Bi-cruciate stabilized total knee arthroplasty can reduce the risk of knee instability associated with posterior tibial slope

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Title

Bi-cruciate stabilized total knee arthroplasty can reduce the risk of knee instability associated with posterior tibial slope

Abstract

Purpose

The purpose of this study was to evaluate the relationship between posterior tibial slope and knee kinematics in bi-cruciate stabilized (BCS) total knee arthroplasty (TKA), which has not been previously reported.

Methods

This computer simulation study evaluated Journey 2 BCS components (Smith & Nephew, Inc., Memphis, TN, USA) implanted in a female patient to simulate weight-bearing stair climbing. Knee kinematics, patellofemoral contact forces, and quadriceps forces during stair climbing (from 86° to 6° of flexion) were computed in the simulation. Six different posterior tibial slope angles (0° to 10°) were simulated to evaluate the effect of posterior tibial slope on knee kinematics and forces.

Results

At 65° of knee flexion, no anterior sliding of the tibial component occurred if the posterior tibial slope was less than 10°. Anterior contact between the anterior aspect of the tibial post and the femoral component was observed if the posterior tibial slope was 6° or more. An increase of 10°

in posterior tibial slope (relative to 0°) led to a 4.8% decrease in maximum patellofemoral contact force and a 1.2% decrease in maximum quadriceps force.

Conclusion

BCS TKA has a wide acceptable range of posterior tibial slope for avoiding knee instability if the posterior tibial slope is less than 10°. Surgeons should prioritize avoiding adverse effects over trying to achieve positive effects such as decreasing patellofemoral contact force and quadriceps force by increasing posterior tibial slope. Our study helps surgeons determine the optimal posterior tibial slope during surgery with BCS TKA; posterior tibial slope should not exceed 10° in routine clinical practice.

Keywords

total knee arthroplasty, posterior tibial slope, knee instability, computer simulation, bi-cruciate stabilized type

Introduction

Proper positioning of total knee arthroplasty (TKA) components is important for a good clinical outcome. Compared with coronal and rotational alignments, sagittal alignment, especially the acceptable range for posterior tibial slope, remains controversial [2,14]. Increased posterior tibial slope can contribute to improved deep knee flexion [22] and reduce the quadriceps force required for knee motion [20]. On the other hand, there are disadvantages such as posterior articular wear of the insert [27] and knee instability, which can result in anterior tibial translation [9,19,21,28]. Although each of these advantages and disadvantages can be appreciated, the optimal range of posterior tibial slope will also vary by implant design. Evaluating the effect of posterior tibial slope on clinical results might be difficult because of the large variation in cutting errors [1]. In addition, inter-individual differences in muscular strength and soft tissue conditions also can obscure the effect of posterior tibial slope on patellofemoral contact forces and quadriceps forces. A computer simulation model might be useful for evaluating the effect of posterior tibial slope on several factors when other conditions remain constant.

Bi-cruciate stabilized (BCS) TKA was designed to overcome the disadvantage of paradoxical motion of the femoral component with conventional posterior-stabilized (PS) TKA [6]. The design concept behind BCS TKA is promoting normal knee kinematics by incorporating both anterior and posterior post-cam mechanisms to replicate function of both the anterior cruciate ligament (ACL) and posterior cruciate ligament (PCL). In addition, BCS TKA has asymmetrical tibial articular geometry, which means a conforming medial compartment and a less conforming lateral compartment in sagittal alignment. These features can stabilize the knee in the sagittal alignment; several studies have reported that BCS TKA has in vivo kinematics that are closer to those of the normal knee than conventional PS TKA [11,24]. However, the relationship between posterior tibial slope and knee kinematics in BCS TKA has not been reported previously. In addition, computer simulation can be used to measure patellofemoral contact forces and quadriceps forces, which cannot be measured in vivo. It is useful for surgeons to appreciate postoperative knee kinematics including instability, patellofemoral contact forces, and quadriceps forces with varying posterior tibial slopes after BCS TKA. The purpose of this study was to determine the acceptable range for posterior tibial slope with BCS TKA based on computer simulation. The hypothesis is that BCS TKA has a more acceptable range of posterior tibial slopes for avoiding knee instability based on the design concept and that it causes knee instability if the posterior tibial slope is excessive, but is unlikely to cause knee instability when implanted using regular surgical technique.

Materials and Methods

Computer simulation

This study evaluated the Journey 2 BCS components (Smith & Nephew, Inc., Memphis, TN, USA) implanted in a female patient that is 162 cm in height and 58 kg in weight to simulate weight-bearing stair climbing. All the components were implanted in an appropriate size (femoral component: size 3, tibial component: size 3, insert: 9 mm, patella component: 29 mm). Initial coordinates were determined using a computer-assisted design software program (Rhinoceros; Robert McNeel and Associates, Seattle, WA, USA) as reported in our previous studies [16,17,19]. The origin of the initial coordinates was the center of the asymmetrical tibial insert, which is the intersection of the perpendicular bisector that made rectangles in both the anterior-posterior and the medial-lateral dimensions. The most distal condylar points of the femoral component were set on the surface of the tibial insert in the superior-inferior dimension.

The implant geometry was imported into a dynamic musculoskeletal modeling program (LifeMOD/KneeSIM 2010; LifeModeler, Inc., San Clemente, CA, USA; Fig. 1). This model has been reported as a useful tool for kinematic evaluation [15,19,23]. KneeSIM uses rigid body dynamics to simulate weight-bearing stair climbing. The masses of the limb segments and body weight generate a flexion moment on the knee, whereas the quadriceps muscle exerts an extension moment. This musculoskeletal model of the knee included the medial collateral and lateral collateral ligaments, quadriceps muscle and tendon, patellar tendon, and hamstring muscles. The proximal attachment points of the medial collateral ligament and lateral collateral ligament were defined as the most prominent epicondyles of the femur. Collateral ligaments were modeled as nonlinear springs with material properties obtained from a published report [3]. Contact between the tibiofemoral and patellofemoral articular surfaces was simulated. The hip and ankle joints had all three rotational degrees of freedom. The ankle section had no translational degrees of freedom. The hip section was constrained in the mediolateral and anteroposterior (AP) directions but was free to translate vertically in the direction of gravity under axial forces that generate a flexion moment at the knee.

Evaluation of knee kinematics and forces during computer simulation

Knee kinematics, patellofemoral contact force, and quadriceps force were computed during stair climbing (from 86° to 6° of knee flexion) in the simulation. For knee kinematics, AP translation of the femoral component relative to the tibial insert and the lowest points of the medial and lateral condyles on the surface of the tibial insert were evaluated. AP translation of the femoral component relative to the tibial insert was defined as anterior (positive) or posterior (negative) to the midline of the tibial tray.

Six different angles (0° to 10°) of posterior tibial slope were simulated to evaluate the effect of posterior tibial slope on knee kinematics and forces in this study. A posterior tibial slope of zero degrees was defined as perpendicular to the tibial mechanical axis, defined as the line connecting the center of the insert to the center of the ankle. We changed the posterior tibial slope angle at 2° intervals ranging from 0° to 10° based on the origin of the coordinates (the center of the tibial insert) in the sagittal alignment.

The anterior post-cam mechanism in BCS TKA was also evaluated using a finite element (FE) model when anterior contact between the anterior aspect of the tibial post and the femoral component occurred with the knee near full extension. FE simulations were performed using FEMAP (Siemens PLM Software, Plano, TX, USA). The femoral component, which is similar to the Co-Cr-Mo alloy femoral component, was modeled as a linear elastic body. The tibial insert consisting of ultra-high molecular weight polyethylene was modeled as a nonlinear elastoplastic body. The Young’s modulus was set at 220 GPa for the femoral component and 0.9 GPa for the tibial insert. Poisson’s ratio was set at 0.31 and 0.45, respectively. The mesh of the femoral component and the tibial insert was generated based on 0.5 mm tetrahedral elements. The generated mesh contained a total of 597,570; 637,093; and 493,919 nodes for the femoral component and 500,530; 523,973; and 600,790 nodes for the tibial insert. The resulted from having 405,722; 432,970; and 408,017 total elements for the femoral component and 346,627; 363,242; and 341,972 total elements for tibial insert, for simulations with posterior tibial slopes of 6°, 8° and 10°, respectively. The maximum von Mises stress on the anterior aspect of the tibial post was analyzed.

Validation of the computer simulation model

Clinical (in vivo) data were used to validate the computational model. Fifteen knees (3 male and 12 female) received the Journey 2 BCS implant used in our computer simulation. Seven of these knees were chosen to validate the computer model after matching for sex and implant size (femoral component: size 3, tibial component: size 3, insert: 9 mm, patella component: 29 mm). Mean age was 71.9 ± 2.5 years, mean posterior tibial slope was 3.1° ± 1.8°, and mean postoperative follow-up was 13.0 ± 1.8 months. Continuous sagittal radiographic images were obtained in each patient during stair climbing using a flat-panel detector (Hitachi, Clavis, Tokyo, Japan), and analyzed using a 2D-3D image-matching technique [8]. The lowest points of the medial and lateral condyles relative to the tibial insert in the computer simulation were compared to clinical data. This study was approved by the institutional review board of Kyushu University (No. 25–74). Informed consent was obtained from all patients prior to study participation.

Statistical analysis

To investigate the reliability and reproducibility of measurement in this simulation, intraobserver and interobserver reliabilities were assessed by intraclass correlation coefficients [ICC (1,1) and

ICC (2,1), respectively] [26]. All measurements were done by two orthopedic surgeons (MH and YM) at an interval of more than 1 week. The ICC (1,1) and ICC (2,1) of the measurement in this simulation were perfect.

Results

Knee kinematics in the simulation

The femoral components translated anteriorly during stair climbing (from 86° to 6° of flexion) (Fig. 2). Increases in posterior tibial slope resulted in a more posterior position of the femoral component relative to the tibial insert and reduced the amount of AP translation. At 65° of knee flexion, anterior sliding of the tibial component occurred only when the posterior tibial slope was 10° (Fig. 2). At 65° of knee flexion, there was an area of contact between the posterior aspect of the tibial post and the femoral component if the posterior tibial slope was less than 10°, but there was no engagement of the post-cam system at 10° (Fig. 3).

Anterior contact between the anterior aspect of the tibial post and the femoral component was observed with the knee near full extension if the posterior tibial slope was 6° or more (Fig. 2). In contrast, no anterior contact occurred when the posterior tibial slope was less than 6°, and there was no contact between the anterior aspect of the tibial post and the femoral component.

Patellofemoral contact force and quadriceps force in the simulation

Both patellofemoral contact force and quadriceps force increased rapidly at 65° of knee flexion when maximum vertical load was placed on the knee joint (Fig. 4, 5). After peaking, the forces decreased gradually with knee extension. Increasing posterior tibial slope decreased both types of maximum forces at 65° of knee flexion (Table 1). An increase of 10° in posterior tibial slope (relative to 0°) led to a 4.8% decrease in maximum patellofemoral contact force and a 1.2% decrease in maximum quadriceps force.

Knee contact conditions in the simulation

Figure 6 shows contours of equivalent maximum von Mises stress on the anterior aspect of the tibial post when anterior contact occurred near full knee extension. The area of contact was a horizontal band on the anterior aspect of the tibial post. Concentrated stress on the center of the anterior aspect of tibial post was observed when the posterior tibial slope was above 6°. Maximum equivalent von Mises stress increased by increasing posterior tibial slope.

Validation: comparing simulation and in vivo knee kinematics

In the computer model, the lowest points of both the medial and lateral condyles in the femoral component were similar to the measured in vivo data (Fig 7, 8). The lowest point on the medial condyle of the femoral component was located almost in the center of tibial insert and the lowest point of the lateral condyle had moved from a posterior position to the center during knee extension. The predicted knee kinematics were almost within the range of inter-specimen variability.

Discussion

The most important findings of the present study were that BCS TKA has a wide acceptable range of posterior tibial slope that avoids knee instability, even though increased posterior tibial slope can result in knee instability similar to anterior sliding of the tibial component. This study showed that no anterior sliding of the tibial component occurs if the posterior tibial slope is less than 10°. Kim et al. reported that many postoperative knees achieved postoperative sagittal alignment of the tibial component between 0° to 7° [10]. Therefore, BCS TKA is unlikely to cause knee instability when implanted using regular surgical techniques even though the computer simulation showed that anterior sliding of the tibial component occurs with 10° of posterior tibial slope.

Increases in posterior tibial slope induce a more posterior position of the femoral component. A more posterior contact position between the femorotibial components leads to a greater quadriceps lever arm, which improves the efficiency of movement and contributes to lower quadriceps and patellofemoral contact forces [7,25]. In the present study, increasing posterior tibial slope decreased both maximum forces at 65° of knee flexion, but the rate of decrease from 0° to 10° was relatively small (4.8% for maximum patellofemoral contact force and 1.2% for maximum quadriceps force). In contrast, increasing posterior tibial slope results in anterior sliding of the tibial component, which should be avoided for long-term TKA success [5,13]. Hamai et al. reported that increasing posterior tibial slope was linked to anterior sliding of the femoral component during mid-flexion of the knee using a 2D-3D image-matching technique [9]. Surgeons should prioritize adverse effects over the positive effect of increasing posterior tibial slope for long-term survival.

This study used KneeSIM as the modeling program; several papers have reported that it yields reproducible simulations of knee kinematics [4,15,17,18]. From our simulated model validated with in vivo data, the lowest points of both the medial and lateral condyle translated anteriorly with BCS TKA. The amount of translation on the lateral side was greater than on the medial side during stair climbing. Knee kinematics in the simulation showed similar trends and was almost within the range of inter-specimen variability for the clinical in vivo data. BCS TKA incorporates both anterior and posterior post-cam mechanisms to reduce abnormal kinematics resulting from AP instability by replicating cruciate ligament function. In addition, the tibial articular geometry with BCS TKA guides posterior motion during knee flexion with less posterior motion in the medial compartment than the lateral compartment. Our simulation model showed knee kinematics that were consistent with the design concept of BCS TKA.

A 15% decrease in maximum patellofemoral contact force and a 6% decrease in maximum quadriceps force with a 15° increase in posterior tibial slope (relative to 0°) were shown in our previous computer simulation study of conventional PS TKA [19]. Anterior sliding of the tibial component at 65° of knee flexion occurred with more than 5° of posterior tibial slope and anterior contact between the tibial post and the femoral component was observed near full extension with more than 10° of posterior tibial slope. The knee kinematics of BCS TKA and conventional PS TKA are different even through all the same parameters were used in both studies.

BCS TKA had a more acceptable range of posterior tibial slope than conventional PS TKA with regards to anterior sliding of the tibial component at 65° of knee flexion. One of the main reasons is the shape of the tibial insert, which was designed to be medially concave and laterally convex in the sagittal plane. In addition, the posterior lip is higher than the anterior lip on the medial side of the tibial insert. These features can result in a sustainable stable condition during stair climbing by increasing posterior tibial slope. Our study suggested that BCS TKA has an area of contact area between the posterior aspect of the tibial post and the femoral component that prevents excessive posterior translation of the medial compartment if the posterior tibial slope was less than 10°.

Anterior contact between the anterior aspect of the tibial post and the femoral component occurred above 6° of posterior tibial slope with BCS TKA in this study. The BCS TKA anterior post-cam mechanism is located more anteriorly than that of conventional PS TKA to replicate the function of the ACL in a normal knee. Anterior contact between the tibial post and the femoral component in PS TKA was observed near full extension with posterior tibial slope of 10° or more in our previous computer simulation; however, this contact was unexpected because it was not part of the design concept. In contrast, the results of BCS TKA were not surprising given the design of the anterior post-cam mechanism. Kuwashima et al. reported that Journey 2 BCS demonstrated no excessive peak stress at any flexion angle based on contact stress analysis of the anterior tibial post because of the concave femoral anterior cam and convex aspect of the tibial post in the axial plane [12]. They also reported that the percentage of contact area was lower than with other PS designs. Anterior contact is considered relatively safe because our study also suggested that contact stress was less with a smaller posterior tibial slope.

There are several limitations to this study. First, only weight-bearing stair climbing was analyzed because we compared the computer simulation with the same activity that had available clinical data and our previous study using conventional PS TKA. Second, only one model size (small female knee) was simulated with BCS TKA in this study. The condition of the knee might be more complicated with cruciate-retained TKA due to the effect of posterior tibial slope on PCL function [21,28]. As the design concept for BCS TKA is to promote normal knee kinematics, it may be affected more by posterior tibial slope differences than PS TKA. Finding from our model cannot be generalized to the entire patient population. In addition, the soft tissue material properties were not subject specific; instead, they were based on the literature. Tanaka et al. suggested that more exact simulation results can be generated by making an individual model with patient-specific ligament insertion points, and that it would be possible to simulate various postoperative conditions accurately for individual patients [23]. However, our computer model was validated adequately with in vivo data, which contains inter-specimen variability. In addition, we evaluated the effect of posterior tibial slope with other factors held constant. Third, clinical conditions of the knee (e.g., postoperative knee scores) affected by posterior tibial slope were not evaluated in the present study because the postoperative course of patients with in vivo data was as short as one year. More research on long-term survival is necessary. Despite these limitations, our study demonstrated the relationship between posterior tibial slope and knee kinematics in BCS TKA. It was found that BCS TKA is more stable and that it is not as affected by increases in posterior tibial slope. Our study helps surgeons determine the optimal posterior tibial slope during surgery with BCS TKA; posterior tibial slope should not exceed 10° in routine clinical practice.

Conclusion

BCS TKA is not associated with anterior sliding of the tibial component if the posterior tibial slope is less than 10°. Surgeons should prioritize avoiding adverse effects over attempting to achieve positive effects from increasing posterior tibial slope, even if BCS TKA is unlikely to cause knee instability when implanted using regular surgical techniques.

Acknowledgments

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List of Abbreviations

TKA: total knee arthroplasty, BCS: bi-cruciate stabilized, PS: posterior-stabilized, ACL: anterior cruciate ligament, PCL: posterior cruciate ligaments, AP: anteroposterior, FE: finite element, 2D: 2-dimensional, 3D: 3-dimensional, ICC: intraclass correlation coefficients, PTS: posterior tibial slope

Competing Interests

Hideki Mizu-uchi: Zimmer Biomet; Paid presenter or speaker. Ken Okazaki: Zimmer Biomet;

Paid presenter or speaker. Smith & Nephew; Paid presenter or speaker. Johnson & Johnson; Paid presenter or speaker. Pfizer Inc.; Research support. Cyfuse Inc.; Research support.

Authors' Contributions

MH collected and analyzed the data and drafted the manuscript. HM conceived of the study, participated in its design, collected and analyzed the data and coordination and helped to draft the manuscript. HM is also the corresponding author. KO collected and analyzed the data, and assisted in drafting the manuscript. TK, KM and YM collected and analyzed the data. SH assisted in drafting the manuscript. YN gave final approval to the manuscript.

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Figure legends

Fig. 1:

Lateral view of the LifeMOD/KneeSIM 2010 knee simulator model used in the present study

Fig. 2:

AP translation of the femoral component relative to the tibial insert during stair climbing

AP: anteroposterior

PTS: posterior tibial slope

Fig. 3:

Posterior contact of the BCS TKA post-cam at 65° of knee flexion (posterior view)

BCS: bi-cruciate stabilized

TKA: total knee arthroplasty

Fig. 4:

Patellofemoral contact force from 86° to 6° of knee flexion during simulated stair climbing

PTS: posterior tibial slope

Fig. 5:

Quadriceps force from 86° to 6° of knee flexion during simulated stair climbing

PTS: posterior tibial slope

Fig. 6:

Maximum equivalent stress distribution in the tibial insert with anterior contact (anterior view)

(a) Posterior tibial slope of 6°

(b) Posterior tibial slope of 8°

(c) Posterior tibial slope of 10°

Fig. 7:

Lowest point on the condyle of the femoral component relative to the tibial insert during stair climbing from 70° to 10° of knee flexion based on simulated and in vivo data

(a) Lowest point on the medial condyle

(b) Lowest point on the lateral condyle

AP: anteroposterior

Fig. 8:

Lowest point on the condyle plotted on the tibial insert during stair climbing from 70° to 10° of knee flexion

(a) Simulated data (posterior tibial slope of 4°)

(b) In vivo data (posterior tibial slope of 4.6°)

Figure 1

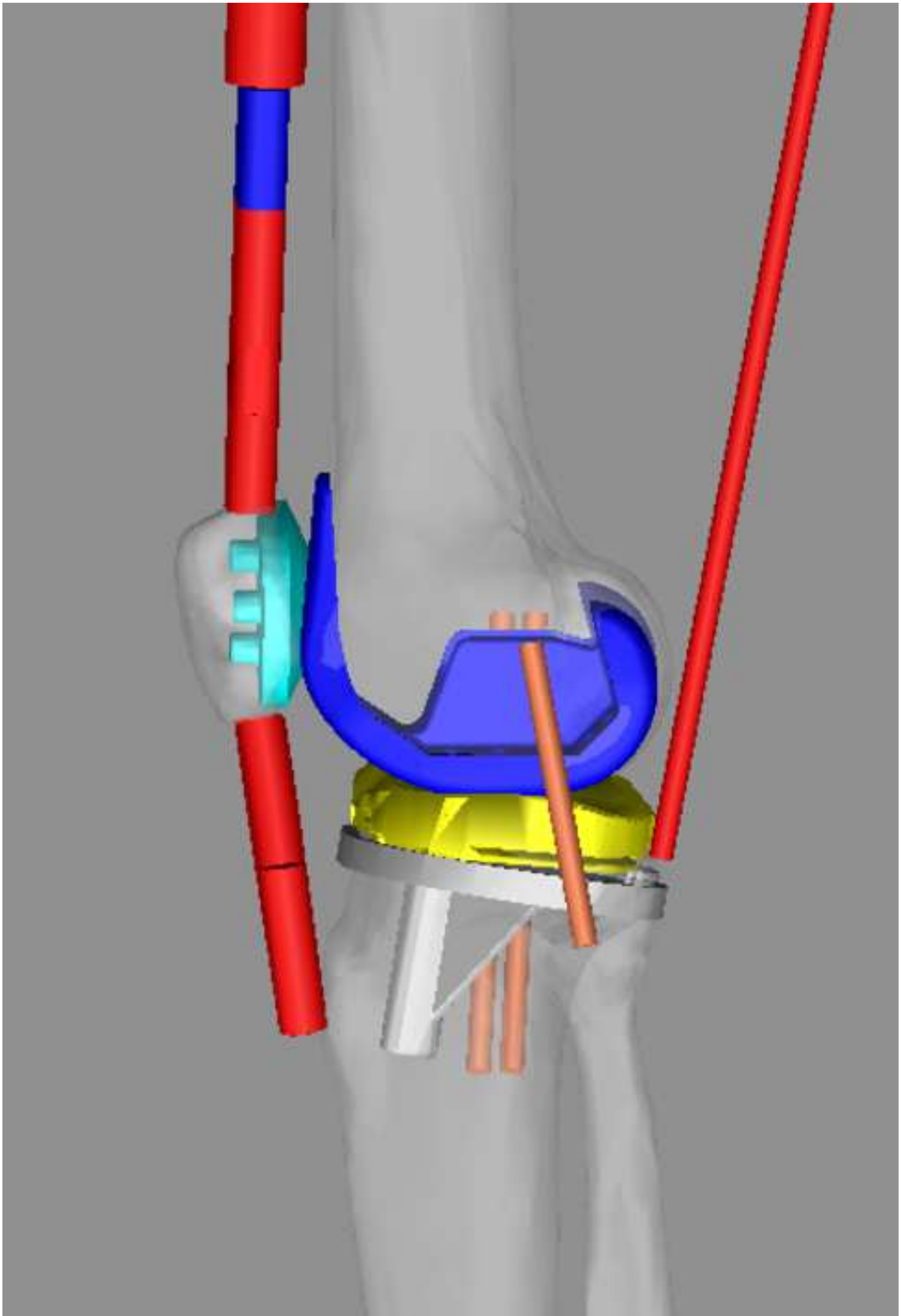
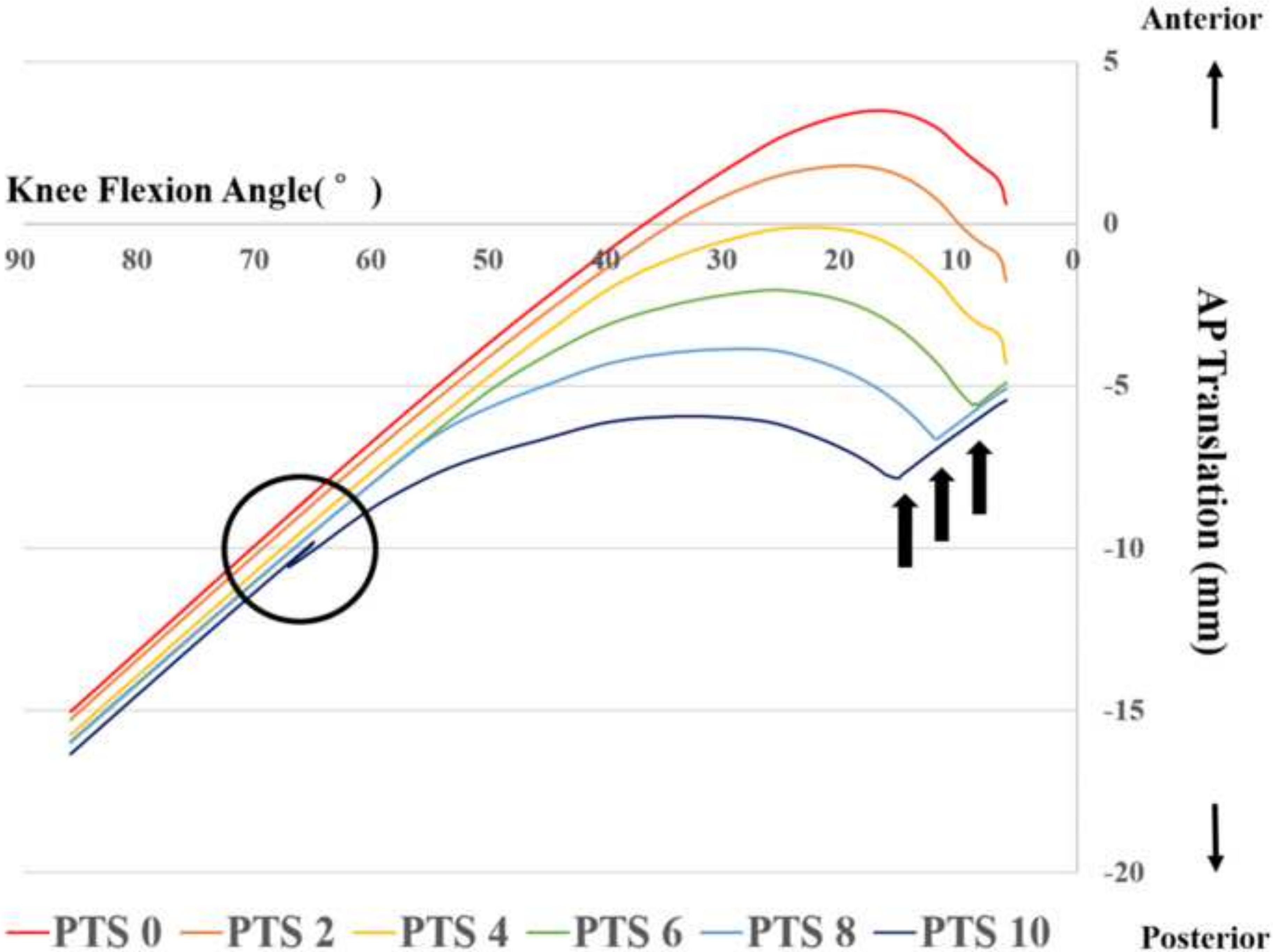


Figure 2



Anterior
Knee Flexion Angle(°)
90
80
70
60
50
40
30
20
10
0
5
0
-5
-10
-15
-20
AP Translation (mm)
Posterior
PTS 0
PTS 2
PTS 4
PTS 6
PTS 8
PTS 10

Figure 3

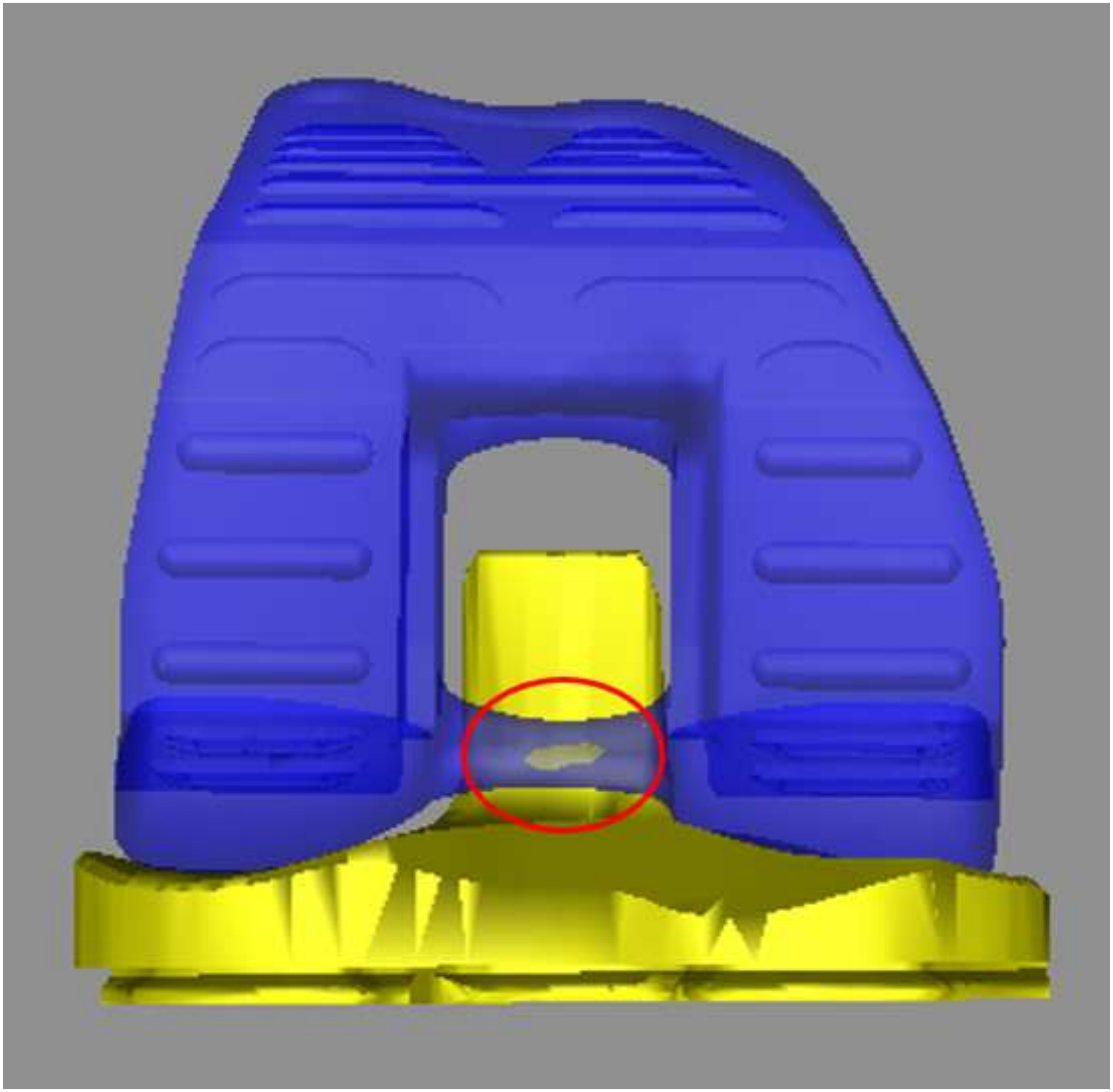


Figure 4

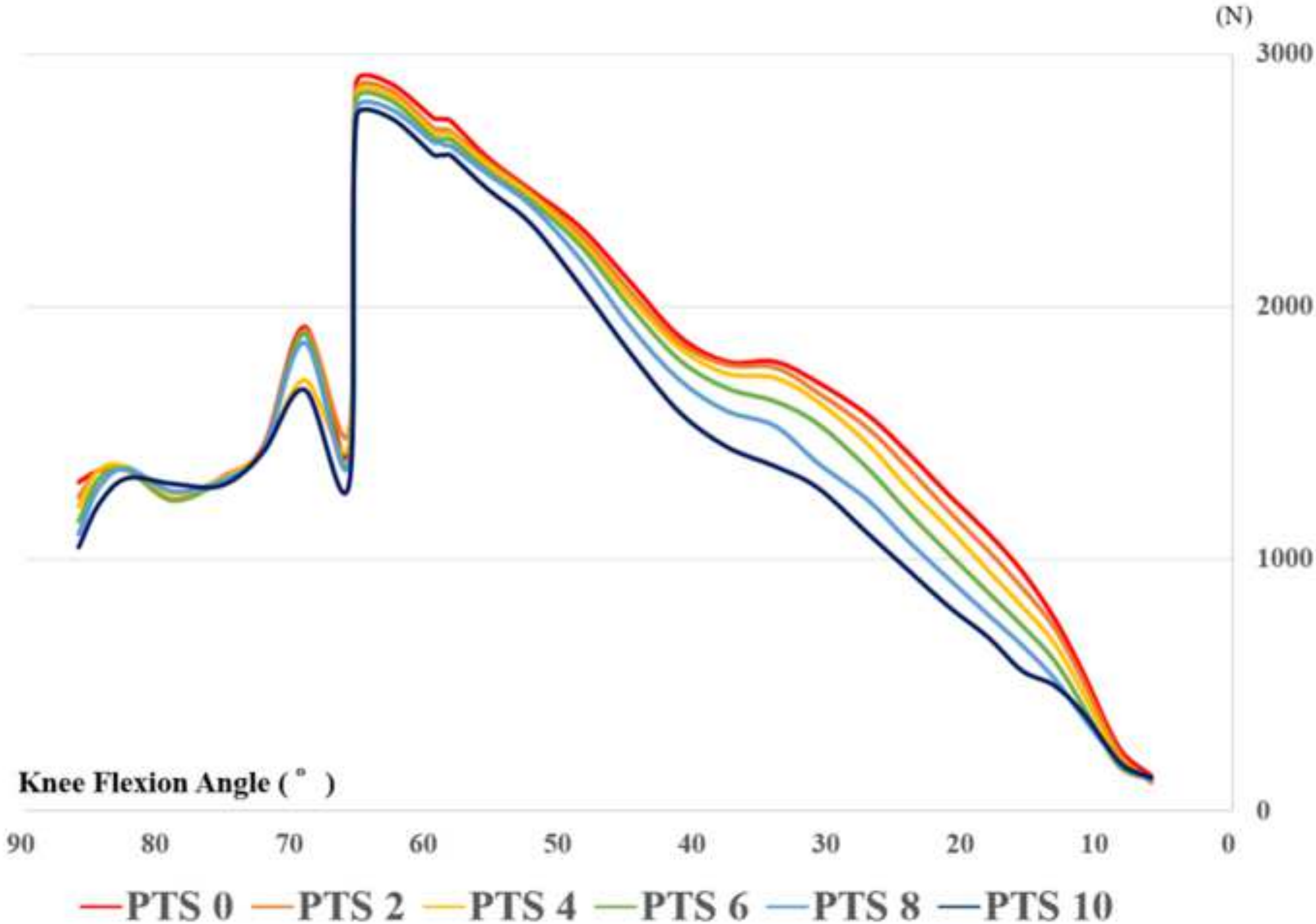
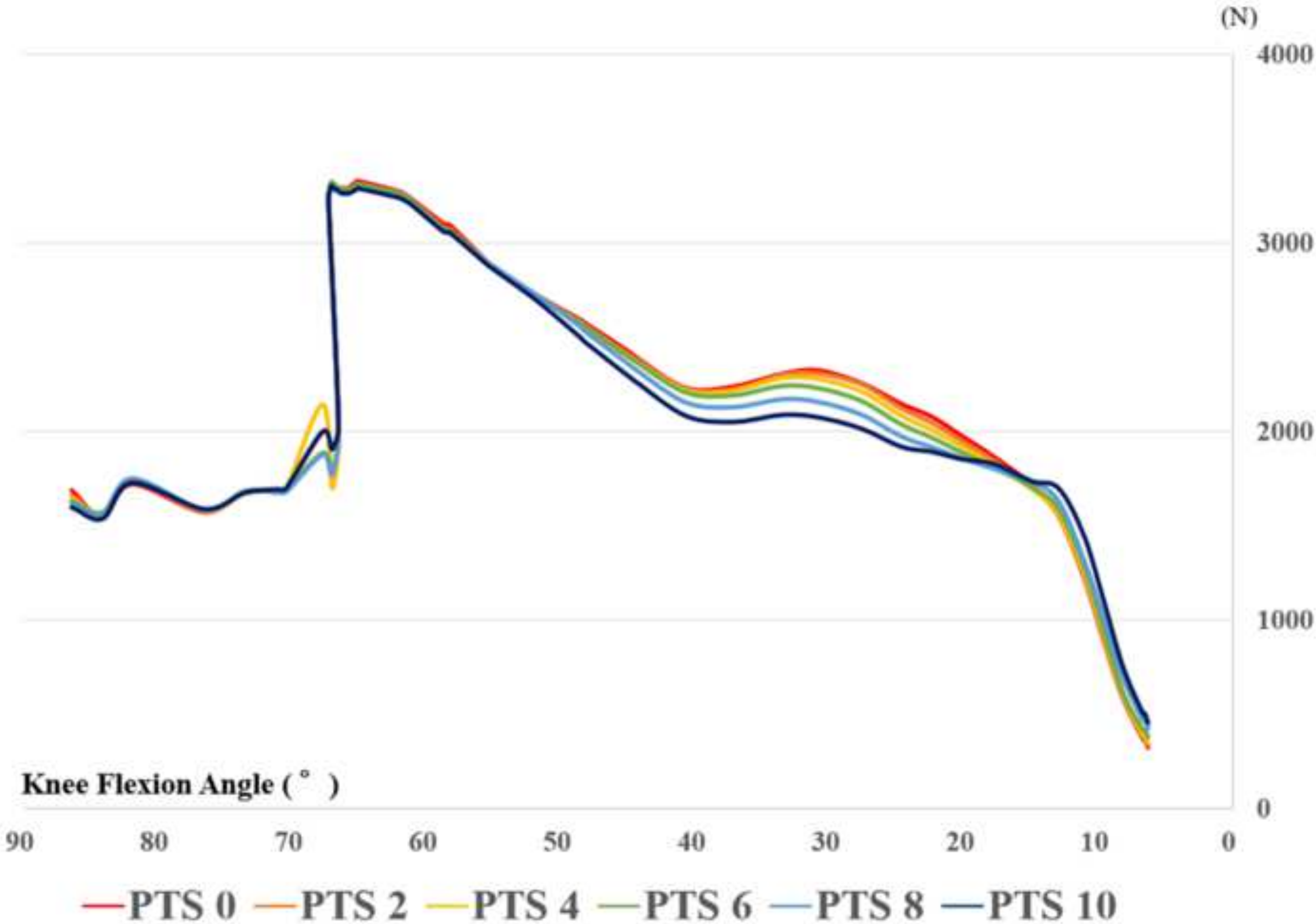

(N)
3000
2000
1000
0
Knee Flexion Angle (°)
90
80
70
60
50
40
30
20
10
0
PTS 0
PTS 2
PTS 4
PTS 6
PTS 8
PTS 10

Figure 5



(N)
4000
3000
2000
1000
0
Knee Flexion Angle (°)
90
80
70
60
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40
30
20
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0
PTS 0
PTS 2
PTS 4
PTS 6
PTS 8
PTS 10

Figure 6

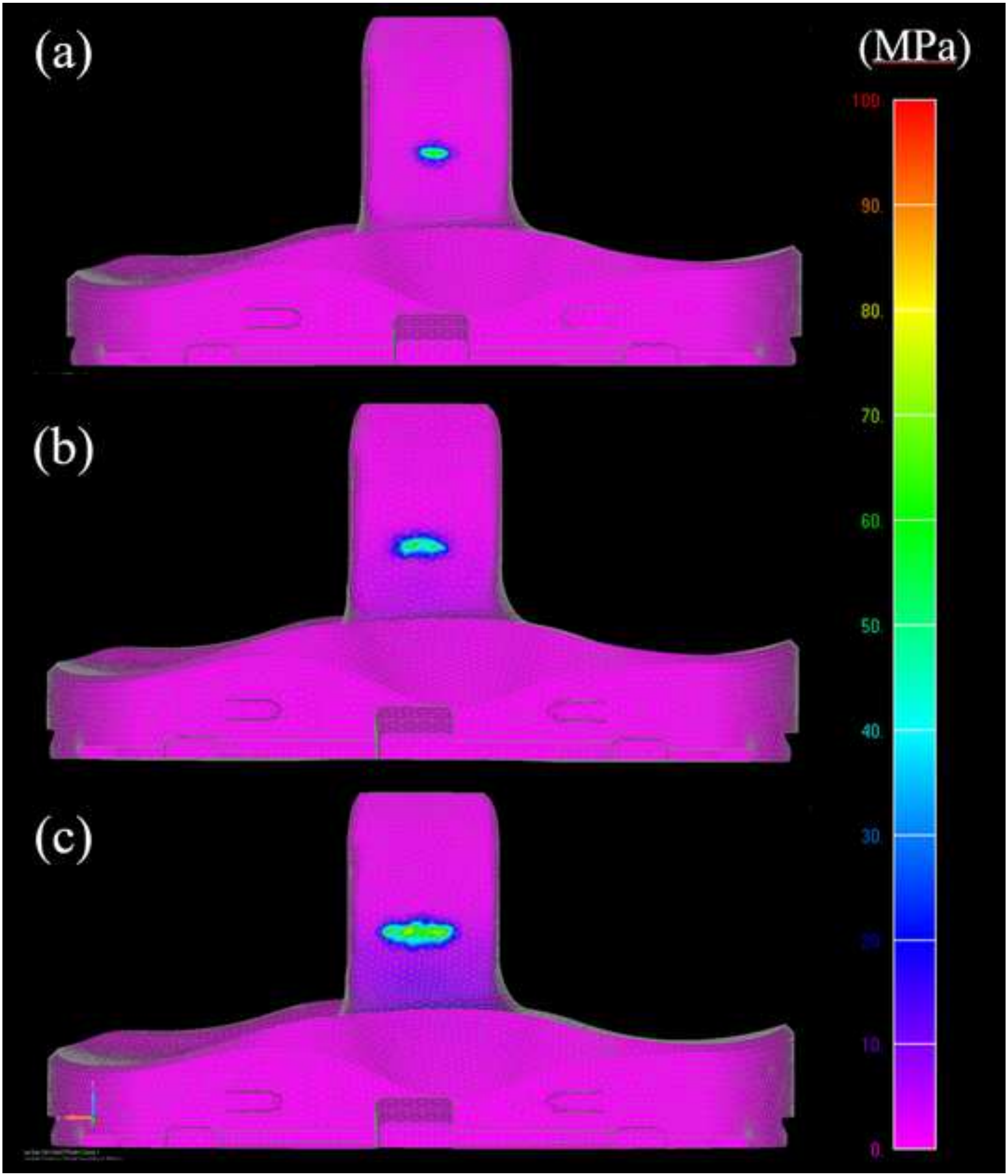
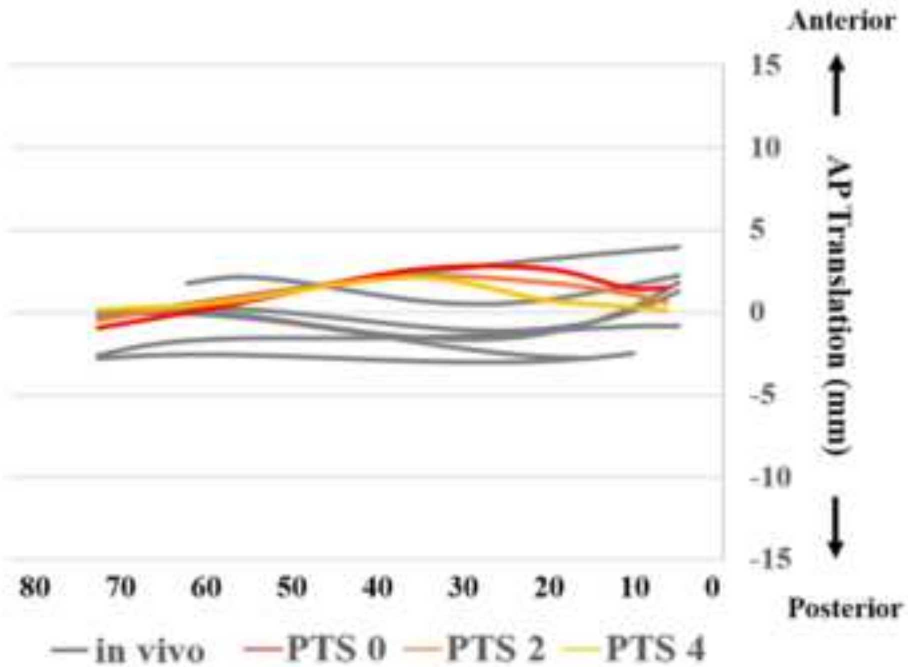
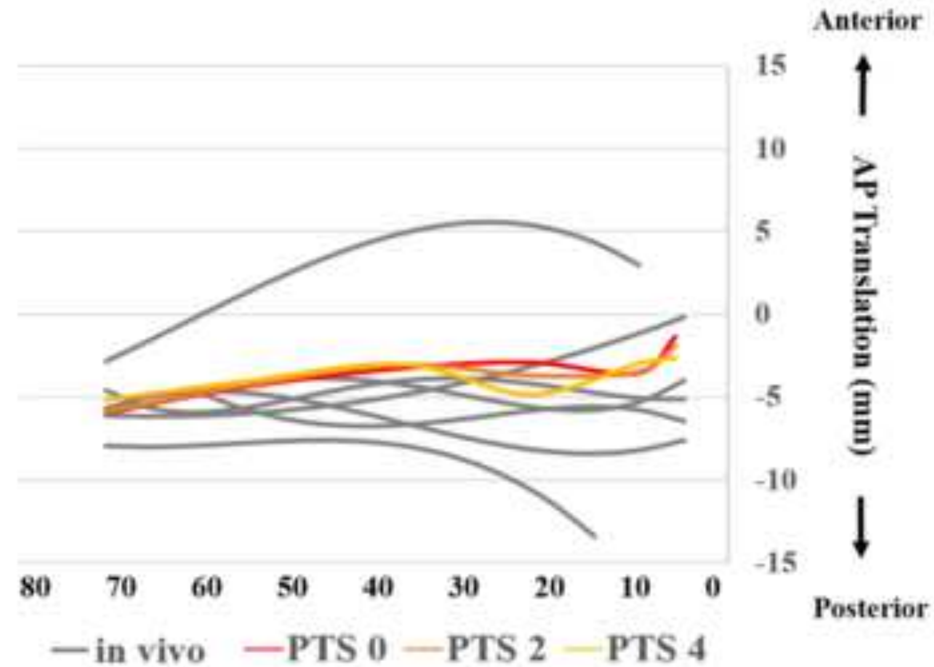


Figure 7



Anterior
15
10
5
0
-5
-10
-15
AP Translation (mm)
Posterior
80
70
60
50
40
30
20
10
0
—in vivo
—PTS 0
—PTS 2
—PTS 4

(a)



Anterior
15
10
5
0
-5
-10
-15
AP Translation (mm)
Posterior
80
70
60
50
40
30
20
10
0
—in vivo
—PTS 0
—PTS 2
—PTS 4

(b)

Figure 8

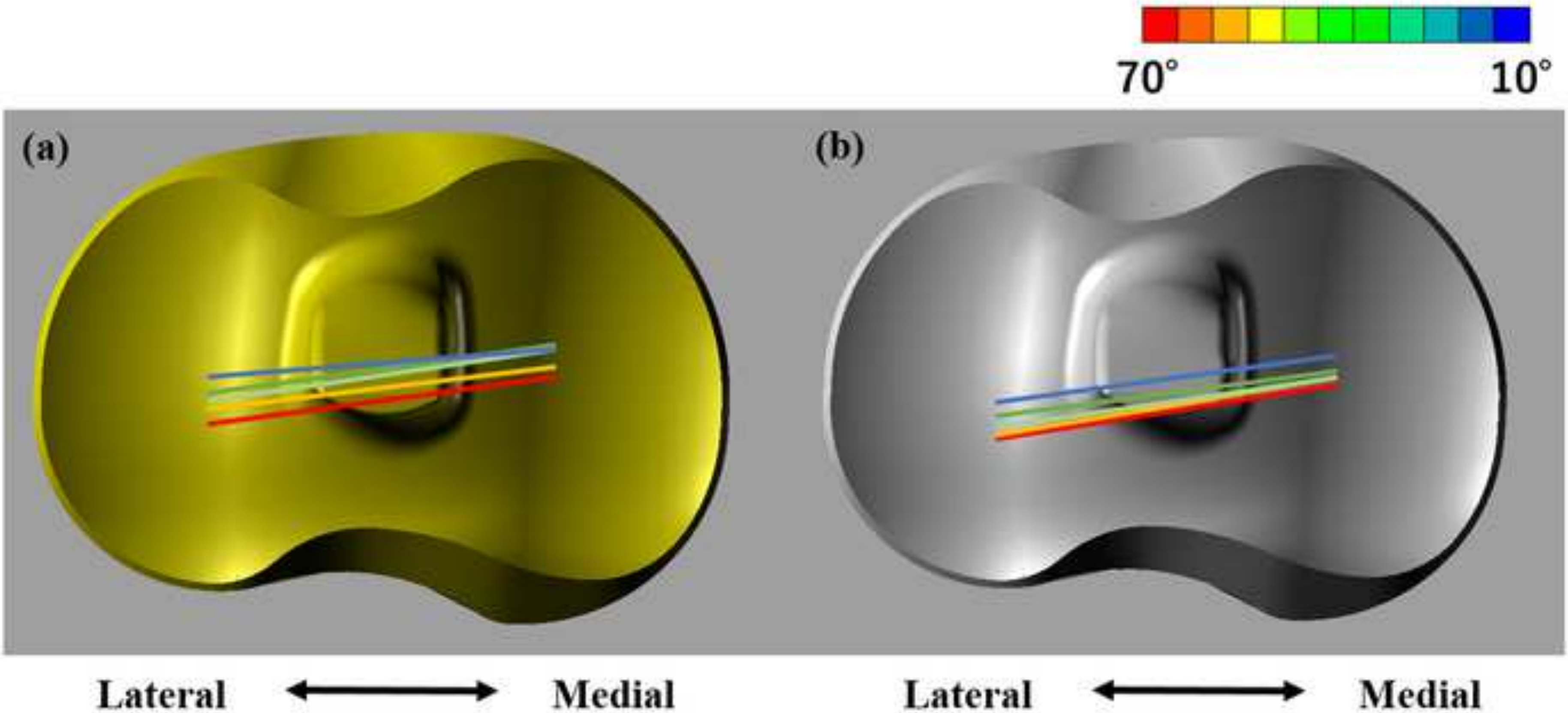


Table 1

Maximum patellofemoral contact force and quadriceps force at 65° of knee flexion in the simulation

Force	Posterior tibial slope					
	0°	2°	4°	6°	8°	10°
PF contact (N)	2935.8	2903.7	2882.9	2865.6	2826.3	2793.9
Quadriceps (N)	3369.2	3354.4	3350.8	3342.5	3330.5	3328.8

PF: patellofemoral