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Impact of intraoperative adjustment method for increased flexion gap on knee kinematics after posterior cruciate ligament-sacrificing total knee arthroplasty

Mutsumi Watanabe[a], Shinichi Kuriyama[a,*], Shinichiro Nakamura[a], Kohei Nishitani[a], Yoshihisa Tanaka[a], Kazuya Sekiguchi[b], Hiromu Ito[a], Shuichi Matsuda[a]

[a] *Department of Orthopaedic Surgery, Graduate School of Medicine, Kyoto University, 54 Kawahara-cho, Shogoin, Sakyo-ku, Kyoto 606-8507, Japan*
[b] *Department of Orthopaedic Surgery, Yawata Central Hospital, 39-1 Gotanda, Yawata-shi, Kyoto, Japan*

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ABSTRACT

Background: In general, the flexion gap is larger than the extension gap with posterior cruciate ligament-sacrificing total knee arthroplasty. Several methods compensate for an excessive flexion gap, but their effects are unknown. The purpose of this study was to compare three methods to compensate for an increased flexion gap.
Methods: In this study, squatting in knees with excessive (4 mm) and moderate (2 mm) flexion gaps was simulated in a computer model. Differences in knee kinematics and kinetics with joint line elevation, setting the femoral component in flexion, and using a larger femoral component as compensatory methods were investigated.
Findings: The rotational kinematics during flexion with setting the femoral component in flexion were opposite to those in the other models. Using a larger femoral component resulted in the most physiological motion. The peak anterior translation was 10 mm in the joint line elevation model compared with approximately 6 mm in the other models. In the joint line elevation model, patellofemoral contact stress was excessively increased at 90° of knee flexion. In contrast, tibiofemoral contact stress was higher during knee extension with setting the femoral component in flexion due to anterior impingement. There were few differences in the effect of the three compensatory methods with a moderate flexion gap.
Interpretation: A larger femoral component should be used to compensate for an excessive flexion gap because it has less negative impact on posterior cruciate ligament-sacrificing total knee arthroplasty, whereas any compensation method might be acceptable for a moderate flexion gap.

1. Introduction

Achieving suitable intraoperative ligament balance is considered important for acquiring a good postoperative outcome for total knee arthroplasty (TKA) (Arima et al., 1998; Tanzer et al., 2002; Watanabe et al., 2015; Yercan et al., 2005a, 2005b). Adjustment of the extension-flexion gap is generally made by releasing ligamentous attachments and resecting bones. However, it is sometimes difficult to acquire an equal flexion-extension gap, especially in flexion during posterior cruciate ligament (PCL)-sacrificing (PS)-TKA; the gap during flexion is generally larger than during extension because the PCL was resected (Dorr and Boiardo, 1986; Kadoya et al., 2001; Kaneyama et al., 2014; Nowakowski et al., 2012; Yagishita et al., 2003).

Three methods to compensate for an increased flexion gap with PS-TKA have been attempted. In the first method, the amount of distal femur re-cutting corresponds to only the amount of the increased flexion gap, and a thicker tibial insert is adjusted. As a result, joint line elevation (JLE) occurs. Second, setting the femoral component to the flexion position in the sagittal plane is probably most commonly used method to compensate for discrepancies between extension and flexion gaps (Tsukeoka et al., 2013; Tsukeoka and Lee, 2012). Finally, using a femoral component with a larger anterior-posterior (AP) dimension is another possible adjustment method.

Previous studies have reported that JLE adversely affects function, resulting in mid-flexion instability and increased intercomponent contact forces (Fornalski et al., 2012; Konig et al., 2010; Martin and Whiteside, 1990). Setting the femoral component in flexion causes anterior impingement of the post-cam mechanism and patellar

* Corresponding author.
E-mail address: kuriyama@kuhp.kyoto-u.ac.jp (S. Kuriyama).

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instability (Keshmiri et al., 2016; Nedopil et al., 2017). Increasing the size of the femoral component sometimes causes medial-lateral (ML) overhang that results in postoperative knee pain (Mahoney and Kinsey, 2010), but narrower designs for the femoral component have recently become available. However, it is still unresolved which intraoperative method is the better solution for an increased flexion gap.

The purpose of this study was to compare the effects of selecting JLE, setting the femoral component in flexion, and increasing the size of the femoral component in PS-TKA. We hypothesized that these compensatory methods need to be changed according to how much the flexion gap increases while considering postoperative kinematics and kinetics.

2. Methods

2.1. Musculoskeletal computational model

A dynamic musculoskeletal modeling program for the knee (LifeMOD/Knee SIM 2010; LifeModeler Inc., San Clemente, CA, USA) was used to simulate two cycles of squatting (0°-130°-0° of knee flexion) (Kuriyama et al., 2014b; Kuriyama et al., 2015; Mizu-Uchi et al., 2015; Watanabe et al., 2017; Zumbrunn et al., 2016). This program modeled tibiofemoral and patellofemoral contact, medial collateral ligament (MCL), lateral collateral ligament (LCL), elements of the knee capsule, quadriceps muscles and tendons, patellar tendon, and hamstring muscles. The geometry of the femur, tibia, fibula, and patella, which was reconstructed from computed tomography images of a healthy volunteer with neutral coronal alignment, was introduced into KneeSIM. The mean difference between in vivo conditions based on fluoroscopic images and simulated kinematics during squatting were 1.7 mm for the AP position and 2.5° for axial rotation. The LCL included a single bundle and the MCL included anterior and posterior bundles (Sugita and Amis, 2001; Warren et al., 1974). All ligament bundles were modeled as nonlinear springs with properties based on a published report (Blankevoort et al., 1991). The insertion point of each ligament was adjusted according to findings from a previous study (LaPrade et al., 2005). The stiffness and length of each ligament bundle were determined based on a previous study (LaPrade et al., 2005). The most prominent points of the femoral epicondyles were the proximal attachment points for the LCL and MCL. The distal attachment points of the LCL and MCL were defined as the tip of the fibular head and the midpoint between the tibial attachments of the anterior and posterior bundles, respectively (Belvedere et al., 2012; Liu et al., 2011; Park et al., 2005) (Fig. 1). The stiffness coefficients for the LCL, anterior MCL bundle, and posterior MCL bundle were determined to be 59, 63, and 63 N/mm respectively, based on previous investigations (Robinson et al., 2005; Sugita and Amis, 2001). During squatting, the hip joint was modeled as a revolute joint parallel to the knee's axis of flexion that could slide vertically. The ankle joint was modeled as a combination of several joints that jointly allowed free translation in the ML direction and free rotation in the axial and varus-valgus directions as well as in the direction of flexion.

Parasolid models of a knee after fixed-bearing PS-TKA (NexGen LPS-Flex; Zimmer Biomet, Warsaw, IN, USA) were imported into the KneeSIM program. The size of the femoral component and the thickness of the tibial insert can be changed for each condition. In the coronal plane, the femoral component was aligned to the mechanical axis of the femur from the center of the knee to the center of the femoral head, and the tibial component was aligned to the mechanical axis of the tibia from the center of the knee to the center of the ankle joint. In the sagittal plane, the femoral component was aligned to the distal anatomical axis of the femur and the tibial component was aligned to the proximal anatomical axis of the tibia with a posterior slope of 7°. The rotation alignment of the femoral and tibial components was set to be in line with the femoral epicondylar axis and the tibial AP axis, respectively. Previous studies have shown that the peak tibiofemoral contact force in patients after TKA is approximately 2–6 times their body weight during squatting (D'Lima et al., 2007; Innocenti et al., 2011; Nagura et al., 2006; Smith et al., 2008). The model parameters were set so that a constant vertical force was converted into approximately 4000 N during squat loading on the bicondylar knee joint, which corresponds to approximately five times 80 kg of body weight.

2.1.1. Model of the normal PS-TKA

At first, a model of the normal PS-TKA was developed based on general TKA placement (Fig. 2). The level of distal femoral resection was set to the sulcus cut line, which is the deepest portion of the femoral trochlear groove. This technique usually maintains the lateral joint line (Kuriyama et al., 2014a). The proximal position of the tibial component was set so that the lateral joint surface will be maintained. The posterior femoral condyle of the femoral component was aligned to the surgical epicondylar axis, and the bone resection thickness was usually greater on the medial condyle than on the lateral condyle (Minoda et al., 2016). In addition, the AP positioning and sizing of the femoral component was carefully selected so that anterior femoral notching was minimized as much as possible; this femoral component size was defined as the standard femoral component size. In this study, two conditions with an increased flexion gap that were investigated: an excessive flexion gap (flexion gap that is 4 mm wider than the extension gap) and moderate flexion gap (2 mm wider). In these two conditions, we compared knee kinematics and kinetics of models with JLE, setting the femoral component in flexion, and increasing the size of the femoral component.

2.1.2. Modeling the excessive flexion gap

The model of an excessive flexion gap was defined as a femoral component being positioned 4 mm anterior to the suitable positioning of the normal TKA with no changes in the position of the joint line. Surgeons might set the position of the femoral component more anteriorly to avoid intraoperative anterior femoral notching. In this situation, the posterior condylar offset is 4 mm smaller in the model with an excessive flexion gap than in the normal model (Fig. 2). In order to compensate for the 4 mm flexion gap, three models were developed: 4 mm of JLE, femoral component in 7° of flexion, and larger femoral component.

2.1.3. Compensatory model with a 4 mm JLE

The standard femoral component was set to be 4 mm more proximal and a tibial insert that was 4 mm thicker was used (Fig. 2). The ligamentous attachments and lengths in this model were not changed from initial setting.

2.1.4. Compensatory model with the femoral component set in 7° of flexion

The standard femoral component was set to 7° of flexion in the sagittal plane to compensate for a 4 mm flexion gap without changing the joint line level and the AP position of the anterior flange of the femoral component (Fig. 2).

2.1.5. Compensatory model with a larger femoral component

A femoral component that is 4 mm larger in the AP dimension than the standard femoral component was used (Fig. 2).

2.1.6. Models to compensate for a moderate flexion gap

When the standard femoral component was set to be 2 mm posterior to the setting with an excessive flexion gap, the difference between the extension and flexion gaps was 2 mm. This constituted a model of a moderate flexion gap. In order to compensate for a flexion gap of 2 mm, three models were developed as follows: a model with 2 mm of JLE in which the femoral component was set to be 2 mm more proximal and 2 mm thicker tibial insert was used, a model with the femoral component set in 3° of flexion in the sagittal plane, and a model with a larger femoral component identical to the method for the excessive flexion

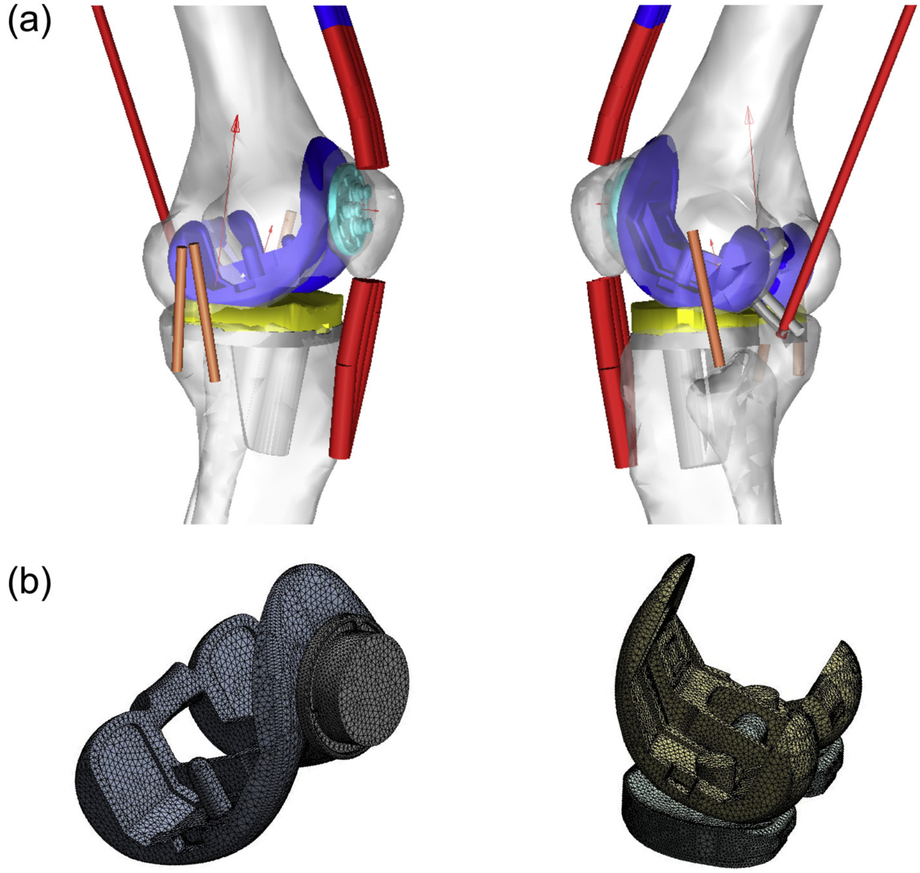

Fig. 1. Insertion point of each soft tissue and force vectors at the tibiofemoral and patellofemoral joints using a model with an excessive flexion gap from the KneeSIM program (a), and finite element analysis models of the patellofemoral and tibiofemoral joints including the post-cam mechanism from the ANSYS program (b).

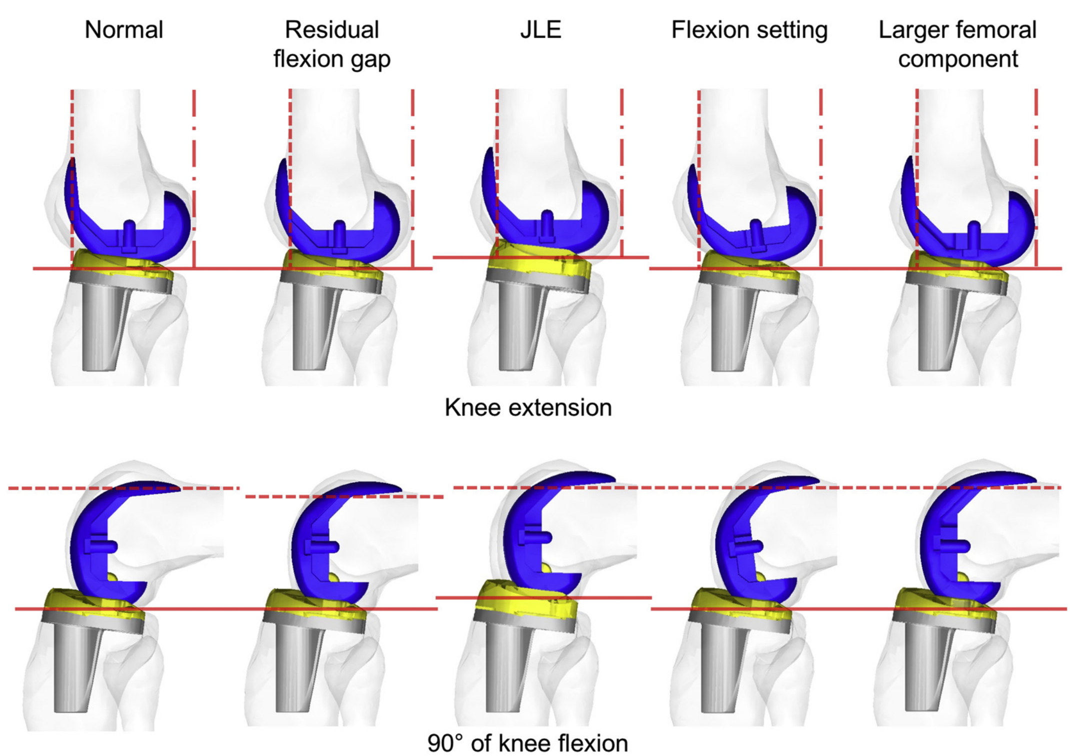

Fig. 2. Lateral view of each model, including the normal posterior cruciate ligament-sacrificing total knee arthroplasty during knee extension and at 90° of knee flexion. Solid horizontal lines indicate the tibiofemoral joint line. Vertical lines indicate the tangent line of the femoral posterior condyle. Dotted lines indicate the tangent line of the anterior cortex of the distal femur. JLE, joint line elevation.

gap mentioned above.

2.2. Measurement methods

This study examined the angle between the line connecting the lowest points of the medial and lateral femoral condyles and the tangent to the posterior medial and lateral condyles of the tibial insert. A positive value was defined as external rotation of the femur against the tibial insert when the lowest lateral point was located more posteriorly than the lowest medial point. The amount of AP movement at the lowest points of the medial and lateral femoral condyles or the center of the femoral component was also examined. A negative value corresponded to the femur being posterior to the center of the tibial insert. In addition, tibiofemoral and patellofemoral contact forces and patellar kinematics were measured. Only analyses of the lowest points were examined up to 110° knee flexion, because the lowest point was moved to the post-cam structure starting at 116° of knee flexion based on the femoral component design. Except for the analyses of the lowest point, values from 0° to 120° of knee flexion were used to avoid the influence of changes in knee motion while going from knee extension to flexion.

2.3. Finite element analysis

Contact stress at the tibial inserts against the femoral component was calculated with three-dimensional finite element (FE) analysis using ANSYS Workbench version 14.5 (ANSYS, Inc., Canonsburg, PA, USA). We aligned the position of each component using solid modeling software (Solid Edge, Siemens AG, Munich, Germany) based on the intercomponent positions calculated using KneeSIM. The magnitude and direction of each force calculated using KneeSIM (Fig. 1) were also used to predict contact stress (Ishikawa et al., 2015; Tanaka et al., 2016; Tanaka et al., 2018). The stress state at the component interfaces was examined at 0°, 30°, 60°, 90°, and 110° of knee flexion. The elastic modulus of the femoral component (Co-Cr-Mo alloy) was 240 GPa. The tibial insert and patellar component were modeled as nonlinear elastoplastic materials. We imported a true stress versus strain curve for GUR 4150 HP, which was based on a previous study (Kurtz et al., 1999), and a Poisson's ratio of 0.46 into the FE analysis. Convergence was defined as a relative change of < 5% from the maximum stress provided by the analysis. Meshes of the femoral component, tibial insert, and patellar component (Fig. 1) were generated based on a 10-node quadratic tetrahedral element of 0.8 mm (Tanaka et al., 2016). Using the ‘contacts’ tool in ANSYS software, the ‘no separation’ contact type was applied. Cylindrical support was used for static analysis, and the load obtained from the musculoskeletal model was applied to the cylinder supporting the femoral and patellar components in the tibiofemoral and patellofemoral joints, respectively. The fixed support on the undersurface of the tibial insert in the tibiofemoral joint was maintained. A fixed support was also applied to the back of the contact surface of the femoral component of the patellofemoral joint. Contact was considered to occur when the perpendicular distance between the surfaces of the femoral component, tibial insert, and patellar component was < 0.3 mm. Soft tissue was not included in the analysis. The measurements in this study are given to one decimal place because our simulation system could calculate each value with at least this degree of accuracy.

3. Results

3.1. Normal PS-TKA model

In the model of the normal PS-TKA, the line connecting the medial and lateral lowest femoral points moved up to 2.6° externally against the tibial insert at 110° of knee flexion (Fig. 3). The peak amount of anterior movement of the femoral component was as small as 2.1 mm (Fig. 4). During knee extension and 90° of knee flexion, the peak lateral tibiofemoral contact stresses were 7.9 MPa and 7.2 MPa, and the peak medial tibiofemoral contact stresses were 5.2 MPa and 8.0 MPa, respectively (Figs. 5, 6). The peak patella contact stress at 90° of knee flexion was 80 MPa (Fig. 7). ML patellar translation based on the femoral component was 8.4 mm from the lateral to the medial side during knee flexion (Fig. 8).

3.2. Compensation for an excessive flexion gap

In the model with an excessive flexion gap, there was less rotational movement of the tibiofemoral joint compared with the normal PS-TKA model. Bicondylar rollback almost occurred (Fig. 3). The compensatory model with a 4 mm JLE had similar rotational movement as the uncompensated model. However, the model with the femoral component in 7° of flexion had opposite internal movement from 30° to 90° of knee flexion. In contrast, the model with a larger femoral component had more external movement, up to 3.0°, at 110° of knee flexion. A relatively large amount of anterior movement of the medial and lateral lowest femoral points against the tibial insert occurred in all models with an excessive flexion gap (Fig. 3). Regarding kinematics of the center of the femoral component, anterior paradoxical rollbacks in all models occurred until approximately 100° of knee flexion with engagement of the post-cam mechanism (Fig. 4). The peak amount of anterior movement was 6.8, 10.2, 5.8, and 6.1 mm in models with a remaining excessive gap, 4 mm of JLE, a 7° flexion setting, and larger femoral component, respectively. There was slightly less anterior movement in the 7° flexion setting and larger femoral component models, but the model with 4 mm of JLE had twice as much anterior movement as the other models (Fig. 4).

Regarding the contact force on the anterior tibial post against the intercondylar box of the femoral component during knee extension, the model with the femoral component set in 7° of flexion had the highest value (742 N), while the model with a larger femoral component had the lowest value (289 N) (Table 1). Moreover, tibiofemoral contact stresses during knee extension were also the highest in the model with the 7° flexion setting because the femoral component was anterior to the tibial insert, while peak tibiofemoral contact forces were variable in each model (Fig. 5, Tables 1, 2). On the other hand, the model with the larger femoral component had the lowest tibiofemoral contact stresses during knee extension and 90° of flexion (Figs. 5, 6), and less quadriceps muscle force was required (Table 1). The patellofemoral joint contact force was also variable in each model (Table 1). However, during 90° of knee flexion, the contact stress of the patellofemoral joint increased markedly in the model with 4 mm of JLE (Fig. 7, Table 2). The patella with the femoral component set in 7° of flexion had the least contact stress and was located more laterally during knee flexion compared with other models, including the normal PS-TKA model (Figs. 7, 8).

3.3. Compensation for a moderate flexion gap

In the model of a moderate flexion gap, the line connecting the medial and lateral lowest femoral points moved more externally against the tibial insert during flexion than in models with an excessive flexion gap (Fig. 9). In the three compensatory models, trends in rotational movement were similar to those in the uncompensated model. The amount of anterior movement was 4.8, 4.7, 4.1, and 6.1 mm in the model with a moderate gap remaining, 2 mm of JLE, 3° flexion setting, and larger femoral component, respectively (Fig. 4). Although the contact force at anterior post of the tibial insert increased in the model with the femoral component set in 3° of flexion, it was lower than in the model with a 7° flexion setting (Table 1). No differences in contact stress at the tibiofemoral and patellofemoral joints were observed in the three models (Figs. 5–7, Table 2). There was less lateral shifting of the patella in the model with a 3° flexion setting compared with the model with a 7° flexion setting (Fig. 8).

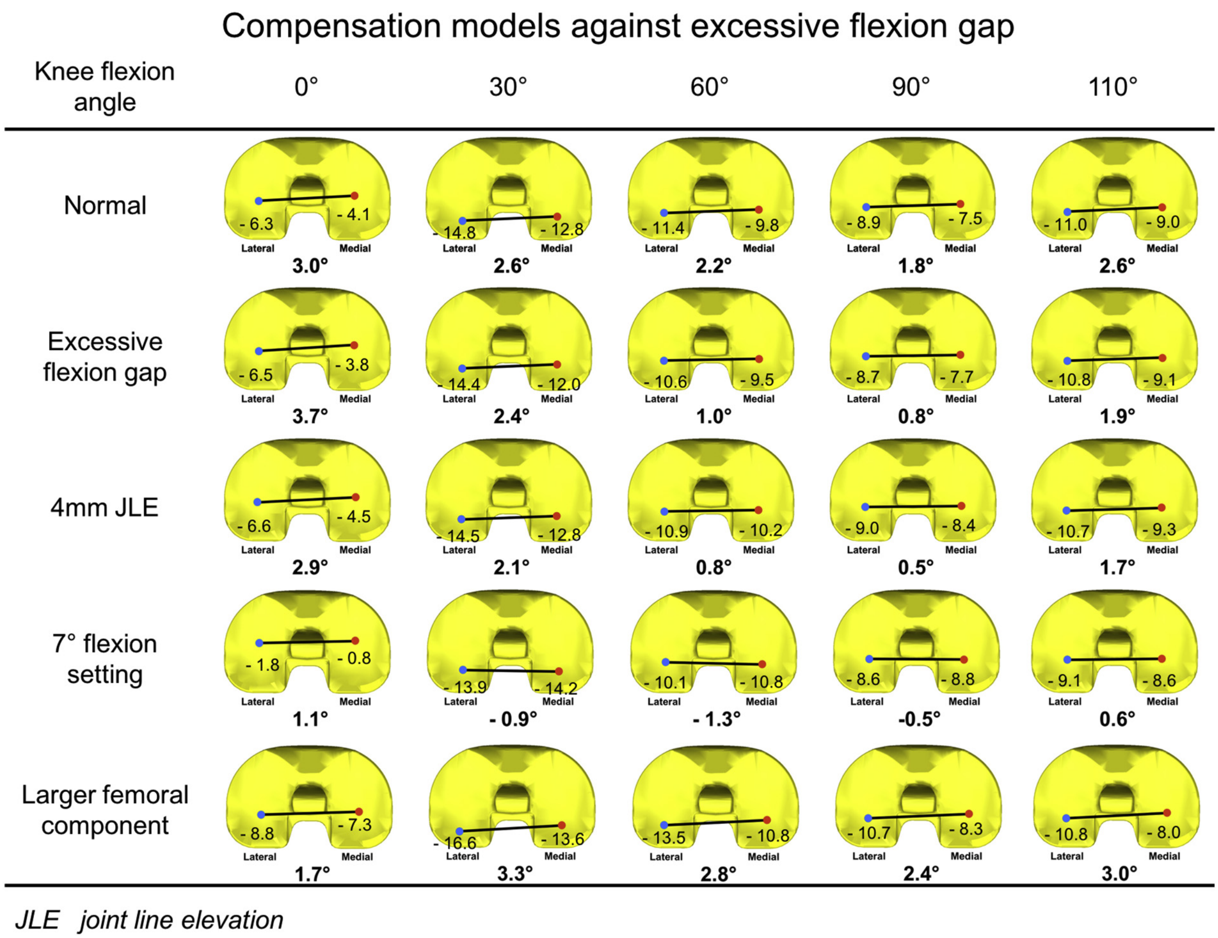

Fig. 3. Lowest points of the medial and lateral femoral condyles relative to the tibial insert at 0°, 30°, 60°, 90°, and 110° of knee flexion in the normal posterior cruciate ligament-sacrificing total knee arthroplasty model and models that compensate for an excessive flexion gap. Blue and red dots show lowest points of lateral and medial femoral condyles, respectively. The external rotation angle (°) of the line connecting the two lowest points with respect to the posterior line of the tibial insert is positive. The number below each lowest point indicates the anteroposterior translation (mm) from the center of the tibial insert. If the lowest point of the femoral component is located behind the center of the tibia, the value is negative. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

4. Discussion

In general, the flexion gap increases with PCL resection and becomes wider than the extension gap as a result of PS-TKA (Dorr and Boiardo, 1986; Kadoya et al., 2001; Kaneyama et al., 2014; Matsumoto et al., 2014; Nowakowski et al., 2012; Yagishita et al., 2003). An increased flexion gap could cause less favorable postoperative knee kinematics (Arima et al., 1998; Minoda et al., 2015; Tanzer et al., 2002; Watanabe et al., 2015; Yercan et al., 2005a, 2005b). A decreased posterior tibial slope might reduce the negative effect of the increased flexion gap (Lombardi et al., 2008; Okazaki et al., 2014). However, decreasing the posterior tibial slope is technically difficult, and errors in re-cutting the proximal tibia simultaneously affect the flexion gap and the extension gap. Hence, this study focused on compensating for an increased flexion gap during PS-TKA by changing femoral component settings. A major strength of this study was to compare the influence on knee kinematics and kinetics of three compensatory methods.

Previous studies have demonstrated problems with JLE, setting the femoral component in flexion, or increasing the size of the femoral component to address the increased flexion gap in TKA (Fornalski et al., 2012; Keshmiri et al., 2016; Konig et al., 2010; Mahoney and Kinsey, 2010; Martin and Whiteside, 1990; Nedopil et al., 2017). Additional resection of the distal femur and using a thicker tibial insert is a simple method that induces JLE. However, it might cause mid-flexion instability and patellofemoral joint problems (Kazemi et al., 2011; Partington et al., 1999; Porteous et al., 2008). Previous studies have reported that 8 mm of JLE affected the patellofemoral joint but there were no significant effects at 4 mm of JLE (Fornalski et al., 2012; Martin and Whiteside, 1990). In contrast, our study revealed that 4 mm of JLE also caused increased tibiofemoral and patellofemoral joint contact stress at 90° of knee flexion due to harmful anterior movement during mid-flexion, while femoral rotational kinematics relative to the tibial insert were not affected. Increased contact stress on the tibiofemoral and patellofemoral joints can be explained by paradoxical anterior movement. Appropriate femoral rollback relative to the tibial insert during flexion contributes to decreased patellofemoral loading and makes the posterior tibial slope effective in reducing tibiofemoral contact stress. Precise analysis using our computer simulation might reveal the critical effects of 4 mm of JLE, which were not detectable in previous studies.

Setting the femoral component to the flexion position in the sagittal plane might be the most common maneuver. It has the advantage of being capable to compensate for the extension-flexion gap difference without changing knee joint line (Matziolis et al., 2012; Tsukeoka et al., 2013). Nevertheless, several studies have warned about the problems of setting the femoral component in flexion, such as anterior impingement of the post-cam mechanism and patellar instability during knee extension (Keshmiri et al., 2016; Nedopil et al., 2017). In this study, setting the femoral component in 7° of flexion decreased paradoxical anterior movement. In contrast, the flexion setting caused increased

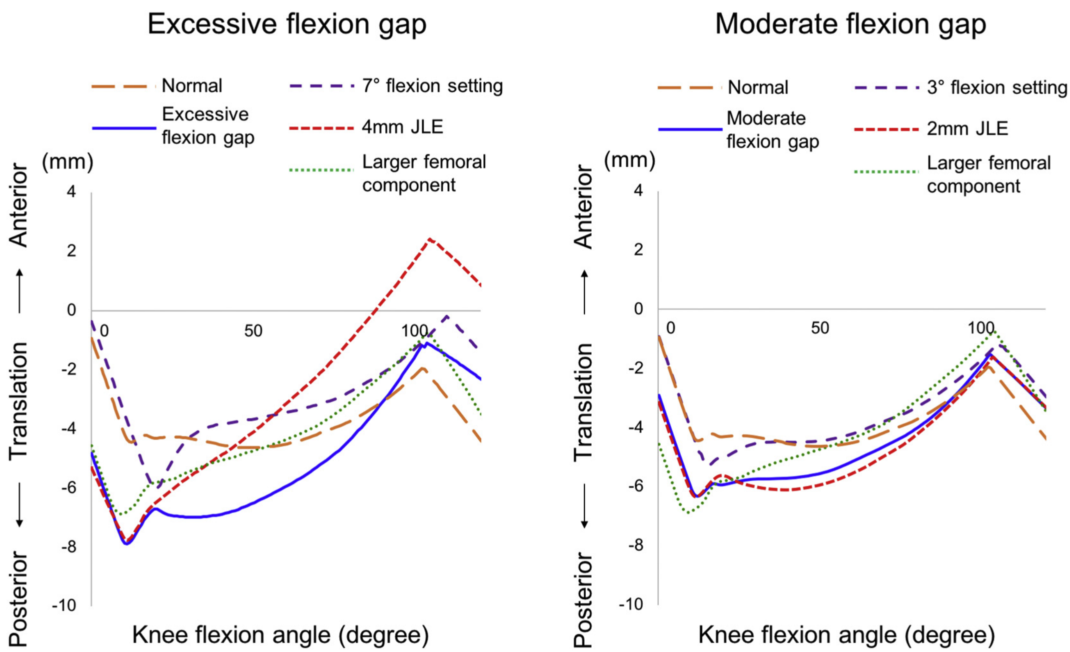

Fig. 4. Anterior-posterior translation (mm) of the center of the femoral component relative to the center of the tibial insert during knee flexion in the normal posterior cruciate ligament-sacrificing total knee arthroplasty model and each compensatory model. The center of the femoral component being located behind the center of the tibia is represented by a negative value.

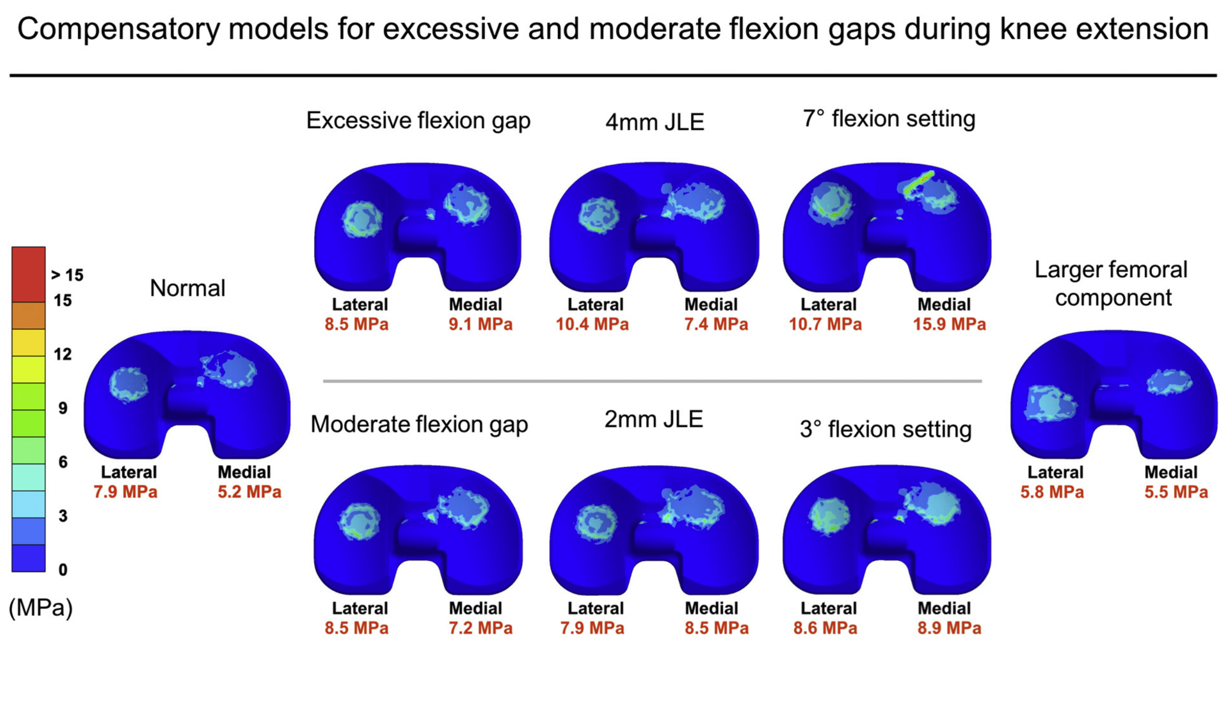

Fig. 5. Tibiofemoral contact stress (MPa) during knee extension based on finite element analysis. The numbers below each figure indicate peak contact stresses on the lateral and medial tibiofemoral joint.

tibiofemoral contact stress during knee extension due to anterior impingement. In the flexion setting, after early anterior impingement between the intercondylar box of the femoral component and the tibial post, the femoral component pushed out more anteriorly against the tibial insert during knee extension. Therefore, increased contact stress between the femoral component and the anterior lip of the tibial insert occurred. In addition, rotational kinematics in the opposite direction and lateral shifting of the patella were observed in this study. We suppose that opposing tibiofemoral kinematics and unstable patellofemoral joints are caused by changes in the geometry of the femoral component due to an excessive flexion setting.

This study demonstrated that adjusting the excessive flexion gap by increasing the size of the femoral component is appropriate in PS-TKA because physiological external rotation of the femoral component was maintained and tibiofemoral and patellar femoral contact stresses were relatively stable. Moreover, less quadriceps muscle force was required

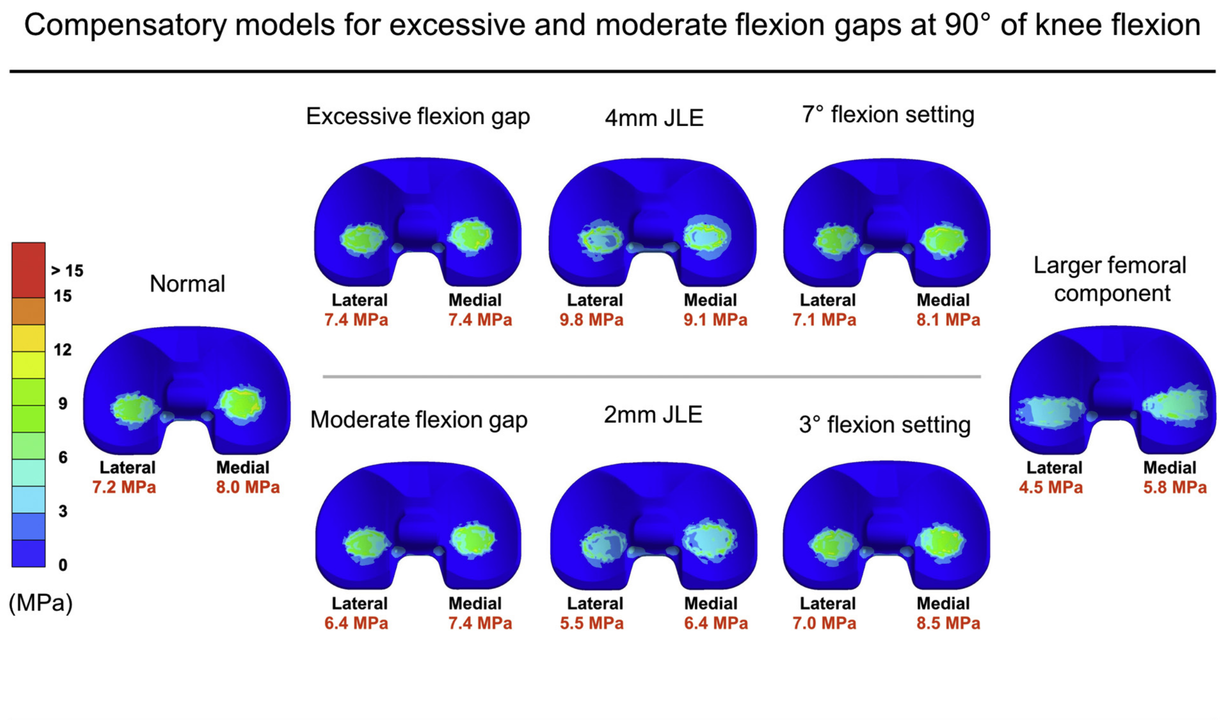

Fig. 6. Tibiofemoral contact stress (MPa) at 90° of knee flexion based on finite element analysis. The numbers below each figure indicate peak contact stresses on the lateral and medial tibiofemoral joint.

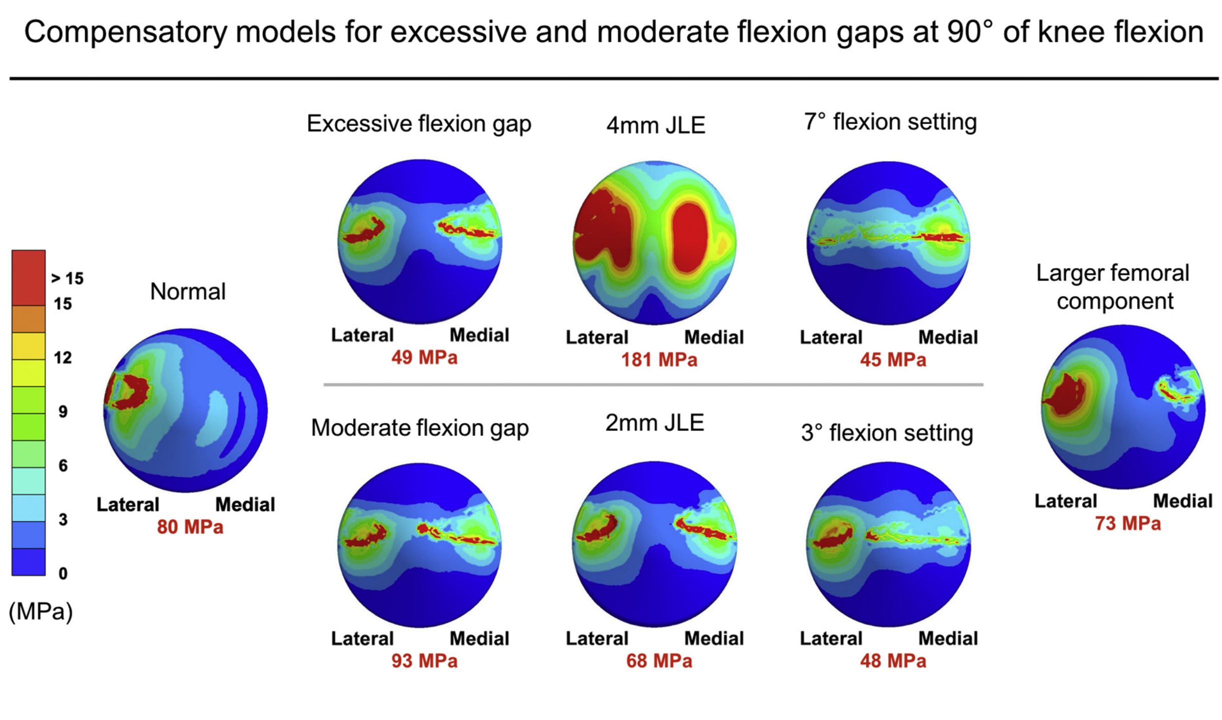

Fig. 7. Patellofemoral contact stress (MPa) at 90° of knee flexion based on finite element analysis.

as the size of the femoral component increased, suggesting increased efficiency of the extensor mechanism. Using the modified gap technique, resection of the proximal tibia and distal femur is performed first. The extension-flexion gap is measured after PCL resection. If there is an excessive flexion gap, a femoral component that is larger by the size of the flexion gap should be used. Some commercially available femoral components have many AP and ML size variations to prevent increased patellofemoral contact force and ML overhang (Clarke and Hentz, 2008; Greene, 2007; Kawahara et al., 2012). Using the measured resection technique in PS-TKA, the femoral component size can be decided based on the posterior reference technique. Surgeons should take into account the 2–4 mm increase in the flexion gap during PCL resection (Kadoya et al., 2001; Kaneyama et al., 2014; Nowakowski et al., 2012; Yagishita et al., 2003).

If there is a moderate flexion gap, this study found that there is a small difference in the impact on knee kinematics and kinetics between

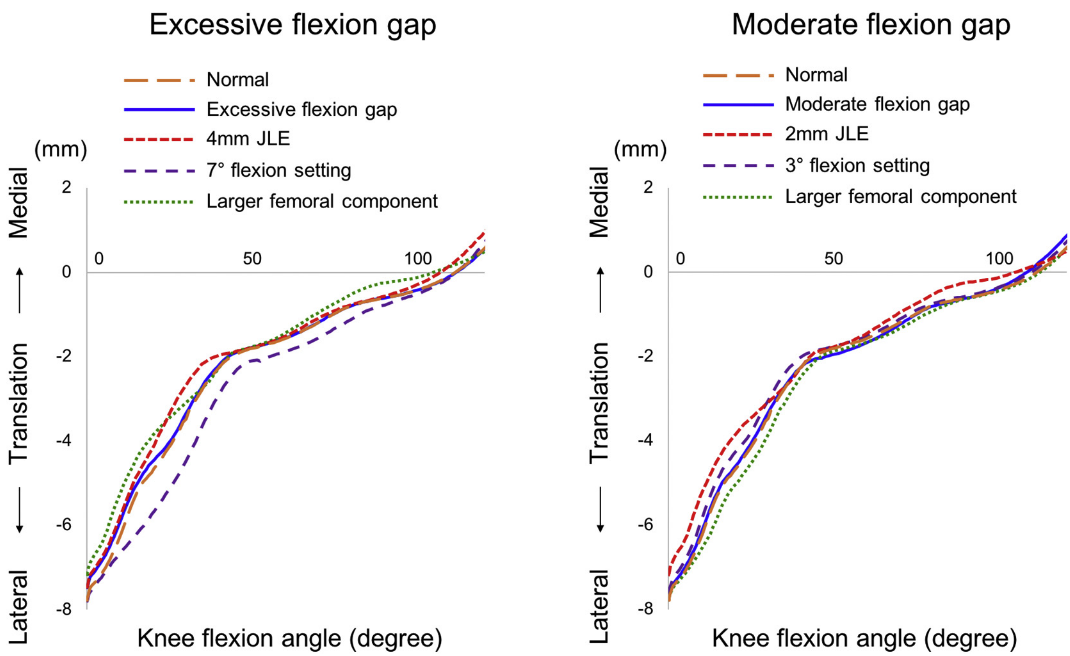

Fig. 8. Medial-lateral patellar translation (mm) based on the femoral component in the normal posterior cruciate ligament-sacrificing total knee arthroplasty model and each compensatory model. Lateral translation of the patella relative to the center of the femoral component is represented by a negative value.

Table 1
Peak tibiofemoral, anterior tibial post, and patellofemoral contact forces (N) at each knee flexion angle and maximum quadriceps muscle force (N) in the normal posterior cruciate ligament-sacrificing total knee arthroplasty model and each compensatory model.

	Flexion angle	Normal	Excessive gap	Moderate gap	4 mm JLE	2 mm JLE	7° flexion	3° flexion	Larger component
TF joint (medial)	0°	628	635	678	624	632	703	696	615
	30°	703	581	686	598	573	632	685	632
	60°	947	737	876	707	764	862	1042	789
	90°	941	783	762	896	851	840	874	922
	110°	375	405	436	538	424	845	458	410
TF joint (lateral)	0°	691	780	737	683	795	772	762	645
	30°	243	246	208	202	335	117	190	298
	60°	362	361	295	383	422	274	185	518
	90°	569	626	644	516	569	560	583	546
	110°	1267	1228	1205	1079	1228	742	1130	1213
Anterior tibial post	0°	464	449	474	417	499	742	607	289
PF joint	0°	64	86	75	81	76	131	93	71
	30°	174	207	189	199	190	188	180	170
	60°	659	709	679	697	684	667	646	635
	90°	1459	1459	1449	1372	1434	1340	1417	1254
	110°	1988	1999	1990	1899	1969	1971	2007	1775
Quadriceps muscle force		1913	1916	1965	1903	1923	1991	1940	1771

JLE, joint line elevation; TF, tibiofemoral; PF, patellofemoral.

the 3 compensatory methods. Therefore, the clinical relevance of this study is that a larger femoral component should be used to compensate for an excessive flexion gap > 2 mm, but it is possible to compensate for a moderate flexion gap ≤ 2 mm with additional resection of the distal femur and a thicker tibial insert or setting the femoral component in flexion.

There are several limitations to this study. First, this simulation comprised a virtual model with a generic knee joint. Whether the kinematics on an individual knee can be replicated by components designed for the average population is unclear. However, the simulation can use the same conditions, accurately adjust for several conditions, and reveal the effects without the influence of unintentional modifications. Next, since this study included only one implant design and bone geometry, the results of this study are not necessarily applicable to other designs such as the 'Gender Solution' in the NexGen series and other knees. Third, squatting with the Oxford knee rig used in the study is not physiological, whereas a previous study validated the difference in knee kinematics during squatting from 0° to 120° between in vivo and KneeSIM data using a 2D-3D registration approach (Tanaka et al., 2016). Fourth, the FE analysis was not a dynamic analysis. It was a quasi-static analysis, and the contact stress values calculated from this study might not necessarily reflect numerical values in vivo because of the lack of validation for the FE analysis in this study. Further simulation studies using various subjects and designs should be conducted in the future to evaluate inter-subject variations in knee kinematics and kinetics.

5. Conclusion

A larger femoral component should be used to compensate for an excessive flexion gap of approximately 4 mm because it has less negative impact on postoperative knee kinematics and kinetics with PS-TKA. On the other hand, any compensatory method might be acceptable for a moderate flexion gap of ≤ 2 mm.

Table 2
Maximum contact stress (MPa) at the tibiofemoral and patellofemoral joints in the normal posterior cruciate ligament-sacrificing total knee arthroplasty model and each compensatory model at 0° 30°, 60°, 90°, and 110° of knee flexion.

	Flexion angle	Normal	Excessive gap	Moderate gap	4 mm JLE	2 mm JLE	7° flexion	3° flexion	Larger component
TF joint (medial)	0°	5.2	9.1	7.2	7.4	8.5	15.9	8.9	5.5
	30°	7.2	5.0	6.1	6.6	5.7	5.7	4.8	4.6
	60°	10.4	6.1	7.1	7.8	7.7	7.7	8.1	4.9
	90°	8.0	7.4	7.4	9.1	6.4	8.1	8.5	5.8
	110°	7.8	5.6	5.9	8.8	7.8	7.6	6.3	5.0
TF joint (lateral)	0°	7.9	8.5	8.5	10.4	7.9	10.7	8.6	5.8
	30°	5.3	5.1	4.6	6.5	7.3	2.9	5.0	3.8
	60°	6.4	6.0	5.7	6.3	8.3	4.7	5.9	4.4
	90°	7.2	7.4	6.4	9.8	5.5	7.1	7.0	4.5
	110°	14.0	10.8	10.4	10.9	12.8	8.5	10.4	9.2
PF joint	0°	2.7	2.5	2.3	3.0	2.4	4.0	3.1	2.4
	30°	4.9	5.1	6.5	5.0	4.3	16.8	4.1	4.3
	60°	13.3	59.5	44.4	57.9	36.6	33.0	49.3	36.9
	90°	79.9	49.4	93.0	181.4	68.2	45.5	48.3	73.0
	110°	98.6	167.8	202.4	99.6	252.6	83.6	255.8	64.0

JLE, joint line elevation; TF, tibiofemoral; PF, patellofemoral.

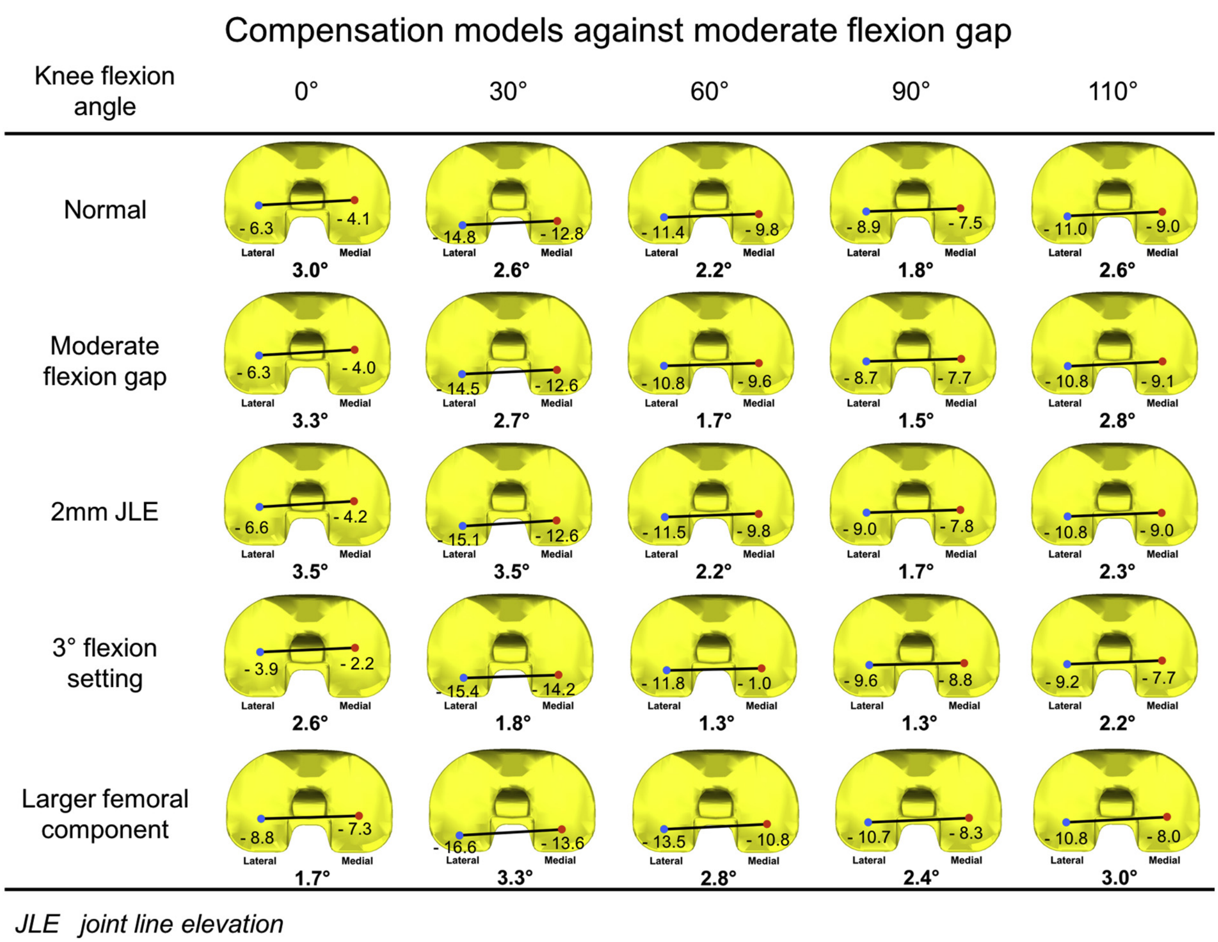

Fig. 9. Lowest points of the medial and lateral femoral condyles relative to the tibial insert at 0°, 30°, 60°, 90°, and 110° of knee flexion in the normal posterior cruciate ligament-sacrificing total knee arthroplasty model and models that compensate for a moderate flexion gap. Please see Fig. 3 for additional information.

Conflicts of interest

None.

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