Tibial Slope and its Impact on Force in Anterior Cruciate Ligament Grafts:

ACL Force Increases Linearly as Posterior Tibial Slope Increases

Abstract

Background: Previous work has reported that increased tibial slope is directly correlated to increased anterior tibial translation, possibly predisposing patients to higher rates of ACL tears and causing higher rates of ACL graft failures over the long term. However, the effect of changes in sagittal plane tibial slope on ACLR graft force has not been well defined.

Purpose/Hypothesis: The purpose of this study was to quantify the effect of changes in sagittal plane tibial slope on ACLR graft force at varying knee flexion angles. Our null hypothesis was that changing the sagittal plane tibial slope would not affect force on the ACL graft.

Study Design: Controlled laboratory study.

Methods: Ten male, fresh-frozen, cadaveric knees had a posterior tibial osteotomy performed and an external fixator placed for testing and accurate slope adjustment. Following ACLR, specimens were compressed with a 200 N axial load at flexion angles of 0°, 15°, 30°, 45°, and 60°, and the graft loads were recorded through a force transducer clamped to the graft. Tibial slope was varied between -2° and 20° of posterior slope at 2° increments under these test conditions.

Results: ACL graft force, in the loaded testing state, increased in a linear fashion as slope increased. This effect was independent of flexion angle. The final model utilized a 2-factor linear mixed-effects regression model and overall noted a significant highly positive and linear relationship between tibial slope and ACL graft force in axially loaded knees at all flexion angles tested (slope coefficient = 0.92, SE = 0.08, p<0.001). Significantly higher graft force was also observed at 0° of flexion compared to all other flexion angles for the loaded condition (all p < 0.001).
**Conclusion:** We found that tibial slope has a strong linear relationship to the amount of graft force experienced by an ACL graft in axially loaded knees. Thus, a flatter tibial slope had significantly less loading of ACL grafts, while steeper slopes increased ACL graft loading. Our biomechanical findings support recent clinical evidence of increased ACL graft failure with increased tibial slope secondary to increased graft loading.

**Clinical Relevance:** Evaluation of the effect of increasing tibial slope on ACL graft force can guide surgeons when deciding if a slope decreasing proximal tibial osteotomy should be performed prior to a revision ACLR. Overall, as slope increases, ACL graft force increases, and in our study, flatter slopes had lower ACL graft forces and were protective of the ACLR graft.

**Key Terms:** tibial slope, anterior cruciate ligament reconstruction, anterior tibial translation, ACL graft forces, closing wedge osteotomy

**For peer review only:**

**What is known about the subject:** Clinical studies have documented that patients with increased sagittal plane tibial slope have higher rates of both ACL tears and ACL graft failures.

**What this study adds to existing knowledge:** This study clearly demonstrates the impact and graft loading with increased and decreased tibial slope on reconstructed ACL grafts, which has not been clearly delineated previously.
Introduction

It has been reported that sagittal plane tibial slope has a role in the risk of anterior cruciate ligament (ACL) tears and that increased slope can significantly impact the rate of subsequent ACL reconstruction graft failure. Native posterior tibial slope has been described to average approximately 7°-10°. Previous work has reported that an increased sagittal plane tibial slope of >12° is directly correlated to increased anterior tibial translation (ATT), predisposing patients to ACL tears and reportedly higher ACL graft failures over the long term. Therefore, it has been suggested that decreasing the posterior tibial slope in patients with increased sagittal plane tibial slope will potentially protect anterior cruciate ligament reconstruction (ACLR) grafts and reduce the risk of failed revision ACLR surgeries in patients with high amounts of posterior tibial slope.

The native and reconstructed ACL serves as a primary restraint to ATT at lower flexion angles between 0-30°. The determination of ACL loading at lower flexion angles in those with increased tibial slope may help aide surgeons and physical therapists in rehabilitation protocols if patients are at an increased risk of ACLR failure. In addition, evaluation of the effect of increasing tibial slope on ACL graft force can also guide surgeons when deciding if a slope reducing proximal tibial osteotomy should be performed prior to a revision ACLR.

While the clinical effects of increased tibial slope have been reported, there is a paucity of biomechanical information on the impact of changes in tibial slope on the forces experienced by ACL grafts. Therefore, the purpose of this study was to quantify the effect of changes in sagittal plane tibial slope on ACLR graft force at varying tibial slopes and knee flexion angles. Our null hypothesis was that changing the sagittal plane tibial slope would not affect force on the ACL graft at any flexion angle.

Materials & Methods:

Specimen Preparation
Ten male, fresh-frozen, cadaveric knees with an average age of 53 years (range, 33-64) and average BMI of 23.3 kg/m² (range, 16.2-36.9) were used. Specimens with prior surgery or evidence of meniscal, cartilage, ligament damage, or osteoarthritis were excluded. The cadaveric specimens utilized were donated to a tissue bank for the purpose of medical research and then purchased by our institution. Institutional review board approval was not required because the use of cadaveric specimens is exempt at our institution.

The skin was removed and all posterior subcutaneous tissues were dissected off the specimen >2 cm distal to the joint line. The popliteus muscle belly was reflected from its origin to visualize the posterior cortex of the tibia at the site of the tibial osteotomy. To maintain the rotational stability of the knee joint, the popliteus tendon was anchored to the posterior cortex of the tibia using suture anchors. The ACL was resected while all other ligamentous structures were left intact. The femur, tibia, and fibula were cut 20 cm distal to the joint line. The distal tibia and fibula were potted up to a point 11 cm distal to the tibial tubercle in a cylindrical mold using poly (methyl methacrylate) (PMMA; Fricke Dental International, Streamwood, IL, USA) with the tibial plateau oriented parallel to the base.

**Surgical Technique**

The ACL was reconstructed by an anatomic, single-bundle technique as previously reported.10,15 Next, the native sagittal plane tibial slope was measured on a true lateral radiograph. Posterior tibial slope measurements were made on lateral radiographs under standard fluoroscopy using a previously validated technique. The native baseline tibial slope was defined as the angle between the medial tibial plateau and a line parallel to the mid-diaphysys of the tibia and was measured using radiographs. The tibial mid-diaphyseal line was centered through the tibial shaft using two lines, one 5 cm distal to the joint line and one 15 cm distal to the joint line and the midpoint of these two lines represented the mid-diaphyseal line and a line was drawn parallel to the tibial plateau. The angle between these two lines was subtracted from 90° to calculate the resultant tibial slope (Figure 1).22
Figure 1. Lateral radiograph demonstrating described technique for calculating sagittal plane tibial slope. The resultant tibial slope angle was subtracted from 90° to determine the posterior tibial slope (in degrees).

A posterior tibial osteotomy, allowing opening and closing of the wedge, was performed 2.5 cm distal to the joint line and progressed parallel to the joint with a saw blade under live radiographic visualization, ensuring the osteotomy did not break through the anterior cortex of the tibia, while leaving a 5-6 mm anterior bone hinge. Based upon the native slope, a wedge of 15 mm was resected to allow adequate slope changes.

Tibial slope was fixed with a medial and lateral external fixation device (Synthes Medium external Fixator, Synthes USA, Westchester, PA) that allowed the slope to be varied as desired. All specimens had two 7.6 cm generic box nails placed along the most anterior aspect of the hinge in a vertical fashion to ensure that no failure of the hinge occurred while the slope was varied. Prior to testing, tibial slopes were measured fluoroscopically on each specimen at five positions spanning -2° to 20° of slope, and these positions were marked on the external fixation device. All other slopes were obtained by linearly interpolating between the measured positions.

Graft Preparation Protocol
Grafts were preconditioned with a force of 250 N ten times to ensure proper conditioning to minimize creep during testing. After preconditioning, the ACL graft was fixed in the femoral tunnel with an 8 x 20 mm interference screw (Smith & Nephew, Andover, MA). The ACL graft consisted of two semitendinosus and two gracilis allografts which were sized to ensure that they would extend far enough out of the tibial tunnel to allow the graft to be clamped to a calibrated external load cell sensor, (Sensortronics, Vishay Precision Group, Malvern, PA, USA). Load sensor calibration was confirmed prior to testing by comparing sensor output to the output from a calibrated dynamic tensile testing machine (Elecropuls E10000; Instron, Norwood, MA, USA) at 5 N load increments ranging from 5 – 100 N.

Mechanical Testing Protocol

The potted tibia was rigidly secured in a custom pivoting base that was allowed to freely translate on the testing table of the Instron testing machine. The orientation of the tibia was modified by the pivoting base to ensure that the mid-shaft of the tibia was orientated perpendicular to the testing table. The femur was secured to a custom fixture previously utilized to vary knee flexion, which was rigidly mounted to the actuator by passing a 10 mm rod transversely through the femoral epicondyles. The 10 mm rod acted as the load-bearing pivot axis. Next, a 7 mm rod was passed through the distal femoral shaft to serve as a fixation point when selecting the knee flexion angle during testing (Figures 2 and 3). All specimens were loaded by compressing the joint with a 200 N axial load at flexion angles of 0°, 15°, 30°, 45°, and 60° and the graft loads were recorded between 20 N and 200 N of joint compression. Pilot testing demonstrated osteotomy failure during testing with 300 N of axial compression, while the specimens did not fail at axial loads of 200 N. Therefore, 200 N was chosen as the axial load for this study. Tibial slope was varied between -2° and 20° of posterior slope in 2° increments. The order of tibial slopes and flexion angles was randomized for all tests. Each time the tibial slope was changed, the graft was re-tensioned to 88 N with the knee in full extension. The unloaded force was the force recorded in the ACL graft prior to axial compression, and the loaded force
was the force that was recorded in the ACL graft after being compressed axially for 20 seconds.

Throughout testing and preparation, the knees were sprayed with normal saline to prevent soft tissue desiccation.

**Figure 2.** (A) Anterior and (B) posterior views of testing set up in a right knee.

**Figure 3.** Schematic representation of the mechanical testing setup for a right knee.
**Statistical Methods**

Two-factor linear mixed-effect models were used to assess the effects of posterior tibial slope and knee flexion angle on ACL graft force for both the unloaded and axially loaded states. Random intercepts were used to allow a different baseline force for each specimen, and to account for the repeated measures nature of the experimental design. Final model specification, including the decision to include an interaction effect and whether tibial slope required a polynomial relationship, was determined among candid models via the Bayesian Information Criterion (BIC). Tukey’s method was used to make pairwise comparisons among the 5 tested flexion angles. Residual diagnostics were performed to confirm model assumptions and model fit. The statistical computing software R was used for all analyses (access date August 17, 2018; R, R Foundation for Statistical Computing with lme4). As a simplification of the full analysis, statistical power and sample size was considered for a single flexion angle and based on the null hypothesis of zero relationship between tibial slope and ACL force within the context of ordinary linear regression. Assuming an alpha level of 0.05, ten specimens, each measured at 12 different values of tibial slope, were sufficient to detect an effect size of $r = 0.25$ with 80% statistical power. This was a conservative estimate because we expected the two-factor random intercept model to account for additional variance within the repeated measures data, and thus lead to higher statistical power.

**Results**

Results are reported in terms of both the loaded and unloaded graft forces that were seen in the ACL graft both before and after axial loading. Means and standard deviations are presented for each combination of posterior tibial slope and knee flexion angle, for both the loaded and unloaded states, in Tables 1 and 2.
### Table 1: Mean unloaded graft force ± standard deviation (N) by posterior tibial slope and knee flexion angle

<table>
<thead>
<tr>
<th>Flexion Angle</th>
<th>-2°</th>
<th>0°</th>
<th>2°</th>
<th>4°</th>
<th>6°</th>
<th>8°</th>
<th>10°</th>
<th>12°</th>
<th>14°</th>
<th>16°</th>
<th>18°</th>
<th>20°</th>
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<tbody>
<tr>
<td>0°</td>
<td>70.8 ± 16.6</td>
<td>67.3 ± 19.3</td>
<td>65.6 ± 16.5</td>
<td>68.0 ± 13.8</td>
<td>69.2 ± 17.4</td>
<td>70.4 ± 13.0</td>
<td>74.6 ± 16.8</td>
<td>65.0 ± 11.2</td>
<td>67.7 ± 16.8</td>
<td>68.5 ± 16.1</td>
<td>80.4 ± 14.1</td>
<td>71.4 ± 13.7</td>
</tr>
<tr>
<td>15°</td>
<td>29.4 ± 12.2</td>
<td>28.2 ± 12.4</td>
<td>24.9 ± 12.3</td>
<td>25.8 ± 11.0</td>
<td>22.3 ± 8.1</td>
<td>21.6 ± 7.7</td>
<td>23.1 ± 10.7</td>
<td>20.7 ± 11.2</td>
<td>21.4 ± 12.7</td>
<td>17.8 ± 9.0</td>
<td>23.1 ± 10.0</td>
<td>19.2 ± 8.8</td>
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<tr>
<td>30°</td>
<td>20.8 ± 16.4</td>
<td>16.7 ± 13.1</td>
<td>22.3 ± 20.1</td>
<td>17.2 ± 13.8</td>
<td>19.9 ± 19.5</td>
<td>18.2 ± 14.4</td>
<td>17.7 ± 14.0</td>
<td>17.6 ± 11.3</td>
<td>15.4 ± 11.9</td>
<td>13.0 ± 9.4</td>
<td>18.1 ± 14.4</td>
<td>13.5 ± 7.4</td>
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<tr>
<td>45°</td>
<td>22.2 ± 19.6</td>
<td>21.2 ± 21.5</td>
<td>21.1 ± 19.2</td>
<td>20.5 ± 17.4</td>
<td>20.8 ± 17.4</td>
<td>21.2 ± 19.5</td>
<td>21.2 ± 19.5</td>
<td>18.6 ± 16.8</td>
<td>20.1 ± 19.5</td>
<td>14.7 ± 12.6</td>
<td>19.8 ± 20.4</td>
<td>14.3 ± 13.3</td>
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<tr>
<td>60°</td>
<td>21.5 ± 19.8</td>
<td>20.7 ± 23.7</td>
<td>21.3 ± 20.5</td>
<td>25.1 ± 25.0</td>
<td>21.5 ± 19.0</td>
<td>19.0 ± 16.2</td>
<td>19.1 ± 18.6</td>
<td>20.5 ± 17.0</td>
<td>17.2 ± 17.9</td>
<td>15.1 ± 15.6</td>
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### Table 2: Mean loaded graft force ± standard deviation (N) by posterior tibial slope and knee flexion angle

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<th>16°</th>
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<tbody>
<tr>
<td>0°</td>
<td>59 ± 16.1</td>
<td>57.8 ± 18.6</td>
<td>59.8 ± 19.3</td>
<td>60.6 ± 14.1</td>
<td>62.1 ± 20.3</td>
<td>70.4 ± 20.7</td>
<td>67.7 ± 16.8</td>
<td>65.0 ± 11.2</td>
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<tr>
<td>15°</td>
<td>22.7 ± 9.0</td>
<td>22.7 ± 10.2</td>
<td>22.9 ± 5.0</td>
<td>21.9 ± 6.1</td>
<td>22.1 ± 5.8</td>
<td>22.4 ± 6.9</td>
<td>21.4 ± 12.7</td>
<td>20.7 ± 11.2</td>
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Contributions of each experimental factor were assessed using a two-factor mixed-effects model for each state. According to the BIC, the best model included a linear effect for tibial slope and no interaction term between slope and flexion angle. This allowed us to interpret and visualize the effects.
of posterior tibial slope and flexion angle upon graft force independently for both the loaded and unloaded states (Figure 4).

Numerical model results are presented in Table 3. The model for unloaded testing found an independently significant, linearly decreasing effect on graft force regardless of flexion angle (coefficient $= -0.23$, SE = 0.07, $p = 0.002$). Most importantly, during testing with the axial load, tibial slope had an independently significant, linearly increasing effect on graft force regardless of flexion angle (coefficient $= 0.92$, SE = 0.08, $p < 0.001$). Meanwhile, significantly higher graft force was observed at zero degrees flexion compared to all other flexion angles for both the loaded (all $p < 0.001$) and unloaded (all $p < 0.001$) conditions.

**Figure 4.** Modeled independent effects plots for two linear mixed-effects models: A1 - tibial slope effect on ACL graft in axially loaded knees; A2 - knee flexion effect on ACL graft in axially loaded knees; B1 –
tibial slope effect on ACL graft in unloaded knees; B2 - knee flexion effect on ACL graft in unloaded knees. Tibial slope effects plots assume flexion was held constant at the average flexion effect, while flexion effects plots assume tibial slope was held constant at 9°. Shaded region and error bars represent the 95% confidence region for continuous predictors and 95% confidence intervals for factor predictors, respectively.

Table 3. Summary of two-factor linear mixed-effects models for unloaded and loaded knee conditions. Baseline status is a knee with 0° of posterior tibial slope at 0° of flexion, and coefficients indicate how the linear relationship changed as slope and flexion were altered.

**Discussion**

The most important finding of this study was that increased tibial slope had a significantly positive linear effect on ACL graft force when the knee was axially loaded. This increasing effect was consistent after adjusting for flexion angle. There was also a minimal negative relationship with tibial slope and the forces experienced by the unloaded ACL graft. Overall, across all flexion angles tested, we found that our null hypothesis was disproven because ACL graft forces significantly increased when loaded as posterior tibial slope increased.

Our finding that decreased posterior tibial slope reduces the forces experienced by the ACL graft supports recent clinical literature reporting that slope-reducing anterior wedge osteotomy can improve long-term patient outcomes, and may be beneficial in protecting the ACLR graft in cases of second revision surgery. In these cases, a slope-decreasing osteotomy has been reported to result in
improved re-revision ACL graft outcomes when altering the native slopes from 13.2° to 4.4°, and 13.6° to 9.2°, respectively. Our results corroborate these clinical findings, where slopes that are increased caused substantially increased graft force and flatter tibial slopes caused a decrease in ACL graft force. Overall our findings strengthen the indication to perform a slope-decreasing proximal tibial osteotomy to protect an ACL graft with either a biplanar or closing wedge osteotomy. When considering clinical and biomechanical data together, based upon the findings in our study, the goal of a slope decreasing osteotomy should be to decrease the tibial slope to <6° in those circumstances.

Our study evaluated the effect of both increased and decreased sagittal tibial slope on ACL graft loading at varying flexion angles with an applied axial force. Previous testing of ATT or ACL loads with increased or decreased slope has mainly been conducted using the native ACL. Previous work has noted that increased posterior tibial slope led to increased ATT and increased native ACL strain. Yamaguchi et al. evaluated ATT and native ACL strain and reported less strain on the ACL and less ATT with decreased tibial slope. Overall, our study confirmed previous findings of the impact of tibial slope on native ACL force and expanded this data to include the impact of both increased and decreased tibial slope on ACLR graft loading.

Increased posterior tibial slope is one of multiple factors which has been reported to contribute to ACLR graft failure. Previous studies have noted that posterior tibial slope greater than 8.4° in one study and 12° in a further study has led to an increased risk of ACLR failures clinically. Salmon et al. reported that at 20 years postoperatively, patients with a posterior tibial slope of greater than 12° had an 11 times higher rate of ACLR graft failure. Our findings validate the findings of these clinical studies and may highlight the need to increase the frequency of slope reducing proximal tibial osteotomies prior to or concurrent with revision ACLR in patients with a tibial slope of >12°. Furthermore, the results of this study reporting that graft force was significantly higher at full extension may help with the management of athletes during rehabilitation and when returning to play. These results suggest that
Training strategies should be employed to control knee flexion during high-risk maneuvers such as pivoting or jump-landing, especially for those who have increased posterior tibial slope.

We acknowledge some limitations to this biomechanical study. This was a cadaveric study where inherent limitations occur while conducting testing at time-zero and biological healing effects cannot be replicated. Due to the multiple testing steps for both flexion angles and degrees of slope, laxity in the surrounding soft tissue can occur. However, we tried to limit both soft tissue laxity and time-variability of graft stiffness by randomized the order of testing for knee flexion angles and degrees of slope for each specimen. We also recognize that a 200 N axial loading force is less than what is experienced in vivo; however, we chose to utilize this protocol to reduce the incidence of fracturing the anterior tibial cortex hinge and to maintain consistency with prior biomechanical studies. Another limitation of this study was that plain radiographs were utilized to measure the native tibial slope. Three-dimensional imaging, such as MRI, may provide more detailed information about the degree of tibial slope in patients. However, previous studies have reported that standard radiographic measurements are highly reliable and reproducible."

Conclusion

We found that tibial slope has a strong linear relationship to the amount of graft force experienced by an ACL graft. Thus, a flatter tibial slope had significantly less loading of ACL grafts, while steeper slopes increased ACL graft loading. Our biomechanical findings support recent clinical evidence of increased ACL graft failure with increased tibial slope secondary to increased graft loading.

References


