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A computational simulation study to determine the biomechanical influence of posterior condylar offset and tibial slope in cruciate retaining total knee arthroplasty

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Objectives

Posterior condylar offset (PCO) and posterior tibial slope (PTS) are critical factors in total knee arthroplasty (TKA). A computational simulation was performed to evaluate the biomechanical effect of PCO and PTS on cruciate retaining TKA.

Methods

We generated a subject-specific computational model followed by the development of ± 1 mm, ± 2 mm and ± 3 mm PCO models in the posterior direction, and -3°, 0°, 3° and 6° PTS models with each of the PCO models. Using a validated finite element (FE) model, we investigated the influence of the changes in PCO and PTS on the contact stress in the patellar button and the forces on the posterior cruciate ligament (PCL), patellar tendon and quadriceps muscles under the deep knee-bend loading conditions.

Results

Contact stress on the patellar button increased and decreased as PCO translated to the anterior and posterior directions, respectively. In addition, contact stress on the patellar button decreased as PTS increased. These trends were consistent in the FE models with altered PCO. Higher quadriceps muscle and patellar tendon force are required as PCO translated in the anterior direction with an equivalent flexion angle. However, as PTS increased, quadriceps muscle and patellar tendon force reduced in each PCO condition. The forces exerted on the PCL increased as PCO translated to the posterior direction and decreased as PTS increased.

Conclusion

The change in PCO alternatively provided positive and negative biomechanical effects, but it led to a reduction in a negative biomechanical effect as PTS increased.

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Keywords: Total knee arthroplasty, Posterior condylar offset, Posterior tibial slope

Article focus

- There has been a lack of biomechanical information on the effect of posterior condylar offset (PCO) and posterior tibial slope (PTS) on postoperative outcomes in cruciate-retaining total knee arthroplasty (TKA).
- The contact stress on the patellar button and the forces on the posterior cruciate ligament (PCL), patellar tendon and quadriceps muscles were investigated in relation to PCO and PTS changes under deep knee-bend conditions using computational simulation.

Key messages

- An increase in PCO of more than 2 mm and a decrease in PTS should be avoided to prevent degeneration of the PCL.
- Forces on the PCL increase with translation of PCO in the posterior direction.
- Contact stress on the patellar button and forces on the patellar tendon and quadriceps muscles increased with translation of PCO in the anterior direction.
- Contact stress on the patellar button and forces on the PCL, patellar tendon and quadriceps muscles decreased as PTS increased.

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Strengths and limitations

- One strength is that we evaluated the biomechanical effect of PCO and PTS by using finite element (FE) analysis, which used to be impractical for direct measurement in an experimental approach.
- Limitations include the fact that the FE analysis study did not include clinical data, and further study on posterior-stabilized TKA and mobile-type bearing TKA is required.

Introduction

One of the important goals of total knee arthroplasty (TKA) is to improve the functional flexion angle while mantaining stability. TKA leads to an improvement in the functioning of the knee joint that is hampered by osteoarthritis, but TKA cannot restore the full range of movement (ROM).[1,2] The postoperative ROM can be influenced by many factors such as patient factors, prosthesis design, surgical technique and rehabilitation. In terms of patient factors, the preoperative flexion angle is known to be the most influential factor in determining the postoperative flexion angle.[3,4] In terms of surgical factors, posterior condylar offset (PCO), intraoperative soft-tissue balancing, posterior cruciate ligament (PCL) tension and posterior tibial slope (PTS) have been associated with postoperative ROM.[5-10] In addition to these factors, the surgical techniques and bone resections are likely to influence postoperative ROM.[11] Patient and surgical factors have in the past been suggested as the key factors influencing the postoperative ROM in TKA; however, PCO and PTS are as important as patient and surgical factors.[7,12,13] At high flexion, mobility is restricted because of impingement between the posterior femoral cortex and tibial plateau borders.[14,15] Thus, carefully planned bone resections may increase ROM prior to impingement and substantially improve the knee flexion range.[11]

Braun et al[16] reported that restoration of the tibial slope has been shown to delay tibiofemoral impingement. Bellemans et al[7] showed that the magnitude of PCO correlated with the flexion angle after posterior cruciate-retaining (CR)-TKA. Massin and Gournay[11] suggested that a high PCO increased the PTS and a more posterior femorotibial contact point can increase flexion. Increases in the PTS and posterior translation of the PCO have both been shown to increase knee flexion and delay tibiofemoral impingement, but our understanding of these variables, their interrelationships and their role in TKA is inadequate.[7,11,16-19]

It is difficult to perform restoration of the PCO and PTS in knee joints because of the mismatch between the prosthesis and bone geometry, tightness in the flexion gap and wide variation in the preoperative PTS.[3,20] The advantage of a computational simulation using a single subject is that we could determine the effects of component alignment within the same subject without the effect of variables such as weight, height, bone geometry, ligament properties and component size.[21] In addition, to our knowledge, there are no reports on PCO preservation and the corresponding change in PTS, particularly its biomechanical effects on TKA. It is challenging to perform experimental measurements to evaluate directly the contact stress distribution in the polyethylene (PE) insert and patellar button, and ligament force and quadriceps muscle force. However, these can be evaluated in computational simulation using finite element (FE) analysis.

The purpose of this study is to determine the biomechanical effects of PCO change on the corresponding change in PTS under deep knee-bend conditions. We analyzed the contact stress in the patellar button, as well as the forces on the PCL, patellar tendon and quadriceps muscle using FE analysis.

Materials and Methods

Normal healthy model. Based on the previously validated and published FE model for a knee joint,[22-25] the FE model used in this study includes the following features: a 3D non-linear FE model of a normal knee joint that was developed using data from the CT and MRI scans of a healthy 37-year-old male subject. The CT and MRI models were developed with a slice thickness of 0.1 mm and 0.4 mm, respectively. The medical history of the subject revealed neither musculoskeletal disorders nor any related diseases arising from malalignment in the leg, all of which indicated a healthy knee joint.

The reconstructed CT and MRI models were combined with a positional alignment of each model by using commercial software (Rapidform version 2006; 3D Systems Korea Inc., Seoul, South Korea) that models bone structures as rigid bodies using four-node shell elements.[26] Additionally, the major ligaments were modelled with non-linear and tension-only spring elements.[27,28] The ligament insertion points were referenced to the anatomy from the MRI sets of the subject and descriptions found in the literature.[29-31]

The interfaces between the cartilage and the bones were modelled to be fully bonded. The contacts between the femoral cartilage and meniscus, meniscus and tibial cartilage and femoral and tibial cartilages were modelled for both the medial and lateral sides, resulting in six contact pairs.[24] The contacts in all articulations adopted a finite sliding frictionless hard contact algorithm with no penetration.[24] Material properties for articular cartilage, meniscus and ligaments are shown in Table I.

Convergence was defined as a relative change of $< 5\%$ between two adjacent meshes. The mean element size of the simulated cartilage and menisci was 0.8 mm.

Change in PCO and corresponding PTS FE models. The surgical simulation of a TKA was performed by two experienced surgeons (Y-GK and SKK). Computer-assisted design models of a posterior CR design from the Genesis

Table I. Material properties for finite element model

	Intact model			**Total knee arthroplasty model**	
	Young's modulus (MPa)		Poisson's ratio	Young's modulus (MPa)	Poisson's ratio
Cartilage	15		0.47	×	
Meniscus	120 circumferential direction		0.20 circumferential and radial direction	×	
	20 axial and radial direction		0.30 axial direction	×	
CoCrMo alloy	×			195 000	0.30
UHMWPE	×			685	0.47
Ti_6Al_4V alloy	×			117 000	0.30
PMMA	×			1940	0.40
	Stiffness (n)	Reference strain	Slack length (mm)		
aACL	5000	0.06	33.74	×	
pACL	5000	0.10	28.47	×	
aPCL	9000	-0.10	33.81	○	
pPCL	9000	-0.03	34.92	○	
LCL	4000	0.06	57.97	○	
aMCL	2500	-0.02	86.54	○	
iMCL	3000	0.04	84.72	○	
pMCL	2500	0.05	51.10	○	
PFL	4000	0.06	43.54	○	
OPL	2000	0.07	80.21	○	
lCAP	2500	0.06	55.59	○	
mCAP	2500	0.08	60.13	○	
ALS	2000	0.06	31.69	○	
aCM	2000	-0.27	37.53	○	
pCM	4500	-0.06	34.48	○	

CoCrMo, cobalt chromium molybdenum alloy; UHMWPE, ultra-high-molecular-weight polyethylene; Ti_6Al_4V, titanium alloy; PMMA, poly (methyl methacrylate); ACL, anterior cruciate ligament; PCL, posterior cruciate ligament; LCL, lateral collateral ligament; MCL, medial collateral ligament; PFL, popliteofibular ligament; OPL, oblique popliteal ligament; CAP, posterior capsule; ALS, anterolateral structures; CM, deep medial collateral ligament; a, anterior; p, posterior; i, inferior; l, lateral; m, medial

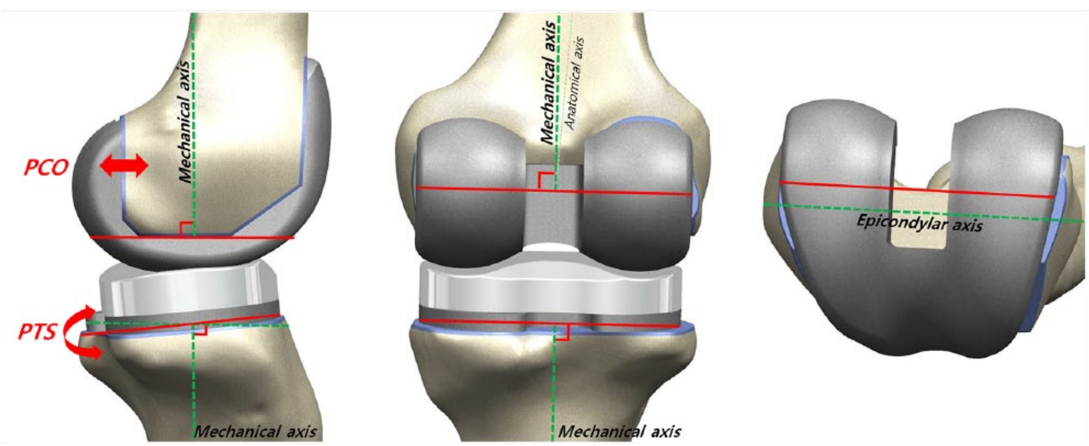


Fig 1a Fig 1b Fig 1c

Diagram showing the orientation of the total knee arthroplasty used in this study in the a) sagittal plane, b) coronal plane and c) transverse plane (PCO, posterior condylar offset; PTS, posterior tibial slope).

II Total Knee System (Smith & Nephew Inc., Memphis, Tennessee) were virtually implanted in the bone geometry. Based on the dimensions of the femur and tibia, devices with sizes seven and six were selected for the femoral component and tibial baseplate, respectively.

In the neutral position, the femoral component was aligned such that the distal bone resection was perpendicular to the mechanical axis of the femur and the anterior and posterior resections were parallel to the clinical epicondylar axis in the transverse plane (Fig. 1). A PCO model identical to the original subject was developed, followed by modifications of each PCO in the model. The femoral component position was adjusted in the anteroposterior direction to avoid notching of the anterior cortex as per the standard surgical protocol. Seven models were developed with - 3 mm, - 2 mm, - 1 mm, 0 mm, + 1 mm, + 2 mm and + 3 mm translation in the posterior direction from the original anatomy (Fig. 2a).

The tibial default alignment was rotated by 0° in relation to the anteroposterior axis, the coronal alignment was 90° to the mechanical axis (Fig. 1) and the sagittal alignments were - 3°, 0°, 3° and 6° to the posterior slope, with an 8 mm resection below the highest point of the lateral plateau (Fig. 2b). This is the lowest point of the PE

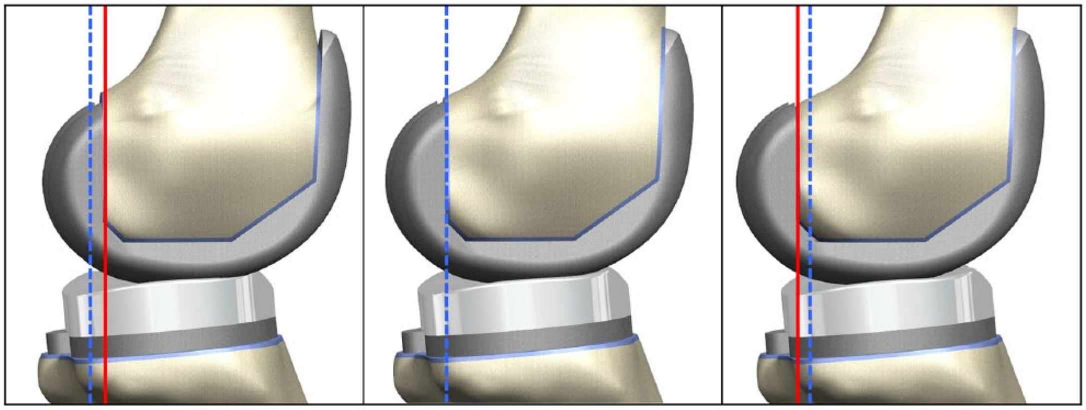


Fig. 2a

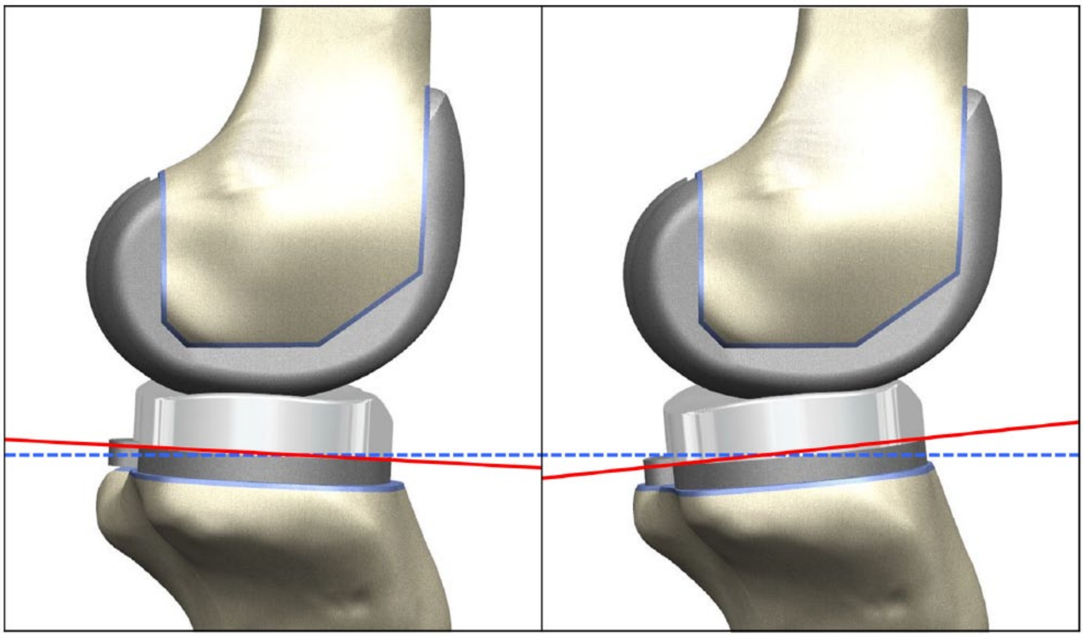


Fig. 2b

Schematic of the knee models with respect to change in a) posterior condylar offset (left) at anterior 3 mm (- 3 mm), (middle) 0 mm and (right) posterior 3 mm (+ 3 mm), and b) posterior tibial slope at - 3° and + 6°.

insert articular surface adjacent to the lowest points of the femoral articular surfaces in extension.

Contact conditions were applied between the femoral component, PE insert and patellar button in TKA. The coefficient of friction between the PE material and metal was chosen to be 0.04 for consistency with previous explicit FE models.[32] The materials of the femoral component, PE insert, tibial component and bone cement corresponded to a cobalt chromium alloy, ultra-high-molecular-weight polyethylene, a titanium alloy, and poly (methyl methacrylate), respectively. As in previous studies,[32-36] these materials were assumed to be homogeneous and isotropic in this case, except for the PE insert. The PE insert was modelled as an elastoplastic material (Table I).

A cement layer was considered with a constant penetration depth of 3 mm into the bone according to the test for different cementing techniques at the femoral and tibial resection surfaces in contact with the femoral and tibial components, respectively.[37,38] The interfaces between the prosthesis and bone were rigidly fixed with consideration of the cement used.[34,39].

The PCO change and the corresponding PTS model topologies provided 6° of freedom to the tibiofemoral (TF) and patellofemoral joints. The FE investigation included two types of loading conditions corresponding to the loads used in the experiments in the study for TKA model validation and model predictions under deep knee-bend loading conditions. The intact model was validated in a previous study,[22-24] and the TKA model was validated by comparing it with models used in a previous study.[40] A conservative ankle force of 50 N and hamstring forces of 10 N were constantly exerted with a linearly rising force and a maximum of approximately 600 N at a

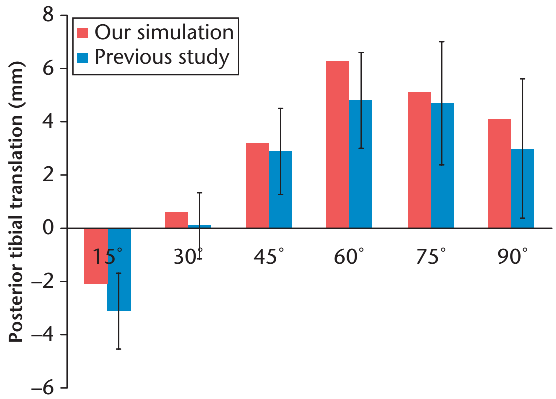


Fig. 3a

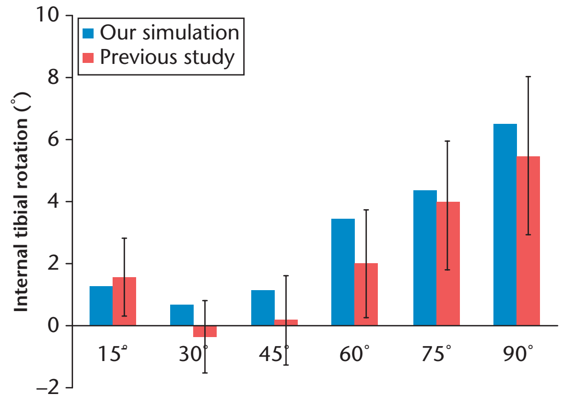


Fig. 3b

Graphs showing the comparison of a) posterior tibial translation and b) internal tibial rotation with previous experiments for total knee arthroplasty model. The error bars represent 1 standard deviation in the experiment.

90° flexion of the quadriceps actuators was exerted for the TKA model under first loading condition.[40]

The second loading condition was deep knee-bend loading applied to evaluate the effect of the change in PCO and the corresponding PTS. The computational analysis was performed with anteroposterior force applied to the femur with respect to the compressive load applied to the hip.[41-43] A proportional integral derivative controller was incorporated into the computational model to allow for the control of the quadriceps in a manner similar to that in a previous experiment.[44] The control system was used to calculate the instantaneous quadriceps displacement required to match a target flexion profile, which was the same as the experiment.[44] Internal-external and varus-valgus torques were applied to the tibia.[41-43]

The FE model was analyzed using the software Abaqus (version 6.11; Dassault Systèmes Simulia Corp., Providence, Rhode Island). The results for the contact stress on the patellar button were evaluated and forces on the PCL, patellar tendon and quadriceps were assessed by analyzing the changes in PCO and the corresponding PTS conditions.

Results

Validation of TKA model. The FE model for the tibia was translated by - 2.1 mm, 0.6 mm, 3.2 mm, 6.3 mm, 5.1 mm and 4.1 mm in the posterior direction, and was internally rotated by 1.3°, 0.7°, 1.2°, 3.5°, 4.8° and 6.6° at 15°, 30°, 45°, 60°, 75° and 90° of flexion, respectively (Fig. 3). There was good agreement between the simulation results and those from a previous experimental study[40] for the range of values under the loading condition applied to the prosthetic implant. In addition, the increase and decrease in tibial translation and rotation from computational simulation showed good concordance with previous experiments by Wünschel et al.[40]

Comparison of contact stresses on the patellar button with change in PCO and corresponding PTS in FE models under deep knee-bend conditions. Figure 4 shows the contact stress on the patellar button with respect to the change in PCO and PTS in the FE model under deep knee-bend conditions.

Contact stresses increased and decreased as the PCO was translated in the anterior and posterior directions, respectively, compared with the neutral position. The contact stress increased and decreased by a mean of 23% and 17%, respectively, in PCO - 3° and + 3° models, compared with the neutral position with respect to PTS change. There was a mean decrease of 23% in the contact stress in the PTS 6° model compared with the PTS - 3° model, with respect to the change in PCO.

Comparison of PCL, patellar tendon and quadriceps muscle forces with change in PCO and PTS in FE models under deep knee-bend conditions. The PCL, patellar tendon and quadriceps muscle forces, with respect to changes in PCO and PTS under deep knee-bend conditions, are shown in Figures 5, 6 and 7. The force exerted on the PCL increased and decreased as the PCO was translated in the posterior and anterior directions, respectively, compared with the neutral position. The force exerted on the PCL remarkably increased in the + 2 mm PCO model compared with the + 1 mm PCO model, which were 102% and 17%, respectively, of the force exerted in the neutral position with + 6° PTS. However, there was no difference in the force exerted on the PCO between the + 2 mm and + 3 mm PCO models with respect to the change in PTS. Additionally, the force exerted on the PCL decreased as PTS increased in all PCO models. There was an identical trend found in the force exerted on the patellar tendon and quadriceps muscle forces. It increased and decreased as PCO was translated in the anterior and posterior directions compared with the neutral model. However, the quadriceps muscle forces decreased as PTS increased, with respect to the change in PCO. A 35% lesser force was required in the 6° PTS model compared with the - 3° PTS model. The - 3 mm PCO model with the - 3° PTS model required the highest quadriceps muscle force. In addition, the force on the patellar tendon decreased

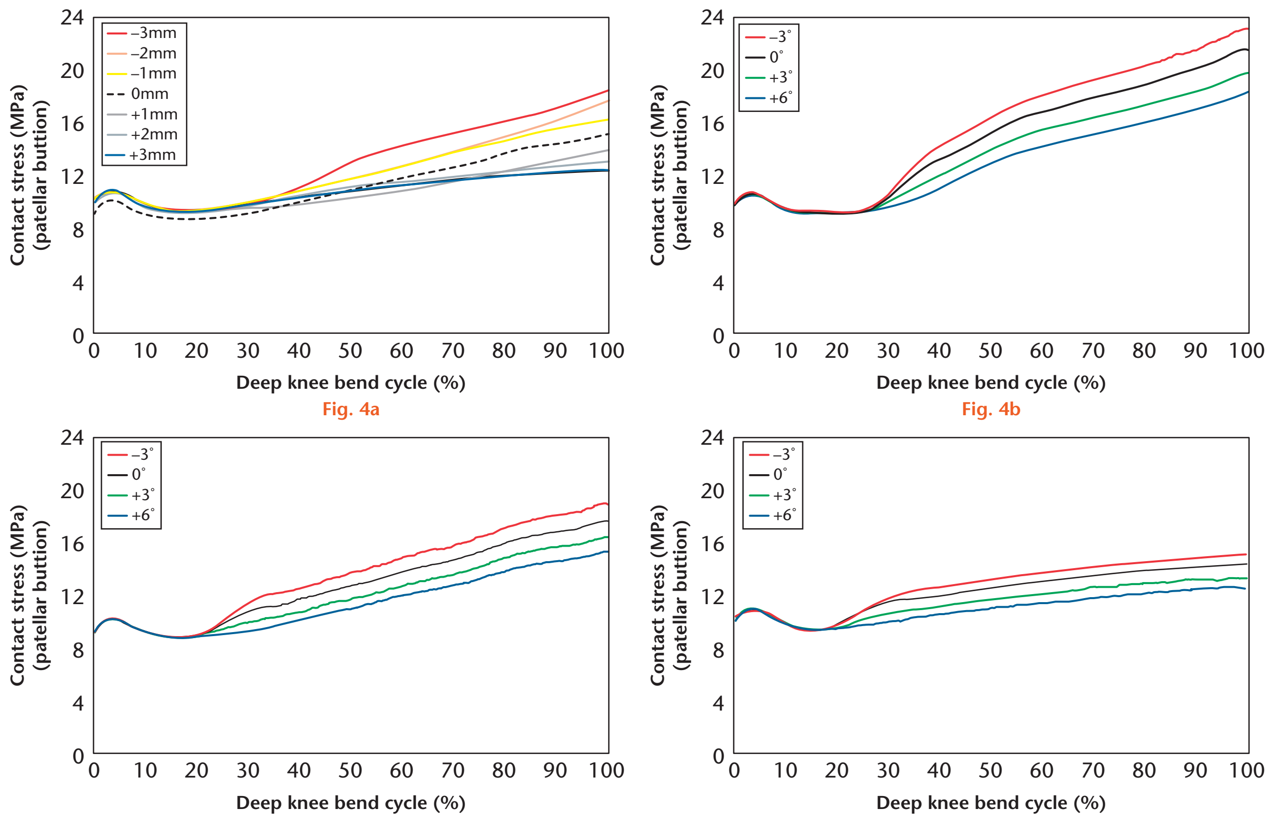


Fig. 4a Fig. 4b Fig. 4c Fig. 4d

Comparison of contact stress on the patellar button: a) with respect to different posterior condylar offset (PCO) in the posterior tibial slope (PTS) 6°; with respect to different PTS in the b) PCO - 3 mm, c) PCO 0 mm and d) PCO + 3 mm models.

as PTS increased with respect to the change in PCO. The force on the patellar tendon decreased by a mean of 15% in the 6° PTS model compared with the - 3° PTS model with respect to the change in PCO. Additional results regarding the rest of the PCO conditions (- 2 mm, - 1 mm, + 1 mm and + 2 mm PCO) can be found in the Supplementary Material.

Discussion

The most important finding of this study was that the biomechanical effect on CR-TKA varied with respect to the translation of the PCO from the posterior to the anterior direction, and that the negative biomechanical effect reduced as PTS increased.

Although the concepts of PCO and PTS were discovered a long time ago, there has recently been a specific interest in determining their functional consequences in TKA. Bellemans et al[7] defined the concept of PCO. They determined fluoroscopically that in 72% of the CR-TKAs considered in their study, the maximum active flexion was limited by direct impingement of the posterior aspect of the tibial component against the posterior aspect of the femur. They also showed in a cadaveric study that surgeons can expect to achieve a mean flexion of 1.7° with every each degree of PTS.[45]

On the other hand, previous studies showed that there was no difference in the postoperative ROM between the knees with 0° and 5° of PTS in posterior-stabilized TKA.[10,46] Bauer et al[47] reported that the correlation between PTS and the maximal knee flexion that could be observed in CR-TKA was not found in posterior-stabilized TKA. Bai et al[48] reported that a reduction in PTS decreased the flexion gap in TKA, and an extremely high PTS may lead to instability and loosening. However, they could not find the relationship between PTS and PCO. There currently exists no consensus on the native PCO or PTS, as a wide range of normal variables have been reported in previous studies.[6-11,14-19,46-48] To our knowledge, no study has investigated the effects of both PCO and PTS on muscle or ligament forces with respect to component alignment and the contact stress on the PE insert and patellar button using computational simulation. Our variables of interest were affected by PCO and PTS in a different manner.

The purpose of this study was to use computational simulation to evaluate the biomechanical effect of PCO models with - 1 mm, - 2 mm, - 3 mm, 0 mm, + 1 mm, + 2 mm and + 3 mm posterior translation and PTS models with 3°, 0°, - 3° from the 6° of the subject's own posterior slope under deep knee-bend conditions. Additionally, the

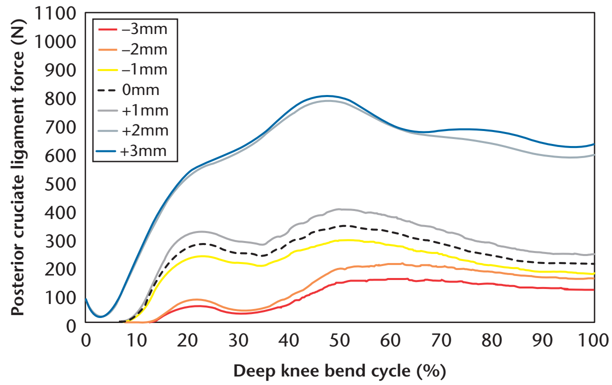


Fig. 5a

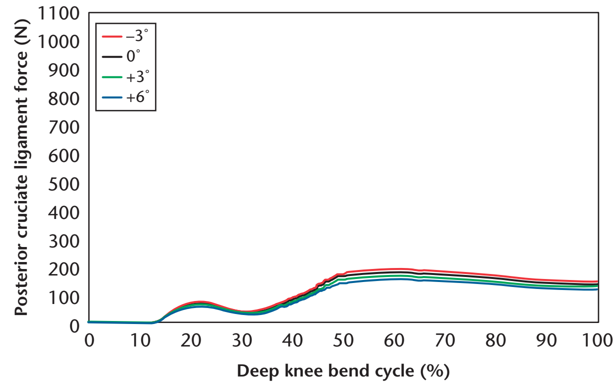


Fig. 5b

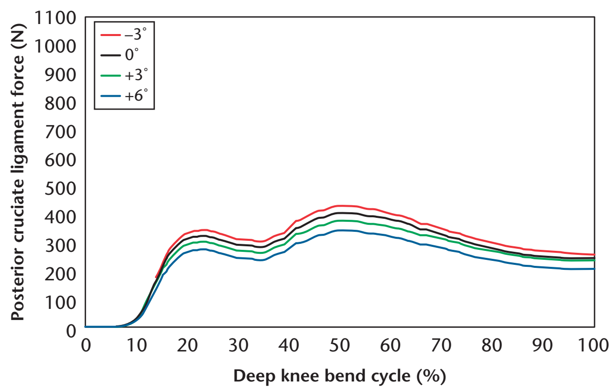


Fig. 5c

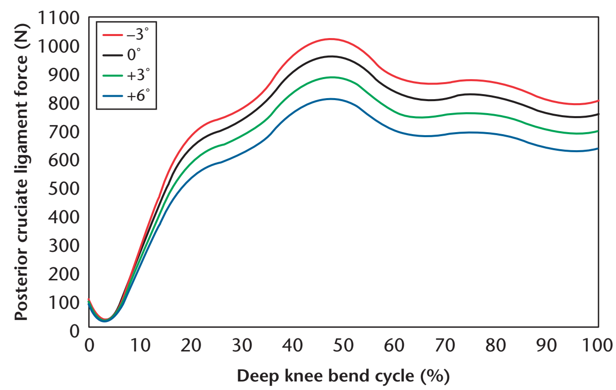


Fig. 5d

Comparison of posterior cruciate ligament force: a) with respect to different posterior condylar offset (PCO) in the posterior tibial slope (PTS) 6°; with respect to different PTS in the b) PCO - 3 mm, c) PCO 0 mm and d) PCO + 3 mm models.

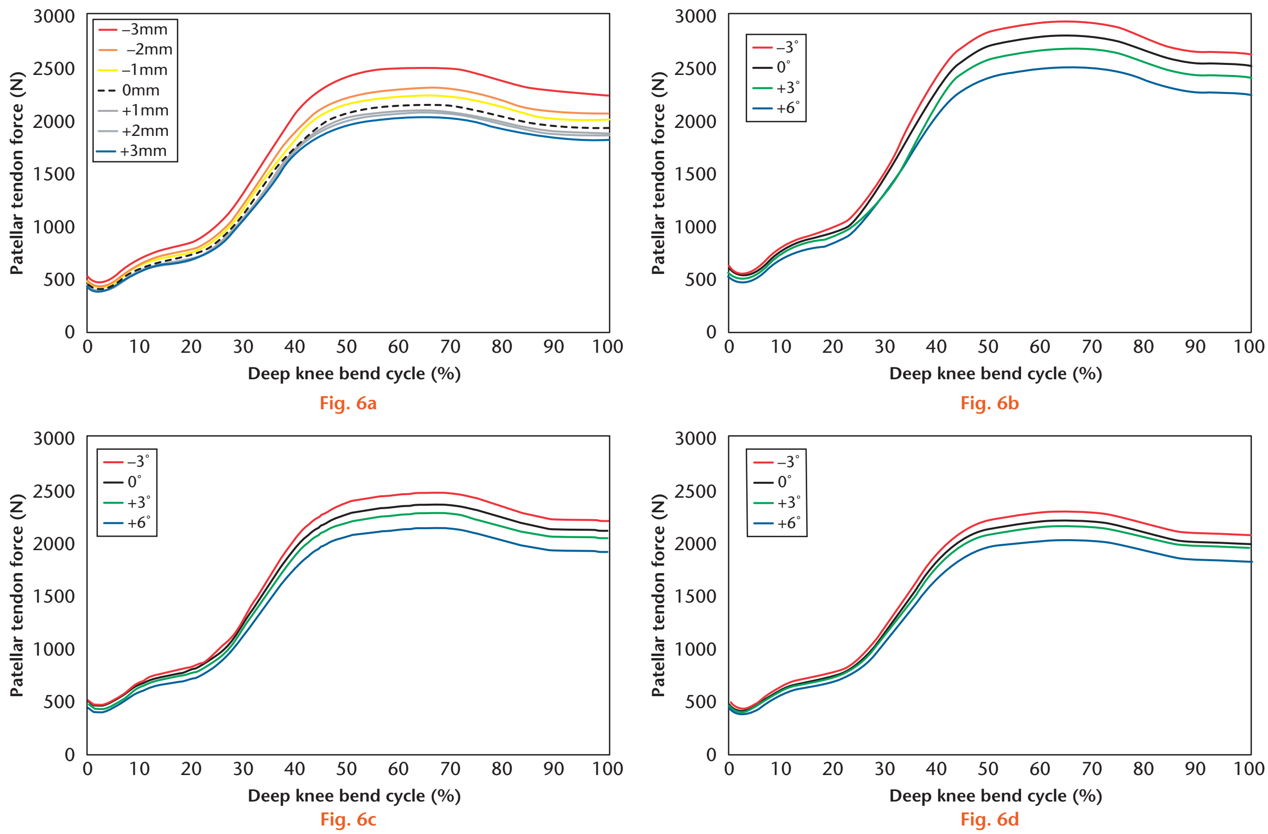
second purpose was to determine if there are critical values that can be applied during surgery.

We hypothesized that the best biomechanical effect could be found in TKA with the subject's own anatomy. To support this hypothesis, we performed 28 simulations with different PCO and PTS combinations.

We found that contact stress on the patellar button increased and decreased as PCO translated to anterior and posterior directions, respectively, compared with the neutral position. Furthermore, the force on the quadriceps muscle and patellar tendon also increased and decreased as PCO translated to anterior and posterior directions, respectively. The force exerted on the PCL increased and decreased as PCO was translated to the posterior and anterior directions, respectively.

An interesting finding was the PCO change when accompanied by a PTS increase. The negative biomechanical effect reduced as PTS increased. The contact stress on the patellar button in all PCO models decreased as PTS increased. This trend was similar to that observed in previous studies[49,50] where the contact stress on the patellar button decreased with an increase in PTS. In addition, a decreased PTS angle increased the PCL tension in all PCO models used in this study. Achieving the appropriate PCL tension is extremely important in CR-TKA. A cadaver study found that the tension in the PCL measured at a 5° posterior slope was significantly greater than at 8° and 10°.[51] Similarly, we showed that a decrease or increase of 3° in the PTS angle decreased or increased the forces exerted on the PCL, respectively, by approximately 25% compared with the PCL force observed in a neutral PTS and the same PCO conditions. There have been no previous reports of the relationship between the PCL force and PCO and PTS. These results suggest that surgical techniques with accurate anatomy and bone preparation are very important to maintain PCL tension in CR-TKA and that the PCL force can be changed by the PCO and PTS without releasing the PCL.

As with the biomechanical factors mentioned previously, the patellar tendon force and quadriceps muscle forces decreased as PTS increased. In the present study, an increase of 6° in the PTS led to a decrease of 34% in the maximum quadriceps muscle and patellar tendon force under the same PCO conditions. Increasing the PTS and PCO led to a more posterior position of the femoral component. A more posterior contact position between the TF components led to an increase in the quadriceps lever arm, which improved the movement efficiency by reducing the quadriceps muscle force and patellar button contact stress.[52,53] Therefore, increasing the PTS and PCO can reduce the quadriceps muscle and patellar tendon force required under deep knee-bend conditions.



Comparison of patellar tendon force: a) with respect to different posterior condylar offset (PCO) in the posterior tibial slope (PTS) 6°; with respect to different PTS in the b) PCO - 3 mm, c) PCO 0 mm and d) PCO + 3 mm models.

Our results showed that the ROM was improved, but it exerted a negative biomechanical influence on the force exerted on the PCL as PCO was translated in the posterior direction. However, there were opposite results for the anterior translation of PCO. Furthermore, all of the biomechanical factors showed positive influences on TKA as PTS increased. Previous studies supported our results.[7,45,52,54] In other words, flexion improved if PCO was translated to the posterior direction and PTS increased,[7,45,55] and the quadriceps muscle and patellar tendon force and contact stress on the patellar button decreased if PTS increased.[49-51,56]

This study showed the importance of preserving the original anatomy of the patient in TKA. Although increased PTS provides good biomechanical effects, it should be moderated because it may cause knee instability.[48] Also, with patients who have degenerative changes of the PCL, the surgeon should be careful not to increase the PCO by more than 2 mm and decrease the PTS. In this case, there is a possibility of affecting the longevity of the implant. Furthermore, the surgeon should be careful not to reduce the PTS, and also be careful to ensure that the PCO does not exceed 2 mm when moving the implant posteriorly to reduce a flexion gap or to resolve anteroposterior-mediolateral mismatches.

There were several limitations to our study. First, there was a virtual and variable model used in this simulation, and the material properties of soft tissues were obtained from relevant cadaveric studies. These are common methods in computational studies.[26,33-35,49,50,54] Second, the geometries of the different structures were only based on one subject without considering any variations in the anatomy. However, this method was commonly used in the biomechanical field.[26,33-35,49,50,54] The intact knee model had undergone a series of rigorous validation steps and the results showed a good accordance with previous experimental or FE studies.[24] In addition, the experimental data from subjects identical to the FE model for cartilage contact area, contact centroid weight-bearing MRI and the TF joint kinematic with laxity test were also used for the validation of the intact knee model.[23,25] The TKA model was also validated using kinematic data from a previous cadaveric study,[40] which showed that they were not perfectly matched but had a similar trend. Therefore, the TKA models used in this study and the following analyses can be considered as reasonable. Third, the balance of all of the collateral ligaments and of the PCL was assumed to be appropriate in our FE model. In fact, it is very important to adjust the PCL balancing in CR-TKA.[57] However, our study evaluated the actual effect of the PCO and PTS as the other factors were

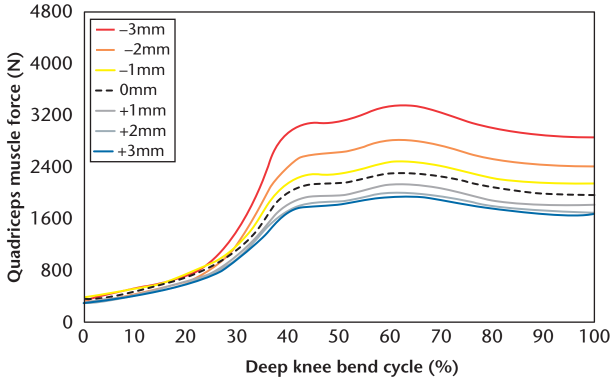


Fig. 7a

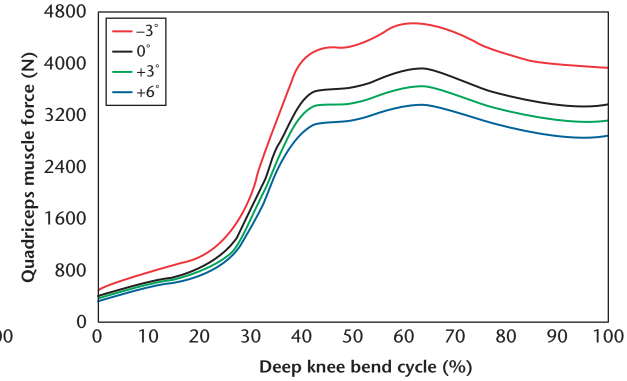


Fig. 7b

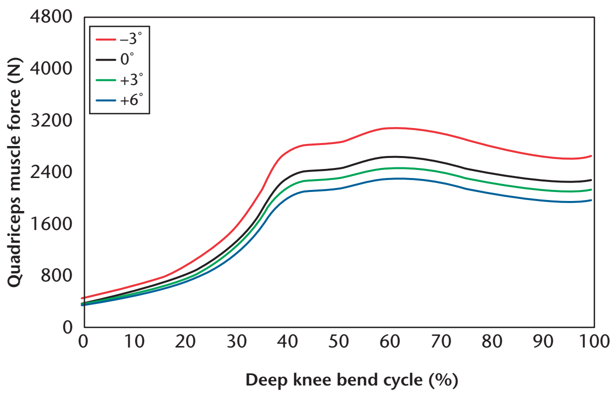


Fig. 7c

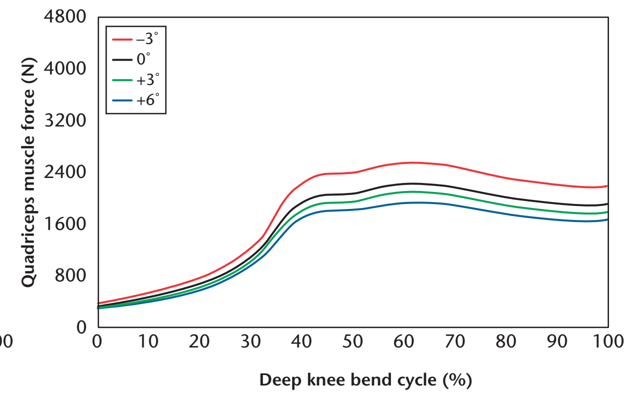


Fig. 7d

Comparison of quadriceps force: a) with respect to different posterior condylar offset (PCO) in the posterior tibial slope (PTS) 6°; with respect to different PTS in the b) PCO - 3 mm, c) PCO 0 mm and d) PCO + 3 mm models.

controlled. Fourth, the results could not substitute for clinical results. Finally, we performed a simulation only for one model fixed-bearing CR-TKA. Therefore, the results from this experiment cannot be considered representative of all fixed-bearing CR-TKAs. Other types of fixed bearings or mobile bearings may provide different results. For instance, the J-curved prosthesis was used in this study but the Scorpio (Stryker Orthopaedics, Mahwah, New Jersey) implant has femoral condyles that have a single radius of curvature, unlike many other CR implant designs that have separate centres of curvature for the medial and lateral condyles in the coronal plane. Therefore, different prostheses and bearing types should be analyzed in the future.

In conclusion, change in PCO and PTS was associated with increase and decrease in the contact stress on the patellar button, forces on the PCL and quadriceps muscle and patellar tendon forces. The change in PCO showed various biomechanical effects, and the efficacy of the prosthesis improved as PTS increased. Our computer simulation results indicate that surgeons should maintain the original anatomy of patients in TKA to the extent possible.

Supplementary material

Supplementary figures showing the rest of the PCO conditions (- 2 mm, - 1 mm, + 1 mm and + 2 mm PCO) that cannot be fully expressed in the manuscript figures are available alongside this article at www.bjr.boneandjoint.org.uk

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- K-T. Kang: Co-first author, Designed the study, Writing the paper, Evaluated the result using finite element analysis.
- Y-G. Koh: Co-first author, Designed the study, Writing the paper, Surgical simulation.
- J. Son: Developed 3D model.
- O-R. Kwon: Data analysis.
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Conflicts of Interest Statement

- None declared