Contributions of Femoral Fixation Methods to the Stiffness of Anterior Cruciate Ligament Replacements at Implantation

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Summary: One purpose of this study was to determine the stiffness of three femoral fixation methods used commonly in anterior cruciate ligament (ACL) reconstruction to secure a double-looped semitendinosus and gracilis (DLSTG) graft and then assess how the stiffness of these methods affects the stiffness of the young human femur-fixation method-graft complex at the time of reconstruction. A second purpose was to define principles for adjusting the stiffness of the ACL replacement (defined as the femoral fixation method plus DLSTG graft plus tibial fixation method) to match that of the native ACL. The stiffness of a DLSTG graft and the stiffness of the femur-fixation method-DLSTG graft complex for three endoscopic fixation methods were measured. Fixations of the DLSTG graft to a button, anchor, and post, both with and without compaction of bone, were tested in young, human femur. The stiffness of each fixation was calculated by modeling the DLSTG graft and fixation method as a series of springs. The stiffness of the DLSTG graft averaged 954 ± 292 N/mm. The stiffness of the DLSTG graft-fixation complex was lowered fourfold to 40-fold by adding fixation. The method of fixation determined the reduction in stiffness. The stiffness of the femur-button-DLSTG graft complex averaged 23 ± 2 N/mm, the femur-anchor-DLSTG graft complex averaged 25 ± 3 N/mm, and the femur-post with bone graft-DLSTG graft complex averaged 225 ± 23 N/mm ($P = .0001$). The knot in the suture loop was the least stiff component and determined the stiffness when the DLSTG graft was fixed with both the button and anchor. Compaction of bone significantly increased stiffness by an average of 41 ± 14 N/mm ($P = .027$). Because the stiffness of femoral fixation methods are 4 to 40 times less than the stiffness of the graft, increasing the stiffness of an ACL replacement would be best achieved by selecting fixation methods with higher stiffness and not by either shortening the graft or increasing the cross-sectional area of the graft. **Key Words:** Femoral fixation— Hamstring graft.

The stiffness of an anterior cruciate ligament (ACL) replacement (i.e., femoral fixation method plus ACL graft plus tibial fixation method) can influence

the restoration of the normal limits of anterior translation, which is a primary goal of an ACL reconstruction. For a given pretension, increasing the stiffness of the ligament replacement at the time of implantation can reduce the limit of anterior tibial translation and overconstrain the knee. Alternatively, reducing the stiffness of the ligament replacement can increase the limit of anterior tibial translation and result in excessive laxity or instability. Because the stiffness of the ligament replacement affects the limit of anterior translation, it has been stated that achieving normal knee kinematics at the time of reconstruction may be more dependent on matching the stiffness of the

ligament replacement to that of the native ACL instead of matching it's ultimate strength.[1]

Another disadvantage of using a ligament replacement that is less stiff than the native ACL is that the pretension required to restore the normal limits of anterior translation produces a force in the graft that is higher than in the native ACL.[2] To provide a basis for comparison, the stiffness of the native femur-ACL-tibia complex in humans has been reported to be 182 N/mm,[3] 242 N/mm,[4] and 303 N/mm.[5] The maximum force in a patellar tendon graft, with a stiffness of 18% to 28% of the native ACL (51 N/mm),[5] was threefold higher compared with the forces in the native ACL when the graft was pretensioned to restore normal anterior-posterior laxity.[2] The higher forces in the graft required to restore normal laxity may exceed the ability of a fixation method to prevent either slippage or failure.

Because there are benefits to matching the stiffness of a ligament replacement to that of the native ACL, an understanding of the different factors that influence the stiffness of ligament replacements at the time of implantation may be useful. The stiffness of the ligament replacement can be affected by the cross-sectional area of the graft, the length of the graft, the material of the graft, the stiffness of the femoral fixation method, the stiffness of the tibial fixation method, and the stiffness of the bone. What remains to be determined is whether the stiffness of the ligament replacement at the time of implantation is influenced more by the stiffness of the fixation methods or by the stiffness of the graft, which is controlled by both the area and length for a specified material such as human tendon.

The first specific objective of this study was to measure the stiffness of a double-looped semitendinosus and gracilis (DLSTG) graft and then measure the combined stiffness of the graft and three endoscopic fixation methods used to fix this graft in young human femur. A second objective was to compute the stiffness of the fixation methods and then determine if any differences were significant. The final objective was to determine whether the stiffness of a ligament replacement at the time of implantation is affected more by changing the stiffness of the fixation methods than by changing the stiffness of the graft.

METHODS AND MATERIALS

Stiffness and Ultimate Load of DLSTG

The stiffness and ultimate load of the DLSTG were derived from measurements obtained using a materials testing machine with a 5 kN load cell (5566 Materials Testing Machine; Instron Corp, Canton, MA). The gracilis and semitendinosus tendons were harvested from six human donors ranging in age from 40 to 65 years (average, 55 years). Muscle was removed and sutures were sewn into the end of each tendon. The cross-sectional area of each tendon at the midsection was measured using an area micrometer.[6] The centers of both tendons were looped over a rigid 6.35-mm diameter steel peg attached to the actuator. A 5-N weight was attached to each suture to pretension all four limbs. A liquid nitrogen freeze clamp, designed to prevent slippage of the graft,[7] was applied to the body of the tendon 60 mm from the peg. An extensometer, constructed using a linear variable displacement transformer (LVDT), was bolted to the peg and freeze clamp to measure graft elongation. The graft was then preconditioned by applying a cyclic tensile load between 70 to 500 N at 1 Hz for 10 cycles. Permanent deformation of the graft during preconditioning was avoided by limiting the upper load to 500 N, which was eight times less than the average failure load of the graft (4,213 N). Therefore, the preconditioning load was confined within the elastic region of the load-elongation curve. The graft was then loaded to failure at a strain rate of 10% per second. This rate was chosen because it lies between the limits of strain rates used in other studies.[3,4,8-10] A personal computer was used to administer the tests and acquire both load and elongation data (Instron Series IX Software).

Stiffness and Ultimate Load of Femur-Fixation-Graft Complex

To calculate the stiffness of the fixation method, the stiffness of the graft, in isolation, had to be determined before testing the femur-fixation device-DLSTG complex. The gracilis and semitendinosus tendons were harvested from seven donors ranging in age from 31 to 67 years (average, 64 years). The DLSTG graft was prepared, mounted in the materials testing machine, and preconditioned as previously described. The graft was then loaded to a subfailure load of 700 N at a strain rate of 10% per second, and stiffness was calculated from the linear region of the load-elongation curve.

The same DLSTG graft was used to test all three fixations within a femur. A pilot study showed that the fixations failed at less than one third of the ultimate load of the graft, which was well within the elastic region of the load-deformation curve of the DLSTG.[11] Seven distal femurs were obtained from young donors ranging in age from 28 to 45 years (average, 35 years).

The femur was thawed, soft tissue was removed, and the diaphysis was centered, bolted, and cemented with bone cement inside a steel tube.

To avoid carry-over effects, the button (EndoButton; Smith & Nephew Endoscopy, Andover, MA) was tested first followed by the anchor (Mitek Ligament Anchor; Mitek Surgical Products, Norwood, MA) and post (Bone Mulch Screw; Arthrotek, Inc, Ontario, CA). The location of the femoral tunnel was selected by drilling a 2.4-mm guide wire into the intercondylar roof 6 mm from the over-the-roof position at 1 o'clock for the left knee and 11 o'clock for the right knee. An 8-mm diameter 25-mm long endoscopic femoral tunnel was drilled. The tunnel was extended through the anterolateral femoral cortex with a 4.5-mm diameter cannulated reamer (Smith & Nephew Endoscopy).

The graft was attached to the button with a loop of 6.5-mm wide polyester suture (MeadoxMedicals, Oakland, NJ). The loop was adjusted to match the length of the 4.5-mm diameter tunnel and tied with a square knot. To assess whether the knot slipped under load, an ink mark was placed on each suture limb as it exited the knot. The centers of both tendons were placed in the suture loop, the button was pulled through the femoral tunnel with a suture, turned, and seated on the cortex of the femur (Fig 1A).

The steel tube, containing the femur-button-DLSTG graft complex, was inserted into a fixture attached to the crosshead of the materials testing machine. The graft and tunnel were positioned in line with the actuator by adjusting the orientation of the fixture, which allowed two translational and three rotational degrees of freedom. All four limbs of the graft were simultaneously pretensioned by attaching a 5-N weight to each suture. The extensometer was clamped to a pin drilled into the lateral condyle, and bolted to the freeze clamp. The length of the graft was adjusted to 60 mm and the freeze clamp was applied. The femur-button-DLSTG graft complex was loaded to failure at a strain rate of 10% per second.

Next, fixation of the DLSTG graft with an anchor was tested by linking the anchor to the graft by a loop of 6.5-mm wide polyester suture using the same technique described for the button. The 4.5-mm diameter segment of the femoral tunnel was expanded to 6 mm. The anchor-DLSTG graft complex was pulled through the femoral tunnel and seated with at least two of the four fixation arcs gaining purchase on the cortex (Fig 1B). The length of the graft was adjusted to 60 mm and the freeze clamp was applied. The femur-anchor-DLSTG graft complex was tested to failure using the previously described protocol.

Finally, fixation of the DLSTG graft with the post, with and without bone graft, was tested. The post was inserted through an 8-mm diameter tunnel in the lateral femoral condyle drilled perpendicular to the femoral tunnel using a guide (U-guide; Arthrotek).[12,13] The post extended from the end of a 10.5-mm diameter threaded screw. The body of the screw was hollow to allow compaction of cancellous bone into the femoral tunnel through the screw. The screw was inserted partially across the femoral tunnel, a suture was looped over the post, and the screw was advanced until 2 mm of the post was imbedded in the medial wall of the tunnel. The two tendons were tied to the suture and the graft was pulled through the femoral tunnel until the middle of both tendons rested on the post (Fig 1C).
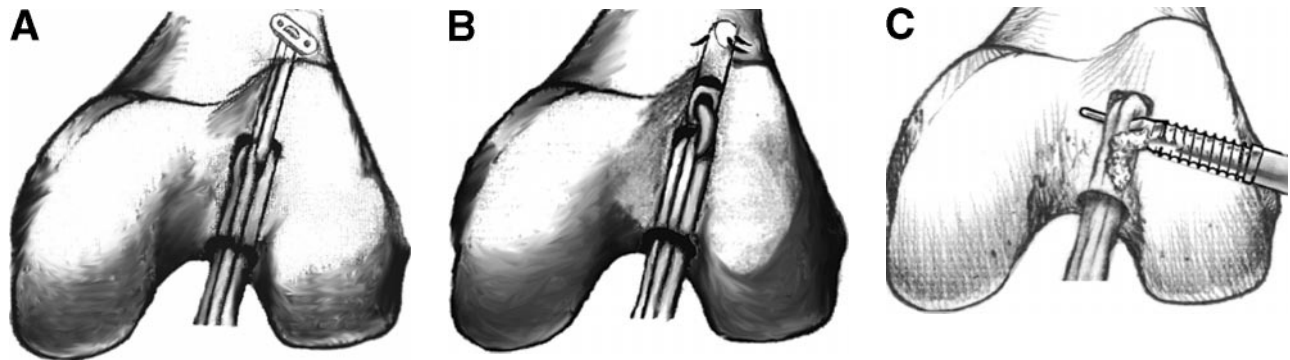
FIGURE 1. (A) How the button was used to fix the DLSTG to the femur. The suture loop connected the DLSTG graft to the button. The least stiff component of the femur-fixation-DLSTG graft construct was the knot in the suture bridge which was the primary determinant of the stiffness of the entire construct (24 ± 2 N/mm). (B) How the anchor was used to fix the DLSTG graft to the femur. The suture loop connected the DLSTG graft to the anchor. The least stiff component of the femur-anchor-DLSTG graft construct was the knot in the suture, which was the primary determinant of the stiffness of the entire construct (26 ± 2 N/mm). (C) How the post was used to fix the DLSTG graft to the femur. The femur-post with bone graft-DLSTG graft construct was the stiffest fixation method (225 ± 23 N/mm). Compacting bone into the femoral tunnel increased the stiffness by 41 ± 14 N/mm.

The length of the graft was adjusted to 60 mm and the freeze clamp was applied.

The stiffness of the femur-post without bone graft-DLSTG graft complex was determined by using the previously described testing protocol with the exception that the maximum load was limited to 500 N to prevent failure. Bone graft, collected from tunnel reamings, was then compacted into the femoral tunnel through the body of the screw. The femur-post with bone graft-DLSTG graft complex was then tested to failure using the previously described protocol.

Stiffness and Ultimate Load of Suture Loop

The stiffness of the polyester suture loop was determined by passing the loop around 6.35-mm steel pegs mounted in the base and crosshead of the materials testing machine. The loops were prepared to match the length of the loops used with the button and anchor and were tied with a single square knot. Testing on 14 loops (7 for each fixation device) was performed to failure using the previously described protocol.

Data Analysis

The stiffness of each DLSTG (K_{DLSTG}) and femur-fixation-DLSTG complex ($K_{f-f-DLSTG}$) were derived from the linear region of the load-elongation plot using a regression analysis. The stiffness of the femoral fixation (K_{f-f}) was calculated starting with the equation

$$1/K_{f-f-DLSTG} = 1/(1/K_{f-f} + 1/K_{DLSTG})$$

derived from a spring-in-series model of the graft-fixation complex and then rearranging this equation to solve for

$$K_{f-f} = K_{f-f-DLSTG} \times K_{DLSTG}/(K_{DLSTG} - K_{f-f-DLSTG}).$$

The ultimate load was defined as the maximum load carried by either the graft or the femur-fixation-DLSTG graft complex. The cross-sectional area of each graft was obtained by doubling the sum of the cross-sectional areas of the gracilis and semitendinosus tendons.

A repeated measures, one-factor analysis of variance was used to determine if the independent variable, the femoral fixation method (K_{f-f}), affected the stiffness and ultimate load of the femur-fixation-DLSTG graft complex ($K_{f-f-DLSTG}$) (SAS statistical package; SAS Institute, Carey, NC). Each graft-femur combination was treated as a random, independent block. Duncan's method of multiple comparison was used to compare the stiffness and ultimate loads of the femur-button-DLSTG graft, femur-anchor-DLSTG graft, and femur-post with bone graft graft-DLSTG graft complexes. The effect of bone compaction on the

stiffness of the fixation with the post was analyzed using a paired *t* test. The effect of the suture loop on the stiffness of the femur-button-DLSTG graft and femur-anchor-DLSTG graft complex was analyzed using a paired *t* test.

A final analysis was to compare the stiffness of the ligament replacement for different femoral and tibial fixation methods. For the purposes of this study, the ligament replacement was defined as being composed of three components; the femoral fixation method, graft, and tibial fixation method. The stiffness of a ligament replacement K_{LR} was calculated by using the equation $K_{LR} = 1/(1/K_{f-f} + 1/K_{DLSTG} + 1/K_{f-t})$ where K_{f-t} represented the stiffness of the fixation in the tibia. The magnitude of the effect of varying the stiffness of femoral and tibial fixation methods was determined by comparing the calculated stiffness of the ligament replacement using reported values for the most and least stiff femoral and tibial fixation methods in all four possible combinations.

RESULTS

Stiffness and Ultimate Load of DLSTG

The stiffness of the DLSTG graft averaged 954 ± 292 N/mm (mean ± standard error) and the ultimate load averaged 4213 ± 250 N. The upper limit of the linear region of the load-elongation curve ranged from 3,356 to 5,225 N. A toe-in region was not observed. Failure occurred in midsubstance between the fixation peg and clamp. The graft did not fail where it contacted the post. The cross-sectional area of the DLSTG graft averaged 43 ± 4.5 mm^2, which was slightly smaller than the 50-mm^2 cross-sectional area of the femoral tunnel.

Stiffness and Ultimate Load of Graft-Fixation-Femur Complex

The stiffness of the femur-fixation-DLSTG graft complex was determined by the type of fixation ($P = .0001$). The stiffness of the femur-post with bone graft-DLSTG graft complex averaged 225 ± 23 N/mm and was significantly greater than the stiffness of the femur-button-DLSTG graft complex which averaged 23 ± 2 N/mm, and the femur-anchor-DLSTG graft complex, which averaged 25 ± 3 N/mm (Fig 2) ($\alpha = 0.05$). Compaction of cancellous bone in the femoral tunnel significantly increased the stiffness by an average of 41 ± 14 N/mm ($P = .027$).

The ultimate load of the femur-fixation-DLSTG graft complex was determined by the type of fixation ($P = .0001$). The ultimate load of the femur-post with
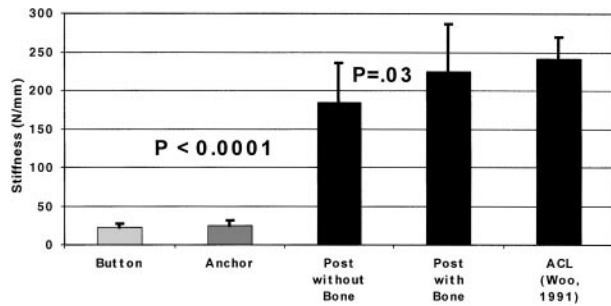
FIGURE 2. Stiffness comparison of four femur-fixation-DLSTG graft complexes to the native ACL. The stiffnesses of the femur-button-DLSTG graft complex and the femur-anchor-DLSTG graft complex were the same because the suture loop was the most elastic component. The stiffness of the femur-post without bone-DLSTG graft complex and the femur-post with bone graft-DLSTG graft complex were significantly greater than the stiffness of the femur-button-DLSTG graft complex and the femur-anchor-DLSTG graft complex with the button and anchor ($P < .0001$). Compaction of bone graft into the femoral tunnel between the post and DLSTG graft significantly increased the stiffness 41 N/mm ($P = .03$). The stiffness of the native ACL from young specimens is provided for comparison.

bone graft graft-DLSTG graft complex averaged 1126 ± 80 N and was significantly greater than the ultimate load of the femur-button-DLSTG graft complex, which averaged 430 ± 27 N and the femur-anchor-DLSTG graft complex which averaged 312 ± 35 N ($P < .05$).

Failure occurred at the fixation site and not in the midsubstance of the graft. The femur-button-DLSTG graft complex failed by the knot in the suture loop slipping and tightening before the suture tore. In one case, the knot slipped and untied. The femur-anchor-DLSTG graft complex failed by the arcs bending backward allowing the anchor to migrate distally in the femoral tunnel. The femur-post with bone graft graft-DLSTG graft complex failed by the post either bending or fracturing at the junction with the screw, which allowed the graft to slip off the tip of the post. The range of the ultimate load of the femur-fixation-DLSTG graft complex was from 210 to 1542 N, which was always below the upper limit of the linear region of the load-displacement curve which ranged from 3,356 to 5,225 N.

Stiffness of Each Fixation Method

From the springs-in-series analysis, the calculated stiffness of the post with bone graft averaged 575 ± 117 N/mm. The calculated stiffness of the button and suture loop averaged 24 ± 2 N/mm, which was not significantly different from the calculated stiffness of the anchor and suture loop, which averaged 26 ± 3 N/mm.

The length of the suture loop averaged 44 ± 4 mm in the femur-button-DLSTG graft complex and 24 ± 3 mm in the femur-anchor-DLSTG complex. The stiffness of just the suture loop was not significantly different from the stiffness of the femur-button-DLSTG graft complex ($P = .80$) and the femur-anchor-DLSTG graft complex ($P = .16$).

Effect of Varying the Stiffness of the Femoral and Tibial Fixation Methods on the Stiffness of a Ligament Replacement

Because the stiffest femoral fixation method was the fixation post with bone compaction (575 N/mm), and the least stiff femoral fixation method was either the button or anchor due to the elastic suture bridge (24 N/mm) that was common to both fixation methods, the stiffnesses of these femoral fixation methods were used as the extremes in the computations of ligament replacement stiffness. From another study in our laboratory in which the stiffnesses of different tibial fixation methods were measured in young human tibia (average age, 35 years), the stiffest tibial fixation method was the WasherLoc (506 ± 197 N/mm) (Arthrotek) and the least stiff tibial fixation method was No. 5 suture (Ethibond; Ethicon Inc, Cincinnati, OH) sewn to each of the four limbs of the graft and tied around a 4.5-mm diameter bicortical screw (70 ± 19 N/mm).[14] Assuming a graft length of 80 mm with a scaled stiffness of 715 N/mm, the calculated stiffness of the ligament replacement using the springs-in-series analysis for each of the four fixation combinations yielded 196 N/mm for the stiffest femoral and tibial fixation methods, 50 N/mm for the stiffest femoral and least stiff tibial methods, 22 N/mm for the least stiff femoral and stiffest tibial fixation methods, and 17 N/mm for the least stiff femoral and tibial fixation methods.

DISCUSSION

Understanding the principles that determine the stiffness of an ACL ligament replacement may be important because stiffness affects the anterior-posterior laxity of the knee at the time of implantation. For example, stability cannot be restored if a low stiffness ligament replacement is used unless the graft is tensioned higher than the native ACL. Alternatively, a high stiffness ligament replacement is undesirable because it may overconstrain the knee.

Methodological Issues

Four methodological issues should be understood before the findings from this study are discussed. The

decision to use and reuse the same DLSTG graft for testing all three methods in a femur was based on a pilot study that predicted that the femur-button-DLSTG graft and femur-anchor-DLSTG graft complexes would fail far below the yield point of the DLSTG graft and not permanently deform the graft. Both of these femur-fixation-DLSTG graft complexes failed below 550 N, which was only 13% of the average ultimate load of the DLSTG graft (4,213 N).[11] The decision to reuse the DLSTG graft did not compromise the results of the study because plastic deformation of the graft did not occur during failure loading of these fixations.

The DLSTG graft was tested instead of a rope, wire, or prosthetic graft because a standardized material that has the same geometry, and material and structural properties as the DLSTG graft could not be found. Using the DLSTG graft rather than a substitute also allowed any interaction between the graft, fixation method, and bone tunnel to contribute to the stiffness of the complex. Because the relative contribution of each component to the stiffness of the femur-fixation-DLSTG graft complex was the focus of the study, it was important to use the DLSTG graft rather than a substitute.

Sequential, nonrandomized testing of the three graft-fixation complexes did not compromise the conclusions from this study. A pilot study, in which a stiff wire was used instead of the DLSTG graft, determined that a reduction in stiffness of only 2% occurred when the stiffness of the wire-post-femur complex was tested before and after loading a wire-button-femur and wire-anchor-femur complex to failure. A review of the mechanisms of failure determined that damage to the femoral cortex during the failure test of the femur-button-DLSTG graft complex was not observed and was unlikely because the suture broke at a relatively low load, averaging 430 N. The failure mechanism of the anchor was attributable either to the device failing or the suture breaking and not from the cortical bone yielding. These observations indicate that sequential testing of the three fixation methods did not produce carry-over effects that would have altered the results.[11]

A post hoc study was performed to determine whether the use of hamstring tendons from older donors affected the results of this study. The tensile modulus (MPa), which normalizes for differences in cross-sectional area and length, was compared for hamstring tendons harvested from 10 younger (age 20 to 45 years; average, 29 years) and 10 older (age 46 to 67 years; average, 56 years) donors. No significant difference ($P = .175$) was observed between the modulus of the semitendinosus (903 ± 123) and gracilis (989 ± 164) tendon harvested from the younger donors compared with the modulus of the DLSTG graft (875 ± 122) harvested from the older donors. Because the age range of the hamstring tendons from the younger donors (age 20 to 45 years; average, 29 years) was not significantly different from the age range of patients undergoing ACL reconstruction (age 15 to 48 years; average, 33 years[15]) ($P = .308$), and the modulus was the same for tendons from younger and older donors, the results from this study can be applied to younger patients requiring an ACL reconstruction and to other studies.

Interpretation of Results

The stiffness of the DLSTG graft was greater than the native ACL and the bone-patellar tendon-bone graft (Fig 3). In our study, the stiffness of the DLSTG graft averaged 954 N/mm for a 60-mm long graft. For comparison the stiffness of relatively young, human femur-ACL-tibia complex ranges between 182 N/mm^{3} to 242 N/mm[4] to 306 N/mm.[5] The average stiffness of the central third bone–patellar tendon–bone is 455 N/mm for a 10-mm wide graft,[16] 685 N/mm for a 13.8-mm wide graft,[3] and 555 N/mm for a 15-mm wide graft.[16]

The stiffness was 6 to 15 times less than the stiffness of the native ACL (182 to 306 N/mm) when the DLSTG graft was attached to the femur using a suture bridge to link the graft to either the button or the anchor. Fixing the DLSTG graft to the femur using either of these two methods reduced the stiffness of the DLSTG graft by itself from 954 N/mm to 24 N/mm with the addition of the femoral fixation. Tightening or slippage of the square knot prior to failure was the primary determinant of the stiffness of the suture loop.
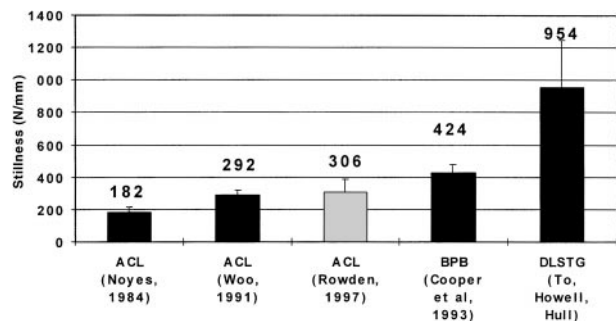

FIGURE 3. Stiffness comparison of the native ACL to a 10-mm wide BPB and double-looped semitendinosus and gracilis (DLSTG) graft. The stiffnesses of commonly used autogenous grafts, the bone–patellar tendon–bone (BPB) and DLSTG graft, are greater than the stiffness of the native ACL.

This explains why two suture loops with the same type of knot, but different lengths (i.e., 44 mm for the femur-button-DLSTG graft complex and 24 mm for the femur-anchor-DLSTG graft complex) had similar stiffness. Therefore, the knot in the suture, and neither the length of the suture nor the fixation device, determined the stiffness of the femur-button-DLSTG graft and femur-anchor-DLSTG graft complexes.

In contrast, the stiffness was comparable to the native ACL (182 to 306 N/mm) when the DLSTG graft was looped around a post inside the femoral tunnel and cancellous bone graft was compacted into the femoral tunnel. Fixing the DLSTG graft to the femur using the post and bone compaction reduced the stiffness of the DLSTG graft by itself from 954 N/mm to 224 N/mm with the addition of the femoral fixation. The mechanism by which compaction of cancellous bone increased the stiffness of the femur-fixation-DLSTG graft complex by 41 N/mm requires comment.

Based on the springs-in-series analysis the primary mechanism for increasing the stiffness of a complex consisting of a series of springs is to increase the stiffness of the spring that is least stiff (assuming a large disparity in stiffness). Of the two springs in the femur-post with bone graft-DLSTG graft complex, the fixation method (575 N/mm) was just 60% of the stiffness of the DLSTG graft (954 N/mm). Therefore, the increase in stiffness from bone compaction could not have resulted from shortening of the DLSTG graft because the DLSTG graft was already stiffer than the fixation method. The most likely mechanism for the increase in stiffness of the fixation complex from compaction of cancellous bone inside the femoral tunnel was that the bone provided additional support to the post that spanned the femoral tunnel. The supplementary bracing of the post by compacted bone can be expected to increase the stiffness of the complex by reducing the deflection of the post when a tensile load is applied to the DLSTG graft.

The increase in stiffness resulting from cancellous bone compaction may not be effective for all methods of graft fixation. Because the mechanism responsible for increasing the stiffness is not that of shortening the graft but of changing the stiffness of the fixation method, the ability of bone compaction to increase stiffness may be dependent on the type of fixation. Therefore, the findings of this study cannot be used to routinely recommend bone compaction as a technique to increase stiffness for all fixation methods. Additional studies are required to identify the fixation methods in which bone compaction can increase the stiffness of the fixation complex.

Because surgeons have many choices of femoral and tibial fixation methods that provide a wide variety of stiffness, it is worthwhile to discuss several principles for adjusting the stiffness of a ligament replacement at the time of implantation. These principles can be illustrated using the springs-in-series analysis and the known stiffness of the DLSTG graft and femoral and tibial fixation methods currently in use.[14,17] To justify the application of these principles to human ACL reconstruction, the stiffness of the femoral fixation methods in this study and the tibial fixations methods in another study[14] were measured in young human bone (i.e., average age 35 years). Because the DLSTG graft is at least 1.7 times stiffer than all of the tested femoral and tibial fixation methods,[14,17] the stiffness of a ligament replacement using a DLSTG graft will be determined by the stiffness of the fixation methods and not by the stiffness of the graft. These same principles can also be applied to ligament replacements using bone–patellar tendon–bone as long as the graft is stiffer than the femoral and tibial fixation methods.

If a surgeon chooses a femoral and tibial fixation method with substantially different stiffness and if the stiffnesses of both fixation methods are less than the stiffness of the graft, then the stiffness of the ligament replacement will be determined by that fixation method with the least stiffness. Therefore, replacing the fixation method that is the least stiff with another with higher stiffness is the only method to increase the stiffness of the ligament replacement.

If a surgeon chooses femoral and tibial fixation methods with equal stiffness and if the stiffness of both fixation methods is less than the stiffness of the graft, then the stiffness of the ligament replacement will be approximately one half the stiffness of the fixation method. In this situation, the most effective method to improve the stiffness of the ligament replacement would be to increase the stiffness of both fixation methods. Replacing only one fixation method with another that is more stiff will at most double the stiffness of the ligament replacement.

For a surgeon to increase the stiffness of the ligament replacement by either shortening the length of the graft or increasing the cross-sectional area of the graft, the stiffness of both the tibial and femoral fixation methods must be greater than the stiffness of the graft. Only when the stiffness of each of the fixation methods is greater than the stiffness of the graft can increasing the stiffness of the ligament replacement be affected by changing the graft structure.

According to our study, the technique of "anatomic fixation" cannot be expected to improve the stiffness of a ligament replacement unless methods of fixation are used that are as least as stiff as the graft. Morgan[18] recommended shortening the graft to match the length of the native ACL to improve stability. Anatomic fixation was thought to minimize the potential for creep failure that can occur within the tunnel as the points of graft fixation are moved away from the anatomic origin and insertion.

Morgan devised the bone-hamstring-bone technique because it theoretically combined the advantages of the high tensile strength of double-looped hamstring tendons and interference screw fixation of the bone–patellar tendon–bone graft.[18] Unfortunately, the clinical application of the bone-hamstring-bone technique was questioned because both the stiffness (average, 34 N/mm) and the pullout strength (average, 354 N) were too low in a study in porcine knees. The low stiffness of the complex was caused by the poor fixation stiffness provided by the interference screws that allowed the graft to slip at low loads at the site of the ligament anchorage.[19]

Similar to the bone-hamstring-bone technique, low stiffness has been measured for anatomic fixation using bioabsorbable and titanium soft-tissue interference screw fixation of hamstring tendons in bovine tibia. A study that evaluated the stiffness of interference screw fixation of a three-stranded semitendinosus graft in bovine tibia found that the stiffness of the fixation complex was four to six times less than the stiffness of the native ACL (182 to 306 N/mm). The reported stiffness was 39.7 ± 10.9 N/mm for fixation with a 7×25 mm round headed titanium interference screw (RCI; Smith & Nephew Donjoy, Carlsbad, CA) and 57.9 ± 13.8 N/mm for fixation with a 8×23 mm biodegradable interference screw (Sysorb; Sulzer Orthopaedics, Munsingen, Switzerland).[20] From these data it can be concluded that anatomic fixation of a hamstring graft at the articular surface of the tibia with either a titanium or biodegradable interference screw does not provide a stiffness that matches the native ACL at the time of implantation even though the length of the graft was shortened to 25 mm.[20] The springs-in-series analysis can be used to explain this finding because it is likely that the stiffness of the fixation with the interference screw was much less than the stiffness of the graft.

In contrast, another study showed that moving the site of tibial fixation of a bone–patellar tendon–bone graft more proximally in porcine knees had a signifi-cant improvement on the resulting anterior displacement and internal rotation of the tibia as well as the in situ forces of the graft.[1] However, these results are potentially misleading because two different fixation methods that may have had different stiffness were used. Proximal and central fixation within the tibial tunnel were achieved with an interference screw while the distal fixation required two staples because the bone plug extended outside the tibial tunnel. The authors made the assumption that the stiffness of the fixation provided by the interference screw and two staples were the same.

A significant difference in knee kinematics and in situ graft forces was not found between the two locations in which the interference screws were used. Instead, a significant difference in knee kinematics and in situ graft forces was found between the sites of fixation that used two different fixation methods. An alternative explanation, supported by the spring-in-series analysis and findings in our study, was that the difference in knee kinematics and in situ graft forces was caused by the staple fixation providing less stiffness than the interference screw. For the authors to have justified the suggestion that proximal graft fixation may provide the most acute stability of the reconstructed knee, a significant difference in knee kinematics and in situ graft forces should have been demonstrated between the two locations in which the stiffness of the fixation methods were the same (i.e., interference screws).

In conclusion, surgeons are familiar with the axiom that the weakest fixation device determines the strength of a ligament replacement at the time of implantation. In keeping with this principle, it is also correct to state that the least stiff fixation method determines the stiffness of the ligament replacement at the time of implantation. Because the stiffness varies widely for currently available femoral and tibial fixation methods, surgeons should understand that the stiffness of the ligament replacement at the time of implantation is determined by the fixation method, because the fixation methods are less stiff than the DLSTG graft. With currently available fixation methods, the best technique for matching the stiffness of the ligament replacement to the native ACL using a DLSTG graft is to increase the stiffness of the fixation methods and not by either shortening the graft or by increasing the cross-sectional area of the graft.

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