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Title: Biomechanical Study of Bone Defects and Unicompartamental Prosthesis
Implantation Position in Spontaneous Osteonecrosis of the Knee

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Abstract

Background

Finite element analysis was used to study the effects of the anterior load-bearing column of the femoral component and bone defects at different locations on stress changes in the internal structures of the knee joint in fixed-bearing unicompartmental knee arthroplasty (FB-UKA). On this basis, the critical value of bone defect depth in UKA for treating SONK was further analyzed, providing a theoretical basis for clinicians to choose between UKA and TKA.

Methods

A validated normal knee model was established. An elliptical bone defect (major axis 6 mm, minor axis 4 mm) was created on the weight-bearing area of the medial femoral condyle. Two groups of finite element models were constructed. Group 1 evaluated the effect of defect depth: with alignment unchanged, defect depths were set to 1/3, 2/3, and 3/3 of the anterior load-bearing pillar length. Group 2 evaluated the effect of medial translation: defect depth was fixed at full pillar length, and the femoral component was translated medially by 1 mm, 3 mm, and 5 mm from the defect center. In total, seven distinct UKA models with bone defects were established. Peak von Mises stresses on the femoral component, polyethylene liner, tibial component, and both femoral-side and tibial-side cement mantles were analyzed.

Results

As defect depth increased, peak stresses on all components and cement mantles exhibited an upward trend. However, when the defect depth was fixed at full pillar length and the component was translated medially by 3 mm, the peak stress on the femoral-side cement increased markedly, exceeding the typical compressive strength of polymethyl methacrylate (PMMA) bone cement (approximately 110 MPa).

Conclusions

For FB-UKA in SONK, when the defect depth does not exceed the full pillar length

and medial translation is ≤ 3 mm, cement mantle fatigue, postoperative loosening, and dislocation of the femoral component can be reduced. This recommended range improves long-term prosthesis survival and maintains overall knee stability.

Keywords: Spontaneous osteonecrosis of the knee; Unicompartmental knee arthroplasty; Finite element analysis; Biomechanics

Background

Spontaneous Osteonecrosis of the Knee (SONK) was first systematically described by Ahlbäck et al.^[1] in 1968 as a spontaneous subchondral bone necrosis independent of trauma or steroid use, predominantly affecting middle-aged women^[2]. In SONK, about 94% of cases are confined to a single compartment (especially the medial compartment), while the remaining compartments are relatively normal^[3]. This provides conditions for the application of unicompartmental knee arthroplasty (UKA). Different from total knee arthroplasty (TKA), UKA retains healthy structures such as articular cartilage and soft tissue, which helps patients obtain proprioception closer to the physiological state, faster functional rehabilitation and higher subjective satisfaction after surgery^[4].

Among UKA prostheses, based on bearing design, there are mobile-bearing (MB-UKP) and fixed-bearing (FB-UKP) types. FB-UKP is more widely used due to its simpler design, easier surgical technique, and stable liner position with less risk of subluxation^[5, 6].

Although UKA is an option for collapsed SONK, its clinical use remains limited^[7]. The low prosthesis survival rate is the most prominent issue restricting the widespread adoption of UKA in orthopedic clinical practice. Studies have shown that the 10-year survival rate of UKA prostheses is generally 80%–85%, which is lower than that of TKA prostheses^[8, 9]. Among the various reasons leading to UKA failure requiring revision surgery, aseptic loosening of the prosthesis, wear of the polyethylene liner, and progression of osteoarthritis in the contralateral compartment rank as the top three^[10, 11]. Among these, aseptic loosening of the prosthesis and the subsequent dislocation or subluxation that may result are among the most common

causes of early postoperative revision.

From a biomechanical perspective, the primary reason for the susceptibility of the femoral component to loosening in SONK patients can be attributed to fatigue fracture of the bone cement^[12]. Bone cement is not a true adhesive but rather a space filler used to achieve mechanical interlocking. When the depth of the bone defect increases or when the load-bearing column at the anterior edge of the femoral component becomes misaligned relative to the bone defect, the stress distribution and concentration within the cement mantle, the prosthesis, and the surrounding bone are significantly altered. Under long-term cyclic loading, these localized high-stress regions accelerate the initiation and propagation of microcracks in the cement material, ultimately leading to fatigue fracture of the bone cement^[13]. Therefore, the depth of the bone defect and the relative positional relationship between the bone defect and the femoral component are key factors that determine whether the postoperative stress distribution within the knee joint is reasonable and whether the bone cement can maintain its integrity over the long term^[14, 15].

Currently, finite element studies on FB-UKP are scarce, and stress changes in knee structures under different bone defects remain unclear^[14, 16].

To investigate the biomechanical effects of defect depth and relative position changes on FB-UKA, this study used finite element analysis to establish two high-fidelity models: Model 1 evaluated the effect of defect depth (1/3, 2/3, 3/3 of pillar length); Model 2 simulated medial translation of the femoral component relative to the defect center (1 mm, 3 mm, 5 mm). The study focused on stress changes in the femoral component, polyethylene liner, tibial component, and cement mantles. The aim is to provide theoretical guidance for FB-UKA in SONK: (1) determine safe defect depth thresholds based on stress limits; (2) establish a safe implantation range for the femoral component relative to the defect.

Materials and methods

Establishment of a normal knee joint model

Imaging data were obtained from a 37-year-old female volunteer with no history

of musculoskeletal disease. Anteroposterior and lateral X-rays of the right knee and full-length standing X-rays of both lower limbs showed no abnormalities, with symmetrical tibiofemoral joint spaces and no pathological changes or developmental anomalies. A 64-slice computed tomography (CT) scanner (Siemens, Germany) was used for full-length lower limb scanning with a slice thickness of 0.625 mm. A 3.0T magnetic resonance imaging (MRI) scanner (Philips, Netherlands) was used for sagittal scanning of the right knee (slice thickness 1 mm), covering 15 cm of the distal femur and proximal tibia.

The images were imported into Mimics 21.0 software (Materialise, Belgium) to create masks for the knee joint bones, cartilage, ligaments, menisci, and other soft tissues. After three-dimensional reconstruction, the masks were smoothed, and all masks were aligned and registered to form a complete lower limb model. Meanwhile, the lateral distal femoral angle was measured to be approximately 84° , and the medial proximal tibial angle was approximately 92° , both within the normal range. A region 15 cm above and below the center of the right knee joint was selected. In Geomagic Wrap 2021 software (3D Systems, USA), each structural model underwent noise reduction, smoothing, and surface fitting, and was saved as an STP file. Subsequently, the models were imported into Creo 7.0 software (PTC, USA) to construct solid models of each structure, which were assembled into a complete knee joint model (Figure 1), optimized, and saved as an STP file. This STP file was then imported into Ansys Workbench 2024 software (ANSYS, USA) for mesh generation, material property assignment, and boundary condition setting. All structures were meshed using 10-node tetrahedral elements, followed by convergence verification, with the convergence criterion defined as a change of less than 5% between two adjacent mesh refinements. According to previous studies^[17, 18], ligaments were defined as hyperelastic isotropic materials, represented by the neo-Hookean model^[16, 18]. The remaining structures were defined as linear elastic isotropic materials with corresponding parameters assigned (Table 1). A total of six contact pairs were established: femoral cartilage–tibial cartilage, tibial cartilage–meniscus, and femoral

cartilage–meniscus. Based on the anatomical characteristics of the knee joint and previous research methods^[19], the bone–ligament connections, bone–cartilage connections, and tibial cartilage–meniscus connections were all set as bonded contacts; whereas, frictionless finite-sliding surface contacts were used between the femoral cartilage and meniscus, and between the femoral cartilage and tibial cartilage, to simulate their relative motion. The boundary conditions were set with the femur and tibia in relative extension, constraining only the flexion–extension movement of the femur while leaving other directions unconstrained. The distal tibia and fibula were fully constrained^[20], thereby establishing a normal knee joint model.

Tab. 1 Material property assignments for knee joint structures

Structures	Young' s modulus (MPa)	Poisson' s ratio (u)
Cortical Bone	17000	0.30
Cancellous Bone	350	0.25
Meniscus	27.5	0.33
Articular Cartilage	15	0.46
Patellar Ligament	87	0.30
Anterior Cruciate Ligament	116	0.30
Posterior Cruciate Ligament	87	0.30
Medial Collateral Ligament	48	0.30
Lateral Collateral Ligament	48	0.30
Cobalt-Chromium- Molybdenum Alloy	210000	0.29
Ultrahigh-Molecular-Weight Polyethylene	850	0.40



Fig. 1 Complete knee model

Construction of Finite Element Models for Different Positional Relationships between Bone Defect and Femoral Component in Fixed-Bearing Unicompartmental Knee Arthroplasty

By measuring the right knee joint of the subject and referencing the dimensions of the Link-Sled® fixed-bearing prosthesis (LINK, Germany), the final selected femoral component dimensions were: width 18 mm, length 52 mm; tibial component dimensions were: width 25 mm, height 9 mm, diameter 50 mm. A SIMSCAN GEN2 handheld 3D scanner (Scantech Co., Ltd., China) was used to scan this prosthesis model. After obtaining the point cloud data of the prosthesis surface, post-processing and surface fitting optimization were performed. A solid model of the prosthesis was then established in Creo 7.0 and imported into the normal knee joint model. Boolean operations were used to assemble the prosthesis with the knee joint. In accordance with UKA surgical standards, the following principles were followed: the sagittal posterior tibial slope was set at 5° ^[21]; the coronal alignment was neutral, i.e., 0° varus/valgus; the horizontal tibial osteotomy thickness was approximately 7 mm; the tibial component was positioned to cover the osteotomy surface as much as possible, perpendicular to the tibial component in the coronal plane with no axial rotation^[14]; the femoral component was positioned to cover the central area of the medial femoral condyle, wrapping around the distal portion of the posterior femoral condyle as much as possible. After implantation, the joint line height remained the same as before osteotomy, with a 1 mm bone cement layer between the prosthesis and the osteotomy

surface. The trimmed and smoothed solid prosthesis model was then imported into Ansys Workbench 2024 software. Tetrahedral elements were used for meshing. The metal part was made of cobalt-chromium-molybdenum alloy, the bearing part was ultra-high molecular weight polyethylene, and the bone cement was polymethyl methacrylate. All three were defined as linear elastic isotropic materials, with parameters assigned according to previous studies (Table 1). Thus, a UKA finite element model with the femoral component in the standard position was established (Figure 2).

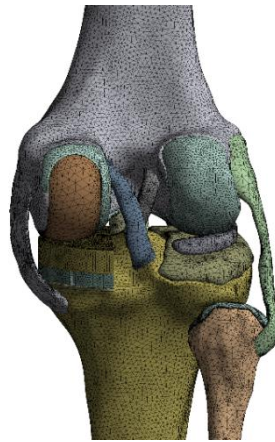


Fig. 2 Three-dimensional finite element model of UKA with the prosthesis in the standard position

An elliptical bone defect with a major axis of 6 mm and a minor axis of 4 mm was created on the weight-bearing surface of the medial femoral condyle. Using the intact model with the prosthesis in the standard position as a baseline, the anterior load-bearing peg of the femoral component was aligned with the center of the bone defect, and the following two groups of UKA models were constructed:

Group 1: To evaluate the effect of bone defect depth. Keeping the alignment between the center of the anterior load-bearing peg of the femoral component and the center of the bone defect unchanged, three models were sequentially established with bone defect depths corresponding to 1/3, 2/3, and 3/3 of the peg length (Fig. 3).

Group 2: To evaluate the effect of prosthesis position changes. The bone defect depth was fixed at the full length of the peg. Using the center of the bone defect as the starting reference point, the femoral component was translated medially by 1 mm, 3 mm, and 5 mm, respectively, to construct three models (Fig. 4).

In summary, a total of 7 different UKA finite element models with bone defects were established in this study. Bonded contact was applied at both the prosthesis–cement interface and the cement–osteotomy surface interface to simulate the stable postoperative state. A frictionless surface contact condition with finite sliding was set between the femoral component and the polyethylene liner to simulate the low-friction characteristics under joint fluid lubrication. The contact relationships for the remaining structures were the same as those in the normal knee joint model. The boundary conditions were set to the relative extension position of the femur and tibia. Only the flexion-extension movement of the femur was constrained, while movements in other directions were not constrained. The distal tibia and fibula were fully constrained.

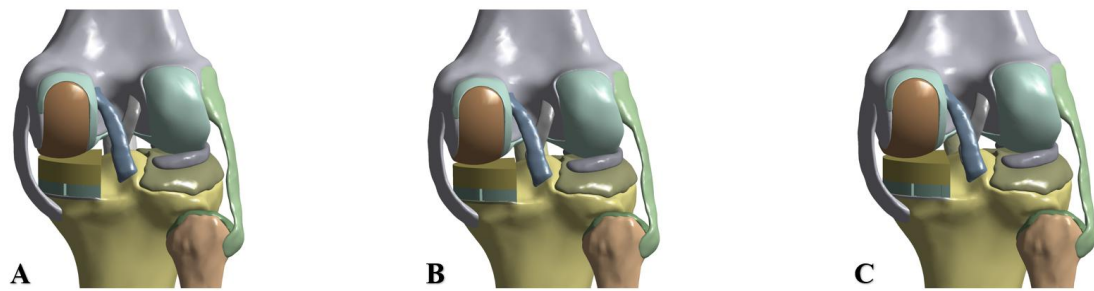


Fig. 3 UKA finite element models with different bone defect depths. (A–C) Bone defect depths corresponding to 1/3, 2/3, and 3/3 of the length of the anterior load-bearing pillar of the femoral component, respectively.

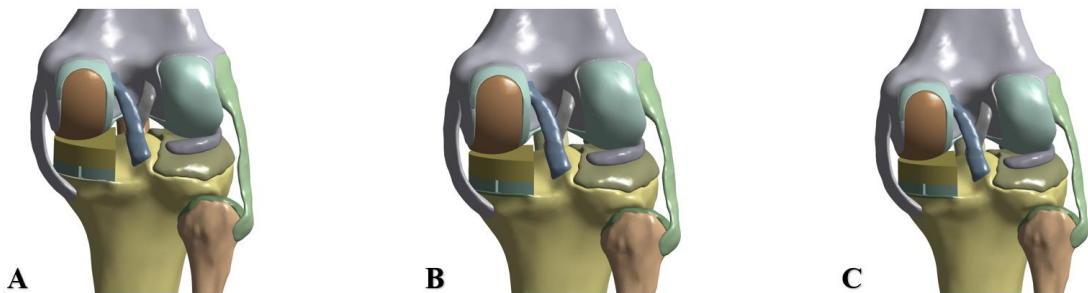


Fig. 4 UKA finite element models with different medial translation distances of the femoral component. (A–C) The anterior load-bearing pillar of the femoral component is translated medially from the center of the bone defect by 1 mm, 3 mm, and 5 mm, respectively.

Stress loading and model validation

To ensure the accuracy and comparability of the finite element results, all UKA models adopted a uniform loading scheme. A concentrated load of 1000 N was applied vertically downward along the mechanical axis of the femur, with the midpoint of the femoral condyle as the reference point. During loading, the boundary conditions of each model remained unchanged, and convergence verification was performed after loading^[22].

Two methods were used to validate the normal knee joint model^[20, 23] (Figure 5). Protocol 1 simulated static axial loading: all degrees of freedom at the distal tibia and fibula were constrained, while adduction/abduction and rotation of the femur in the coronal plane were also restricted. Subsequently, a uniform load of 1000 N was applied along the femur–tibia axis, with the center of the femoral condyle as the origin. The equivalent stress distribution in the tibial cartilage and menisci of the medial and lateral compartments of the knee joint was analyzed, including the maximum stress values and the stress proportion of each structure. Protocol 2 simulated the anterior drawer test: all degrees of freedom at the proximal femur were fully constrained, while only the flexion degree of freedom was constrained for the tibia and fibula. An anterior thrust of 134 N was applied at the tibial intercondylar eminence, and the anterior displacement of the intersection point between the line of action of this force and the anterior edge of the tibia was recorded. The results of the two validation methods were compared with those from previous studies. If the differences were small, the model was considered reasonable.

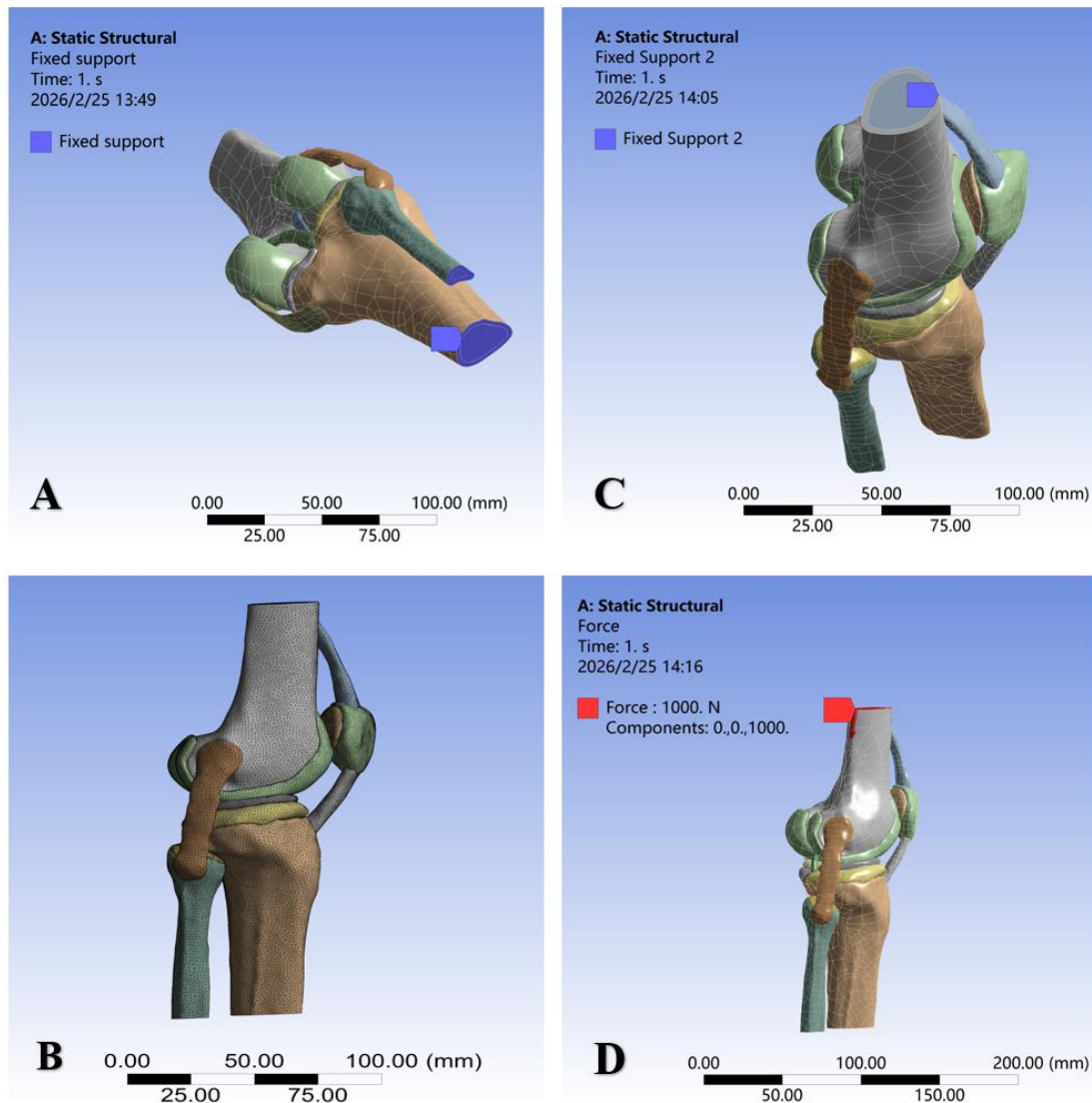


Fig. 5 Validation of the finite element model of the normal knee joint. (A, B) Boundary conditions; (C) Mesh generation; (D) Load application.

Results

Results of the model validation

The normal knee joint model exhibited good convergence after loading. The model had a total of 474,625 nodes and 271,845 elements. The validation results of Method 1 showed that the peak equivalent stresses on the medial and lateral tibial cartilage surfaces were 1.8063 MPa and 1.8595 MPa, respectively (Fig. 6); the peak equivalent stresses on the medial and lateral meniscus surfaces were 5.44 MPa and 6.2323 MPa, respectively (Fig. 7). The load distribution ratios of the medial and lateral compartments were 58.46% and 41.54%, respectively, both of which were similar to those reported in previous studies. The validation results of Method 2

showed that the anterior displacement of the point on the tibia where the load was applied was 4.86 mm, which was consistent with the standard displacement of approximately 5 mm in the anterior drawer test, and the results were also similar to those of previous studies^[20, 24].

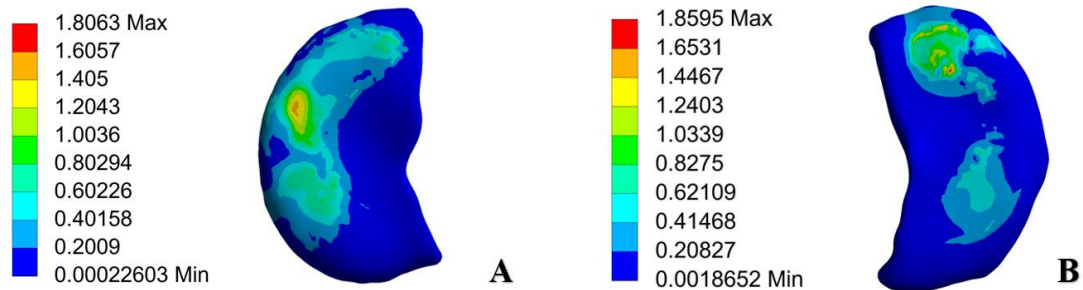


Fig. 6 Peak von Mises stress and stress distribution on the tibial cartilage surface. (A) Stress on the medial tibial cartilage surface; (B) Stress on the lateral tibial cartilage surface.

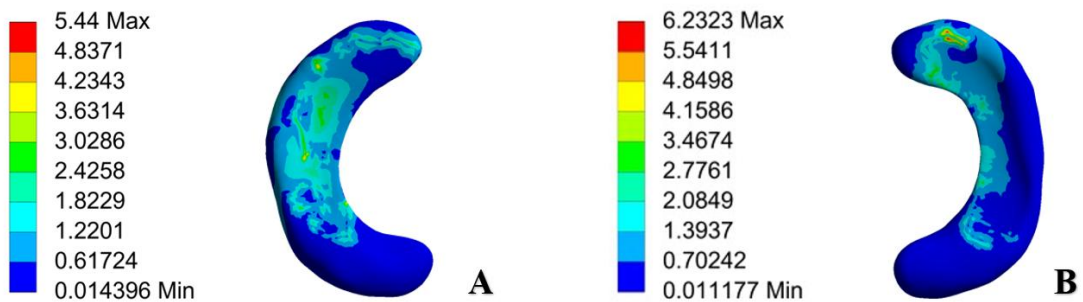


Fig. 7 Peak von Mises stress and stress distribution on the meniscus surface. (A) Stress on the medial meniscus surface; (B) Stress on the lateral meniscus surface.

Stress changes in the knee joint structures in different UKA finite element models

Upper surface of the femoral prosthesis

Comparing the three models with different bone defect depths, it was found that the surface stress on the femoral prosthesis increased with the depth of the bone defect: when the bone defect depth increased from 1/3 to 2/3 of the length of the anterior load-bearing column of the prosthesis, the peak equivalent stress on the prosthesis rose from 385.88 MPa to 517.13 MPa, an increase of 34.01%; when the depth increased from 2/3 to the full length (3/3), the peak equivalent stress increased from 517.13 MPa to 537.58 MPa, an increase of only 3.95%, indicating a significant

slowdown in stress growth (Fig. 8A, B, C).

Comparing the three models with different femoral prosthesis translation distances, it was found that the medial translation distance of the femoral prosthesis exhibited a nonlinear relationship with the surface stress on the prosthesis: when the medial translation distance increased from 1 mm to 3 mm, the peak equivalent stress rose sharply from 799.71 MPa to 1565 MPa, an increase of 95.70%; when the distance further increased from 3 mm to 5 mm, the peak equivalent stress increased from 1565 MPa to 2294.2 MPa, an increase of 46.59%, with a reduced rate of stress growth (Fig. 8D, E, F).

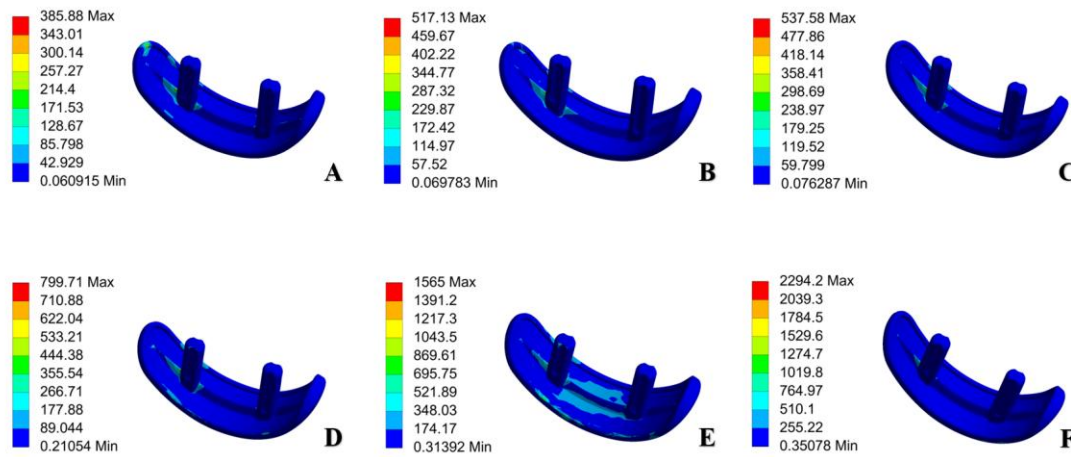


Fig. 8 Peak von Mises stress and stress distribution on the femoral prosthesis surface. A–C represent bone defect depths involving 1/3, 2/3, and 3/3 of the load-bearing column at the anterior edge of the femoral prosthesis, respectively; D–F represent the femoral prosthesis being translated inward by 1 mm, 3 mm, and 5 mm, respectively.

Upper surface of the polyethylene liner

Comparing the three models with different bone defect depths, it was found that the surface stress on the polyethylene liner increased with the depth of the bone defect: when the depth increased from 1/3 to 2/3 of the length of the anterior load-bearing column of the prosthesis, the peak equivalent stress on the liner rose from 72.363 MPa to 79.677 MPa, an increase of 10.11%; when the depth further increased to the full length, the peak stress rose sharply to 118.36 MPa, an increase of 48.55% (Fig. 9A, B, C).

Comparing the three models with different femoral prosthesis translation distances, it was found that the surface stress on the liner exhibited a nonlinear increase with the medial translation distance: when the medial translation distance increased from 1 mm to 3 mm, the peak stress slowly rose from 36.575 MPa to 41.938 MPa, an increase of 14.66%; when the distance further increased from 3 mm to 5 mm, the peak stress sharply rose to 150.75 MPa, an increase of as high as 259.46%. Unlike the bone defect model, the high-stress region of the liner in the translation model was not only concentrated at the contact interface but also exhibited a significant anterolateral shift, which may exacerbate local wear (Fig. 9D, E, F).

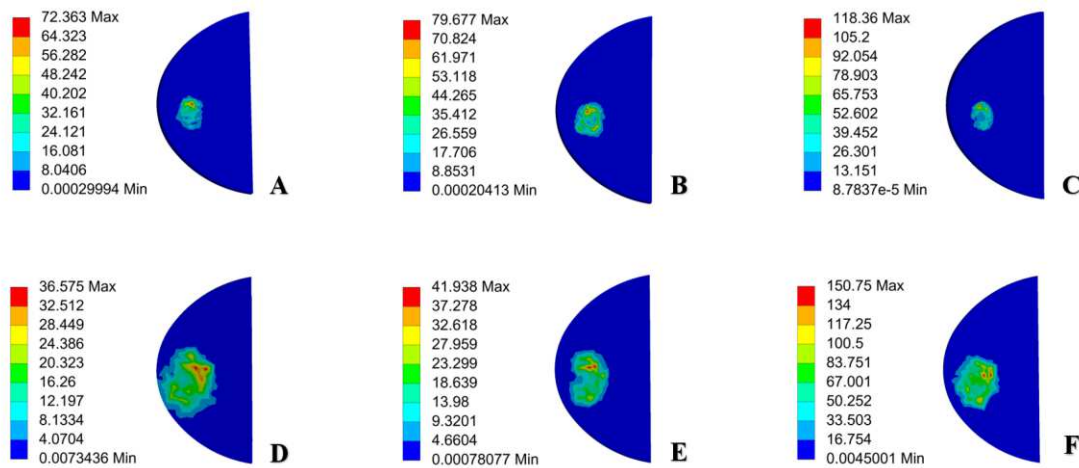


Fig. 9 Peak von Mises stress and stress distribution on the surface of the polyethylene liner.

A–C represent bone defect depths involving 1/3, 2/3, and 3/3 of the load-bearing column at the anterior edge of the femoral prosthesis, respectively; D–F represent the femoral prosthesis being translated inward by 1 mm, 3 mm, and 5 mm, respectively.

Upper surface of the tibial component

By comparing three models with different bone defect depths, it is found that the peak equivalent stress of the tibial prosthesis increases progressively with the increase of bone defect depth: when the bone defect depth increases from 1/3 to 2/3 of the length of the load-bearing column, the peak stress increases from 56MPa to 59.399MPa, with an increase of 6.07%; when it further increases to the full length, the peak stress reaches 60.865MPa, with an increase of 2.47% (Fig. 10A, B, C).

By comparing three models with different femoral prosthesis translation distances, it is found that the stress of the tibial prosthesis also increases progressively: when the medial translation of the femoral prosthesis increases from 1mm to 3mm, the peak equivalent stress on the surface of the tibial prosthesis increases from 137.40MPa to 145.97MPa, with an increase of 6.24%; when it is further translated medially to 5mm, the peak stress increases to 152.96MPa, with an increase of 4.79% (Fig. 10D, E, F).

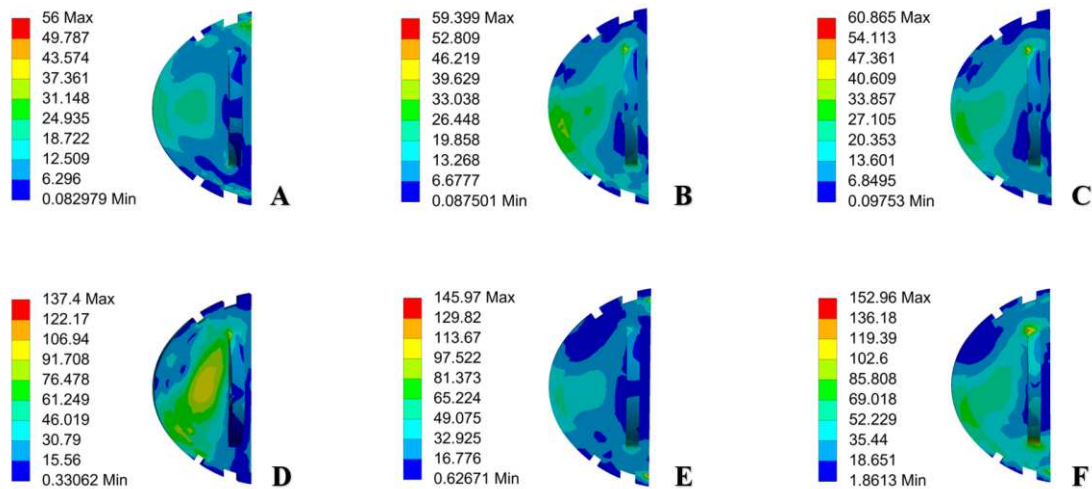


Fig. 10 Peak von Mises stress and stress distribution on the surface of the tibial prosthesis.

A–C represent bone defect depths involving 1/3, 2/3, and 3/3 of the load-bearing column at the anterior edge of the femoral prosthesis, respectively; D–F represent the femoral prosthesis being translated inward by 1 mm, 3 mm, and 5 mm, respectively.

Upper surface of the femoral cement mantle

As the depth of the bone defect increases, the stress on the femoral side bone cement mantle also increases: when the defect depth increases from 1/3 to 2/3 of the length of the anterior load-bearing column of the femoral prosthesis, the peak stress rises from 1.3587 MPa to 1.9535 MPa (an increase of 43.78%); when the defect reaches the full length, the peak stress surges to 18.009 MPa (an increase of 821.88%) (Fig. 11A, B, C).

As the medial translation distance of the femoral prosthesis increases, the stress

also increases: when the medial translation increases from 1 mm to 3 mm, the peak stress rises from 76.392 MPa to 129.4 MPa (an increase of 69.39%); when increased to 5 mm, the peak stress reaches 409.54 MPa (an increase of 216.49%). At a medial translation of 3 mm, the peak stress of 129.4 MPa already exceeds the maximum compressive strength of PMMA bone cement (110 MPa) (Fig. 13), significantly increasing the risk of fatigue fracture (Fig. 11D, E, F).

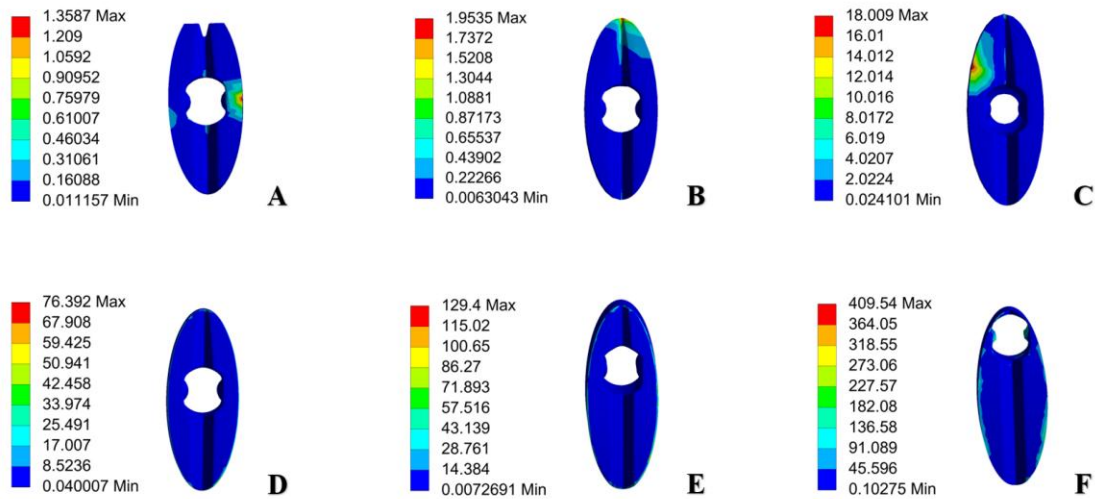


Fig. 11 Peak von Mises stress and stress distribution on the surface of the femoral cement mantle.

A–C represent bone defect depths involving 1/3, 2/3, and 3/3 of the load-bearing column at the anterior edge of the femoral prosthesis, respectively; D–F represent the femoral prosthesis being translated inward by 1 mm, 3 mm, and 5 mm, respectively.

Upper surface of the tibial cement mantle

As the depth of the bone defect increases, the surface stress of the cement mantle on the tibial side increases: when the bone defect depth increases from 1/3 to 2/3 of the length of the anterior load-bearing column of the femoral prosthesis, the peak equivalent stress increases from 11.378MPa to 19.432MPa, with an increase of 70.79%; when the bone defect depth reaches the full length, the peak stress increases to 28.038MPa, with an increase of 44.29% (Fig. 12A, B, C).

As the medial translation distance of the femoral prosthesis increases, the surface stress of the cement mantle on the tibial side increases: when the medial translation

distance increases from 1mm to 3mm, the peak equivalent stress increases from 23.153MPa to 52.124MPa, with an increase of 125.13%; when the medial translation reaches 5mm, the peak stress reaches 79.294MPa, with an increase of 52.13%(Fig. 12D, E, F).

In both the bone defect model and the prosthesis translation model, the peak equivalent stress of the bone cement on the tibial side does not exceed its maximum compressive strength (Fig. 14).

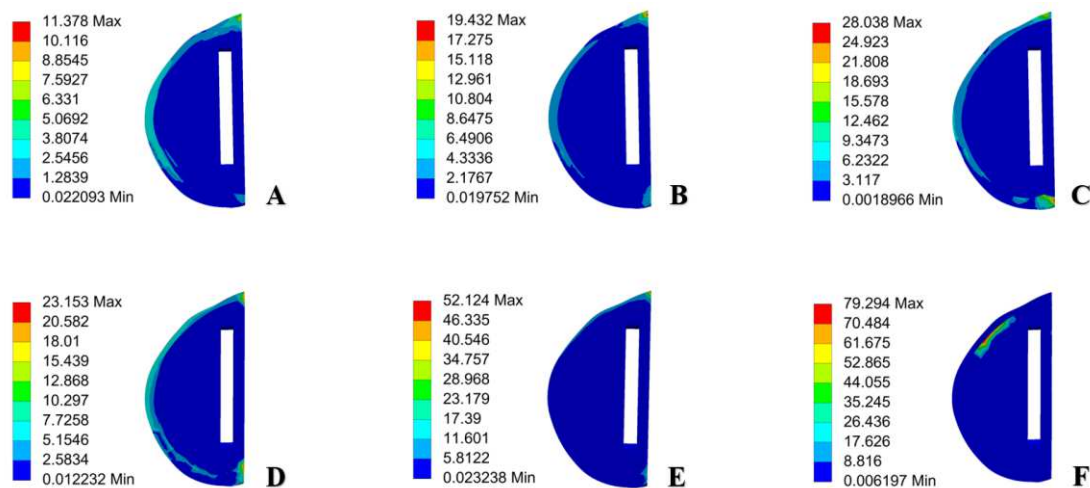


Fig. 12 Peak von Mises stress and stress distribution on the surface of the tibial cement mantle.

A–C represent bone defect depths involving 1/3, 2/3, and 3/3 of the load-bearing column at the anterior edge of the femoral prosthesis, respectively; D–F represent the femoral prosthesis being translated inward by 1 mm, 3 mm, and 5 mm, respectively.

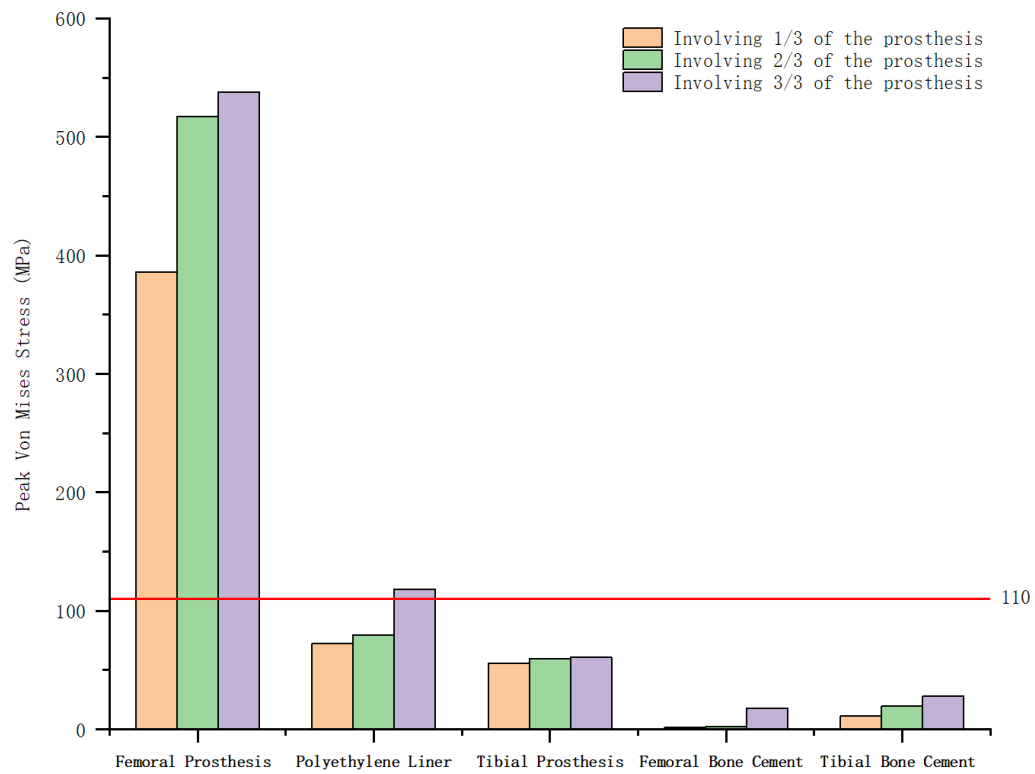


Figure 13 Comparison of peak equivalent stress among different structures in the bone defect model.

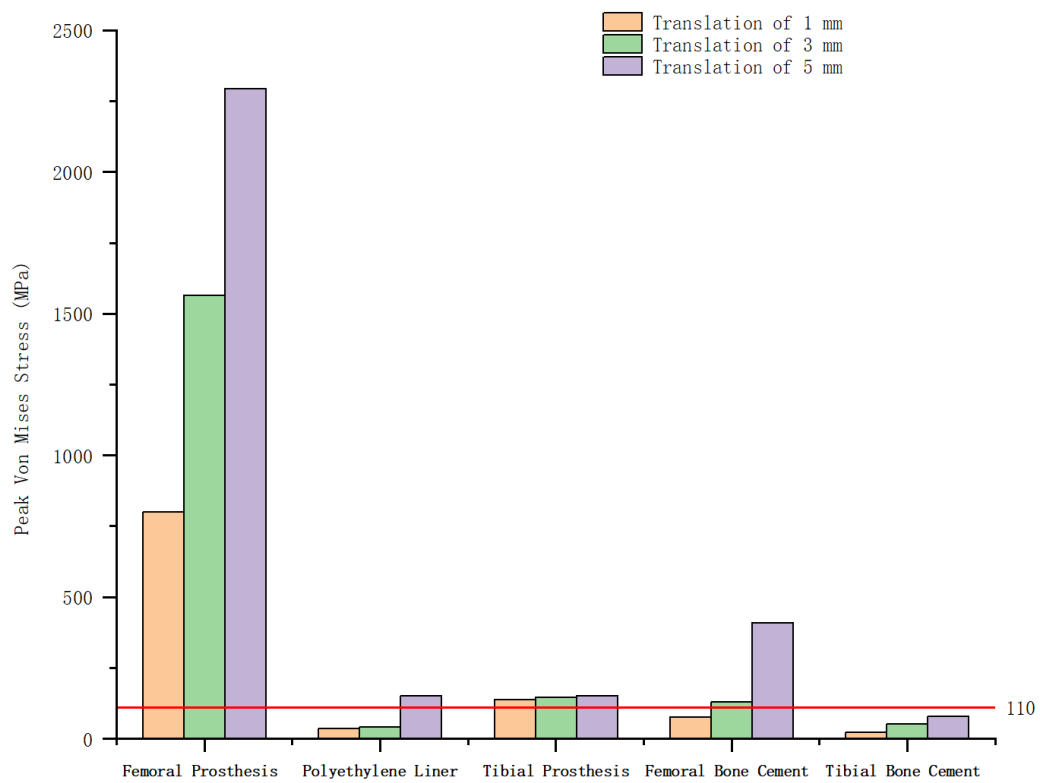


Figure 14 Comparison of peak equivalent stress among different structures in the translation model.

Discussion

With the aging population, the incidence of spontaneous osteonecrosis of the knee (SONK) is increasing among middle-aged and elderly women^[25], and it has become a significant public health issue. For moderate to advanced SONK that is unresponsive to conservative treatment, unicompartmental knee arthroplasty (UKA) is increasingly favored due to its advantages such as less trauma, faster recovery, and better preservation of proprioception. However, despite continuous technological advancements, the long-term revision rate of UKA remains relatively high. Its long-term success is highly dependent on three key factors: precise reconstruction of the lower limb mechanical axis, optimal three-dimensional positioning of the prosthesis, and a reasonably balanced distribution of intra-articular contact stress^[26].

In the clinical practice of performing UKA for SONK, there remains a lack of biomechanically evidence-based quantitative consensus regarding the safe depth threshold for femoral bone defects and the optimal coronal placement range of the femoral component, particularly for fixed-bearing prostheses^[27, 28]. Fixed-bearing prostheses require high precision in osteotomy, and even minor deviations can affect prosthesis stability^[29, 30]. Therefore, this study employed finite element analysis to systematically investigate the effects of varying depths of femoral bone defects and different degrees of medial translation of the femoral component in the coronal plane on the stress distribution and peak stress of each component of the knee joint when using the Link-Sled fixed-bearing prosthesis for the treatment of SONK.

The results showed that bone defect depth is a key factor influencing stress. As the defect deepened, the peak von Mises stresses of the femoral component, polyethylene liner, tibial component, and bone cement all increased. When the defect reached the full length of the anterior load-bearing peg of the femoral component, the stresses in all components reached their maximum values, with the femoral component exhibiting the highest peak stress (537.58 MPa) and the bone cement showing the greatest increase in stress. The loss of bony support in the defect area, filled with weaker bone cement, led to abnormal stress concentration at the

prosthesis–cement interface, significantly increasing the risk of micromotion and implant loosening. In addition, medial translation of the femoral component generally increased stresses in a nonlinear manner. Under a fixed defect depth, a 3 mm medial translation resulted in peak stresses of 1565 MPa for the femoral component and 129.4 MPa for the femoral-side bone cement, representing increases of 95.7% and 216.49%, respectively. The stress in the bone cement exceeded the typical compressive strength of PMMA bone cement (approximately 110 MPa), making it highly prone to fatigue fracture.

In summary, through refined finite element computational biomechanical simulations, this study clearly identified two key quantitative risk thresholds: first, when the depth of the femoral bone defect after SONK lesion debridement reaches or exceeds the full length of the anterior load-bearing peg of the selected femoral component; second, when the femoral component is medially offset by 3 mm from the ideal center in the coronal plane. Under these two conditions, significant stress concentration occurs within the knee joint, and the stress in the primary load-bearing structure (the bone cement layer) approaches or exceeds its material fatigue limit. It is recommended that intraoperatively, bone grafting or metal augmentation be used to ensure solid bony support, and that precise positioning be achieved with the aid of navigation, robotics, or extensive experience, keeping the medial translation error within a safe range (<3 mm).

Of course, this study has some limitations. Although the use of data from a single volunteer ensured comparability between different models, the results may lack population representativeness^[31, 32]. In addition, the simulation primarily focused on the static stance phase load and did not fully replicate the complex muscle co-contraction and load changes during dynamic activities such as walking and stair climbing. Finally, the analysis of this study mainly concentrated on the stresses in the prosthesis, insert, and bone cement, while the long-term adaptive bone remodeling process of the surrounding cortical and cancellous bone under stress shielding or stress concentration has not been thoroughly investigated.

Conclusions

For FB-UKA in SONK, when the defect depth does not exceed the full pillar length and medial translation is ≤ 3 mm, cement mantle fatigue, postoperative loosening, and dislocation of the femoral component can be reduced. This recommended range improves long-term prosthesis survival and maintains overall knee stability.

Abbreviations

SONK Spontaneous Osteonecrosis of the Knee

UKA Unicompartamental Knee Arthroplasty

TKA Total Knee Arthroplasty

MB-UKP Mobile-Bearing Unicompartamental Knee Prosthesis

FB-UKP Fixed-Bearing Unicompartamental Knee Prosthesis

CT Computed Tomography

MRI Magnetic Resonance Imaging

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Authors' contributions

D.Z.A. and F.Y.Q. jointly conceived and designed the experiments and drafted the manuscript. Z.H. and Z.B. acquired and analyzed the data. W.G.L. and L.S.H. provided research guidance. All authors reviewed and approved the final manuscript and are accountable for all aspects of the work to ensure accuracy and integrity.

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Availability of data and materials

The datasets used and/or analyzed during the current study are available from the corresponding author on reasonable request.

Declarations

Ethics approval and consent to participate

This study was approved by the ethics committee of Shandong University, China. All procedures were conducted in accordance with the principles of the Declaration of Helsinki. Participants in this study provided informed consent and agreed to the use of their clinical data for research and publication purposes. All data were anonymized by removing sensitive personal information to protect privacy.

Consent for publication

All participants involved in this study have provided informed consent for the publication of the research findings, including any individual data, images, or videos that may be included in the manuscript.

Competing interests

The authors declare that they have no conflict of interest.

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