Tibial Slope and Its Effect on Force in Anterior Cruciate Ligament Grafts

Anterior Cruciate Ligament Force Increases Linearly as Posterior Tibial Slope Increases

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Background: Previous work has reported that increased tibial slope is directly correlated with increased anterior tibial translation, possibly predisposing patients to higher rates of anterior cruciate ligament (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 ACL reconstruction (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 linearly as slope increased. This effect was independent of flexion angle. The final model utilized a 2-factor linear mixed-effects regression model and 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, $P < .001$). Significantly higher graft force was also observed at 0° of flexion as compared with all other flexion angles for the loaded condition (all $P < .001$).

Conclusion: The authors found that tibial slope had 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 steeper 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 before 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.

Keywords: tibial slope; anterior cruciate ligament reconstruction; anterior tibial translation; ACL graft forces; closing wedge osteotomy

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 affect the rate of subsequent ACL reconstruction (ACLR) graft failure.[§] Native posterior tibial slope has been described to average approximately 7° to 10°.[10,20] Previous work reported that an increased sagittal plane tibial slope >12° is directly

The American Journal of Sports Medicine
2019;47(2):296–302
DOI: 10.1177/0363546518820302

[§] References 1, 5, 6, 9, 10, 12, 13, 15, 21, 24.

correlated with increased anterior tibial translation (ATT), predisposing patients to ACL tears and reportedly higher rates of ACL graft failures over the long term.[7] Therefore, it has been suggested that decreasing the posterior tibial slope of patients with increased sagittal plane tibial slope could potentially protect ACLR grafts and reduce the risk of failed revision ACLR surgery among patients with high amounts of posterior tibial slope.[6,19,26]

The native and reconstructed ACL serves as a primary restraint to ATT at lower flexion angles between 0° and 30°. The determination of ACL loading at lower flexion angles among those with increased tibial slope may help aid 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 guide surgeons when deciding if a slope-reducing proximal tibial osteotomy should be performed before a revision ACLR.

While the clinical effects of increased tibial slope have been reported, there is a paucity of biomechanical information on the effect 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.

METHODS

Specimen Preparation

Ten male fresh-frozen cadaveric knees were used (mean age, 53 years [range, 33-64 years]; mean body mass index, 23.3 kg/m^2 [range, 16.2-36.9 kg/m^2]). Specimens with prior surgery or evidence of meniscal, cartilage, or 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 with 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 with poly(methyl methacrylate) (Fricke Dental International) with the tibial plateau oriented parallel to the base.

Surgical Technique

The ACL was reconstructed by an anatomic single-bundle technique as previously reported.[11,16] 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 with a previously validated technique.[23] The native baseline tibial slope was defined as the angle between the medial tibial plateau and a line parallel to the middiaphysis of the tibia and was measured with radiographs. The tibial middiaphyseal line was centered through the tibial shaft with 2 lines, one 5 cm distal to the joint line and one 15 cm distal to the joint line. The midpoint of these 2 lines represented the middiaphyseal line, and a line was drawn parallel to the tibial plateau. The angle between these lines was subtracted from 90° to calculate the resultant tibial slope (Figure 1).[23]

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 that the osteotomy did not break through the anterior cortex of the tibia while leaving a 5- to 6-mm anterior bone hinge. Based on 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) 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. Before testing, tibial slopes were measured fluoroscopically on each specimen at 5 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 10 times to ensure proper conditioning to minimize creep during testing.[14] After preconditioning, the ACL graft was fixed in the femoral tunnel with an 8 × 20–mm interference screw (Smith & Nephew). The ACL graft consisted of 2 semitendinosus and 2 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

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One or more of the authors has declared the following potential conflict of interest or source of funding: R.F.L. receives royalties from Arthrex and Smith & Nephew; is a paid consultant for Arthrex, Ossur, and Smith & Nephew; and receives research support from Arthrex, Linvatec, Ossur, and Smith & Nephew. AOSSM checks author disclosures against the Open Payments Database (OPD). AOSSM has not conducted an independent investigation on the OPD and disclaims any liability or responsibility relating thereto.

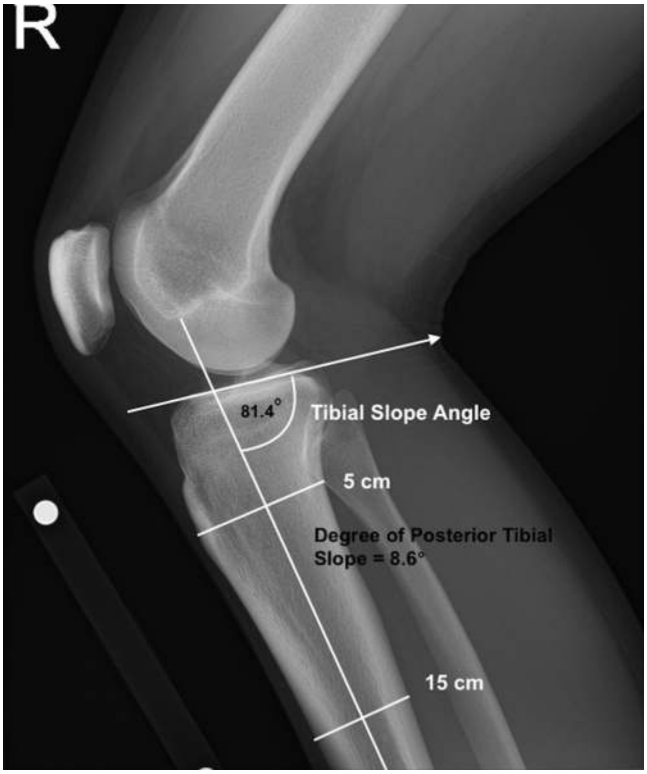

Figure 1. Lateral radiograph demonstrating described technique for calculating sagittal plane tibial slope. The resultant tibial slope angle was subtracted from 90° to determine the posterior tibial slope (in degrees).

external load cell sensor (Sensortronics; Vishay Precision Group). Load sensor calibration was confirmed before testing by comparing sensor output with the output from a calibrated dynamic tensile testing machine (ElecroPuls E10000; Instron) at 5-N load increments ranging from 5 to 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 midshaft of the tibia was oriented 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.[17] The 10-mm rod acted as the load-bearing pivot axis. Next, a 7-mm rod was passed through the distal femoral shaft to fixate the knee at the desired 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. For each time that the tibial slope was changed, the graft was retensioned to 88 N with the knee in full extension. The unloaded force was the force recorded in the ACL graft before axial

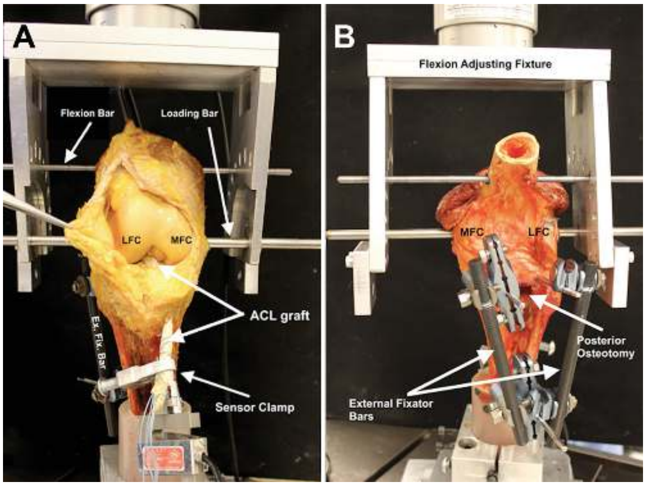

Figure 2. (A) Anterior and (B) posterior views of testing setup in a right knee. ACL, anterior cruciate ligament; LFC, lateral femoral condyle; MFC, medial femoral condyle.

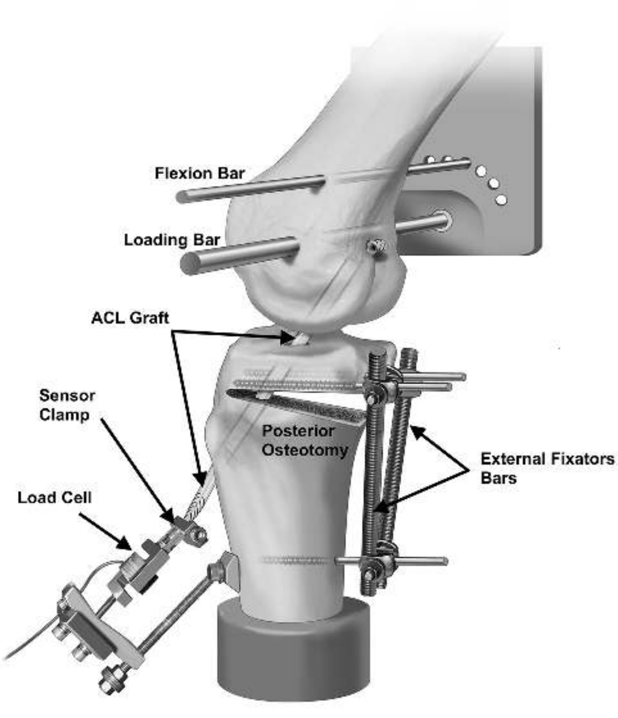

Figure 3. Schematic representation of the mechanical testing setup for a right knee. ACL, anterior cruciate ligament.

TABLE 1
Unloaded Graft Force by Posterior Tibial Slope and Knee Flexion Angle[a]

Slope	0°	15°	30°	45°	60°
−2°	70.8 ± 16.6	29.4 ± 12.2	20.8 ± 16.4	22.2 ± 19.6	21.5 ± 19.8
0°	67.3 ± 19.3	28.2 ± 12.4	16.7 ± 13.1	21.2 ± 21.5	20.7 ± 23.7
2°	65.6 ± 16.5	24.9 ± 12.3	22.3 ± 20.1	21.1 ± 19.2	21.3 ± 20.5
4°	68.0 ± 13.8	25.8 ± 11.0	17.2 ± 13.8	20.5 ± 17.4	25.1 ± 25.0
6°	69.2 ± 17.4	22.3 ± 8.1	19.9 ± 19.5	18.0 ± 17.4	19.0 ± 16.2
8°	70.4 ± 13.0	21.6 ± 7.7	18.2 ± 14.4	18.1 ± 17.4	19.8 ± 20.4
10°	74.6 ± 16.8	23.1 ± 10.7	17.7 ± 14.0	17.5 ± 13.8	20.5 ± 17.0
12°	65.0 ± 11.2	20.7 ± 11.2	17.6 ± 11.3	18.6 ± 16.8	19.1 ± 18.6
14°	67.7 ± 16.8	21.4 ± 12.7	15.4 ± 11.9	20.1 ± 19.5	17.2 ± 17.9
16°	68.5 ± 16.1	17.8 ± 9.0	13.0 ± 9.4	14.7 ± 12.6	15.1 ± 15.6
18°	80.4 ± 14.1	23.1 ± 10.0	16.1 ± 9.6	15.3 ± 12.9	16.6 ± 15.8
20°	71.4 ± 13.7	19.2 ± 8.8	13.5 ± 7.4	14.3 ± 13.3	14.4 ± 16.8

[a]Values are presented as N, mean ± SD.

compression, and the loaded force was the force 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.

Statistical Methods

Two-factor linear mixed-effects models were used to assess the effects of posterior tibial slope and knee flexion angle on ACL graft force for 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.[4] The Tukey 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 (accessed August 17, 2018; R Foundation for Statistical Computing with lme4).[3,18] As a simplification of the full analysis, statistical power and sample size were 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. With an assumed alpha level of .05, 10 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 2-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 the loaded and unloaded graft forces that were seen in the ACL graft before and after axial loading. Means and SDs are presented for each combination of posterior tibial slope and knee flexion angle for the loaded and unloaded states in Tables 1 and 2, respectively.

Contributions of each experimental factor were assessed with a 2-factor mixed-effects model for each state. According to the Bayesian information criterion, 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 the loaded and unloaded states (Figure 4).

Numerical model results are presented in Table 3. The model for unloaded testing revealed an independently significant, linearly decreasing effect on graft force regardless of flexion angle (coefficient = -0.23, SE = 0.07, $P = .002$). Most important, 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, $P < .001$). Meanwhile, significantly higher graft force was observed at 0° of flexion as compared with all other flexion angles for the loaded (all $P < .001$) and unloaded (all $P < .001$) conditions.

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 ACLR graft supports recent

TABLE 2
Loaded Graft Force by Posterior Tibial Slope and Knee Flexion Angle[a]

Slope	0°	15°	30°	45°	60°
−2°	59 ± 16.1	22.7 ± 9.0	15.2 ± 11.4	18.7 ± 15.2	21.3 ± 20.4
0°	57.8 ± 18.6	22.7 ± 10.2	13.6 ± 8.2	18.8 ± 16.1	22.6 ± 22
2°	59.8 ± 19.3	22.9 ± 5.0	18.2 ± 9.7	20.1 ± 13.0	23.7 ± 18
4°	60.6 ± 14.1	21.9 ± 6.1	19.8 ± 7.2	23.5 ± 11.0	30.2 ± 20.8
6°	62.1 ± 20.3	22.1 ± 5.8	22.6 ± 10.9	19.9 ± 10.2	25.1 ± 17.3
8°	70.4 ± 20.7	22.4 ± 6.9	23.7 ± 8.8	22.8 ± 16.7	29.6 ± 20.3
10°	72.4 ± 18.2	29.2 ± 6.0	28.6 ± 13.2	28.5 ± 13.3	33.7 ± 20.5
12°	64.7 ± 15.3	29.2 ± 9.8	26.5 ± 7.7	28.3 ± 10.2	28.7 ± 19.4
14°	70.2 ± 19.6	29.8 ± 12.1	29.8 ± 12.3	33.6 ± 16.3	31.2 ± 20.7
16°	72.3 ± 19.4	34.3 ± 10.3	32.7 ± 11.3	31.3 ± 9.3	34 ± 16.2
18°	85.2 ± 18.5	42.3 ± 13.9	38.4 ± 11.7	38.7 ± 10.6	39.7 ± 16.4
20°	78 ± 20.0	41.1 ± 16.6	37.5 ± 16.2	34.8 ± 15.3	31.6 ± 16.2

[a] Values are presented as N, mean ± SD.

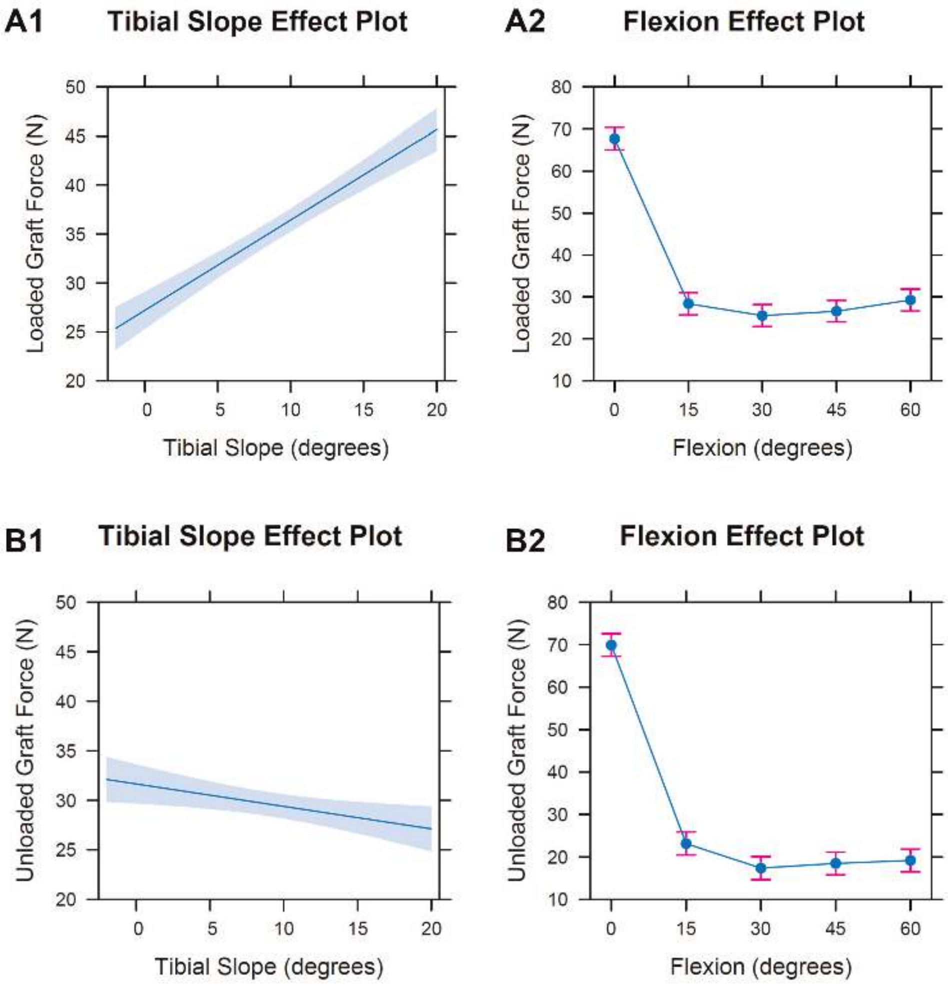

Figure 4. Modeled independent-effects plots for 2 linear mixed-effects models: (A1) tibial slope effect on ACL graft in axially loaded knees, (A2) knee flexion effect on ACL graft in axially loaded knees, (B1) tibial slope effect on ACL graft in unloaded knees, and (B2) knee flexion effect on ACL graft in unloaded knees. Tibial slope effect plots assume that flexion was held constant at the mean flexion effect, while flexion effect plots assume that tibial slope was held constant at 9°. Shaded region and error bars represent the 95% confidence region for continuous predictors and 95% CIs for factor predictors, respectively. ACL, anterior cruciate ligament.

TABLE 3
Two-Factor Linear Mixed-Effects Models for Unloaded and Loaded Knee Conditions[a]

	Model A: Loaded Knee				Model B: Unloaded Knee			
	Coefficient	SE	*t* Value	*P* Value (2-Tailed)	Coefficient	SE	*t* Value	*P* Value (2-Tailed)
(Intercept)	59.41	2.01	29.506	<.001	71.96	3.1	23.22	<.001
Slope	0.92	0.08	11.369	<.001	−0.23	0.07	−3.189	.002
Flexion								
15°	−39.33	1.77	−22.198	<.001	−46.77	1.55	−30.108	<.001
30°	−42.16	1.77	−23.796	<.001	−52.55	1.55	−33.825	<.001
45°	−41.12	1.77	−23.211	<.001	−51.44	1.55	−33.113	<.001
60°	−38.44	1.77	−21.697	<.001	−50.73	1.55	−32.658	<.001

[a]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.

clinical literature reporting that slope-reducing anterior wedge osteotomy could improve long-term patient outcomes and may be beneficial in protecting the ACLR graft in cases of second revision surgery.[6,22] In these cases, a slope-decreasing osteotomy was reported to result in improved re-revision ACL graft outcomes when the native slopes were altered from 13.2° to 4.4° and from 13.6° to 9.2°.[6,22] 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 a biplanar or closing wedge osteotomy. When clinical and biomechanical data are considered together based on 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 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 was mainly conducted with the native ACL.[1,8,10,26] Previous work noted that increased posterior tibial slope led to increased ATT and increased native ACL strain.[1,8,10] Yamaguchi et al[26] 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 effect of tibial slope on native ACL force and expanded these data to include the effect of increased and decreased tibial slope on ACLR graft loading.

Increased posterior tibial slope is 1 of multiple factors that has been reported to contribute to ACLR graft failure.[15] Previous studies noted that posterior tibial slope >8.4° and >12° led to an increased risk of clinical ACLR failure.[5,19] Salmon et al[19] reported that at postoperative 20 years, patients with a posterior tibial slope >12° had an 11-times higher rate of ACL 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 before or concurrent with revision ACLR among patients with a tibial slope >12°.[19,25] Furthermore, the results of this study, reporting that graft force was significantly higher at full extension, may help with the treatment 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 during testing at time zero and where biological healing effects cannot be replicated. Owing to the multiple testing steps for flexion angles and degrees of slope, laxity in the surrounding soft tissue can occur. However, we tried to limit soft tissue laxity and time variability of graft stiffness by randomizing 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 magnetic resonance imaging or computed tomography, may provide more detailed information about the degree of tibial slope in patients. However, it was reported that standard radiographic measurements are highly reliable and reproducible.[2] One other issue to consider in decreasing tibial slope is the potential increase in knee hyperextension, especially for those patients with preexisting genu recurvatum. In those cases, one needs to carefully weigh the benefits of decreasing tibial slope with the potential negative effects of increased genu recurvatum on the ACLR graft.

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.

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