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THE EFFECTS OF MODIFIED POSTERIOR TIBIAL SLOPE ON ACL STRAIN AND KNEE KINEMATICS: A HUMAN CADAVERIC STUDY

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Abstract

Increases to the posterior tibial slope can lead to an anterior shift in tibial resting position. However, the effect of this shift on anterior cruciate ligament (ACL) strain has not been investigated sufficiently. This study examined the relationship between increased tibial slope and ACL strain, as well as the subsequent kinematics of the tibiofemoral joint. We hypothesized increases in slope would shift the tibia anterior relative to the femur and increase ACL strain. ACL strain measurements and tibiofemoral kinematics were compared for 5 intact and experimental knees subject to anterior opening wedge osteotomy. Combinations of both compressive and AP loading were applied. As slope increased, the resting position of the tibia shifted anteriorly, external tibial rotation increased, and tibial translation remained unchanged. Contrary to our hypothesis, ACL strain decreased. The clinical implication of these findings is that alterations to the posterior tibial slope should not increase strain in the ACL.

Keywords

Osteotomy; ACL; knee biomechanics; basic science

Introduction

Medial opening wedge osteotomy frequently is performed to correct coronal deformities in the knee joint.[2] Corrections intended to alter only the alignment of the coronal plane typically augment the slope of the tibial plateau as well, with increases to posterior tibial slope of up to 10° reported.[17] In addition, intended alterations, from anteromedial opening wedge osteotomy for treatment of either posterolateral instability or anterior opening wedge osteotomy to help correct posterior cruciate ligament deficient knees, also increase posterior tibial slope. Increases in posterior tibial slope have been shown to directly lead to increased anterior tibial translation.[9,12] Thus, considering the anterior cruciate ligament (ACL) supports between 80% and 90% of anteriorly applied tibial loads,[8] it seems intuitive that osteotomy-induced increases

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in posterior tibial slope would increase ACL loading, hence placing it at increased risk for failure.

In the native knee, greater posterior tibial slope has been correlated positively with ACL rupture in both men and women.[7] Giffin et al[12] were the first to study the effects of increased posterior tibial slope on cruciate ligament forces. In their study, a robotic testing system was used to examine the effects of combinations of compressive loading (200 N) and AP loading (134 N), which are magnitudes of force typical in clinical examinations, over the full range of knee motion. Posterior tibial slope was increased approximately 5° via a 5-mm anterior opening wedge osteotomy. However, in direct opposition to their initial hypothesis, increases in posterior tibial slope did not lead to a concomitant increase in ACL loading. Small anterior translations, as a result of low loading conditions, were proposed as the likely reason for this observation, with larger changes in ACL loading purported to occur for greater increases in tibial slope. However, this tenet was not examined explicitly in the study. A lack of significant findings by Giffin et al[12] also may have stemmed from the fact that data were obtained for relatively low joint loading states. It may be that reasonable increases in anterior tibial translation, and thus ACL loading, would be more likely under higher load states that are more representative of physiological loading conditions.

Given these facts, our study was undertaken to extend the previous research of Giffin et al[12] to determine whether greater changes in posterior tibial slope, in conjunction with increased external load states, significantly impacts ACL strain. A further objective was to evaluate the effect of increased slope on knee kinematics, with measures of anterior tibial translation, anterior tibial shift, and axial knee rotation. We hypothesized that increasing tibial slope, by increasing osteotomy plate size, would significantly increase ACL strain. In addition, we expected increased osteotomy plate size would significantly increase anterior tibial shift and externally rotate the knee axially without significantly increasing anterior tibial translation.

Materials and Methods

Five fresh-frozen adult cadaveric knee specimens were used in this study. Specimens were thawed overnight at room temperature before instrumentation and testing. Each specimen consisted of all bone and soft tissue approximately 15 cm above and below the joint line. All specimens appeared normal upon visual and fluoroscopic inspection. The femur and tibia were potted in aluminum tubing with polymethylmethacrylate (Patterson Dental, St Paul, Minn) and transfixed bicortically with perpendicular transfixion pins. Each knee was mounted into a custom loading apparatus capable of producing variable compressive and AP-directed tibial shear loads relative to the femur. The tibial cylinder was rigidly secured to a concrete floor, and the femoral cylinder was secured to the loading apparatus frame at 15° of flexion to approximate the typical knee angles of mid-stance during walking (Figure 1).[14]

To quantify ligament load in terms of strain, a microminiature differential variance reluctance transducer (DVRT) (Microstrain Inc, Williston, Vt) was secured to the anteromedial bundle of the ACL. The anteromedial bundle was chosen to enable direct comparison between previous studies and current data[3] and to minimize the risk of gauge impingement.[5] A 3-cm vertical incision was made first on the anteromedial aspect of the knee joint, at the level of the joint space. Prior to gauge attachment, a femoral notchplasty (1 cm) was performed to further reduce the risk of gauge impingement. The DVRT was secured to the inferior portion of the anteromedial bundle midsubstance via a gauge insertion tool (Microstrain Inc). Each specimen was preconditioned prior to testing by moving it through 10 complete flexion-extension cycles[13] and repositioned to 15° of flexion with respect to the shafts of the femur and tibia.

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A set of predetermined joint forces, based on previously published studies,[4,15,16,21,22] were applied to the knee using the custom-made loading apparatus via a system of weights and pulleys (Figure 1). Specifically, there were 3 levels of AP load (18 N, 108 N, and 209 N) and 2 levels of compressive load (216 N and 418 N). Loading was applied at each of the 3 plating heights: native, 5-mm plate, and 10-mm plate. The loading protocol was randomized, with the only restrictions on randomization being that each test series began with the intact native knee prior to osteotomy (native plating height) and that the knee was in a relaxed state (0 N of compression load and 18 N of AP load) at the beginning and end of each testing sequence. All possible combinations of these 6 loads were applied to knee specimens at each of the 3 plating conditions (native, 5 mm, and 10 mm).

The anterior opening wedge osteotomy procedures were preformed via a surgical technique recommended by the osteotomy plating system (Arthrex Inc, Naples, Fla). A longitudinal incision was made along the medial border of the patellar tendon. Through the incision, the osteotomy site was prepared on either side of the patellar tendon insertion. The tibial tuberosity was not detached, and all peripheral capsuloligamentous structures attached to the epiphysis were left intact. The osteotomy was started approximately 4 cm distal to the joint line just superior to the tibial tuberosity. The cut was performed with an oscillating saw and routed slightly superior to finish distal to the posterior cruciate ligament and capsular insertion below the level of the joint line.

The osteotomy was distracted open against the posterior cortical hinge using the Arthrex guide. Lateral radiographs verified that the posterior cortex remained intact. A 5-mm or 10-mm 4-hole Arthrex plate was positioned along the medial side of the patellar tendon. With the knee in extension, the plate was fixed proximally with 3 cancellous screws and distally with 2 cortical screws.

The DVRT strain gauge was connected via a fine wire cable to a signal conditioner (Microstrain Inc) and relayed to a data acquisition board (National Instruments, Austin, Tex), where the signals were converted from analog to digital. These microminiature DVRT sensors had a resolution of 2 μm and a typical gauge length of 8.5 mm. Output signals were displayed and collected at a sampling rate of 100 Hz using a custom MatLab (Mathworks Inc, Natick, Mass) application. Relative zero strain was calculated for the ACL with the knee in what was assumed a relaxed state (0 N of compression load and 18 N of AP load) at 15° of flexion. Relative zero strain served as an estimate of absolute zero strain for the ensuing knee-specific strain comparisons. Strain gauges were retained in the ligament throughout testing on each specimen, allowing strain to be compared directly between different osteotomy plates within the knee. Strain was calculated using the following standard equation:

$$\varepsilon_k = \frac{y_k - y_0}{y_0}$$

where y_k and e_k correspond to the length of the DVRT and the associated ACL strain estimate for the k^{th} data sample, respectively, and y0 corresponds to the DVRT length at a relaxed state.

Overhead digital photography was used to capture the precise two-dimensional location of the markers in the axial plane for each plate and load condition. The proximal tibia was immobilized, and 2 markers were rigidly attached to the distal femur and aligned medially and laterally at the level of the joint line to assess tibial motion (Figure 1). Motion calculations were based on the deviation in position of the midpoint between these markers from the baseline relaxed load (0 N of compression load and 18 N of AP load) and intact joint state. Analysis of tibiofemoral joint motion was separated specifically into anterior tibial shift (change in resting position of the tibiofemoral joint after osteotomy), anterior tibial translation (AP distance

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between the baseline measurement and that resulting from each loading protocol), and axial rotation (external tibial rotation relative to the baseline position measurement).

To evaluate changes to the tibial slope, lateral radiographs were obtained to quantify initial tibial slope prior to osteotomy and altered tibial slopes resulting from each osteotomy procedure (Figure 2). Tibial slope was calculated by measuring the angle between 2 reference lines, with 1 line approximating the posterior tibial slope and the other perpendicular to the diaphyseal axis.[9]

Data were analyzed using the Minitab (State College, Pa) statistical program. Stepwise linear regressions (forward and backward) were applied to these data to determine the statistical significance of each response from the control factors. The control factors were the osteotomy plate size (3 levels), AP load (3 levels), and compressive load (2 levels). Measured responses included anterior tibial shift, anterior tibial translation, axial rotation, and ACL strain. Interactions between the control factors also were evaluated, although the interactions were found to have no statistically significant effect on the responses. The randomized experimental design guarded against confounding experimental variables (plate and loadings) with any uncontrolled systemic variables. Uncontrolled variables contributed only to the overall test viability and not the trending in the measured responses due to the experimental variables. Consequently, the experimental design minimized the number of measurements needed to identify significant trending in the measured responses, while controlling for the effect of any uncontrolled variable. An alpha level of 0.05 was adopted for all statistical treatments to denote statistical significance.

Results

The native tibial slope and changes resulting from 5-mm and 10-mm osteotomy plates are shown in the Table. Mean (±SD) posterior slope of the 5 native knees was 12.1°±2.4°. The 5-mm osteotomy plate resulted in a mean tibial slope of 15.6°±2.6°, an increase from the mean native slope of 3.5°±0.6°. The 10-mm osteotomy plate resulted in a mean posterior tibial slope of 21.7°±3.0°, an increase from the mean native slope of 9.6°±2.0°. A significant relationship was identified between tibial slope and osteotomy plate size ($P < .001$).

Anterior tibial motion was separated into anterior tibial shift and anterior tibial translation. Anterior tibial shift, defined as the change in resting position of the tibiofemoral joint after osteotomy, was influenced significantly only by plate size ($P < .05$). Specifically, anterior tibial shift increased from 1.44±0.77 mm in the native knee to 6.15±4.80 mm and 9.23±7.55 mm after osteotomy plating of 5 and 10 mm, respectively (Figure 3). Conversely, anterior tibial translation, an indication of the magnitude of tibiofemoral joint motion or joint laxity, was not significantly increased by plating condition (Figure 3). Rather, it was influenced significantly only by AP load ($P < .001$). The smallest AP load of 18 N resulted in 5.0±7.2 mm of translation, whereas larger loads of 108 and 209 N resulted in anterior tibial translation of 12.6±7.3 and 18.5±8.3 N, respectively.

Axial rotation is a measure of tibiofemoral rotation, where positive values represent external rotation of the tibia relative to the femur, as would be experienced during external knee rotation. Axial rotation was influenced significantly by the main effect of osteotomy plate size ($P = .011$). Axial rotation increased from 1.00°±2.90° in the native knee to 3.77°±0.04° and 4.14° ±5.71° after plating of 5 and 10 mm, respectively.

Strain data collection was limited to 4 specimens as 1 knee was excluded because of mechanical failure of a strain gauge during experimentation. Even with only 4 specimens, the use of an experimental design ensured that statistically significant correlations could be identified. Anterior cruciate ligament strain was statistically dependent on both plate size ($P = .014$) and

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AP load ($P < .001$). Specifically, strain in the ACL decreased from a native knee measure of 1.88±1.35% to 1.06±1.15% and -0.38±1.24% after plating of 5 and 10 mm, respectively (Figure 4A). In addition, ACL strain increased from -0.10±1.01% for 18 N of AP loading to 1.14 ±1.40% and 1.52±1.72% for increased loadings of 108 and 209 N, respectively (Figure 4B).

Discussion

Previous research has confirmed that increases to posterior tibial slope as a by-product of opening wedge osteotomy result in an anterior shift in the position of the tibia relative to the femur.[8,12] Giffin et al[12] found that under loading conditions typical of physical examinations, ACL strain was not impacted significantly by these tibial slope alterations. However, the extent to which this relationship held true for physiologic joint load states remained unclear. Therefore, our study was undertaken to examine the effect of larger alterations to posterior tibial slope, with increased loading conditions on ACL strain and resultant axial plane tibiofemoral joint motion.

Our results indicate an increased posterior tibial slope as a result of osteotomy plating leads to a shift in the resting position of the tibiofemoral joint but does not significantly affect tibiofemoral translation. This finding is consistent with the study by Giffin et al,[12] which also found an increase in resting position but no change in translation (under combined loading conditions) following smaller increases to posterior tibial slope. Our testing increased loading magnitudes and osteotomy plate sizes and therefore found larger magnitudes of anterior tibial shift.

Our findings also indicate increasing plate size leads to an external tibial rotation relative to the femur. Rotation of the tibia relative to the femur following opening wedge osteotomy is a topic that has not been specifically addressed previously, although it has been determined previously that the tibia rotates externally to the femur as the knee is extended.[18] When constraining the alignment of the femur relative to the tibia, as in our experimental setup, opening wedge osteotomy positions the contacting surfaces of the joint in an alignment representative of a knee in further extension (Figure 5). Therefore, it seems feasible that the observed external tibial rotation occurs as a result of this further extended interface.

Anterior cruciate ligament strain typically increases as the tibia shifts anteriorly to femur, a necessary translational by-product of moving the knee toward full extension. [8,19] Therefore, it seems somewhat intuitive that for a static joint position, increased anterior tibial shift, resulting from an anterior opening wedge osteotomy, would increase ACL strain.[10] However, in contrast to this hypothesis, Giffin et al[12] concluded there were no significant changes in ACL forces following osteotomy. In fact, they reported a reduction in the in situ forces in the ACL from 38±17 N to 3±18 N, although this decrease was not significant.[12]

We hypothesized increased plate size and increased force magnitudes would lead to increased ACL strain. However, similar to Giffin et al,[12] we found this to be untrue. We actually found increased osteotomy plate size significantly decreased ACL strain. Anterior cruciate ligament strain did increase with anterior tibial translation under loading (Figure 6). Although this trend was maintained for each osteotomy plating condition, there was a downward shift in the range of strain values following osteotomy procedures of increasing plate size. This shift demonstrates how ACL strain can decrease with increased posterior tibial slope yet retain the normal behavior of the ligament during anterior tibial movement.

We are unsure as to the precise reasons for the inverse relationship between tibial slope and ACL strain. Giffin et al[12] postulated changes to the anatomy may play a role, and we agree with this assessment. Recent studies comparing tibiofemoral contact pressures following increases to posterior slope have counterintuitively found the point of contact between the femur and tibia to shift anteriorly on the tibia as the tibia translates anteriorly to the femur.[1,20] It is possible that this shift in the tibiofemoral contact point is a function of the geometry of the femoral condyles, possibly bringing the attachment sites of the ACL closer to each other. This action would effectively increase ligament laxity while maintaining the same relative joint position.

In addition, the bony contacts and restraints appear to be forcing the knee into an external rotation as the knee shifts to its new point of contact following osteotomy. This change in axial rotation also may potentiate a reduction in ACL strain, as small changes in external tibial rotation are known to decrease ACL strain.[6] Finally, altered joint anatomy also may lead to increased loading in secondary tissue restraints, decreasing the contribution of the ACL. Further work is necessary to test each of these postulates and thus gain a precise understanding of the relationship between altered tibial slopes and resultant ACL loading.

Limitations

Although we are confident our observations can be translated into a realistic in vivo clinical environment, we acknowledge these observations have been made within several methodological constraints. One limitation of this study relates to the adoption of relative rather than absolute strain data to assess ACL loading. However, we are confident the strain gauge remained intact within the ligament throughout the entire testing procedure, allowing for precise and direct comparisons between osteotomy plating conditions within each specimen. To account for the use of relative strain data, our statistical analysis focused on evaluating specimen-specific changes and avoided direct specimen-to-specimen comparison. In the future, more complex and potentially accurate methods for de- fining zero strain will be explored[11]; however, we expect these methods will produce similar results.

A further potential limitation relates to our use of two-dimensional motion tracking and a single static knee angle. As previously described, an overhead transverse im- age was used to denote AP translation, anterior tibial shift, and axial knee rotation changes to each specimen. However, we acknowledge the knee may in fact be moving in a more complex three-dimensional motion pattern. In the future, we will evaluate the use of a three-dimensional motion capture system and robotics to load the specimen throughout the range of motion to accommodate for this potential confounding factor.

In addition, the knee is a dynamic joint. Thus, examining the relationship between tibial slope, anterior tibial translation, and ACL loading across a wider range of joint positions will likely provide additional insight into potential pathomechanical relationships between these factors. However, examining this relationship for a single joint position, as in our study, still provides valuable information regarding the potentially deleterious clinical implications of altering the posterior tibial slope with either an anterior or medial opening wedge osteotomy.

Conclusion

On the basis of current results, it appears inadvertent alterations of posterior tibial slope, as a by-product of medial opening wedge osteotomy, do not pose an increased threat to increased ACL loading. However, we are unsure whether this is a result of an increased role of bony or soft-tissue restraints, or a result of detensioning via external tibial rotation. Future studies are necessary to evaluate the change in orientation of the attachment points of the ACL and to address the possible contribution of other soft-tissue restraints.

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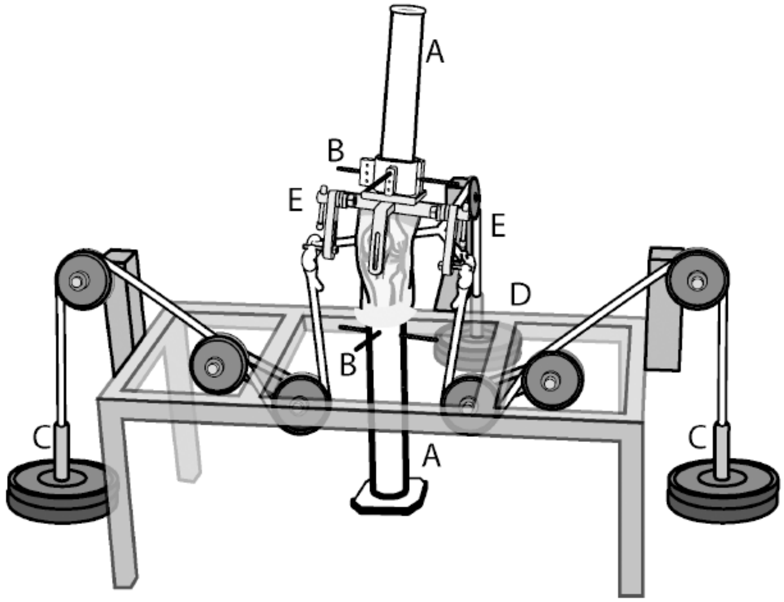

Figure 1.
Schematic showing the axial and AP loading of intact knee specimens before and after anterior opening wedge high tibial osteotomy (A, potted bone; B, transfixion pins; C, axial load application via weights; D, AP load application via weights; E, femoral markers). An overhead camera was used to track the motion of the femoral markers. Reprinted with the permission of the Cleveland Clinic Center for Medical Art & Photography © 2008. All Rights Reserved.

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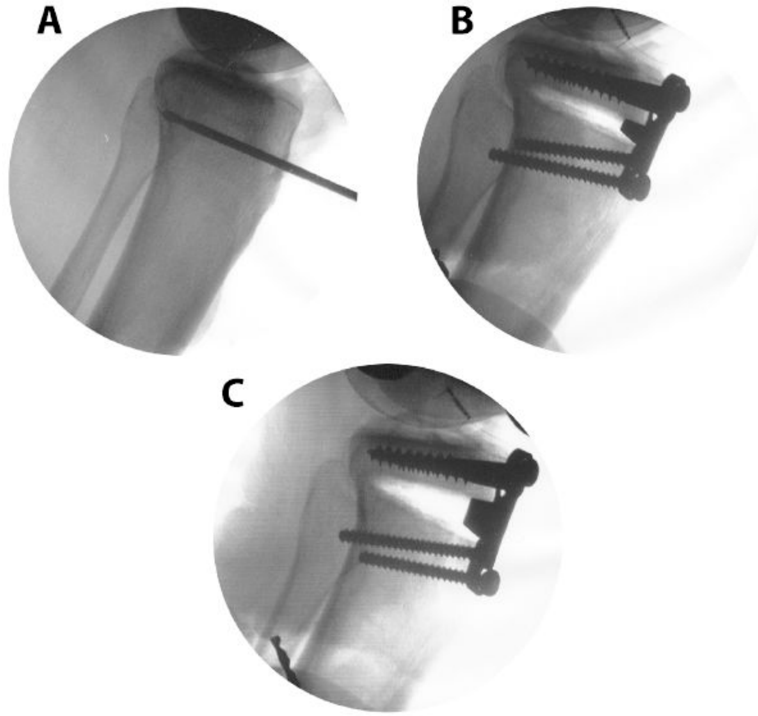

Figure 2.

Sagittal radiographs showing the effects of high tibial wedge osteotomy on tibial slope with an intact knee (A) and osteotomies with a 5-mm (B) and a 10-mm (C) anterior opening wedge plate. Reprinted with the permission of the Cleveland Clinic © 2008. All Rights Reserved.

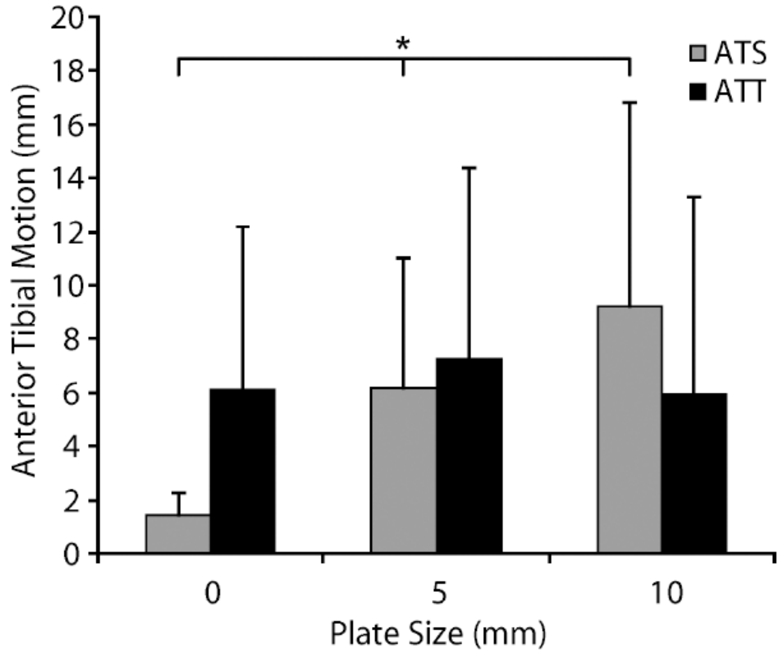

Figure 3.
Graph showing the effect of plate size on anterior tibial motion (mean6SD). Anterior tibial shift (ATS) was significantly dependant on plate size (*P < .05) whereas anterior tibial translation (ATT) was not. The 0-mm plate size indicates a native knee.

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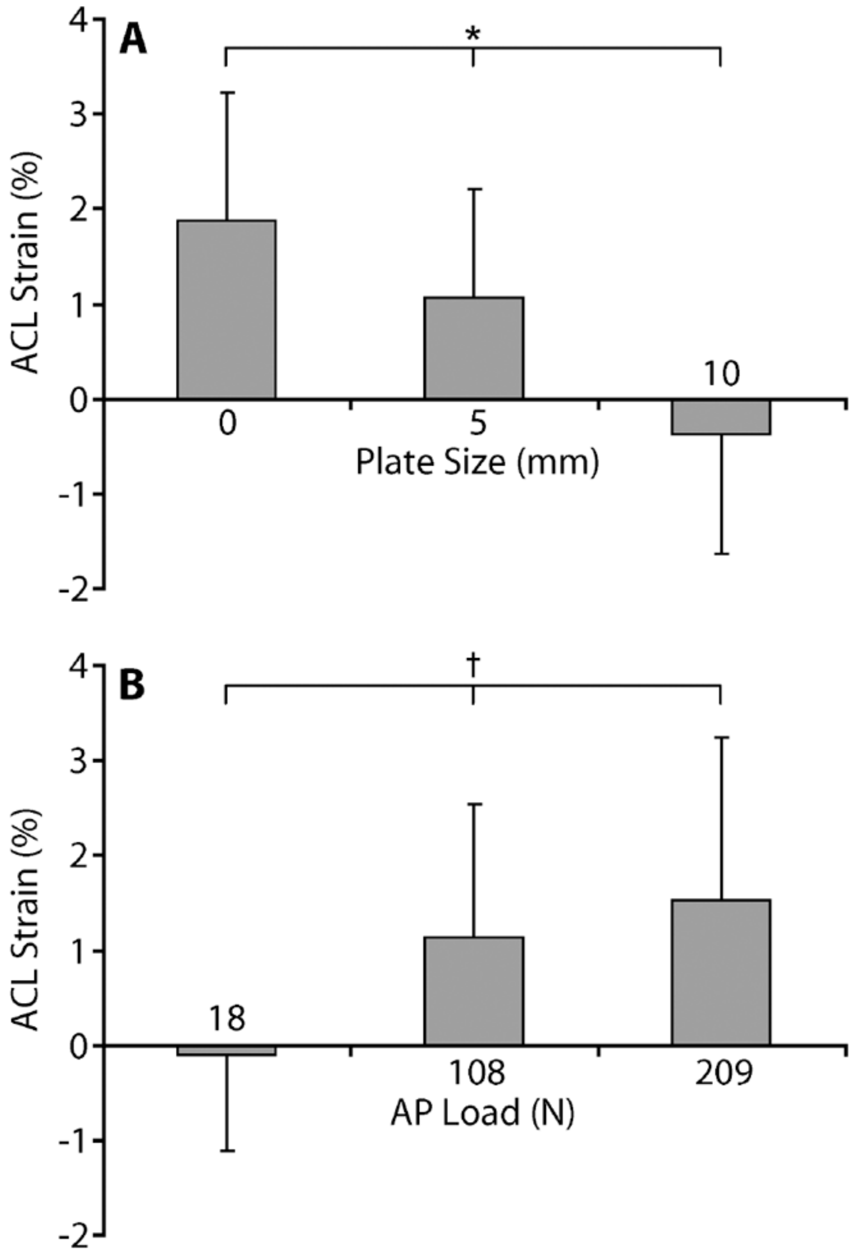

Figure 4.
Graph showing the effects of plate size and AP load on relative anterior cruciate ligament (ACL) strain (mean6SD). Anterior cruciate ligament strain decreased significantly as a function of osteotomy plate size (*P = .014) (A) and increased significantly as a function of AP loading (†P < .001) (B). The 0-mm plate size indicates a native knee.

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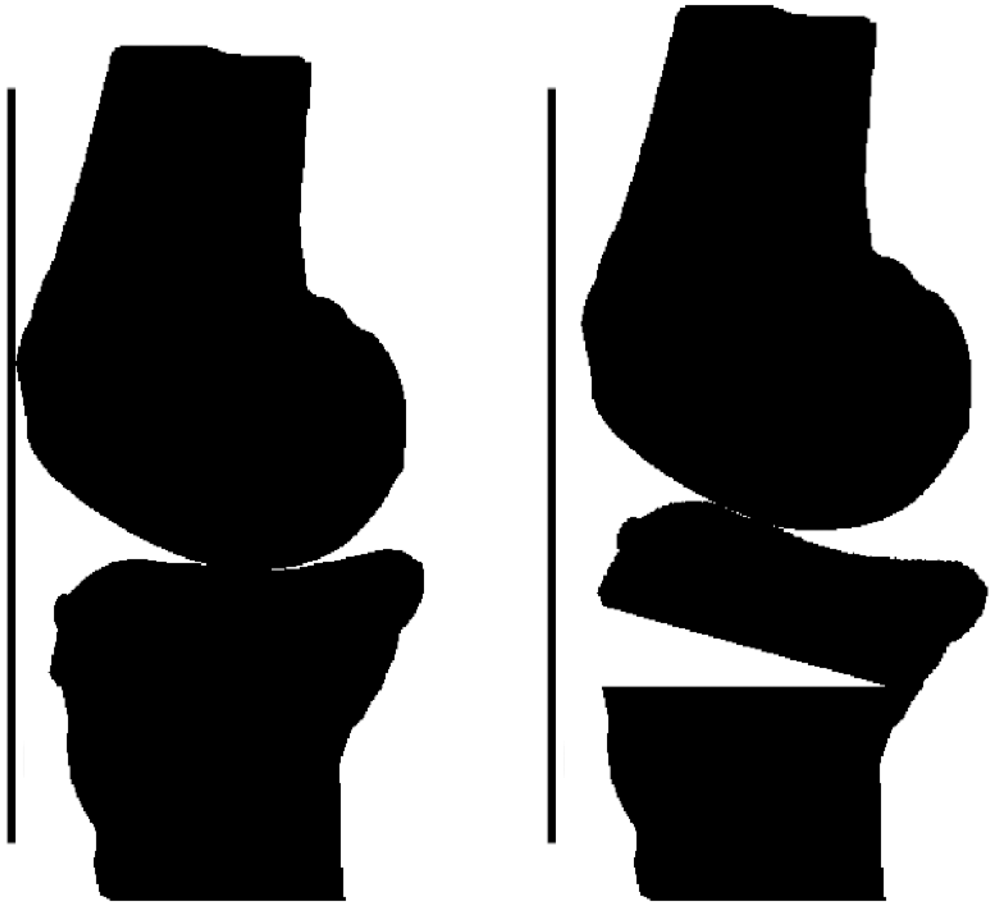

Figure 5.
Diagram shows that when constraining alignment of the femur relative to the tibia, an anterior opening wedge osteotomy places the articulating surfaces of the joint in a position mimicking further extension. Reprinted with the permission of the Cleveland Clinic © 2008. All Rights Reserved.

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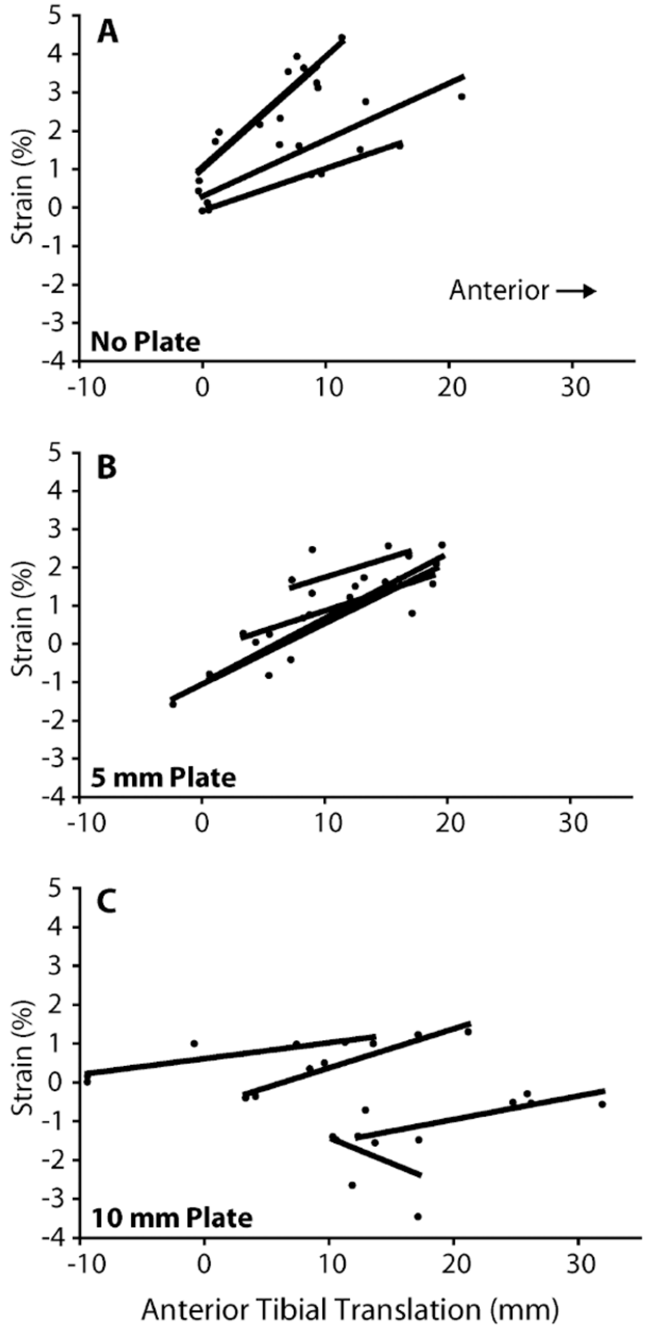

Figure 6.

Scatterplots showing anterior cruciate ligament strain as a function of anterior translation for the intact knee (A), and knees with 5-mm (B) and 10-mm (C) osteotomy plates. Regression lines shown are specific to each of the 4 specimens.

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TABLE

Changes in Tibial Slope as a Result of Osteotomy Plating

Knee Specimen	Native Slope (°)	5 mm Plate (°)	10 mm Plate (°)
1	13	16	20
2	12	15	24
3	15.5	20	24
4	12	15	24
5	8	12	16.5