Empirical Relationship Between Lengthening an Anterior Cruciate Ligament Graft and Increases in Knee Anterior Laxity: A Human Cadaveric Study

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Lengthening of an anterior cruciate ligament (ACL) graft construct can occur as a result of lengthening at the sites of tibial and/or femoral fixation and manifests as an increase in anterior laxity. Although lengthening at the site of fixation has been measured for a variety of fixation devices, it is difficult to place these results in a clinical context because the mathematical relationship between lengthening of an ACL graft construct and anterior laxity is unknown. The purpose of our study was to determine empirically this relationship. Ten cadaveric knees were reconstructed with a double-looped tendon graft. With the knee in 25° of flexion, the position of the proximal end of the graft inside the femoral tunnel was adjusted by moving the femoral fixation device until the anterior laxity at an applied anterior force of 134 N *matched that of the intact knee. In random order, the graft construct was lengthened 1, 2, 3, 4, and* 5 mm *by moving the femoral fixation device distally along the femoral tunnel and anterior laxity was measured. The increase in the length of the graft construct was related to the increase in anterior laxity by a simple linear regression model. Lengthening the graft construct from 1 to* 5 mm *caused an equal increase in anterior laxity (slope* =1.0 mm/mm, *r^2* =0.800, *p* < 0.0001). *Because an anterior laxity increase of* 3 mm *or greater in a reconstructed knee is considered unstable clinically and because many fixation devices in widespread use clinically allow* 3 mm *or greater of lengthening in in vitro tests, our empirical relationship indicates that lengthening at the site of fixation probably is an important cause of knee instability following ACL reconstructive surgery. Our empirical relation also indicates that an important criterion in the design of future fixation devices is that lengthening at the sites of fixation in in vitro tests should be limited to less than* 3 mm. [DOI: 10.1115/1.2378931]

Introduction

Almost every clinical study has reported that some knees become unstable after anterior cruciate ligament (ACL) reconstruction, as indicated by the anterior laxity, which often increases over time postoperatively. The cause of the increase in anterior laxity often can be traced to an increase in the length of the graft construct (i.e., graft plus fixation devices). Lengthening of an ACL graft construct after reconstruction can result from lengthening at the sites of tibial and femoral fixation [1–4] and/or from lengthening of the graft between the sites of fixation [1,4–6].

Knowing the mathematical relationship between lengthening of an ACL graft construct and increased anterior laxity would be useful in evaluating structural properties of fixation devices. Arguably the most important structural property of a fixation method is its ability to resist lengthening at the site of fixation (e.g., slippage) under cyclic loading. Lengthening at the site of fixation has been measured for a variety of fixation devices through in vitro testing and varies from less than 1 mm to greater than 5 mm [3,4,7–10]. The amount of lengthening depends on the type of fixation device, the number of load cycles, the load of each cycle, the age of the bone, and the species of bone [3,4,7–10]. A 1 to 5 mm increase in graft construct lengthening might occur *in vivo* depending on the choice of fixation devices, the bone quality, and the aggressiveness of the rehabilitation program. Knowing the mathematical relationship between graft construct lengthening and increased anterior laxity would improve our understanding of the effect of lengthening at the site of fixation on the surgical outcome. Because the mathematical relationship between lengthening of an ACL graft construct and increases in anterior laxity has not been determined by any previous study to our knowledge, our purpose in the present study was to determine empirically this relationship.

Materials and Methods

Experiments. Ten cadaveric knees (average 72 range 56 to 89 years) were harvested and stored at −20°C. Radiographs and inspection of the knee at the time of ACL reconstruction did not reveal moderate or severe degenerative arthritis, chondrocalcinosis, or torn menisci. The intact knee was thawed overnight and prepared for testing using a previously described technique [11]. The knee was placed supine in a testing apparatus that allowed unconstrained motion of the tibia with respect to the femur (Fig. 1). An aluminum cylinder containing the femur was clamped in the femoral fixture, with the flexion-extension axis of the knee perpendicular to the sagittal plane. The length, ankle height, weight, and center of gravity of the shank-foot complex were sized to match that of an 81 kg, 180 cm male [12]. The knee was manually extended until resistance was felt, which defined full extension [13]. The flexion angle of the knee relative to full extension was measured with a goniometer (Stryker Howmedica, Mahwah, NJ, ±1° accuracy).

The anterior translation of the intact knee at 25° of flexion in response to a 134-N anterior force applied perpendicular to the longitudinal axis of the tibia was determined with a custom-made arthrometer [11] (Fig. 1). A custom-made arthrometer rather than a commercial arthrometer was used because the knee specimens did not include a full shank, as required for the attachment of a commercial arthrometer. The loading protocol of the custom-made arthrometer matched that of a commercial arthrometer (KT-1000, MEDmetric Corp., San Diego, CA). An 89-N posterior force was applied to the tibia three times and removed between each appli-
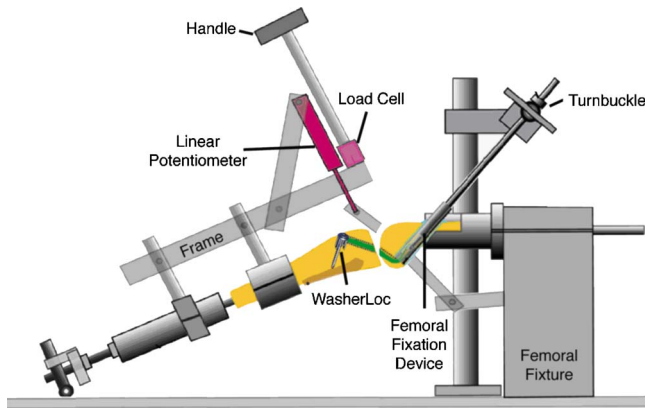
Fig. 1 Diagram of the reconstructed knee placed in the testing apparatus in 25° of flexion. The rigidly-fixed femur was aligned so that the flexion-extension axis is perpendicular to the sagittal (i.e., vertical) plane. The origin of the tibial coordinate system is the point where the longitudinal axis of the tibia (defined by the axis of the intramedullary rod) intersects a line that is perpendicular to the longitudinal axis of the tibia at the level of the joint line and that lies in the sagittal plane. The line lying in the sagittal plane and perpendicular to the longitudinal axis of the tibia at the level of the joint line is the A-P axis. The custom-made arthrometer was installed on the tibia with the load handle centered over the knee. Anterior and posterior forces were applied by pulling and pushing on the load handle. A load cell in series with the handle recorded the load. Anterior-posterior translation was measured with a linear potentiometer (ETI Systems, Carlsbad, CA) connected to a rigid arm, which was connected to the femoral fixture. The linear potentiometer was positioned above the tibia so that its axis was collinear with the A-P axis. The linear potentiometer was connected through spherical bearings at both ends so that it did not constrain the motion of the knee

cation. The neutral position of the tibia was the position of the tibia after the removal of the third posterior force. A 134-N anterior force was applied to the tibia. Anterior laxity was the difference in the position of the tibia between the 134-N anterior force and the neutral position.

The trabecular bone in the metaphysis of the cadaveric tibia was replaced with polyurethane foam to provide structural properties of the tibial fixation device (WasherLoc, Arthrotek Inc., Warsaw, IN) in specimens from elderly donors that are similar to those in young human tibia [14]. The ACL was excised and tibial and femoral drill holes were made using a previously described transtibial technique that positions the graft without roof impingement, without PCL impingement, and with a tension pattern during passive flexion-extension that is similar to that of the intact ACL [11,15]. The details of the technique have been described previously [11,15]. The tibial tunnel was drilled to 8 mm in diameter and serially dilated in 0.5 mm increments to 9 mm in diameter. An open-end femoral tunnel was drilled to 16 mm in diameter and a low-friction femoral bushing with an outer diameter of 16 mm and an inner diameter of 9 mm at its distal end was inserted into the femoral tunnel until flush with the intercondyler roof and fixed with bone cement [12]. The foam-reinforced knee was replaced in the testing apparatus.

Ten double-looped tendon grafts were made from bovine extensor tendons [14]. These grafts have structural properties similar to those of a young human double-looped semitendinosus and gracilis graft [5]. The tendons were trimmed until when looped over a suture they passed snugly through a 9 mm diameter cylinder and not through an 8 mm diameter cylinder (Sizing Sleeve, Arthrotek, Inc). Four centimeters of the end of each strand were whip-stitched using a No. 1, braided, absorbable suture (Polysorb, United States Surgical/Syneture, Norwalk, CT) [14]. A fresh

double-looped tendon graft was selected at random for each specimen and was looped over a crossbar of the femoral fixation device (stainless steel, 2.4 mm diameter, 10 mm length). The femoral fixation device was inserted into the femoral tunnel, and connected to a turnbuckle that was connected to the baseplate of the testing apparatus. The turnbuckle was adjusted until the crossbar was positioned 25 mm inside the femoral tunnel [12] (Fig. 1).

The graft was fixed to the foam-reinforced tibia with use of a WasherLoc. The WasherLoc was used because it slips minimally under cyclic load [3,4,7]. A 17 mm diameter counter bore was drilled into the distal end of the tibial tunnel to recess the WasherLoc (16 mm diameter WasherLoc, Arthrotek, Warsaw, IN). The sutures from the two strands of each tendon were passed from proximal to distal through the tibial tunnel. The knee was placed in full extension. Two tibial load cells (225 N, SM-50, Interface, Scottsdale, AZ), each connected to pneumatic cylinder (Illinois Pneumatics, Rosco, IL), were used to apply equal tensile force to each strand of the graft in line with the tibial tunnel. The graft was tensioned to 110 ± 1 N and fixed in the counterbore with the WasherLoc and a 6.5 mm diameter cancellous screw that purchased the lateral cortex of the tibia.

Three cyclic loading treatments were applied in succession to precondition both the knee and graft. Each cyclic loading treatment consisted of flexing the knee to 25° and cyclically loading the knee 20 times between a posterior force of 26 N and an anterior force of 100 N. A flexion angle of 25° was selected because this flexion angle is commonly used when testing knee anterior stability clinically using commercially available knee arthrometers (e.g., KT-1000, MEDmetric Corp., San Diego, CA). Thus, the increases in anterior laxity caused by lengthening of the graft construct would be meaningful clinically. The anterior force of 100 N generated an intra-articular tension of 170 N in the graft, which is similar to the tension in the ACL during level walking [16]. The position of the proximal end of the graft was adjusted with the turnbuckle to restore the anterior translation to within ± 0.1 mm of the intact knee at an applied anterior force of 134 N.

To determine the effect of increasing the length of the double-looped tendon graft construct on the anterior laxity, the graft was lengthened within the femoral bushing by 1, 2, 3, 4, or 5 mm in random order using the turnbuckle. Rotating the turnbuckle moved the femoral fixation device distally inside the femoral tunnel (accuracy of ± 0.06 mm), thus lengthening the graft. Anterior laxity was measured after each graft lengthening at an applied anterior force of 134 N.
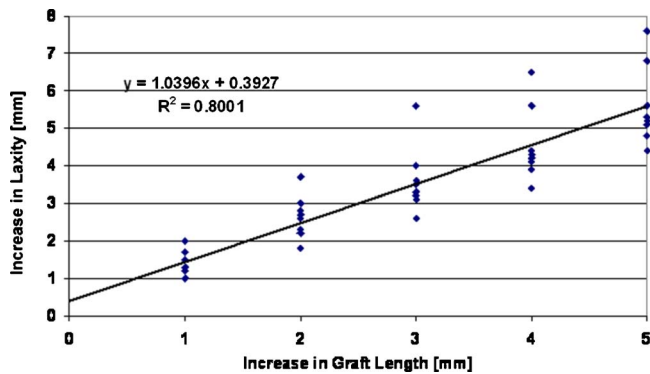
Statistical Analysis. A simple linear regression was used to determine the empirical relationship between lengthening the graft construct and the increase in anterior laxity. The intercept and slope, the coefficient of determination, and the level of significance of both the intercept and slope were calculated with commercially available software (SAS, release 8.0, SAS Institute, Cary, NC). Because the graft construct was lengthened 5 mm in eight specimens and 4 mm in two specimens, 48 data points were available for the linear regression (Fig. 2).

Results

The increase in the anterior laxity was linearly related to the increase in length of the graft construct (Fig. 2). The slope of the regression line was 1.0 mm/mm ($R^2 = 0.800$, $p < 0.0001$), indicating that a 1.0 mm increase in graft construct length caused an equal increase in anterior laxity. The y intercept of 0.4 mm was not significantly different from zero ($p = 0.1191$) and was within the acceptable tolerance (± 0.5 mm) of matching the anterior laxity to that of the intact knee.

Discussion

Because lengthening of an ACL graft construct causes an increase in anterior knee laxity and because the mathematical rela-

$$y = 1.0396x + 0.3927$$
$$R^2 = 0.8001$$

Fig. 2 Graph of the simple linear regression analysis showing that lengthening the graft 1 to 5 mm causes an equal millimeter-for-millimeter increase in anterior laxity

tionship between lengthening of a graft construct and increases in laxity is unknown, the objective of our study was to determine an empirical relationship for lengthening of a double-looped tendon graft in the range 1–5 mm. The most important result from our study is that lengthening of a double-looped tendon graft from 1 to 5 mm causes an equal increase in anterior laxity. We would like to discuss how this relationship might be used to improve our clinical understanding of the effect of structural properties of fixation devices on anterior laxity.

In vitro studies of soft tissue fixation devices have focused on determining the amount of lengthening at the site of fixation as an important structural property indicating a fixation device's performance. Surgeons that prefer the use of soft tissue ACL grafts such as the double-looped semitendinosus and gracilis autograft and/or the single-looped tibialis allograft recognize the importance of minimizing lengthening at the sites of fixation [2,3,9,10,17]. Minimizing lengthening is especially challenging for these two types of soft tissue grafts because the tendon heals more slowly to the tunnel wall compared to a bone plug graft [18,19] particularly in the tibia [20] and because the function of the knee is more quickly gained as a result of the increased time associated with healing of the patellar tendon. Thus, lengthening is more likely to occur in a knee reconstructed with a soft tissue graft than with a bone-patellar tendon-bone graft [2]. For these reasons, fixation devices used with a soft tissue graft must have better structural properties than fixation devices used with a bone-patellar tendon-bone graft [18].

Trying to sort out which fixation devices have an excessive amount of lengthening is problematic because of differences in the age of the bone, species of bone, and cyclic loading protocols between studies. The material used in a study should be selected so that it provides structural properties of fixation devices that are representative of those in young human bone. Material considerations suggest that skeletally mature porcine and bovine bone should not be used because the amount of lengthening is underestimated, especially for fixation devices that purchase cancellous bone (i.e., interference screws, Intrafix) [3,17]. Similarly, cadaveric bone from elderly donors should not be used because the amount of lengthening is overestimated [14]. This leaves as bone options cadaveric tibia from elderly donors where the trabecular bone has been replaced with foam [14], bone from calf tibia (age 16–24 months) [21–23], and bone from young human donors, which is difficult to obtain [3].

The methodology of a study should be designed to provide a conservative or worst-case measurement of lengthening at the site of fixation. Thus methodology considerations suggest that a conservative or worst-case cyclic loading protocol requires a peak force that either matches or exceeds the tensile force in the intact

ACL during level walking (i.e., ≥ 170 N), and that there is a sufficient number of cycles such that lengthening either stabilizes or fixation fails [24].

Lengthening at the site of fixation for a variety of devices was reviewed with these testing criteria in mind. Unfortunately, few of the published studies met all the testing criteria of (1) the use of either young human bone, foam bone, or calf bone, (2) the use of a loading protocol to at least 170 N in the graft, (3) and the use of a sufficient number of cycles such that lengthening either stabilized or fixation failed. Fixation devices that allowed the graft to lengthen 3 mm or greater in any type of bone other than elderly human cadaveric bone were classified as "excessive" based on the assumption that 3 mm of lengthening renders the reconstructed knee unstable [25–28]. On the femoral side, lengthening with the RigidFix (6.0 mm), and interference screw (5.4 mm) was excessive in bovine tibia [8], and lengthening with the closed loop endobutton (3.9 mm), RigidFix (3.7 mm) and three types of interference screws (range 3.2 to 4.0 mm) was excessive in porcine tibia [10]. On the tibial side, sutures tied to a post (4.9 mm), double staples (3.3 mm), and a single-spiked washer (3.5 mm) was excessive in porcine tibia, an interference screw (3.7 mm, 4.0 mm) was excessive in young human tibia [3,17], and a clawed washer (6.7 mm) was excessive in bovine tibia [2]. Considering the large number of fixation devices that allow lengthening greater than 3 mm *in vitro* and the widespread use of these devices clinically, lengthening at the site of fixation probably is an important cause of knee instability following ACL reconstructive surgery.

Our finding that an increase in graft length causes an equal increase in anterior laxity at 25° of knee flexion is interesting physically. Because of the steep angle that the graft makes with the tibial plateau in both the sagittal and coronal planes (approximately 70° in each plane), it might be expected intuitively that the increase in anterior laxity would be greater than the increase in graft length. However when anterior force is applied to the tibia, the graft tension increases by an amount approximately equal to the anterior force for high stiffness fixations such as those used in this study [29]. The tension increase causes a corresponding increase in the compressive force on the tibial plateau, which in turn increases the resistance of the knee to anterior translation [30–32]. This may explain the one-to-one relationship between an increase in the graft length and an increase in anterior laxity found in our study.

To evaluate the effect of graft lengthening on knee kinematics, we measured the A-P load displacement of the knee under the application of anterior tibial force. While it is well known that the ACL is a primary restraint to anterior force applied to the tibia, it is also effective in resisting internal tibial torque particularly near full extension [33]. Therefore, it may also be of interest to assess how graft lengthening affects the response of the knee to rotational loads.

In summary, the empirical relationship reported in this study is useful in the evaluation of structural properties of fixation devices, particularly lengthening at the site of fixation after cyclic loading. In evaluating current fixation devices, our empirical relationship indicates that lengthening at the site of fixation likely is an important cause of knee instability following ACL reconstructive surgery. Also, our empirical relation suggests that an important criterion in the design of future fixation devices is to limit the lengthening at the sites of fixation in *in vitro* tests to less than 3 mm.

References

[1] Cuppone, M., and Seedhom, B. B., 2001, "Effect of Implant Lengthening and Mode of Fixation on Knee Laxity After ACL Reconstruction With an Artificial

Ligament: A Cadaveric Study," J. Orthop. Sci., **6**, pp. 253–261.

[2] Giurea, M., Zorilla, P., Amis, A. A., and Aichroth, P., 1999, "Comparative Pull-out and Cyclic-Loading Strength Tests of Anchorage of Hamstring Tendon Grafts in Anterior Cruciate Ligament Reconstruction," Am. J. Sports Med., **27**, pp. 621–625.

[3] Magen, H. E., Howell, S. M., and Hull, M. L., 1999, "Structural Properties of Six Tibial Fixation Methods for Anterior Cruciate Ligament Soft Tissue Grafts," Am. J. Sports Med., **27**, pp. 35–43.

[4] Roos, P. J., Hull, M. L., and Howell, S. M., 2004, "Lengthening of Double-Looped Tendon Graft Constructs in Three Regions After Cyclic Loading: A Study Using Roentgen Stereophotogrammetric Analysis," J. Orthop. Sci., **22**, pp. 839–846.

[5] Haut Donahue, T. L., Gregersen, C., Hull, M. L., and Howell, S. M., 2001, "Comparison of Viscoelastic, Structural, and Material Properties of Double-Looped Anterior Cruciate Ligament Grafts Made From Bovine Digital Extensor and Human Hamstring Tendons," ASME J. Biomech. Eng., **123**, pp. 162–169.

[6] Boorman, R. S., Thornton, G. M., Shrive, N. G., and Frank, C. B., 2002, "Ligament Grafts Become More Susceptible to Creep Within Days After Surgery: Evidence for Early Enzymatic Degradation of a Ligament Graft in a Rabbit Model," Acta Orthop. Scand., **73**, pp. 568–574.

[7] Coleridge, S. D., and Amis, A. A., 2004, "A Comparison of Five Tibial-Fixation Systems in Hamstring-Graft Anterior Cruciate Ligament Reconstruction," Knee Surg. Sports Traumatol. Arthrosc., **12**, pp. 391–397.

[8] Ahmad, C. S., Gardner, T. R., Groh, M., Arnouk, J., and Levine, W. N., 2004, "Mechanical Properties of Soft Tissue Femoral Fixation Devices for Anterior Cruciate Ligament Reconstruction," Am. J. Sports Med., **32**, pp. 635–640.

[9] Kousa, P., Jarvinen, T. L. N., Vihavainen, M., Kannus, P., and Jarvinen, M., 2003, "The Fixation Strength of Six Hamstring Tendon Graft Fixation Devices in Anterior Cruciate Ligament Reconstruction—Part II: Tibial Site," Am. J. Sports Med., **31**, pp. 182–188.

[10] Kousa, P., Jarvinen, T. L. N., Vihavainen, M., Kannus, P., and Jarvinen, M., 2003, "The Fixation Strength of Six Hamstring Tendon Graft Fixation Devices in Anterior Cruciate Ligament Reconstruction—Part I: Femoral Site," Am. J. Sports Med., **31**, pp. 174–181.

[11] Grover, D. M., Howell, S. M., and Hull, M. L., 2005, "Early Tension Loss in an Anterior Cruciate Ligament Graft. A Cadaver Study of Four Tibial Fixation Devices," J. Bone Jt. Surg., Am. Vol., **87-A**, pp. 381–390.

[12] Grover, D. M., 2004, "There is an Early Tension Loss in a Tendon ACL Graft That Increases the Maximal Anterior Translation: A Cadaveric Study of Four Tibial Fixation Devices," MS thesis, Biomedical Engineering, University of California, Davis, CA.

[13] Melby, A., III, Noble, J. S., Askew, M. J., Boom, A. A., and Hurst, F. W., 1991, "The Effects of Graft Tensioning on the Laxity and Kinematics of the Anterior Cruciate Ligament Reconstructed Knee," Arthroscopy: J. Relat. Surg., **7**, pp. 257–266.

[14] Bailey, S. B., Grover, D. M., Howell, S. M., and Hull, M. L., 2004, "Foam-Reinforced Elderly Human Tibia Approximates Young Human Tibia Better Than Porcine Tibia—A Study of the Structural Properties of Three Soft Tissue Fixation Devices," Am. J. Sports Med., **32**, pp. 755–764.

[15] Simmons, R., Howell, S. M., and Hull, M. L., 2003, "Effect of the Angle of the Femoral and Tibial Tunnels in the Coronal Plane and Incremental Excision of the Posterior Cruciate Ligament on Tension of an Anterior Cruciate Ligament Graft: An in Vitro Study," J. Bone Jt. Surg., Am. Vol., **85-A**, pp. 1018–1029.

[16] Morrison, J. B., 1969, "Function of the Knee Joint in Various Activities," Biomed. Eng., **4**, pp. 573–580.

[17] Nurmi, J. T., Sievanen, H., Kannus, P., Jarvinen, M., and Jarvinen, T. L. N., 2004, "Porcine Tibia is a Poor Substitute for Human Cadaver Tibia for Evaluating Interference Screw Fixation," Am. J. Sports Med., **32**, pp. 765–771.

[18] Tomita, F., Yasuda, K., Mikami, S., Sakai, T., Yamazaki, S., and Tohyama, H., 2001, "Comparisons of Intraosseous Graft Healing Between the Doubled Flexor Tendon Graft and the Bone-Patellar Tendon-Bone Graft in Anterior Cruciate Ligament Reconstruction," Arthroscopy: J. Relat. Surg., **17**, pp. 461–476.

[19] Kyung, H. S., Kim, S. Y., Oh, C. W., and Kim, S. J., 2003, "Tendon-to-Bone Tunnel Healing in a Rabbit Model: The Effect of Periosteum Augmentation at the Tendon-to-Bone Interface," Knee Surg. Sports Traumatol. Arthrosc, **11**, pp. 9–15.

[20] Grassman, S. R., McDonald, D. B., Thornton, G. M., Shrive, N. G., and Frank, C. B., 2002, "Early Healing Processes of Free Tendon Grafts Within Bone Tunnels is Bone-Specific: A Morphological Study in a Rabbit Model," The Knee, **9**, pp. 21–26.

[21] Hoffmann, R. F., Peine, R., Bail, H. J., Sudkamp, N. P., and Weiler, A., 1999, "Initial Fixation Strength of Modified Patellar Tendon Grafts for Anatomic Fixation in Anterior Cruciate Ligament Reconstruction," Arthroscopy: J. Relat. Surg., **15**, pp. 392–399.

[22] Weiler, A., Hoffmann, R. F., Siepe, C. J., Kolbeck, S. F., and Sudkamp, N. P., 2000, "The Influence of Screw Geometry on Hamstring Tendon Interference Fit Fixation," Am. J. Sports Med., **28**, pp. 356–359.

[23] Weiler, A., Hoffmann, R. F., Stahelin, A. C., Bail, H. J., Siepe, C. J., and Sudkamp, N. P., 1998, "Hamstring Tendon Fixation Using Interference Screws: A Biomechanical Study in Calf Tibial Bone," Arthroscopy: J. Relat. Surg., **14**, pp. 29–37.

[24] Roos, P. J., Hull, M. L., and Howell, S. M., 2004, "How Cyclic Loading Affects the Migration of Radio-Opaque Markers Attached to Tendon Grafts Using a New Method: A Study Using Roentgen Stereophotogrammetric Analysis (RSA)," ASME J. Biomech. Eng., **126**, pp. 62–69.

[25] Howell, S. M., and Taylor, M. A., 1996, "Brace-Free Rehabilitation With Early Return to Activity for Knees Reconstructed With a Double-Looped Semitendinosus and Gracilis Graft," J. Bone Jt. Surg., Am. Vol., **78**, pp. 814–825.

[26] Howell, S. M., and Deutsch, M. L., 1999, "Comparison of Endoscopic and Two-Incision Techniques for Reconstructing a Torn Anterior Cruciate Ligament Using Hamstring Tendons," Arthroscopy: J. Relat. Surg., **15**, pp. 594–606.

[27] O'Neill, D. B., 1996, "Arthroscopically Assisted Reconstruction of the Anterior Cruciate Ligament. A Prospective Randomized Analysis of Three Techniques," J. Bone Jt. Surg., Am. Vol., **78**, pp. 803–813.

[28] Yasuda, K., Ohkoshi, Y., Tanabe, Y., and Kaneda, K., 1992, "Quantitative Evaluation of Knee Instability and Muscle Strength After Anterior Cruciate Ligament Reconstruction Using Patellar and Quadriceps Tendon," Am. J. Sports Med., **20**, pp. 471–475.

[29] Eagar, P., Hull, M. L., and Howell, S. M., 2004, "How the Fixation Method Stiffness and Initial Tension Affect Anterior Load-Displacement of the Knee and Tension in Anterior Cruciate Ligament Grafts: A Study in Cadaveric Knees Using a Double-Loop Hamstrings Graft," J. Orthop. Res., **22**, pp. 613–624.

[30] Ahmed, A. M., Hyder, A., Burke, D. L., and Chan, K. H., 1987, "In-Vitro Ligament Tension Pattern in the Flexed Knee in Passive Loading," J. Orthop. Res., **5**, pp. 217–230.

[31] Markolf, K. L., Bargar, W. L., Shoemaker, S. C., and Amstutz, H. C., 1981, "The Role of Joint Load in Knee Stability," J. Bone Jt. Surg., Am. Vol., **63**, pp. 570–585.

[32] Markolf, K. L., Burchfield, D. M., Shapiro, M. M., Davis, B. R., Finerman, G. A., and Slauterbeck, J. L., 1996, "Biomechanical Consequences of Replacement of the Anterior Cruciate Ligament With a Patellar Ligament Allograft. Part I: Insertion of the Graft and Anterior-Posterior Testing," J. Bone Jt. Surg., Am. Vol., **78**, pp. 1720–1727.

[33] Hame, S. L., Oakes, D. A., and Markolf, K. L., 2002, "Injury to the Anterior Cruciate Ligament During Alpine Skiing: A Biomechanical Analysis of Tibial Torque and Knee Flexion Angle," Am. J. Sports Med., **30**, pp. 537–540.