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The effect of graft stiffness on knee joint biomechanics after ACL reconstruction—a 3D computational simulation

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Abstract

Objective. The objective was to determine the effect of varying graft stiffness and initial graft tension on knee kinematics and graft tension after anterior cruciate ligament reconstruction.

Design. A 3D computational knee model was used.

Background. Many factors influencing the biomechanical outcome of anterior cruciate ligament reconstruction have been investigated. However, there are no reports on the effect of variations in graft stiffness on knee behavior.

Methods. A 3D computational knee model was used to simulate anterior cruciate ligament reconstruction using three different grafts with stiffnesses similar to the anterior cruciate ligament (graft 1), a 10mm bone-patellar tendon-bone graft (graft 2), and a 14mm bone-patellar tendon-bone graft (graft 3). The initial graft tension was set to 0 or 40 N with the knee at 30° of flexion. A 134 N anterior tibial drawer load and a 400 N quadriceps load were applied to the knee, and kinematics and graft tension were calculated.

Results. When fixed with no initial tension, graft 1 was found to under-constrain the knee, while graft 2 slightly over-constrained the knee, and graft 3 over-constrained the knee when compared to the intact knee. When an initial graft tension of 40 N was used, all of the reconstructed knees were more constrained than when an initial tension of 0 N was used.

Conclusions. This study suggests that graft stiffness has a direct impact on knee biomechanics after anterior cruciate ligament reconstruction. An optimal anterior cruciate ligament reconstruction can be achieved if the anterior cruciate ligament is replaced by a graft with similar structural stiffness.

Relevance

This study showed that if the graft material and fixation sites are selected such that the anterior cruciate ligament structural stiffness is retained, normal knee kinematics can be restored.

Keywords: Anterior cruciate ligament; Knee; Biomechanics; Computational model; Kinematics

1. Introduction

Reconstruction of the anterior cruciate ligament (ACL) has been widely used to restore anterior knee stability after injury to the ACL. Currently, ACL reconstruction is most often performed using a bone-patellar tendon-bone graft or a hamstring tendon graft (Frank and Jackson, 1997). While many clinical studies have shown that ACL reconstruction using these grafts is successful in restoring anterior knee stability (Marder et al., 1991; O'Neill, 2001; Aglietti et al., 1994), recent long-term clinical studies have reported an increased rate of radiographic arthrosis in ACL reconstructed knees (O'Neill, 2001), even when accounting for pre-existing chondral or meniscal injuries (Jomha et al., 1999). Abnormal joint kinematics after ACL reconstruction has been proposed as one possible etiology for this late degeneration (Buckwalter and Lane, 1997).

The two most commonly used grafts for ACL reconstruction, the bone-patellar tendon-bone graft (BPTB) graft and the quadrupled semitendinosus and gracilis (QST/G) graft, are two to four times stiffer than

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the intact ACL (Butler et al., 1986; Hamner et al., 1999; Muellner et al., 1998; Noyes et al., 1984; Cooper et al., 1993). A recent in vitro biomechanical study by Shoemaker et al. (1993) has reported a change in rotational kinematics of the tibia after ACL reconstruction using BPTB grafts. Tashman and Anderst (2001) have reported that the ACL reconstructed knee is externally rotated in comparison to the contralateral, uninjured limb under in vivo dynamic loading conditions. The behavior of the reconstructed knee depends on several variables (Jomha et al., 1999; Aglietti et al., 1997; Black et al., 2000; Markolf et al., 1996a,b; Breitfuss et al., 1996; Rowden et al., 1997; Woo et al., 1997; Yasuda et al., 1997; Kurosaka et al., 1987; Magen et al., 1999). There have been many investigations into the effect of variations in initial graft tension (Yasuda et al., 1997; Bylski-Austrow et al., 1990; Fleming et al., 1992; Burks and Leland, 1988), graft placement (Bylski-Austrow et al., 1990; Hefzy and Grood, 1986), fixation technique (Rowden et al., 1997; Kurosaka et al., 1987; Magen et al., 1999), and graft selection (Aglietti et al., 1997; Woo et al., 1997) on anterior knee stability. However, there is little data reported on the effect of variations in graft stiffness on the kinematics of the ACL reconstructed knee. ACL reconstruction using a graft with supraphysiologic stiffness may restrict, or over-constrain, the motion of the knee potentially increasing joint contact stresses when compared to the intact knee.

In this study, we hypothesized that ACL reconstruction using a graft of increased stiffness would result in over-constraint of the knee. A 3D computational knee model was used to assess the effect of variations in graft stiffness and initial graft tension on the biomechanics of an ACL reconstructed knee. The model was placed under an anterior tibial drawer load and a quadriceps load at multiple flexion angles. The knee kinematics and ACL/graft tension were determined with the knee in intact, ACL-deficient, and ACL-reconstructed states. Graft stiffnesses were selected to model the stiffness of an intact ACL, a 10mm BPTB graft, and a 14 mm BPTB graft, the last of which is approximately three times stiffer than the intact ACL (Butler et al., 1986; Noyes et al., 1984).

2. Methods

The 3D computational model was constructed from magnetic resonance (MR) images of a cadaver knee (65 years old) (Li et al., 1999). The bone and cartilage contours were digitized from the MR images and used to reconstruct the bone and cartilage geometry using MSC/Patran® (The MacNeal-Schwendler Corp., CA, USA). Aside from the bone and cartilage geometry, the model included ligaments and meniscus elements (Fig. 1). The cartilage was assumed to be linearly elastic with a Young's modulus of 5 MPa (Blankevoort et al., 1991).

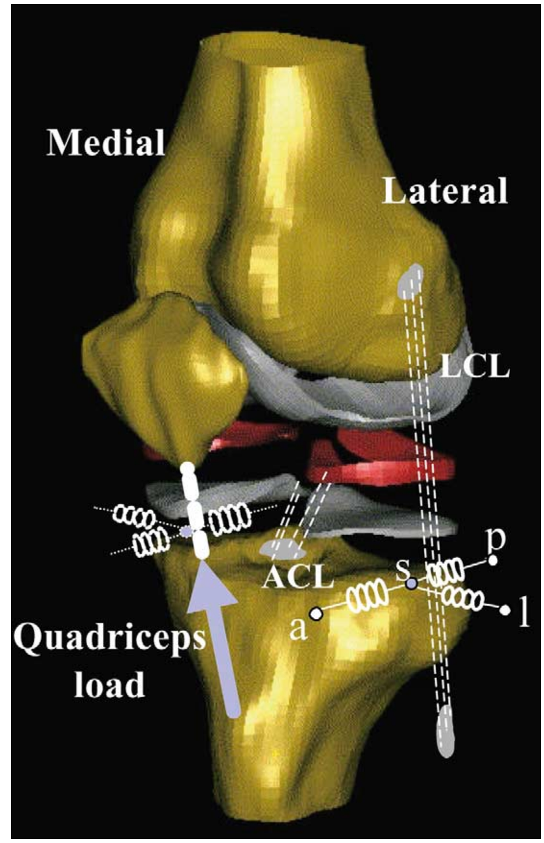

Fig. 1. The computational knee joint model constructed using MR images. The knee model includes the femur, tibia, patella, cartilage layers and menisci, where the menisci are simulated as equivalent resistant springs. At the lateral side **a–s** represents the anterior compressive spring, **p–s** posterior spring, and **s–l** the lateral spring. There are similarly three springs used at the medial side.

The ligament insertions on the cadaver specimen were digitized and positioned on the model relative to bony landmarks (Li et al., 1999). The ACL and posterior cruciate ligament (PCL) were modeled as having four bundles, the medial collateral ligament (MCL) including deep (capsular) fibers as having five bundles, and the lateral collateral ligament (LCL) as having three bundles (Butler et al., 1986; Li et al., 1999; Blankevoort et al., 1991; Mommersteeg et al., 1996). Each bundle was represented by a nonlinear tensile spring with the following load–displacement relationship:

$$f = \begin{cases} 0, & \in < 0, \\ 1/4k \in^2 / \in_l, & 0 \leqslant \in \leqslant 2 \in_l \\ k(\in - \in_l), & \in > 2 \in_l \end{cases}$$

where $\in$ is the strain, $\in_l$ is a ligament parameter, and k represents the axial modulus in the linear region of the load–displacement relationship. The strain is defined as $\in = (l - l_0)/l_0$, where l is the deformed bundle length and l_0 is the zero load length of the bundle (Blankevoort and Huiskes, 1996). The axial modulus (the product of Young's modulus and the cross-sectional area of the bundle) is the force per unit strain generated in the ligament bundle. It is equivalent to the familiar "structural" stiffness (force per unit elongation) multiplied by l_0. The axial moduli for the ligament bundles were adapted from previous reports (Butler et al., 1986; Blankevoort and Huiskes, 1996; Andriacchi et al., 1983). The menisci were modeled with compression springs connected to the tibia and femur and distributed in the anterior–posterior (AP) and medial–lateral directions (Li et al., 1999) (Fig. 1).

The model was optimized and validated using experimental results (Li et al., 1999). A robotic testing system was used to experimentally measure the behavior of the knee used to construct the model. Measurements were obtained with the knee under AP loads and internal–external (IE) moments. In order to optimize the model, the zero load lengths of the ligament bundle elements and the stiffness of each meniscus element were adjusted until the model predictions of the knee kinematics under simulated AP loads matched the kinematics obtained experimentally under the same loading conditions. To validate the model, the model predictions under simulated IE moments were compared to the experimental measurements obtained from the cadaver knee under IE loading and from other results reported in the literature. The comparison demonstrated that the predicted kinematics of the knee were consistent with the experimental data. More details of the construction and validation approach have been reported previously (Li et al., 1999).

To simulate ACL reconstruction using single bundle grafts, the ACL in the model was replaced with a graft modeled with a single nonlinear spring element. Grafts with three different axial moduli were used in this study. Graft 1 had a modulus similar to the intact ACL, graft 2 had a modulus similar to that of a 10mm BPTB graft as reported by Cooper et al. (1993), and graft 3 had a modulus similar to that of a 14mm BPTB graft as reported by Butler et al., Cooper et al., and Noyes et al. (Butler et al., 1986; Noyes et al., 1984; Cooper et al., 1993) (Table 1). The ACL deficient case was analyzed by using a graft with zero axial modulus. The initial graft

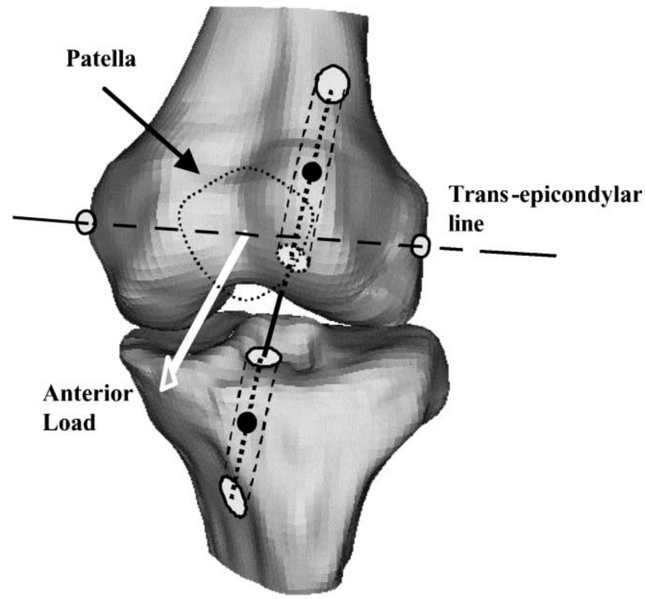

Fig. 2. Definition of the transepicondyle line and tunnel and fixation positions of the ACL graft. Graft is fixed to the bone halfway through the tunnels (denoted by black dots).

tension was set to 0 or 40 N with the knee at 30° of flexion (Marder et al., 1991; Aglietti et al., 1994) by adjusting the zero load length of the graft. The graft tunnels were modeled according to a current ACL reconstruction technique first described by Clancy et al. (1982), Shelbourne and Gray, 1997 (Fig. 2). The graft was assumed to be rigidly fixed to the bone at the midpoint of the length of each tunnel. Contact between the graft and the tunnel was assumed to otherwise be frictionless.

A simulated 134 N anterior drawer load was applied to the tibia while the knee was fixed at 0°, 30°, and 60°, and 90° of flexion (Aglietti et al., 1997; Yasuda et al., 1997). The anterior drawer load was applied at the knee center, which was defined as the midpoint of the transepicondylar line (Fig. 2). The femur was held fixed in space while the tibia was allowed to move in all

Table 1
Axial moduli of the ACL and various grafts (kN) adopted from literature

Graft	(Hamner et al., 1999)[a,b]	(Noyes et al., 1984)[a,b]	(Noyes et al., 1984)[b,c]	(Cooper et al., 1993)[b,c]	(Woo et al., 1997)[c,d]	(Butler et al., 1986)[c,e]	Model
ACL	–	4.9	–	–	6.5	13.3	10
BPTB, 10mm	–	–	–	22.8	–	23	20
BPTB, 14mm	–	18.4	31.0	27.8	–	32.2	30
Semitendinosus	6.4	5.0	15.1	–	–	–	–
QST/G	23.3	–	–	–	–	–	–

[a] Soft tissue was held directly by grips.
[b] Hamner et al., Noyes et al. and Cooper et al. report structural stiffness based on lengths of 30, 26.9, and 50 mm, respectively.
[c] Soft tissue was left attached to bone or markers were used to measure local strain.
[d] Length measurements from Noyes et al. were used to convert reported structural stiffnesses to axial moduli.
[e] Area measurements from Noyes et al. were used to convert reported Young's Moduli to axial moduli.

degrees of freedom except flexion. A large sliding contact algorithm was used when analyzing the tibiofemoral articulation. The model was analyzed in intact ACL, ACL deficient (complete rupture), and ACL reconstructed states using the three grafts and two initial tensions described above. The calculated anterior tibial translation (ATT), internal tibial rotation (ITR), and ACL/graft tension were recorded. The process was repeated at all targeted flexion angles with a 400 N (approximately half body weight) quadriceps load applied to the tibia instead of the anterior drawer load to simulate isometric extension of the knee. The vector of the quadriceps force was determined by the orientation of the inferior pole of the patella with respect to the tibial tubercle using radiographs of the cadaver knee used to build the model (Fig. 1). The patellofemoral joint was not included in the model.

3. Results

3.1. Anterior drawer load

Under the 134 N anterior drawer load, ATT in the intact knee increased with flexion from 0° to 30° of flexion with a peak ATT of 5.3 mm and then decreased slightly from 30° to 90° of flexion (Table 2). ATT of the ACL deficient knee followed the same trend, but the magnitude was 30–40% greater than that of the intact knee. The peak ATT for the ACL deficient knee was 7.2 mm, which was 36% greater than the intact value. ITR of the knee in all states had local minimums at 0° and 60° of flexion and local maximums at 30° and 90° (Table 2). ITR of the deficient knee was 15–50% greater than that of the intact knee. The intact ACL tension increased slightly with increasing flexion from 0° to 30° peaking at 97 N and then gradually decreased from 30° to 90° of flexion (Table 2).

After ACL reconstruction, the ATT, ITR, and graft tension of the knee followed the same trends with respect to flexion angle as the intact knee under the anterior drawer load (Table 2). All ACL reconstructions reduced the ATT compared to the ACL deficient knee at all the selected flexion angles. When the initial graft tension was set to 0 N, ACL reconstruction using graft 1 produced a peak ATT of 5.7 mm at 30° of flexion, which was 8% greater than that of the intact knee at the same flexion angle. Reconstruction using grafts 2 and 3 produced peak ATTs of 5.1 and 4.8 mm, respectively, at 30° of flexion. These translations were 3% and 10% less than the ATT of the intact knee, respectively. The ITR at 30° of flexion when using graft 1 was 5.7°, 10% greater than the intact ITR (Table 2). The ITRs when using grafts 2 and 3 were 5.0° and 4.5°, which were 4% and 13% less than the intact ITR, respectively. Peak graft tension was also observed at 30° of flexion in all three grafts. When graft 1 was used, the peak tension was 77 N, which was 21% less than the intact ACL (Table 2). Graft 2 had a peak tension of 105 N, which was 8% greater than the intact ACL, and graft 3 had a peak tension of 121 N, which was 24% greater that that of the intact ACL.

When the initial graft tension was set to 40 N, ACL reconstruction using grafts 1, 2, and 3 produced peak ATTs of 4.3, 3.9, and 3.6 mm, respectively, at 30° of flexion under the anterior drawer load (Table 2). These ATTs were 18%, 27%, and 32% less than that of the intact knee. The ITRs at 30° of flexion for grafts 1, 2, and 3 were 3.9°, 3.3°, and 3.0°, which were 24%, 36%,

Table 2
Anterior tibial translation, internal tibial rotation and ACL/graft tension of its of the knee in intact, ACL deficient and ACL reconstructed conditions in response to a 134 N anterior tibial load

Flexion angle	Intact	Initial tension 0 N			Initial tension 40 N			Deficient
		Graft 1	Graft 2	Graft3	Graft 1	Graft 2	Graft 3	
Anterior tibial translation (mm)								
0°	3.5	3.8	3.5	3.2	2.8	2.5	2.3	4.7
30°	5.3	5.7	5.1	4.8	4.3	3.9	3.6	7.2
60°	5.0	5.4	4.9	4.5	4.1	3.7	3.5	7.0
90°	4.9	5.3	4.9	4.7	4.2	3.9	3.8	6.4
Internal tibial rotation (deg.)								
0°	3.6	4.0	3.6	3.3	2.9	2.5	2.2	5.1
30°	5.2	5.7	5.0	4.5	3.9	3.3	3.0	7.9
60°	3.2	3.3	3.1	3.1	3.0	2.8	2.7	3.7
90°	5.6	5.9	5.5	5.3	4.7	4.5	4.3	7.4
ACL/graft tension (N)								
0°	84.8	65.4	93.6	111.0	134.6	156.5	168.5	0.0
30°	97.5	77.3	105.2	121.2	138.7	155.3	163.2	0.0
60°	89.9	72.3	95.5	108.6	126.2	140.1	148.0	0.0
90°	72.9	51.7	70.3	81.5	103.5	115.4	122.1	0.0

Two initial graft tensions (0 and 40 N) were simulated.

Table 3
Anterior tibial translation, internal tibial rotation and ACL/graft tension of the knee in intact, ACL deficient and ACL reconstructed conditions in response to a 400 N quadriceps load

Flexion angle	Intact	Initial tension 0 N			Initial tension 40 N			Deficient
		Graft 1	Graft 2	Graft 3	Graft 1	Graft 2	Graft 3	
Anterior tibial transition (mm)								
0°	3.6	4.0	3.5	3.3	2.9	2.5	2.3	4.8
30°	4.9	5.3	4.7	4.3	3.8	3.4	3.1	7.0
60°	2.1	2.0	2.0	1.9	0.7	0.9	1.0	2.1
90°	−2.5	−2.5	−2.5	−2.5	−2.5	−2.5	−2.5	−2.5
Internal tibial rotation (deg.)								
0°	4.8	5.4	4.7	4.4	3.9	3.3	2.9	6.3
30°	6.8	7.5	6.5	5.9	5.1	4.4	3.9	10.0
60°	4.5	4.4	4.4	4.3	2.5	2.9	3.0	4.5
90°	−4.3	−4.3	−4.3	−4.3	−4.3	−4.3	−4.4	−4.3
ACL/graft tensiion (N)								
0°	92.4	70.0	102.1	122.4	144.6	168.7	181.2	0.0
30°	97.7	77.7	103.7	117.7	132.6	144.7	150.0	0.0
60°	0.0	1.4	2.4	3.0	19.9	8.4	18.1	0.0
90°	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0

Two initial graft tensions (0 and 40 N) were simulated.

and 43% less than the intact ITR, respectively. The graft tensions at 30° of flexion were 139, 155, and 163 N, respectively. These were 42%, 59% and 67% greater than that of the intact ACL (Table 2).

3.2. Quadriceps load

Under the quadriceps load, ATT in the intact knee increased with increasing flexion peaking at 30° with an ATT of 4.9 mm (Table 3). The ATT then decreased as flexion increased from 30° to 90°. ATT in the ACL deficient knee followed the same trend as the intact knee with a 7.0 mm peak ATT, a 43% increase from the intact knee. ITR in the intact knee also increased with increasing flexion, peaking at 30°, and then decreased from 30° to 90° of flexion (Table 3). The ITR of the intact knee at 30° of flexion was 6.8°. In the ACL deficient knee, the ITR at 30° of flexion was 10.0°, 47% greater than the intact ITR. The absence of the ACL had very little effect on ATT and ITR at 90° of flexion. The intact ACL tension increased slightly from 0° to 30° of flexion, with a peak tension of 98 N (Table 3). Beyond 30° of flexion, the ACL tension dramatically decreased to 0 N at 90° of flexion. As was the case under the anterior tibial drawer load, the most extreme behavior under the quadriceps load tended to occur at 30° of flexion.

After ACL reconstruction, the ATT, ITR, and graft tension of the simulated ACL reconstructions generally followed the trends of the intact knee under the quadriceps load (Table 3). With an initial graft tension of 0 N, ACL reconstruction using graft 1 produced a peak ATT at 30° of flexion of 5.3 mm, which was 9% greater than that of the intact knee. When the knee was reconstructed using grafts 2 and 3, the peak ATTs of 4.7 and 4.3 mm, respectively, also occurred at 30° of flexion. These values were 4% and 12% less than the intact knee, respectively. At 30° of flexion, the ITR of the knee reconstructed with graft 1 was 7.5°, 10% greater than that of the intact knee. The ITRs when grafts 2 and 3 were used were 6.5° and 5.9°, which were 4% and 13% less than the intact ITR, respectively. Graft 1 experienced a peak tension of 78 N, which was 20% less than the intact ACL tension, at 30° of flexion (Table 3). The tension in graft 2 was 104 N, 6% greater than the intact ACL tension. The tension in graft 3 was 118 N, which was 20% greater than the intact ACL tension.

When the ACL was reconstructed with a 40 N initial graft tension, the resulting ATT and ITR under the quadriceps load were less than those obtained when a 0 N initial graft tension was used (Table 3). Grafts 1, 2, and 3 produced ATTs of 3.8, 3.4, and 3.1 mm, respectively, at 30° of flexion. These translations were 22%, 31% and 36% less than that of the intact knee. The ITRs at 30° of flexion when grafts 1, 2, and 3 were used were 5.1°, 4.4°, and 3.9°, which were 25%, 36%, and 43% less than the intact ITR, respectively. The tension of graft 1 at 30° of flexion was 133 N, 36% greater than that of the intact ACL tension (Table 3). The tensions in grafts 2 and 3 were 145 and 150 N, which were 48% and 53% greater than the intact ACL tension.

4. Discussion

ACL reconstruction has been shown to be effective in restoring anterior knee stability (Marder et al., 1991; O'Neill, 2001; Aglietti et al., 1994). However, clinical studies with long-term follow-up have reported an increased incidence of complications after ACL reconstruction, such as early joint degeneration or patellofemoral joint pain, suggesting that ACL reconstruction may not be as efficient as expected in preventing long-term joint degeneration (O'Neill, 2001; Jomha et al., 1999; Daniel et al., 1994; Dye et al., 1999; Jarvela et al., 2001). Current ACL grafts (BPTB, QST/G) have been shown to have axial moduli two to four times greater than that of the native ACL (Butler et al., 1986; Hamber et al., 1999; Muellner et al., 1998; Noyes et al., 1984) (Table 1). The supraphysiologic stiffness of the grafts may be a factor resulting in an over-constrained knee after ACL reconstruction. However, it is difficult to experimentally investigate the effect of variations in graft stiffness on knee kinematics.

The computer model of the knee used in this study has been extensively validated in our previous study (Li et al., 1999). Comparisons between published data and predicted kinematics and ligamentous tensions under various loading conditions have demonstrated that this model can be used to calculate knee biomechanics in response to external loads. For example, under the quadriceps load, this model predicted that the anterior tibial translation increased from full extension to 30° of flexion and then decreased as flexion continued to increase. This result was consistent with those of (Hirokawa et al., 1992) and (Li et al., 1999).

The computer simulation demonstrated that the axial modulus of the graft has a considerable effect on kinematics of the ACL reconstructed knee. When the initial graft tension was set to 0 N, both grafts 2 and 3 overcorrected the knee kinematics, and the graft tensions were higher than that of the intact ACL (Table 2). Only graft 1, which had an axial modulus similar to the ACL, resulted in an under-corrected knee. When initial graft tension was set to 40 N, all three grafts over-constrained the knee by more than 15%, and the corresponding graft tensions were more than 35% greater than the tension in the intact ACL. Comparing the results of these various ACL reconstructions, graft 2 (which had an axial modulus two times that of the ACL) produced kinematics closest (2% over-constraint) to the intact knee when 0 N initial graft tension was used.

This simulation of ACL reconstruction fixed the graft at the mid-length of the tunnels making the actual graft length approximately twice that of the intact ACL. Thus, the linear structural stiffness (force per unit elongation of the whole structure) of graft 1 was less than that of the ACL, even though its axial modulus was similar to the ACL. Consequently, graft 1 offered less constraint to knee motion than the intact ACL. However, under a 40 N initial tension, the graft was prestretched through a significant portion of the toe region of the force–displacement curve (Butler et al., 1986; Blankevoort and Huiskes, 1996). Thus, preloading the graft with this initial tension resulted in an over-constrained knee under the loading conditions used in this study. While graft 2 had twice the axial modulus of the intact ACL, its structural stiffness was actually similar to that of the native ACL since the graft was approximately twice as long as the ACL. Therefore, under 0 N initial graft tension, graft 2 produced similar kinematics to the intact knee. However, preloading the graft with an initial tension of 40 N again over-constrained the knee kinematics under both loading conditions. The axial modulus of graft 3 was such that its structural stiffness was greater than the intact ACL even though the graft was longer than the intact ACL. Consequently, the knee was over-constrained by this graft even under 0 N initial graft tension.

Both in vitro and in vivo studies have observed over-constraint of knee kinematics after ACL reconstruction (Bylski-Austrow et al., 1990; Fleming et al., 1992; Beynnon et al., 2001). Bylski-Austrow et al. (1990) found cadaver knees to be over-constrained after simulated reconstruction using a graft system with stiffness similar to that of the intact ACL. The graft was fixed at all combinations of 0° and 30° of flexion with initial tensions of 22 and 44 N, and the knees were placed under a 100 N tibial anterior drawer load. Both initial tensions led to over-constrained knees at almost all flexion angles from 0° to 90° when the graft was fixed at 30° of flexion. Fleming et al. (1992) also performed a cadaver study of ACL reconstruction using a graft with stiffness similar to that of the ACL. The graft was fixed with initial tensions of 0, 9, 18, and 27 N at 30° of flexion. The conclusion of the study was that any initial tension $\geqslant$9 N over-constrains the knee under a 150 N tibial anterior drawer load. As BPTB and QST/G grafts have stiffnesses greater than the ACL, it follows that these studies are consistent with our conclusion that an ACL reconstruction using BPTB or QST/G grafts fixed with any significant initial tension at 30° of knee flexion may produce an over-constrained knee. Beynnon et al. (2001) found knees reconstructed with a 10mm BPTB graft to have 2 mm less anterior laxity than intact knees under a 90 N anterior tibial load immediately after reconstruction in living patients. The initial tension in the graft as well as the graft fixation sites were not reported in this study, so it is difficult to directly compare their results to ours. However, the initial postoperative laxity data reported in this study confirms that ACL reconstruction using a BPTB or QST/G graft with a significant initial tension can produce an over-constrained knee in the initial postoperative period.

An over-constrained knee may be the result of a "tight" graft, meaning a graft with excessive structural stiffness or initial tension or an inappropriate combination of the two. The structural stiffness of the graft is not only dependent on the axial modulus of the graft, but also the graft length. In this study, graft 2 had a similar structural stiffness of the intact ACL when the graft was fixed at the mid-length position of the tunnels even though its axial modulus was two times that of the ACL. However, if the graft were fixed at the articular inlet of the tunnels, the graft would have a greater structural stiffness than the ACL. Therefore, if the graft is fixed close to the articular intlets, a graft with lower axial modulus than a 10mm BPTB graft should be used. Conversely, if the graft were fixed at the outlet of the tunnels, the graft would have a structural stiffness less than the ACL. If this fixation is used, a graft with axial modulus greater than a 10mm BPTB graft should be used. Ishibashi et al. (1997) has observed the effect of tunnel fixation sites on anterior stability of the knee. Anterior knee laxity decreased as the fixation sites were moved closer to the articular openings of the tunnels. The distance between fixation sites may need to be adjusted depending on the graft material to assure the reconstruction has a structural stiffness similar to that of the intact ACL.

Over-constrained knee kinematics may result in increased joint contact forces. Decreased anterior tibial translation arising from over-constrained ACL reconstruction will reduce the moment arm of the patellar tendon. Under this condition, a greater quadriceps force will be required to produce the same extension moment as in the intact state. This increased quadriceps force results in increased patellar contact pressure (Cain and Schwab, 1981; Cross and Powell, 1984). Similarly, decreased internal tibial rotation may also result in higher contact pressures in the lateral facet of the patellofemoral joint (Li et al., 2002). Therefore, restoration of normal knee kinematics after ACL reconstruction may be necessary to protect the knee from over loading.

A limitation of this computer simulation is that we assumed the graft to be rigidly fixed to the bone. It has been shown that graft fixation may not be rigid (Rowden et al., 1997; Kurosaka et al., 1987). When this is the case, the structural stiffness of the overall ACL reconstruction may be reduced. We are currently developing an elastic fixation element to simulate non-rigid fixation. As in all other in vitro experiments, this model does not account for changes in the reconstruction that take place over time, such as healing or remodeling (Beynnon et al., 2001; Beynnon et al., 1997). Therefore the data in this study can only relate to knee function immediately after reconstruction.

Despite these limitations, this model has unique advantages. The model was constructed from anatomical data of a human knee, and was validated using experimental data (Li et al., 1999). It allows for precise control and parametric evaluation of various factors involved in ACL reconstruction, such as graft stiffness, initial tension, and graft location. This flexibility can be used to determine how a particular variable affects the kinematics and kinetics of the knee after ACL reconstruction and help develop guidelines for an optimal reconstruction. The model can be used indefinitely, and the results of a certain experiment on the model are not affected by previous experiments. We believe that 3D computational modeling will be a powerful tool in the investigation of the effect of various soft tissue injuries and surgical reconstructions on knee joint behavior.

In conclusion, a computational knee joint model was used to investigate the effect of variations in graft stiffness along with initial graft tension on kinematics in the ACL reconstructed knee. ACL reconstructions were simulated using grafts with axial moduli similar to the ACL, a 10mm BPTB graft, and a 14mm BPTB graft, under initial tensions of 0 or 40 N. The graft with *structural stiffness* similar to that of the intact ACL best restored intact knee kinematics when it was placed anatomically and fixed with 0 N initial tension at 30° of flexion. All three grafts over-constrained the knee when fixed with a 40 N initial graft tension. The structural stiffness of a graft, along with the initial graft tension, plays a important role in determining the outcome of ACL reconstruction. The structural stiffness depends on the axial modulus of the graft and the graft length, which is determined by the choice of fixation points. Therefore, a combination of axial modulus and fixation points should be chosen such that the structural stiffness of the ACL reconstruction is similar to that of the intact ACL.

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