

Biomechanical Comparison of Kinematic and Mechanical Knee Alignment Techniques in a Computer Simulation Medial Pivot Total Knee Arthroplasty Model

Young Dong Song, MD¹ Shinichiro Nakamura, MD, PhD¹ Shinichi Kuriyama, MD, PhD¹ 
Kohei Nishitani, MD, PhD¹ Hiromu Ito, MD, PhD^{1,2} Yugo Morita, MD¹ Yusuke Yamawaki, MD¹
Shuichi Matsuda, MD, PhD¹

¹ Department of Orthopaedic Surgery, Graduate School of Medicine, Kyoto University, Kyoto, Japan

² Department of Advanced Medicine for Rheumatic Diseases, Graduate School of Medicine, Kyoto University, Kyoto, Japan

Address for correspondence Shinichiro Nakamura, MD, PhD, 54 Shogoin-Kawaharacho, Sakyo-ku, Kyoto 606-8507, Japan (e-mail: shnk@kuhp.kyoto-u.ac.jp).

J Knee Surg

Abstract

Several concepts may be used to restore normal knee kinematics after total knee arthroplasty. One is a kinematically aligned (KA) technique, which restores the native joint line and limb alignment, and the other is the use of a medial pivot knee (MPK) design, with a ball and socket joint in the medial compartment. This study aimed to compare motions, contact forces, and contact stress between mechanically aligned (MA) and KA (medial tilt 3° [KA3] and 5° [KA5]) models in MPK. An MPK design was virtually implanted with MA, KA3, and KA5 in a validated musculoskeletal computer model of a healthy knee, and the simulation of motion and contact forces was implemented. Anteroposterior (AP) positions, mediolateral positions, external rotation angles of the femoral component relative to the tibial insert, and tibiofemoral contact forces were evaluated at different knee flexion angles. Contact stresses on the tibial insert were calculated using finite element analysis. The AP position at the medial compartment was consistent for all models. From 0° to 120°, the femoral component in KA models showed larger posterior movement at the lateral compartment (0.3, 6.8, and 17.7 mm in MA, KA3, and KA5 models, respectively) and larger external rotation (4.2°, 12.0°, and 16.8° in the MA, KA3, and KA5 models, respectively) relative to the tibial component. Concerning the mediolateral position of the femoral component, the KA5 model was positioned more medially. The contact forces at the lateral compartment of all models were larger than those at the medial compartment at >60° of knee flexion. The peak contact stresses on the tibiofemoral joint at 90° and 120° of knee flexion were higher in the KA models. However, the peak contact stresses of the KA models at every flexion angle were <20 MPa. The KA technique in MPK can successfully achieve near-normal knee kinematics; however, there may be a concern for higher contact stresses on the tibial insert.

Keywords

- total knee arthroplasty
- computer simulation
- medial pivot knee
- kinematics
- contact stresses

received
May 29, 2021
accepted after revision
October 22, 2021

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Thieme Medical Publishers, Inc.,
333 Seventh Avenue, 18th Floor,
New York, NY 10001, USA

DOI <https://doi.org/10.1055/s-0041-1740392>.
ISSN 1538-8506.

A problem with total knee arthroplasty (TKA) is the consequent difficulty experienced in restoring normal knee kinematics. To overcome this problem, a medial pivot knee (MPK) design has been developed to restore normal knee kinematics, with ball-and-socket conformity on the medial compartment and a flat unconstrained lateral tibial surface, allowing anteroposterior (AP) translations of the lateral condyle to permit rotational freedom of the joint.¹ These contrasting conformities were hypothesized to lead to more normal knee kinematics.^{2,3} Although many clinical studies on MPK^{4–6} have shown excellent results reflecting normal kinematic characteristics, contrasting results have been found in terms of survival rate,⁷ clinical outcomes,⁸ and postoperative flexion, compared with posterior-stabilized (PS) TKA.⁹

A kinematically aligned (KA) TKA technique has been proposed to improve patient satisfaction by restoring normal knee kinematics through the restoration of the native cartilage thickness and soft tissue tension, thereby retaining the native joint lines and limb alignments specific to each patient.^{10–12} The femoral and tibial resection is made with more valgus and varus, respectively, compared with a mechanically aligned (MA) technique. Consequently, KA TKA often has a varus joint line relative to MA TKA.¹³ In a biomechanical analysis of the effects of the KA TKA technique with conventional cruciate-retaining (CR)¹⁴ and PS¹⁵ implants, greater posterior movement and external rotation of the femoral component were observed in the medial joint line tilt 3° [KA3] and 5° [KA5] model than in MA TKA. This may explain the better clinical results and patient satisfaction observed in previous clinical studies.^{16–19}

Although both the KA technique and MPK have been developed to restore normal knee kinematics and to improve patient satisfaction after TKA, it is difficult to predict the outcome of the simultaneous use of both methods, with each having its own risks. Therefore, combining these approaches does not always guarantee better results. Varus alignment of the tibial component leads to accelerated polyethylene wear, leading to early revision.^{20,21} For that reason, there remains a need to ensure the safety of the varus alignment of the tibial component because of a risk of failure including aseptic loosening. However, recent studies^{18,22,23} demonstrated a negligible incidence of component failure rate from KA TKA, substantiated during a 10-year follow up study. Unfortunately, a quantitative kinetic impact of the joint-line tilting in MPK remains unclear, unlike CR and PS implants. Furthermore, no previous study has performed a comparative kinematic and kinetic analysis in which the joint-line tilting of MPK varies during deep-knee-bend activities.

The aims of this study were to compare motions, contact forces, and the stresses of the two KA models (tilt 3° [KA3] and 5° [KA5]) with those of the MA (tilt 0°) model in MPK, and to determine the benefits and risks of the KA technique in MPK. It was hypothesized that the KA3 and KA5 models would reveal near-normal knee kinematics compared with the MA model during deep-knee-bend activities.

Methods

A musculoskeletal computer simulation was used to evaluate the results of various statuses of the joint-line tilting. This model provides dynamic simulation of the knee (LifeMOD/KneeSIM 2010; LifeModeler Inc., San Clemente, CA), including tibiofemoral and patellofemoral contact, lateral collateral ligament (LCL), medial collateral ligament (MCL), elements of the knee joint capsule, quadriceps muscle and tendon, patellar tendon, and hamstring muscles (–Fig. 1). The KneeSIM program uses implant geometry to analyze the performance of the femoral, tibial, and polyethylene inserts under a variety of conditions. This program has been validated in previous biomechanical studies and is able to estimate individual in vivo knee kinematics and contact forces.^{15,24,25} The LCL is a single fiber bundle, whereas the MCL consists of anterior and posterior bundles.^{26–28} All ligament bundles were modeled as nonlinear springs with material properties based on a previous report.²⁹ The proximal attachment points of the LCL and MCL were defined as the most prominent points of the femoral epicondyles. Their distal attachment points were defined as the tip of the fibular head and the midpoint between the tibial attachments of the anterior and posterior bundles, respectively. The stiffness coefficients of the LCL, MCL anterior bundle, and MCL posterior bundle were determined to be 59, 63, and 63 N/mm, respectively. The initial strain of each ligament was determined on the basis of results of previous cadaver studies.^{27,30,31}

In the current study, the model parameters for a fixed-bearing, cruciate-sacrificing prosthesis (GMK Sphere; Medacta International AG, Castel San Pietro, Switzerland) were imported into the program and tested during a simulated weight-bearing deep knee bend, as described previously.^{32,33} The medial compartment is constrained in a medial tibial insert with a 1:1 bone and socket ratio, while the flat lateral

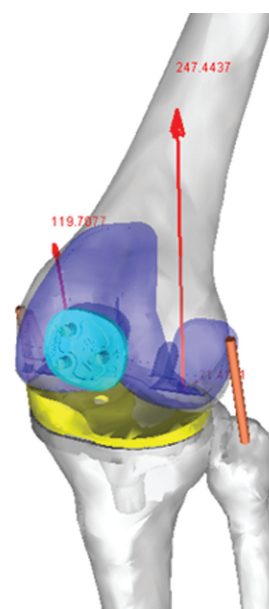


Fig. 1 The KneeSIM model. Red arrows indicate the direction of tibiofemoral contact forces, and orange bars indicate the MCL and LCL. LCL, lateral collateral ligament; MCL, medial collateral ligament.

tibial side with lateral femoral components allows more posterior translations of the femoral component and achieves larger external rotation. Implantation was performed using virtual bones of the normal knee model. The implant was size 5 (overall AP length: 69 mm, overall mediolateral [ML] length: 72 mm) for the femoral component and size 4 (AP length: 47 mm, ML length: 74 mm) for the tibial insert, determined to match the size of the virtual bones in the model.

For the MA TKA model, the femoral component was aligned perpendicular to the mechanical axis of the femur in the coronal plane and parallel to the distal anatomical axis of the femur in the sagittal plane. The tibial component was aligned perpendicular to the mechanical axis of the tibia with a 3° posterior tibial slope in the sagittal plane. The rotational alignments of the femoral and tibial components were determined according to the femoral trans-epicondylar axis and the Akagi line,³⁴ respectively. Since individual knee anatomies are heterogeneous, two coronal joint-line tilting models were created to embody the surgical goal of the KA technique for various morphologies. In comparison with the MA model, the joint obliquity was determined to be 3° and 5° based on clinical data.^{12,16,35–37} In the coronal plane, the femoral component in the KA3 and KA5 models was tilted at 3° and 5° at the valgus. Correspondingly, the tibial component in the KA3 and KA5 models was also tilted at 3° and 5° at the varus (►Fig. 2). In the axial plane, the femoral component in both the KA3 and KA5 models had the same condition, parallel to the posterior condylar axis of the femur (►Fig. 3). The tibial component in both the KA3 and KA5 models was set parallel to the AP axis of the elliptical-shaped lateral tibial condyle (►Fig. 4).³⁸

During movement, the hip joint was allowed to flex and extend to slide vertically, whereas the ankle joint was allowed free translation in the ML direction, and free varus–valgus and axial rotation. A 4,000-N load was applied at the hip, the active driving elements of which were the forces

of the quadriceps and hamstring muscles. The simulation was driven by a controlled actuator arrangement similar to a physical machine. A closed-loop controller applied tension to the quadriceps and hamstrings to match the firing of a prescribed flexion angle at each point. The co-contraction between these muscles was defined. The models were subjected to a 4.5 second-cycle of a squat motion (0–130° flexion).

Kinematics were analyzed in the AP positions, ML positions, and measured external rotation angles of the femoral components using the tibial component coordinate system. We used both flexion facet centers as the reference system of the femur. Flexion facet centers were set as the center of circles to fit the articular surface of the posterior condyles in the sagittal plane.³⁹ The AP positions of the medial flexion facet center (MFFC) and the lateral flexion facet center (LFFC) were evaluated. Concerning the ML positions, the medial end of the tibial insert was set to zero position and ML translation

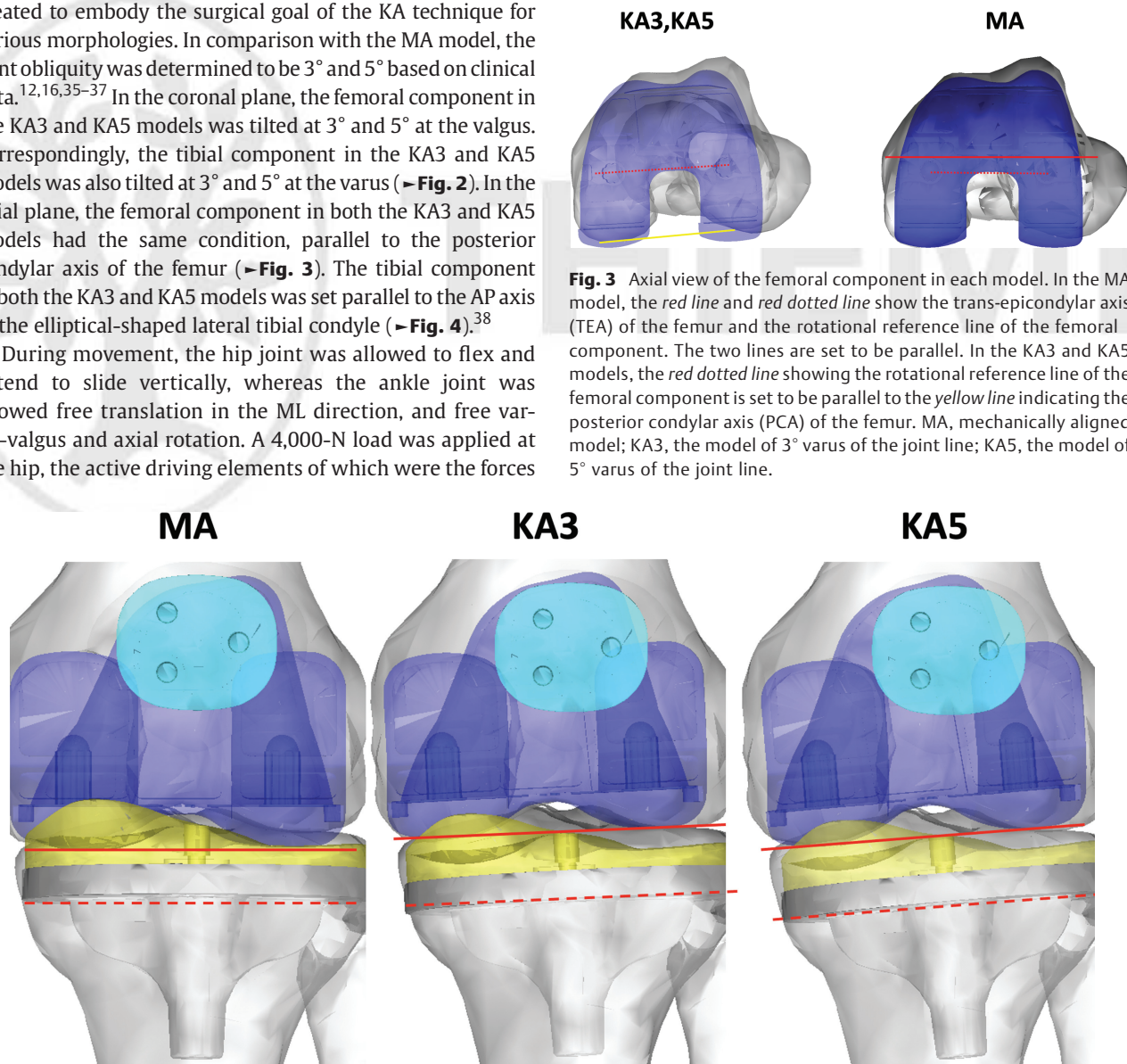


Fig. 2 Coronal view of each model. Red line indicates alignment of the femoral component and red dotted line indicates alignments of the tibial component. MA, mechanically aligned model; KA3, the model of 3° varus of the joint line; KA5, the model of 5° varus of the joint line.

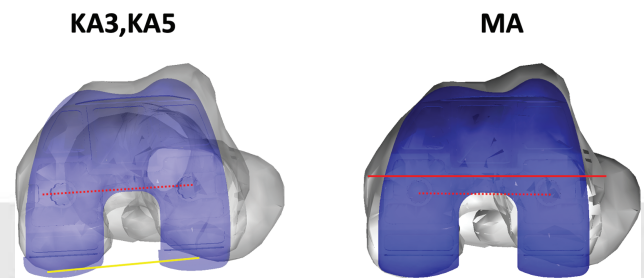


Fig. 3 Axial view of the femoral component in each model. In the MA model, the red line and red dotted line show the trans-epicondylar axis (TEA) of the femur and the rotational reference line of the femoral component. The two lines are set to be parallel. In the KA3 and KA5 models, the red dotted line showing the rotational reference line of the femoral component is set to be parallel to the yellow line indicating the posterior condylar axis (PCA) of the femur. MA, mechanically aligned model; KA3, the model of 3° varus of the joint line; KA5, the model of 5° varus of the joint line.

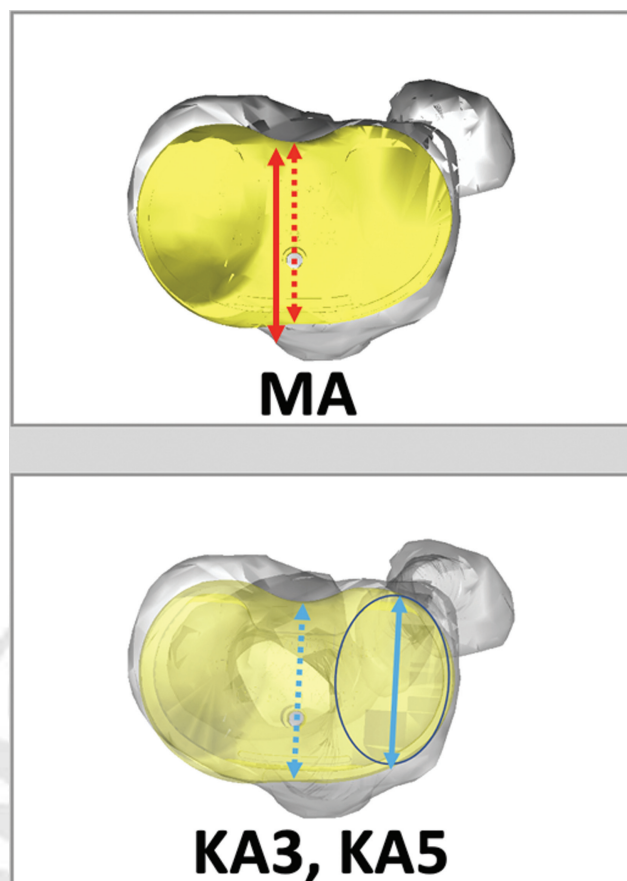


Fig. 4 Axial view of the tibial component. In the MA model, the red dotted line is the AP axis of the tibial component, which is parallel to the Akagi line (red straight line). In the KA3 and KA5 models, the blue elliptical line shows the boundary of the articular surface of the lateral tibial condyle, and the blue straight line is the long axis connecting the most anterior and distal points on the boundary. The blue dotted line is the AP axis of the tibial component, which is parallel to the blue straight line in KA3 and KA5. AP, anteroposterior; MA, mechanically aligned model; KA3, the model of 3° varus of the joint line; KA5, the model of 5° varus of the joint line.

was defined as the horizontal distance from the medial end of the tibia to the MFFC (►Fig. 5). The axial rotations of the femoral component were determined relative to the tibial component. The anterior position for the AP positions, lateral position for the ML positions, and external rotation for the axial rotations were denoted as positive in the kinematic analyses. The tibiofemoral contact force at the medial and lateral compartments was analyzed.

The contact stress at the tibial inserts against the femoral component was calculated with a three-dimensional finite element (FE) analysis using Mechanical Finder version 11.0 (Research Centre for Computational Mechanics Inc., Tokyo, Japan). We aligned the position of each component using a solid modeling software (Solid Edge; Siemens AG, Munich, Germany) based on the simulation program. The magnitude and direction of each force determined with the simulation program were also used to calculate the contact stress.^{14,15,25} The stress at the component interfaces was examined at 0°, 30°, 60°, 90°, and 120° of knee flexion. Meshes of the femoral component and the tibial insert were generated based on a

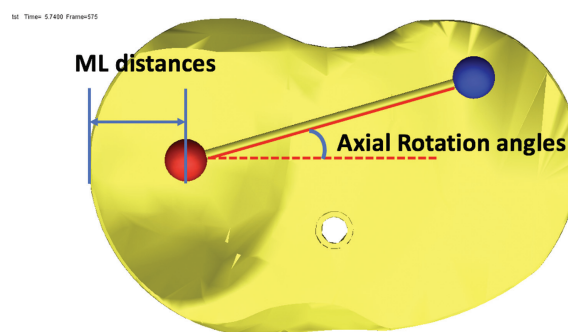


Fig. 5 ML distance and axial rotation angle are measured in the axial plane image from KneeSIM. ML distance is measured as the horizontal length between the medial end of the tibial insert and the MFFC (center of the red ball), with a positive value indicating a lateral position of the femoral component. For the axial rotation angle, we measured the angle determined by a connecting line (the red line) between the MFFC and LFFC (center of the blue ball) and the ML axis (the red dotted line) of the tibial insert, with a positive value indicating external rotation of the femoral component. LFFC, lateral flexion facet center; MFFC, medial flexion facet center; ML, mediolateral.

10-node quadratic tetrahedral element by setting the mesh size between 4 mm (maximal) and 1 mm (minimal). The generated mesh contained a total of 30,635 nodes as a result of 140,762 solids. The elastic modulus of the femoral component (Co–Cr–Mo alloy) was set to 240 GPa and Poisson's ratio was set to 0.3, whereas the values for the tibial insert were 650 MPa and 0.46, respectively. The coefficient of friction was set to 0.1 under contact conditions between the tibial insert and femoral component, following a previous study.⁴⁰

Results

All three models exhibited similar patterns of AP movement of the MFFC. During knee flexion, the AP movement of the MFFC from 0° to 120° of knee flexion was 0.2 mm (MA), 1.2 mm (KA3), and –4.8 mm (KA5). However, the LFFC showed different results from the medial compartment. The AP movement of the LFFC was –0.3, –6.8, and –17.7 mm in the MA, KA3, and KA5 models, respectively (►Fig. 6A). Concerning the ML positions, MFFC in MA and KA3 models showed similar positions during whole knee flexion relative to the medial edge of the tibial insert, whereas MFFC in the KA5 model was positioned medially (►Fig. 6B and C). Concerning the rotation, the axial rotation from 0° to 120° was 4.2°, 12.0°, and 16.8° in the MA, KA3, and KA5 models, respectively (►Fig. 6D).

In all three models, from 0° to 40°, the medial tibiofemoral contact forces were greater than those in the lateral compartment; however, from 50° to 120°, the lateral tibiofemoral contact forces were greater than the medial tibiofemoral contact forces (►Fig. 6E). Concerning the contact stress in FE analyses, the peak contact stresses from 0° to 60° of knee flexion were almost similar among all models. At 90° and 120° of knee flexion, the contact stresses in the KA3 and KA5 models were higher than the contact stress in the MA model (►Table 1, ►Fig. 7). However, the peak contact stress of the KA3 and KA5 models was <20 MPa at all flexion phases.

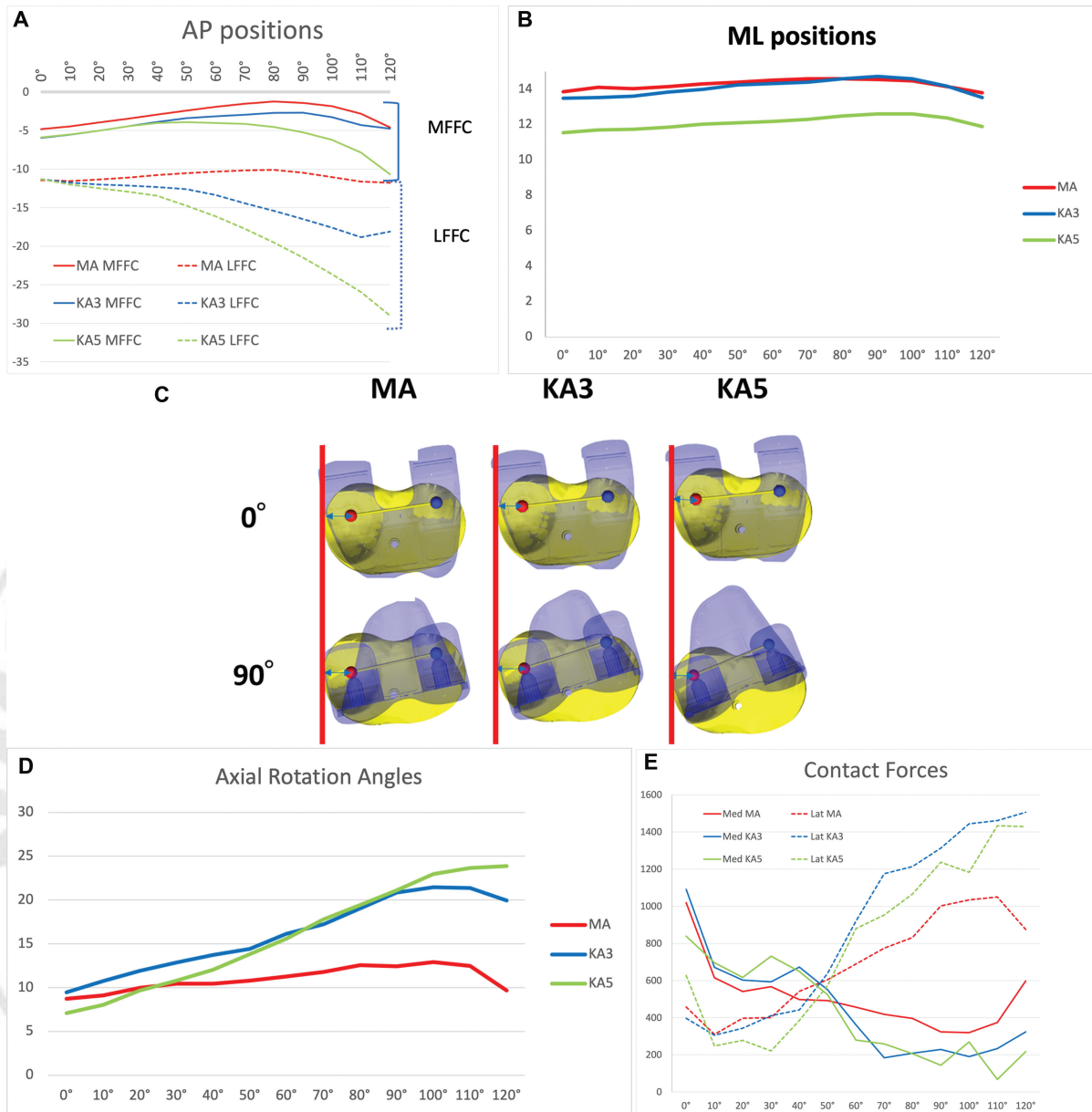


Fig. 6 Diagrams showing the kinematic and kinetic results from 0° to 120° of knee flexion in the three models (red lines: MA model, blue lines: KA3 model, green lines: KA5 model). (A) AP positions of the MFFC and LFFC (y-axis, mm) plotted against the knee flexion angles (x-axis, degree) during deep knee bends. (B) ML positions of the MFFC (y-axis, mm) plotted against the knee flexion angles (x-axis, degree) during deep knee bends. (C) MFFC (red dots) in MA and KA3 models shows similar positions in 0° and 90° of knee flexion relative to the medial edge of the tibial insert (red lines), whereas MFFC in the KA5 model is positioned medially. (D) Axial rotation angles of the femoral component (y-axis, degree) plotted against the knee flexion angles (x-axis, degree) during deep knee bends. (E) Diagrams showing the tibiofemoral contact forces (y-axis, N) from 0° to 120° of knee flexion (x-axis, degree) in the three models. AP, anteroposterior; LFFC, lateral flexion facet center; MFFC, medial flexion facet center; ML, mediolateral.

Table 1 The peak contact stress of the three models during deep knee bend

		0°	30°	60°	90°	120°
MA	Med (MPa)	2.71	4.17	2.08	2.4	6.05
	Lat (MPa)	4.14	5.82	6.06	4.61	5.63
KA3	Med (MPa)	2.23	1.91	4.24	3.25	11.67
	Lat (MPa)	6.67	3.59	7.11	12.21	13.08
KA5	Med (MPa)	2.15	2.07	7.33	10.83	15.71
	Lat (MPa)	3.99	3.76	7.77	16.44	17.01

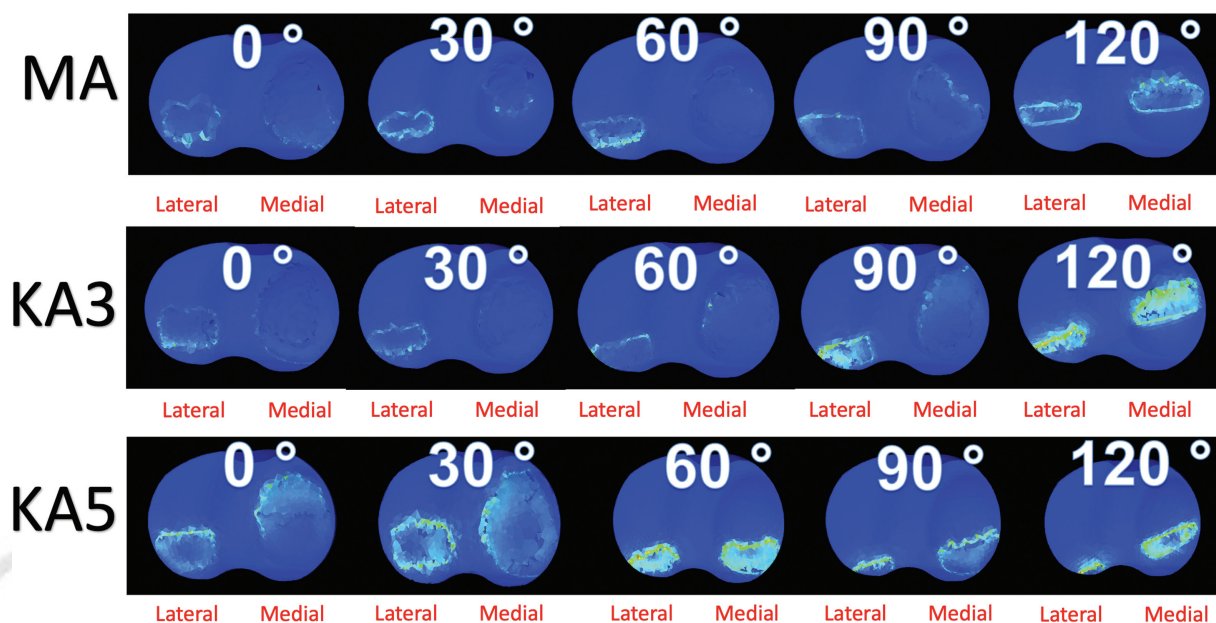


Fig. 7 Contact stresses on the tibiofemoral joint at 0°, 30°, 60°, 90°, and 120° of knee flexion in the MA, KA3, and KA5 models. The contact area decreases with increasing knee flexion. The peak contact stress at 60° and more of knee flexion is higher in the KA models than in the MA model. KA, kinematically aligned; MA, mechanically aligned model.

Discussion

The most important finding of the present study is that the quantity of lateral posterior movement and external rotation of the femoral component of the KA models were greater than those of the MA model in MPK using a computer simulation model. In terms of kinetics, lateral contact forces and stresses were greater at >90° of knee flexion in the KA models, although the peak contact stress of the KA models was 20 MPa. The joint line tilting in MPK was able to achieve near-normal knee kinematics with an acceptable increase in contact force and contact stress on the tibial insert.

The kinematics of normal knees has almost been identified by previous *in vivo* and *in vitro* experiments in terms of AP movement and axial rotation. A static magnetic resonance imaging study reported that the lateral femoral condyle moved posteriorly by 22 mm and the femur rotated externally through an angle of 20° from hyperextension to 120° in normal knees.⁴¹ Similar results were observed in a fluoroscopic imaging study.⁴² The current study showed similarities in terms of the directions of AP movement and axial rotation, indicating comparable postoperative kinematics of MPK in KA TKA with a normal knee. However, the magnitudes of the AP movement of the lateral condyle and the axial rotation were relatively less than those in the normal knee. One of the possible reasons for the discrepancy in magnitude is the existence of the screw-home mechanism.

Previous computer simulation studies^{14,15} reported that KA TKA with conventional CR and PS implants had better near-normal knee kinematics than MA TKA. In this study, kinematic results of KA models with the MPK system are evidently in agreement with previous results. Recently, an *in*

vivo fluoroscopic study⁴³ reported that CS TKA designs showed 2.5 ± 2.0 mm AP movement of the lateral condyle, 2.5 ± 2.2 mm AP movement of the medial condyle, and $6.6 \pm 3.3^\circ$ axial rotation during weight-bearing deep knee flexion. These are similar to the results obtained in the MA model of the current study. In addition, the kinematic results of the KA3 model in this study were similar to those of other studies. Similar kinematic results were also reported in the study by Alesi et al,⁴⁴ in which MPK with an adjusted MA technique and slight varus on the coronal deformity showed greater anterior translation of the lateral condyle (7.1 ± 0.6 mm) than that of the medial condyle (2.9 ± 0.7 mm) in the sit-to-stand task. Moreover, Kono et al⁴⁵ reported kinematic data from bicruciate-stabilized (BCS) TKA, with built-in joint-line tilting, in which a 3.1 ± 1.3 mm medial rollback and an approximately 10 mm lateral rollback during a deep-standing squat were observed on fluoroscopic analysis. These findings suggest that the effect of joint-line tilting with MPK tends to provide predictions that are parallel to either the adjusted MA technique with MPK or BCS TKA. This could be explained by the asymmetric conformity of the tibial insert and the medial tilt of the joint line. Moreover, the results obtained in the current study confirm another *in vivo* kinematic analysis that showed that a varus alignment of the limb and tibial component results in a greater axial rotation from full extension to maximum flexion.⁴⁶

Tibiofemoral contact forces are a crucial issue in the design and clinical outcomes of TKA. These forces strongly affect the wear and stress distribution of the implant and the implant–bone interface. Previous studies reported that the net tibiofemoral joint contact forces became greater at higher

flexion angles during the deep knee bend of the normal knee.^{47–49} These studies used motion analysis and ground reaction forces to calculate the forces and moments around the knee joint. Recently, new computer simulation studies with normal knee⁵⁰ and a PS TKA prosthesis⁵¹ have shown similar trends to previous findings, like the current study on MPK. However, in the current study, the lateral contact force and stress on the tibiofemoral joint at $>60^\circ$ of knee flexion were higher than those applied on the medial compartment in all three models. Moreover, the force of the KA models was higher than that of the MA model, which was inconsistent with the results reported by previous computer simulation studies on CR¹⁴ and PS¹⁵ implants. This may be related to the different conformities of polyethylene in MPK. In addition, the medially positioned femoral component in KA models relative to the tibial component due to medial slope of the tibial component in the coronal plane resulted in a reduced adduction moment, which, in turn, increased the lateral compartment contact force and decreased the medial compartment contact force. This may explain the increased tibiofemoral contact stress observed in the deep knee bend in the MPK system.

Neutral mechanical alignment has long been considered a goal for a successful TKA. It has generally been accepted that malalignment causes increased loads on the implant and implant–bone interface, leading to implant failures. In previous biomechanical studies,⁵² malalignment of the tibial component was associated with more detrimental effects than malalignment of the femoral component. Ritter et al²¹ retrospectively reviewed 6,070 knees in 3,992 patients and reported a failure rate of 3.2% for the tibial varus and 7.8% for the femoral valgus, percentages which were higher than the rate of 0.2% for neutral alignment. However, in recent years, several clinical studies^{53–55} have suggested that these alignment issues may not matter in the era of modern TKA because of improvements in the properties of materials, including polyethylene. In this study, the lateral contact stress at 90° and 120° of knee flexion was approximately two to three times higher in the KA models than in the MA model. This is similar although lower than the results reported by previously published computer simulation studies.^{14,15} This discrepancy may be due to the design characteristics of the MPK system, which has a larger contact area on the medial side, leading to lower contact stress. However, it remains unclear whether higher shear forces due to a higher AP constraint from a larger contact surface can lead to earlier loosening of the tibial component. When considering the results from other computer simulation studies using the MPK system⁵⁶ and the strength of polyethylene inserts, the contact stress in KA models might be generally acceptable.

This study has several important limitations. First, only one knee model was used and KA, using the calipered technique, was not completely implemented in this study. However, we implanted the components to represent the KA technique to restore the native joint lines and limb alignment. We have been conservative in outlining the kinematics and kinetic effect of the joint-line tilting in the coronal plane. The second limitation is the use of only one MPK design. Dennis et al⁵⁷ reported that different models of the same

concept can have different knee kinematics. However, the focus of the current study was on the effects of the joint-line tilting; hence, it was important to restrict the analysis to one specific implant to reduce the bias from different model designs. Third, the FE analysis was not a dynamic analysis. The contact stress calculated from this study may not necessarily reflect numerical values in vivo because of the lack of validation of the FE analysis in this study. Finally, it is difficult to represent motion, contact forces, and contact stress in a normal knee completely in a computer simulation,⁵⁸ and the use of a normal knee model rather than an arthritic knee model is another limitation. The major problems in creating a computer simulation model of a normal knee and an arthritic knee are the (1) anatomical complexity of the cartilage surface, and specifically worn cartilage, (2) difficulty in measuring the in vivo kinetics and kinematics of the knee, and (3) unpredictable influences from various ligaments and muscles. However, our results provide compelling evidence that fine adjustments were made to the stiffness, length patterns, and slack of each ligament, based on the relevant anatomical literature and previously validated kinematics and kinetics of implanted and normal knees.

Conclusions

The medial tilt of the joint line in MPK may exhibit greater posterior movement and external rotation of the femoral component with increased lateral contact forces and stresses using a computer simulation model and an FE analysis. These results suggest that utilizing the KA technique in MPK is of use in achieving near-normal knee kinematics, with slight detrimental effects, such as higher contact stress, on the tibial insert.

Conflict of Interest

None declared.

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THIEME