

INITIAL TENSION AND ANTERIOR LOAD-DISPLACEMENT BEHAVIOR OF HIGH-STIFFNESS ANTERIOR CRUCIATE LIGAMENT GRAFT CONSTRUCTS

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Background: Because the tension that exists in an anterior cruciate ligament graft when the knee is unloaded (the initial tension) affects the surgical outcome and because high initial tension has a number of adverse consequences, the primary purpose of this study was to determine quantitatively how much less initial tension was required for a high-stiffness construct than for a low-stiffness construct. A secondary purpose was to determine how the stiffness of the graft construct affects the anterior load-displacement behavior of the knee from 0° to 90° of flexion.

Methods: Anterior-posterior load-displacement was measured in each of ten intact cadaveric knee specimens, the anterior cruciate ligament was excised, and the anterior cruciate ligament was reconstructed with a double-loop bovine tendon graft. Graft constructs of different stiffness were created with use of six springs, ranging in stiffness from 25 to 275 N/mm to simulate the fixation stiffness. After adjusting the initial tension of the graft so that the anterior-posterior laxity of the reconstructed knee matched that of the intact knee, the 0-N posterior limit and the 225-N anterior limit were measured at 0°, 30°, 60°, and 90° of flexion.

Results: The highest stiffness fixation (275 N/mm) required an average of 73 N of initial tension, which was more than three times less than the average of 242 N of initial tension required by the lowest stiffness fixation (25 N/mm). The 225-N anterior limit was overconstrained an average of 1.0 mm with the highest stiffness fixation (275 N/mm), which was 3.6 mm less than the overconstraint with the lowest stiffness fixation (25 N/mm). Likewise, the posterior limit was overconstrained an average of 2.6 mm with the highest stiffness fixation (275 N/mm), which was 3.8 mm less than the overconstraint with the lowest stiffness fixation (25 N/mm).

Conclusions: The initial tension for a high-stiffness graft construct is more than three times less than that for a low-stiffness construct. The initial tension for a high-stiffness graft construct better restores both the 225-N anterior limit and the 0-N posterior limit to normal than the initial tension for a low-stiffness graft construct over the range of flexion from 0° to 90°.

Clinical Relevance: Because a high-stiffness graft construct requires substantially less initial tension than a low-stiffness graft construct, the tension pattern in a high-stiffness graft construct better matches the pattern in the intact anterior cruciate ligament. This tension pattern may avoid adverse consequences to both the knee joint function and the graft, which have been linked to high initial graft tension when the initial tension is maintained postoperatively. When the initial tension is not maintained postoperatively, a high-stiffness construct may be advantageous in avoiding a recurrence of knee instability.

It is important to determine how much initial tension should exist in an anterior cruciate ligament graft when the knee is unloaded because this tension affects the surgical outcome of the reconstructed knee¹⁻⁹. If the initial tension is too low, it adversely affects knee joint function as a result of excessive anterior laxity⁸. If the initial tension is too high, it

adversely affects knee joint function as a result of posterior subluxation of the tibia^{1,10-12} and inhibited knee extension¹. High initial tension also has been linked to adverse consequences to the graft, which include excessive graft wear at the femoral tunnel margin⁴, poor revascularization, myxoid degeneration, and inferior mechanical properties^{5,9}. Therefore,

it is important to identify factors that influence the initial tension. By judiciously controlling these factors, initial tension can be optimized so that knee stability is restored while the adverse consequences associated with high initial tension are avoided.

One factor affecting the initial tension required to restore knee stability to normal is the stiffness of the graft construct. At the time of surgery, the stiffness of the construct is controlled primarily by the stiffness of fixation because the fixation method is typically less stiff than the graft^{13,14}. Depending on the stiffness of the fixation method, the stiffness of a double-loop tendon graft-fixation complex can be varied tenfold at implantation (i.e., from 24 to 259 N/mm)^{13,14}. After surgery, the stiffness of graft constructs has not been determined, to our knowledge, but this stiffness is also expected to vary widely in the in vivo environment¹⁵. In either case, the amount of initial tension required to restore normal knee stability with a high-stiffness construct should be less than that with a low-stiffness construct. While this relationship between the stiffness of the construct and the initial tension can be appreciated intuitively, we found no studies that determined quantitatively the initial tension for a specific graft-construct stiffness and the stiffness that minimizes the initial tension while restoring the anterior load-displacement behavior to normal.

The first objective of our study was to determine the optimal initial tension carried by a double-loop anterior cruciate ligament graft in full extension such that motion of the tibia relative to the femur with 225 N of anterior force applied to the tibia (i.e., a 225-N anterior limit of motion) was normal at 30° of flexion for a range of graft-construct stiffnesses. A related second objective was to determine how well the 225-N anterior limit of motion at flexion angles other than 30° was maintained for different graft-construct stiffnesses. While the stability of the reconstructed knee is of primary interest as a dependent variable, the unloaded position of the knee also is of interest. This is because knee joint function is affected adversely as a result of posterior subluxation of the tibia as noted above. Thus, the third objective was to determine how the unloaded position of the tibia was affected by the graft-construct stiffness over flexion angles ranging from 0° to 90°.

Materials and Methods

Specimen Selection and Preparation

Ten fresh-frozen cadaveric knee specimens (average age of donors, sixty-five years; range, thirty-seven to seventy-five years) were obtained from tissue banks. The knee joints were evaluated radiographically and visually at the time of anterior cruciate ligament reconstruction. Only specimens with no evidence of degenerative arthritis were included in the study.

The specimens were prepared for experimentation by completely removing all tissue 50 mm proximal to and 50 mm distal to the joint line down to the bone. The bone was scraped to remove the periosteum. The fibula was fixed in its relative position with a screw anchored in the tibia. The fibula was

then cut off approximately 70 mm distal to the joint line. After reaming the medullary canals of the tibia and the femur until only cortical bone remained, steel rods that were 10, 11, or 12 mm in diameter were fixed in the canals with polymethylmethacrylate. The knee was wrapped in saline-solution-soaked gauze to prevent desiccation of the remaining tissues.

Determination of Anterior and Posterior Limits of the Intact Knee

Each knee was aligned, preconditioned, and tested in a load application system¹⁶. The load application system is a six-degree-of-freedom apparatus that can apply loads to the knee in all degrees of freedom and measure the corresponding displacements according to a joint coordinate system¹⁷. Flexion-extension is adjustable over the full physiologic range, and unconstrained motion is allowed in the remaining degrees of freedom. For this study, anterior-posterior force was applied and anterior-posterior displacement was measured (resolution, ± 0.1 mm). With use of the steel rods to interface the specimen to the load application system, each specimen was aligned with use of the functional axes method, which aligns the natural axes of joint motion with those of the load application system¹⁶. Once aligned, the shafts of the tibia and the femur were potted in aluminum tubes filled with polymethylmethacrylate, which were then clamped rigidly to the load application system. The knee was preconditioned by applying a 50-N stepwise load to 250 N in both the anterior and posterior directions to the tibia for five cycles at 0° and 90° of flexion. This preconditioning protocol produced a repeatable load-deflection pattern¹⁶. Zero degrees of flexion was defined as the position of the knee with an extension moment of 2.5 N-m¹⁸.

Motion of the intact knee was measured at 0°, 30°, 60°, and 90° of flexion in random order. The tibia was loaded in 15-N steps to incrementally increase the load from 0 to 45 N of anterior force, decrease the load to 0 N, increase the load from 0 to 45 N of posterior force, decrease the load to 0 N, and increase the load from 0 to 225 N of anterior force¹⁹. With use of a linearly variable differential transformer as a transducer to measure the anterior-posterior displacement of the tibia with respect to the femur, the 0-N posterior limit of motion was defined as the position of the tibia at 0-N force once the load on the tibia was decreased from 45 N of posterior force. The 225-N anterior limit of motion was defined as the position of the tibia at 225 N of anterior force.

Reconstruction of the Anterior Cruciate Ligament

The knee was removed from the load application system, and the joint was exposed with use of medial and lateral parapatellar incisions. The patella and patellar tendon were reflected distally, the joint was inspected for degenerative arthritis, and the anterior cruciate ligament was excised.

A double-loop graft was constructed from bovine extensor tendons with use of the same technique for preparing a double-loop semitendinosus and gracilis hamstring graft. A double-loop bovine tendon graft was used because it has

TABLE I Fixation Stiffness for Eighteen Combinations of Femoral and Tibial Fixation Methods and Corresponding Values of Initial Tension from Regression Model*

Stiffness of Femoral Fixation	Stiffness of Tibial Fixation†					
	No. 5 Sutures Tied to Post at 70 N/mm‡	Double Staples at 174 N/mm‡	One 20-mm Washer at 192 N/mm‡	Two Tandem Washers at 318 N/mm§	Metal Interference Screw at 340 N/mm§	WasherLoc at 506 N/mm§
Endo button at 24 N/mm	18 N/mm (286 N)	21 N/mm (265 N)	21 N/mm (265 N)	22 N/mm (259 N)	22 N/mm (254 N)	23 N/mm (254 N)
Mitek anchor at 26 N/mm	19 N/mm (278 N)	23 N/mm (254 N)	23 N/mm (254 N)	24 N/mm (249 N)	24 N/mm (249 N)	25 N/mm (244 N)
Bone mulch screw with bone compaction at 575 N/mm	62 N/mm (157 N)	134 N/mm (109 N)	144 N/mm (105 N)	205 N/mm (88 N)	214 N/mm (87 N)	269 N/mm (78 N)

*The fixation stiffness of these combinations ranged from 18 to 269 N/mm. Initial tensions ranged from a high of 286 N for the lowest stiffness combination to a low of 78 N for the highest stiffness combination. †The values for initial tension are given in parentheses. ‡Stiffness of fixation was determined with use of porcine tibia¹³. §Stiffness of fixation was determined with use of either human tibia¹³ or femur¹⁴ from donors with an average age of thirty-five years.

similar structural properties and is longer than a double-loop semitendinosus and gracilis graft²⁰. The added length of the double-loop bovine tendon graft ensured firm fixation in a freeze clamp, which is not always possible with a loop of gracilis tendon. The bovine tendons were harvested, muscle was removed, and excess tendon was trimmed so that two tendons when folded in half side-by-side fit snugly inside a 9-mm-diameter sizing cylinder (Sizing Sleeves; Arthrotek, Warsaw, Indiana). The free ends of each tendon were trimmed to a slight taper and were sewn with 1-0 suture with use of a crisscross stitch to facilitate passage of the graft in the knee^{21,22}.

To position the tibial tunnel, a 2.4-mm-diameter Kirschner wire was placed with use of a tibial drill-guide (Howell Tibial Guide; Arthrotek). The tibial drill-guide customized the placement of the Kirschner wire in the sagittal plane by accounting for variations in knee extension and slope of the intercondylar roof. This placement prevented the graft from impinging against the intercondylar roof in full extension²³⁻²⁶. The placement of the Kirschner wire was assessed with use of anterior-posterior and lateral radiographs of the knee in maximum extension before drilling the tibial tunnel. In the coronal plane, the placement of the Kirschner wire was considered acceptable when it met two radiographic criteria: (1) the Kirschner wire entered the intercondylar notch between the medial and lateral tibial eminences and (2) the Kirschner wire formed an angle of 70° with the articular surface of the medial tibial plateau. In the sagittal plane, the Kirschner wire was considered to have acceptable placement when it lay 4 to 5 mm posterior and parallel to the slope of the intercondylar roof²⁶. The tibial tunnel was drilled over the Kirschner wire with use of a 9-mm-diameter cannulated reamer.

The femoral tunnel was placed by inserting a 9-mm-diameter endoscopic femoral aimer (Size-Specific Femoral Aimer; Arthrotek) into the intercondylar notch through the tibial tunnel. The knee was flexed until the hook of the aimer locked into place in the over-the-top position, and a 2.4-mm-diameter

Kirschner wire was drilled into the femur. A 30-mm closed-end femoral tunnel was drilled with use of a 9-mm cannulated end reamer.

To allow the spring attached to the free end of the double-loop bovine tendon graft to simulate the combined stiffness of a femoral and tibial fixation method, the femoral fixation used in the specimen had to be much stiffer than the stiffest spring (275 N/mm). Accordingly, a special procedure was developed to create an ultra-high-stiffness femoral fixation. A 10-mm lateral-medial tunnel was positioned 25 mm inside the femoral tunnel with use of the U-Shaped Drill Guide (Arthrotek). A 4-mm-diameter steel rod was centered in the lateral-medial tunnel with a plug inserted in the femoral tunnel. The lateral-medial tunnel was then packed with polymethylmethacrylate and forced into the cavities of the trabecular bone with use of threaded end caps²⁷. Once the polymethylmethacrylate had hardened, the 4-mm-diameter rod was removed so that the femoral tunnel plug could be extracted. The femoral tunnel was cleared of polymethylmethacrylate with a curet. The rod was then reinserted in the cement mantle to form the femoral fixation post. The stiffness of the steel rod-cement-bone construct was conservatively estimated as 13,500 N/mm, and the corresponding deflection under a 225-N anterior force was 0.02 mm.

Rationale for Determining the Stiffness of the Springs

To vary the stiffness of the graft construct, six springs (25, 75, 125, 175, 225, and 275 N/mm) were selected as representing the distribution of the stiffness of different combinations of femoral and tibial fixation methods. The overall stiffness of eighteen different combinations of femoral and tibial fixation methods was calculated with use of available values for the stiffness of each femoral and tibial fixation method^{13,14} and a spring-in-series analysis¹⁴. The overall stiffness for these fixation combinations ranged from 18 to 269 N/mm (Table I). The graft construct stiffness corresponding to each spring was

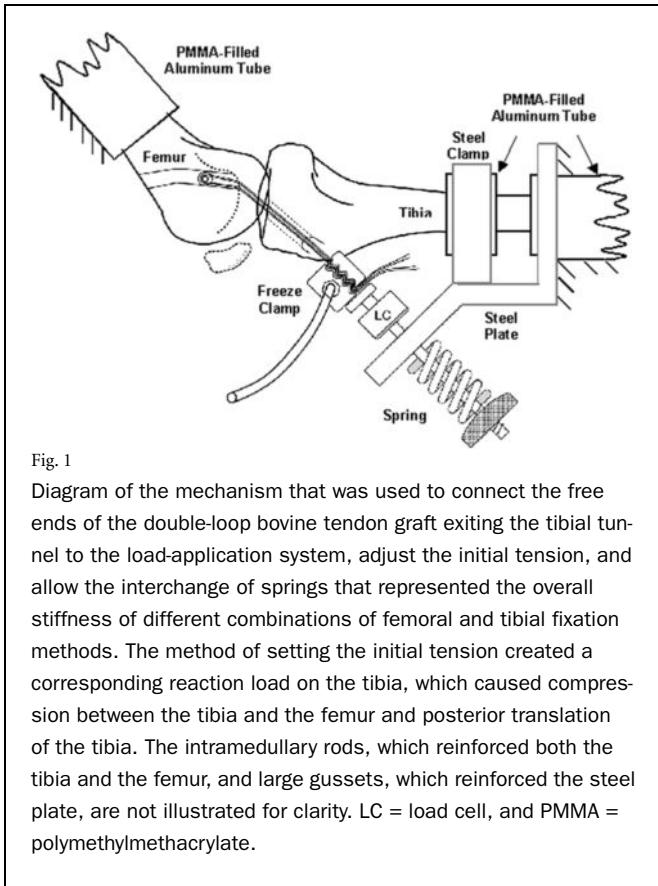


Fig. 1

Diagram of the mechanism that was used to connect the free ends of the double-loop bovine tendon graft exiting the tibial tunnel to the load-application system, adjust the initial tension, and allow the interchange of springs that represented the overall stiffness of different combinations of femoral and tibial fixation methods. The method of setting the initial tension created a corresponding reaction load on the tibia, which caused compression between the tibia and the femur and posterior translation of the tibia. The intramedullary rods, which reinforced both the tibia and the femur, and large gussets, which reinforced the steel plate, are not illustrated for clarity. LC = load cell, and PMMA = polymethylmethacrylate.

computed by including a 450-N/mm graft²⁰ in series with the spring and ranged from 24 to 171 N/mm.

Determination of Anterior and Posterior Limits of the Reconstructed Knee

A custom fixture that measured the graft tension and allowed the initial graft tension to be adjusted and the effective stiffness of the fixation method to be varied was added to the tibial unit of the load application system (Fig. 1). Upon exiting the tibial tunnel, the four limbs of the graft were gripped with a freeze clamp²⁸. A load-cell (Futek Advanced Sensor Technology, Irvine, California) attached to the freeze clamp measured the graft tension. To adjust the initial tension, a threaded shaft and knurled end cap were attached to the load-cell. The shaft passed through a spherical alignment bearing in a steel plate attached to the tibial unit of the load application system. A coil spring was sandwiched between the steel plate and the end cap that threaded onto the shaft so that the spring was compressed when the graft was in tension. Turning the knurled end cap allowed adjustment of the initial tension. When the end cap was removed from the shaft, a coil spring of a different stiffness could be installed. Because the tibia was clamped rigidly to the steel plate, which was bolted to the load application system that allowed unconstrained motion in five degrees of freedom, the initial tension created a compressive load between the tibia and the femur and also caused posterior translation of the tibia.

Following the reconstruction, the knee specimen was clamped in the load application system in the identical position to that for the intact specimen. After the graft was secured in the freeze clamp, an arbitrary initial tension of >250 N was applied to the graft with the knee at 0° of flexion with the stiffest spring installed. The reconstructed knee was subjected to the same preconditioning protocol that was used for the intact knee.

After the graft was preconditioned, the initial tension was set to 100 N at 0° of flexion and the knee was passively flexed to 120° while the graft tension was checked to ensure that it did not increase prematurely at an early flexion angle. A premature increase in tension is an indication that the graft is impinged by the intercondylar roof as a result of a femoral tunnel being placed too anteriorly²⁹. If a premature increase in tension was observed, then the specimen was removed from the study. This check was performed for the 25, 175, and 275 N/mm springs.

The reconstructed knee specimen was subjected to the same anterior-posterior-anterior loading cycle used for the intact knee. Once one of six springs (25, 75, 125, 175, 225, and 275 N/mm) was randomly selected, the initial tension was varied at 0° until the 225-N anterior limit matched within 0.5 mm that of the intact knee at 30° of flexion. When the loading cycle was completed, the knee was returned to 0°, the initial tension was adjusted as required to maintain the initial tension to the value that matched the 225-N anterior limit of the intact knee at 30° of flexion, a flexion angle from the remaining three was randomly selected, and the loading cycle was applied again. After all four flexion angles were completed, another spring was randomly selected and the procedure was repeated until measurements were obtained for all six springs.

Repeatability checks were performed systematically during the experiment to ensure that carryover effects were not important. For every other spring, the check included five anterior-posterior-anterior loading cycles at 0° with the initial tension for the spring that restored the 225-N anterior limit to that of the intact knee. Five cycles were chosen because the anterior-posterior load-displacement behavior within a spring was repeatable after that number. After all of the springs were tested, a repeatability check was again performed for the first spring with its appropriate initial tension.

Statistical Analysis

On the basis of a power law where the dependent variable was the average initial graft tension, nonlinear regression analysis was used to determine the optimal initial graft tension as a function of fixation stiffness and of graft-construct stiffness. A two-factor repeated analysis of variance was used to determine whether stiffness affected the difference in the 225-N anterior limit of motion between the reconstructed and intact knees. The two factors were the stiffness at six levels (25, 75, 125, 175, 225, and 275 N/mm) and the flexion angle at three levels (0°, 60°, and 90°). The same analysis of variance was used to determine whether stiffness affected the difference in the 0-N posterior limit of motion except that the flexion an-

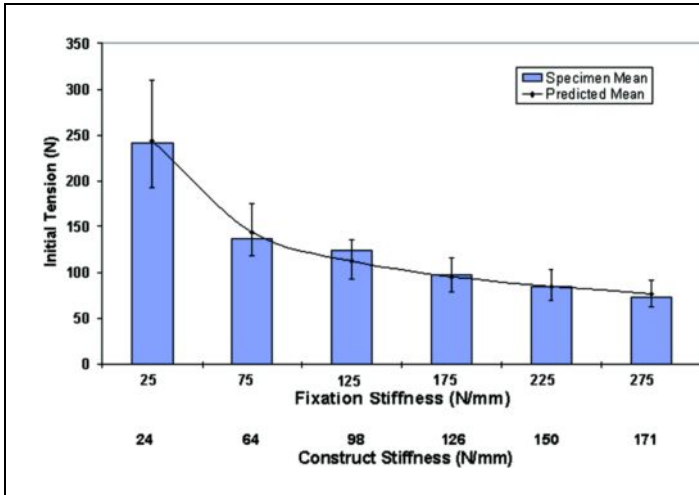


Fig. 2

The average value of the initial tension required to restore the 225-N anterior limit to normal at 30° of flexion for the different fixation and graft construct stiffnesses. The stiffness of the graft construct corresponding to each spring was calculated by incorporating a representative stiffness of 450 N/mm for a double-loop bovine tendon graft²⁰ in a springs-in-series analysis. The power law for the fixation stiffness is: initial tension = 1150(fixation stiffness)^{-0.48}. The power law for graft construct stiffness is: initial tension = 1547(graft construct stiffness)^{-0.58}. The best-fit regression model for the fixation stiffness is illustrated (R² = 0.982). Error bars represent the 95% confidence intervals.

gle was at four levels (0°, 30°, 60°, and 90°). Because the first analysis of variance did not reveal a significant interaction (p = 0.2768) and the second analysis of variance revealed a significant (p < 0.0001) but unimportant interaction, the differences in the 225-N anterior limit and the differences in the 0-N posterior limit were averaged over all flexion angles tested at each stiffness and a Tukey's test was performed comparing these averages for all pairs of stiffness. Significance was set at p < 0.05.

Results

As the fixation stiffness increased, the initial tension to restore the 225-N anterior limit to normal at 30° of flexion decreased so that the initial tension was lowest for the highest stiffness fixation (Fig. 2). The drop in the initial tension was

greatest when the fixation stiffness increased from 25 to 75 N/mm, and the drop steadily decreased as the fixation stiffness was increased incrementally by 50 N/mm. Overall, the highest stiffness fixation (275 N/mm) required an average of 73 N of initial tension, which was more than three times less than the average 242 N of initial tension required by the lowest stiffness fixation (25 N/mm). A power law was effective in describing the relationship between the average initial tension and the fixation stiffness and the average initial tension and graft-construct stiffness.

The fixation stiffness significantly affected the difference in the 225-N anterior limit from that of the intact knee (p < 0.0001), and the highest stiffness fixation restored the 225-N anterior limit closest to normal from 0° to 90° of flexion (Fig. 3). When averaged over the three flexion angles, the highest

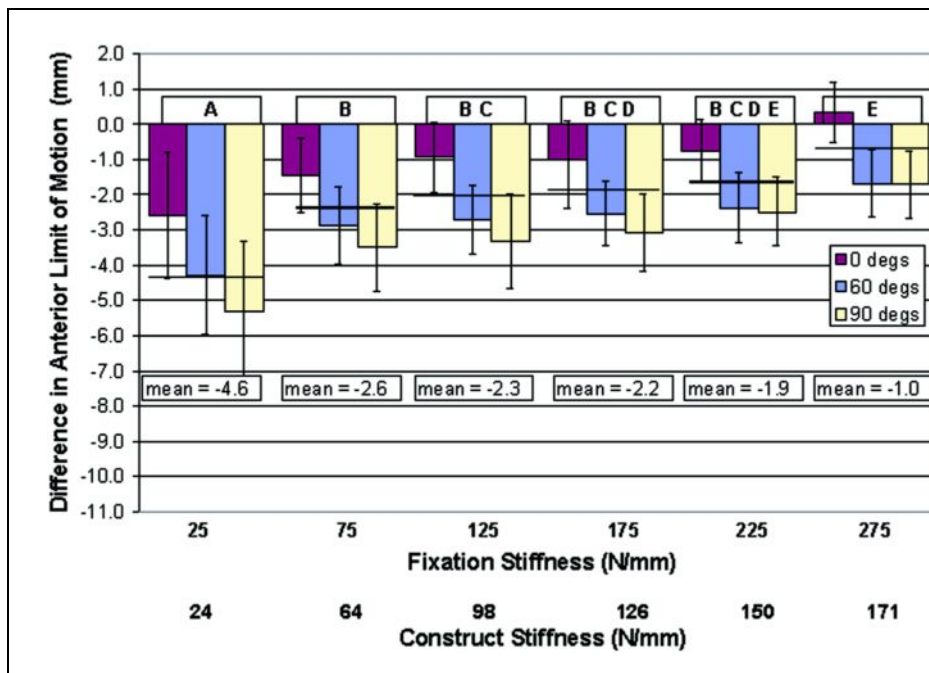
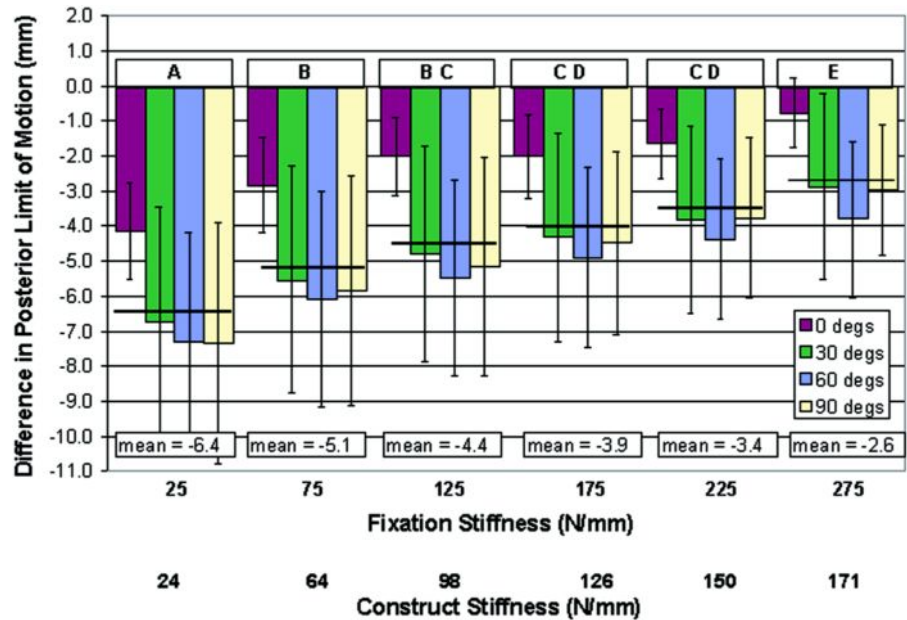


Fig. 3

The effect of stiffness on the difference in the anterior limit between the intact knee and the reconstructed knee. The heavy horizontal lines through each group of three flexion angles at each stiffness indicate the average of the pooled data for the three flexion angles at that stiffness. The letters at the top of each group of columns indicate the results of the Tukey's test on the average of the pooled data. Average differences in the anterior limit for groups of columns with different letters were significantly different (p < 0.05). Error bars represent the 95% confidence intervals.

Fig. 4

The effect of stiffness on the average difference in the posterior limit between the intact knee and the reconstructed knee. Heavy horizontal lines through each group of four flexion angles at each stiffness indicate the average of the pooled data for the four flexion angles at that stiffness. The letters at the top of each group of columns indicate the results of the Tukey's test on the average of the pooled data. Average differences in the posterior limit for groups of columns with different letters were significantly different ($p < 0.05$). Error bars represent the 95% confidence intervals.



stiffness fixation (275 N/mm) overconstrained the 225-N anterior limit by 1.0 mm, which was 3.6 mm less than the 4.6 mm of overconstraint of the 225-N anterior limit with the lowest stiffness fixation (25 N/mm).

The fixation stiffness also significantly affected the difference in the 0-N posterior limit from that of the intact knee ($p < 0.0001$), and the highest fixation stiffness restored the 0-N posterior limit closest to normal from 0° to 90° of flexion (Fig. 4). The highest stiffness fixation (275 N/mm) overconstrained the 0-N posterior limit by 2.6 mm, which was 3.8 mm less than the 6.4 mm of overconstraint of the 0-N posterior limit with the lowest stiffness fixation (25 N/mm).

Discussion

Because the initial tension is an important variable affecting surgical outcome and because high initial tension has a number of adverse consequences, the primary purpose of this study was to determine quantitatively how much less initial tension was required to restore stability to normal at 30° of flexion for a high-stiffness graft construct compared with a low-stiffness graft construct. A secondary purpose was to determine how the graft construct stiffness affected the anterior load-displacement behavior of the knee from 0° to 90° of flexion by examining the 225-N anterior limit and the 0-N posterior limit. The key findings of this study were that a high-stiffness construct required more than three times less initial tension than that of a low-stiffness construct, while at the same time it best restored both the 225-N anterior limit and 0-N posterior limit to normal.

Methodological Issues

The method used to set the initial tension provided a reaction load on the tibia, which may be an important practical consid-

eration in anterior cruciate ligament reconstructive surgery. For tibial fixation devices that require wrapping of either a suture or a tendon around a post, the tension applied by the surgeon manually is inherently reacted by the tibia. For tibial fixation devices that do not require wrapping of the graft, such as double staples, metal interference screws, and WasherLocs (Arthrotek), the manual application of tension does not create a reaction load on the tibia, provided that the tendons are not still attached to the tibia and that the manual pull is reacted solely by foot-ground reaction loads. However, a reaction load on the tibia can be created by means of fixtures manufactured and marketed commercially for this purpose; examples include the Intrafix (Mitek, Norwood, Massachusetts), the Graft Tensioner (Arthrotek), and the Tension Isometer (MEDmetric, San Diego, California).

In the present study, even though the initial tension was developed through the application of extra-articular tension, the extra-articular tension was representative of the initial tension or intra-articular tension. As demonstrated in an earlier study²², friction between either the graft bundles or between the graft bundles and the tibial tunnel does not create a substantial reduction in the intra-articular tension.

Although viscoelastic effects can change the initial tension in a graft through load relaxation^{20,30,31}, viscoelastic effects were eliminated as a confounding factor in the experimental design. The initial tension was always maintained so that the 225-N anterior limit matched that of the intact knee at 30° of flexion. Thus, the initial tension was the tension required to restore the 225-N anterior limit to that of the intact knee in the absence of viscoelastic effects. This was appropriate because the 225-N anterior limit of the intact knee at 30° of flexion was determined in the absence of viscoelastic effects. Recognizing that knee laxity in the clinical setting is usually

determined in the absence of viscoelastic effects, we designed the experiment to mimic the clinical setting.

The stiffness of the graft construct was changed by varying the stiffness of the fixation with use of a coil spring rather than an actual fixation method. The advantage of the use of a coil spring to simulate the stiffness was that the effects of this independent variable could be isolated for study. In addition to providing stiffness, actual fixation methods also allow varying degrees of lengthening in the region of fixation (e.g., slippage) particularly under repeated loading^{13,14,32}. Accordingly, use of actual fixation methods would have confounded the design of our study because any change in the load-deflection behavior could have been caused either by lengthening in the region of fixation or by the stiffness.

The range of graft construct stiffnesses considered is believed to be meaningful not only before biological incorporation of the graft in the bone tunnels but also after, when the fixation stiffness and/or graft stiffness can be expected to change¹⁵. Because a graft construct stiffness of <24 N/mm was believed to be unlikely following biological incorporation, only construct stiffnesses greater than this value were considered. Because the initial tension became increasingly insensitive to the graft construct stiffness as the stiffness increased, it was not necessary to test springs with stiffness higher than 275 N/mm.

Importance and Interpretation of Results

One important finding of the present study is that the fixation stiffness (and hence the graft construct stiffness), when varied over a practical range, profoundly affected the initial tension required to restore the 225-N anterior limit of motion to normal (see Fig. 2). The importance of this finding is that the 169-N greater initial tension for a low-stiffness construct compared with a high-stiffness construct is of sufficient magnitude to adversely affect both knee joint function and the graft.

One adverse effect upon knee joint function is an increased risk of an unstable knee, particularly during the early healing period before biological incorporation of the graft, because the fixation carries the majority of the graft force. High initial tension causes high graft forces to occur during both passive flexion as the knee is fully extended^{11,18} and when the knee is loaded¹³. During cyclic loading, fixation devices exhibit viscoplastic behavior, which is a permanent lengthening of the graft construct that increases with time^{13,32}. Because the viscoplastic effect is often greater for lower-stiffness fixation devices¹³ and because graft forces are higher with these fixation devices, the risk for permanent lengthening of the construct is increased.

Another adverse effect to knee joint function is that high initial tension causes the tibia to subluxate posteriorly on the femur^{1,10,12}, which in turn loads the posterior structures of the knee^{1,12}. Posterior subluxation decreases the moment arm of the patellar tendon, thus increasing the force that the quadriceps must produce to cause extension¹⁰. Also, a greater extensor moment is needed to reach full extension¹ because graft

tension increases as the knee is moved into full extension. These two effects combine to increase the quadriceps force required to actively extend the knee¹⁸.

An adverse effect to the graft is excessive wear at the entrance to the femoral tunnel⁴. In the study by Graf et al.⁴, increasing the initial tension by threefold decreased the fatigue life of patellar tendon grafts almost fourfold. This relative increase in initial tension is comparable with that determined in the present study for low-stiffness compared with high-stiffness graft constructs. Although the graft that we used was a different type from that used by Graf et al., the much higher initial tension for a low-stiffness construct may decrease the fatigue life of double-looped hamstring tendon grafts. Other adverse effects to the graft that have been linked to high initial tension include poor revascularization and myxoid degeneration that result in inferior mechanical properties^{5,9}.

The power law relation described in this study between the initial tension and graft construct stiffness (Fig. 2) is important because it demonstrates that the initial tension is not particularly sensitive to the construct stiffness as long as the construct stiffness is greater than approximately 125 N/mm. However, if the construct stiffness falls below approximately 100 N/mm, then the initial tension starts to increase dramatically. Thus, the power law may be useful in determining a practical lower limit on the construct stiffness where the initial tension can still be maintained close to the minimum value for the highest-stiffness graft construct.

Caution should be exercised in applying the power law to set the amount of initial tension applied to a graft intraoperatively, for several reasons. Additional factors affect the amount of tension carried by the graft immediately postoperatively. These factors include seating of the fixation device and settling in of the graft in the fixation device³³. Viscoelastic effects such as load relaxation^{20,30,31} can also affect the initial tension, though these effects will be recoverable for the most part provided that the initial tension is minimal in the mid-region of the flexion arc, from about 20° to 90°²⁰. This requirement will be satisfied by a high-stiffness fixation, which requires a relatively small amount of initial tension. As a result of these additional factors, the tension carried by the graft immediately postoperatively may not be directly indicated by the extra-articular tension.

Another reason for caution is that even if the tension carried by the graft immediately postoperatively is indicated by the extra-articular tension, the tension carried by the graft postoperatively may not be maintained. This is because lengthening of the graft construct can occur as a result of lengthening in the region of the fixation due to repeated loading from rehabilitative exercises³³ and/or lengthening in the region between the fixations due to remodeling of the graft in the in vivo environment^{34,35}. The potential for lengthening in the region of fixation due to repeated loading is reduced for high-stiffness constructs.

A final reason is that the stiffness of the graft construct can change in the in vivo environment. After the graft has been incorporated into the bone tunnels, the stiffness of fixa-

tion may either increase or decrease depending on the type of fixation device¹⁵. Thus, it is possible that a low-stiffness construct at the time of implantation may become a high-stiffness construct after biological incorporation of the graft. Thus, it might be advantageous to achieve a low initial tension intraoperatively rather than a high initial tension, assuming that the initial tension is maintained.

Another important finding of this study is that both the 225-N anterior limit and 0-N posterior limit were more closely restored to normal over the flexion range from 0° to 90° for a high-stiffness construct (Figs. 3 and 4). On the average, the 225-N anterior limit differed by 1.0 mm from normal while the 0-N posterior limit differed by 2.6 mm from normal with both differences being overconstrained.

Because both limits became more overconstrained as the stiffness decreased, the difference in anterior laxity from that of the intact knee was relatively constant for the various stiffnesses. Defining the anterior laxity as the anterior displacement that the tibia undergoes from the 0-N posterior limit to the 225-N anterior limit, the anterior laxity was 1.8 mm greater than that of the intact knee at the lowest stiffness and 1.6 mm greater than that of the intact knee at the highest stiffness. Therefore, even though a high stiffness produced both a 225-N anterior limit and a 0-N posterior limit closer to normal, this improvement in anterior load-displacement behavior, particularly for the 0-N posterior limit, may not be evident clinically with a knee arthrometer because this device measures anterior laxity and not the limits of motion^{36,37}.

In summary, the present study indicates that the initial tension is more than three times less for a high-stiffness construct than for a low-stiffness construct. Furthermore, both the anterior and posterior limits of motion are closer to normal for a high-stiffness construct than the limits achieved with a low-stiffness construct. When the initial tension is maintained, the clinical relevance is that a high-stiffness graft construct likely avoids many adverse consequences to both knee joint function and the graft that can result from high initial tension. When the initial tension is not maintained, the clinical relevance is that a high-stiffness construct likely will better prevent a recurrence of knee instability. ■

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