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RESEARCH ARTICLE

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Increasing the posterior tibial slope lowers in situ forces in the native ACL primarily at deep flexion angles

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Abstract

High tibial osteotomy is becoming increasingly popular but can be associated with unintentional posterior tibial slope (PTS) increase and subsequent anterior cruciate ligament (ACL) degeneration. This study quantified the effect of increasing PTS on knee kinematics and in situ forces in the native ACL. A robotic testing system was used to apply external loads from full extension to 90° flexion to seven human cadaveric knees: (1) 200 N axial compressive load, (2) 5 Nm internal tibial + 10 Nm valgus torque, and (3) 5 Nm external tibial + 10 Nm varus torque. Kinematics and in situ forces in the ACL were acquired for the native and increased PTS state. Increasing PTS resulted in increased anterior tibial translation at 30° (1.8 mm), 60° (1.7 mm), and 90° (0.9 mm) flexion and reduced in situ force in the ACL at 30° (57.6%), 60° (69.8%), and 90° (75.0%) flexion in response to 200 N axial compressive load. In response to 5 Nm internal tibial + 10 Nm valgus torque, there was significantly less (39.0%) in situ force in the ACL at 90° flexion in the increased compared with the native PTS state. Significantly less in situ force in the ACL at 60° (62.8%) and 90° (67.0%) flexion was observed in the increased compared with the native PTS state in response to 5 Nm external tibial + 10 Nm varus torque. Increasing PTS affects knee kinematics and results in a reduction of in situ forces in the native ACL during compressive and rotatory loads at flexion angles exceeding 30°. In a controlled laboratory setting PTS increase unloads the ACL, affecting its natural function.

KEYWORDS

ACL, HTO, kinematics, osteotomy, slope

1 | INTRODUCTION

Frontal and sagittal lower limb alignment has gained attention in recent years, as it has been shown to affect pressure distribution across the articular cartilage, anterior cruciate ligament (ACL) graft forces, failure rates, and clinical outcomes.[1–5] The sagittal alignment is primarily determined by the morphology of the proximal tibia and particularly by the posterior tibial slope (PTS).[6] Clinically, a high (steep) PTS has been found to be an independent risk factor for primary and recurrent ACL injury and was correlated with anterior tibial translation.[7–10]

The investigation was performed at the Orthopaedic Robotics Laboratory, Departments of Orthopaedic Surgery and Bioengineering, University of Pittsburgh, Pittsburgh, Pennsylvania, USA.

The frontal alignment is determined by the morphology of both the proximal tibia and the distal femur.[6] Increasing varus alignment has been shown to increase peak contact pressure in the medial tibiofemoral compartment, result in a higher degree of medial meniscal extrusion, and increase ACL forces by up to 74% compared with neutral alignment.[3,11] As a result, high tibial osteotomy (HTO) in the medial opening wedge (MOW-HTO) technique has been recommended in varus malalignment to protect the articular cartilage in the medial compartment and reduce the risk of ACL graft failure in case of ACL reconstruction.[2,11,12] However, MOW-HTO has also been associated with an unintentional increase in PTS, which subsequently leads to ACL fiber degeneration.[13–15] Moreover, PTS-increasing osteotomies are performed in patients with posterior cruciate ligament insufficiency without knowing the effects of PTS change on the native ACL.[16,17] Yet, HTOs with both intentional and unintentional PTS changes, are becoming increasingly popular to counteract negative effects of frontal and sagittal plane malalignments.[18–22] However, these osteotomy procedures alter the bony morphology, moving the position of tibial ACL insertion site relative to the femoral ACL insertion site, and causing a change of the in situ forces in the native ACL. Altered, nonphysiologic in situ forces in the ACL may result in ACL fiber degeneration and subsequent injury.

The purpose of this study was to quantify the effect of increasing PTS on knee kinematics and the in situ forces in the native ACL in human cadaveric knees using a 6 degrees of freedom (6 DOF) robotic testing system. It was hypothesized that increasing PTS would increase anterior and proximal tibial translation, while maintaining medial-lateral tibial translation and internal-external, and varus-valgus tibial rotation. It was further hypothesized that increasing PTS would decrease the in situ force in the native ACL.

2 | METHODS

2.1 | Specimen preparation

Seven fresh-frozen human cadaveric knees (mean age 51.9 ± 19.8 years; range, 21–75 years; 71% male) were used for this controlled laboratory study, which was approved by the institutional review board of the University of Pittsburgh (CORID #331).

Specimens were thawed at room temperature for 24 h before testing. Before testing, each specimen underwent manual, fluoroscopic, and arthroscopic examination. Specimens were excluded if ligamentous injuries, meniscal injuries, cartilage injuries greater than Grade 2 according to the International Cartilage Repair Society grading system,[23] or osteoarthritis greater than Grade 2 to according to the Kellgren–Lawrence grading scale[24] were detected. In addition, specimens with a medial PTS greater than 12° as determined on strict lateral radiographs were excluded. The threshold value of 12° was chosen based on previous research showing that only 3% of a population between 18 and 92 years have a medial PTS ≥ 12°.[25] The femur and tibia were cut 20 cm proximal and distal from to the joint line, respectively. All soft tissue was removed 10 cm proximally and distally from the joint line and the fibula was fixed to the tibia using a bicortical screw to maintain its anatomical position. Subsequently, an epoxy compound (Bondo; 3 M) was used to pot the femoral and tibial bone.

2.2 | PTS states and surgical procedures

Two PTS states were assessed for each specimen: (1) osteotomized knee with native PTS, (2) osteotomized knee with increased PTS. Both PTS states were assessed after the osteotomy was performed to avoid any influence of the surgical approach on the primary outcome measures (kinematics, forces). As a result, the detected changes in knee kinematics and in situ forces in the ACL could exclusively be attributed to the change in PTS.

After potting and before the experimental testing protocol started, an external fixator (Hoffmann 3 Modular External Fixation; Stryker GmbH) was attached to the proximal tibia. Under fluoroscopic guidance, one pin (ø 6 mm) each was placed in the subchondral bone parallel to the PTS of the medial and lateral tibial plateau. A third pin was placed centrally in the epoxy compound cylinder of the tibia. Each subchondral pin was connected to the pin in the epoxy compound cylinder of the tibia by a connecting rod (ø 11 mm) via pin-to-rod couplings. As a result, the external fixator formed a triangle for rigid force transmission along the long axis of the tibia without affecting knee kinematics and in situ forces in the ACL (Figure 1).[2,5,26,27]

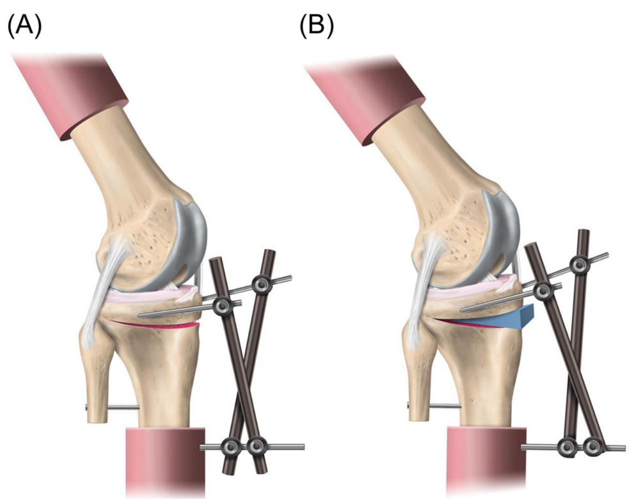

FIGURE 1 Schematic illustration of the osteotomy. Osteotomized knee with native posterior tibial slope (PTS) state (A) and osteotomized knee with increased PTS state (B). A supra-tuberositary anterior opening wedge osteotomy with a posterior cortical hinge was performed. To increase the PTS an epoxy compound wedge (blue) was placed in the osteotomy gap. The osteotomy was secured with an external fixator. The quadriceps tendon, patella, and patellar tendon were removed in this figure to show the level and orientation of the osteotomy.

Next, a 3–5 cm longitudinal incision 3 cm medial and lateral to the tibial tuberosity was made. The patellar tendon and its insertion zone at the tibial tuberosity were exposed. The cortical bone of the tibia proximal to the tibial tuberosity and distal to Gerdy's tubercle was visualized. The tibial insertion of the medial collateral ligament (MCL) had to be slightly released (less than one-third of the tibial insertion site) for the subsequent osteotomy. Two K-wires (one medial, one lateral to the patellar tendon) were drilled from anterior-inferior to posterior-superior to guide the subsequent osteotomy. K-wire placement was performed under fluoroscopic control. Next, the osteotomy was performed using an oscillating saw starting just proximal to the tibial tuberosity and distal to Gerdy's tubercle, thus avoiding violation of the joint capsule and the anterolateral structures of the knee. The osteotomy cut the anterior, medial and lateral cortex of the tibia and ended approximately 5 mm anterior to the posterior cortex in the area of the tibial insertion site of the posterior cruciate ligament, creating a posterior cortical hinge running from medial to lateral. An intact posterior cortical hinge was required to reliably change the PTS in the sagittal plane and was confirmed by fluoroscopy.

To increase the PTS, the proximal pin-to-rod couplings of the external fixator were released and the osteotomy gap was gradually opened. A custom-made epoxy compound wedge (wedge base height, 10 mm) was placed in the osteotomy gap until the base of the wedge touched the anterior tibial cortex and the proximal pin-to-rod couplings of the external fixator were resecured. Accordingly, the osteotomy gap was opened by 10 mm (anterior cortex), resulting in a knee size-dependent change in PTS.[16,28] To decrease the PTS, the steps were performed in reverse order. After each PTS change, a layer-by-layer wound closure was performed.

2.3 | Experimental setup and protocol

Specimens were attached to a 6 DOF robotic testing system (MJT model FRS2010) using custom-made aluminum clamps. Femur and tibia were attached to the lower and upper plates of the robotic manipulator, respectively. A universal force/moment sensor (UFS; ATI Delta IP60 model SI-660-60), located on the upper manipulator of the robotic testing system, was used to provide feedback to the controller. The robotic testing system was controlled using a LabView Program (Technology Services Inc). The translational and rotational position repeatability of the robotic testing system was shown to be less than ±0.015 mm and ±0.01°, respectively. The measurement uncertainty of the UFS was found to be approximately 1% of full scale.[29]

The medial-lateral translation axis and flexion–extension rotation axis were defined by the femoral insertion sites of the medial and lateral collateral ligament. The proximal–distal translation axis and internal-external rotation axis were defined by the anatomical tibial shaft axis. The anterior-posterior translation axis and varus-valgus rotation axis were defined as the cross-product of the two mentioned axes.[30] Given that a supra-tuberositary HTO was performed, the anatomical tibial shaft axis and therefore the joint coordinate system was not affected by the HTO.

The path of passive flexion-extension from full extension (defined as 1 Nm extension moment) to 90° of knee flexion of the osteotomized intact knee with the native PTS state (i.e., external fixator attached, surgical approach, MCL release, and osteotomy performed) was determined. Minimized forces and moments (within 0.5 N and 0.2 Nm, respectively) across all axes of the defined coordinate system throughout the range of motion were required to determine the path of passive flexion extension.[31–34] Three loading conditions were applied to the osteotomized intact knee with the native PTS state while continuously flexing the knee from full extension to 90° of knee flexion, and the resulting 6 DOF kinematics were recorded. The three loading conditions were (1) 200 N axial compressive load, (2) 5 Nm internal tibial torque combined with 10 Nm valgus tibial torque, and (3) 5 Nm external tibial torque combined with 10 Nm varus tibial torque. The loading conditions were chosen as it has been shown that axial compressive loads and tibial rotational torques affect the in situ forces in the ACL.[35–38] In addition, the combined loading conditions have recently been confirmed as appropriate biomechanical loading conditions to evaluate in situ forces in the ACL.[34,39] To control for viscoelasticity of the soft tissues, each loading condition was repeated five times, and the data from the fifth cycle were used for final analysis.

After acquisition of the 6 DOF kinematics for each loading condition of the osteotomized intact knee with the native PTS state, the specimen was removed from the robotic testing system leaving the connecting clamps attached and thus maintaining the previously defined coordinate system. The PTS was increased and the osteotomized intact knee with the increased PTS state was mounted to the robotic testing system. The testing protocol was repeated as for the native PTS state.

After the 6 DOF kinematics of the intact knee were recorded for both PTS states (osteotomized knee with native PTS, osteotomized knee with increased PTS) and all three loading conditions, the ACL was arthroscopically removed. In the ACL deficient knee, the previously acquired 6 DOF kinematics for each loading condition of the osteotomized intact knee with the increased PTS state were repeated using the position-control mode. The new forces and moments were measured by the UFS and the in situ forces in the ACL in the osteotomized knee with the increased PTS state were calculated by applying the principle of superposition.[40,41] Next, the native PTS state was restored and the previously acquired 6 DOF kinematics for each loading condition of the osteotomized knee with the native PTS state were repeated using the position-control mode. Again, the new forces and moments were measured by the UFS and the in situ forces in the ACL in the osteotomized knee with the native PTS state were calculated by applying the principle of superposition (Table 1).[40,41] Given the ability of the robotic testing system to accurately replicate knee kinematics after ACL removal (i.e., position-control mode), it is possible to determine the magnitude and direction of the in situ force of the removed ACL. This can be done by

TABLE 1 Experimental protocol and data acquired

Knee state	Loading condition/replay kinematics[a]	Data acquired
Intact, osteotomized native PTS	(1) 200 N Axial compression (2) 5 Nm ITT + 10 Nm ValTT (3) 5 Nm ETT + 10 Nm VarTT	(a) Kinematics of the intact knee with osteotomized native PTS
Intact, increased PTS	(1) 200 N Axial compression (2) 5 Nm ITT + 10 Nm ValTT (3) 5 Nm ETT + 10 Nm VarTT	(b) Kinematics of the intact knee with increased PTS
ACL deficient, increased PTS	Replay recorded kinematics of (b)	In situ force: ACL with increased PTS
ACL deficient, osteotomized native PTS	Replay recorded kinematics of (a)	In situ force: ACL with osteotomized native PTS

Abbreviations: ACL, anterior cruciate ligament; ETT, external tibial torque; ITT, internal tibial torque; PTS, posterior tibial slope; ValTT, valgus tibial torque; VarTT, varus tibial torque.

[a]Replay (a) and (b) refer to the reproduction of previously recorded kinematics of (a) intact knee with osteotomized native PTS state and (b) intact knee with increased PTS state.

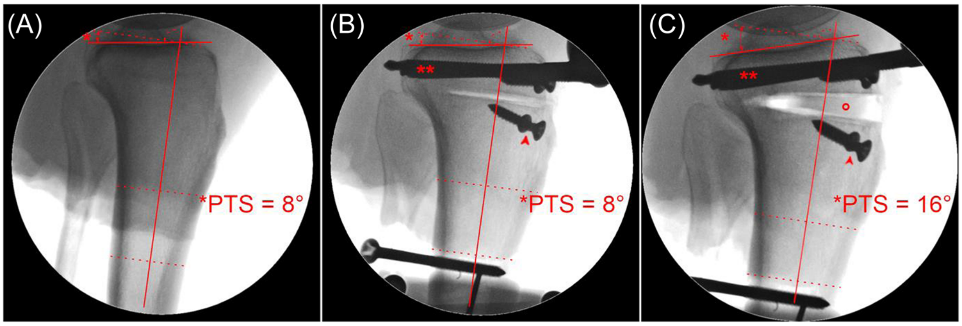

FIGURE 2 Posterior tibial slope (PTS) states and measurement. Fluoroscopy of the native (A), osteotomized native (B), and increased (C) PTS states without external load application. Rods proximal (**) and distal (not depicted) to the osteotomy gap were used to secure the osteotomy with an external fixator. The distance between screws (arrowhead) proximal and distal to the osteotomy gap was measured for repeatable PTS adjustment. °, a wedge was placed within the osteotomy gap.

quantifying the change in force vector before and after ACL removal.[40,41]

The specimens were kept moist throughout the entire testing protocol using physiological saline solution.[42]

2.4 | PTS measurements and repeatability of PTS adjustment

PTS measurements were performed on strict lateral radiographs using a previously described technique (Figure 2).[18,43] All PTS measurements reported were performed by observer one (Philipp W. Winkler). Intraclass correlation coefficients (ICC) were calculated to determine intra- and interrater reliability of measurements performed. The medial PTS was measured three times on 17 human cadaveric knees by observer one (Philipp W. Winkler), with 2 weeks between measurements. In addition, the medial PTS was measured on the same 17 knees by observer two (Nyaluma N. Wagala) and three (Jonathan D. Hughes). The images used for reliability testing were blinded by an independent coworker (Calvin K. Chan; not involved in measurements) and randomly arranged for each measurement cycle. Excellent intrarater reliability (ICC, 0.994; 95% confidence interval [CI]: 0.984–0.998) and good to excellent interrater reliability (ICC, 0.916; 95% CI: 0.755–0.977) was found for medial PTS measurements.[44] ImageJ version 1.52a (Wayne Rasband, National Institutes of Health) was used for PTS measurements.

To apply the principle of superposition to calculate in situ forces in the ACL in both PTS states within the same specimen, the two PTS states need to be precisely and reliably restored. Preliminary testing displayed good to excellent repeatability of multiple PTS adjustments to an accuracy of ±0.2° and measurements of the in situ force in the ACL within the repeatability of the robotic testing system, enabling the principle of superposition to be applied reliably.

To ensure reliable PTS adjustments during testing, a three-step protocol was followed for each PTS change: (1) The two PTS states were marked on the pins and connecting rods of the external fixator to ensure that the pin-to-rod couplings were always at the exact same level for the corresponding PTS state. (2) The PTS was measured after each PTS change and the experimental protocol was continued only if the PTS was within ±0.2° of the corresponding

PTS state. (3) Reference screws were placed medially and laterally to the patellar tendon, both proximally and distally of the osteotomy (Figure 2). The distances between the proximal and distal reference screws (medial and lateral) were measured with a caliper at each PTS change, allowing a repeatability of distance restoration of ±0.1 mm.

2.5 | Statistical analysis

A priori power analysis was based on preliminary testing and performed using G*Power (Erdfelder, Faul, Buchner, Lang, HHU Düsseldorf, Düsseldorf, Germany). To achieve a statistical power of 0.8 a minimum of four specimens was required to detect statistically significant differences in in situ forces in the ACL between the osteotomized native and increased PTS states (effect size, 3.17; α, 0.05).

Continuous variables were first assessed by the Shapiro–Wilk test. Accordingly, normally distributed data were expressed as mean, standard deviation, and range. Changes in knee kinematics were expressed as the difference in kinematics between the passive path of motion and the path of motion after load application at each flexion angle in the respective PTS state. A two-factor repeated measures analysis of variance (ANOVA) followed by Tukey post hoc tests was conducted to assess the effects of the factors "PTS" (osteotomized native vs. osteotomized increased) and "knee flexion angle" (full extension vs. 30° vs. 60° vs. 90°) and a potential PTS × knee flexion angle interaction on knee kinematics and the in situ forces in the ACL. Statistical analysis was performed using SPSS software version 26.0 (IBM-SPSS) and the level of significance was set at $p < 0.05$.

3 | RESULTS

The experimental protocol was successfully completed in all seven fresh-frozen human cadaveric knees without complications such as posterior cortical hinge fracture, tibial plateau fracture, or osteotomy fixation failure.

The native PTS before osteotomy was 9.1 ± 2.3° (range, 8.0–11.1°) and was not different from the osteotomized knee in the native PTS state (9.2 ± 2.2° [range, 8.0–11.2°]; $p > 0.05$). Anterior open wedge osteotomy increased the PTS significantly from 9.2 ± 2.2° (osteotomized native PTS state) to 17.8 ± 1.7° (range, 15.8–20.0°; osteotomized increased PTS state; $p < 0.001$).

3.1 | Knee kinematics

Anterior tibial translation was significantly higher in the increased PTS state compared with the native PTS state at 30° (4.5 ± 1.4 mm vs. 2.6 ± 2.2 mm, $p < 0.05$), 60° (2.6 ± 1.1 mm vs. 0.9 ± 1.1 mm, $p < 0.05$), and 90° (1.1 ± 0.8 mm vs. 0.1 ± 0.8 mm, $p < 0.05$) flexion in response to 200 N axial compressive load. At the same loading condition at 30° flexion, proximal tibial translation increased by 0.5 mm after increasing the PTS ($p < 0.05$). In response to 5 Nm internal tibial torque combined with 10 Nm valgus tibial torque, there was significantly more valgus tibial rotation at full extension (4.3 ± 1.7° vs. 3.6 ± 1.1°, $p < 0.05$), 30° (7.8 ± 2.3° vs. 5.8 ± 1.3°, $p < 0.05$), 60° (9.5 ± 2.2° vs. 6.6 ± 1.4°; $p < 0.05$), and 90° (9.5 ± 1.9° vs. 7.5 ± 1.3°; $p < 0.05$) flexion in the increased compared with the native PTS state. A detailed breakdown of the kinematic data can be found in Online Supporting Information: S1.

3.2 | In situ forces in the native ACL

Increasing PTS caused a significant reduction of the in situ force in the ACL by 57.6%, 69.8%, and 75.0% at 30°, 60°, and 90° flexion, respectively, in response to 200 N axial compressive load (all $p < 0.05$). In response to 5 Nm internal tibial torque combined with 10 Nm valgus tibial torque the in situ force in the ACL was 39.0% less in the increased compared with the native PTS state at 90° flexion ($p < 0.05$). Similarly, increasing the PTS resulted in a significant reduction of the in situ force in the ACL by 62.8% and 67.0% at 60° and 90° flexion, respectively, in response to 5 Nm external tibial torque combined with 10 Nm varus tibial torque (both $p < 0.05$; Figure 3).

4 | DISCUSSION

The most important finding of this study was that an isolated increase in PTS in a native knee joint affected translational and rotatory knee kinematics and resulted in a significant reduction of the in situ forces in the native ACL, primarily noted at flexion angles exceeding 30°.

In line with the hypothesis of this study, an isolated increase in PTS was shown to cause anterior and proximal tibial translation, which is consistent with previous research.[1,28,45,46] One study evaluated changes in knee kinematics in 10 human cadaveric knees after increasing the PTS by 4.4°. In response to a 200 N axial compressive load, a significant increase in anterior tibial translation by 2 mm was observed at 30° and 90° flexion.[45] Increased anterior and also proximal tibial translation after gradually increasing the PTS by up to 20° is supported by another study investigating seven human cadaveric knees in response to an isokinetic flexion-extension motion.[1]

Contrary to the hypothesis and depending on the loading condition, an increase in varus or valgus tibial rotation was found after increasing the PTS. A supra-tuberositary anterior opening wedge osteotomy, also referred to as a flexion osteotomy, was performed to increase the PTS. Accordingly, the distal insertion sites of the collateral ligaments were moved to a slightly flexed position in the increased PTS state. As the collateral ligaments slacken in a flexed

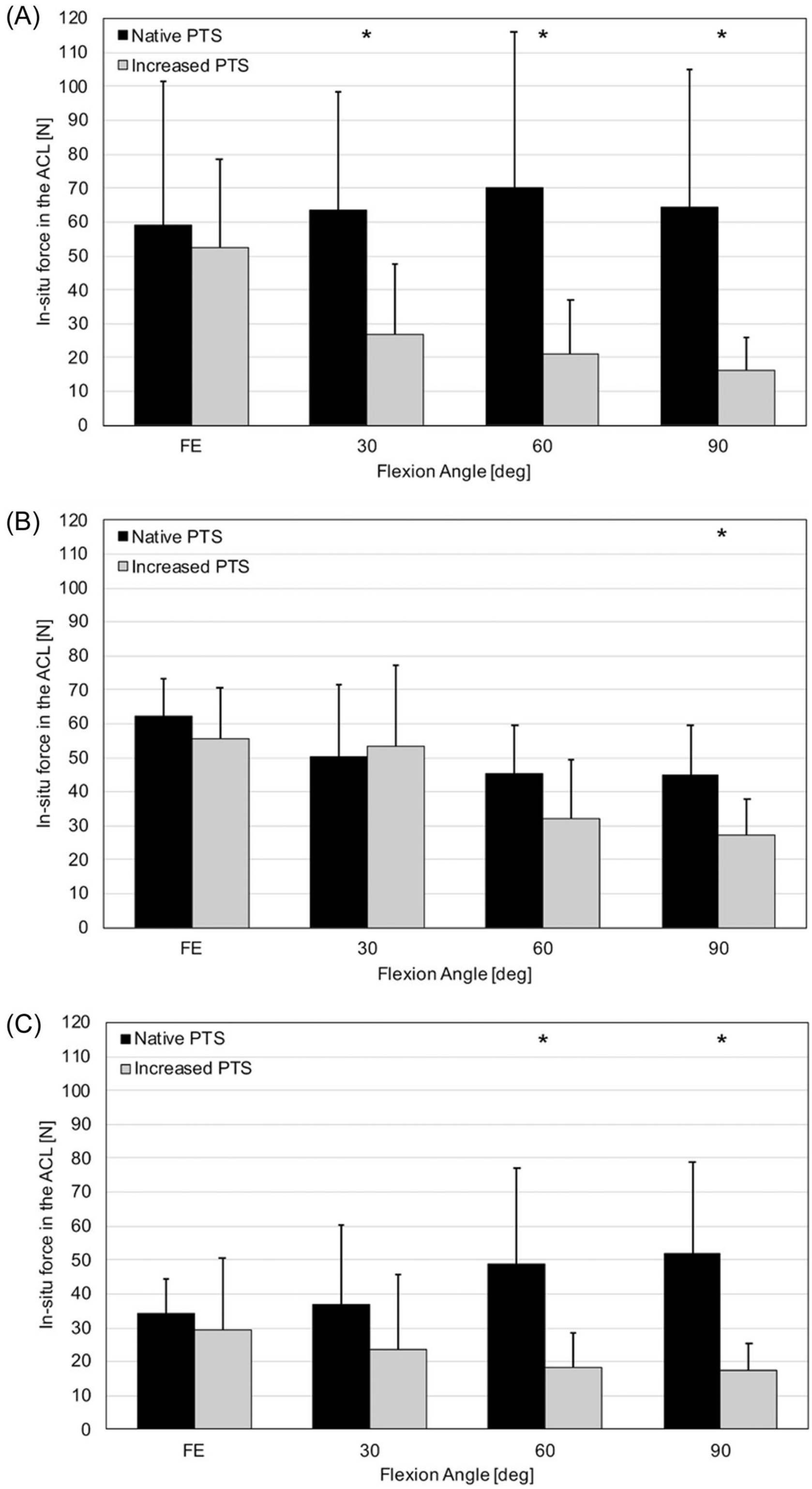

FIGURE 3 In situ forces in the native ACL. In situ force in the ACL versus flexion angle in the native and increased posterior tibial slope (PTS) state in response to 200 N axial compressive load (A), 5 Nm internal tibial torque + 10 Nm valgus tibial torque (B), and 5 Nm external tibial torque + 10 Nm varus tibial torque (C). *Statistically significant difference between the native and increased PTS state ($p < 0.05$). ACL, anterior cruciate ligament.

knee position,[47] this might be the reason for the observed increase in varus/valgus tibial rotation after an increase in PTS.

In agreement with the hypothesis, a reduction of the in situ forces in the ACL in several loading conditions and flexion angles was found after increasing the PTS. This finding may be attributed to a three-dimensional change in the position of the tibial ACL insertion site, which causes the tibial and femoral ACL insertion sites to converge. Consequently, the tensile strain within the ACL decreases, which is reflected by lower in situ forces. This finding is supported by previous studies.[28,46] One study investigated the native ACL in nine human cadaveric knees in response to a 200 N axial compressive load and found a significant reduction in native ACL strain after gradually increasing the PTS from 5° to 15°.[46] This was confirmed by another study that showed a 120% decrease in the native ACL strain after increasing the PTS by an average of 9.6° in four intact human cadaveric knees.[28] However, it should be questioned whether a reduction in native ACL strain of more than 100% is physiologically feasible considering the inability of ligamentous structures to transmit compressive loads. Of note, the reduced in situ forces in this study were primarily observed at flexion angles exceeding 30°, where typical cutting or pivoting maneuvers would not be expected. This may be related to the smaller radii of the femoral condyles at higher flexion angles, resulting in an even more pronounced approximation of the ACL insertion sites. Taken together, a modification of the PTS affects ACL function. In case of PTS increase, increasing proximal and anterior tibial translation result in a reduction of in situ forces in the native ACL. It is important to distinguish the results of this study from previous investigations that evaluated the effect of PTS on ACL graft forces.[5]

The clinical relevance of lower in situ forces in the ACL that reflect stress deprivation is based on the principle of stress shielding. Animal studies have shown that stress deprivation adversely affects the mechanical properties of soft tissues.[48–50] It has been histologically confirmed that stress shielding of tendon tissue causes irregular crimp patterns and increasing cross-sectional area, cellularity, and vascularity.[48–50] The reported histomorphological changes may reflect the first stages of fiber degeneration in tissues subject to stress deprivation. Consequently, the reduction of in situ forces in the ACL after increasing PTS may be responsible for observed ACL fiber degeneration in patients with increased PTS.[13,51–53] One propensity score-matched study indicated an association between mucoid degeneration of the native ACL and increased PTS. A 17% higher probability of mucoid ACL degeneration for each 1° increase in lateral PTS was found.[53] In addition, partial to complete ACL rupture was observed during second-look arthroscopy in 39% of patients at an average of 1.6 years after HTO, in which the PTS was unintentionally increased by almost 2°.[13] Although clinical data are consistent, the reason for mucoid ACL degeneration has not yet been determined. Based on the findings of this study, it seems that an increase in PTS causes histomorphological changes of the native ACL induced by a reduction of in situ forces.

Quantification of in situ forces in the ACL in the same knee joint in two different PTS states is a unique characteristic of this study, which was feasible due to good to excellent repeatability of multiple PTS changes based on a reliable three-step protocol for PTS adjustment. Previous studies investigating ACL forces in different PTS states have either used invasive measurement devices or different groups of specimens, each of which carries the risk of bias.[28,45,46] In the future, the proposed three-step protocol will aid in further investigation of the effects of increasing PTS on other intra-articular structures such as the posterior cruciate ligament, menisci, and cartilage.

Given that the ACL is a vascularized and vital intra-articular structure that is capable of remodeling in response to altered stress levels, the nature of this time-zero study is an important limitation. Histomorphological changes of the ACL could not be assessed and are of high interest for future studies. Another limitation is that only two PTS states (native, increased) were investigated. Accordingly, it was not possible to determine a critical PTS level beyond which the observed changes in knee kinematics and ACL forces become relevant. Furthermore, it should be noted that the applied external loads reflect forces that occur during low-intensity tasks such as clinical examination (Lachman Test, Pivot-Shift Test) or activities of daily living.[39] Noncontact ACL injuries most commonly occur during athletic tasks, associated with considerably higher loads.[54] Therefore, it might be possible that certain effects of PTS increase were masked in this study owing to low external loads applied.

5 | CONCLUSION

In this study, an isolated increase in PTS by an anterior opening wedge HTO affected both translational and rotatory knee kinematics and resulted in a reduction of in situ forces in the native ACL by up to 75% in response to isolated compressive and combined rotatory loads. Increasing the PTS in a controlled laboratory setting unloads the ACL, affecting its natural function.

AUTHOR CONTRIBUTIONS

All listed authors have contributed substantially to this work: Philipp W. Winkler, Nyaluma N. Wagala, Gian Andrea Lucidi, Calvin K. Chan, Sene K. Polamalu, and Jonathan D. Hughes collected data, performed statistical analysis, literature review, and primary manuscript preparation. Volker Musahl and Richard E. Debski assisted with interpretation of the results, initial drafting of the manuscript, as well as editing and final manuscript preparation. All authors read and approved the final manuscript.

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CONFLICT OF INTEREST

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ETHICS STATEMENT

This study was approved by the Institutional Review Board of the University of Pittsburgh (CORID #331).

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SUPPORTING INFORMATION

Additional supporting information can be found online in the Supporting Information section at the end of this article.

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