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Reduction in tibiofemoral conformity in lateral unicompartmental knee arthroplasty is more representative of normal knee kinematics

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Aims

Commonly performed unicompartmental knee arthroplasty (UKA) is not designed for the lateral compartment. Additionally, the anatomical medial and lateral tibial plateaus have asymmetrical geometries, with a slightly dished medial plateau and a convex lateral plateau. Therefore, this study aims to investigate the native knee kinematics with respect to the tibial insert design corresponding to the lateral femoral component.

Methods

Subject-specific finite element models were developed with tibiofemoral (TF) and patellofemoral joints for one female and four male subjects. Three different TF conformity designs were applied. Flat, convex, and conforming tibial insert designs were applied to the identical femoral component. A deep knee bend was considered as the loading condition, and the kinematic preservation in the native knee was investigated.

Results

The convex design, the femoral rollback, and internal rotation were similar to those of the native knee. However, the conforming design showed a significantly decreased femoral rollback and internal rotation compared with that of the native knee ($p < 0.05$). The flat design showed a significant difference in the femoral rollback; however, there was no difference in the tibial internal rotation compared with that of the native knee.

Conclusion

The geometry of the surface of the lateral tibial plateau determined the ability to restore the rotational kinematics of the native knee. Surgeons and implant designers should consider the geometry of the anatomical lateral tibial plateau as an important factor in the restoration of native knee kinematics after lateral UKA.

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Keywords: Unicompartmental knee arthroplasty, Tibial insert, Finite element method

Article focus

- This study investigated the restoration of native knee kinematics with respect to different tibiofemoral (TF) conformities for the tibial insert design in lateral patient-specific (PS) unicompartmental knee arthroplasty (UKA).

Key messages

- In lateral PS UKA, the tibial insert conformity is an important factor that influences the kinematics.
- The convex tibial insert design that mimicked the anatomy showed kinematics similar to those of the native knee.

Strengths and limitations

- The kinematics of the tibial insert design that mimicked the patient anatomy were similar to those of the native knee under a deep knee bend loading condition.
- A computational simulation was used in this study. It was performed without clinical data.

Introduction

Unicompartmental knee arthroplasty (UKA) is an effective treatment for unilateral end-stage osteoarthritis (OA) and osteonecrosis of the knee. Studies have shown long-term survival with excellent functional outcomes

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in UKA.[1-3] Recent studies reported that the long-term survival of UKA is similar to that of total knee arthroplasty (TKA).[4-8] Several studies have also demonstrated better functional outcomes after UKA than TKA.[9,10] This could be attributed to the greater similarity between the kinematics of the native knee and UKA than between the native knee and TKA, because UKA preserves the knee ligaments.[11] Thus, this preservation could be an important factor in restoring the kinematics of the knee in UKA.[11] However, the indications are not well-defined, especially in the lateral compartment.[12] The clinical outcome of UKA in both the compartments is difficult to examine because published studies have not shown a comparison of the different implant designs.[12] In addition, lateral unicompartmental OA occurs less often than medial unicompartmental OA.[13] Lateral UKA is more technically challenging than that of medial UKA owing to the exposure difficulties as well as the limitations of traditional UKA, which does not consider the anatomical differences between the medial and lateral compartments.[14] In general, the lateral tibial plateau is rounder than that of the medial side.[15] The tibial components do not match the anteroposterior (AP) and/or medial:lateral ratio of the tibial plateau. Trained surgeons use methods to compensate for these limitations. The tibial component is translated to the medial side, thereby not covering the lateral aspect of the tibial plateau, and the femoral component is intentionally moved laterally.[15] Common UKA is designed for the medial compartment given the larger volume of patients who require medial repair.[15,16] Owing to the required surgical compromises, the use of standard medial UKA for a lateral compartment indication may cause inadequate rotational alignment of the tibial component and suboptimal bone coverage or overhang of the tibial or femoral components.[17] In addition, the lateral condyle is smaller, and oversizing of the femoral component can result in patellofemoral impingement.[15] Customized patient-specific (PS) implants can overcome the shortcomings of current off-the-shelf implants and improve osseous coverage on the tibial and femoral sides.[14,15] A limitation of a PS design is the variability in the coronal curvature of the femoral component, which could create point loading in select flexion angles when a curved tibial insert is used. Therefore, a flat polyethylene tibial component is paired with a constant coronal curvature femoral component to ensure constant loading conditions over a large area irrespective of the flexion angle.[18] However, in an anatomical native knee, the medial and lateral tibial plateaus have asymmetrical geometries with a slightly dished medial plateau and convex lateral plateau.[19] In addition, tibiofemoral (TF) conformity is one of the most influential factors of kinematics.[20]

Therefore, this study aimed to investigate the native knee kinematics with respect to the tibial insert design corresponding to the lateral femoral component. Subject-specific finite element (FE) models were developed with TF and patellofemoral joints for one female and four male subjects. Three different TF conformity designs were implemented in these models. Flat, convex, and conforming tibial insert designs were applied to the same design of the femoral component. A deep knee bend was the loading condition, and the preservation of the kinematics in the native knee was investigated. The convex tibial design insert that mimicked the anatomy was hypothesized to demonstrate kinematics similar to those of a normal knee.

Patients and Methods

Design of patient-specific lateral UKA. This research was approved by the authors' institutional review board (IRB). Four male subjects (Subject 1: age 36, height 178 cm, weight 75 kg; Subject 2: age 34, height 173 cm, weight 83 kg; Subject 3: age 32, height 182 cm, weight 79 kg; and Subject 4: age 34, height 173 cm, weight 71 kg) and one female subject (Subject 5: age 26, height 163 cm, weight 65 kg) participated in this study. None of the patients had any medical history of lower limb problems. Patient-specific lateral UKA was designed using a previous 3D knee joint model.[21,22] The PS design was initiated through CT and MRI scans. The image data were imported into Mimics version 14.1 (Materialise, Leuven, Belgium) for editing and 3D reconstruction. Planes were introduced through the intersection of the condyles in the sagittal and coronal views. Intersection curves were used to extract the articulating surface geometry in both planes, which were imported into Unigraphics NX (Version 7.0; Siemens PLM Software, Torrance, California) and fitted with rational B-splines (Figure 1).[18,23] The patient's bone defined the sagittal geometry of the femoral component. Thus, it was PS, and the resultant sagittal implant radii varied along the AP dimension of the implant.[18,23] The coronal curvatures of the patient were measured at multiple positions along the femoral condyle. A mean curvature was then derived for each patient. Using this approach, a patient-derived constant coronal curvature was obtained (Figure 1). The tibial component was designed based on the CT and MRI data of the patient's tibia to ensure complete cortical rim coverage. However, unlike the femoral components, three different tibial insert designs were applied: a flat design, a convex design that mimicked the anatomy, and an increasing conformity design. This method was applied to all subject-specific models. The radius of the convex design that mimicked the anatomy was identical to that of the lateral tibial plateau in the anatomy analysis.[24]

Design of the finite element model. The previously mentioned 3D medical imaging data used for a PS UKA design were also used in the development of the FE model (Figure 2).[20,21,23,25] The contours of the tibia, femur, and fibula bones were obtained from the CT to construct the 3D bone model geometry of the knee. The contours of the menisci and cartilage were obtained from the MRI to

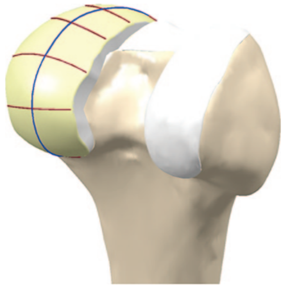

Fig. 1a

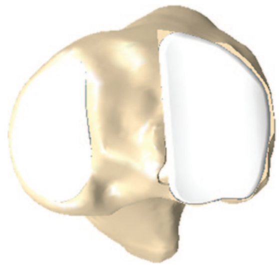

Fig. 1b

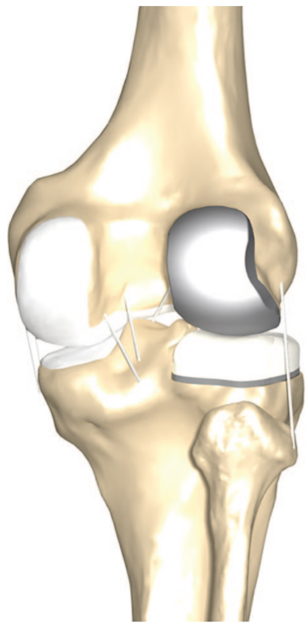

Fig. 1c

Design process of patient-specific (PS) unicompartmental knee: a) spline curves used to model the femoral component; b) polyethylene insert that provides an anatomical fit and perfect coverage; and c) PS unicompartmental knee arthroplasty (UKA) model design.

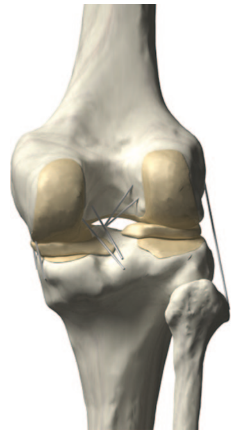

Fig. 2a

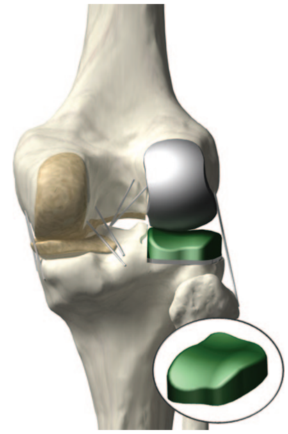

Fig. 2b

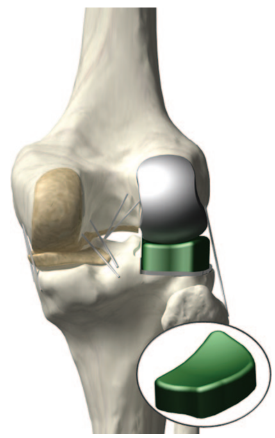

Fig. 2c

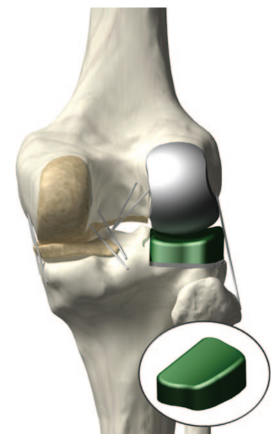

Fig. 2d

The three different finite element (FE) models used in the analysis: a) intact; b) convex design; c) conforming design; and d) flat design.

construct the 3D menisci and cartilage geometry. The ligament insertion points were set with respect to the anatomy obtained from the MRI sets of the subject and the descriptions provided in previous studies.[26-29] The ligaments were simulated as non-linear force elements, and their parabolic and linear equations were as follows, where f is the tension in the ligament, the parameter ε denotes the ligament strain, and k is the stiffness coefficient in each ligament:

$$\text{if } \varepsilon < 0, f(\varepsilon) = 0$$

$$\text{if } 0 \leq \varepsilon \leq 2\varepsilon 1$$

$$f(\varepsilon) = k\varepsilon 2 / 4\varepsilon 1$$

$$\text{if } \varepsilon > 2\varepsilon 1,\ f(\varepsilon) = k(\varepsilon - \varepsilon 1)$$

The linear range threshold was specified as ε1 = 0.03. In all the test scenarios, the soft-tissue elements remained at the same position. The bony structures were modelled as rigid bodies using four-node shell elements.[23] The interfaces between the articular cartilage and bones were modelled as fully bonded.[23] Six pairs of TF contact between the femoral cartilage and meniscus, meniscus and tibial cartilage, and femoral and tibial cartilage were modelled for the medial and lateral sides of the joint.[23]

The heights of the tibial insert for the three different designs were matched to the original bone anatomy using a sagittal cross-sectional image, then aligned with the mechanical axis, and positioned at the medial edge with a square (0°) inclination in the coronal plane of the tibia.[23] The rotating axis was defined as the line parallel to the lateral edge of the tibial baseplate passing through the centre of the femoral component fixation peg. For the implanted model, a 1 mm cement gap was simulated between the component and bone. The materials used for the femoral component, tibial insert, tibial baseplate, and bone cement were cobalt chromium molybdenum

Table I. Material properties for finite element (FE) model

Material	Young's modulus (MPa)	Poisson's ratio
CoCr alloy	220000	0.30
UHMWPE	685	0.47
Ti6Al4V	110000	0.30
PMMA	1940	0.4

CoCr, cobalt chromium; UHMWPE, ultra-high-molecular-weight polyethylene; Ti6Al4V, titanium alloy; PMMA, poly(methyl methacrylate)

Table II. Comparison of mean contact stress on menisci for validation of the model under a loading condition

Meniscus	Mean contact stress, MPa	
	Previous study[31]	Present study
Medial	2.9	3.3
Lateral	1.4	1.6

alloy (CoCrMo), ultra-high-molecular-weight polyethylene (UHMWPE), titanium alloy (Ti6Al4V), and poly(methyl methacrylate) (PMMA), respectively (Table I).[23,28,30] The femoral component contacted the tibial insert. The coefficient of friction between the femoral component and tibial insert was chosen as 0.04.[28]

Loading and boundary conditions. This FE simulation included two types of loading conditions corresponding to the loads used in the experiment for a model validation and the predictions for daily activity loading scenarios. Under the first loading condition, a load of 1150 N was applied to the model to obtain the contact stresses, and was then compared with those reported in a published FE knee joint study.[31] The second loading condition corresponding to deep knee bend loading was applied to evaluate the knee joint mechanics. A computational analysis was performed using an AP force applied to the femur with respect to the compressive load applied to the hip with a constrained femoral internal-external (IE) rotation, free medial-lateral translation, and knee flexion determined through a combination of vertical hip and load of the quadriceps. Thus, a six degrees-of-freedom TF joint was created.[32,33] A proportional-integral-derivative (PID) controller was incorporated into the computational model to control the quadriceps in a manner similar to that used in a previous study.[21] A control system was used to evaluate the instantaneous displacement of the quadriceps muscle, which was required to match the same target flexion profile as used in the experiment. Internal-external and varus-valgus torques were applied to the tibia, while the remaining tibial degrees-of-freedom were constrained.[32,33]

The FE model was simulated using the software ABAQUS (version 6.11; Simulia, Providence, Rhode Island, USA). This study investigated and compared the kinematics of the PS UKA designs for three different conformities using a native knee. The kinematics were calculated based on the Grood and Suntay definition of a joint coordinate system.[34]

Statistical analysis. Single cycles of the deep knee bend loading condition were divided into 11 timepoints (0.0 to 1.0 phases). To assess the three different tibial insert designs – flat, convex, and conforming – each design condition was compared with the native knee in a pairwise manner using non-parametric repeated-measurement Friedman tests at each phase of the cycle. Post hoc comparisons were performed using a Wilcoxon's signed rank test with a Holm correction to control the familywise error rate for the tests conducted within each phase of the cycle. Statistical analyses were performed using SPSS for Windows (version 20.0.0; IBM, Armonk, New York, USA). The statistical significance was set to $p < 0.05$ for all comparisons.

Results

Native knee validation. The five subject-specific FE models were validated by comparison between the mean contact stress on the menisci from the previous study and that obtained in our FE model (Table II).[31] These differences could be due to the geometrical differences between the different studies, such as the thickness of the cartilage and menisci. The consistency between the validation results and the results reported in the literature demonstrated the validity of the results obtained through the FE model used in this study.

Comparison of the kinematics in the lateral PS UKA designs in the three different conformities and native knee during the deep knee bend condition. Figure 3 shows the femoral rollback and tibial internal rotation in the three different tibial insert designs under the deep knee bend condition. The conforming design showed a significantly decreased femoral rollback compared with that of the native knee. The conforming design was smaller than the native knee, but a mean significant decrease of 2.1 mm in the femoral rollback was identified during the deep knee bend loading condition. In addition, the conforming design showed significantly smaller tibial internal rotation angles during the deep knee bend loading condition.

The flat design demonstrated a significant mean increase of 1.5 mm in the femoral rollback as compared with the native knee. The flat designs showed an increased tibial internal rotation during knee flexion. In addition, the tibial internal rotation in the flat design was 2.6° greater than that of the native knee; however, it was not significant except for the early flexion. The convex design showed a decreased femoral rollback. However, this was the closest femoral rollback to the native knee during the deep knee bend condition. The mean difference between the femoral rollback of convex design and that of the native knee was found to be 0.8 mm. The tibial internal rotational kinematics of the convex design were similar to those of the native knees throughout knee flexion. In addition, the mean difference between the internal rotation convex design and native knee was 0.7°.

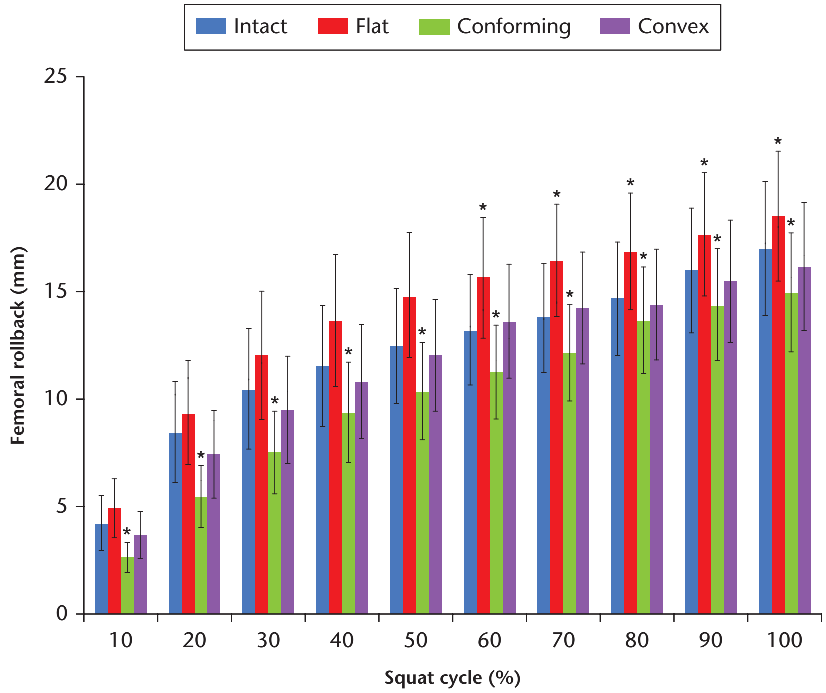

Fig. 3a

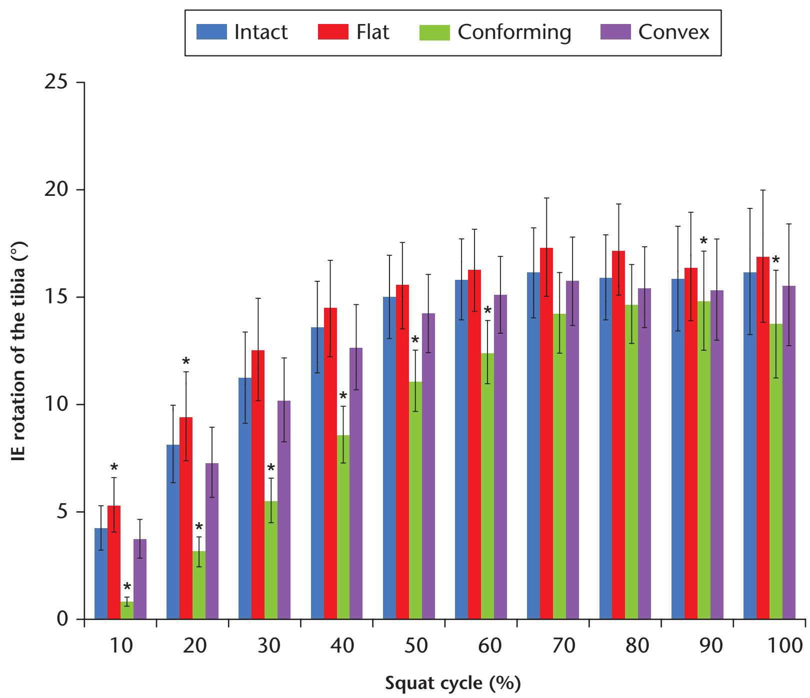

Fig. 3b

Comparison of three different unicompartmental knee arthroplasty (UKA) designs during the squat cycle: mean (SD) of a) femoral rollback (anteroposterior mean); and b) tibial rotation (internal-external mean). *$p < 0.05$.

Discussion

The most important finding of this study was that the convex design that mimicked the anatomy preserved the native knee kinematics in lateral PS UKA. The results indicated that the femoral rollback kinematics of the native knee were not restored when the flat and conforming designs were used in lateral PS UKA. However, it was preserved after the convex design lateral PS UKA. This is the first study to evaluate the lateral PS UKA kinematics of a tibial insert design using a computational analysis.

In practice, surgeons do not prefer implants with insufficient tibial coverage because they induce the possibility of tibial implant collapse because the load can be transmitted to the cancellous bone as opposed to the cortical bone.[35] Another concern with insufficient coverage is posterior loosening due to the femoral rollback. We believe that the main reason for failure in lateral UKA is the improper selection, indication, or technique (overcorrection), rather than an inadequate tibial implant, except for the mobile bearing implants in the lateral compartment.[36] Carpenter et al[37] utilized the morphometric data to compare the size, match, and fit between PS implants and incrementally sized off-the-shelf UKA implants from several different implant manufacturers. They showed that PS implants provided superior cortical bone coverage and fit, while minimizing the overhang and insufficient coverage of the off-the-shelf implants.[37]

Demange et al[14] determined that using a PS UKA implant resulted in more precise component positioning and better tibial bone coverage than when using commercially available standard medial implants for the treatment of lateral unicompartmental arthritis. They showed that PS lateral UKA demonstrated better tibial coverage and provided excellent short-term clinical and radiological results compared with standard lateral UKA.[14] Recently, the biomechanical effects of PS and standard off-the-shelf prostheses for UKA were compared.[23] The PS UKA yielded mechanics closer to those of the normal knee joint.[23] In addition, the decreased contact stress on the opposite compartment could reduce the overall risk of progressive OA.[23] Based on surgical technique, UKA is more demanding than TKA, particularly on the lateral side. The biomechanics of the lateral compartment are different to those of the medial compartment. For instance, femoral rollback is greater in the lateral compartment.[38] The wear of the original articular surface, as well as that of the polyethylene component after arthroplasty, are different in both compartments.[39] The shape of the femoral condyles and the tibial plateau are different in the medial and lateral sides.[40] As previously mentioned, surgeons must modify their technique to perform lateral UKA with conventional implants, which are designed for common medial compartment procedures.[15] The design and shapes of the PS implants are different for the medial and lateral compartments. Owing to the PS shape of the implant with a j-curve (designed to mimic the native femur and the full cortical coverage of the femoral implant), shifting the femoral component laterally, as described by Sah and Scott,[41] becomes unnecessary. As previously discussed, earlier customized UKA design of the femoral component has a PS sagittal curvature corrected for deformity and a constant curvature in the coronal plane. The constant coronal curvature was matched to the curvature of the tibial insert to minimize polyethylene wear. However, anatomical medial and lateral tibial plateaus have asymmetrical geometries, with a slightly dished medial plateau and a convex lateral plateau.[19] In addition, the biomechanics of the medial and lateral menisci are different.[42] The medial meniscus is significantly less mobile than the lateral meniscus due to its attachment to the medial collateral ligament (MCL) and the larger insertion areas. Therefore, the medial meniscus contributes more to joint stability than the lateral meniscus, which closely follows the AP excursion of the femur.[42] The dished medial plateau and greater stability of the medial meniscus limit the AP motion and posterior rollback of the medial femoral condyle. In contrast, the convex lateral plateau, combined with lateral meniscus mobility, enables a greater range of AP motion with a greater posterior rollback of the lateral femoral condyle. Thus, in activities of high flexion, such as a deep knee bend, the knee shows an overall medial pivot motion with a greater rollback of the lateral femoral condyle.[43] In addition, previous studies demonstrated that different types of kinematics were found with respect to the TF joint conformity.[20,44,45]

This study showed that different types of kinematics were found with respect to the TF joint conformity. The conforming design showed limited kinematics in the femoral rollback and internal rotation compared with that of the native knee. In addition, the flat design showed increased femoral rollback and tibial internal rotation kinematics compared with those of the native knee. The flat design also showed increased femoral rollback and tibial internal rotation kinematics compared with those of the conforming design and native knee owing to more degrees of freedom. The tibial internal rotation was attributed to the anterior cruciate ligament (ACL) and the asymmetry between the medial and lateral femoral condyles. The screw-home mechanism was reported as a sharp internal rotation near extension.[46,47] The proximal tibia rotates externally on the distal femur by approximately 15° during the final 20° of extension, with the screw-home mechanism locking the knee in extension and limiting the anterolateral rotatory movement.[48] However, in this study, the screw-home mechanism was not preserved in the conforming UKA design despite the restoration of the ACL; it was restored in the UKA with flat and convex designs. Recently, Wada et al[11] showed that the rotational kinematics of flat surface fixed-bearing lateral UKA were similar to those of the native knees throughout the knee flexion. Their results were similar to those obtained in this study for lateral flat surface fixed-bearing UKA.

In terms of TF conformity, the loss of the upslope of the tibia in the conforming design of UKA changed the rotational kinematics after lateral UKA, especially in the early flexion phase. A recent dynamic computational simulation showed that a bicruciate-retaining TKA implant, designed with biomimetic inserts mimicking the native tibial surface geometry, achieved more kinematic improvement than that of cruciate-retaining TKA, contemporary bicruciate TKA, or bi-UKA.[49]

In terms of the clinical relevance, this study also showed the importance of preserving the native geometry for kinematics restoration of the native knee. In other words, although the femoral component was designed to be PS, the tibial insert cannot preserve intact the tibial plateau anatomy, thus it is challenging to preserve natural kinematics. However, although a convex lateral compartment was developed that mimicked the patient's anatomy, perfect native knee kinematics could not be restored. The role of the menisci is important in the TF joint, and the lateral meniscus is more mobile than that of the medial side. Moreover, the material properties of the tibial insert are different to those of the native TF joint. However, these results suggest that the articular surface in knee arthroplasty should be further considered.

Two strengths of this study should be highlighted. First, in contrast to the current biomechanical UKA model, this study included deep knee bend loading, as opposed to a simple vertical static loading condition.[30] Second, the biomechanical effect was evaluated using a single model in the previous FE model; however, five subject-specific FE models with different PS tibial insert designs for UKA were developed to investigate their biomechanical effects.[23,30]

Nevertheless, this study had several limitations. First, the UKA had fully bonded, and the micromotion, which may occur between the tibial compartment and tibial insert, was not considered in this study. Second, only the initial model was validated. However, this method has been widely used in biomechanics.[30,31,50] Third, the results do not predict the clinical results and patient satisfaction. Fourth, although five subject-specific FE models were developed, unlike in other earlier studies, there was only one female subject. We would recommend increasing the number of female subjects in future studies. Finally, the model assumed the material properties and attachment points of the ligaments based on variable values from the references. However, the objective was not to determine the actual values of the muscle and ligament forces. This study aimed to determine the effect of variability in lateral PS UKA with respect to tibial insert conformity for the variables of interest.

In conclusion, these results showed that the geometry of the surface of the lateral tibial plateau determined the ability to restore the rotational kinematics of the native knee. Surgeons and implant designers should consider the anatomical lateral tibial plateau geometry in the restoration of native knee kinematics after lateral UKA.

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- Y-G. Koh: Designed the study, Evaluated the results using finite element analysis, Wrote the manuscript.
- J-A. Lee: Developed the 3D model.
- H-Y. Lee: Evaluated the results using finite element analysis.
- H-J. Kim: Evaluated the results using finite element analysis.
- H-S. Chung: Evaluated the results using finite element analysis.
- K-T. Kang: Supervised the study, Analyzed the data.
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