Original Article

Modified Lemaire Tenodesis Forces in Cadaveric Specimens Are Not Affected by Random Small-Scale Variations in the Femoral Insertion Point During Active Knee Joint Flexion-Extension

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Purpose: To directly measure lateral extra-articular tenodesis (LET) forces supporting anterior cruciate ligament reconstruction (ACLR) during dynamic flexion-extension cycles induced by simulated active muscle forces, to investigate the influence of random surgical variation in the femoral LET insertion point around the target insertion position, and to determine potential changes to the extension behavior of the knee joint in a cadaveric model. **Methods:** After iatrogenic anterior cruciate ligament deficiency and simulated anterolateral rotatory instability, 7 fresh-frozen cadaveric knee joints were treated with isolated ACLR followed by combined ACLR-LET. The specimens were tested on a knee joint test bench during active dynamic flexion-extension with simulated muscle forces. LET forces and the degree of knee joint extension were measured. Random variation in the LET insertion point around the target insertion position was postoperatively quantified by computed tomography. **Results:** In extension, the median LET force increased to 39 $\pm$ 2 N (95% confidence interval [CI], 36 to 40 N). In flexion over 70°, the LET was offloaded (2 $\pm$ 1 N; 95% CI, 0 to 2 N). In this study, small-scale surgical variation in the femoral LET insertion point around the target position had a negligible effect on the graft forces measured. We detected no difference in the degree of knee joint extension after combined ACLR-LET (median, 1.0° $\pm$ 3.0°; 95% CI, −6.2° to 5.2°) in comparison with isolated ACLR (median, 1.1° $\pm$ 3.3°; 95% CI, −6.7° to 6.1°; P = .62). **Conclusions:** LET forces in combined ACLR-LET increased to a limited extent during active knee joint flexion-extension independent of small-scale variation around 1 specific target insertion point. Combined ACLR-LET did not change knee joint extension in comparison with isolated ACLR under the testing conditions used in this biomechanical study. **Clinical Relevance:** Low LET forces can be expected during flexion-extension of the knee joint. Small-scale deviations in the femoral LET insertion point around the target insertion position in the modified Lemaire technique might have a minor effect on graft forces during active flexion-extension.

Knees with anterior cruciate ligament (ACL) injury often have rotational instability due to accompanying anterolateral insufficiencies.[1-4] Anterolateral stabilization in addition to ACL reconstruction (ACLR) can be performed to address high-grade rotational knee joint instability.[5] Recent clinical studies have shown less residual rotational knee joint instability after additional anterolateral stabilization and a lower ACLR graft repeated rupture rate in comparison with isolated ACLR.[6,7]

A widely used surgical technique for anterolateral stabilization is lateral extra-articular tenodesis (LET)

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using a strip of the iliotibial band. In the modified Lemaire LET technique, the iliotibial strip is routed deep to the lateral collateral ligament (LCL) and fixed near the femoral insertion of the LCL. For modified Lemaire LET, Kittl et al.[8] reported a length change of about 5% from 0° to 90° of flexion when measuring the distance from the Gerdy tubercle to the femoral LET insertion point. It is not known how LET forces are influenced by this length change during active flexion-extension. Furthermore, the influence of random surgical variation around the target insertion point in the modified Lemaire technique on LET forces remains unknown.

Biomechanical in vitro studies have reported possible kinematic changes after combined ACLR-LET compared with the intact knee joint over the years. The major concerns raised are reduced internal tibial rotation,[9-11] altered anterior translation,[9,12] and general overconstraint of the lateral compartment of the knee joint.[13] However, it is not known how knee joint extension behavior after combined ACLR-LET is possibly affected directly after surgery owing to the lateral-compartment constraint induced by the LET.

The purposes of this study were to directly measure LET forces supporting ACLR during dynamic flexion-extension cycles induced by simulated active muscle forces, to investigate the influence of random surgical variation in the femoral LET insertion point around the target insertion position, and to determine potential changes to the extension behavior of the knee joint in a cadaveric model. We hypothesized that LET forces would only increase to a limited extent and would be independent of random surgical variations around the target insertion point and that the addition of LET would not influence knee joint extension in comparison with isolated ACLR.

Methods

Preparation of Specimens

Seven fresh-frozen human knee joints with a median donor age of 89 years (range, 70-92 years; 2 men and 5 women) were used in the study. The specimens were donated to the local anatomic institute by individuals who had provided informed consent before death for the use of their bodies for scientific and educational purposes.[14] Quantitative computed tomography scans (LightSpeed VCT 64; GE Healthcare, Chicago, IL) were performed, and arthritic knees (Kellgren-Lawrence grade > 2) were excluded. Before testing, the specimens were stored at −20°C and thawed for 24 hours at room temperature. The knees underwent removal of skin and subcutaneous tissue. The knee capsule and inserting tendons were left intact. For active dynamic cyclic motion, the quadriceps tendons (vastus lateralis, vastus medialis, and vastus intermedius—rectus femoris) and hamstring tendons (biceps femoris and semimembranosus-semitendinosus) were armed with high-strength sutures (No. 5 FiberWire; Arthrex, Naples, FL) (Fig 1).

The approximated flexion-extension axis of the knee joint, described by Stannard and Schmidt,[15] was used as a reference for standardized embedding as well as positioning inside the test bench. A mediolateral pin was inserted into the femur, following the procedure described by Stannard and Schmidt. Correct placement was verified under lateral and anteroposterior radiographic visualization. Cutting of the femur was conducted at a distance of 150 mm from the flexion-extension axis. Cutting of the tibia was conducted at a distance of 110 mm from the flexion-extension axis. Cutting of the fibula was conducted at its neck, with tibial fixation of the fibular head using two 3.5-mm screws.

Embedding in polymethyl methacrylate cement (Technovit 3040; Kulzer, Wehrheim, Germany) was performed, accounting for individual knee physiology. A horizontal orientation of the tibial plateau could be ensured via the mediolateral pin inserted parallel to the tibial plateau. The tibial long axis was orientated as coaxially as possible inside the tibial potting mold. After embedding of the tibia, the femur was free to move and embedded in accordance with its natural Q-angle. Care was taken to position the posterior cortex of the long bone parallel to the tibial long axis (full extension) and avoid the introduction of artificial constraining forces during embedding. Physiological movement of the tibia inside the test bench was verified during pretest dynamic flexion-extension of the intact knee joint. The orientation of the femur was adjusted as necessary to align the anatomic flexion-extension axis of the knee joint with the flexion-extension axis of the test bench and ensure a natural flexion-extension motion in the sagittal plane of the test bench.

Surgical Technique

Insufficiency Model

A medial parapatellar arthrotomy was used to verify meniscal and ligament integrity, as well as to resect the ACL. To cut the anterolateral ligament as well as the capsule, a cut was created anterior and parallel to the LCL, from the lateral epicondyle to the joint line.[12,16] Care was taken to leave the meniscus and popliteus tendon intact. Finally, the distal Kaplan fibers of the deep iliotibial band at the lateral femoral condyle were cut[17,18] to complete the simulated anterolateral rotatory instability model.

ACL Reconstruction

One experienced orthopaedic surgeon (R.M.) carried out all of the surgical procedures. At the center of the femoral ACL footprint, a full 9-mm tunnel was drilled through the medial arthrotomy using an anteromedial

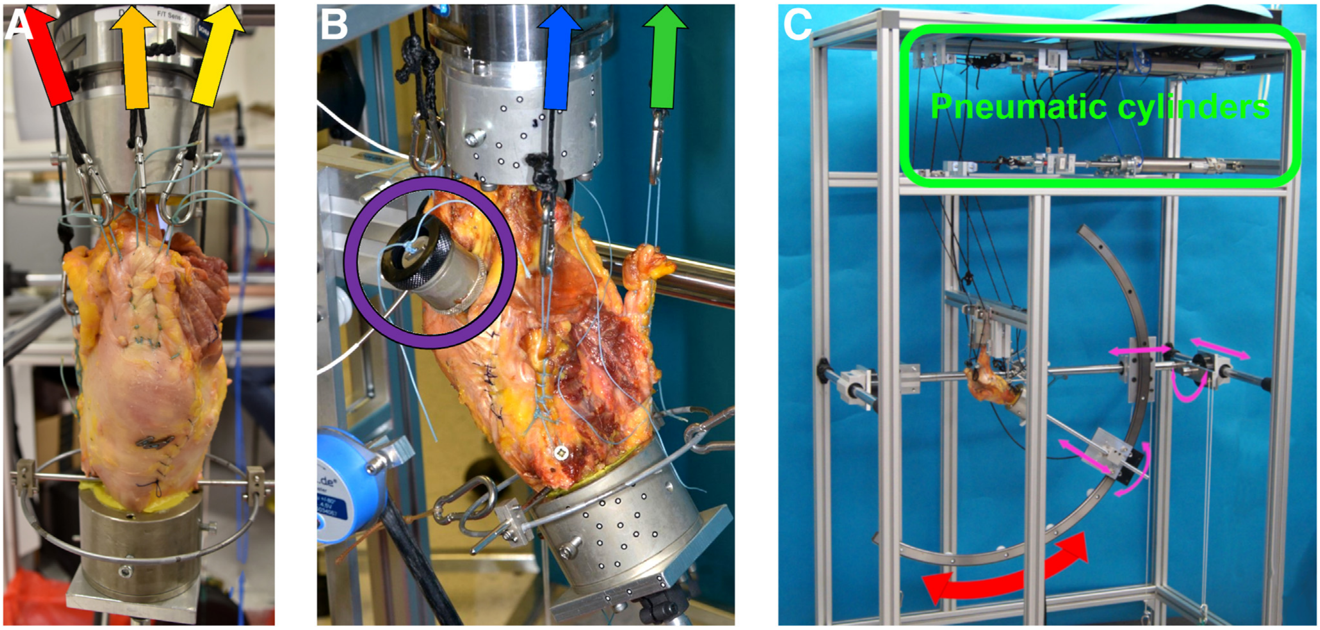

Fig 1. Right knee joint specimen inside test bench. (A) Armed quadriceps tendons (vastus lateralis [red arrow], vastus intermedius−rectus femoris [orange arrow], and vastus medialis [yellow arrow]). (B) Armed hamstring tendons (biceps femoris [green arrow] and semimembranosus-semitendinosus [blue arrow]) and cortically fixed custom load cell device (purple circle) for measuring lateral extra-articular tenodesis force. (C) Knee joint test bench with pneumatic actuator cylinders. Active dynamic flexion-extension cycles (red arrow) were induced while all other tibial degrees of freedom (pink arrows) were unconstrained.

drill guide (Arthrex) with an offset of 7 mm. At the tibial ACL stump, a full 9-mm tunnel was drilled using a tibial drill guide set at an angle of about 55° to result in a minimal tunnel length of 40 mm. Bovine digital extensor tendons folded into a 2-stranded graft with a diameter of 9 mm were used to perform ACLR. Although these are not human tendons, their biomechanical behavior is comparable and limits the variation in tendon properties[19,20] in the ACLR. Baseball stitches (No. 2 Ethibond; Ethicon, Raritan, NJ) were used to suture the free ends. After preconditioning for 10 minutes under 89 N, the ACLR graft was fixed at 30° of flexion at 80 N using 9 × 30−mm PEEK (polyether ether ketone) interference screws (Arthrex).[21-23] Tibial graft strands were additionally fixed using a small fragment screw with a washer at the tibial cortex. Finally, the medial parapatellar arthrotomy was sutured.

Lateral Extra-articular Tenodesis

A full 7-mm femoral tunnel was drilled, aiming for 8 mm proximal and 4 mm posterior to the lateral epicondyle of the femur in accordance with the modified Lemaire technique.[8,24] Without the use of drilling guidance, random surgical variation around this target insertion point was accepted and investigated in our study. LET tunnel angulation primarily aimed for 30° proximal and 30° anterior but could be slightly adjusted if a risk of tunnel collision with the ACLR was seen after insertion of a Kirschner wire. A central 10-mm broad strip of the iliotibial band with fibers oriented toward the Gerdy tubercle was prepared, leaving the attachment to the Gerdy tubercle intact. After the graft had been secured with whipstitch sutures (No. 2 FiberWire; Arthrex), it was shuttled deep to the LCL and through the femoral tunnel. The free end was connected medially to a load cell, which was incorporated into a custom device screwed into the tunnel (Fig 1B). The LET was finally fixed at 60° of flexion at 20 N in neutral tibial rotation, as this has been reported to approximate physiological knee joint motion with nearly isometric graft behavior using the modified Lemaire technique.[8,22] This is also the tensioning protocol used in our clinical practice.

Biomechanical Measurements

To actively move the knee joint dynamically, a validated test bench controlled by a custom LabVIEW Virtual Instrument (version 11.0; National Instruments, Austin, TX) was used (Fig 1C).[25] In contrast to the rigidly mounted femur, the tibia had unconstrained motion, with 6 *df*.[25] Five pneumatic cylinders (DSNU; Festo, Vienna, Austria) actuated knee joint flexion-extension over a rope-and-pulley system. The armed quadriceps and hamstring muscles were therefore connected to the rope-and-pulley system of the test bench (Fig 1 A and B). The initial control settings for flexion-extension actuation were adjusted to result in a 0° to 75° open-chain flexion-extension cycle for the intact knee joint resulting in maximal quadriceps forces

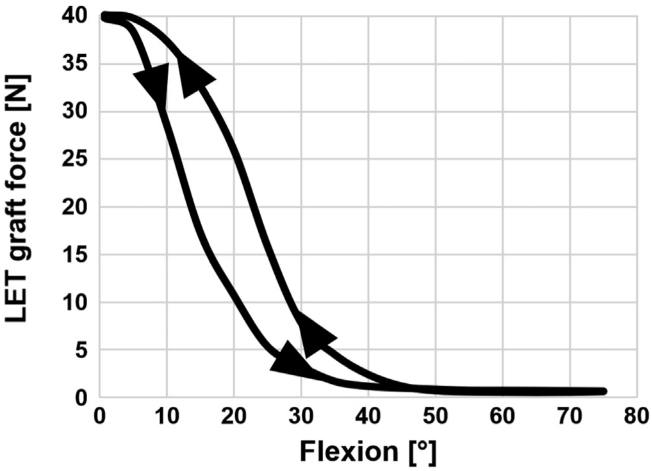

Fig 2. Mean lateral extra-articular tenodesis (LET) force using modified Lemaire technique during 5 cycles of muscle-induced dynamic flexion-extension in 1 example specimen. Arrowheads indicate the direction of motion from extension to flexion and back to extension showing the hysteresis.

of 237 N (86 N for vastus lateralis, 85 N for vastus medialis, and 66 N for vastus intermedius—rectus femoris) and maximal hamstring forces of 711 N (360 N for biceps femoris and 351 N for semimembranosus-semitendinosus), which are in the range of estimated muscle forces during the swing phase of gait.[26] Keeping the initial control settings constant for the remainder of testing allowed detection of varying degrees of knee joint extension between the intact knee joint and insufficient knee joint, as well as between isolated ACLR and combined ACLR-LET. Muscle loads during knee flexion-extension were adjusted by a custom LabVIEW Virtual Instrument controlling valves inside a pneumatic control unit (Shadow Robot, London, England) until the actuation of the native state was reached. Miniature load cells (Burster Präzisionsmesstechnik, Gernsbach, Germany) measured the LET forces. An inclinometer (NG4U; Seika Mikrosystemtechnik, Wiggensbach, Germany) measured the flexion angle, including the degree of maximal extension. In combination with a PICAS measuring amplifier (Peekel Instruments, Bochum, Germany), LabVIEW was used to record graft forces as well as the knee joint flexion angle at 50 Hz.

The femoral LET insertion point was localized postoperatively by computed tomography in a 3-dimensional model of each specimen created in Volume Viewer (version 16.0; GE Healthcare). After superimposition of the posterior femoral condyles in the sagittal plane, anterior-posterior and proximal-distal distances between the lateral epicondyle of the femur and the center of the insertion point were measured. The center of the insertion point was identified by fitting an ellipse to the footprint of the tunnel on a true lateral view. The absolute distance between the target insertion point and the actual insertion point was calculated using the Pythagorean theorem. To minimize the effect of subjective influences, the position was measured by 3 observers (M.S., R.M., and B.G.) and the mean value was calculated.

Testing Protocol

After preliminary investigation of the intact knee joint, as well as the insufficient knee joint, the isolated ACLR condition was tested first, followed by testing of the combined ACLR-LET condition. Initial graft preconditioning and relaxation were conducted before each test by dynamic cycling of the joint 10 times, followed by graft retensioning. Each specimen was then actively flexed and extended 11 times. Cycles 6 to 10 were evaluated to investigate LET forces as well as the degree of knee joint extension.

Data Analysis

Maximal LET force values and knee joint extension angles were calculated as the average of the 5 investigated cycle maximums for each specimen. Femoral LET insertion point values were analyzed based on the deviation from the target value of the modified Lemaire tenodesis technique,[24] described in the literature as having graft behavior close to isometry.[8]

Statistical analysis was conducted using IBM SPSS Statistics software (version 27; IBM, Armonk, NY). Because of the limited sample size (N = 7), a non-normal distribution was assumed. A paired-sample Wilcoxon signed rank test was used to compare the degree of knee joint extension between isolated ACLR and combined ACLR-LET. Two additional paired-sample Wilcoxon signed rank tests were used to compare the degree of knee joint extension between the intact knee joint and the insufficient knee joint, as well as the isolated ACLR condition. Multiple-comparison correction was performed using Bonferroni correction. The significance level was set at .05. Medians with standard errors and 95% confidence intervals (CIs) are reported. Bootstrapping (n = 1,000) was used to calculate standard errors and 95% CIs. Correlations between femoral LET insertion points and LET forces were analyzed using the 1-sided Spearman rank correlation coefficient.

Results

LET Force

At maximal extension, the LET force was highest and increased to 39 ± 2 N (95% CI, 36 to 40 N). During flexion, the force dropped and the LET was offloaded in flexion over 70° (2 ± 1 N; 95% CI, 0 to 2 N). Marked hysteresis occurred during flexion-extension cycles, with LET force values being higher for flexion-to-

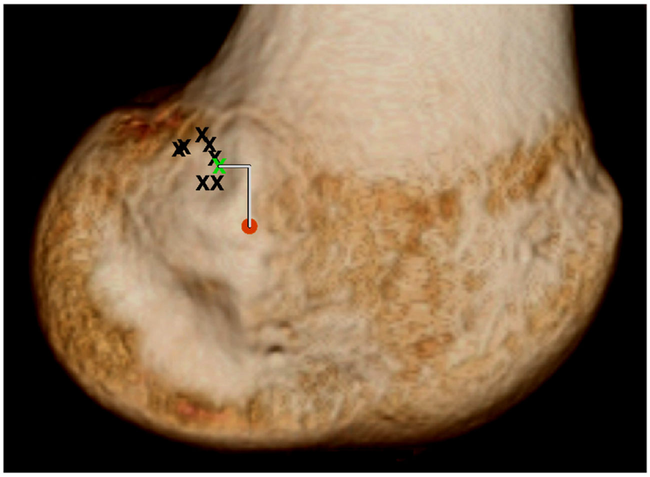

Fig 3. Positions of femoral lateral extra-articular tenodesis insertion points (black X's) relative to lateral epicondyle (orange circle) in right femur. The green X indicates the target insertion point for the modified Lemaire lateral extra-articular tenodesis (8 mm posterior and 4 mm proximal to the apex of the lateral epicondyle[8,21]).

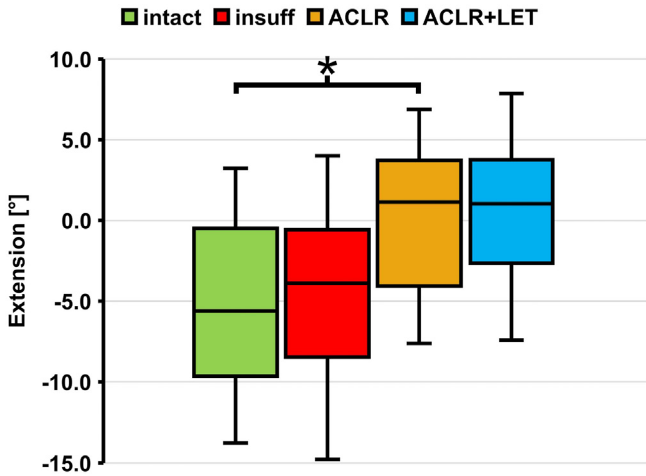

Fig 4. Maximum knee joint extension angle during active flexion-extension: intact knee, insufficient knee (insuff), isolated anterior cruciate ligament reconstruction (ACLR), and combined ACLR and lateral extra-articular tenodesis (ACLR+LET). Negative values indicate hyperextension. No statistically significant difference was evident between isolated ACLR and combined ACLR-LET (P = .62) or between the intact state and insufficient state (P > .99). The degree of knee joint extension significantly decreased after isolated ACLR in comparison with the intact state (P = .036, asterisk).

extension motion in comparison with extension-to-flexion motion (Fig 2).

Femoral LET Insertion Point

Proximal-distal deviation of the femoral LET insertion point from the target value of the modified Lemaire tenodesis technique (green X in Fig 3) was 2.3 ± 1.6 mm (95% CI, −2.2 to 2.9 mm). Anterior-posterior deviation of the femoral LET insertion point from the target value was 2.2 ± 1.1 mm (95% CI, 0.6 to 4.7 mm). Individual femoral insertion points for all tested LETs are shown in Figure 3.

Correlation Between LET Force and Insertion Point

No correlation was observed between maximal LET force and proximal-distal deviation (P = .255), posterior-anterior deviation (P = .153), or absolute distance (P = .215). The position of the femoral LET insertion point had a negligible effect on LET forces.

Degree of Knee Joint Extension

There were no significant differences in the maximal knee joint extension angle after combined ACLR-LET (1.0° ± 3.0°; 95% CI, −6.2° to 5.2°) in comparison with isolated ACLR (1.1° ± 3.3°; 95% CI, −6.7° to 6.1°; P = .62) (Fig 4). Furthermore, we found no differences in knee joint extension after simulated insufficiency (−3.9° ± 3.6°; 95% CI, −12.4° to 0.9°) compared with the intact knee joint (−5.6° ± 3.7°; 95% CI, −10.9° to 0.7°; P > .99). Maximum extension decreased after isolated ACLR in comparison with the intact knee joint (P = .04) under the conditions used in this biomechanical study with steady muscle-force activation between tested states. A decrease in extension after isolated ACLR therefore does not represent clinical extension loss but indicates affected extension behavior after ACLR compared with the native knee joint. All of the relevant numerical values for maximum LET force, femoral LET insertion point, and degree of knee joint extension are summarized in Table 1.

Discussion

During active dynamic flexion-extension, LET forces peaked at maximal extension. Measured values were independent of random small-scale variations in the femoral LET insertion point around the target insertion position. Knee joint extension behavior after combined ACLR-LET in comparison with isolated ACLR did not change.

LET Force

Measured LET forces reached peak values at the maximum degree of extension in this study. Our study is in line with a study by Kittl et al.[8] investigating the length change patterns of different femoral LET insertion points. Kittl et al. investigated maximal elongation of a suture routed deep to the LCL and spanned between the LET attachment positions used in our study at full extension. Elongation of the suture decreased with increasing knee joint flexion. LET forces using the same attachment positions and course deep to the LCL are therefore expected to peak at full extension and decrease with flexion, as shown by our data.

Table 1. Summary of LET Parameters Investigated in Each Specimen

	Maximum LET Force, N	Deviation of Femoral LET Insertion Point From Target Insertion Point, mm			Maximum Extension Angle, °	
		Proximal (+)/Distal (−)	Posterior (+)/Anterior (−)	Absolute (Direct)	Isolated ACLR*	Combined ACLR-LET*
Specimen 1	40	2.9	1.3	3.1	−1.4	1.0
Specimen 2	39	4.2	2.3	4.8	1.3	2.4
Specimen 3	26	2.3	5.4	5.9	1.1	0.9
Specimen 4	36	1.1	0.6	1.3	6.1	7.9
Specimen 5	39	2.6	4.7	5.4	−6.7	−7.4
Specimen 6	55	−2.2	2.2	3.1	6.9	5.2
Specimen 7	40	−2.2	0.3	2.2	−7.6	−6.2
Median	39	2.3	2.2	3.1	1.1	1.0
Minimum	26	1.1	0.3	1.3	−7.6	−7.4
Maximum	55	4.2	5.4	5.9	6.9	7.9

NOTE. The target insertion point of the LET was 8 mm proximal and 4 mm posterior to the lateral epicondyle of the femur, in accordance with the modified Lemaire technique.[8,21]

ACLR, anterior cruciate ligament reconstruction; LET, lateral extra-articular tenodesis.

*Negative values indicate hyperextension at maximum knee joint extension angle.

Measured LET forces reached a peak value of 55 N in this study. Because other published data on directly measured LET forces during unloaded flexion-extension cycles are missing, the respective length change pattern reported by Kittl et al.[8] was used to estimate reference values and compare them with the measured values. Kittl et al. detected a maximum length change of 5.1% over knee flexion for the LET configuration used in our study. With an LET cross-sectional area of approximately 6 mm^2, as in our study, and assuming a Young's modulus of 270 MPa,[27] LET forces can be calculated to be around 80 N for a strain of 5.1%. Deviation of absolute LET forces can be explained by inter-specimen variation in the Young's modulus of the iliotibial band.[28]

LET force estimation based on length change data allows an initial plausibility evaluation of the measured values. However, the experimental conditions in our study are not exactly the same as those in the study by Kittl et al.[8] The comparability of the measured and estimated LET forces is therefore limited. Knee joint kinematics during dynamic flexion-extension induced by active muscle forces might differ from the kinematics during static testing with constant muscle forces[8] and could have influenced measured LET forces. Another aspect to bear in mind is the fact that Kittl et al. measured the length change pattern of a single suture under a small but constant tensioning force (0.5 N). The suture was therefore always tensioned over the range of motion. Contrary to this, the LET fixed at 60° of flexion was able to become slack over the course of flexion-extension with no or only low graft forces. This might also explain the lower graft forces observed in our study in comparison with the estimated values based on the study of Kittl et al., given that the transition from the graft being slack to being taut was in the tested flexion range. Low graft force values over 70° of flexion (<2 N) support this assumption and can be explained as the graft was fixed at 60° of flexion. For higher flexion values, graft slackening is expected.

Slackening of the LET at deeper flexion angles might be favorable given that natural internal rotation increases with flexion.[23,29] The biomechanical behavior observed in the applied Lemaire technique—with a tighter LET close to extension stabilizing the joint and a slacker LET at deeper flexion angles allowing natural internal tibial rotation—might be desirable for LET techniques.[30] Overall, the LET forces measured during active dynamic flexion-extension cycles varied only slightly and reached a low maximum value of 55 N. Undesirable graft stretching over 6%, as described by Kittl et al.,[8] is therefore unlikely to occur. Different methods of actuation with varying quadriceps and hamstring muscle forces could have influenced absolute LET forces and should be taken into account when one is interpreting the results.

Tunnel angulation and associated bone-graft friction might have affected the LET forces measured. However, the bone-graft interface is assumed to have a low friction coefficient owing to the wet environment. Undisturbed translation of the LET inside the femoral tunnel was ensured during testing. Direct measurement of LET forces in line with the lateral routing was technically not possible.

Femoral LET Insertion Point

Accurate femoral tunnel positioning in LET procedures is not easy and is prone to error. In particular, the widely used method of intraoperative palpation of anatomic landmarks involves a risk of suboptimal placement.[31] An experimental in vitro study quantified the variation in tunnel positioning in LET using the Lemaire technique as ±5.5 mm in the anterior-posterior direction and ±5 mm in the proximal-distal direction (with standard deviations reported).[31] These data are in line with the variation observed in our study (maximum anterior-posterior deviation of 5.4 mm and maximum proximal-distal deviation of 4.2 mm). Fluoroscopy could improve and simplify LET tunnel positioning[32,33] but requires time-consuming additional steps. However, small-scale variations in the femoral LET insertion point had a negligible effect on LET forces in our study. A systematic comparison of different insertion positions was not part of the study design. The aim of this study was to investigate whether surgical variation around a specific target insertion point has a significant influence on LET forces.

Although the pulley effect of the LCL in LET routing deep to the LCL was not examined in this study, it could be the reason for the negligible effect of small-scale variations in the femoral LET insertion point on LET forces. While being deflected, small-scale variations in the femoral insertion point of the LET only slightly change the course of the LET between the LCL insertion and the femoral LET insertion. The main course of the LET between the LCL insertion and the tibial attachment remains unchanged. Therefore, LET forces may not have been significantly affected by small-scale variations in the femoral insertion point. Studies comparing measured LET graft forces during active dynamic flexion-extension between superficial routing and deep routing to the LCL are needed to clarify this assumption.

Degree of Knee Joint Extension

Achieving full active knee extension is crucial after ACLR.[34] Although there are no published clinical data on loss of extension directly related to the ACLR itself, the extension behavior might be affected after ACLR. Our results showed altered extension behavior between the intact knee joint and ACLR-treated knee joint. Emphasis is raised once again that the decreased degree of knee joint extension after ACLR compared with the intact knee joint does not account for a clinically relevant loss of extension. ACLR graft fixation at 30° of flexion might be the reason for altered knee joint extension behavior. Fixation at 0° of flexion could have eradicated this effect but is reported to be inferior in terms of restoring knee stability.[35]

Combined ACLR-LET did not further change the extension behavior in comparison with isolated ACLR. Williams et al.[36] compared the flexion-extension behavior of isolated ACLR versus combined ACLR-LET in a short-term clinical follow-up study. No differences between isolated ACLR and combined ACLR-LET could be found. Knee joint extension also varied in the range of ±5° to ±7° after treatment compared with the intact contralateral knee. To fully understand the effect of an LET on knee joint motion, 3-dimensional analysis during active dynamic flexion-extension is needed. Although the range of extension did not change after the addition of the LET, the rotational behavior of the tibia might have changed but was not investigated in our study.

Limitations

This study has some limitations. No active in vivo processes were included in this biomechanical in vitro study. Each measurement only depicts the time-zero condition of treatment. Only the knee joint, without skin, subcutaneous tissue, and the lower extremity (ankle and foot), was investigated. The small sample size limits the significance of the results. There was no control group treated by another technique or with the LET routed superficially instead of deep to the LCL. For the ACLR, bovine tendons were used. The donor age of the specimens tested is not representative of typical patients treated with combined ACLR-LET. Tunnel angulation and associated bone-graft friction might have affected the LET forces measured. The varying Young's modulus between the specimens' iliotibial bands is also an influencing factor for comparison of absolute LET forces. Examination of knee joint motion was limited to the degree of flexion-extension instead of a full 3-dimensional analysis. Finally, ground reaction forces were not simulated, and care should be taken when extrapolating the results to LET forces during clinical scenarios.

Conclusions

LET forces in combined ACLR-LET increased to a limited extent during active knee joint flexion-extension independent of small-scale variation around 1 specific target insertion point. Combined ACLR-LET did not change knee joint extension in comparison with isolated ACLR under the testing conditions used in this biomechanical study.

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