ELSEVIER

Journal of Orthopaedic Research 21 (2003) 708–715

Journal of Orthopaedic Research

www.elsevier.com/locate/orthres

The effect of initial graft tension on the biomechanical properties of a healing ACL replacement graft: a study in goats

Steven D. Abramowitch, Christos D. Papageorgiou, John D. Withrow, Thomas W. Gilbert, Savio L-Y. Woo *

Department of Orthopaedic Surgery, Musculoskeletal Research Center, E1641 Biomedical Science Tower, 210 Lothrop Street, P.O. BOX 71199, Pittsburgh, PA 15213, USA

Accepted 27 November 2002

Abstract

While a number of in vitro studies have shown that the tension on an anterior cruciate ligament (ACL) replacement graft at the time of fixation has an affect on joint stability, most in vivo studies have reported little or no long-term difference in outcome. The objectives of this study were to (1) establish a large animal model in which differences in knee stability are present at time-zero after ACL reconstruction with grafts fixed at a low (5 N) and high (35 N) initial tension and to (2) quantitatively determine if these initial effects remain after six weeks of healing and if the tensile properties of an ACL replacement graft are influenced by initial graft tension. Seventeen skeletally mature female Saanan breed goats were used. Using the robotic/UFS testing system, the knee kinematics and in situ forces in the replacement graft in response to an externally applied 67 N anterior–posterior (A–P) tibial load were evaluated at time-zero and after six weeks of healing. Afterward, the femur-ACL graft–tibia complexes (FGTCs) from the six-week group were tested under uniaxial tension so that the stress relaxation and structural properties of the FGTC were obtained.

At time-zero, knees fixed with a high initial graft tension could better reproduce the A–P translation of the intact knee in response to the 67 N A–P tibial load. Further, in situ forces in these grafts were also closer to those in the intact ACL under the same external loading condition. After six weeks of healing, the A–P translation of the knee and in situ forces in the replacement grafts became similar for the low and high tension groups, while both were significantly different from controls. Further, the percentage of stress relaxation as well as the stiffness, ultimate load at failure, ultimate elongation at failure, and energy absorbed of the FGTCs for both reconstruction groups were not significantly different from each other, but were significantly different from controls. These results demonstrate that while the high initial graft tension could better replicate the normal knee kinematics at time-zero, these effects may diminish during the early graft healing process.

Introduction

An anterior cruciate ligament (ACL) reconstruction is a complex procedure, as many surgical variables can affect the ability of the graft to successfully restore knee function. The variables that are most frequently studied include the placement of the graft, knee flexion angle at the time of fixation, tension of the graft at the time of fixation, method of fixation, and selection of graft material [16,35,40,41]. With regards to the initial tension applied at graft fixation, the optimal amount is still unknown. Some have recommended a low initial graft tension to reduce the risk of greater stresses in the graft, to avoid over-constraining the knee, as well as to prevent excessive contact force of the articular surfaces [13,43,47], while others have advocated high tension to restore knee stability [3,30,32,46].

Cadaveric studies aimed at resolving this issue have demonstrated mixed results with some suggesting that the magnitude of initial graft tension at the time of fixation does not have an affect on the kinematics of the knee and forces in the ACL graft [19,49], while others have demonstrated that increasing initial graft tension leads to a significant decrease in knee laxity, and still others have pointed out that excessive tension may lead to an over-constrained knee [4,13,14,29–31]. A recent study by Fleming et al. [14] examined the time-zero effect of initial graft tension on A–P translation as a function

* Corresponding author. Tel.: +1-412-648-2000; fax: +1-412-648-2001.

E-mail address: ddecenzo@pitt.edu (S.L-Y. Woo).

0736-0266/03/$ - see front matter © 2003 Orthopaedic Research Society. Published by Elsevier Science Ltd. All rights reserved.
doi:10.1016/S0736-0266(02)00265-6

of knee flexion in a goat model. They found that levels of tension applied at various angles of knee flexion all resulted in significant increases or decreases in knee laxity compared to the intact knee [14].

There are a few in vivo studies that have attempted to answer whether or not initial graft tension affects outcome [25,43,46–48]. An early study in dogs revealed at three months post-operatively there were neither differences in knee laxity nor in the structural properties of graft–bone complexes with the graft fixed at 1 and 39 N [47]. King et al. [25] performed a study comparing knee laxity with three levels of tension (tight, anatomic, and loose) in an MCL autograft rabbit model out to 48 weeks. At time-zero, large differences were observed in knee laxity, however, those differences diminished greatly after 12 weeks of healing, and were undetectable after 24 weeks. For all three conditions, the joint laxity was nearly restored to normal levels [25]. Later, Yasuda et al. [46] evaluated patients at two years whose ACLs were reconstructed with a semitendinosus graft fixed at either 20, 40, or 80 N of tension. The only relationship that was found was an inverse correlation between graft tension and knee laxity for grafts fixed at 20 and 80 N [46].

Thus, the effects of initial graft tension remain unclear. While some clinical studies have shown that the tension on an ACL replacement graft at the time of fixation has an affect on joint stability, the function of the ACL graft cannot be directly determined. Issues including the contribution of muscles to joint stability, swelling of soft tissues, and a lack of standardized techniques to evaluate the ACL graft's contribution to joint function have clouded the ability for researchers to quantitatively assess the effects of initial graft tension during the early healing process. As the graft will undergo changes during the early stages of healing [20, 33,34,37,47], it is important to know how the effects of initial graft tension change with time. A multiple time point study that quantitatively examines the ACL graft's contribution to knee kinematics as function of initial graft tension will help to delineate the relationship between initial graft tension and the remodeling process of the ACL graft.

As previous studies have shown that initial graft tension can affect knee laxity at time-zero using the goat model [14], the objectives of this study were to (1) use this model to establish an ACL reconstruction in which there are differences in knee stability and ACL graft function at time-zero for grafts fixed at a low (5 N) and high (35 N) initial tension and to (2) quantitatively determine if these initial effects remain after six weeks of healing and if the tensile properties of an ACL replacement graft are influenced by initial graft tension. The data used for evaluating these effects include kinematics of the reconstructed knees and in situ forces in the ACL grafts, in addition to the viscoelastic behavior and structural properties of the healing femur-ACL graft–tibia complex (FGTC) under uniaxial tension. As previous studies have shown that the remodeling process of an autograft causes significant changes to the graft's mechanical behavior, we hypothesize that the effects of high initial graft tension demonstrated at time-zero, including knee kinematics under external loading conditions and in situ forces in the graft that were closer to those of the normal knee when compared to a lower graft tension, will be diminished after six weeks such that knee kinematics and in situ forces in the ACL graft will be similar for both the high and low initial graft tension groups [9,25,33].

Materials and methods

Seventeen female Saanan breed goats (wt. 44.3 ± 6.2 kg; mean± SD) were used for this study. The goat was chosen for this study because, unlike the canine and rabbit models, ACL reconstructions have been deemed successful out to three years with very low rates of graft failures or severe articular cartilage degeneration [23,33,34]. All goats were required to pass a veterinary examination to rule out any physical or physiological abnormalities prior to testing. Further, skeletal maturity was assessed by dental evaluation, and radiographs were taken to ensure the closure of the epiphyseal plate. The research protocol followed the National Institutes of Health guidelines for animal care and was approved by the University of Pittsburgh's Institutional Animal Care and Use Committee.

Six knees from three goats were utilized for the time-zero evaluations. The knees were subjected to ACL reconstructions using bone–patellar tendon–bone autografts fixed with high and low initial tensions, therefore each knee served as its own control [28]. For the six week groups, 14 goats were used and equally divided into two groups. The ACL in the right knee of seven goats was reconstructed with grafts fixed at a high initial tension (35 N), while the ACL in the right knee of the remaining seven goats was reconstructed with grafts fixed at a low initial tension (5 N). Their contralateral legs served as control.

All surgical procedures were performed using sterile techniques under general endotracheal anesthesia using isoflurane. An anterior midline skin incision was made to harvest a 6 mm wide central third patellar tendon graft. After the ACL was completely resected, an osseous tunnel, approximately 20 mm in length, was drilled through the center of the femoral insertion site of the ACL. The tibial tunnel, approximately 20 mm in length, was made using a 5 mm canulated drill guide system centered on the tibial insertion of the ACL. The femoral and tibial bone blocks were trimmed to snugly pass through a 5.5 and 5 mm sizer, respectively.

The cross-sectional area of the patellar tendon graft was measured 1 cm from the patellar bone block using a laser micrometer (accuracy of 0.1 mm^2) prior to use in the ACL reconstruction [44]. Sutures (2-0 TICRON) were passed through the femoral bone block and were feed through the femoral tunnel. Since the femoral bone block was trimmed to be slightly larger than the femoral tunnel, a press-fit was obtained by impacting the femoral bone block with a pusher and hammer into the femoral tunnel. Finally, the sutures attached to the femoral bone block were then tied over a titanium button on the lateral femoral cortex for additional fixation. A 2-0 TICRON suture was then attached through the tibial bone block and was used for passing the graft through the tibial tunnel. Relative to the size of the knee joint, the bone–patellar tendon–bone graft for the goat is much longer than that for the human. Thus, the tibial bone block was pulled completely through the tibial tunnel. A spring scale was fixed to the suture and the graft was then tensioned in the tunnel with five cycles of knee flexion–extension while the graft was held under a constant load of 35 N.

Tension was then released and the knee was taken through one more cycle of knee flexion to allow for application of the fixation tension from an initial lax position of the graft. Following this cycle, the graft was pulled to the appropriate tension and the graft was fixed on the medial surface of the tibia using a 6.4 mm fixation staple with spikes (Smith and Nephew, Memphis, TN) with the knee held at 30° of flexion. The 2-0 TICRON suture used for passing the graft through the tunnel was tied to a cortical screw that was fixed distal to the staple on the medial surface of the tibia for additional support. Once the graft was fixed, the knee was examined to ensure that a full range of knee flexion/extension was possible. The wounds were closed using standard suture technique.

Post-operatively, all animals were allowed free cage activity (cage area, 3 m^2). The status of weight bearing and general health condition of all goats was monitored during recovery. After six weeks, the goats were humanely euthanized. Both hind limbs were disarticulated at the hip joint, sealed in double plastic bags, and immediately stored at −20 °C [45].

To examine the effects of initial graft tension on knee kinematics and in situ forces in the ACL graft, all knees were tested using a robotic/universal force-moment sensor (UFS) testing system [38,39] (Fig. 1). The robotic manipulator (Puma Model 762, Unimate, Inc.) is capable of achieving position control in 6 degree of freedom (DOF), with a repeatability of 0.2 mm for translations and 0.2° for rotations. Further, the UFS (Model 4015, JR3, Inc., Woodland, California) can measure three orthogonal forces and moments with a repeatability of 0.2 and 0.01 N-m for forces and moments, respectively. With the force feedback from the UFS, the robotic/UFS testing system can also operate in a force-control mode [17,18,38]. For both human and goat cadaveric knees, this system has been used successfully to apply external loads to the joint at preselected angles of knee flexion, while the kinematics of the resulting 5 DOF (medial–lateral, proximal–distal, anterior posterior (A–P) translations, internal–external and varus valgus rotations) joint motions are measured [22,34,39]. By repeating these positions with a high level of accuracy, the principle of superposition can be applied, allowing the in situ forces in the ligament to be determined [17,18,38].

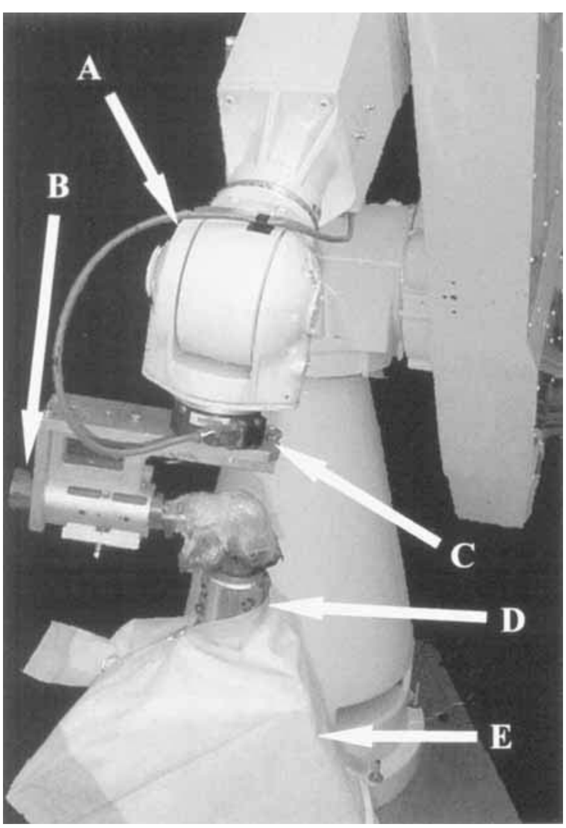

Fig. 1. A photograph of a specimen mounted on the robotic/UFS testing system. (A) 6 DOF robotic manipulator; (B) Tibia fixed in tibial clamp; (C) UFS; (D) Femur mounted in femoral clamp; (E) Pedestal rigidly attached to floor.

Prior to biomechanical testing, each specimen was thawed for 24 h at room temperature. The specimens were kept moist with 0.9% saline during dissection and biomechanical testing. The tibia and femur were cut 20 cm from the joint line and the surrounding skin and muscles were dissected 10 cm proximal and distal to the knee joint. To minimize bending or twisting during testing, the shafts of the tibia and femur were potted in an epoxy compound. The potted specimens were secured with bolts in thick-walled aluminum cylinders, which could be mounted onto the robotic/UFS testing system. The femur was rigidly fixed relative to the base of the robotic manipulator and the tibia was mounted to the end-effector of the robot through the UFS. The center of the knee (defined as the midpoint between the femoral insertions of the MCL and LCL) was measured relative to the UFS.

The robotic/UFS testing system operating in force-control mode, was used to determine the 5 DOF knee kinematics in response to a 67 N A–P tibial load at selected angles of knee flexion (30°, 60°, 90°) [22,27,34]. To determine the in situ forces in the ACL at time-zero, the ligament was transected. Then, the recorded kinematics of the intact knee were repeated. Using the principle of superposition, the vector difference in force measured before and after the ACL was sectioned represented the in situ force in the intact ACL [15,17,38]. After the ACL was reconstructed with an initial graft tension of 5 N, the 5 DOF kinematics of the reconstructed knee were again determined. The initial graft tension of 5 N was chosen because it represented the lowest measurable tension that could be applied in order to remove slack from the graft. The graft was then released and the recorded kinematics of the knee reconstructed with 5 N of initial graft tension were repeated to obtain the in situ force in the ACL graft. The graft was once again tensioned, this time to 35 N. A load of 35 N was chosen based on the results of Yoshiya et al. [47], which demonstrated that joint degeneration occurred in a dog model after tensioning of an ACL graft to 39 N [47]. Using the same testing protocol, the kinematics of the knee reconstructed with 35 N of initial graft tension were recorded and the in situ force in the ACL graft was again determined. The order of tensioning was not varied to ensure that the creep behavior of the ACL graft was consistent for each experiment and the results are comparable to the operative procedures performed on specimens assigned to the healing portion of this study.

For the specimens evaluated after six weeks of healing, the 5 DOF kinematics of the reconstructed and control knees were determined using the same methodology for those tested at time-zero. To determine the in situ force in the ACL graft, all soft tissue structures in and around the knee except the ACL graft were dissected, leaving the FGTC. The articulating surfaces of the femoral condyles were also removed to prevent possible bone-to-bone contact. The recorded kinematics were repeated by the robotic manipulator and the UFS directly recorded the in situ force that was in the ACL graft since it is the only structure attaching the femur to the tibia. For the contralateral legs, the same methodology was used to determine the 5 DOF knee kinematics and the in situ force in the ACL. The difference in protocol between the time-zero and six week specimens was to allow for tensile testing the healing FGTC at six weeks. Compiled data from our research center ensured that both methods used to determine the in situ force in ligaments produced comparable results [1].

After the FGTCs were removed from the robotic/UFS testing system, they were further prepared to undergo uniaxial tensile testing to determine the structural properties and viscoelastic behavior of the FGTC after healing. First, the tibia and femur were cut 10 cm from the joint line and mounted in customized clamps so that the cross-sectional area of the ACL replacement graft could be measured by means of a laser micrometer system [44].

Subsequently, each specimen was mounted in customized clamps and tested on an Instron uniaxial tensile testing machine (model 4502; Instron, Canton, MA, USA) using an apparatus that enabled the orientation of the ACL to be adjusted and aligned along the axis of the applied tensile load in both the frontal and sagittal planes [26]. The size of this clamp prevented the use of a saline bath. Thus, specimens were kept moist with repeated applications of saline using a spray bottle. For the intact ACL, care was taken to align each specimen in an anatomical orientation such that the natural insertion angles of the ACL were maintained, allowing for a smooth transition of load from bone to ligament as well as a more uniform load distribution within the ligament [26]. For each ACL reconstructed specimen, all fixation hardware was removed and the ACL graft attached to the bone outside the tibial tunnel was transected prior to testing. This was done so the tissue that healed outside of the tibial tunnel did not contribute to the results obtained from tensile testing.

After a preload of 2 N was applied, the gauge length reference position of the Instron crosshead was reset to 0 mm. Each FGTC underwent preconditioning by cyclically loading between 0 and 0.75 mm for 10 cycles at 10 mm/min. This was followed by a static stress relaxation test. In this test, each specimen was again elongated at a crosshead rate of 10 mm/min to 1.5 mm (rise time of approximately 9 s), and the load was measured over a 60-min period. The percentage of stress-relaxation was defined as the difference between the final load (at 60 min) and the initial peak load, divided by the initial peak load, multiplied by 100%. After 60 min of recovery, the 10 cycles of preconditioning were once again repeated. The FGTC was loaded until failure. The stiffness was defined as the slope of the linear region of the load-elongation curve such that the Pearson's product correlation coefficient was greater than 0.99 when comparing the straight line and the experimental data. The ultimate load was the maximum load achieved before failure.

In order to account for specimen variability due to differences in size and geometry of each animal, ΔA–P translation, defined as the A–P translation for one reconstructed knee condition (i.e. knee fixed with 5 N of initial graft) minus the A–P translation of its control, was calculated and compared. The mean and variance for ΔA–P translation were computed for statistical comparisons between the tension groups and between time periods. As only the magnitude of the in situ force vector was considered for this study, the control in situ force values were not subtracted from those for the reconstructed knee conditions. All statistical analyses done on the kinematic and force magnitude data were performed using a repeated measures two-factor analysis of variance (ANOVA) with knee condition and time of healing as factors. This was followed by comparisons using multiple contrasts at each flexion angle independently. For these tests, significance was set at $p < 0.05$.

As the viscoelastic behavior and structural properties were only determined for FGTCs at six weeks, unpaired t tests were used to compare the experimental/control ratios between groups (i.e. 5 to 35 N groups) and paired t tests were used to compare non-normalized data within a group (i.e. 5 N reconstruction at six weeks to its control) for these data. A Bonferroni correction was used to account for multiple comparisons. For these tests, significance was set at $p < 0.017$.

Results

All goats tolerated surgery well and were weight-bearing within a few hours after surgery. However, visual inspection revealed that the animals walked with a limp for three to four weeks. Upon dissection, all ACL replacement grafts were found to remain intact and were encapsulated by a white fibrinous tissue. One specimen from the low initial graft tension group was damaged during specimen preparation after removal from testing on the robotic/UFS testing system for knee kinematics. That specimen and its control were not subjected to uniaxial tensile testing. Thus, the sample size for tensile data for the low initial tension group was reduced to $n = 6$.

For comparison between specimens, the kinematic data is represented as ΔA–P translation ($p < 0.05$; Fig. 2). A translation of zero represents no difference from the respective control (i.e. the intact knee at time-zero and the contralateral leg after six weeks of healing). At time-zero, the ΔA–P translations for the high tension group

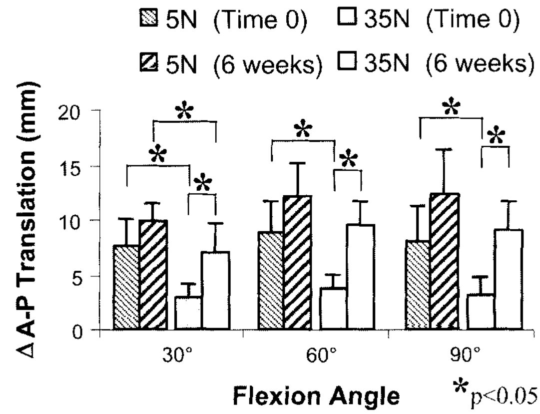

Fig. 2. ΔA–P translation for the low and high tension groups at time-zero and after six weeks of healing in mm (mm, mean ± SD).

were found to be significantly less ($p < 0.05$) than the values for the low tension group (2.8 ± 1.2 vs. 7.5 ± 2.6 mm; 3.8 ± 1.2 vs. 8.8 ± 3.0 mm; and 3.1 ± 1.5 vs. 7.9 ± 3.3 mm; at knee flexion angles of 30°, 60°, and 90°, respectively; mean ± SD). After six weeks of healing, ΔA–P translations for the high tension group demonstrated significant increases of 145%, 147%, 187%, at flexion angles 30°, 60°, and 90°, respectively over those at time-zero ($p < 0.05$; Fig. 2); whereas, the ΔA–P translations for the low tension group also appeared to have increased after six weeks of healing, but no significant changes were detected, possibly due to the small sample size. At six weeks there were no statistically significant differences in ΔA–P translations between the low and high tension groups at all tested flexion angles except 30° of flexion (i.e. 6.9 ± 2.7 mm for the high tension group, and 9.7 ± 1.8 mm for the low tension group; $p < 0.05$; mean ± SD). For the reconstructed knees of both initial graft tension groups at both time-zero and after six weeks of healing, there was a significant increase in A–P translation when compared to the controls for all flexion angles tested (see Table 1).

At time-zero, the in situ forces in the grafts of the high tension group in response to a 67 N anterior load were not significantly different from the in situ forces in the intact ACLs for all flexion angles tested, while the in situ forces in the grafts of the low tension group were less at all flexion angles but only significantly different at 30° of knee flexion (Fig. 3). The differences between the low and high tension groups were approximately 35% at flexion angles of 30°, 60°, and 90° ($p < 0.05$). However, by six weeks, the in situ forces in the grafts of the low and high tension groups became similar ($p > 0.05$), and both were less than the in situ forces in the intact ACLs with significant differences at both 30° and 90° of knee flexion ($p < 0.05$; Fig. 4).

The cross-sectional areas for grafts were significantly larger than the contralateral controls (intact ACL) for both tension groups after six weeks of healing. Further,

Table 1
A–P translations (in mm) of the intact and reconstructed knees at time-zero and six weeks for the low and high tension groups at flexion angles of 30°, 60°, and 90° of knee flexion (mean ± SD)

		Time-zero		Six weeks	
		5 N tension	35 N tension	5 N tension	35 N tension
(a) 30° of knee flexion					
Intact	4.1 ± 1.0			5.3 ± 1.6	4.9 ± 1.3
Reconstructed		11.6 ± 2.3	6.9 ± 0.4	15 ± 2.6	11.8 ± 2.3
(b) 60° of knee flexion					
Intact	3.5 ± 2.2			5.1 ± 1.1	5.0 ± 1.3
Reconstructed		12.3 ± 2.1	7.3 ± 1.8	17.2 ± 3.5	14.3 ± 2.3
(c) 90° of knee flexion					
Intact	2.9 ± 2.0			4.1 ± 1.1	4.1 ± 1.3
Reconstructed		10.8 ± 3.3	4.9 ± 2.1	16.4 ± 4.8	13.0 ± 2.6

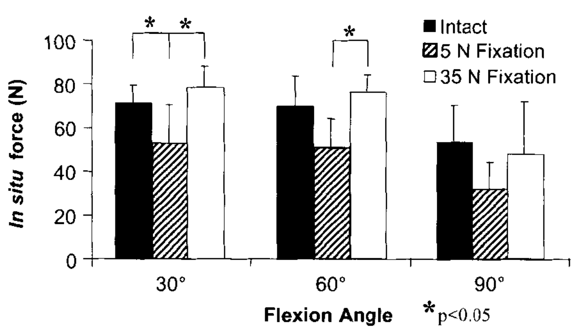

Fig. 3. In situ forces at time-zero in response to a 67 N anterior tibial load (N, mean ± SD).

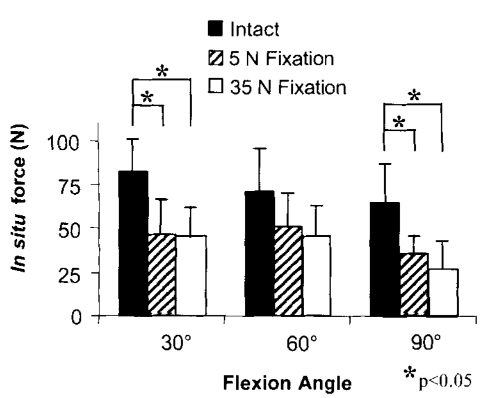

Fig. 4. In situ forces at six weeks in response to a 67 N anterior tibial load (N, mean ± SD).

for both tension groups, the cross-sectional areas of the grafts were significantly increased after six weeks of healing as compared to their initial values at the time of reconstruction ($p < 0.017$, see Table 2). No significant differences could be demonstrated between the two tension groups.

In terms of the viscoelastic behavior, the shape of the stress relaxation curves under uniaxial tension for the two tension groups showed a steeper initial slope (faster rate of relaxation) compared to their controls, but no noticeable differences between themselves. The percentage of stress relaxation for the healing FGTCs of the low and high tension groups (57% ± 15% vs. 60% ± 8%, respectively; mean ± SD) was significantly larger than their controls (28% ± 7% vs. 25% + 13%, respectively; mean ± SD). Again, there were no statistically significant differences for the percentage of stress relaxation of low and high tension groups ($p < 0.017$).

Table 2
Cross-sectional area (in mm^2) of the graft at the time of implantation and after six weeks of healing, as well as that of the control ACL

	Graft at implantation	Graft at six weeks	Control ACL
5 N	17.5 ± 3.6*	54.7 ± 8.4*†	29.9 ± 5.5
35 N	19.7 ± 1.7*	46.2 ± 14.6*†	26.0 ± 1.9

$^*p < 0.05$ compared to the control, $^\dagger p < 0.05$ compared to graft at implantation (mean ± SD).

The values for stiffness, ultimate load at failure, ultimate elongation at failure, and energy absorbed to failure are shown in Table 3. No significant differences were seen between the low and high tension groups for these parameters, but significant differences were detected between the two tension groups and their respective controls for all parameters ($p < 0.017$). Further, the modes of failure were observed, with approximately half of the FGTCs failed in the midsubstance of the graft and the other half failed by pullout from the tibia. This was the case for both initial tension groups.

Discussion

This study examined changes in the ACL graft's contribution to knee kinematics as a function of initial graft tension and the biomechanical properties of ACL replacement grafts during the early phase of healing. Using a goat model in which initial graft tension had an

Table 3
Structural properties for the control and reconstructed specimens for the low and high tension groups (mean ± SD)

	5 N		35 N	
	Control	Experimental	Control	Experimental
Stiffness (N/mm)	143.3 ± 32.4	47.9 ± 29.3*	141.3 ± 49.9	41.3 ± 26.1*
Ultimate load (N)	1553.8 ± 297.1	178.3 ± 115.4*	1368 ± 408.6	159.0 ± 129.8*
Ultimate elongation (mm)	12.7 ± 4.5	5.5 ± 2.4*	12.0 ± 4.5	4.9 ± 1.9*
Energy absorbed (N mm)	10,033.4 ± 4685.9	534.0 ± 433.9*	9054.3 ± 5904.6	464.0 ± 605.8*

*$p < 0.017$ compared to the control ACL.

effect on kinematics and in situ forces in the ACL graft at time-zero, this study was able to quantitatively assess if these effects remained throughout early graft healing. Using the robotic/UFS testing system, the 5 DOF knee kinematics and in situ forces in the ACL graft in response to 67 N A–P loads were found to be closer to control values for the high tension group at time-zero. However, after six weeks of healing, the early effects caused by high initial graft tension were diminished as the graft remodeled, confirming our hypothesis. Viscoelastic behavior and structural properties of the healing FGTCs were not affected by the magnitude of initial graft tension.

The subsequent increase in A–P translation of the reconstructed knee after six weeks in response to external loads for both low and high tension groups is in agreement with the results of previous studies [23, 33,34,37]. Further, the findings regarding the effects of initial graft tension at six weeks supports the work of Yoshiya et al. [47] who found no differences in knee laxity after three months of healing using a canine model [47]. The present study extends the previous work by quantifying changes of in situ forces in the ACL graft during the early stages of healing/remodeling as a function of initial graft tension. King et al. also found that the effects of initial graft tension diminished with healing time using a rabbit MCL autograft model and that joint laxity returned to control levels by 12 weeks [25]. The present study used a goat ACL autograft model. Further studies are needed to determine how the differences in environment (i.e. intra- vs. extra-articular) affect these processes.

Clinically, a study by Yasuda et al. [46] found an inverse correlation in knee laxity and magnitudes of initial graft tension in the range of 20–80 N, and remains one of the few studies to demonstrate this effect in patients [46]. Although the work presented in this paper only demonstrated statistically significant differences at 30° of knee flexion, it should be noted that the results of this study show similar trends in knee laxity as average ΔA–P translation of the high tension group were less than those for the low tension. It is unclear whether an even greater initial graft tension than the one used in this study would have resulted in significant differences in knee laxity. On the other hand, further increasing the initial graft tension could lead to joint degeneration [4,13,29–31].

One factor that likely contributes to the increased A–P translation of the knee in response to external loads between time-zero and six weeks is an altered viscoelastic behavior of the healing autograft. Our study, as well as others, demonstrated that the percentage of stress relaxation increased during remodeling of the ACL graft, while some studies have shown that the amount of creep and viscoplastic deformation of healing autografts is increased [9,25,33]. These processes are reflected by a decrease in the in situ forces in the graft and a transfer of loads to surrounding soft tissue structures around the knee. It should also be noted that other factors including slippage of the graft from the fixation may have also contributed to these results.

The long-term success of ACL reconstructions using the goat model, robust activity level, and large size make it an attractive model for this study. As neither ACL reconstructions in the goat nor the human are completely understood, more work needs to be done to determine the relevance of this model to the human situation. Although short- and long-term outcome studies have shown unsatisfactory results in as many as 20–25% of patients [2,5–7,24,36] and others have shown that A–P translations in patients increase with time [8,21,42], the results of this study may not correlate directly with clinical studies due to inherent differences between species. As with humans, studies using the goat model have shown that there are many surgical and non-surgical variables that can affect knee kinematics following reconstruction [10–12]. Post-operative rehabilitation protocols or the stresses subjected to the ACL graft as it remodels are difficult to control using animals. Other differences from clinical studies may be attributed to the methodology used to evaluate A–P translation of the knee; or that these results were observed in cadaveric knees and not directly from patients. Thus, the contribution of muscle forces to joint stability were not considered. It should be noted, however, that this is not a limitation of our study as this allowed for the function of the ACL graft to be uniquely determined.

Still, the question of optimal initial graft tension for an ACL replacement graft has yet to be completely elucidated. However, this study does provide some insight into the relationship between initial graft tension and the remodeling process of an ACL replacement graft by demonstrating that the effects of a high initial graft tension are diminished after the first six weeks of healing. Even though the time periods and loading conditions chosen for evaluation do not completely represent the time-course of healing for an ACL reconstruction, the knowledge gained in this study does suggest that large changes occurred to the graft that contributed to changes in knee function even at this early phase of healing. The assessment of more complex loading conditions at multiple time periods will add a wider knowledge of the healing characteristics of an ACL reconstruction and the role initial graft tension plays in that process.

Acknowledgements

Technical support of Eiichi Tsuda, M.D., as well as the financial support by the NIH Grant AR41820 are gratefully acknowledged.

References

[1] Abramowitch SD, Gabriel MT, Ma CB, Woo SL-Y. Validating the principle of superposition as a method to determine in situ forces in the anterior cruciate ligament of the goat. In: 49th Annual Meeting of the Orthopaedic Research Society, to be presented. New Orleans, LA, 2003.
[2] Aglietti P, Buzzi R, Giron F, Simeone AJ, Zaccherotti G. Arthroscopic-assisted anterior cruciate ligament reconstruction with the central third patellar tendon. A 5–8-year follow-up. Knee Surg Sports Traumatol Arthrosc 1997;5:138–44.
[3] Amis AA, Jakob RP. Anterior cruciate ligament graft positioning, tensioning and twisting. Knee Surg Sports Traumatol Arthrosc 1998;6:S2–12.
[4] Andersen HN, Jorgensen U. The immediate postoperative kinematic state after anterior cruciate ligament reconstruction with increasing preoperative tension. Knee Surg Sports Traumatol Arthrosc 1998;6:S62–9.
[5] Anderson AF, Snyder RB, Lipscomb Jr AB. Anterior cruciate ligament reconstruction. A prospective randomized study of three surgical methods. Am J Sports Med 2001;29:272–9.
[6] Aune AK, Holm I, Risberg MA, Jensen HK, Steen H. Four-strand hamstring tendon autograft compared with patellar tendon–bone autograft for anterior cruciate ligament reconstruction. A randomized study with two-year follow-up. Am J Sports Med 2001;29:722–8.
[7] Bach Jr BR, Tradonsky S, Bojchuk J, Levy ME, Bush-Joseph CA, Khan NH. Arthroscopically assisted anterior cruciate ligament reconstruction using patellar tendon autograft. Five–nine-year follow-up evaluation. Am J Sports Med 1998;26:20–9.
[8] Barber-Westin SD, Noyes FR, Heckmann TP, Shaffer BL. The effect of exercise and rehabilitation on anterior–posterior knee displacements after anterior cruciate ligament autograft reconstruction. Am J Sports Med 1999;27:84–93.
[9] Boorman RS, Shrive NG, Frank CB. Immobilization increases the vulnerability of rabbit medial collateral ligament autografts to creep. J Orthop Res 1998;16:682–9.
[10] Cummings JF, Grood ES. The progression of anterior translation after anterior cruciate ligament reconstruction in a caprine model. J Orthop Res 2002;20:1003–8.
[11] Cummings JF, Grood ES, Butler DL, Levy MS. Subject variation in caprine anterior cruciate ligament reconstruction. J Orthop Res 2002;20:1009–15.
[12] Cummings JF, Grood ES, Levy MS, Korvick DL, Wyatt R, Noyes FR. The effects of graft width and graft laxity on the outcome of caprine anterior cruciate ligament reconstruction. J Orthop Res 2002;20:338–45.
[13] Fleming B, Beynnon B, Howe J, McLeod W, Pope M. Effect of tension and placement of a prosthetic anterior cruciate ligament on the anteroposterior laxity of the knee. J Orthop Res 1992; 10:177–86.
[14] Fleming BC, Abate JA, Peura GD, Beynnon BD. The relationship between graft tensioning and the anterior–posterior laxity in the anterior cruciate ligament reconstructed goat knee. J Orthop Res 2001;19:841–4.
[15] Fox RJ, Harner CD, Sakane M, Carlin GJ, Woo SL-Y. Determination of the in situ forces in the human posterior cruciate ligament using robotic technology. A cadaveric study. Am J Sports Med 1998;26:395–401.
[16] Frank CB, Jackson DW. The science of reconstruction of the anterior cruciate ligament. J Bone Joint Surg Am 1997;79:1556–76.
[17] Fujie H, Livesay GA, Woo SL-Y, Kashiwaguchi S, Blomstrom G. The use of a universal force–moment sensor to determine in situ forces in ligaments: a new methodology. J Biomech Eng 1995; 117:1–7.
[18] Fujie H, Mabuchi K, Woo SL-Y, Livesay GA, Arai S, Tsukamoto Y. The use of robotics technology to study human joint kinematics: a new methodology. J Biomech Eng 1993;115:211–7.
[19] Gertel TH, Lew WD, Lewis JL, Stewart NJ, Hunter RE. Effect of anterior cruciate ligament graft tensioning direction, magnitude, and flexion angle on knee biomechanics. Am J Sports Med 1993;21:572–81.
[20] Grana WA, Egle DM, Mahnken R, Goodhart CW. An analysis of autograft fixation after anterior cruciate ligament reconstruction in a rabbit model. Am J Sports Med 1994;22:344–51.
[21] Howell SM, Taylor MA. Brace-free rehabilitation, with early return to activity, for knees reconstructed with a double-looped semitendinosus and gracilis graft. J Bone Joint Surg Am 1996; 78:814–25.
[22] Ishibashi Y, Rudy TW, Livesay GA, Stone JD, Fu FH, Woo SL-Y. The effect of anterior cruciate ligament graft fixation site at the tibia on knee stability: evaluation using a robotic testing system. Arthroscopy 1997;13:177–82.
[23] Jackson DW, Grood ES, Goldstein JD, Rosen MA, Kurzweil PR, Cummings JF, et al. A comparison of patellar tendon autograft and allograft used for anterior cruciate ligament reconstruction in the goat model. Am J Sports Med 1993;21:176–85.
[24] Jomha NM, Borton DC, Clingeleffer AJ, Pinczewski LA. Long-term osteoarthritic changes in anterior cruciate ligament reconstructed knees. Clin Orthop 1999:188–93.
[25] King GJ, Edwards P, Brant RF, Shrive NG, Frank CB. Intraoperative graft tensioning alters viscoelastic but not failure behaviours of rabbit medial collateral ligament autografts. J Orthop Res 1995;13:915–22.
[26] Lyon RM, Woo SL-Y, Hollis JM, Marcin JP, Lee EB. A new device to measure the structural properties of the femur-anterior cruciate ligament–tibia complex. J Biomech Eng 1989;111:350–4.
[27] Ma CB, Janaushek MA, Vogrin TM, Rudy TW, Harner CD, Woo SL-Y. Significance of changes in the reference position for measurements of tibial translation and diagnosis of cruciate ligament deficiency. J Orthop Res 2000;18:176–82.
[28] Ma CB, Papageogiou CD, Debski RE, Woo SL-Y. Interaction between the ACL graft and MCL in a combined ACL + MCL

knee injury using a goat model. Acta Orthop Scand 2000;71:387–93.

[29] Markolf KL, Burchfield DM, Shapiro MM, Cha CW, Finerman GA, Slauterbeck JL. Biomechanical consequences of replacement of the anterior cruciate ligament with a patellar ligament allograft. Part II: forces in the graft compared with forces in the intact ligament. J Bone Joint Surg Am 1996;78:1728–34.

[30] Markolf KL, Burchfield DM, Shapiro MM, Davis BR, Finerman GA, Slauterbeck JL. 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 Joint Surg Am 1996;78:1720–7.

[31] Melby AD, Noble JS, Askew MJ, Boom AA, Hurst FW. The effects of graft tensioning on the laxity and kinematics of the anterior cruciate ligament reconstructed knee. Arthroscopy 1991; 7:257–66.

[32] More RC, Markolf KL. Measurement of stability of the knee and ligament force after implantation of a synthetic anterior cruciate ligament. In vitro measurement. J Bone Joint Surg—Am Vol 1988;70:1020–31.

[33] Ng GY, Oakes BW, Deacon OW, McLean ID, Lampard D. Biomechanics of patellar tendon autograft for reconstruction of the anterior cruciate ligament in the goat: three-year study. J Orthop Res 1995;13:602–8.

[34] Papageoriou CD, Ma CB, Abramowitch SD, Clineff TD, Woo SL-Y. A multi-disciplinary study of the healing of an intra-articular anterior cruciate ligament graft in a goat model. Am J Sports Med 2001;29:620–6.

[35] Penner DA, Daniel DM, Wood P, Mishra D. An in vitro study of anterior cruciate ligament graft placement and isometry. Am J Sports Med 1988;16:238–43.

[36] Ritchie JR, Parker RD. Graft selection in anterior cruciate ligament revision surgery. Clin Orthop 1996:65–77.

[37] Rodeo SA, Arnoczky SP, Torzilli PA, Hidaka C, Warren RF. Tendon-healing in a bone tunnel. A biomechanical and histological study in the dog. J Bone Joint Surg—Am Vol 1993;75:1795–803.

[38] Rudy TW, Livesay GA, Woo SL-Y, Fu FH. A combined robotic/universal force sensor approach to determine in situ forces of knee ligaments. J Biomech 1996;29:1357–60.

[39] Sakane M, Fox RJ, Woo SL-Y, Livesay GA, Li G, Fu FH. In situ forces in the anterior cruciate ligament and its bundles in response to anterior tibial loads. J Orthop Res 1997;15:285–93.

[40] Sapega AA, Moyer RA, Schneck C, Komalahiranya N. Testing for isometry during reconstruction of the anterior cruciate ligament. Anatomical and biomechanical considerations. J Bone Joint Surg—Am Vol 1990;72:259–67.

[41] Shelbourne KD, Rowdon GA. Anterior cruciate ligament injury. The competitive athlete. Sports Med 1994;17:132–40.

[42] Shelbourne KD, Whitaker HJ, McCarroll JR, Rettig AC, Hirschman LD. Anterior cruciate ligament injury: evaluation of intraarticular reconstruction of acute tears without repair. Two to seven year followup of 155 athletes. Am J Sports Med 1990;18: 484–8, discussion 488–9.

[43] van Kampen A, Wymenga AB, van der Heide HJ, Bakens HJ. The effect of different graft tensioning in anterior cruciate ligament reconstruction: a prospective randomized study. Arthroscopy 1998;14:845–50.

[44] Woo SL-Y, Danto MI, Ohland KJ, Lee TQ, Newton PO. The use of a laser micrometer system to determine the cross-sectional shape and area of ligaments: a comparative study with two existing methods. J Biomech Eng 1990;112:426–31.

[45] Woo SL-Y, Orlando CA, Camp JF, Akeson WH. Effects of postmortem storage by freezing on ligament tensile behavior. J Biomech 1986;19:399–404.

[46] Yasuda K, Tsujino J, Tanabe Y, Kaneda K. Effects of initial graft tension on clinical outcome after anterior cruciate ligament reconstruction. Autogenous doubled hamstring tendons connected in series with polyester tapes. Am J Sports Med 1997; 25:99–106.

[47] Yoshiya S, Andrish JT, Manley MT, Bauer TW. Graft tension in anterior cruciate ligament reconstruction. An in vivo study in dogs. Am J Sports Med 1987;15:464–70.

[48] Yoshiya S, Kurosaka M, Ouchi K, Kuroda R, Mizuno K. Graft tension and knee stability after anterior cruciate ligament reconstruction. Clin Orthop 2002:154–160.

[49] Zeminski J, Yagi M, Kanamori A, Fu FH, Woo SL-Y. High initial graft tension has no apparent benefit on the biomechanical outcome of ACL reconstruction. In: 46th Annual Meeting of the Orthopaedic Research Society, Orlando, FL, 2000, p. 484.