BJR

■ **KNEE**

Patient-specific medial unicompartmental knee arthroplasty has a greater protective effect on articular cartilage in the lateral compartment

A FINITE ELEMENT ANALYSIS

K-T. Kang,
J. Son,
D-S. Suh,
S. K. Kwon,
O-R. Kwon,
Y-G. Koh

Yonsei Sarang Hospital, Seoul, South Korea

Objectives

Patient-specific (PS) implantation surgical technology has been introduced in recent years and a gradual increase in the associated number of surgical cases has been observed. PS technology uses a patient's own geometry in designing a medical device to provide minimal bone resection with improvement in the prosthetic bone coverage. However, whether PS unicompartmental knee arthroplasty (UKA) provides a better biomechanical effect than standard off-the-shelf prostheses for UKA has not yet been determined, and still remains controversial in both biomechanical and clinical fields. Therefore, the aim of this study was to compare the biomechanical effect between PS and standard off-the-shelf prostheses for UKA.

Methods

The contact stresses on the polyethylene (PE) insert, articular cartilage and lateral meniscus were evaluated in PS and standard off-the-shelf prostheses for UKA using a validated finite element model. Gait cycle loading was applied to evaluate the biomechanical effect in the PS and standard UKAs.

Results

The contact stresses on the PE insert were similar for both the PS and standard UKAs. Compared with the standard UKA, the PS UKA did not show any biomechanical effect on the medial PE insert. However, the contact stresses on the articular cartilage and the meniscus in the lateral compartment following the PS UKA exhibited closer values to the healthy knee joint compared with the standard UKA.

Conclusion

The PS UKA provided mechanics closer to those of the normal knee joint. The decreased contact stress on the opposite compartment may reduce the overall risk of progressive osteoarthritis.

■ K-T. Kang, PhD, Researcher, Department of Mechanical Engineering,

■ J. Son, MS, Researcher, Department of Mechanical Engineering, Yonsei University, 50 Yonsei-ro, Seodaemun-gu, Seoul 03722, South Korea.

■ D-S. Suh, MD, Orthopaedic Surgeon, Department of Orthopaedic Surgery,

■ S. K. Kwon, MD, Orthopaedic Surgeon, Department of Orthopaedic Surgery,

■ O-R. Kwon, MD, Orthopaedic Surgeon, Department of Orthopaedic Surgery,

■ Y-G. Koh, MD, Orthopaedic Surgeon, Department of Orthopaedic Surgery, Joint Reconstruction Center, Yonsei Sarang Hospital, 10 Hyoryeong-ro, Seocho-gu, Seoul 06698, South Korea.

Correspondence should be sent to O-R. Kwon; email: osorkwon@gmail.com and Y-G. Koh; email: osygkoh@gmail.com

Article focus

■ This study compares patient-specific (PS) and standard unicompartmental knee arthroplasties (UKAs) with a healthy knee through a finite element (FE) analysis.

Key messages

■ The contact stress on the lateral compartment (lateral meniscus and articular cartilage) in the PS UKA showed closer values to those of a healthy knee control.

■ No significant difference was found between the contact stresses on the medial PE insert in the PS and standard UKAs.

Strengths and limitations

■ The computational simulation enabled the estimation of the contact stress on the

PE insert and the lateral compartment under gait cycle conditions.

- Only one standard UKA prosthesis was examined through computational simulation. Thus, other prostheses may show different results.
- The FE analysis used in our study was performed without clinical data.

Introduction

Standard off-the-shelf hip and knee prostheses, which are used in the two most common orthopaedic surgeries worldwide, have been successfully implanted in many patients for over 30 years.[1] Many previous anatomical studies evaluated the wide range of variability in the size and shape of the tibiofemoral joint.[2-4] In addition, osteoarthritis (OA) of the knee joint has become increasingly common in young and middle-aged patients.[5] The rate of unicompartmental knee arthroplasty (UKA) has increased three times faster than the rate of total knee arthroplasty (TKA).[5,6] However, most femoral and tibial components used in UKA provide a range of five to six implant sizes despite the high degree of differentiation between patients.[7]

Each prosthetic system generally provides pre-designed geometries according to the manufacturer. Moreover, the prosthesis size is adjusted up or down intra-operatively in an attempt to optimise the resected bony surfaces.[7] The prosthetic fit influences the success rate in UKA.[8] Femoral and tibial overhang may cause increased pain and impingement of the soft tissue.[9-11] In contrast, femoral and tibial resection underhang contributes to component loosening and subsidence.[3,7,12]

Patient-specific (PS) prostheses are used to treat patients in situations where standard off-the-shelf prostheses would not be a perfect fit.[13] Fitzpatrick et al[14] reported that PS resurfacing implants enable a femoral bone-preserving approach and enhance the cortical bone support in the tibia, overcoming the critical design limitations of commercial off-the-shelf implants. Carpenter et al[7] also reported that significantly less cortical rim overhang and undercoverage were observed with PS UKA in 30 patients. As such, PS UKA can restore normal anatomy, joint line position and normal joint function, with the potential to result in more normal knee kinematics. Harrysson et al[1] reported that PS TKA showed a more even stress distribution on the bone-implant interface using finite element (FE) analysis. This even distribution theoretically reduces uneven bone remodelling that can lead to premature loosening.[1] Similarly, van den Heever et al[15] demonstrated through FE analysis that PS UKA resulted in lower contact stresses at the tibiofemoral joint compared with a fixed-bearing design and displayed a more uniform stress distribution at the bone-implant interface than any of the other implant designs in UKA.

The most important advantage of PS UKA or PS TKA is the theoretically perfect coverage of the resected bone, which reduces the rate of underhang or overhang. The curvature parameters are the most important consideration for contact stress optimisation in PS UKA and TKA prosthetic design, even though patient kinematics would be enhanced if PS anatomical geometry is used accordingly. However, the effect of PS UKA has not yet been compared with that of standard UKA from a biomechanical perspective.

This study aimed to investigate the difference in the biomechanical effects of PS UKA and standard UKA. A validated FE model was used to evaluate the contact stresses on the polyethylene (PE) insert, articular cartilage and lateral meniscus. Stresses were compared with those of a normal healthy knee joint to assess the biomechanical effect of the prostheses. Both PS and standard UKAs involved fixed-bearing-type prostheses. The loading condition studied was normal-level walking.

Materials and Methods

Our study was approved by the Institutional Review Board of Yonsei Sarang Hospital, Seoul, South Korea (Project No.: 17-D02-01, Protocol No.: FEA_1.0.).

Normal knee model. A non-linear 3D FE model of the intact knee was developed from CT and MRI of the knee joint of a healthy 36-year-old male subject. Computed tomography and MRI were performed using a 64-channel CT scanner (SOMATOM Sensation 64; Siemens Healthcare GmbH, Erlangen, Germany) and a 3T MRI system (Discovery MR750w; GE Healthcare, Milwaukee, Wisconsin), respectively. The medical records of the subject were reviewed and no musculoskeletal disorders or limb alignment problems were found. Thus, the investigated knee was considered healthy without any history of related diseases. This computational knee joint model was established and validated in previous studies.[16-18]

A 3D reconstruction procedure was initially required to develop the 3D FE model. The medial images were processed and segmented in software (Mimics 17.0; Materialise, Leuven, Belgium) designed to generate the 3D structures of the lower extremity. The reconstructed CT and MRI models were combined with the positional alignment of each FE model using Rapidform (3D Systems Korea, Inc., Seoul, South Korea), in which the bony structures were considered to be rigid bodies.[19] The articular cartilage and the menisci were modelled as isotropic and transversely isotropic, respectively, with linear elastic material properties (Table I).[20] All major ligaments were modelled with nonlinear and tension-only spring elements (Table II). The interfaces between the articular cartilage and the bones were modelled to be fully bonded.[21-23] The six pairs of tibiofemoral contact between the femoral cartilage and the meniscus, the meniscus and the tibial cartilage and the femoral cartilage and the tibial

Table I. Material properties for the finite element model

	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
Ti6Al4V alloy	117 000	0.30
PMMA	1940	0.40

MPa, Megapascal; CoCrMo, cobalt chromium molybdenum; UHMWPE, ultra-high-molecular-weight polyethylene; Ti6Al4V, Titanium aluminium vanadium alloy; PMMA, polymethylmethacrylate (acrylic)

Table II. Material properties in the ligaments

Ligament	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

ACL, anterior cruciate ligament; PCL, posterior cruciate ligament; LCL, lateral collateral ligament; MCL, medial collateral ligament; PFL, patellofemoral ligament; OPL, oblique popliteal ligament; CAP, posterior capsule; ALS, anterolateral structures; CM, deep medial collateral ligament

Fig. 1a   Fig. 1b

Fig. 1c   Fig. 1d

Design process of patient-specific unicompartmental knee arthroplasty for the a) subject's anatomic 'J' curve; b) spline curves used to model the femoral component; c) femoral component; and d) polyethylene insert that provides an anatomical fit and a perfect coverage.

cartilage were modelled for both the medial and lateral sides of the joint.[17] A finite sliding frictionless hard contact algorithm with no penetration was adopted for the contact points in all articulations.[17] Convergence was defined as a relative change of > 5% between two adjacent meshes. The average element size of the simulated articular cartilage and menisci was 0.8 mm.

PS UKA design. The design process of PS UKA is described as follows: 3D-reconstructed PS bone geometry and surface data were used to develop the PS UKA geometry. The 3D images were converted to stereolithography files using Mimics software, then loaded into the digital computer-aided design software, 3-matic (version 9.0; Materialise, Leuven, Belgium). This software allows the user to combine geometry from mixed sources into one project. The initial graphics exchange specification files exported from 3-matic were entered into Unigraphics NX software (version 7.0; Siemens PLM Software, Torrance, California) to develop the PS UKA.

The PS UKA geometry was designed to ensure a minimum component thickness of 3.0 mm for the femoral component and 2.0 mm for the metal backing of the tibial component. The sagittal geometry of the subject's bone was used for the geometry of the PS femoral component.[24,25] Planes were introduced to the condyles in the sagittal view, in which the curves were used to extract the articulating surface geometry (Fig. 1).[15] The subject's anatomic 'J' curves were mimicked through this procedure, providing a better approximation of the normal articulating geometry (Fig. 1). The coronal curvatures were measured in multiple positions along the length of the femoral condyle by following the derivation of a mean value for the patient.[24] The average curvature was adjusted to maintain a minimum thickness of > 3 mm, which allowed for the derivation of the patient's constant coronal curvature (Fig. 1). On the tibial plateau, the tibial component geometry was defined by the patient's tibial profile. The tibial plateau and the PE inserts were designed for minimal bone resection to provide a smooth articulating surface for the femoral component. The implant provided complete cortical rim coverage (> 95%) because it was PS. This result would not be achieved with the standard off-the-shelf UKA prosthesis (Fig. 1).[14]

FE models of PS and standard UKAs. The standard UKA (Zimmer Biomet Inc., Warsaw, Indiana) FE model was

Fig. 2a   Fig. 2b   Fig. 2c
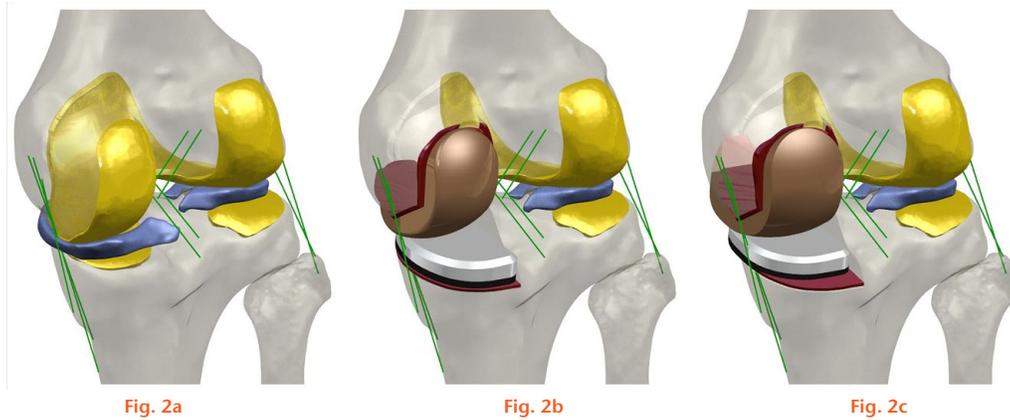
Finite element models used in the analysis: (a) normal knee; (b) patient-specific unicompartmental knee arthroplasty (UKA); and (c) standard UKA.

developed using a 3D laser scanner. The detailed procedure is described in a previous study.[16] The height of the PE insert was matched to the original bone anatomy by a sagittal cross-sectional image,[16,26,27] then aligned with the mechanical axis and positioned at the medial edge of the tibia. The neutrally aligned tibial component was defined as having a square (0°) inclination in the coronal plane.[16,26,27] A rotating axis was defined as the line parallel to the lateral edge of the tibial component passing through the centre of the femoral component fixation peg. A transverse resection with a posterior slope of 7° was made at the same anterior depth in both the PS and standard UKA models. Moreover, the malalignment or overcorrection in the femoral compartment was not considered (Fig. 2). A posterior slope of 7° was referred to herein as the subject's anatomical posterior slope. With respect to the implanted model, a 1 mm cement gap was simulated between the component and the bone.

The PE insert and the femoral and tibial components were modelled as elastoplastic and linear elastic isotropic materials, respectively.[27-29] The materials of the femoral component, PE insert, tibial component and bone cement were cobalt chromium molybdenum alloy (CoCrMo), ultra-high-molecular-weight polyethylene (UHMWPE), titanium alloy (Ti6Al4V) and polymethyl methacrylate (PMMA), respectively (Table I). The femoral component made contact with the PE insert. The selected coefficient of friction between the PE and metal was 0.04.[29]

This FE investigation included two types of loading conditions corresponding to the loads used in the experimental part of the study for model validation and predictions for daily activity loading scenarios. The intact model was validated in our previous study,[16-18] while the UKA model was validated by comparison with a previous experimental study.[30] The validation of the UKA model was performed with flexion angles of 0°, 30°, 60° and 90° through a passive flexion simulation. The anterior and posterior drawer loads of 130 N were separately applied to the tibia at the knee centre, in a similar fashion to that of the experimental study.[30] The daily activity loading was applied as a

second loading to compare the biomechanical effects of the PS and standard UKAs. The computational analysis was performed with the anteroposterior force applied to the femur with respect to the compressive load applied to the hip.[31-34] 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 the experiment.[35]

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.[35] Internal-external and varus-valgus torques were applied to the tibia.[31-34] The FE model was analysed using Abaqus software (version 6.11; Dassault Systemes Simulia Corp.; Providence, Rhode Island). The contact stresses on the PE insert, articular cartilage and lateral meniscus were compared in the PS and standard UKAs and the normal healthy knee.

Results

UKA model validation. In the anterior drawer test at 130 N, the anterior tibial translations in the standard UKA model were 6.1 mm, 9.9 mm, 8.7 mm and 8.5 mm, while those in the posterior drawer test were 5.8 mm, 4.3 mm, 3.8 mm and 4.9 mm, respectively, at 0°, 30°, 60° and 90° of knee flexion (Fig. 3). Agreement was observed between the results from our simulation and those from the previous experimental study within the ranges of values under the anterior and posterior drawer loadings.[30]

Comparison of the maximum contact stress on the PE insert in the PS and standard UKA FE models. In order to evaluate the effect of the PS UKA compared with the standard UKA on the PE insert, the maximum contact stresses on the PE inserts were examined under gait loading conditions (Fig. 4). The maximum contact stresses on the PE insert were found at approximately 17% of the gait cycle in both PS and standard UKAs. The maximum contact stress on the PE insert was greater during the stance phase in the PS UKA and during the swing phase of the gait cycle in the standard UKA. The maximum contact stress in the

Fig. 3a

Fig. 3b
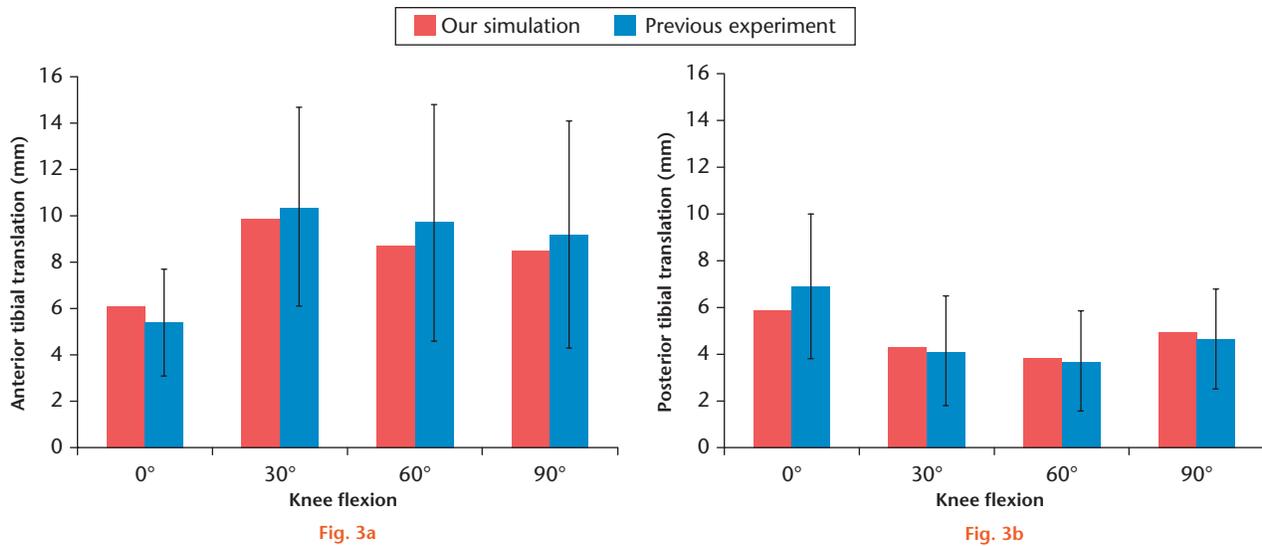
Tibial translations in the anterior (a) and posterior (b) drawer tests under 130 N. The error bars indicate 1 standard error.

Fig. 4
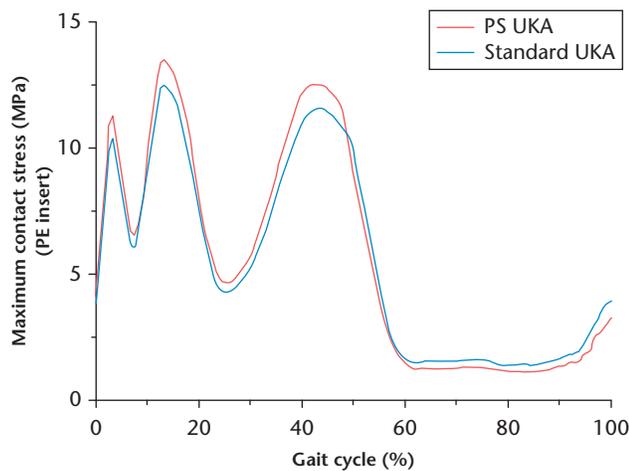
Maximum contact stress on the polyethelene (PE) insert in patient-specific (PS) unicompartmental knee arthroplasty (UKA) and standard UKA.

PS UKA in the stance phase was 11.5% greater than that in the standard UKA. In contrast, the maximum contact stress in the standard UKA during the swing phase was 19.3% greater than that in the PS UKA. However, no significant biomechanical effect of the PS UKA was observed on the PE insert that surpassed the standard UKA.

Comparison of the maximum contact stress on the articular cartilage and the lateral meniscus in the PS and standard UKA FE models. The results were compared within the lateral compartment of the normal healthy knee to evaluate the biomechanical effect of the PS UKA on the articular cartilage and lateral meniscus compared with that of the standard UKA. Figure 5 shows the maximum contact stresses on the articular cartilage and lateral meniscus in the normal healthy knee and in the PS UKA and the standard UKA models during the gait cycle. A complicated trend was found for the articular cartilage.

The maximum contact stresses on the articular cartilage in both the PS and standard UKAs were lower than those on the normal healthy knee during the stance phase. The maximum contact stress on the articular cartilage decreased by 16% to 18% and by 15% to 17% in the PS and standard UKAs, respectively, compared with the normal healthy knee in the stance phase. However, both UKA types showed greater stresses than the normal healthy knee in the swing phase. The contact stresses in the standard UKA was 32% greater, whereas that in the PS UKA was 7% greater, than the stresses in the normal healthy knee. Similar to those in the articular cartilage, lower contact stresses were found on the lateral meniscus in both PS and standard UKAs in the stance phase compared with the normal healthy knee. However, in double support, the contact stresses on the lateral meniscus in the PS and standard UKAs exceeded the contact stress in the normal healthy knee at approximately 50% of the gait cycle. The contact stresses on the lateral meniscus in the swing phase increased by 28% and 5% in the standard and PS UKAs, respectively, compared with the normal healthy knee.

Discussion

The PS surgical technique for UKA or TKA was implemented to improve functional outcomes and satisfaction rates.[7,36] The reported benefits of the PS implant system include a more normal femoral rollback, enhanced coverage of the tibial plateau and a reduction in blood transfusions.[7,23,37,38] UKA has also been offered to younger and more active patients as a more conservative treatment option than TKA.[15]

Fitzpatrick et al[14] described the analysis of various theoretical designs for unicompartmental, commercially available implants on 34 tibias. The theoretical design in which the shape and size could both be altered was
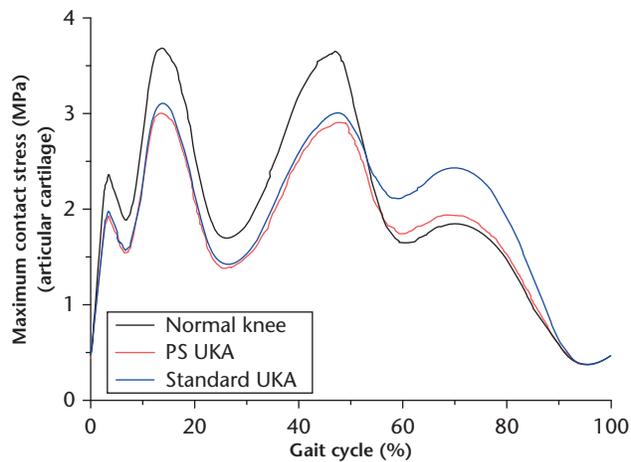
Fig. 5a
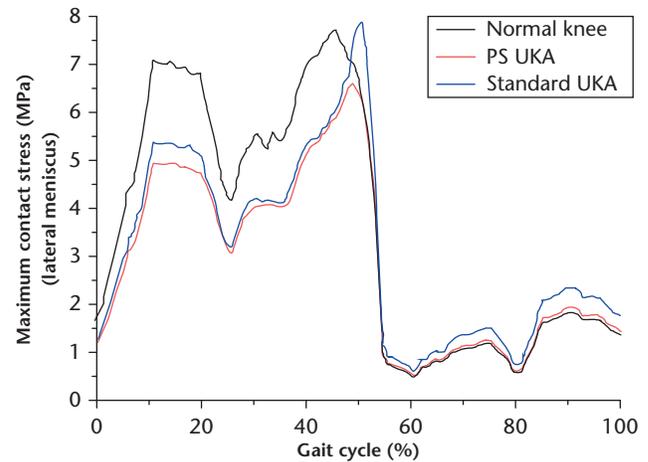
Fig. 5b

Maximum contact stress on a) articular cartilage (of the lateral compartment only) and; b) lateral meniscus in the normal knee, patient-specific (PS) unicompartmental knee arthroplasty (UKA), and standard UKA.

found to provide significantly better cortical rim coverage than the standard off-the-shelf implants, regardless of the shape. However, a recent study found that the PS UKA provides statistically superior cortical bone coverage and fit because it minimises the rates of overhang and under-coverage that occur with standard off-the-shelf implants.[7] In addition, Patil et al[37] reported that PS implants provide kinematics closer to normal knee mechanics, even without using custom blocks, compared with standard off-the-shelf implants. Koeck et al[39] found that the PS UKA restores a more reliable mechanical axis of the lower extremity by obtaining a medial proximal tibial angle of 90° to avoid malpositioning and provide maximal tibial coverage. However, from a mechanical point of view, the majority of failures following UKA were caused by two problems: the wear of the PE insert; and progressive OA in the other compartment.[16,40] To the best of our knowledge, no study has evaluated the effect of the PS UKA, compared with the standard UKA, on the restoration of normal healthy knee mechanics with respect to the other compartment.

We hypothesised that if the implanted PS UKA prosthesis is as similar as possible to the normal healthy knee joint, it could result in post-operative biomechanical improvements as well as improved functional outcomes in the opposite compartment. A 3D non-linear FE model of the knee joint, which was primarily developed with bony structures and soft tissues, including the major ligaments, menisci and articular cartilage, was evaluated under gait-cycle loading conditions in order to verify our hypothesis. The maximum contact stress, which was closely associated with degenerative OA of the knee joint, was estimated.[19,41] Contact stress is one of the main determinants of the long-term survival of a fixed-bearing UKA because higher contact stresses lead to a higher propensity for abrasive damage on the PE insert and, accordingly, accelerated wear.[24] The trends in the contact

stresses on the PE insert, articular cartilage and lateral meniscus were consistent with the results of previous studies.[16,29,40] However, the study results showed no difference in the biomechanical effect of a PS UKA compared with that of a standard UKA. The PS UKA showed a lower contact stress in the swing phase, but a greater contact stress in the stance phase.

The more notable advantage of the PS UKA over the standard UKA was found in the lateral compartment. The contact stresses on the articular cartilage with the PS and standard UKAs were lower than those of the normal healthy knee in the initial stance phase. This could be caused by a difference in the stiffness of the PE insert and the articular cartilage because the axial load was mainly applied in the stance phase. The stiffness of the PE insert was greater than that of the articular cartilage. Hence, the contact stress was lower following UKA than it was in the normal healthy knee during the stance phase. The biomechanical advantage of the PS UKA over the standard UKA in relation to the impact on the articular cartilage was found in the swing phase during joint flexion. In other words, the PS UKA, using the patient's anatomical sagittal 'J' curves in the design, produces a biomechanical effect closer to that of a normal healthy knee than does the standard UKA. Likewise, we reported in a previous study that keeping the joint line close to the intact anatomical position led to better biomechanical outcomes in the knee joint.[16] Furthermore, in our model, the PS and standard UKAs were virtually implanted in the ideal anatomical position, with the original cartilage replicating the original joint line, without malalignment of the components. Therefore, the beneficial biomechanical effect of the PS UKA may be related to the preservation of the PS geometry in the prosthetic design.

Similar to the findings for the articular cartilage, the contact stress on the lateral meniscus also decreased in the stance phase after UKA. In addition, this difference in

the lateral meniscus was greater than that in the articular cartilage because of the difference in the stiffness between the lateral meniscus and the PE insert. Furthermore, the PS UKA was more effective at reducing the contact stress on the lateral meniscus than was the standard UKA, as was shown in the improvement in the swing phase. No improvement was observed in the biomechanical effect on the PE insert in the PS UKA compared with the standard UKA, as was initially assumed in this study. However, we do expect to prevent progressive degenerative OA in the lateral compartment because the PS UKA produced mechanics similar to those of the normal healthy knee.

Several strengths of the present study should be highlighted. First, in contrast to the previous UKA studies, the FE model in this study included the tibia, femur and related soft tissues.[26,27,42] Secondly, in contrast to the current biomechanical UKA model, the present study included the application of gait cycle loading as opposed to a simple vertical static loading condition.[26,27,42-45] Thirdly, the current study not only validated the intact model, but also performed a kinematic validation on the UKA FE model.

Nevertheless, several limitations should also be noted. First, the anatomy for the UKA design was based on, and virtually implanted in, only one subject. However, the advantages of a computational simulation with a single subject are the determination of the component alignment effect within the same subject and the elimination of variables, such as weight, height, bony geometry, ligament properties and component size.[46] In addition, a larger number of subjects will be evaluated in future studies. Secondly, the results do not necessarily predict clinical results and patient satisfaction. However, the contact stresses on the PE insert, cartilage and lateral meniscus are the key factors that should be investigated for the evaluation of the biomechanical effect in computational biomechanics.[16-20,26-30,35,42-44,46] In addition, as previously mentioned, the contact stress is closely associated with degenerative OA. Thirdly, only one standard UKA prosthesis was examined. Prostheses from other manufacturers with different types of PE insert and tibial trays may provide different results. Fourth, the bony structures were assumed to be rigid. A bone is composed of cortical and cancellous tissue, but the evaluation of the effect on the bone was not the purpose of this study. Moreover, the bone had minimal influence on this study because it had a greater stiffness than the relevant soft tissues.[19] Finally, only the gait cycle was simulated, and more demanding activities, such as sitting and standing from a seated position, ascending and descending stairs, or squatting, would be required for a more reliable investigation in the future.

In this study, the contact stresses on the PE insert and the stresses in the lateral compartment were evaluated to investigate the biomechanical effect of the PS UKA compared with the standard UKA. In conclusion, the contact stress on the other lateral in the PS UKA was lower than that in a UKA with a standard off-the-shelf implant, indicating that the PS UKA may have an advantage in preventing progressive degenerative OA. However, in terms of the PE insert, the PS UKA could not be expected to extend the life expectancy of the insert any more than a UKA with a standard off-the-shelf implant.

References

1. **Harrysson OL, Hosni YA, Nayfeh JF.** Custom-designed orthopedic implants evaluated using finite element analysis of patient-specific computed tomography data: femoral-component case study. *BMC Musculoskelet Disord* 2007;8:91.

2. **Hashemi J, Chandrashekar N, Gill B, et al.** The geometry of the tibial plateau and its influence on the biomechanics of the tibiofemoral joint. *J Bone Joint Surg [Am]* 2008;90-A:2724-2734.

3. **Servien E, Saffarini M, Lustig S, Chomel S, Neyret P.** Lateral versus medial tibial plateau: morphometric analysis and adaptability with current tibial component design. *Knee Surg Sports Traumatol Arthrosc* 2008;16:1141-1145.

4. **Kang KT, Son J, Kwon OR, et al.** Morphometry of femoral rotation for total knee prosthesis according to gender in a Korean poputation using three-dimensional magnetic resonance imaging. *Knee* 2016;1:1-2.

5. **Riddle DL, Jiranek WA, McGlynn FJ.** Yearly incidence of unicompartmental knee arthroplasty in the United States. *J Arthroplasty* 2008;23:408-412.

6. **Pailhé R, Cognault J, Massfelder J, et al.** Comparative study of computer-assisted total knee arthroplasty after opening wedge osteotomy versus after unicompartmental arthroplasty. *Bone Joint J* 2016;98-B:1620-1624.

7. **Carpenter DP, Holmberg RR, Quartulli MJ, Barnes CL.** Tibial plateau coverage in UKA: a comparison of patient specific and off-the-shelf implants. *J Arthroplasty* 2014;29:1694-1698.

8. **Schotanus MG, Sollie R, van Haaren EH, et al.** A radiological analysis of the difference between MRI- and CT-based patient-specific matched guides for total knee arthroplasty from the same manufacturer: a randomised controlled trial. *Bone Joint J* 2016;98-B:786-792.

9. **Chau R, Gulati A, Pandit H, et al.** Tibial component overhang following unicompartmental knee replacement–does it matter? *Knee* 2009;16:310-313.

10. **Gudena R, Pilambaraei MA, Werle J, Shrive NG, Frank CB.** A safe overhang limit for unicompartmental knee arthroplasties based on medial collateral ligament strains: an in vitro study. *J Arthroplasty* 2013;28:227-233.

11. **Konan S, Sandiford N, Unno F, et al.** Periprosthetic fractures associated with total knee arthroplasty: an update. *Bone Joint J* 2016;98-B:1489-1496.

12. **Mensch JS, Amstutz HC.** Knee morphology as a guide to knee replacement. *Clin Orthop Relat Res* 1975;112:231-241.

13. **Slamin J, Parsley B.** Evolution of customization design for total knee arthroplasty. *Curr Rev Musculoskelet Med* 2012;5:290-295.

14. **Fitzpatrick C, FitzPatrick D, Lee J, Auger D.** Statistical design of unicompartmental tibial implants and comparison with current devices. *Knee* 2007;14:138-144.

15. **van den Heever DJ, Scheffer C, Erasmus P, Dillon E.** Contact stresses in a patient-specific unicompartmental knee replacement. *Clin Biomech (Bristol, Avon)* 2011;26:159-166.

16. **Kwon OR, Kang KT, Son J, et al.** Importance of joint line preservation in unicompartmental knee arthroplasty: finite element analysis. *J Orthop Res* 2017;35:347-352.

17. **Kim YS, Kang KT, Son J, et al.** Graft extrusion related to the position of allograft in lateral meniscal allograft transplantation: biomechanical comparison between parapatellar and transpatellar approaches using finite element analysis. *Arthroscopy* 2015;31:2380-2391.e2.

18. **Kang KT, Kim SH, Son J, et al.** Probabilistic evaluation of the material properties of the in vivo subject-specific articular surface using a computational model. *J Biomed Mater Res B Appl Biomater* 2017;105:1390-1400.

19. **Peña E, Calvo B, Martinez MA, Palanca D, Doblaré M.** Why lateral meniscectomy is more dangerous than medial meniscectomy. A finite element study. *J Orthop Res* 2006;24:1001-1010.

20. **Haut Donahue TL, Hull ML, Rashid MM, Jacobs CR.** How the stiffness of meniscal attachments and meniscal material properties affect tibio-femoral contact pressure computed using a validated finite element model of the human knee joint. *J Biomech* 2003;36:19-34.

21. **Mesfar W, Shirazi-Adl A.** Biomechanics of the knee joint in flexion under various quadriceps forces. *Knee* 2005;12:424-434.

22. **Takeda Y, Xerogeanes JW, Livesay GA, Fu FH, Woo SL.** Biomechanical function of the human anterior cruciate ligament. *Arthroscopy* 1994;10:140-147.

23. **Blankevoort L, Huiskes R.** Validation of a three-dimensional model of the knee. *J Biomech* 1996;29:955-961.

24. **Steklov N, Slamin J, Srivastav S, D'Lima D.** Unicompartmental knee resurfacing: enlarged tibio-femoral contact area and reduced contact stress using novel patient-derived geometries. *Open Biomed Eng J* 2010;4:85-92.

25. **ConforMIS.** http://www.conformis.com (date last accessed 06 September 2017).

26. **Inoue S, Akagi M, Asada S, et al.** The valgus inclination of the tibial component increases the risk of medial tibial condylar fractures in unicompartmental knee arthroplasty. *J Arthroplasty* 2016;31:2025-2030.

27. **Pegg EC, Walter J, Mellon SJ, et al.** Evaluation of factors affecting tibial bone strain after unicompartmental knee replacement. *J Orthop Res* 2013;31:821-828.

28. **Innocenti B, Truyens E, Labey L, et al.** Can medio-lateral baseplate position and load sharing induce asymptomatic local bone resorption of the proximal tibia? A finite element study. *J Orthop Surg Res* 2009;4:26.

29. **Godest AC, Beaugonin M, Haug E, Taylor M, Gregson PJ.** Simulation of a knee joint replacement during a gait cycle using explicit finite element analysis. *J Biomech* 2002;35:267-275.

30. **Suggs JF, Li G, Park SE, et al.** Function of the anterior cruciate ligament after unicompartmental knee arthroplasty: an in vitro robotic study. *J Arthroplasty* 2004;19:224-229.

31. **No authors listed.** International Organization for Standardization (ISO). ISO 14243-1:2009. Implants for surgery – Wear of total knee-joint prostheses – Part 1: Loading and displacement parameters for wear-testing machines with load control and corresponding environmental conditions for test, 2009. https://www.iso.org/standard/44262.html (date last accessed 06 September 2017).

32. **Baldwin MA, Clary C, Maletsky LP, Rullkoetter PJ.** Verification of predicted specimen-specific natural and implanted patellofemoral kinematics during simulated deep knee bend. *J Biomech* 2009;42:2341-2348.

33. **Kutzner I, Heinlein B, Graichen F, et al.** Loading of the knee joint during activities of daily living measured in vivo in five subjects. *J Biomech* 2010;43:2164-2173.

34. **Halloran JP, Clary CW, Maletsky LP, et al.** Verification of predicted knee replacement kinematics during simulated gait in the Kansas knee simulator. *J Biomech Eng* 2010;132:081010.

35. **Kang KT, Koh YG, Jung M, et al.** The effects of posterior cruciate ligament deficiency on posterolateral corner structures under gait- and squat-loading conditions: A computational knee model. *Bone Joint Res* 2017;6:31-42.

36. **Schwechter EM, Fitz W.** Design rationale for customized TKA: a new idea or revisiting the past? *Curr Rev Musculoskelet Med* 2012;5:303-308.

37. **Patil S, Bunn A, Bugbee WD, Colwell CW Jr, D'Lima DD.** Patient-specific implants with custom cutting blocks better approximate natural knee kinematics than standard TKA without custom cutting blocks. *Knee* 2015;22:624-629.

38. **Martin GM.** In-vivo tibial fit analysis of patient specific TKA system versus off-the-shelf TKA. *JISRF Reconstr Rev* 2014;4:102.

39. **Koeck FX, Beckmann J, Luring C, et al.** Evaluation of implant position and knee alignment after patient-specific unicompartmental knee arthroplasty. *Knee* 2011;18:294-299.

40. **Palmer SH, Morrison PJ, Ross AC.** Early catastrophic tibial component wear after unicompartmental knee arthroplasty. *Clin Orthop Relat Res* 1998;350:143-148.

41. **Setton LA, Mow VC, Howell DS.** Mechanical behavior of articular cartilage in shear is altered by transection of the anterior cruciate ligament. *J Orthop Res* 1995;13:473-482.

42. **Iesaka K, Tsumura H, Sonoda H, et al.** The effects of tibial component inclination on bone stress after unicompartmental knee arthroplasty. *J Biomech* 2002;35:969-974.

43. **Zhu GD, Guo WS, Zhang QD, Liu ZH, Cheng LM.** Finite element analysis of mobile-bearing unicompartmental knee arthroplasty: the influence of tibial component coronal alignment. *Chin Med J (Engl)* 2015;128:2873-2878.

44. **Innocenti B, Pianigiani S, Ramundo G, Thienpont E.** Biomechanical effects of different varus and valgus alignments in medial unicompartmental knee arthroplasty. *J Arthroplasty* 2016;31:2685-2691.

45. **Innocenti B, Bilgen OF, Labey L, et al.** Load sharing and ligament strains in balanced, overstuffed and understuffed UKA. A validated finite element analysis. *J Arthroplasty* 2014;29:1491-1498.

46. **Thompson JA, Hast MW, Granger JF, Piazza SJ, Siston RA.** Biomechanical effects of total knee arthroplasty component malrotation: a computational simulation. *J Orthop Res* 2011;29:969-975.