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A comparison of Anterior Cruciate Ligament graft tunnel orientation: Anatomic vs. transtibial

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ABSTRACT

Background: Recent Anterior Cruciate Ligament reconstruction techniques have emphasized reproducing the insertion sites of the native Anterior Cruciate Ligament. Anatomic techniques have shown improvements in biomechanical testing, but their superior results have not been shown clinically. The hypothesis of this study is that more oblique tunnels utilized in anatomic reconstructions cause asymmetric loading across the graft.
Methods: Seven cadaver knees were tested in a knee simulator that performed a gait cycle and an anterior–posterior laxity test. Each knee underwent both reconstructions in random order utilizing the same Anterior Cruciate Ligament bone patellar tendon bone graft. Before reconstruction, the graft was split longitudinally and miniature force probes were inserted in the medial and lateral portions.
Findings: During anterior–posterior laxity testing, the transtibial medial bundle averaged 74.8 N compared to 87 N for the anatomic. The lateral bundles averaged 146.2 and 158 N respectively. Both reconstructions exhibited a similar ratio of force distribution between the bundles and there was no statistical difference. The average anterior–posterior motion for the intact knees was 10.8 mm compared to 17.0 mm after the Anterior Cruciate Ligament was sectioned. Anatomic reconstructions had an average of 14.0 mm of laxity compared to 14.9 mm for transtibial reconstructions ($P<0.038$).
Interpretation: Greater obliquity did not lead to an increase in asymmetry of graft loading. The failure of anatomic reconstructions to show clinical improvement over transtibial reconstructions is not due to oblique tunnels causing asymmetric graft loading.

1. Introduction

The goal of every Anterior Cruciate Ligament (ACL) reconstruction is to provide a stable knee that will allow return to previous function and knee kinematics. Previous studies have shown aberrant knee kinematics after reconstruction and long term osteoarthritis continues to be a problem (Logan et al., 2004; McCulloch et al., 2007; Streich et al., 2008). Recently, there is considerable interest in developing and testing more anatomically correct reconstructions with the expectation that this will lead to improvements in outcomes. Single and double bundle techniques have been developed (McCulloch et al., 2007) that more closely reproduce the femoral insertion of the native ACL. Although these techniques have shown some improvements in biomechanical testing (Mae et al., 2001; Yagi et al., 2002), their superior results have yet to be shown clinically (Meredick et al., 2008; Streich et al., 2008), and the ideal location and orientation of bone tunnels in an ACL reconstruction remains controversial. Surgical techniques used to make the femoral tunnel in an anatomic location yield a more oblique tunnel in the coronal plane (Fig. 1). The native ACL insertion is typically too distal on the lateral femoral condyle to be drilled in a transtibial fashion. One hypothesized explanation for the lack of clinical improvement is that the greater tunnel obliquity of these ACL reconstructions may lead to asymmetric loading across the ACL graft.

The goal of this study was to determine if there is a difference between the transtibial reconstruction and a single bundle anatomic reconstruction in term of anterior–posterior (AP) laxity, knee kinematics, and graft loading.

2. Methods

Seven fresh cadaver knees (5 right, 2 left knees; average age 70, range 59 to 77; 2 male, 5 female) were tested in a six degree of freedom knee simulator (Sutton et al., 2010) (Fig. 2) that moved each knee through a standardized gait cycle based on an International Organization for Standardization (ISO) Standard (ISO, 2002) for load-control testing of knee implants. The simulator caused knee flexion–extension while applying axial compressive loads, AP loads, and tibial torques to the knee. To prevent possible overloading of the

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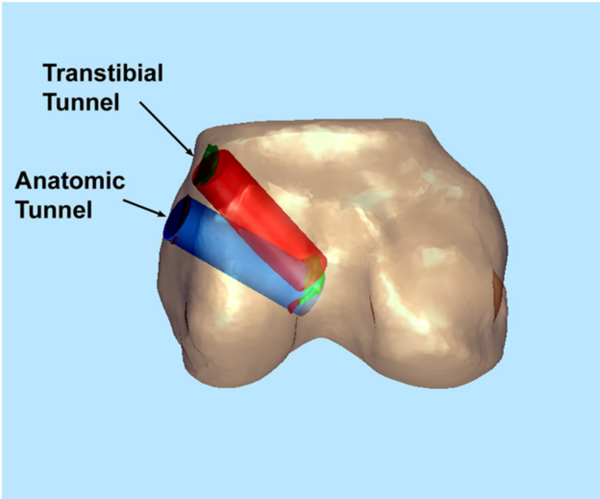

Fig. 1. Three dimensional model of the femoral tunnels taken from a single cadaver specimen.

knee ligaments and other soft tissues, 33% of each axial compressive force, AP force and tibial torque specified in the standard were utilized. Free medial–lateral motion and free abduction–adduction were permitted. In addition to moving the knee through 10 repetitive walking gait cycles, the simulator replicated a cyclic AP laxity test at 30° of knee flexion with first a 50 N anterior force and then a 50 N posterior force for 5 repetitions. During the AP laxity test a constant 150 N compressive axial force was applied while free medial–lateral motion, free tibial rotation and free abduction–adduction were permitted. Potting and alignment of each knee in the simulator was done as previously described (Sutton et al., 2010). Optical sensors (NDI Corporation, Ontario, Canada) were mounted directly to the femur and tibia with cortical pins (Fig. 2) and used to measure angular knee kinematics (with an accuracy of <0.1°) while AP displacements (with an accuracy of <0.01 mm) were measured using an integrated displacement transducer connected in parallel to the knee simulator AP actuator. The alignment of the femoral optical sensor defined the knee coordinate system from which knee flexion, tibial rotation and valgus/varus rotations could be measured. The femoral optical sensor was oriented to be parallel to the femoral coronal plane. This was defined while potting each knee by using the femoral epicondylar axes and the long axis of the femur. The femoral sensor flexion–extension axis was aligned to be parallel with the knee flexion–extension axis as determined while potting the knee (Sutton et al., 2010).

Each knee underwent both an anatomic (bone tunnel centered at the native femoral ACL attachment) and a standard transtibial reconstruction (Fig. 1) utilizing the same ACL 12 mm diameter bone–patellar tendon–bone graft with 25 mm long bone plugs (Fig. 3). The order of performing the anatomic reconstruction and the transtibial reconstruction was randomly selected for each knee. A new bone–patellar tendon–bone graft was harvested for each knee, but used for both repairs in each knee. The anatomic tunnel starting point was centered between the anteromedial and posteriolateral native bundles on the tibia and the femur, while the transtibial starting point was centered between the bundles on the tibia and the standard over the top position on the femur (7 mm over the top guide). The graft was then inserted with the cortical sides facing posterior. The femoral side was fixed first by tying over a post and the knee was then flexed 10 times while holding tension. The knee was then placed in 20° of flexion and fixed by tying the tibial sutures as tightly as possible. The second reconstruction was performed after filling the femoral tunnel with cement and then redrilling the tunnel. Each knee was tested in the knee simulator with all soft tissues intact, after the ACL was sectioned and after each of these ACL reconstructions. To facilitate performing the reconstructions, the knee was removed from the simulator for each ACL reconstruction.

Prior to the reconstructions, the tendinous portion of the graft was split axially perpendicular to the plane of the cortex and miniature force probes (AIFP force probe, MicroStrain, Inc, Williston, VT, USA) were inserted in the medial and lateral portions of the graft while the bone plugs were kept intact(Fig. 3). These bundles were aligned in the coronal plane to provide the greatest sensitivity for the difference in the coronal obliquity of the tunnels. Thus each force probe independently measured the force in one half of the graft (medial or lateral), while the bone plug on each end was secured as typically performed clinically. Instead of inserting the force probes transverse to the length of the tendon, a longitudinal pocket in the tendon was

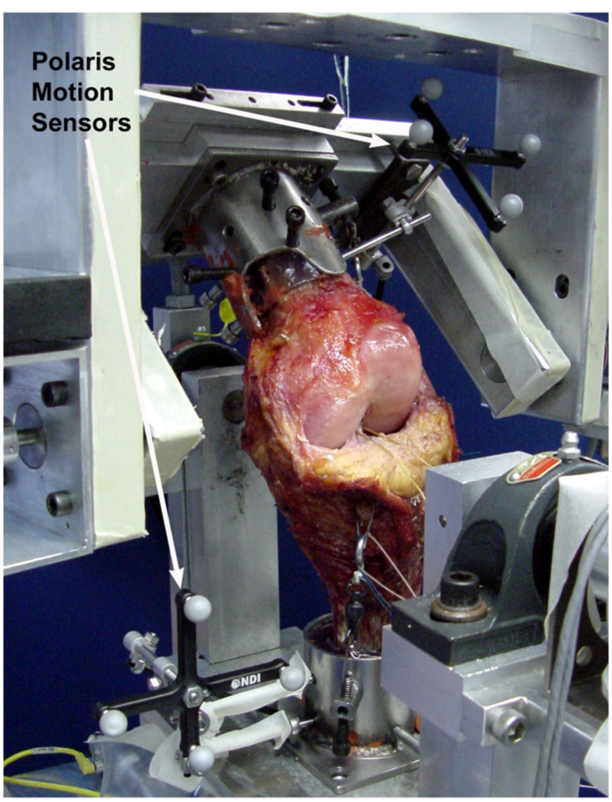

Fig. 2. Knee simulator with cadaver knee flexed showing placement of optical sensors and overall setup.

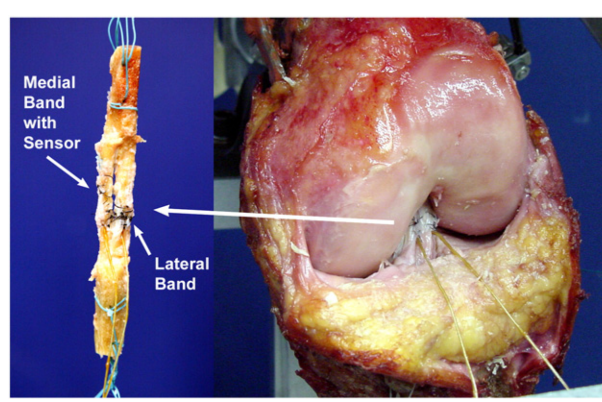

Fig. 3. Placement of tension sensors in medial and lateral bundles of ACL graft and knee shown after implantation.

created and the force probe inserted to enhance force probe stability and security. Upon completion of the experiment, the bone blocks were separated longitudinally using a small oscillating saw creating two separate bone–tendon–bone units corresponding to the medial and lateral parts of the graft. Each bone–tendon–bone unit was potted and axially loaded to determine a calibration relationship between measured probe output with the applied axial force.

During the 5th cycle of the AP drawer test, AP force and displacement data were extracted and a repeated measures analysis of variance (ANOVA) was used to compare the total AP displacement with the knee intact, after the ACL was sectioned and after each of the reconstructions. A comparison of the total AP displacement was used since no zero displacement reference value was determined with the knee at 30° of flexion and the reference at 0° is not applicable. The changes in knee varus/valgus and tibial rotation were also extracted and changes with ACL sectioning and each reconstruction were compared using a repeated measures ANOVA. The peak forces in the medial and lateral portions of the graft that occurred when the peak anterior tibial draw force was applied were compared to each other and for the 2 reconstructions using a two way repeated measures ANOVA.

Knee tibial rotation and varus/valgus data were extracted from the optical sensor measurements at 15% and 40% of the 10th gait cycle (Fig. 4) for all 7 knees. The 15% location is when the knee is maximally flexed during the stance portion of gait. The 40% location is when the knee is maximally extended before the swing phase. A repeated measures ANOVA was used to compare the amount of tibial rotation and varus/valgus alignment at these positions with the knee intact, the ACL cut and after each reconstruction.

In each of these repeated measures ANOVA tests, we assumed the data to be normally distributed and statistical significance was set at $P<.05$.

3. Results

The results for the AP laxity testing (Fig. 5) showed that the anatomic reconstruction had statistically ($P<0.038$) less total AP motion than the knee with the ACL cut, while the transtibial did not. These results are based on only 5 knees as no intact AP laxity tested was performed in the first knee while in another knee the AP zero was inadvertently changed preventing us from using that data. As expected the displacement with the ACL cut was statistically greater ($P=0.049$) than with the knee intact. The average AP motion for the intact knees was 10.8 (SD = 2.7) mm compared to 17.0 (SD = 7.4) mm after the ACL was cut. The anatomic reconstruction showed an average of 14.0 (SD = 5.5) mm of laxity compared to 14.9 (SD = 7.5) mm for the transtibial reconstruction (Fig. 5). During the AP drawer test the amount of tibial rotation was not statistically different after sectioning the ACL or after either of the reconstructions ($P>0.29$). Although there was a trend to have an increase in valgus with the ACL sectioned or with the transtibial reconstruction, it was only with the anatomic reconstruction that the increase in abduction (0.8°) was statistically different compared to the intact knee ($P=0.05$).

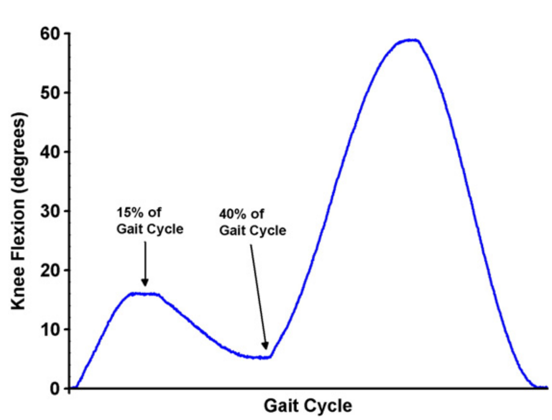

Fig. 4. Knee flexion during the gait cycle measured during one cycle of an illustrative knee to demonstrate when 15% and 40% of the gait cycle occurs.

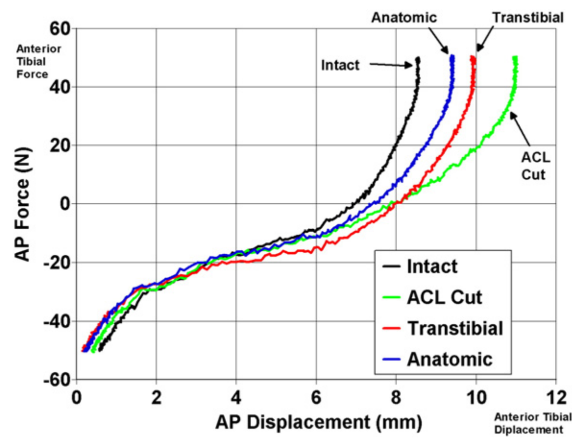

Fig. 5. AP motion in response to 50 N anterior force and 50 N posterior force in an illustrative knee with the knee in 30° of flexion. 0 mm of displacement corresponds to the neutral position with the knee at 0° of flexion.

During AP laxity testing, the force probe data suggested that both reconstructions had greater average force seen in the lateral bundles compared to the medial, however this was not statistically significant ($P=0.186$). These results are based on all 7 knees. The transtibial medial bundle averaged 74.8 (SD = 78.8) N compared to 87 (SD = 104.7) N for the anatomic. The lateral bundles average tension was 146.2 (SD = 78.0) N and 158.1 (SD = 80.6) N respectively. Both reconstructions exhibited a similar ratio of force distribution between the two bundles and there was no statistical difference ($P=0.484$) between the reconstructions.

At 15% of the gait cycle, there was a slight, but statistically significant increase in knee valgus with the transtibial reconstruction ($P=0.015$), as well with the anatomic reconstruction ($P=0.038$). The increase with the transtibial reconstruction was 1.1° and 1.0 with the anatomic. With the ACL cut, the increase was 1.0° but it was not a significant change ($P=0.062$) from the intact. At 40% of the gait cycle, there was also a statistically significant increase in valgus with both reconstructions (transtibial, $P=0.003$; anatomic, $P=0.019$). The increase with the transtibial and anatomic reconstructions was 0.5° for each. With the ACL cut, the increase was 0.3° but it was not a significant change ($P=0.225$) from the intact.

At 15% and 40% of the gait cycle, there was a trend towards greater internal tibial rotation as compared to the native knee, however these changes were not statistically significant. At 15% of the gait cycle, with the ACL cut there was an average increase of 1.7° (SD = 2.0) degrees of internal tibial rotation ($P=0.059$). Both reconstructions also allowed an increase in internal tibial rotation with the anatomic averaging 2.1° (SD = 2.8) degrees ($P=0.068$) and the transtibial 3.2° (SD = 3.9) degrees ($P=0.089$). At 40% of the gait cycle with the ACL cut there was an average increase of 2.2° of internal tibial rotation ($P=0.003$). Both reconstructions also allowed more tibial rotation than in the intact knee with the transtibial averaging 4.0° ($P=0.053$) and the anatomic 3.9° ($P=0.055$).

4. Discussion

This cadaver study supports improved knee stability when a more anatomic reconstruction is performed. Many bench studies have shown that anatomic reconstructions with femoral tunnel placed at the original site of insertion, (further down the lateral femoral condyle from the traditional overtop position) have improved AP laxity and some reduction in internal rotation. Our results further support this by showing a significantly less AP laxity in the anatomic reconstructions and a trend towards less tibial rotation.

In spite of the superior results in cadaver testing, clinical studies have failed to show clinical improvements. Our hypothesis was that the anatomic reconstruction would alter the medial to lateral loading in the ACL graft as compared to the transtibial. We felt that the improvements in stability from more anatomic tunnel placement may be negated by asymmetric loads placed on the anatomic grafts from the oblique tunnel placement leading to early partial failure. Our results did not show a significant change in the proportion of the force seen by the medial and lateral portions of the graft when comparing the two techniques. According to our model, more oblique tunnel placement does not increase the asymmetry of graft loading.

Both reconstructions had greater force in the lateral portion of graft. There is not a clear explanation for this; however, a study by Markolf et al. (2009) found that the posterior lateral bundle of a double bundle reconstructions had significantly higher tension than the anteriomedial and that surgeons should be cautious in tensioning their posterior lateral bundles. A three dimensional finite element model by Song et al. (2004) found the force in a posterior lateral bundle to be 43% higher than the anterior medial bundle in response to an anterior tibial force with the knee at full extension. Further studies are needed in this area.

ACL reconstruction with a transtibial technique or a centered anatomic single bundle reconstruction fails to restore native kinematics. The reconstructed knees showed increases in internal rotation and valgus alignment while going through an ISO standardized gait cycle compared to the native knee. These differences were quite small and may have been secondary to additional soft tissue dissection necessary to perform the reconstructions. In our testing states, we transected the ACL in mid-substance and this may have left some fibers intact and surrounding synovium adding some stability to the knee. Perhaps performing a more thorough excision rather than transecting would better simulate an ACL deficient knee. A previous study has shown abnormal rotation and varus alignment during running in post op ACL knees (Tashman et al., 2004). These abnormal kinematics may lead to long term osteoarthritic changes (Radin et al., 1994). A recent cadaver study by Kondo et al. (2011) compared transtibial, centered anatomic, and double bundle ACL reconstructions in AP laxity, pivot shift, and internal rotation testing. They found that the centered anatomic and the double bundle reconstructions rotational and pivot-shift laxity were significantly better than that of a single bundle over the top reconstruction. They also showed no improvement of the double bundle over the single bundle anatomic reconstructions. Further investigation into developing reconstructions that come closer to restoring native knee kinematics are justified. In our model, the more anatomic reconstruction showed a trend toward more native kinematics than the transtibial.

Our model has several limitations. The tested cadaver specimens had an average age of 70 years, while ACL reconstructions are more commonly performed in a younger population. The calibration of the force transducers located in the ACL grafts was performed with the graft vertical under unilateral tension. Although the in situ graft did bend and press against the tunnel, the transducer was positioned in the portion of the graft that was outside of the tunnels and did not flex or impinge. Our knee simulator does not tension the quads or hamstrings and thus doesn't account for dynamic stability. Also, the soft tissue dissection necessary to safely implant the grafts with the sensors in place removed the majority of the anterior knee capsule and extensor mechanism. Removal of the knee to perform the ACL reconstructions required its reinsertion back into the knee simulator. We feel that minimal or no changes in the knee's alignment occurred during its reinsertion due to the fixture design and tolerances used in reattaching it to the simulator. Lastly there are large standard deviations in our results, indicating that there are large variations in how individual knees responded to loading and the repairs. However, by using a repeated measures ANOVA statistical test, each specimen served as its own control.

Another limitation is our graft tensioning. Our graft tensioning protocol consisted of the single surgeon tying sutures as tightly as possible over a post as is done in a standard ACL reconstruction. Utilization of a tension gage to ensure equal tension on the grafts would be more consistent.

5. Conclusions

In our cadaver model, greater tunnel obliquity did not significantly alter the medial to lateral force distribution in the graft. The clinical significance of this finding is that the reported failure of anatomic reconstructions to show clinical improvement over transtibial reconstructions would appear to not be related to a change in how the graft is loaded. Although both grafts failed to statistically restore native knee kinematics during gait, more anatomic reconstructions had a trend for allowing less internal tibial rotation and also reduced AP motion during AP laxity testing to a greater degree than the transtibial reconstructions. These results suggest the potential clinical value of the more anatomic reconstruction. However, further clinical and cadaver investigation into more anatomic reconstructions is warranted.

Conflict of interest

No author has any financial or personal relationships with other people or institutions that could inappropriately influence or bias this work.

Mr. Werner and Mr. Sutton have a patent on a non related shoulder device.

Mr. Werner has institutional grant funding from several orthopedic companies that is unrelated to this study.

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