BACHELOR THESIS

THE IMPACT OF POSTERIOR TIBIAL SLOPE ON KNEE BIOMECHANICS AS PREDICTOR OF FLEXION CONTRACTURE IN TOTAL KNEE ARTHROPLASTY

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Abstract

Pre-operative knee flexion contracture in patients undergoing total knee arthroplasty (TKA) is a risk factor for post-operative flexion contracture associated with undesirable surgical outcomes such as restricted range of motion and knee instability [1]. Osteophyte formation [2] and variations of the posterior tibial slope (PTS) [3] in osteoarthritic knees may contribute to flexion contracture. However, the effects of these morphological changes are not well understood. This study aims to investigate the effect of PTS on the tibiofemoral (TF) joint kinematics and ligament and muscle forces using a musculoskeletal modeling approach.

A patient-specific musculoskeletal knee model was developed using a previously established framework [4] to represent the pre-operative knee comprising the osteophytic femoral and tibial bones. The PTS was the only varying parameter in the model, while all other variables, such as the muscle and ligament attachment sites and their slack lengths remained unchanged throughout the entire study. The PTS in the pre-operative knee was determined based on anatomical landmarks located at the proximal tibia. The PTS was varied from -9° up to 6° with increments of 3° compared to the baseline by rotating the tibial plateau in the sagittal plane of the patient-specific anatomical frame. We captured the effect of PTS on the TF joint kinematics, quadriceps muscle activity and ligament and contact forces during an unloaded knee extension simulation from 60° to 0°. Differences in the simulated outcomes were quantified using the root-mean-square deviation.

A greater PTS (by +6°) resulted in increased forces of the anterior cruciate ligament (ACL), posterior capsule (PC), deep medial collateral ligament (dMCL), and superficial medial collateral ligament (sMCL) in extension of the knee by +44.3%, +18.6%, +49.8%, and +119.3% respectively. The TF compressive and shear forces and the muscle activity of the quadriceps increased in extension as well and the tibia was translated more anteriorly with respect to the femur for larger slope angles. The results show a possible contribution to flexion contracture. Surgeons should carefully consider the angle of the tibial cut in TKA to avoid residual flexion contracture.

Contents

1 Introduction

Total knee arthroplasty (TKA) is a well-established method for alleviating pain and improving the function of the knee joint in patients with end-stage osteoarthritis (OA). However, post-operative knee flexion contracture is a risk factor for patient dissatisfaction following TKA [1, 5]. Knee flexion contracture is defined as the inability to fully extend the knee joint to 0°. Post-operative flexion contracture is associated with increased pain, restricted range of motion and overall lower function scores [5, 6]. A pre-operative flexion contracture deformity is one of the risk factors for residual flexion contracture after TKA [5]. The cause of flexion contracture is multifactorial; ligament contracture, hamstring shortening, contracture of the posterior capsule, bony impingement [6] and a large posterior tibial slope (PTS) [3] are contributing factors to this complication in OA knees.

OA is characterized by the formation of osteophytes; a pathological condition that changes the bone morphology. Depending on their size and location osteophytes may cause increased tension on the ligaments or the posterior capsule as a result of wrapping around the bony outgrowth [2], narrowing of the joint space [7], and mechanical obstruction [8]. These effects could contribute to a loss of range of motion and an extension deficit. Moreover, osteophyte formation in the posterior part of the femur might increase the pressure on the knee joint posteriorly during weight-bearing conditions, causing a gradual change of the articular surface increasing the PTS over time [3].

In advanced OA knees, the mechanical alignment of the lower limb in the coronal plane tends to shift, exaggerating the original alignment of the knee (i.e. increased varus or valgus deformity) [9]. The inclination of the medial compartment of the proximal tibia is aligned parallel to the ground in the coronal plane under weight-bearing conditions. This parallel mechanism is possibly a result of inefficient absorption of load-bearing by the degenerative cartilage and malfunction of ligaments during large adduction moment in OA knees [10]. Similarly, it is thought that in the sagittal plane the articular surface of the proximal tibia is aligned parallel to the ground during weight-bearing [3]. To achieve sagittal alignment, an increased posterior tibial slope (PTS) would require more knee flexion in a standing position and would therefore contribute to a pre-operative flexion contracture. Furthermore, a recent study showed that patients with pre-operative flexion contracture showed a larger PTS compared to patients with no flexion contracture [3].

Knee kinematics are significantly influenced by the tibial slope. A greater PTS is associated with increased shear force and anterior tibial translation [11, 12, 13]. This resulted in a more posterior position of the tibiofemoral contact point, increased tension on the anterior cruciate ligament (ACL) and affected anteroposterior stability [3, 12, 14]. Furthermore, increasing the tibial slope during TKA resulted in a tight extension gap and loss of extension post-operatively [15, 16]. Patients with pre-operative flexion contracture might benefit from correction of a large PTS during TKA to regain full extension, since every degree of increased PTS results in residual flexion contracture [6]. During pre-operative planning of TKA surgery, the patient's PTS and the angle of the tibial cut should therefore be considered carefully.

A musculoskeletal model could be a useful tool towards defining a pre-operative plan that is personalized to the patient-specific anatomy. Personalized musculoskeletal knee modeling may assist surgeons to optimize their surgical decision-making approach in the correction flexion contracture [17]. The aim of this study was to investigate the effect of PTS on the tibiofemoral joint kinematics, ligament and joint contact forces and muscle activity using a musculoskeletal modeling approach. It was hypothesized that increasing the PTS would result in increased tension in the ACL and increased anterior tibial translation.

2 Methods

A previously developed musculoskeletal model of a severe OA knee was used in this study [4]. This model includes the femur, tibia, and patella bones. In brief, the model consists of the tibiofemoral (TF) and patellofemoral (PF) joints, comprising in total 7 degrees of freedom (DOF). The TF joint was modeled to have 6 DOF and the PF joint was considered as an ideal revolute joint with 1 DOF. Ligaments were defined as one-dimensional spring elements. The quadriceps muscles were included in the model, driving the extension of the knee. The remaining 5 DOF in the TF joint are solved quasi-statically using the force-dependent kinematics (FDK) approach [18]. The FDK allows for concurrent estimation of ligament, muscle and joint contact forces.

2.1 Defining the patient-specific tibial frame

Based on the patient-specific bones, the anatomical frame of the tibia was defined (see Figure 1. The origin of the anatomical frame was defined as the midpoint between the outermost points of the medial and lateral tibial condyle. The y-axis (positive superiorly) was defined as the line connecting the origin of the frame and the midpoint of the medial and lateral malleolus. The tibial z-axis (positive laterally) was defined as the line connecting the outermost points of the medial and lateral tibial condyle. Lastly, the x-axis (positive anteriorly) was defined as the cross-product of the y-axis and the z-axis.

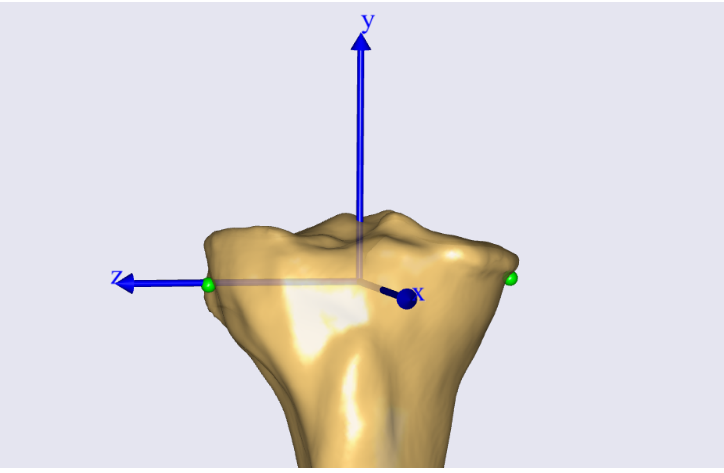

Figure 1: *Patient-specific anatomical frame of the tibia based on anatomical landmarks; the x-axis is pointing anteriorly, the y-axis superiorly, and the z-axis laterally. The green landmarks indicate the outermost points of the tibial condyles.*

2.2 Calculation of the posterior tibial slope

The medial and lateral PTS in the pre-operative knee was determined based on anatomical landmarks located at the joint surface of the proximal tibia. For both the medial and lateral PTS, these landmarks consisted of the anterior-most and the posterior-most points of the proximal tibial joint surface in the respective compartment, as seen in Figure 2. The PTS was defined as the angle between the line connecting the posterior and anterior landmarks and the xz-plane of the patient-specific anatomical frame of the tibia (see Figure 3).

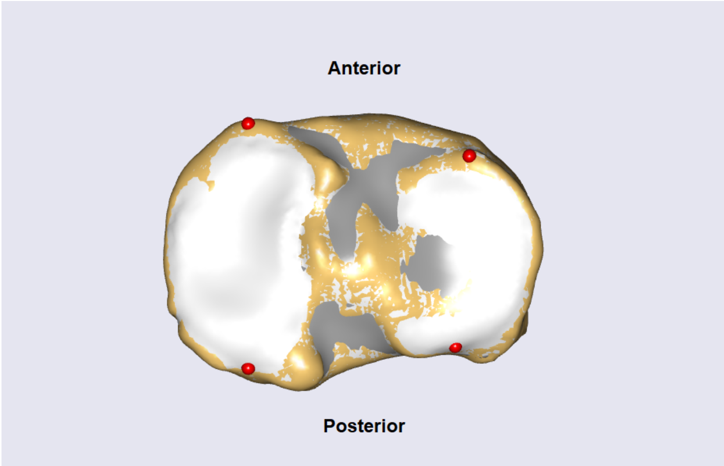

Figure 2: *Bony landmarks for determining the posterior tibial slope (PTS) from a superior view of the tibial joint surface. The landmarks marked in red comprise the anterior-most and poster-most points of the medial and lateral tibial condyles.*

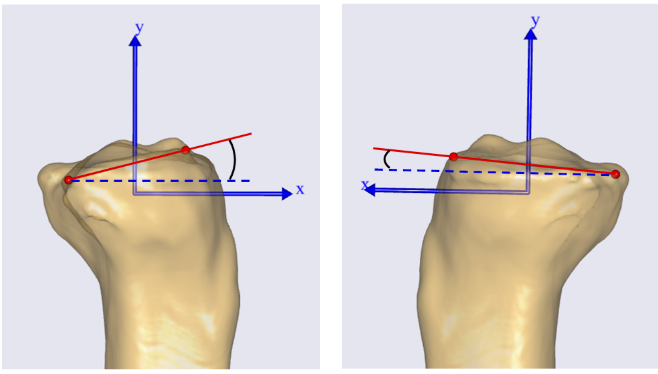

Figure 3: *Calculation of the angle of the lateral (left) and medial (right) posterior tibial slope (PTS). The slope is defined as the angle between the line connecting the bony landmarks from posterior to anterior (indicated in red) and the xz-plane of the tibial frame (indicated with the blue dashed line)*

2.3 Varying the posterior tibial slope

In this model, the PTS was changed by rotating the tibial plateau in the sagittal plane. The tibial plateau included the bony joint surface as well as the articular cartilage. With increments of 3°, the PTS was varied from -9° to +6° compared to the baseline (Figure 4) by rotation around the z-axis of the previously defined patient-specific tibial frame. The origin of this frame was the center of rotation. The medial and lateral PTS were found to be 5.12° and 14.39° respectively. Normal angles of PTS can range from -4° to 17° [19]. To keep the medial and lateral PTS within reasonable bounds, the PTS was decreased by 9° and increased by 6°, resulting in a medial PTS ranging from -3.88° to 11.12° and a lateral PTS ranging from 5.40° to 20.37°. Muscle and ligament attachment sites and their slack lengths were adjusted to the reference case of 0° change in PTS and remained unchanged throughout the entire study, leaving the slope as the only varying parameter in the model.

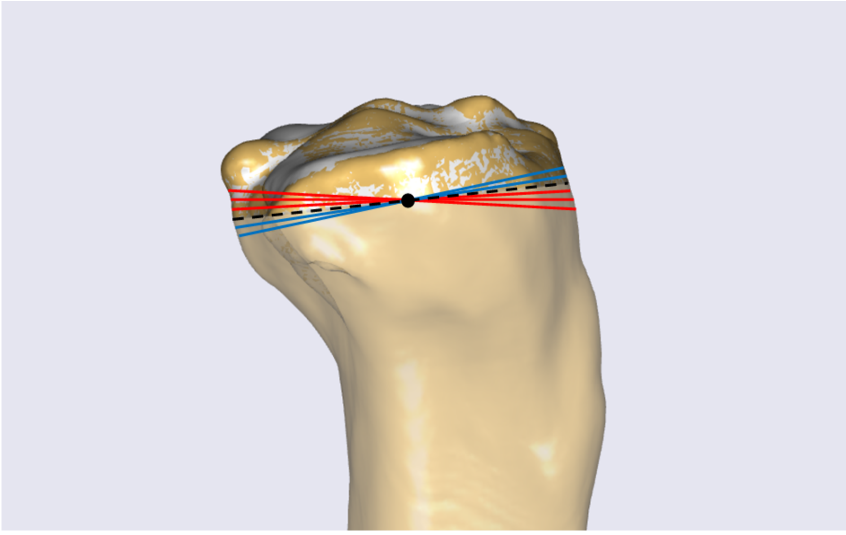

Figure 4: *Variations of the posterior tibial slope (PTS) by rotation of the tibial plateau. The red lines represent the decreasing PTS angles and the blue lines the increasing angles. The dashed line is the reference with a 0° change in PTS and the circle is the center of rotation.*

2.4 Simulating knee extension

The effect of PTS on the TF joint kinematics, ligament and joint contact forces and muscle activity was captured during simulated extension of the knee. The motion of the knee simulated an unloaded leg swing, extending the knee from 60° to 0°. The primary outcome of simulations included forces of the cruciate ligaments, collateral ligaments and posterior capsule, maximum muscle activity of the quadriceps, anterior-posterior (AP) translation and compressive and shear force of the TF joint. The differences in the simulated outcomes for each variation in the angle of the tibial plateau compared to the baseline were quantified using the root-mean-square deviation (RMSD) for the full range of motion.

3 Results

In the reference case, the medial and lateral PTS were found to be 5.12° and 14.39° respectively. Changing the slope by -9°, -6°, -3°, +3° and +6° resulted therefore respectively in PTS variations of -3.88°, -0.88°, 2.12°, 8.12° and 11.12° in the medial tibial compartment and 5.40°, 8.40°, 11.39°, 17.38° and 20.37° in the lateral tibial compartment.

3.1 Ligament forces

The ligament forces of the anterior cruciate ligament (ACL), posterior cruciate ligament (PCL), oblique popliteal ligament (OPL), posterior capsule (PC), deep medial collateral ligament (dMCL), superficial medial collateral ligament (sMCL), lateral collateral ligament (LCL) and anterolateral ligament (ALL) for each of the simulated PTS variations are depicted in Figure 5. When increasing the PTS, the ligament force of the ACL, PC, dMCL, and sMCL increased in extension of the knee. The OPL force decreased slightly for an increased PTS at a 0° knee flexion angle. The LCL and ALL remained slack during extension. At 60° flexion of the knee, a larger slope resulted in decreased forces of the dMCL, sMCL and LCL and an increased ALL force. The PCL mostly remained slack over the full range of motion except for the increased PTS angles, where the ligament force slightly increased at 0° knee flexion.

The RMSDs, quantifying the differences of the slope variations for each ligament over the full

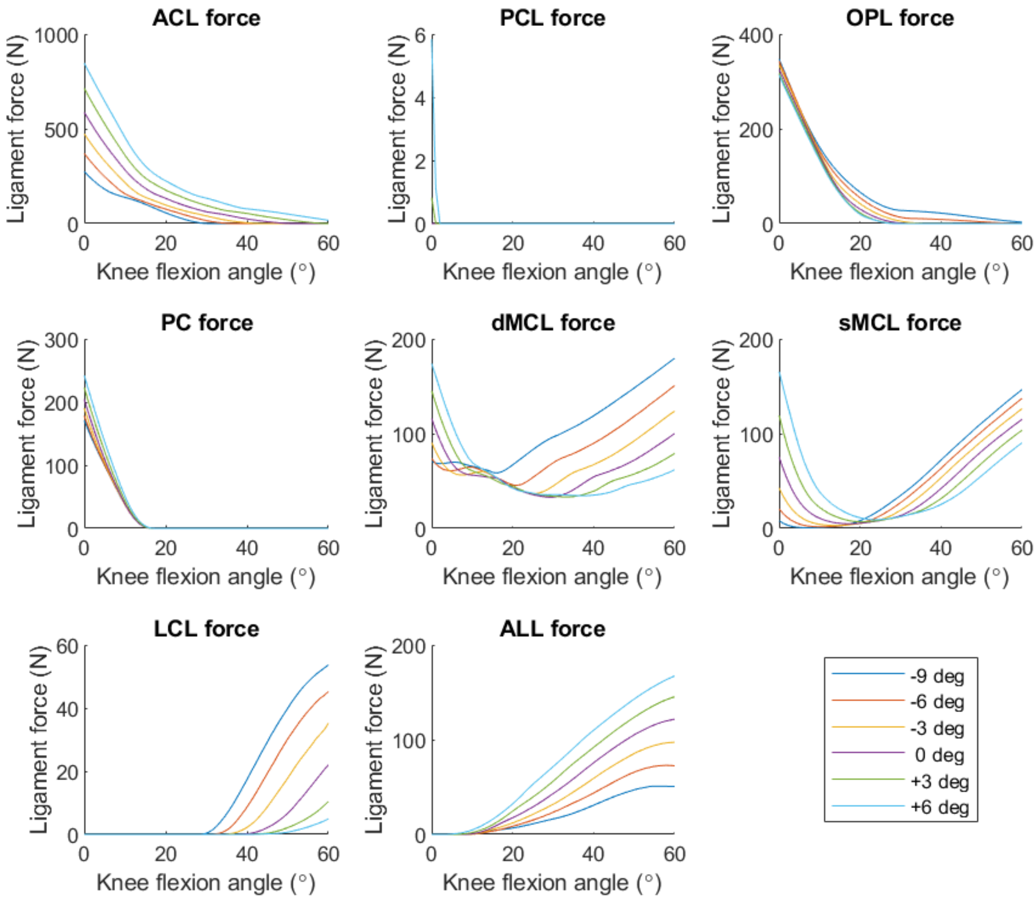

Figure 5: *Ligament forces of the anterior cruciate ligament (ACL), posterior cruciate ligament (PCL), oblique popliteal ligament (OPL), posterior capsule (PC), deep medial collateral ligament (dMCL), superficial medial collateral ligament (sMCL), lateral collateral ligament (LCL) and anterolateral ligament (ALL) during extension of the knee from 60° to 0° for changes in the posterior tibial slope (PTS) by -9°, -6°, -3°, +0°, +3°, and +6°.*

range of motion, are denoted in Table 1. For the ACL, PCL, PC, and ALL increasing the PTS resulted in overall larger ligament forces compared to the reference and increasing the slope by 6° resulted in a RMSD of 116.68 N, 0.77 N, 12.04 N, and 27.91 N respectively. Controversely, OPL and LCL ligament forces decreased when increasing the slope, resulting in a RMSD of 6.87 N and 5.30 N respectively for an increased PTS of 6° with respect to the reference. Furthermore, the RMSD values showed that increasing the slope resulted in larger differences in ligament forces of the ACL, PC, and sMCL compared to decreasing the PTS; a +6° change yielded a RMSD of 116.68 N for the ACL, 12.04 N for the PC and 30.71 N for the sMCL, while a -6° change in PTS respectively yielded 85.14 N, 6.01 N and 19.78 N. However, in the OPL, dMCL and LCL ligaments decreasing the slope resulted in larger differences, where the RMSD was 13.51 N, 32.75 N, and 12.39 N for a -6° change and 6.87 N, 24.55 N, and 5.30 N for a +6° change respectively.

In extension of the knee, the ligament forces of the ACL, PCL, PC, dMCL and sMCL were increased compared to the reference case when the slope increased. As denoted in Table 2, forces of the ACL increased by 44.3%, PC by 18.6%, dMCL by 49.8% and sMCL by 119.3% for a PTS increase of 6°, while the OPL force decreased by -3.6%. The PCL, LCL, and ALL

Table 1: *Root-mean-square deviations of the ACL, PCL, OPL, PC, dMCL, sMCL, LCL and ALL forces during extension of the knee from 60° to 0° for changes in PTS by -9°, -6°, -3°, +3° and +6° compared to the reference case.*

	-9°	-6°	-3°	+3°	+6°
ACL	113.99	85.14	46.31	54.59	116.68
PCL	0	0	0	0.11	0.77
OPL	22.82	13.51	5.89	4.43	6.87
PC	7.22	6.01	3.37	5.46	12.04
dMCL	55.13	32.75	14.63	12.96	24.55
sMCL	26.49	19.78	11.14	14.06	30.71
LCL	18.05	12.39	5.99	3.85	5.30
ALL	38.07	26.41	13.50	13.62	27.91

remained slack in extension for the reference case.

Table 2: *Differences and changes in ligament forces of the ACL, PCL, OPL, PC, dMCL, sMCL, LCL, and ALL in extension of the knee for changes in PTS by -9°, -6°, -3°, +3° and +6° compared to the reference case. *The reference case remained slack for PCL, LCL, and ALL.*

	-9°		-6°		-3°		+3°		+6°	
	ΔF	%	ΔF	%	ΔF	%	ΔF	%	ΔF	%
ACL	-312.21	-53.2	-217.29	-37.0	-113.21	-19.3	126.71	+21.6	260.10	+44.3
PCL*	0	-	0	-	0	-	0.82	-	5.84	-
OPL	21.60	+6.6	17.33	+5.3	9.49	+2.9	-8.33	-2.6	-11.76	-3.6
PC	-33.02	-16.2	-26.21	-12.8	-13.83	-6.8	18.58	+9.1	37.99	+18.6
dMCL	-45.37	-39.1	-42.21	-36.4	-25.49	-22.0	29.30	+25.3	57.78	+49.8
sMCL	-67.43	-89.4	-54.83	-7,2.7	-32.81	-43.5	43.67	+57.9	89.97	+119.3
LCL*	0	-	0	-	0	-	0	-	0	-
ALL*	0	-	0	-	0	-	0	-	0.01	-

3.2 Quadriceps muscle activity

The simulated outcomes of the maximum muscle activity of the quadriceps are depicted in Figure 6. An increase in tibial slope angle resulted in greater muscle activity. RMSD values of 0.1051, 0.0721, 0.0364, 0.0401, and 0.0813 in muscle activity were found for a slope change by -9°, -6°, -3°, +3°, and +6° respectively.

3.3 Tibiofemoral contact forces

In Figure 7 are the lateral and medial TF compressive and shear forces depicted. The magnitude of the TF compressive force increased for larger angles of PTS in both the medial and lateral compartments. The compressive force is acting downwards, hence the negative values. The compressive forces were largest in knee extension. The medial and lateral TF shear force increased in anterior direction when increasing the tibial slope.

3.4 Anterior tibial translation

The simulated outcomes of AP-translation are displayed in Figure 8. Increasing the slope of the tibial plateau resulted in a more anterior translation of the tibia with respect to the femur. RMSDs of 4.07 mm, 2.70 mm, 1.32 mm, 1.12 mm, and 1.95 mm in AP-translation were found for a slope change by -9°, -6°, -3°, +3°, and +6° respectively.

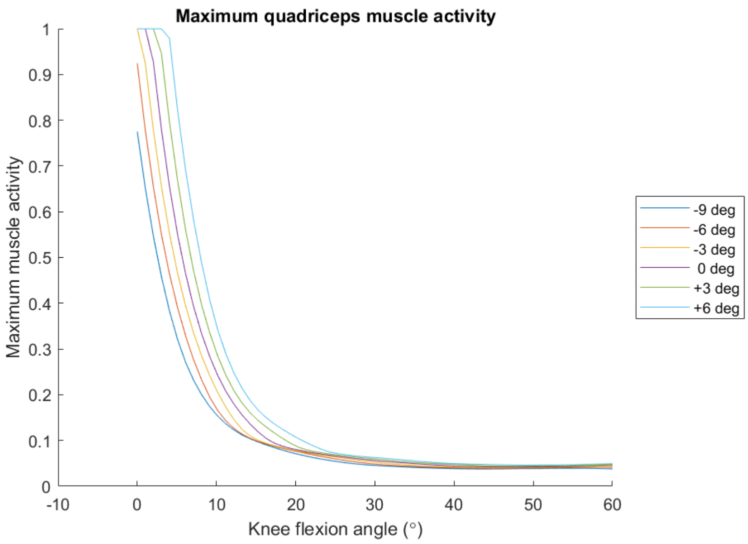

Figure 6: *Maximum muscle activity of the quadriceps during extension of the knee from 60° to 0° for varying angles of the tibial plateau.*

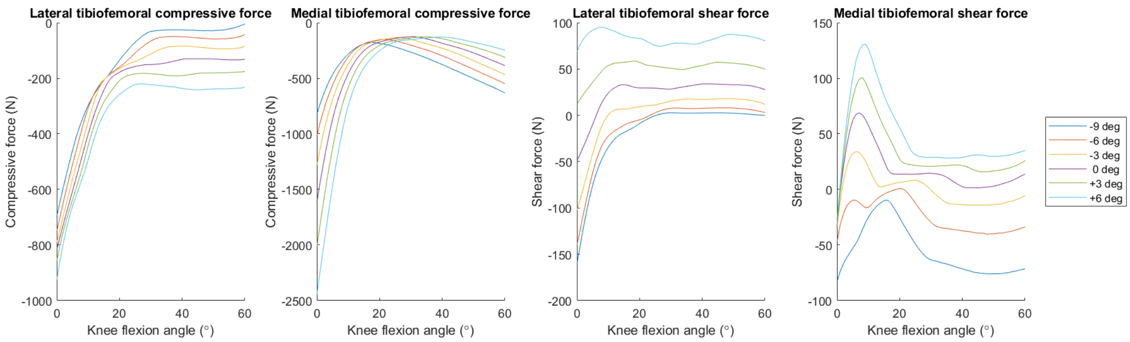

Figure 7: *Medial and lateral tibiofemoral contact forces during extension of the knee from 60° to 0° for varying angles of the tibial plateau.*

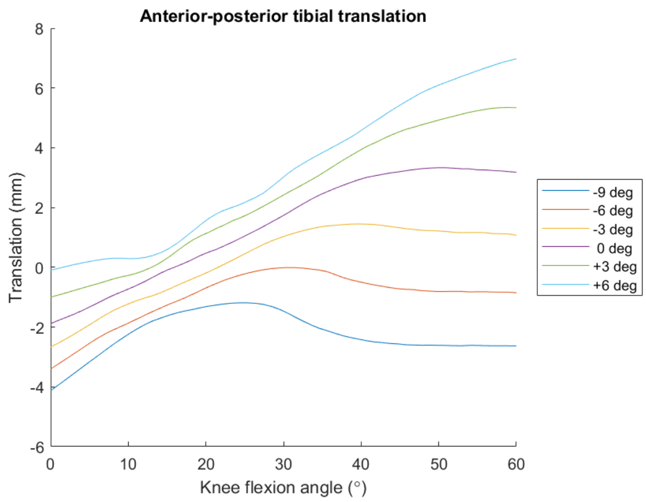

Figure 8: *Anterior-posterior tibial translation during extension of the knee from 60° to 0° for varying angles of the tibial plateau.*

4 Discussion

The purpose of this study was to investigate the effect of PTS on ligament forces, muscle activity of the quadriceps, contact forces, and TF joint kinematics. The results confirm our hypothesis that increasing the PTS would result in increased ACL forces and increased anterior tibial translation. The simulated outcomes showed that increasing PTS resulted in increased ligament forces of the ACL, dMCL, sMCL, and PC, increased activity of the quadriceps muscles, increased TF joint contact forces, and a more anterior translation of the tibia with respect to the femur.

The results showed AP-translation in anterior direction and increased shear forces in the TF joint for larger slope angles. This is in agreement with a previous biomechanical study in human cadaveric knees which found that TF contact pressure was shifted more anteriorly and the tibia was translated more anteriorly with respect to the femur [20]. Dejour and Bonnin [21] reported a 6 mm anterior tibial translation for every 10° increase in the slope. In our study, an anterior translation of 1.95 mm was found for a 6° increase in PTS. An anterior tibial translation results in increased distance between ligament attachments of the ACL and ALL, causing these ligaments to strain. Contrarily, the distance between ligament attachments of the collateral ligaments and the PCL would decrease and slack these ligaments in 60° flexion. The results in this study support this statement; for a larger PTS, forces of the ACL and ALL increased, forces of the collateral ligaments decreased in flexion and the PCL remained slack in flexion.

The largest differences in ligament forces were found in ACL, dMCL, and sMCL. The differences in the PC force were smaller than expected when varying the slope angle: a +6° change resulted in a force increase of 18.6%, while forces in the ACL, dMCL, and sMCL increased by 44.3%, 49.8%, and 119.3% in extension. A possible cause for this might be that attachment sites of the PC bundles located at the tibial joint surface did not follow the rotation of the tibial plateau for the varying slopes, since attachment sites of all ligaments remained unchanged throughout the entire study. The PCL remained slack for the full range of motion and most of the PTS variations. Only in extension the PCL force slightly increased for the +3° and +6° PTS variations. This might be due to improper assignment of the reference strain of the PCL.

In the lower knee flexion angles, increasing TF joint compressive forces and quadriceps muscle activity were found for larger angles of PTS. This is in line with cadaveric studies, where an increase in slope resulted in a significantly higher quadriceps strength necessary to extend the knee [20]. These findings indicate that increased pressure is acting on the TF joint in extension and that higher quadriceps strength might be necessary to gain full extension of the knee, suggesting a possible contribution of increasing PTS to pre-operative flexion contracture. A decrease in the angle of the tibial cut during TKA might therefore prevent residual flexion contracture post-operatively.

Pre-operative flexion contracture requires correction during TKA. Correction of flexion contracture starts with the pre-operative understanding of the problem. Mochizuki et al. [3] reported that the PTS was more exaggerated within a flexion contracture group in their study compared to a non-flexion contracture group. Although many factors may contribute to a pre-operative flexion contracture, recognition of a large PTS could be of value in optimizing the surgical approach to correcting flexion contracture. While implant selection, component positioning, alignment, and gap balancing may all complicatedly affect residual flexion contracture, the angle of the tibial cut should be considered carefully in TKA to avoid unintentional worsening of the tibial slope [3, 6]. Decreasing the tibial cut angle might aid in the improvement of flexion contracture.

To the best of our knowledge, this study is the first to use musculoskeletal modeling to investigate the effect of PTS on the biomechanics of the pre-operative OA knee and its relation to flexion contracture. The use of a musculoskeletal model was a major strength of this study. Modeling allowed for the investigation of the effect of PTS variations, while other variables, such as soft tissue properties, remained unchanged.

This study had several limitations. First of all, the mechanical properties of the ligaments, such as the reference strain and stiffness, are based on literature findings and are not specific to the patient. Secondly, in order to vary the PTS, the center of rotation was chosen to be the estimated center of the tibial plateau. A more anterior or posterior center of rotation might, however, influence ligament forces differently when varying the tibial slope angle. Moreover, the femoral and tibial bones used in this study contained large osteophytes. It is unclear how these osteophytes affected the outcomes of this study. Furthermore, ligament attachment sites remained unchanged throughout the entire study. However, for ligaments with attachments at the tibial joint surface, such as the ACL, PCL, and PC, it might be more realistic to change the attachment sites with rotation of the tibial plateau, keeping the attachment sites at their respective position on the tibial joint surface. Finally, this model simulated only a single patient. The anatomical variability among OA patients was therefore disregarded. Further research could therefore involve simulation of different phenotypes and different stages of OA to overcome this limitation.

5 Conclusion

Changes in PTS have a considerable effect on the ligament forces, TF joint contact forces, quadriceps muscle activity, and TF kinematics. Increasing of the PTS angle resulted in anterior tibial translation, increased quadriceps muscle activity and increased TF compressive and shear forces in extension. For a greater PTS, ligament forces of the ACL, PC, dMCL, and sMCL increased in extension and forces of the collateral ligaments decreased in flexion. The angle of the tibial cut in TKA should be carefully considered to avoid residual flexion contracture.

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