion angle was (MAX) was 55 ± 4º and 88 ± 10º respectively. The red solid line depicts the femoral attachment of ALL-Claes, and the blue dashed line depicts the femoral attachment of the ALL-Kennedy. Values are mean and 95% confidence interval.

Table 1. Step-up motion

Knee flexion angle	ALL-Claes length change (%)	P-value†	Accumulated length change (%)	ALL-Kennedy length change (%)	P-value†	Accumulated length change (%)
Max - 50°	1.2 (0.3 to 2.2)	0.386	1.2	2.5 (1.6 to 3.4)	0.283	2.5
50° - 40°	2.6 (1.5 to 3.7)	<0.001	3.8	1.9 (2.9 to 1.0)	<0.001	4.4
40° - 30°	3.8 (2.8 to 4.9)	<0.001	7.7	3.5 (2.7 to 4.4)	<0.001	7.9
30° - 20°	4.1 (2.8 to 5.4)	<0.001	11.7	4.3 (3.0 to 5.7)	<0.001	12.3
20° - 10°	4.7 (3.6 to 5.8)	<0.001	16.5	5.1 (3.9 to 6.2)	<0.001	17.3
10° - 0°	4.7 (3.5 to 6.0)	<0.001	21.2	5.4 (3.9 to 6.9)	<0.001	22.7
0° - MRI	0.0 (-1.2 to 1.3)	0.987	21.2	1.6 (-0.1 to 3.4)	0.280	24.3

NOTE. Values are expressed as a percentage of the length as measured from non-weight-bearing MR reference length. Values are presented as mean (95% confidence interval), positive values indicate lengthening of the distance between attachment points, negative values indicate shortening.

† *P*-value: shows the statistical difference between each knee flexion angle range (e.g. MAX° - 50°).

Table 2. Sit-to-stand motion

Knee flexion angle	ALL-Claes length change (%)	P-value†	Accumulated length change (%)	ALL-Kennedy length change (%)	P-value†	Accumulated length change (%)
MAX – 80°	0.8 (-0.3 to 2.0)	0.024	0.8	2.2 (1.3 to 3.1)	0.007	2.2
80° – 70°	4.2 (3.0 to 5.4)	<0.001	5.0	3.9 (2.7 to 5.0)	<0.001	6.0
70° – 60°	3.2 (2.0 to 4.5)	<0.001	8.3	3.7 (2.5 to 5.0)	<0.001	9.8
60° – 50°	2.9 (1.8 to 4.1)	<0.001	11.2	2.8 (1.6 to 4.0)	<0.001	12.6
50° – 40°	4.0 (2.9 to 5.1)	<0.001	15.2	3.9 (2.9 to 5.0)	<0.001	16.5
40° – 30°	6.3 (5.1 to 7.5)	<0.001	21.5	6.0 (4.7 to 7.2)	<0.001	22.5
30° – 20°	4.2 (2.5 to 5.8)	<0.001	25.7	4.2 (2.2 to 6.2)	<0.001	26.7
20° – 10°	4.0 (2.4 to 5.7)	<0.001	29.7	4.4 (2.9 to 6.0)	<0.001	31.1
10° – 0°	5.4 (4.5 to 6.2)	<0.001	35.1	6.5 (5.3 to 7.7)	<0.001	37.6
0° – MR scan	0.1 (0.1 to 0.2)	0.826	35.2	1.6 (-0.4 to 3.6)	0.337	39.2
Knee flexion angle						
MAX – 70°	5.0 (3.3 to 6.8)	<0.001	5.0	6.0 (4.5 to 7.6)	<0.001	6.0
70° – 0°	30.1 (23.6 to 36.6)	<0.001	35.1	31.5 (25.4 to 37.7)	<0.001	37.6
MAX – 45°	13.1 (9.0 to 17.2)	<0.001	13.1	14.5 (11.0 to 17.9)	<0.001	14.5
45° – 0°	22.0 (17.2 to 26.8)	<0.001	35.1	23.1 (18.1 to 28.1)	<0.001	37.6

NOTE. Values are expressed as a percentage of the length as measured from non-weight-bearing MR reference length. Values are presented as mean (95% confidence interval), positive values indicate lengthening of the distance between attachment points, negative values indicate shortening.

† *P*-value: shows the statistical difference between each knee flexion angle range (e.g. MAX° - 80°).

DISCUSSION

The principal findings of this study show that both the ALL-Claes and ALL-Kennedy consistently and significantly decreased in length from approximately 90º of flexion to full extension, as is in agreement with our hypothesis. Similar nonisometric length change patterns were found during the step-up and sit-to-stand motion. The ALL length decreased approximately 22% during the step-up motion (approximately 55º of flexion to full extension) and 35% for the sit-to-stand motion (approximately 90º of flexion to full extension). An approximately 6% decrease in length was seen between 90º and 70º of flexion, and a 30% decrease in length was seen between 70º of flexion and full extension.

The nonisometric pattern of the ALL is in agreement with previous in vitro studies,[7, 14, 32] and a comparable length change of the ALL to our previous measurements during a quasi-static lunge was observed.[30] Helito et al.[7] found that the ALL increased in length from full extension to 90º, with a greater length increase from 60º to 90º than from 0º to 60º. These findings are in agreement with the study by Zens et al.,[32] who found the ALL to be a nonisometric structure that increased in length with increasing knee flexion. However, these results are in contrast with the findings described by Dodds et al.[4] In their study, they found the ALL to be near isometric between 0º and 60º of flexion, and the ALL to decrease in length between 60º and approximately 90º of flexion. These differences in length change may be explained due to the different techniques for measurement of the ALL length change that have been used in the cadaveric setting, for example, forced neutral tibial flexion,[3] unconstrained passive flexion,[7] fixed knee flexion angle at 30º,[19] with[4, 14] or without muscle loading conditions and with[4, 19] or without the use of forced internal rotation. Various ways to calculate the ALL length were used, such as a linear variable displacement transducer technique,[4, 14] and measurement based on a highly elastic capacitive polydimethylsiloxane strain gauge technique.[33] In the study by Helito et al.,[7] the ALL insertion sites were marked with metallic spheres and the distance between the 2 spheres was measured; no muscle tensioning was used and tibial rotation was controlled during flexion. Hereby, no native knee joint motion was simulated and the wrapping effect of the ALL was unaddressed.

Most recently, Imbert et al.[10] reported on the length change of 3 different ALL descriptions. The attachment sites anterior-distal to, and at the center of the lateral femoral epicondyle showed increasing length with increasing flexion, similar to the current study findings. The posterior-proximal point in their study was found to decrease in length with increasing flexion; no such length decrease was found in the current study. However, this may be explained due to the apparent difference in posterior-proximal descriptions: 7-7 mm (Imbert et al.[10]) versus approximately 3-3 mm (Kennedy et al.[13]). This could suggest that a more posterior-proximal location changes the length change pattern drastically.

Previous studies have shown that anterolateral extraarticular injuries accompanying anterior cruciate ligament (ACL) tears are frequently seen, and can attribute to the different instability patterns seen after ACL injury.[9, 27] Failure to recognize and manage concomitant injuries at the time of primary ACL reconstruction might result in persistent postoperative instability[11, 12] and put the knee joint at risk of secondary damage.[5, 26] Persistent postoperative instability as revealed by a residual pivot-shift test has been reported in 25% of the patients.[25] Monaco et al.[20] found that extraarticular reconstruction improved axial tibial rotation and stability during the pivot-shift test. Sonnery-Cottet et al.[25] found that combined ACL and extra-articular reconstruction can be an effective procedure in restoring knee stability without specific complications at a minimum follow-up of 2 years. Most recently, it was found that in the presence of anterolateral extraarticular injury, isolated ACL reconstruction was unable to restore internal rotation instability, whereas concomitant ALL reconstruction to the ACL reconstruction was able to significantly reduce internal rotation.[21] These results are promising and show the possible benefits of adding an extra-articular reconstruction to the ACL reconstruction to better restore knee stability.

In our recent pilot study,[31] we found that nonanatomic extra-articular reconstructions showed more biomechanically favorable length change patterns (i.e., smaller length change percentage) compared with the ALL reconstruction, therefore reducing the likelihood of graft stretch. However, only 1 functional activity a single quasi-static leg lunge was performed at discrete flexion angles. It is important to note that the anatomic ALL showed nonisometric behavior with increased length in deeper flexion angles. This means that more isometric, nonanatomic reconstructions potentially overconstrain the lateral compartment of the knee. In the present study, the considerable length change of the ALL as was previously measured during the quasi-static lunge was also seen during 2 fully dynamic activities. This finding further substantiates the probability that an anatomic ALL reconstruction might not be biomechanically favorable. It has been suggested that an increase of 6% in separation distance between insertion points could lead to permanent graft stretching.[23] The ALL changed approximately 6% in length between approximately 90º and 70º of flexion. These data therefore suggest that anatomic ALL reconstruction might have to be performed beyond 70º of knee flexion. Graft tensioning at lower flexion angles potentially results in excessive stretch of the graft and overconstraint of the lateral compartment of the knee. We believe that the findings of this study can contribute to the design of improved treatment protocols for anterolateral rotatory instability. Future studies should focus on the biomechanical changes of adding the anatomical ALL reconstruction to the ACL reconstruction and investigate possible nonanatomic extraarticular attachment points with similar length change patterns to the native biomechanics.

Limitations

The ALL length was measured as the shortest distance between the attachment sites on the 3D models projected to the bony surfaces. Baseline measure of the ALL length was defined as the relaxed, non-weight-bearing knee state as was seen in the MR imaging scan to which the percentage length change was calculated. Therefore, the ALL is likely to be unloaded at this position and the length change is not representative of true ligament strain (i.e., change in length due to an applied force divided by the original length) but rather an increase in the distance between 2 anatomical sites. We could not identify the ALL on the available 3-Tesla MR images; instead the detailed anatomic descriptions by Claes et al.[1] and Kennedy et al.[13] were used to determine the ALL attachment sites. No pivoting motion was performed, and thus, the effect of internal rotation demanding movements on the ALL length change could not be assessed.

Conclusions

The ALL was found to be a nonisometric structure during the step-up and sit-to-stand motion. The length of the ALL was approximately 35% longer at approximately 90º of knee flexion when compared with full extension and showed decreasing length at lower flexion angles. Similar ALL length change patterns were found with its femoral attachment located slightly anterior-distal or posterior-proximal to the FCL attachment.

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Chapter 6

An in vivo simulation of isometry of the anterolateral aspect of the healthy knee

Willem A. Kernkamp
Samuel K. Van de Velde
Tsung-Yuan Tsai
Ewoud R. A. van Arkel
Peter D. Asnis
Rob G. H. H. Nelissen
Robert F. LaPrade
Bertram Zarins
Guoan Li

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ABSTRACT

Background: To assess the isometry of theoretical lateral extra-articular reconstruction (LER), we evaluated theoretical grafts attached to various points on the lateral femoral condylar area and to either Gerdy's tubercle or the anatomic attachment site of the anterolateral ligament to the tibia.

Methods: In 18 subjects, healthy knees with no history of either injury or surgery involving the lower extremity were studied. The subjects performed a sit-to-stand motion (from approximately 90° of flexion to full extension), and each knee was studied using magnetic resonance and dual fluoroscopic imaging techniques. The 3-dimensional wrapping paths of each theoretical LER graft were measured. Grafts showing the least change in length during the sit-to-stand motion were considered to be the most isometric.

Results: The most isometric attachment site on the lateral femoral epicondyle to either of the studied tibial attachment sites was posterior-distal to the femoral attachment site of the fibular collateral ligament. The LER graft had a mean change in length of approximately 3%. Moving the femoral attachment site anteriorly resulted in increased length of the graft with increasing flexion; more posterior attachment sites resulted in decreased length with increasing flexion. Moving the attachment site in the proximal-distal direction had a less profound effect. Moving the tibial attachment site from Gerdy's tubercle to the tibial attachment site of the anterolateral ligament affected the overall isometric distribution on the lateral femoral epicondyle.

Conclusions: The most isometric attachment site on the femur for an LER would be posterior-distal to the femoral attachment site of the fibular collateral ligament. Different length changes for LER grafts were identified with respect to different femoral attachment sites. Desirable graft fixation locations for treating anterolateral rotatory instability were found posterior-proximal to the femoral fibular collateral ligament attachment.

Clinical Relevance: The present data could be used both in biomechanical studies and in clinical studies as guidelines for planning LER surgical procedures.

INTRODUCTION

Recent anatomic studies on the anterolateral aspect of the knee have created renewed interest in lateral extraarticular reconstruction (LER) of knees that have a torn anterior cruciate ligament (ACL).[13, 19, 28, 32, 35] Historically, the LER procedures were tried but were abandoned because of clinical failures.[30] However, an LER theoretically has appeal, as it is peripheral to the center of rotation of the knee and therefore has a lever arm to constrain excess rotatory laxity. The combined LER and intra-articular ACL reconstruction might therefore be able to better control excessive internal rotation of knees and reduce intra-articular forces on ACL grafts. However, there are few data on the biomechanical behavior of these extra-articular reconstructions, especially with respect to isometry.[8, 13, 16, 19, 21 22, 34] Information on this behavior is clinically relevant, enabling proper placement of the graft.

In the in vitro setting, different femoral attachment sites were believed to result in isometric or desirable patterns in length changes.[8, 13, 16, 21, 22, 34] These different results in the cadaveric experiments may be explained by the variety of methods used. Tibiofemoral biomechanics are highly dependent on the muscle-loading conditions of the knee. Length changes between points are highly sensitive to minor shifts in position around the rotational axis of the femur.[29] Even the most advanced in vitro experiments are limited by the difficulty in simulating the complex physiological loading conditions that occur during weight-bearing flexion of the knee.[39] Therefore, care should be taken when translating the in vitro biomechanical measurements during variable loading conditions to the results that would be seen in the knee during in vivo weight-bearing motion. Previously, we measured the theoretical length changes of the anterolateral ligament and 2 nonanatomic LERs during in vivo weight-bearing flexion.[18, 40] The anterolateral ligament was a nonisometric structure that showed a consistent length increase, up to 50%, from 0º to 90º of knee flexion. The nonanatomic LER showed length changes up to 15%. These results are promising and demonstrate the potential benefits of adding an LER to the intra-articular ACL reconstruction to better restore knee laxity and intra-articular graft forces. However, the most isometric point in vivo and most desirable length changes in vivo remain unknown and could improve current surgical techniques.

Therefore, the purpose of this study was to determine the in vivo isometry between various femoral attachment sites and 2 tibial attachment sites: Gerdy's tubercle and the anterolateral ligament attachment. This isometry was determined in healthy subjects during a dynamic sit-to-stand weight-bearing motion.

METHODOLOGY

Patient Selection

This study was approved by our institutional review board. Written consent was obtained from all subjects prior to participation in this study. In 18 subjects, healthy knees with no history of injury or surgery involving the lower extremity (12 male and 6 female subjects; mean age [and standard deviation], 35.4 ± 10.9 years; mean height, 175 ± 9 cm; mean weight, 83.3 ± 18.0 kg; and mean body mass index [BMI], 27 ± 3.5 kg/m2) were analyzed in the study. These subjects were included in our previous study on changes in the length of the anterolateral ligament.[18]

Imaging Procedures

The magnetic resonance imaging (MRI) and dual fluoroscopic imaging techniques for the measurement of ligament kinematics have been described in detail previously.[20, 38] MRI scans of the knee joints were performed in the sagittal plane using a 3-T MRI scanner (MAGNETOM Trio; Siemens Healthcare) with a double-echo water-excitation sequence (thickness, 1 mm; resolution, 512 × 512 pixels).[6] The images were then imported into solid-modeling software (Rhinoceros; RobertMcNeel & Associates) to construct 3-dimensional (3D) surface-mesh models of the tibia, fibula, and femur. The attachment sites of the fibular collateral ligament were identified as previously described and were included in the 3D model.[38]

After the MRI-based computer models were constructed, the knee of each subject was simultaneously imaged using 2 fluoroscopes (BV Pulsera; Philips) as the patient performed a sit-to-stand motion (from approximately 90º of flexion to full extension). 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 3D MRI-based knee model of each subject was imported into the software, viewed from the directions corresponding to the source of fluoroscopic radiation used to acquire the images, and independently manipulated in 6 degrees of freedom in the software until the projections of the model matched 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 errors of <0.1 mm and 0.3º in measuring tibiofemoral joint translations and rotations, respectively.[6, 23, 24]

Tibial and Femoral Attachment Sites

To determine the in vivo lengths of theoretical LERs during motion, various femoral attachment sites and 2 tibial attachment sites—the center of Gerdy's tubercle and the anterolateral ligament attachment (midway between Gerdy's tubercle and the anterior margin of the fibular head)—were used.[4, 17] To account for the geometric variations between knees, all 3D knee models were scaled using the anteroposterior borders of the lateral femoral condyle to the mean anteroposterior length (66.1 mm). Next, the right femoral models were mirrored to the left models with respect to the sagittal plane. Thereafter, the scaled and mirrored 3D models were aligned to find the best-fit position with respect to the lateral femoral condyle using a surface-to-surface registration method.[37] This process resulted in a mean average error of 0.9 ± 0.3 mm between models. An average femoral model was constructed from the scaled, mirrored, and aligned 3D models. The average model was then used to construct a transepicondylar axis (connecting the medial and lateral femoral epicondyles).[27] The direction of the transepicondylar axis of the average 3D femoral model was used to project 156 femoral attachment points to the individual scaled, mirrored, and aligned 3D models. The region of interest for the femoral points was determined by the data from previously published in vitro studies.[8, 19, 21, 34] Approximately the posterior half of the lateral femoral epicondyle was used to project the points to the 3D model with 2.5mm of spacing (Fig. 1). Once the femoral attachment points were determined on each femoral model, the scaled and mirrored 3D models with the projected attachment points were restored to the original coordinates for the measurement of individual graft lengths.

Length-Change Measurements

The length changes for each theoretical graft were measured as a function of knee flexion. The direct line connecting the femoral and tibial attachment points was projected on the osseous surfaces to create a curved line to avoid penetration of the connecting line through bone (a wrapping path). 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 measurement of ligament kinematics.[20, 29, 38] The length of the projected line (curved around the osseous surfaces) was measured as the length of the ligament. For each subject, the length-change data were normalized to percentage change by using the reference length of each tibiofemoral graft from the relaxed, non-weight-bearing MRI position (MR): (length – length MR)/length MR × 100%. A heat map was created to provide visual representation of the isometry distribution over the lateral femoral epicondyle by using the mean maximum percentage length change minus the mean minimum percentage length change of each theoretical tibiofemoral graft during the sit-to-stand motion.

Quadrant Method

A true lateral view of the femur was established at 90º of flexion. A 4 × 4 grid was applied to the lateral femoral epicondyle using a line extended along the posterior cortex of the distal femoral shaft and the posterior condylar offset line (PCOL). This technique has been used in other studies and was found to have intraobserver and interobserver reliabilities of 0.899 and 0.882, respectively.[14] Next, lines perpendicular to the PCOL were drawn to the proximal condylar cartilage border and the osseous femoral joint line, and 3 lines were drawn in between to create an evenly distributed grid (Fig. 2). Similar to the intraarticular quadrant method developed by Bernard et al.,[3] the current method used 4 distances, including condylar width perpendicular to the PCOL with the extended posterior cortex line as border (distance x), sagittal diameter along the PCOL (distance y), distance from the femoral attachment site to the anterior border along line x (distance Δx), and distance from the femoral attachment site to the proximal border along line y (distance Δy). Distances Δx and Δy were expressed as percentages of x and y.

Fig 1.

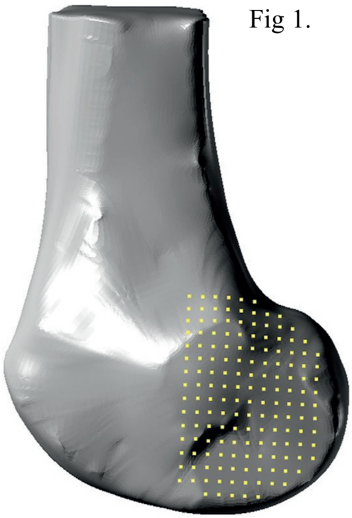

Fig. 2

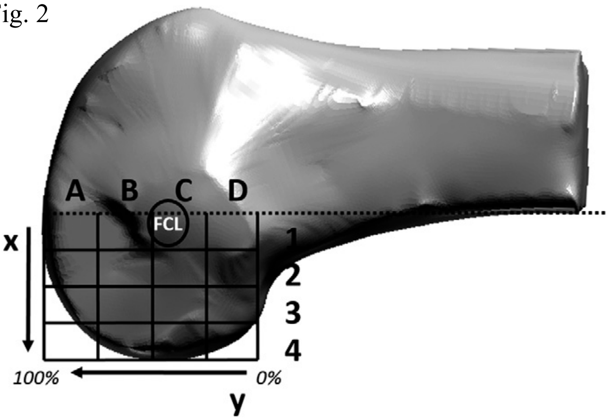

Fig. 1 Lateral view of a 3D femoral model showing the distribution of the femoral attachment sites (dots). Various femoral attachment sites were connected to either Gerdy's tubercle or the tibial attachment site of the anterolateral ligament (midway between Gerdy's tubercle and the fibular head). **Fig. 2** Lateral view of a 3D femoral model in 90º of flexion. A 4 × 4 grid was applied to the lateral femoral epicondyle. A line extended along the posterior cortex of the distal femoral shaft was used as a landmark for the anterior border of the grid, and the PCOL formed the posterior border. The proximal and distal borders were formed by the proximal condylar cartilage border and the osseous femoral joint line, respectively. Distal to proximal: A to D. Anterior to posterior: 1 to 4. Line x: maximum distance perpendicular to the PCOL to the posterior edge of the lateral condyle. Line y: maximum distance from the proximal condylar cartilage border to the osseous femoral joint line. FCL = fibular collateral ligament.

RESULTS

Isometric Point

The mean maximum observed flexion angle during the dynamic sit-to-stand motion was 88º ± 10º.

The most isometric femoral attachment site of the theoretical LER grafts that connected to Gerdy's tubercle was found to be posterior-distal to the femoral fibular collateral ligament attachment site; on average, it was 57% distal and 39% posterior, with a mean length change of 2.2% (95% confidence interval [CI], 1.8% to 2.6%).

When the LER was connected to the anterolateral ligament attachment, the most isometric femoral attachment site was found to be slightly more proximal-anterior to the point described above; on average, it was 50% distal and 31% posterior, with a mean length change of 3.3% (95% CI, 2.9% to 3.7%) (Fig. 3).

Posterior to the femoral fibular collateral ligament attachment site, a zone in the proximal-distal direction (the blue area on the femoral condyle in Fig. 3) demonstrated the lowest percentage change in length during the sit-to-stand motion when connected to Gerdy's tubercle. When connected to the anterolateral ligament attachment, the most isometric zone had a slightly oblique direction from posterior-distal to proximal-anterior.

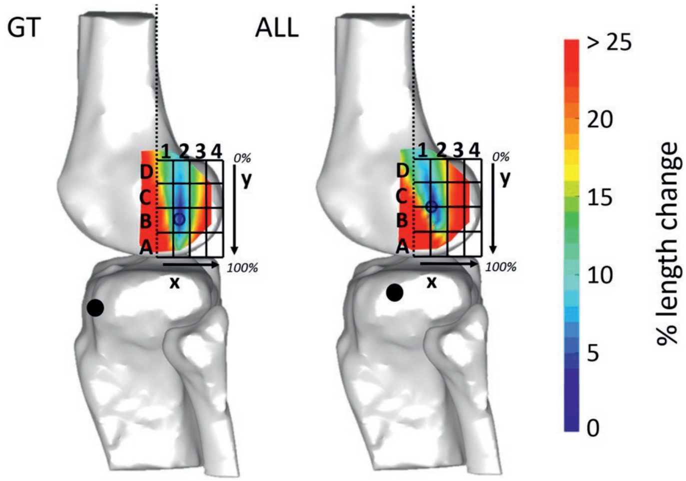

Fig. 3 Heat map illustrating the isometry distribution (mean maximum percentage length change – minimum percentage length change) over the lateral femoral epicondyle for single point-to-point curves when connected to Gerdy's tubercle (GT) or the tibial attachment site of the anterolateral ligament (ALL) during the dynamic sit-to-stand motion. The circle on the femur represents the most isometric attachment site (a 2.2% length change for Gerdy's tubercle and a 3.3% length change for the tibial attachment site of the ALL).

Femoral and Tibial Attachment Sites

Altering the femoral attachment site in the anterior-posterior direction affected the length changes, irrespective of the tibial attachment site (Fig. 4). The areas located anterior to the most isometric zone resulted in increased graft lengths with increased flexion angles; more posteriorly located areas resulted in decreased length with increased flexion. Moving the femoral attachment site in the proximal-distal direction had a less profound effect on the length changes (Fig. 5). Moving the tibial attachment site from Gerdy's tubercle to the anterolateral ligament attachment changed the overall isometric distribution on the lateral femoral epicondyle (Fig. 3). Comparable length changes could be found with respect to the most isometric zone (Fig. 6).

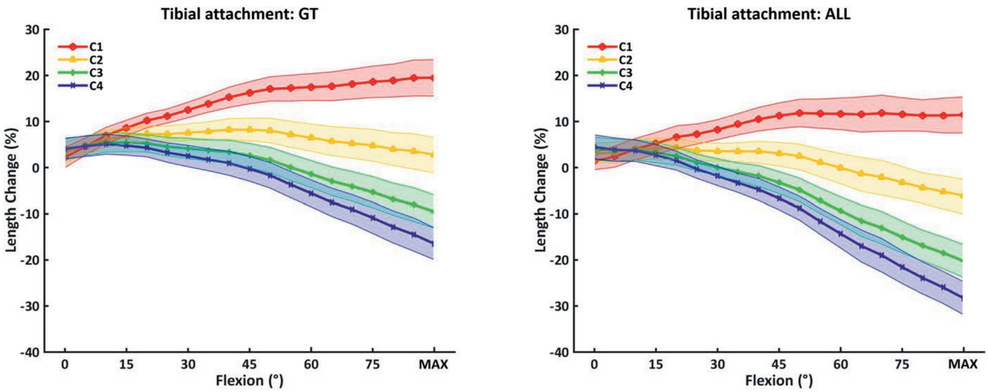

Fig. 4 Normalized length changes in percentage of area (C1 [anterior], C2 [middle anterior], C3 [middle posterior], and C4 [posterior]) during the dynamic sit-to-stand motion when connected to Gerdy's tubercle (GT; left) or the tibial attachment site of the anterolateral ligament (ALL; right). The mean maximum flexion angle (MAX) was $88^o \pm 10^o$. Mean values are shown, with the shaded area indicating the 95% CI.

Graft Length Changes

LER grafts that exhibited the least change and a tight (long) graft during early knee flexion (from full extension to 45º) and a slack (short) state during deep knee flexion (from 45º to approximately 90º) were found in the posterior-proximal area: C3-4 and D3-4 for Gerdy's tubercle, and C2-3 and D3-4 (using the quadrant method) for the anterolateral ligament attachment (Fig. 6).

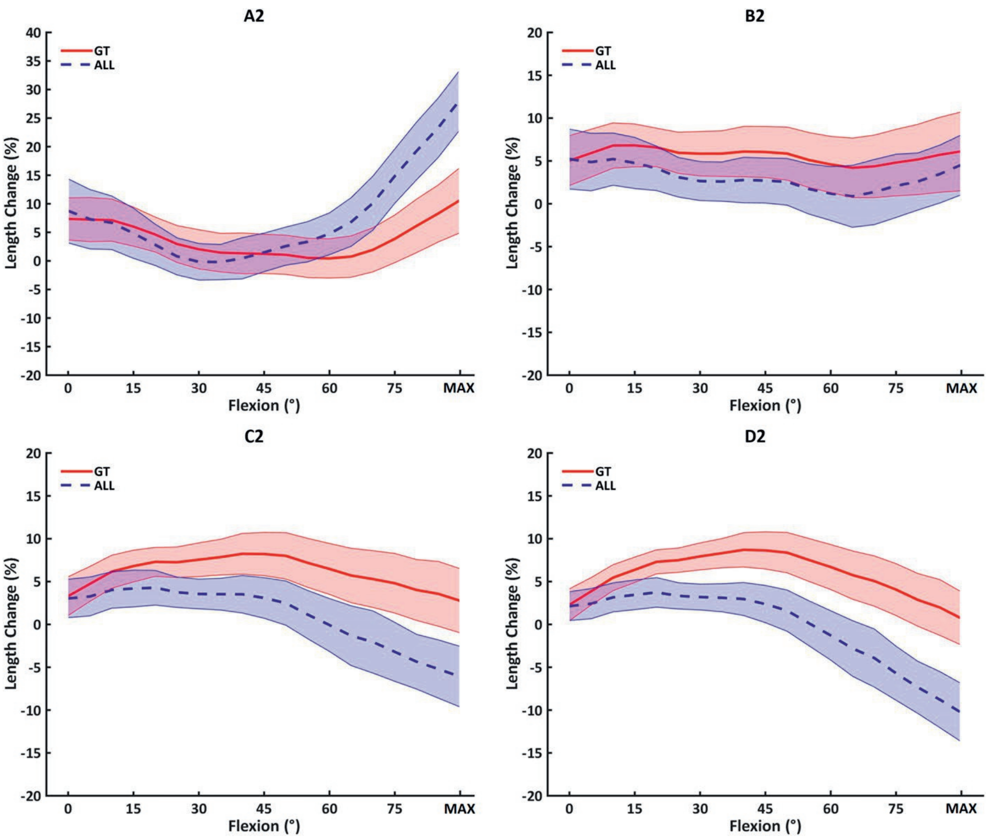

Fig. 5 Normalized length changes in percentage for healthy knees in the proximal-distal direction (A2 to D2) during the dynamic sit-to-stand motion. The mean maximum flexion angle (MAX) was 88º ± 10º. The dashed and solid lines represent the theoretical tibiofemoral grafts connected to Gerdy's tubercle (GT) and the tibial attachment site of the anterolateral ligament (ALL), respectively. Mean values are shown, with the shaded area indicating the 95% CI.

DISCUSSION

The most important finding in this study was that the most isometric location for a theoretical LER graft on the lateral femoral epicondyle was posterior-distal to the femoral fibular collateral ligament attachment site. This was true for both of the tibial attachment sites studied, Gerdy's tubercle and the attachment site of the anterolateral ligament. A graft in this position underwent a length change of approximately 3% during approximately 90º of active knee flexion. A zone, mainly in the proximal-distal direction, was found to show the lowest percentage length change during motion from full extension to approximately 90º of flexion. On the basis of Fig. 3, one might conclude that the most isometric femoral attachment site is in the region of the popliteus sulcus, and thus, an LER at this point might interfere with the popliteus tendon. Desirable length changes for LER, in which a tight graft in early knee flexion and a slackened graft in deep flexion were observed, were located in the posterior-proximal area of the lateral femoral epicondyle. Moving the tibial attachment site changed the overall isometry distribution on the lateral femoral epicondyle.

Several cadaveric studies have been published on the isometry of extra-articular reconstructions attached to the lateral femoral condyle.[9, 13, 16, 19, 21, 22, 34] The findings of the current study are most consistent with those of the cadaveric studies by Draganich et al.[8] and Ankri et al.,[2] in which the most isometric point was posterior-distal with a mean length change of 2% to 6% and 4.3%, respectively. Similar to the authors of previous cadaveric studies,[2, 8] we found that the most isometric zone (demonstrating the least overall length change) was posterior to the fibular collateral ligament attachment site and ran mainly in

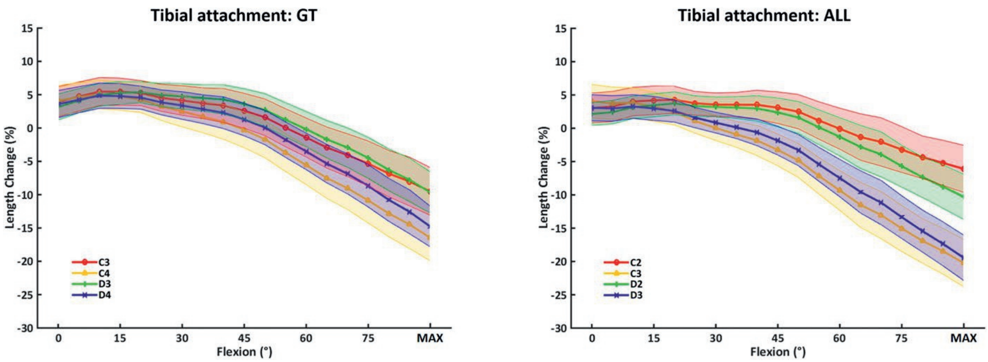

Fig. 6 Desirable changes in graft length (tightness in extension and lower flexion angles, and not limiting the range of motion in deeper flexion) could be found in the posterior-proximal area of the lateral femoral epicondyle. Similar patterns in graft-length change were found with respect to the most isometric zone (areas C3-4 and D3-4 when connected to Gerdy's tubercle [GT; left] as well as C2-3 and D2-3 for the tibial attachment site of the anterolateral ligament [ALL; right]). The mean maximum flexion angle (MAX) was 88º ± 10º. Mean values are shown, with the shaded area indicating the 95% CI.

the proximal-distal direction. Sidles et al.[34] found the most isometric attachment site to be directly posterior to the fibular collateral ligament and the most isometric zone to be more posterior. These differences may be explained not only by the kinematic difference between in vitro and in vivo loading of the knee but also by the different way that their data were normalized ([maximum length – minimum length]/[maximum length + minimum length]); also, the wrapping effect of the tibiofemoral curves was not considered in their study. In the present study, moving the femoral attachment sites in the anterior direction caused changes in which increased flexion resulted in increased length, whereas more posteriorly located points resulted in decreased length with increased flexion. This phenomenon is in agreement with findings of the most recent cadaveric studies by Imbert et al.[13] and Katakura et al.,[16] which measured the isometric characteristics and graft tension, respectively, of 3 different anterolateral ligament locations on the femur.

The rate of injury to the extra-articular structures of the knee at the time of the primary ACL tear has been found to be as high as 90%.[11, 36] It is thought that the combination of an intra-articular ACL tear with injury to anterolateral extraarticular structures might be responsible for the severe rotatory instabilities that can be seen in the clinic.[11, 26, 36] Unaddressed injury to secondary stabilizers may put the knee at risk for persistent postoperative rotatory instability[15] and consequently secondary injuries such as meniscal and chondral lesions, increased failure rates, and early cartilage degenerative changes. The combined LER and ACL reconstruction might be able to better restore anterolateral rotatory instability to normal in some patients,[41] and improve the tensile strength of the reconstruction, decreasing excessive loads through the ACL graft,[9, 10] potentially protecting the ACL graft during the healing phase[35] and subsequently reducing graft failure and recurrence rates.[12, 25]

A minimum degree of isometry reduces the likelihood of unwanted graft behavior, such as graft stretching, failure, and overconstraint of the lateral compartment.[1] In the normal knee, with increased knee flexion angles, internal tibial rotation also increases.[31] Thus, a certain degree of isometry is necessary to reduce undesirable graft behavior, but a true isometric reconstruction technique might overconstrain the knee during deeper flexion angles. Therefore, the ideal LER would provide internal rotatory constraint in lower flexion angles, extension, and slacken at increased flexion angles. Hence, the nonisometric behavior of the anterolateral ligament, with its increased length at increased flexion angles makes it unsuitable for reconstruction. This unsuitability was recently confirmed in the study by Schon et al.,[33] in which anterolateral ligament reconstruction overconstrained knee joint kinematics compared with the native knee at all fixation angles.

Functional length of the graft (determined by its proximal and distal fixation) is an important variable in any reconstruction. Stress-strain curves are characterized by a nonlinear toe region and a linear region. Long grafts have a greater elongation under the same load compared with short grafts for both nonlinear and linear regions; decreasing the length of a graft linearly increases its stiffness.[5] The fixation sites of the graft in LER determine the effective lengths of the graft and thus play an important role in the kinematic response of the knee.

The femoral attachment site directly affects the effective graft length and length changes that occur during knee motion. Moving the femoral attachment site in the anterior-posterior direction results in considerable length changes during motion, whereas alteration in proximal-distal direction has a less profound effect (Fig. 5). If one wanted to achieve a tight (long) graft in extension and a slack (short) graft in flexion, a femoral location posterior-proximal to the fibular collateral ligament attachment would be chosen; to achieve a tight graft in increased flexion, a more anterior location would be chosen (Fig. 4). Moving the tibial attachment site changes the effective graft length and changes the isometry distribution. In addition, the tibial attachment site affects the angle of the graft vector. A more anterior tibial attachment site (e.g., Gerdy's tubercle) holds a mechanical advantage over a posterior site (e.g., the anterolateral ligament attachment) for limiting internal rotation.

Limitations

There are several limitations to this study. Only data from healthy knees during 1 functional activity were used. Future research should also consider knees with a torn ACL and more demanding in vivo functional activities, such as lunging, walking, and running. No pivoting motion was performed in this study and, thus, the effect of rotational moments could not be assessed. Caution should be taken when translating the length changes as observed in this study to actual LER. No reconstruction was performed in the current study. Therefore, no actual restraint due to LER was present. Kinematics, and consequently length changes, could be altered if an LER had been performed. Finally, tunneling grafts deep to the fibular collateral ligament was not considered in this study.

Nevertheless, we believe that the findings of this study offer data that can be used to optimize LER techniques. Future studies should focus on the biomechanical effects of combining LER with ACL reconstruction and should investigate whether the most isometric graft, or a graft that is tight in extension and slack in flexion, would best restore laxity in a knee with a torn ACL.

Conclusions

In summary, the most isometric attachment site on the femur for an LER would be posterior-distal to the femoral attachment site of the fibular collateral ligament. Moving the femoral attachment site anteriorly resulted in increased length of theoretical LERs with increased flexion, whereas more posteriorly attachments resulted in decreased length with increased flexion angles. Desirable graft-fixation locations, stabilizing the knee at low flexion angles but not overconstraining the knee at high flexion, were found posterior-proximal to the femoral fibular collateral ligament attachment.

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Chapter 7

The medial patellofemoral ligament is a dynamic and anisometric structure – An in vivo study on length changes and isometry

Willem A. Kernkamp
Cong Wang
Changzhao Li
Hai Hu
Ewoud R.A. van Arkel
Rob G.H.H. Nelissen
Robert F. LaPrade
Samuel K. Van de Velde
Tsung-Yuan Tsai

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ABSTRACT

Background: Medial patellofemoral ligament (MPFL) reconstruction is associated with a high rate of complications, including recurrent instability and persistent knee pain. Technical errors are among the primary causes of these complications. Understanding the effect of adjusting patellofemoral attachments on length change patterns may help surgeons to optimize graft placement during MPFL reconstruction and to reduce graft failure rates.

Purpose: To determine the in vivo length changes of the MPFL during dynamic, weightbearing motion and to map the isometry of the 3-dimensional wrapping paths from various attachments on the medial femoral epicondyle to the patella.

Study Design: Descriptive laboratory study.

Methods: Fifteen healthy participants were studied with a combined computed tomography and biplane fluoroscopic imaging technique during a lunge motion (full extension to ~110° of flexion). On the medial femoral epicondyle, 185 attachments were projected, including the anatomic MPFL footprint, which was divided into 5 attachments (central, proximal, distal, posterior, and anterior). The patellar MPFL area was divided into 3 possible attachments (proximal, central, and distal). The length changes of the shortest 3-dimensional wrapping paths of the various patellofemoral combinations were subsequently measured and mapped.

Results: For the 3 patellar attachments, the most isometric attachment, with an approximate 4% length change, was located posterior and proximal to the anatomic femoral MPFL attachment, close to the adductor tubercle. Attachments proximal and anterior to the isometric area resulted in increasing lengths with increasing knee flexion, whereas distal and posterior attachments caused decreasing lengths with increasing knee flexion. The anatomic MPFL was tightest in extension, decreased in length until approximately 30° of flexion, and then stayed near isometric for the remainder of the motion. Changing both the femoral and patellar attachments significantly affected the length changes of the anatomic MPFL ($P < .001$ for both).

Conclusions: The most isometric location for MPFL reconstruction was posterior and proximal to the anatomic femoral MPFL attachment. The anatomic MPFL is a dynamic, anisometric structure that was tight in extension and early flexion and near isometric beyond 30° of flexion.

Clinical Relevance: Proximal and anterior MPFL tunnel positioning should be avoided, and the importance of anatomic MPFL reconstruction is underscored with the results found in this study.

INTRODUCTION

The medial patellofemoral ligament (MPFL) is the primary restraint to lateral patellar translation, contributing 50% to 60% of the total restraining force.[16,25,41,45] The MPFL is almost always ruptured during a lateral patellar dislocation.[24,38] Primary patellar dislocations may be treated nonoperatively; however, a redislocation is seen in up to 35% to 50% of patients,[5,9,12,14,15,36,79] which has been related to increased cartilage damage and the early onset of osteoarthritis.[75] Therefore, surgical reconstruction of the MPFL is indicated in patients with recurrent patellar dislocations.[47,77] Moreover, recently, some authors have described that in specific cases, MPFL reconstruction may be beneficial after primary patellar dislocations.[42,50] Although several studies have shown significant improvements in patient outcomes after MPFL reconstruction,[54,66,72] others have described high complication rates, in particular recurrent instability and persistent knee pain.[3-5,13,37,43,58,68] One of the primary causes of these complications is technical surgical errors, of which femoral tunnel malpositioning has been found to be one of the most common.[10,43,51,52,68]

Knowledge about the native anatomy and understanding its function are paramount in ligament reconstruction. The anatomy of the MPFL has been heavily debated, and recent publications have shown variability of its femoral attachment.[1,6,67] It has been shown that nonanatomic graft positioning can lead to decreased range of motion, knee pain, graft failure, tunnel widening, recurrent dislocations, and increased medial patellofemoral joint contact pressure, which has been postulated to cause early degenerative changes of the patellofemoral joint.[10, 13, 18, 19, 52, 56, 63, 64, 68, 71] Few studies have investigated the effects of patellofemoral attachment locations on graft length changes.[22,27,40,59,60,65,78] It has been found that modifying the femoral graft position, mainly in the proximal-distal direction, is more sensitive for graft length changes than altering the patellar position.[65] However, most of these studies were limited by using only a few patellofemoral attachments and the nature of their cadaveric, nonphysiological muscle-loading conditions.[30] Moreover, Kaiser et al.[29] recently underscored the importance of muscle-loading conditions on both tibiofemoral and patellofemoral joint kinematics. Therefore, it is difficult to extrapolate the biomechanical behavior of the MPFL that was measured in these studies to the length change patterns that would be seen during in vivo weightbearing knee flexion.

The purpose of this study was to assess the in vivo isometry and length change patterns of the MPFL in the healthy knee using various patellofemoral attachments. The hypothesis was that attachments outside the anatomic footprint would yield nonphysiological graft length changes (i.e., cannot replicate ‘‘normal’’ MPFL behavior).

METHODS

Participants

This study was approved by the Shanghai Jiao Tong University Institutional Review Board, and written consent was obtained from each participant before taking part in this study project. All participants were examined between June and July 2018. The inclusion criteria consisted of participants aged 18 to 45 years with the ability to perform daily activities independently without any assistance device and without taking pain medication. A standard knee examination was performed on the knee, and participants with increased laxity (as described by Brighton[23]) were excluded. Other exclusion criteria were a positive lateral patellar apprehension test finding, retropatellar tenderness or crepitation, joint effusion, recurrent or chronic knee pain, and either a history of injuries or surgery involving the lower limb. Fifteen healthy participants were included in this study (9 men, 6 women; mean age, 25.1 ± 5.2 years; mean height, 170 ± 10 cm; mean weight, 63.9 ± 11.9 kg; mean body mass index, 22.1 ± 2.7 kg/m2). The mean tibial tuberosity–trochlear groove distance was 13.3 ± 3.0 mm (range, 8.1-17.4 mm).

Imaging

The computed tomography (CT) and dual fluoroscopic imaging techniques for the measurement of ligament kinematics have been described in detail previously.[34,35] CT scans (SOMATOM Definition AS+; Siemens) of the knee joints ranging from approximately 30 cm proximal and distal to the joint line (thickness, 0.6 mm; resolution, 512 × 512 pixels) were obtained. The images were then imported into solid modeling software (3D Slicer, www.slicer.org[21]) to construct 3-dimensional (3D) surface models of the femur, patella, tibia, and fibula. Then, the knee of each participant was simultaneously imaged using 2 fluoroscopes (BV Pulsera; Philips) as the participant performed a lunge motion (full extension to ~110° of flexion). In addition to the lunge motion, the knee was imaged in its relaxed full extension position. Next, the fluoroscopic images were imported into MATLAB (R2018a; Math-Works) and placed in the imaging planes based on the projection geometry of the fluoroscopes during imaging of the participant. Finally, the CT-based knee model of each participant 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 6 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.

Patellofemoral Attachments

To determine in vivo the shortest 3D wrapping paths (i.e., theoretical grafts) during motion, various patellofemoral attachments were used. First, a true medial-lateral view of the femur was established. Second, to account for geometric variations between knees, the quadrant method, as described by Stephen et al.,[65] was applied to the femoral 3D models. The anterior and posterior borders of the quadrant were formed by lines parallel to the posterior femoral cortex at the anterior and posterior bony aspects of the medial femoral condyle (line t). The proximal and distal borders were formed by lines perpendicular to line t, proximally to the tip of line t, and distally at the bony cortex of the medial condyle (line h). The medial-lateral view was used to project 185 femoral attachment points to the medial aspect of the medial femoral condyle (Fig. 1). Based on the recent systematic review by Aframian et al.,[1] an area of interest was created, to which the 185 points were placed on the medial femoral epicondyle, including 5 attachments for the anatomic MPFL (proximal, central, distal, posterior, and anterior), which were placed within the dimple between the adductor tubercle and the medial femoral epicondyle, as described by the meticulous anatomic study of Kruckeberg et al.[32] Three patellar attachments (proximal, central, and distal) were selected to describe the anatomic MPFL length changes (Fig. 1).

Length Change Measurements

The length changes for each theoretical graft were measured as a function of knee flexion using in vivo 6 degrees of freedom knee joint kinematics. To create the path of a true graft, a direct line connecting the patellofemoral attachments (i.e., direct end-to-end distance) was projected on the bony surfaces using the convex hull algorithm to create a curved line, avoiding penetration of the connecting line through bone, that is, a ‘‘wrapping path’’ (Fig. 2). An optimization procedure was implemented 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 lengths.[73] The length of the 3D wrapping path (i.e., the line curved around the bony surfaces) was measured as the length of the theoretical graft. The MPFL length change data were calculated as follows: $Ln = L - L_0 / L_0 \times 100\%$; where Ln is the normalized length change, L is the graft length, and L0 is the reference length (defined as the length of the MPFL with the lower limb in full extension). Then, the offset at 0° of flexion, caused by the normalization procedure, was zeroed for each participant. The length change measurements had an accuracy of 0.3 ± 0.1 mm based on the systematic error of the registration method (i.e., dual fluoroscopic imaging technique). A heat map was created to provide visual representation of the isometry distribution over the medial femoral epicondyle by using the mean maximum percentage length change – mean minimum percentage length change of each theoretical patellofemoral graft during the lunge motion.

The patellofemoral attachment combination yielding the least length change was considered to be the most isometric graft.

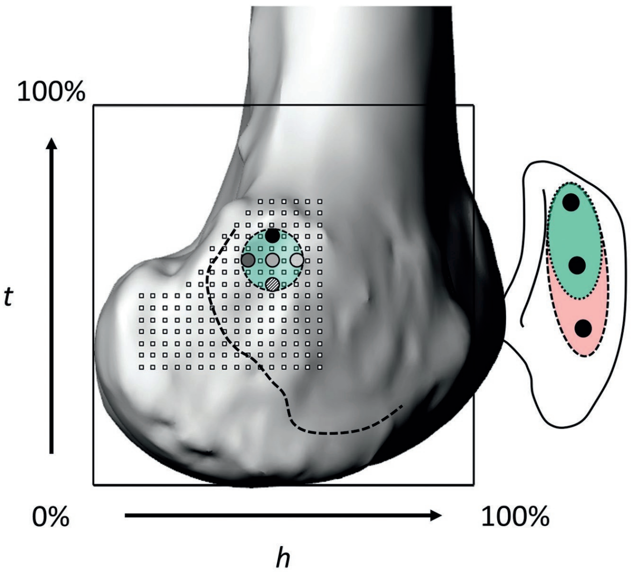

Fig. 1 True medial-lateral view of the knee in extension. The grid, as described by Stephen et al.[65], was applied to the medial femoral condyle. Line t was formed parallel to the posterior femoral cortex line, and line h was formed by the anterior-posterior distance of the medial femoral condyle; lines t and h were identical in length. Line t was connected with the anterior and posterior cortices of the medial femoral condyle. Line h was connected proximally to the tip of line t and distally with the femoral cortex. On the medial femoral epicondyle, 185 points were placed, and 3 patellar attachments (proximal, central, and distal attachments) were selected to describe the anatomic medial patellofemoral ligament (MPFL) length changes. The dashed line on the medial condyle lines shows the true Blumensaat line; the green filled circle on the medial condyle shows the anatomic MPFL attachment within the dimple between the adductor tubercle and the medial femoral epicondyle with its proximal, central, distal, posterior, and anterior attachments.

Statistical Analysis

We analyzed the changes in the length of the anatomic MPFL caused by flexion of the knee using repeated measures 2-way analysis with Tukey honest significant difference post hoc analysis, examining the 5 femoral attachments (i.e., proximal, central, distal, posterior, and anterior) connected to the 3 patellar attachments (i.e., proximal, central, and distal). Analyses were performed in MATLAB. P values 0.05 were considered significant.

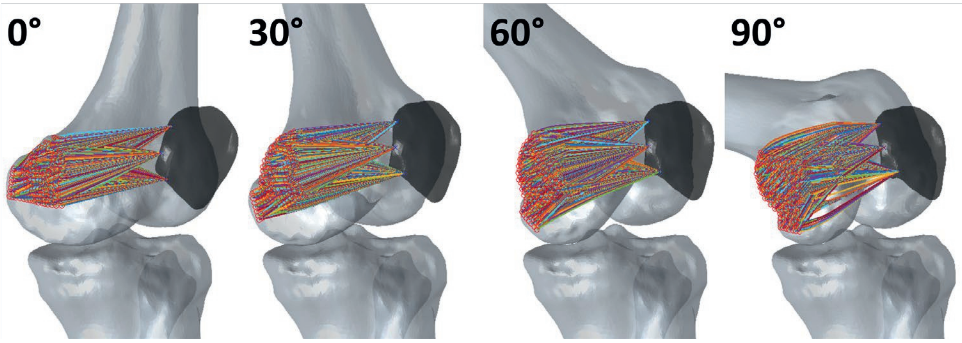

Fig. 2 Illustration of the knee with the 3-dimensional (3D) wrapping paths over the bony geometry of the femur and patella, that is, at 0°, 30°, 60°, and 90° of knee flexion. At each flexion angle, an optimization procedure was implemented to determine the shortest 3D wrapping path of each graft, creating a path of least resistance for the medial patellofemoral ligament.

RESULTS

Isometry

The most isometric femoral attachment was located posterior and proximal to the anatomic MPFL attachment area, that is, near the adductor tubercle (Fig. 3, Video 1 available on the journal's website). This was true for the proximal, central, and distal patellar attachments. The 3D wrapping paths of the femoral attachments proximal and anterior to the isometric zone increased with increasing flexion angles, whereas attachments distal and posterior to the isometric zone decreased with increasing flexion angles (Fig. 4). Moving the patellar attachment proximally caused the most isometric area to move proximally; conversely, the most isometric area moved distally with a distal patellar attachment.

Length Changes of the Anatomic MPFL

In the relaxed full extension position, for the central-to-central patellofemoral attachment, the anatomic MPFL had a mean length of 60.7 mm (95% CI, 58.0-63.4 mm). The central-to-central attachment of the MPFL was longest at full extension and rapidly decreased in length (i.e., slackened) between full extension to 35° of flexion, decreasing in length by 8.5%, and remained near isometric through the remainder of the flexion cycle (Fig. 5). The proximal and anterior femoral attachments tended to increase in length with deeper flexion angles, best seen for the central and distal patellar attachments. The length changes of the other patellofemoral attachment combinations are shown in Fig. 5. Moving the patellar and femoral attachments resulted in significantly different length changes ($P < .001$ for both) (Table 1). Post hoc analyses showed that moving the patellar attachment from central to proximal, central to distal, and proximal to distal caused significant different length changes ($P < .001$ for all). Moving the patellar attachment distally caused the 3D wrapping paths to increase in length at >30° to 110° of flexion (Fig. 5). Moving the femoral attachment from the central to proximal position caused a significant increase in length with knee flexion ($P < .001$). Moving the femoral attachment from central to distal caused a significant decrease in length with knee flexion ($P < .001$). For the proximal patellar attachment, no significant differences were found when moving in the anterior-posterior direction; however, for the central and distal patellar attachments, the length change patterns did alter when moving in the anterior-posterior direction. Detailed information is found in Table 1.

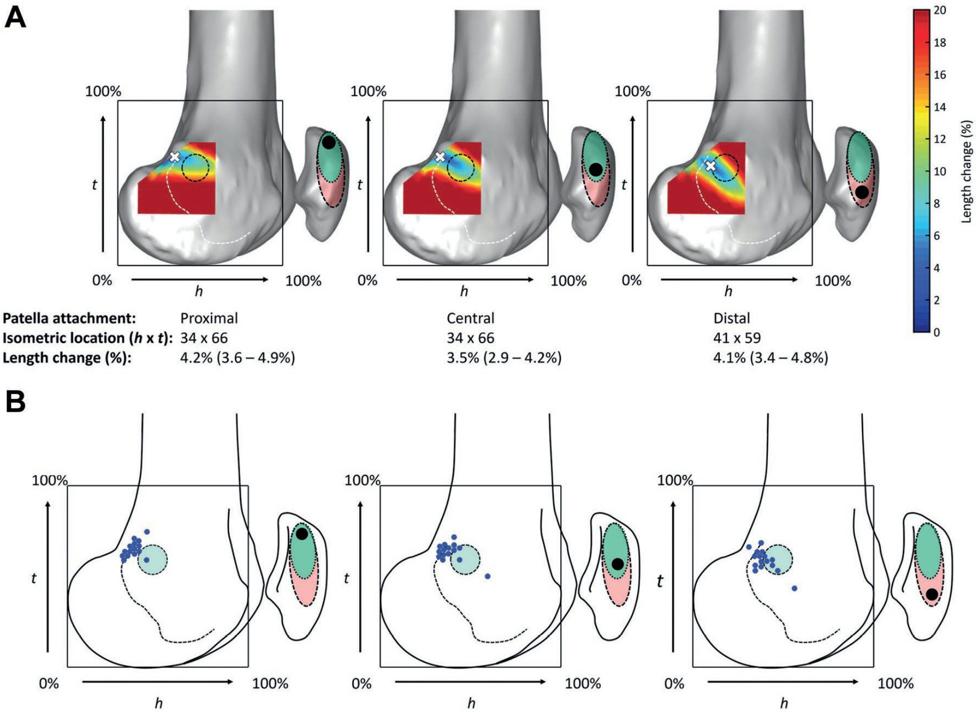

Fig. 3 (**A**) Heat map illustrating the isometry distribution (mean maximum % length change – mean minimum % length change) over the medial aspect of the medial femoral epicondyle for the 3-dimensional wrapping paths around the bony contours when connected to the proximal, central, and distal patellar attachments during the lunge motion. The darkest blue area on the femur shows a near isometric attachment area, while red areas highlight areas with a high degree of anisometry. The white cross represents the most isometric attachment. Values are shown as mean (95% CI). The dashed line (white) on the medial condyle lines shows the true Blumensaat line, and the dashed circle (black) on the medial condyle shows the anatomic medial patellofemoral ligament attachment area. (**B**) The most isometric attachment location per patellar attachment per patient.

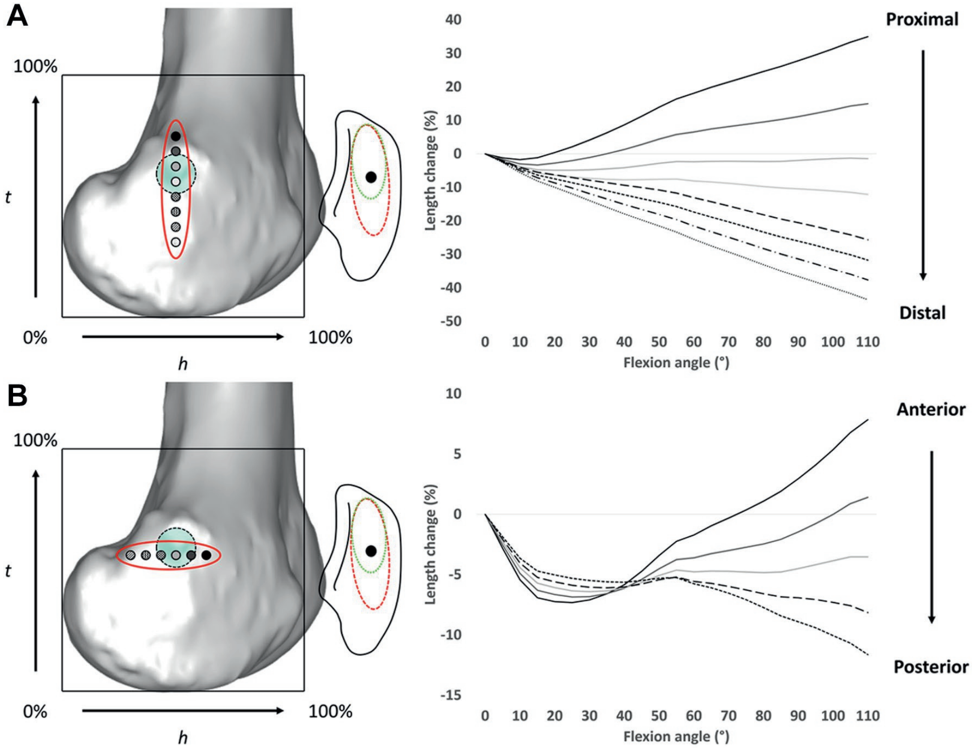

Fig. 4 On the left, a true medial-lateral view of a 3-dimensional femur model with several attachment points illustrated when moving (A) along the proximal-distal direction or (B) along the anterior-posterior direction. The normalized length changes for the attachments, when connected to the central patellar attachment, are shown by the line graphs on the right. Proximal attachments increased in length with increasing flexion angles, whereas distal attachments decreased in length with increasing flexion angles. When moving the attachment along the anterior-posterior direction, posterior attachments would decrease with increasing flexion angles, whereas anterior attachments would increase in length beyond approximately 30° of flexion. The greater the distance of a femoral attachment to the isometric zone, the greater the percentage length change as the knee flexes.

Table 1. Results of the repeated-measures two-way analysis of variance having the Tukey's Honest Significant Difference post hoc analysis for the length changes of the various studied bundles in the lunge motion.

	P-values									
Femur	**Central vs Proximal**	**Central vs Distal**	**Central vs Posterior**	**Central vs Anterior**	**Posterior vs Anterior**	**Posterior vs Proximal**	**Posterior vs Distal**	**Proximal vs Distal**	**Proximal vs Anterior**	**Distal vs Anterior**
Patella										
Proximal	<0.001	<0.001	*N.S. (0.983)*	*N.S. (1.0)*	*N.S. (0.958)*	<0.001	<0.001	<0.001	<0.001	<0.001
Central	<0.001	<0.001	*N.S. (0.357)*	0.031	<0.001	<0.001	<0.001	<0.001	<0.001	<0.001
Distal	<0.001	<0.001	<0.001	<0.001	<0.001	<0.001	*N.S. (1.0)*	<0.001	*N.S. (0.620)*	<0.001

DISCUSSION

The most important finding of this study was that the most isometric femoral MPFL attachment was located posterior and proximal to the anatomic MPFL attachment when connected to the 3 different patellar attachments. In addition, the anatomic MPFL is a dynamic, anisometric structure that was longest (i.e., tightest) in extension; the MPFL decreased in early flexion (i.e., ~30° of flexion) and remained near isometric during deeper flexion angles (i.e., >30° of flexion). Moving the femoral attachments in the proximal-distal direction significantly affected the length changes, whereas moving in the anterior-posterior direction had a much smaller but also significant effect. Similarly, moving the patellar attachments affected the length changes significantly, with more distal attachments causing increasing lengths at >30° of flexion.

Several researchers have attempted to define the isometry and length changes of the MPFL using various methods in both cadaveric[44,59,62,65,74] and in vivo settings.[27,40,52,61,69,78] Previous studies were often limited by using single[27] or only several[11, 22, 28, 44, 59, 61, 62, 65, 74, 78] patellofemoal points within or close to the MPFL attachment and have found different isometric locations. Our approach of analyzing the isometry using various attachments

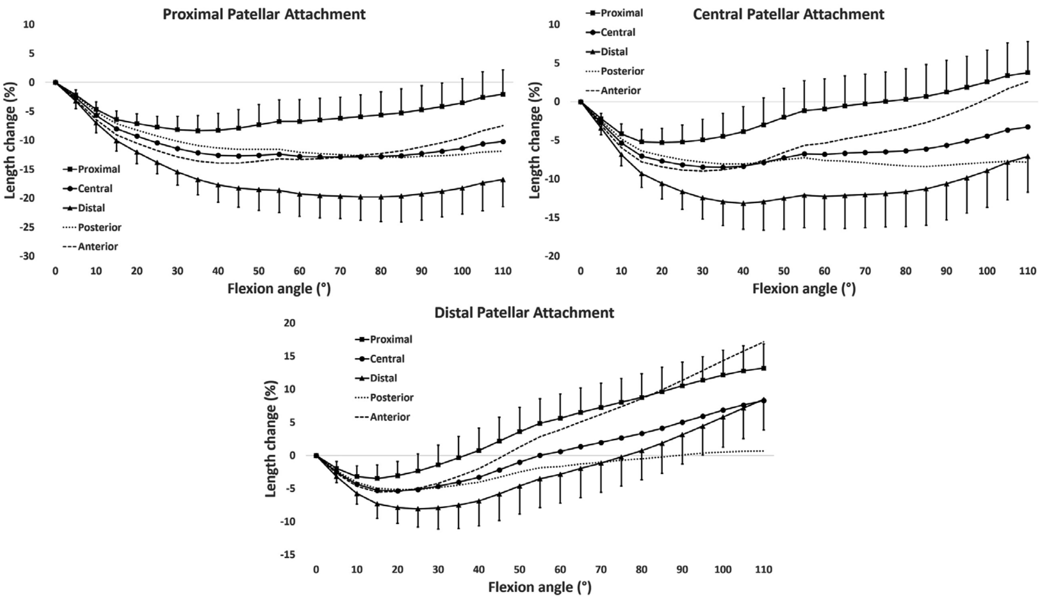

Fig. 5 Normalized length changes as a function of knee flexion for the 5 femoral anatomic medial patellofemoral ligament attachments (proximal, central, distal, posterior, and anterior) when connected to the 3 patellar attachments (proximal, central, and distal) in the lunge motion. Values are shown as mean (95% CI).

within a larger area of interest, including but not limited to the anatomic MPFL footprint on the medial femoral epicondyle, enabled us to find the most isometric area on the medial femoral epicondyle when connected to the 3 patellar attachments. We found the most isometric area to be posterior and proximal to the femoral MPFL attachment, close to/on the adductor tubercle when connected to any of the patellar attachments (see Fig. 3). Thus, the most isometric attachment had a nonanatomic location and yielded non-physiological length changes.

The adductor tubercle has been described as the ''lighthouse of the medial knee'' because when it is found, it allows the surgeon to find all other landmarks.[33,76] Recent anatomic studies have agreed that the MPFL is located in a dimple between the adductor tubercle and the medial femoral epicondyle[6,33,39,46,65,76]; this location cannot be palpated and therefore is hard to find during surgery. Several articles have described radiographic landmarks of the MPFL[7,31,48,55,69,76]; however, others have questioned the accuracy of performing anatomic reconstruction using these radiographic landmarks.[53,80] In this study, the center of the dimple was, on average, 7.2 mm (95% CI, 6.3-8.2 mm) distal and 4.6 mm (95% CI, 4.0-5.2 mm) anterior to the adductor tubercle. We argue that surgeons can use the adductor magnus tendon to routinely locate the adductor tubercle to find the ideal position for MPFL reconstruction.

In agreement with previous studies, we found that the length changes were more sensitive to changes in the proximal-distal direction[11,59,65,78] than in the anterior-posterior direction.[40,59,65] In addition, the effect of moving the patellar attachments distally also showed strong similarities with the patterns found in the cadaveric work by Stephen et al.,[65] leading to greater length changes for the distal patellar attachments, most evident at deeper flexion angles. Therefore, when performing MPFL reconstruction, any errors to be accepted on the femoral side should be made in the anterior-posterior direction (avoiding too anterior positions, which cause increased lengths at deeper flexion angles), not in the proximal-distal direction. On the patellar side, placement should not be more distal than the anatomic MPFL attachment, as this will cause increased length changes at deeper flexion angles.

We confirmed that the anatomic MPFL is a nonisometric structure that was longest (i.e., tightest) in extension, decreased during early flexion, and remained near isometric for the remainder of the flexion cycle. This is in agreement with the cadaveric work by Stephen et al.;[65] however, this was different than others have reported.[11, 22, 27, 28, 44, 59, 61, 62, 74, 78] These differences may be explained by methodological differences inherent to the cadaveric setup, limited selection of analyzed patellofemoral attachments, different loading conditions, and not considering the wrapping effect. The length change patterns of the anatomic MPFL suggest that its role is to prevent dislocations with the knee in extension and early flexion angles (which has been shown to be where the patella luxates most easily[2]) as well as keep the patella medially enough, pulling it toward and enabling its entrance in the trochlea, corroborating the MPFL function descriptions by Bicos et al.[8] At deeper flexion angles, the MPFL slackens, and its stabilization is primarily dependent on the patellofemoral geometry and becomes less important.[26] Perhaps because the patellofemoral geometry takes over the role as primary restraint to lateral patellar displacement at knee flexion angles beyond 30°, it is not necessary for the anatomic MPFL to be an isometric structure, as it is only providing secondary stability at deeper flexion angles.

Another key element for achieving successful MPFL reconstruction is the knee flexion angle for graft fixation.[63] Currently, there is no consensus on graft fixation angles in MPFL reconstruction; fixation angles varying from 0° to 90° have been proposed/used.[17,20,70] As the patellofemoral attachment combination determines its length change pattern, it is important for surgeons to realize that the graft fixation angle recommendations are attachment location specific. For example, graft fixation at 0° of flexion for a graft with a proximal femoral attachment will result in a graft that is slack in extension and early flexion angles and tightens with knee flexion, whereas a graft with a distal femoral attachment would be tight in extension and early flexion and slacken with knee flexion (Fig. 5). Previously, it was shown that minor changes in tunnel positioning and graft tensioning could already cause increased cartilage contact pressure.[63] Therefore, because the anatomic MPFL is longest at 0° of flexion, with a central patellar attachment, this may be the most suitable knee position for graft fixation to prevent overconstraint of the patellofemoral joint as the knee goes into flexion.

High complication rates have been described after MPFL reconstruction.[43,58] However, only few reports have described the potential causes of postoperative complications in the eye of femoral tunnel placement.[10,13,52,56,57] In the case series by Camp et al.,[13] nonanatomic femoral positioning of the MPFL was found to be the only significant risk factor for failure. Sanchis-Alfonso et al.[52] found that failed MPFL reconstruction was significantly anteriorly when compared with clinically successful reconstruction. Bollier et al.[10] found that graft positioning anterior and proximal to the anatomic femoral MPFL attachment caused medial patellofemoral articular overloading, iatrogenic medial subluxation, or recurrent lateral instability. These adverse outcomes may, in part, be explained by the length changes found in this study for such proximal and anterior attachments, which showed an increase in length with increasing flexion angles, causing the MPFL graft to overconstrain the patellofemoral joint and to repetitively elongate, leading to attenuation of the MPFL graft and ultimately failure. However, others were unable to find such strong correlations between femoral tunnel positioning and worse patient outcomes.[57] Future studies should focus on the tunnel location and postoperative outcomes to provide better insight on its clinical importance.

Finally, these data may be used in settings in which the above-suggested ideal femoral and patellar MPFL tunnel placements are impeded, for example, in revision cases with tunnels of the initial surgical procedure present. The heat maps per patellar attachment can serve as a map for surgeons to find the length changes that can most closely replicate the physiological length changes for MPFL reconstruction.

Limitations

There are some limitations to this study. Only healthy participants were studied. Future studies should include groups with patellar instability and patellofemoral abnormalities such as patella alta, trochlear and patellar dysplasia, a laterally positioned tibial tubercle, and the recently described variable short lateral posterior condyle.[49] Data were acquired during a lunge motion. Future studies may analyze different activities with different muscle loading conditions. The length changes were normalized to the MPFL length as measured with the leg in full extension as a reference. The precise reference lengths (zero load length) are unknown because of the in vivo nature of the study; hence, no force or true strain could be measured. Finally, no MPFL reconstruction was performed in the present study, so no definite conclusions could be generated regarding the most optimal graft positions.

Conclusions

The most isometric location for MPFL reconstruction was posterior and proximal to the anatomic femoral MPFL attachment. The anatomic MPFL is a dynamic, anisometric structure that was tight in extension and early flexion and near isometric beyond 30° of flexion.

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Chapter 8

General Discussion

BACKGROUND

Among the most commonly performed ligament reconstructions are the reconstructions of the anterior cruciate ligament (ACL), posterior cruciate ligament (PCL), anterolateral ligament (ALL) / lateral extra-articular tenodesis (LET) and medial patellofemoral ligament (MPFL). An improved understanding of the anatomy and biomechanics of the knee ligaments has led to advanced surgical techniques and rehabilitation protocols with improved patient outcomes.[59,70,74] During the last decades, cruciate ligament reconstruction has been marked by a paradigm shift from isometric to anatomic reconstruction.[8,16] It is believed that restoration of the native anatomy will result in the best postoperative knee kinematics, consequently leading to the most optimal clinical and patient outcome.[54,55,69] In ACL reconstruction, the transtibial drilling technique, pursuing isometric tibiofemoral tunnel positions to minimize graft length changes, made way for tibial-independent techniques which restored more accurately the native anatomy (e.g. anteromedial portal and outside-in retrograde drilling). Others have tried to restore anatomy using a double-bundle reconstruction technique, trying to restore the individual anteromedial and posterolateral bundle of the ACL to better restore rotatory stability of the knee.7 Analogous to the trends seen in ACL reconstruction, also in PCL reconstruction, anatomic reconstruction is pursued. For that matter, outside-in tunnel drilling and both single- and double-bundle PCL reconstruction techniques are advocated.[8]

Despite these efforts to improve cruciate ligament reconstruction, approximately 50% of the patients still develop osteoarthritis (OA),[38] and only 50% of patients return to their pre-injury level of sports participation.[1,2,60] Moreover, failure rates as high as 20-30% are seen.[24,34,39-41,48,51,58,65,72,76] Of these failed ligament reconstructions, almost one-third is caused by technical errors – errors which occur at the time of surgical reconstruction.[75] Of these technical errors, graft tunnel malpositioning and tensioning of the graft are the most frequently encountered problems which are surgically modifiable factors. Another factor contributing to failure of ligament reconstructions is failure to correct associated ligament instabilities. Recently, much attention has been given to the ALL as it is thought to have a key role in rotatory stability of the knee. Therefore, it is advocated that the ALL/LET reconstruction may be able to further improve outcome in the ACL deficient patient. However, currently controversy exists on the anatomy and function of the ALL. Even more, indications to perform an anatomic or non-anatomic LET are not established or even disputed. Nevertheless, tunnel positioning is essential to create a reconstruction that is biomechanically favorable to support stability to the intra-articular ACL deficient patient.

PURPOSE

The purpose of this thesis was to improve our understanding of the complex in vivo anatomic function of knee ligaments and to delineate the effects of having different tunnel position combinations on graft length changes. To do so, we investigated the ligament length changes and isometry of the native ACL, PCL, and MPFL. Furthermore, the isometry was studied of various points on the medial aspect of the lateral femoral condyle when connected to the tibial ACL attachment were studied; the lateral aspect of the medial femoral condyle when connected to the tibial PCL attachment, and the medial aspect of the medial condyle when connected to the patellar MPFL attachment. Because of the current increased popularity of lateral extra-articular procedures and its potential to augment the intra-articular ACL reconstruction to better restore the excessive internal tibial rotational laxity of the knee in ACL-deficient patients. Therefore, we studied the length changes of the native ALL and the isometry of the lateral femoral epicondyle with respect to Gerdy's tubercle and the native ALL attachment of the tibia. Ligament length changes are a useful measurement because they reflect the dynamics of ligament tensioning.[18] Therefore, the ligament length changes help us understanding the function of the native ligaments, the consequences of changing tunnel locations during ligament reconstructions and find the most optimal graft fixation angles to reproduce the native ligament function. Better understanding of the native ligament function and the consequences of tunnel positioning on the graft function are relevant to decrease the number of failures in ligament reconstruction surgery.

METHODOLOGY

For this thesis a combined imaging technique was used of dual fluoroscopy and either magnetic resonance imaging (MRI) or computed tomography (CT). The validation and accuracy of this technique has been described in detail previously.[67]

To study the ligament length changes of the knee in this thesis, different motions were used. Specifically, a step-up motion was used to study the length changes of the ACL and the ALL (Chapter 2, 3, 5), a sit-to-stand motion (i.e., similar to a box squat) was used to analyze the length changes of the ACL, ALL and simulated LETs (Chapter 2, 5 and 6), and a lunge motion was used to investigate the length changes of the PCL an MPFL (Chapter 4 and 7). The step-up motion, covering approximately 0 to 55° degrees of flexion, is a frequently performed daily activity and has been adopted as a closed-kinetic chain exercise in various lower extremity rehabilitation protocols.[70,74] Furthermore, previous in-vivo research found significant differences in the knee joint kinematics of ACL deficient versus intact knees, making it suitable for ACL research.[35] The sit-to-stand motion, covering approximately 0 to 90° of flexion, is another key movement of normal daily activities and is also used in various lower extremity rehabilitation protocols.[13,37,71,74] The sit-to-stand motion was deemed suitable for analyzing the length changes of the ACL, ALL and simulated LETs since it concerns were raised about potential overconstraint at the lateral compartment of extra-articular reconstructions, especially at deeper flexion angles.[11,12,42,56]

Lastly, the deep lunge, a strenuous motion covering approximately 0 to 120° degrees of flexion was used to study the length changes of the PCL and the MPFL. The PCL is known to be loaded at deeper flexion angles, previous research has shown significant differences in knee joint kinematics between intact and PCL deficient knees.[14,36] For the MPFL, postoperative complaints related to overconstraint of the patellofemoral compartment such as medial subluxation, limited knee flexion and postoperative knee pain have been described. These symptoms/complications may occur at both lower and deeper flexion angles. Therefore, the deep lunge was considered an interesting motion to study for the length changes and isometry of the PCL and MPFL.

To determine the in vivo length changes of the native ligaments of the knee, patient-to-patient geometric differences had to be overcome. Therefore, for the cruciate ligament studies (Chapter 2, 3 and 4) the quadrant method as described by Bernard et al.[3] was applied to the 3D models. Since such a quadrant method was not readily available for the lateral femoral epicondyle, we created a novel quadrant method to describe the graft locations on the lateral femoral epicondyle (Chapter 5).[30] For the MPFL study (Chapter 7) the quadrant method as described by Stephen et al.[62] was used. The quadrant method is independent of variation in knee size, it is practical, and reproducible. Thereafter, anatomical studies using the quadrant method to describe the attachment locations of the native ligaments of the knee were used to project the individual points to the femur, tibia and patella.[25,50,52,53]

The length changes of the virtual projected grafts at the 3D models were measured as a function of knee flexion. The direct line connecting the femoral and tibial or patellar attachment points were projected on the bony surfaces of the 3D knee models. This enabled to create a line that avoids penetration bone, and therefore followed bony geometry, that is, a wrapping path. An optimization procedure was implemented to determine the projection angle to find the shortest 3D wrapping path (this to mimic a trajectory of minimal resistance) at every studied flexion angle during the knee motion (i.e., approximately every 5° and 15° of flexion for continues and quasi-static motions respectively). This technique has been described in previous studies for measurements of ligament kinematics.[68] The length of the projected line (i.e., curved around the bony surfaces) was measured as the length of the graft. The length changes were then normalized to a reference as follows: $\varepsilon = L - L_0 / L_0 \times 100\%$, where ε is relative graft strain, L is graft length, and L_0 is the reference length. Thereafter, a heat map was created to provide a visual representation of the difference in isometry distribution across the femoral condyle. For this, the mean maximum length change – mean minimum length change of each theoretical tibiofemoral graft during the motion was calculated.

IMPLEMENTATION

This thesis provides orthopedic surgeons a combination of both descriptive and explanatory biomechanical information. At the same time, it has an educational aspect on some of the most frequently performed ligament reconstructions of the knee, which still has avoidable technical complications. We found that none of the studied anatomically positioned ligaments of the knee yielded isometric behavior. In contrast, the native ligaments exhibited a complex anisomeric behavior during knee motion.

In our ACL studies[31,32] we found that the native ACL was tightest in extension and during deeper flexion angles. A small area of least length change was found in the proximal-distal direction, just posterior to the intercondylar notch on the medial aspect of the lateral femoral epicondyle. Attachments located posteriorly to this isometric zone resulted in decreased graft lengths with increasing flexion angles, whereas more anteriorly attachments had less length changes at the same flexion angles. Tunnel positioning has been proven to be paramount in order to achieve successful outcome in ACL reconstruction. In patients who have technical errors contributing to the ACL graft failure, 80% is believed to have femoral tunnel malposition.[75] More specifically, the femoral tunnel is typically positioned too anteriorly and/or too vertical.[21,23,46,73] This anterior and vertical position of the femoral tunnel is overlapping the isometric zone as was found in our study,[32] and would be outside of the anatomical ACL footprint. Therefore, the anterior-vertical femoral positioning will be unable to mimic the native ACL length changes. This misdirected femoral tunnel is often seen when using the transtibial drilling technique to reach the femoral entry point. The transtibial drilling technique limits the surgeon to adequately choose its femoral entry point and direction of the tunnel, since the tibial tunnel, which is used to drill into the femur dictates to a great extend which position and which direction can be chosen for the femoral side. Over the past decade, the tibial independent drilling techniques such as anteromedial portal and outside in techniques have gained popularity since more freedom exists to choose the entry point and direction of the femoral tunnel.[9,66] The need for an anatomical position of the ACL tunnel is supported by our length change data. The latter holds especially for the femoral tunnel, suggesting that tibial independent drilling techniques are favorable as they may aid in reducing failure rates by their improved femoral tunnel positioning due to its greater freedom to choose the femoral entry point and direction.

In our PCL study[27] we found that the anterolateral bundle was slack in extension and tightened with deeper flexion angles, whereas the posteromedial bundle had some tension in extension, then slackened and tightened at flexion angles over 90° of flexion. A small area of least length change was found in the anterior-posterior direction approximately midway between the Blumensaat line just posterior to the intercondylar notch. Attachments distal to the isometric zone resulted in increased graft lengths with increasing flexion angles, whereas proximal attachments resulted in decreased graft lengths with increasing flexion angles. Similar to the situation in ACL reconstruction, femoral tunnels located at the isometric zone have been shown to lead to graft failure. Few reports are available on the etiology of failed PCL reconstructions. Noyes et al.[49] reported that incorrect tunnel placement was based on too proximal and too posterior placement of the tibial and femoral tunnels, respectively. From our own experience, failed PCL grafts had too proximal and far too anterior tibial tunnels. As in ACL reconstruction, in PCL reconstruction femoral tunnels placed outside of the anatomical footprint, and inside the isometric zone were associated with increased PCL failure rates. Interestingly, only minor adjustments in tunnel location will alter the ligament length changes significantly, especially if done for the femoral tunnels.[27] Therefore, non-anatomically placed tunnels will be unable to reproduce anatomical length changes and will thus increase the risk for graft failure. For that matter, isometric positioning in cruciate ligament reconstruction should be avoided. Isometric positioning of the graft will cause overconstraint or a too slack graft at certain flexion ranges during knee motion, leading to repetitive stretch-shortening cycles causing fatigue and ultimately failure of the graft.

An ALL reconstruction or LET is intended to aid the intra-articular ACL reconstruction to correct excessive internal rotatory laxity at full extension of the knee and during early flexion. We found that the native ALL (as described by Claes et al.[10] and Kennedy et al.[26]) was non-isometric between any of the flexion angles, and in fact, increased in length at deeper flexion angles.[29] This suggests that a structure at this location would be slack/loose in extension and tight at deeper flexion angles, and therefore, it would be unable to correct the excessive rotational laxity at full extension and lower flexion angles that is seen in ACL deficient knees. Moreover, it may overconstrain the knee at deeper flexion angles. We were therefore interested to see whether there were any tibiofemoral combinations that would be able to provide a tight graft in extension, and a slacker graft during deeper flexion angles to avoid overconstraint. To do so, we investigated the isometry of the lateral aspect of the lateral epicondyle to the Gerdy's tubercle and the native ALL attachment of the tibia.[30] In this study, we found that attachments posterior-proximal to the lateral femoral epicondyle were yielded this kind of behavior and would be suitable for LET.

Earlier, we published our anatomic ALL reconstruction,[28] using an autograft harvested of the iliotibial tract. The graft was fixed proximally slightly anterior to the fibular collateral ligament and just posterior-proximal to the popliteus tendon, then tunneled under the iliotibial tract, and distally secured at the anatomic tibial ALL attachment, with the knee in 90° of flexion and slight external rotation. We noticed early failures of this technique, which was altered since then. Both tunnel positions have been changed and instead of a "free" autograft, a strip of iliotibial tract is used from Gerdy's tubercle to the lateral aspect of the lateral condyle (i.e. at a point located posterior and proximal to the lateral femoral epicondyle). Subsequently, no more failures occurred, and we see a benefit of the LET procedure in patients with ACL deficiency and high-grade rotatory knee laxity (i.e., pivotshift 2-3), as was found by others too.[17,61]

Comparable to the cruciate ligaments and ALL studies, the MPFL study[33] also showed that this was an anisometric structure. The MPFL was tightest in extension and decreased in length until approximately 30° of flexion and remained near isometric in length during the remainder of the knee motion. The most isometric area was slightly posterior and proximal to the anatomic femoral MPFL attachment. Attachments proximal and anterior to the isometric area resulted in increasing lengths between femur and tibia with increasing knee flexion, whereas distal and posterior attachments caused decreasing lengths with increasing knee flexion. Moving both the femoral and patellar attachments resulted in significant different length changes of the MPFL. These biomechanical results translate seamlessly to the clinical outcome after MPFL reconstruction. Various case reports and case series [4,5,22,47,63] have tried to delineate the etiology of MPFL reconstruction failure, one of the primary causes of failure being tunnel malposition. Too proximal and anterior femoral tunnel positions have been shown to cause unfavorable functional outcomes, stiffness (at deeper flexion angles), tunnel widening of the medial cortex, higher rates of dislocations and increased graft failure rates.[22,47,57] These adverse outcomes may, in part, be explained by the length changes found in our study. The proximal and anterior tunnel positions resulted in an increase in length with increasing flexion angles, causing the MPFL graft to tighten, thus overconstraining the patellofemoral joint. Additionally, a proximal anterior femoral tunnel position of the MPFL graft would cause repetitive elongation, leading to attenuation of the MPFL graft and ultimately failure of the graft.

CURRENT AND FUTURE PERSPECTIVES

The length changes between fiducial tibial and femoral ligament attachments, as well as the isometry heat maps described in this thesis, can be used as a platform by surgeons performing ACL, PCL, ALL/LET and MPFL reconstructions. Furthermore, our studies improve the surgeons' understanding of the function of the native ligaments of the knee and the effects of changing the tunnel positioning during reconstruction. Ligament length changes are a great measurement because they indicate graft tension and aid in finding appropriate graft tensioning and fixation angles. However, length changes are also limited since they cannot provide the direction of this tension, this is the so-called ligament orientation (i.e., elevation and deviation angles). Therefore, future studies should focus on the orientation of the ligaments during knee motion to give us a better and more complete understanding of their complex function. Additionally, this will further help surgeons understand which areas are "unsafe" for ligament reconstruction because they cannot mimic the anatomic ligament function.

Clinical practice is driven by evidence-based medicine, and even more, no innovation should be done without evaluation.[43,44] Biomechanical studies, both in-vitro and in vivo, are essential to understand the principles of dynamic and static stability. The second phase would be to thoroughly evaluate current clinical practice on potential improvements based on findings from basic biomechanical research. Although during the last decade emphasize has been placed on patient reported outcome, which is important but is also determined by subjective factors like hope and optimism.[19] As for ligament reconstructions, based on findings from the earlier mentioned biomechanical studies, randomized controlled trials have to be designed. Since large numbers are necessary for these studies, it is preferred to have a multicenter setup, randomizing between established techniques and new techniques. In addition to these randomized clinical trials, all patients should be included in national ligament registries. The latter improves outcome and has been chosen by the Swedish and Norwegian ACL registries.[20] In these studies, RCT and registries alike, quantification of well established, important biomechanical variables (e.g., tunnel positioning, flexion fixation angles) should be collected. As for biomechanical studies, there has to be a consensus on minimum variable reporting, for example the number of cadaveric specimen, the type (human/animal), the type of preservation (e.g., fresh frozen, embalmed), clear description of robot and dissection protocols and reconstruction techniques (which should then also be quantified after performing it), and where possible standardized reporting of results and discussion sections. Thus, these standardized reporting protocols and quality checklists enable to increase the power and clinical applicability of meta-analyses. At present, a randomized clinical trial, measuring the knee biomechanics using a combined dual fluoroscopy and MR imaging technique is performed on patients with an acute ACL tear and clinical evidence of anterolateral rotatory instability in which an ACL reconstruction with or without non-anatomic LET is performed. This study analyses the tibiofemoral kinematics, in which the pre- and postoperative anterior translation and internal rotation are compared.

Since the recent rise in popularity of the lateral extra-articular tenodesis or ALL/ALC reconstructions, hundreds of studies on anatomy, biomechanics, surgical techniques and patient related outcome measures were published. Recently, two group meetings by different research teams[17,61] were held in an attempt to create consensus on this subject. However, indications for LET or ALL/ALC reconstructions remain vague and long-term clinical outcomes unclear.[17,61] Thus, extra-articular reconstructions of the knee to augment the intra-articular ACL reconstruction remain heavily debated which emphasizes the need for high quality multicenter, multinational studies.

Future studies on LET or ALL/ALC reconstructions should use devices that have been made to obtain quantitative assessment of the pivotshift test in the clinical setting,[64] or use radiologic modalities such as MR[45] and ultrasound[6] imaging which have been described to be able to demonstrate the presence of extra-articular injuries. Such devices and imaging techniques are interesting as they help us differentiate and objectify between intra-articular (i.e. ACL rupture) and extra-articular injuries (e.g. ALL/ALC injuries) of the knee. Next, in vitro studies should be performed to find the optimal LET or ALL/ALC reconstruction technique to reduce adverse effects such as overconstraint of the lateral compartment. After that, pilot studies analyzing the in vivo outcomes of the optimal reconstruction technique should be performed. Thereafter, high quality RCTs are needed to assess the patient related outcomes after ACL or ACL with extra-articular reconstruction.

Patients with a torn cruciate ligament tears can be categorized as coopers or non-coopers.[15] Little is known about the biomechanical differences between these types of patients. The combined CT/MRI and dual fluoroscopic imaging technique is suitable to further investigate this subject. This may reveal interesting results which may further help us identify the patients that benefit most from cruciate ligament reconstruction.

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SUMMARY

This thesis offers an overview of the length changes and isometry of the most frequently performed ligament reconstructions of the knee joint. The primary goal of this thesis was to improve our knowledge about the complex function of the anatomical ligaments of the knee. This newly gathered knowledge could improve the contemporary ligament reconstructions of the knee to reduce the amount of failed grafts related to tunnel malpositioning.

To study the length changes and isometry of the ligaments of the knee, we used a non-invasive imaging methodology to capture the in vivo biomechanics. Dual fluoroscopy was used to capture the in vivo joint motion and was combined with magnetic resonance (MR) or computed tomography (CT) imaging which were used to reconstruct the bony anatomy of the knee. To overcome the knee-to-knee differences, a quadrant method was used to apply the anatomical attachments of the ligaments to the 3-dimensional knee models.

In ***Chapter 2***, we show the length changes of the center of the anatomical anterior cruciate ligament (ACL) and the "over-the-top" position. Additionally, we show the isometry of several locations on the medial aspect of the lateral femoral condyle to the ACL attachment on the tibia. The most isometric tibiofemoral combination was found distal and anterior, outside of the anatomical ACL attachment area on the femur. We found that the anatomical ACL was anisometric and was tight at extension, and slackens during deeper knee flexion angles. Due to the impaired kinematics in patients with an ACL tear, i.e., the increased anterior tibial translation and internal rotatory laxity, the distance of the femur to the tibia measured from the ACL attachments significantly increases between 0° to 30° of flexion (***Chapter 3***).

In ***Chapter 4*** we show the length changes of the posterior cruciate ligament (PCL) and the isometry of the lateral aspect of the medial femoral condyle to the tibial attachment of the PCL. The anterolateral bundle of the PCL is slack at extension and tightens gradually during knee flexion. The posteromedial bundle is tight at extension, then slackens till 60° of flexion and tightens thereafter. The most isometric location was found proximal to, just outside, the femoral anatomic footprint of the PCL.

Approximately 25% of the patients that undergo an ACL reconstruction have persistent postoperative knee laxity. This excessive laxity exists primarily in the internal rotation direction with the knee in extension and early flexion ranges. Because of the trajectory of the anterolateral ligament (ALL), it is thought that simultaneous reconstruction of the ACL and ALL may overcome this frequently seen persistent postoperative laxity. In ***Chapter 5*** we show that the anatomic ALL with its attachment slightly anterior and distal to the lateral femoral epicondyle was anisometric and progressively increased in length (i.e. tightened) up to approximately 40% between 0° to 90° of flexion. This length increase makes an anatomic ALL reconstruction biomechanically unfavorable as it will fail. Specifically, the

anatomic ALL would be slack in extension and during early knee flexion and becomes increasingly tight at deeper knee flexion. Thus, the ALL is slack where it is intended to correct the excessive rotational laxity and will be too tight during flexion potentially harming the lateral compartment due to overconstraint.

Although the anatomic ALL reconstruction is unable to resolve the persistent postoperative excessive internal rotation, from a biomechanical point of view, it makes sense to solve any rotational abnormalities at a point further away from the center of rotation (the ACL). Thus, a non-anatomical lateral extra-articular tenodesis (LET) that is able to provide stability at extension and early knee flexion stays interesting. Therefore, in ***Chapter 6***, we show the isometry of several locations on the lateral aspect of the lateral femoral condyle connected to the anatomic attachment of the ALL on the tibia and Gerdy's tubercle. In this study, we were interested to see whether an area existed that yielded favorable length change patterns for an LET. Such area was found posterior and proximal to the lateral femoral epicondyle for both the anatomic tibial attachment of the ALL and Gerdy's tubercle.

Lastly, in ***Chapter*** *7*, we studied the length changes of the medial patellofemoral ligament (MPFL) and the isometry of the medial aspect of the medial femoral condyle to the patellar MPFL attachment. The anatomical MPFL is tight in extension and slackens till approximately 30° of flexion and stays near isometric thereafter. The MPFL allows the patella to smoothly enter the trochlea and prevents the patella from dislocating laterally at deeper flexion angles. The most isometric location was found posterior and proximal to the anatomical femoral footprint of the MPFL close to the adductor tubercle.

NEDERLANDSE SAMENVATTING

Dit manuscript geeft een overzicht van de elongatiepatronen en isometrie van de meest frequent uitgevoerde ligament reconstructies van het kniegewricht. Het primaire doel van dit manuscript was om een beter inzicht te krijgen in de complexe functie van de anatomische ligamenten van de knie. Deze nieuwe kennis helpt de orthopaedisch chirurg om ligament reconstructies te verbeteren om zo in de toekomst het aantal gefaalde reconstructies gerelateerd aan tunnel malpositionering te reduceren en de resultaten te verbeteren.

Om de elongatiepatronen en isometrie van de knieligamenten te meten werd een niet-invasieve meetmethode gebruikt die de in vivo biomechanica van de knie kan evalueren. Twee beeldenversterkers (fluoroscopen) die de in vivo beweging simultaan vastleggen werden gecombineerd met magnetische resonantie (MR) of computed tomography (CT) beelden waarmee de benige anatomie van de knieën werd gereconstrueerd. Om persoon-tot-persoon verschillen van de knie te overkomen, werd een kwadrant methode toegepast zodat de genormaliseerde locaties van de ligamenten op de 3-dimensionale kniemodellen konden worden geplaatst.

In ***Hoofdstuk 2*** beschrijven wij de elongatiepatronen van het anatomische centrum van de voorste kruisband (VKB) en de “over-the-top” positie. Daarnaast beschrijven wij de isometrie van de mediale zijde van de laterale femurcondyl richting de VKB-aanhechting op de tibia. De meest isometrische tibiofemorale locatie valt buiten de anatomische aanhechting van de VKB op de femur. De VKB is anisometrisch en is op spanning in extensie, naarmate de knie dieper flecteert wordt de VKB lakser. In patiënten met een afgescheurde VKB neemt bestaat er meer voorwaartse translatie van de tibia ten opzichte van de femur toe gedurende knieflexie. Deze toename in laxiteit leidt tot een significante toename in afstand tussen de aanhechting van de VKB op de femur en tibia tussen 0° tot en met 30° graden (***Hoofdstuk 3***).

In ***Hoofdstuk 4*** beschrijven wij de elongatiepatronen van de achterste kruisband (AKB) en onderzochten wij de isometrie van de laterale zijde van de mediale femurcondyl naar de AKB-aanhechting op de tibia. De anterolaterale bundel van de AKB is laks in extensie en neemt naarmate de knie dieper flecteert toe in spanning, de posteromediale bundel heeft in extensie enige spanning, wordt dan lakser tot ongeveer 60° en neemt daarna weer toe in spanning. De meest isometrische locatie valt net buiten de anatomische aanhechting van de AKB op de femur.

In ongeveer 25% van de patiënten die een VKB-reconstructie heeft ondergaan wordt een postoperatieve restinstabiliteit gezien. Deze persisterende postoperatieve instabiliteit bestaat vooral uit excessieve interne rotatie van het kniegewricht in extensie en vroege knieflexie. Door het verloop van het anterolateraal ligament (ALL) wordt gedacht dat gelijktijdige reconstructie van de VKB en ALL deze frequent geziene restinstabiliteit kan verhelpen. In

Hoofdstuk 5 beschrijven wij dat de anatomische ALL met de aanhechting net anterior en distaal van de laterale femur epicondyl anisometrisch was en op spanning komt naarmate de knie in diepere flexie komt. Tussen 0° en 90° flexie nam de ALL met ongeveer 40% toe in lengte. Hierdoor is een anatomische ALL-reconstructie biomechanisch onaantrekkelijk omdat deze zou falen. Daarnaast biedt de ALL geen stabiliteit in extensie of vroege knieflexie (dit is waar wij graag zien dat de reconstructie functioneert) en komt de ALL juist toenemend onder spanning te staan bij diepere knieflexie. Hierdoor kan het zo zijn dat tijdens flexie te veel spanning komt te staan op het laterale compartiment en daardoor juist schade veroorzaakt in plaats van voorkomt.

Ondanks dat een anatomische ALL-reconstructie niet in staat is de postoperatieve excessieve interne rotatie te verhelpen; vanuit een biomechanisch oogpunt is het zinvol om rotatieafwijkingen op te lossen op een punt verder weg van het rotatiecentrum (de VKB). Daarom blijft het interessant een niet anatomische laterale extra-articulaire tenodese (LET) te verrichten die wel in staat is de knie te stabiliseren in extensie en vroege knieflexie. In ***Hoofdstuk 6*** keken wij naar de isometrie van meerdere locaties op de laterale zijde van de laterale femurcondyl die verbonden werden met de anatomische aanhechting van de ALL op de tibia en Gerdy’s tubercle. Aan de hand van deze resultaten hebben wij gekeken of een gebied gevonden kon worden dat geschikt was voor een LET. Een gebied posterior en proximaal van de laterale femurepicondyl bezat deze eigenschappen voor zowel de anatomische ALL-aanhechting op de tibia en Gerdy’s tubercle.

Tot slot keken wij in **Hoofdstuk 7** naar de elongatiepatronen van de anatomische mediale patellofemorale ligament (MPFL) en isometrie van de mediale zijde van de mediale femurcondyl naar de MPFL-aanhechting op de patella. De anatomische MPFL staat op spanning in extensie, wordt dan iets lakser tot en met 30° en blijft daarna vrijwel isometrisch tot en met 110° flexie. De MPFL zorgt ervoor dat de patella soepel in de trochlea komt en daarna dat deze niet naar lateraal afglijdt. De meest isometrische locatie werd posterior en proximaal gevonden van de anatomische aanhechting van de MPFL op de femur, vlakbij het tuberclulum adductorium.

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CURRICULUM VITAE

Willem Alexander Kernkamp was born on April 13th 1989, in Madrid, Spain. The first part of his childhood he spent in Madrid and his second part he spent in Boxmeer. He has a one-year older brother, Dirk-Jan, and an eleven-year younger sister, Paula. After graduating from secondary school, he started his medical training at the Leiden University Medical Center, the Netherlands. During medical training, he completed a clinical rotation at the emergency department with dr. Sa'ad Lahri and dr. Hennie Lategan at the Stellenbosch University, Khayelitsha District hospital. Furthermore, he completed clinical rotations at the Orthopaedic Surgery department of the Haaglanden Medical Center with dr. E.R.A. van Arkel and at Deventer Hospital with dr. H.P. van Jonbergen. During his clinical rotations he studied the length changes of the anterolateral ligament and wrote a paper about the anterolateral ligament reconstruction.

After graduating *cum laude* from medical school in 2015, he joined the Bioengineering Laboratory at the Massachusetts General Hospital and Harvard Medical School in Boston, Massachusetts (USA), as a postdoctoral research fellow in 2015. With Prof. dr. G. Li, Prof. Prof. dr. T.Y. Tsai, Prof. dr. R.G.H.H. Nelissen, dr. E.R.A. van Arkel, dr. S.K. van de Velde and in collaboration with dr. R.F. LaPrade (Steadman Philippon Clinic) he initiated the study into the ligament length changes of the knee, as outlined in this thesis. In 2018 he went to Shanghai to continue his work on ligament length changes under the supervision of Prof. dr. T.Y. Tsai at the Shanghai Jiao Tong University, Shanghai. Some of the accolades of his stay in Boston include a 2nd place of the O'Donoghue Sports Injury Research Award, and two-time winner of the dr. Eikelaar Award of the Dutch Arthroscopy Association.

At the end of 2018, he returned to the Netherlands to commence his clinical work at Albert Schweitzer Hospital, Dordrecht, under the supervision of dr. P.W. Plaisier. He started his Orthopaedic surgery residency at the 1st of January 2020 in the ROGO Leiden under the supervision of Prof. dr. Nelissen and Dr. E.R.A. van Arkel. At the same time, he will continue to further build a long-lasting collaboration between Boston and Shanghai.

PUBLICATIONS

Publications related to this thesis

W.A. Kernkamp, N.H. Varady, J.S. Li, T.Y. Tsai, P.D. Asnis, E.R.A. van Arkel, R.G.H.H. Nelissen, T.J. Gill, S.K. van de Velde, G. Li: An in-vivo prediction of anisometry and strain in anterior cruciate ligament reconstruction – A combined magnetic resonance and dual fluoroscopic imaging analysis; Arthroscopy 2018

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