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Measuring the effect of femoral malrotation on knee joint biomechanics for total knee arthroplasty using computational simulation

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Objectives

Malrotation of the femoral component can result in post-operative complications in total knee arthroplasty (TKA), including patellar maltracking. Therefore, we used computational simulation to investigate the influence of femoral malrotation on contact stresses on the polyethylene (PE) insert and on the patellar button as well as on the forces on the collateral ligaments.

Materials and Methods

Validated finite element (FE) models, for internal and external malrotations from 0° to 10° with regard to the neutral position, were developed to evaluate the effect of malrotation on the femoral component in TKA. Femoral malrotation in TKA on the knee joint was simulated in walking stance-phase gait and squat loading conditions.

Results

Contact stress on the medial side of the PE insert increased with internal femoral malrotation and decreased with external femoral malrotation in both stance-phase gait and squat loading conditions. There was an opposite trend in the lateral side of the PE insert case. Contact stress on the patellar button increased with internal femoral malrotation and decreased with external femoral malrotation in both stance-phase gait and squat loading conditions. In particular, contact stress on the patellar button increased by 98% with internal malrotation of 10° in the squat loading condition. The force on the medial collateral ligament (MCL) and the lateral collateral ligament (LCL) increased with internal and external femoral malrotations, respectively.

Conclusions

These findings provide support for orthopaedic surgeons to determine a more accurate femoral component alignment in order to reduce post-operative PE problems.

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Keywords: Malrotation, Total knee arthroplasty, Finite element analysis

Article focus

- What would be the effect of malrotation of the femoral component in total knee arthroplasty (TKA) on the polyethylene (PE) insert and patellar button in the walking and squatting positions?
- Would this malrotation affect forces on the medial and lateral collateral ligaments (MCL and LCL)?

Key messages

- External rotation of the femoral component increased the contact stress on the lateral side of the PE insert and on the force exerted on the LCL, while internal rotation increased the contact stress on the medial side of the PE insert and on the patellar button, as well as on the force exerted on the MCL.

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- Femoral component malrotation may increase loosening and lead to failure in TKA, and it may also explain the number of patient complaints and PE problems following TKA.

Strengths and limitations

- Strength: We analysed the contact stresses and ligament forces within the practical ranges of femoral component malrotation in TKA.
- Limitations: Finite element analysis study without clinical data.

Introduction

Total knee arthroplasty (TKA) is the most common treatment, with adequate long-term survival, for severe arthritis in the knee joint.[1] The surgical technique and prosthetic design have been improved to achieve more consistent and satisfactory results.[2,3] Despite the clinical success, some patients complain that the execution of important daily activities is difficult.[4] There are complications that remain, although the majority of TKA surgeries are typically successful. Sharkey et al reported that early revisions are due to infection, loosening, instability, and component malrotation.[5] Although infection and component failure continue to be a concern, failure because of biomechanical problems is the most challenging issue for engineers and surgeons.[6,7]

Malalignment is the most frequent complication following TKA, occurring in approximately one third of patients, and the post-operative outcome is dependent on surgical technique and the anatomical landmarks that are used.[8-11] Malrotation of the femoral component is a common problem, which may necessitate revision surgery because of issues associated with flexion gap asymmetry or patellar maltracking.[8,9] Femoral component alignment to the transepicondylar axis has been shown to result in optimal patellar tracking and extremely low patellar shear forces, similar to normal knee alignment.[12-14] The internal rotation of the femoral component increases the Q-angle, resulting in patellofemoral (PF) maltracking and a greater tendency for lateral patellar subluxation.[15] However, little has been reported on the biomechanical effects of high variability in the malrotation of the femoral components for functional tasks in TKA.[15,16]

Several *in vitro* studies have evaluated the effects of component malalignment in TKA, such as cadaveric experiments in the laboratory using a physiological gait simulator with computational analysis.[17-19] Moreover, Kuriyama et al reported that malrotation of the tibial component increased medial collateral ligament (MCL) tension in TKA.[20] Thompson et al[16] demonstrated that femoral rotation had a greater effect on the quadriceps forces, collateral ligament forces, and varus/valgus kinematics in the deep knee bend position. In addition, Colwell Jr et al[21] reported that the mobile bearings of a rotating platform can reduce malrotation between the tibial and femoral components and may reduce PF maltracking during deep knee bend. Heino Brechter et al[22] described a method for quantifying the PF joint contact area using magnetic resonance imaging (MRI). However, most previous studies have been skewed towards the investigation of patellar maltracking in the deep knee bend.[16,20-22]

Therefore, the purpose of this study is to determine the effects of internal or external rotation of the femoral component in stance-phase gait and in the squatting position. We investigated the maximum contact stresses on the medial and lateral sides of the PE insert and patellar button, as well as on the collateral ligament forces.

Patients and Methods

A 3D non-linear finite element (FE) knee model was developed from the computed tomography (CT) and MRI images of a healthy 36-year-old male subject.[23,24] The contours of the bony structures (including the femur, tibia, fibula, and patella) and the soft tissues (the ligaments and menisci) were reconstructed from the CT and MRI images, respectively. This computational knee joint model has been established and validated in previous studies.[23,24]

The bony structures were modelled as rigid bodies.[25] All major ligaments were modelled with non-linear and tension-only spring elements.[26,27] The force-displacement relationship based on the functional bundles in the actual ligament anatomy is shown in Table I.[28]

The forces across the components of the knee joint were calculated as follows:

$$f(\varepsilon)=\begin{cases}\dfrac{k\varepsilon^2}{4\varepsilon_1}, & 0\le\varepsilon\le\varepsilon_1\\ k(\varepsilon-\varepsilon_1), & \varepsilon>2\varepsilon_1\\ 0, & \varepsilon<0\end{cases}$$

$$\varepsilon=\frac{l-l_0}{l_0}$$

$$l_0=\frac{l_r}{\varepsilon_r+1}$$

where $f(\varepsilon)$ is the current force, k is the stiffness, ε is the strain, and ε_1 is assumed to be constant at 0.03. The ligament bundle slack length l_0 can be calculated by the reference bundle length l_r and the reference strain ε_r in the upright reference position.

Contact conditions were applied between the femoral component, PE insert, and the patellar button in TKA. The coefficient of friction between the PE material and metal was chosen to be 0.04 for consistency with previous explicit FE models.[24,29] Contact was defined using a penalty-based method with a weighting factor. As a

Table I. Properties in 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

aACL, anterior bundle of anterior cruciate ligament; pACL, posterior bundle of anterior cruciate ligament; aPCL, anterior bundle of posterior cruciate ligament; pPCL, posterior bundle of posterior cruciate ligament; LCL, lateral collateral ligament; aMCL, anterior bundle of medial collateral ligament; iMCL, intermediate bundle of the superficial medial collateral ligament; pMCL, posterior bundle of medial collateral ligament; PFL, popliteofibular ligament; OPL, oblique popliteal ligament; Lcap, lateral posterior capsule; Mcap, medial posterior capsule; ALS, anterolateral structures; aCM, anterior deep medial collateral ligament; pCM, posterior deep medial collateral ligament

Table II. Material properties for finite element model

	Young's modulus (MPa)	Poisson's ratio
CoCrMo ally	195 000	0.30
UHMWPE	940	0.46

CoCrMo, cobalt chrome molybdenum; UHMWPE, ultra-high molecular weight polyethylene; MPa, megapascal

result, contact forces were defined as a function of the penetration distance of the master into the slave surface. The PE insert and patellar button were modelled as an elastoplastic material (Table II).[24] The femoral and tibial components were fully bonded to the femur and tibia bone models, respectively. All implant components were modelled as linear elastic isotropic materials (Table II).[24]

Surgical simulation for TKA was performed by two experienced surgeons (Y-GK and KKP). A neutral position FE model was developed according to the following surgical preferences: default alignment for the femoral component rotation was parallel to the transepicondylar axis with the coronal alignment perpendicular to the mechanical axis and the sagittal alignment at 3° flexion with a 9.5 mm distal medial resection. To develop the malrotation models, ten different malrotation cases were considered with respect to the neutral position: neutral, internal and external 2°, 4°, 6°, 8° and 10° malrotations (Fig. 1). The tibial default alignment was rotated 0° to the anteroposterior axis, the coronal alignment was 90° to the mechanical axis, and the sagittal alignment was 5° of the posterior slope with an 8 mm resection below the highest point of the lateral plateau. The implant used was the Genesis II Total Knee System (Smith & Nephew, Inc., Memphis, Tennessee).

To evaluate the effect of internal and external malrotation on the femoral component of the TKA model, the stance-phase gait and squat loading conditions were applied to both the tibiofemoral and PF joint motions.[30-32] The FE model was analysed using ABAQUS software (version 6.11; Simulia, Providence, Rhode Island). The results for the maximum contact stress on the PE insert were assessed, and the patellar button pressure and collateral ligament forces were evaluated in both internal and external malrotation conditions.

Results

Effects of malrotation on the maximum contact stress of the PE insert and patellar button. Figure 2 shows the maximum contact stress on the PE inserts in the neutral position and the malrotation FE models during the stance-phase gait cycle. The medial peak contact stress on the PE insert increased by 14% with internal malrotation of 10°, whereas it decreased by 21% with external malrotation of 10° in the stance-phase gait loading conditions. There was an opposite trend on the lateral side. The lateral peak contact stress on the PE insert increased by 34% with external malrotation of 10°, whereas it decreased by 35% with internal malrotation of 10° in the stance-phase gait loading conditions.

The maximum contact stress on the PE inserts in the neutral position and the malrotation FE models in the squat loading condition are presented in Figure 3. Compared with the stance-phase gait loading condition, the effect of malrotation on the peak contact stress of the PE insert increased in the squat loading condition. The medial peak contact stresses on the PE insert increased by 103% and decreased by 50% with internal and external malrotations of 10°, respectively, in the squat loading conditions. There was an opposite trend shown on the lateral side in the stance-phase gait and squat loading conditions. The lateral peak contact stresses on the PE insert increased by 80% and decreased by 74% with external and internal malrotations of 10°, respectively, in the squat loading condition.

Figure 4 shows the maximum contact stress distribution on the PE insert in stance-phase gait and squat loading conditions. In internal malrotation, the peak contact stress was unsymmetrically concentrated lateral-posterior and medial-anterior under stance-phase gait loading conditions. In contrast, the peak contact stress was concentrated medial-posterior and lateral-anterior in external malrotation under stance-phase gait loading conditions. The peak contact stresses were skewed more lateral-posterior and medial-posterior in internal and external malrotations, respectively, under squat loading conditions.

The peak contact stress on the patellar button in the stance-phase gait and squat loading conditions is shown in Figure 5. The peak contact stresses on the patellar button increased by 14% and decreased by 22% with internal and external malrotations of 10°, respectively, in the stance-phase gait loading conditions. The peak contact stresses on the patellar button increased by 51% and

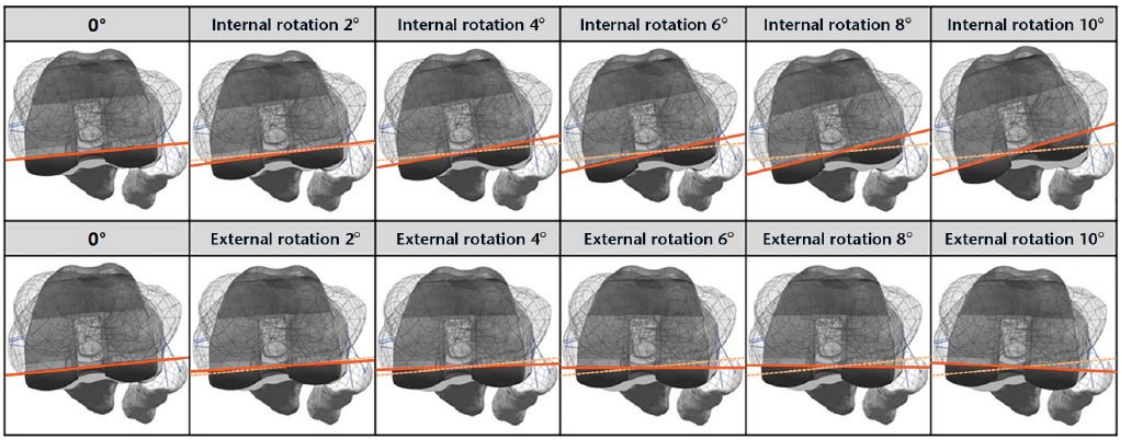

Fig. 1

Schematic of finite element model in neutral position and internal-external femoral malrotation conditions.

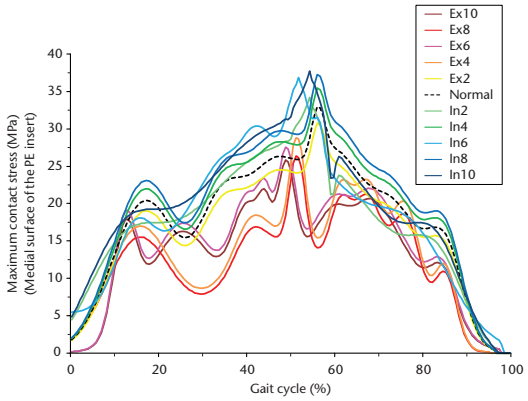

Fig. 2a

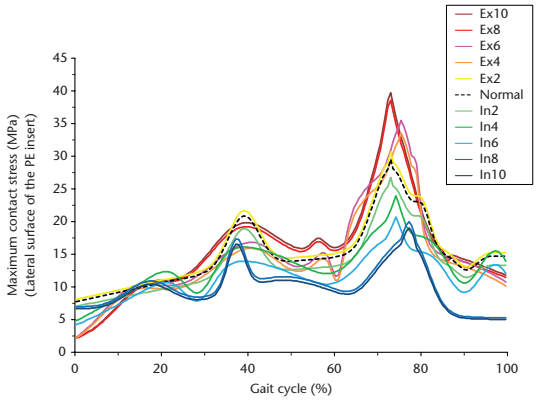

Fig. 2b

Effects of malrotation on the femoral components in total knee arthroplasty with respect to maximum contact stress on the polyethylene insert under gait cycle loading conditions.

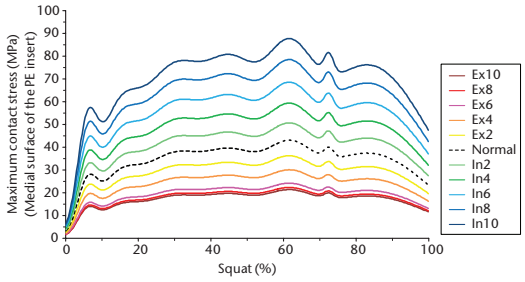

Fig. 3a

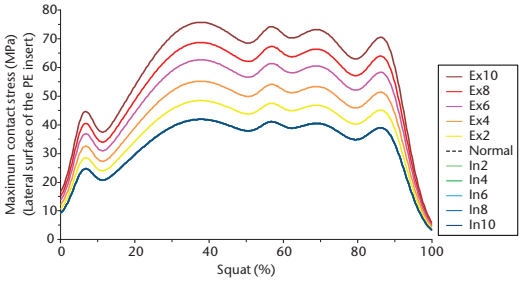

Fig. 3b

Effects of malrotation on the femoral components in total knee arthroplasty with respect to maximum contact stress on the polyethylene insert under squat loading conditions.

decreased by 25% with internal and external malrotations of 10°, respectively, in the squat loading condition.

Effects of malalignment on the collateral ligament forces. Figure 6 shows the ligament forces on the MCL, LCL, PFL, and anterolateral structures (ALS) in the neutral position and malrotation FE models during the stance-phase gait cycle. The ligament forces on MCL increased by 139% with internal malrotation of 10°, and decreased by 70% with external malrotation of 10° in the stance-phase gait loading condition. The ligament forces on LCL, PFL, and ALS increased by 53%, 26%, and 10%, respectively, with external malrotation. Furthermore, they decreased by 66%, 37%, and 15%, respectively, with internal malrotation in the stance-phase gait loading condition.

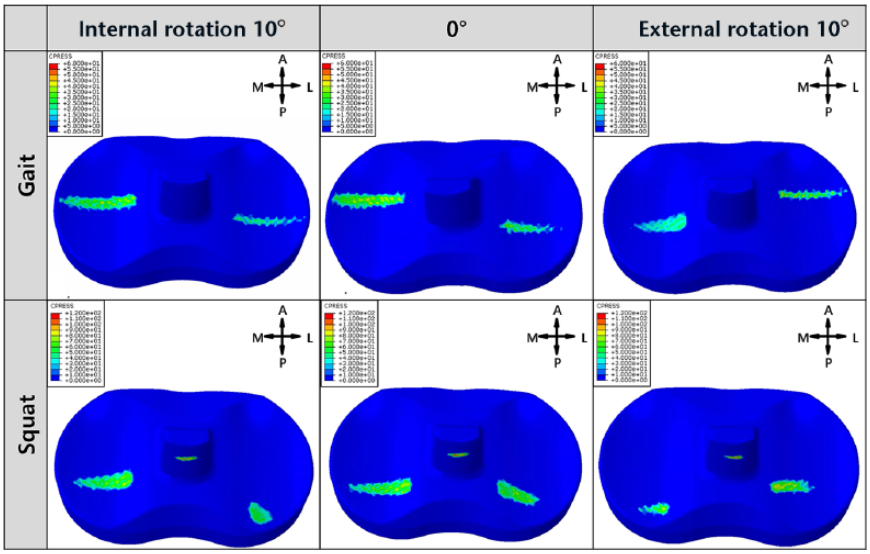

Fig. 4

Results of maximum contact stress distribution on the polyethylene insert in femoral component malrotation with normal, internal and external malrotation of 10° under gait and squat loading conditions.

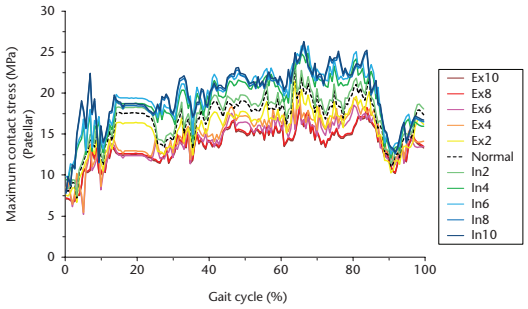

Fig. 5a

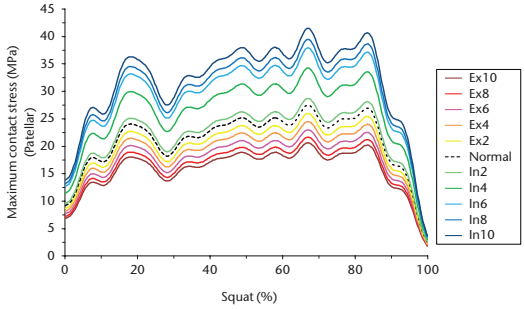

Fig. 5b

Comparison of maximum contact stresses on the patellar button in femoral component malrotation during the gait and squat loading conditions.

The ligament forces on MCL, LCL, PFL, and ALS in the neutral position and under malalignment in the squat loading condition are shown in Figure 7. The ligament forces on MCL increased by 397% and decreased by 98% with internal and external malrotations of 10°, respectively, in the squat loading condition. The ligament forces on LCL, PFL, and ALS increased by 103%, 46%, and 11%, respectively, and decreased by 72%, 43%, and 17%, respectively, with external and internal malrotations of 10° in the squat loading condition.

Discussion

We investigated the effect of walking on stance-phase gait and squat in the internal and external malrotations of the femoral component. Malrotation of the components in TKA has been attributed to several clinical complications.[19] Despite the reported high variability in malrotation of the femoral components in TKA, the biomechanical effects of this variability for functional tasks remain unknown.[16] Recently, malrotation was shown to lead to an elevation in the PF joint stress, PE insert wear, and ligament instability.[8-11,16,20] Therefore, the contact stress on the PE insert and patellar button, as well as the collateral ligament forces under malrotation conditions, may be appropriate for daily activities. The majority of previous studies have investigated the dynamic effects of such variability on joint loading during squats, but there has been no studies that have considered both gait and squat loading conditions.[11,14,16,18-22] Furthermore, to our knowledge, analysis of the contact stress on a PE insert and patellar button, as well as the ligament forces on the knee joint with respect to malrotation during gait and squat loading conditions, has not been reported in previous studies. The level of femoral malrotation reflects the maximum reasonable amount of rotational uncertainty encountered during surgery.[33-35]

This study investigated several biomechanical consequences of improper femoral component malrotation during stance-phase gait and squat loading simulations. Interestingly, our variables of interest were affected differently by the internal and external femoral rotations. Our

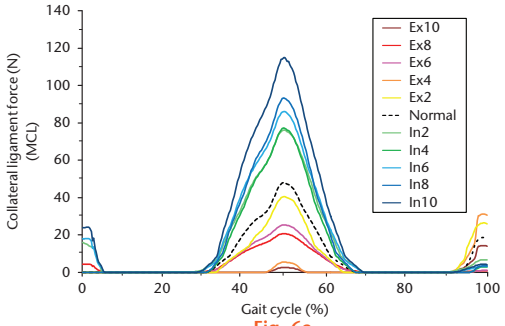

Fig. 6a

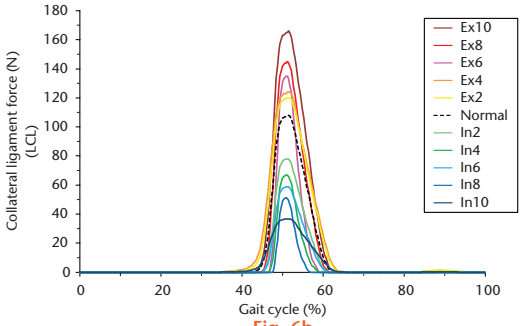

Fig. 6b

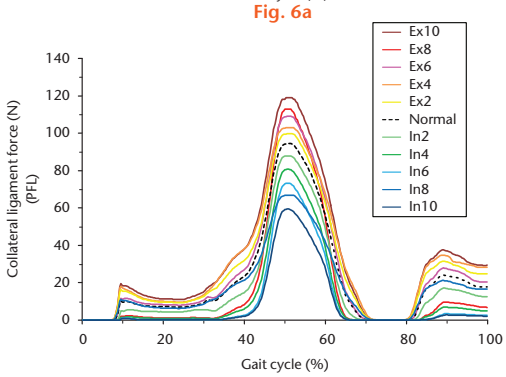

Fig. 6c

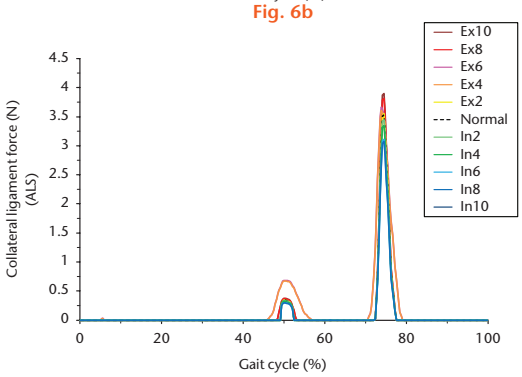

Fig. 6d

Effects of the malrotation on the femoral components in total knee arthroplasty with respect to collateral ligament force under gait cycle loading conditions.

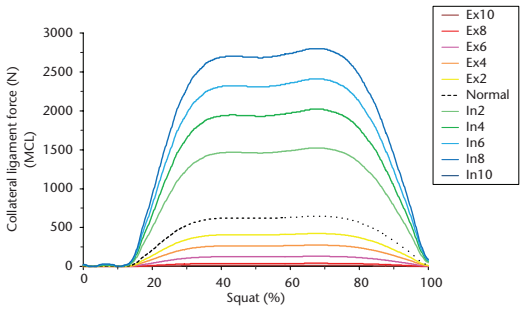

Fig. 7a

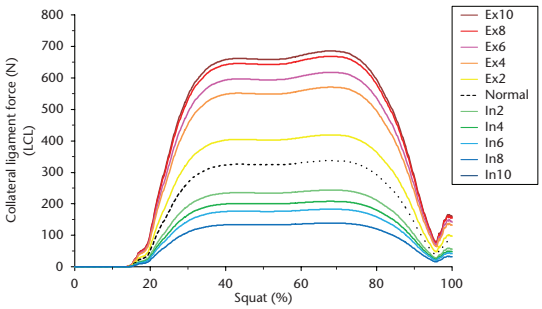

Fig. 7b

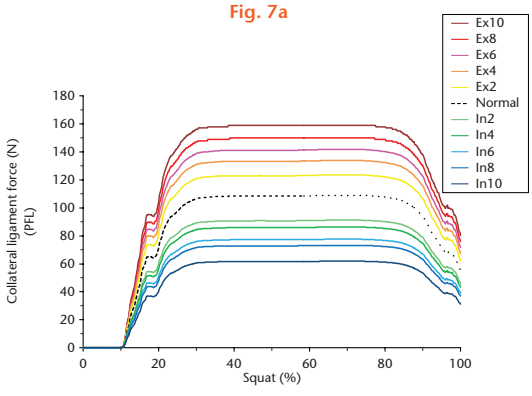

Fig. 7c

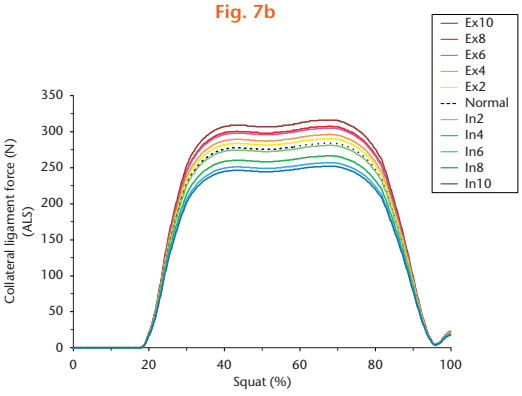

Fig. 7d

Effects of malrotation on the femoral components in total knee arthroplasty with respect to collateral ligament force under squat loading conditions.

findings regarding the maximum contact stress on the PE insert increase in femoral malrotation were consistent with the results of previous studies for gait loading conditions.[19]

Contact stress on the medial and lateral side of the PE insert increased with internal and external femoral malrotations, respectively, in the stance-phase gait loading condition. Moreover, a similar trend was found for the squat loading condition. One interesting finding was that the increase in the contact stress on the lateral side of the PE insert with external femoral

malrotation was greater than the increase in the contact stress on the medial side of the PE insert with internal femoral malrotation during the stance-phase gait cycle. However, in the squat loading condition, the increase in the contact stress on the medial side of the PE insert with internal femoral malrotation was greater than the increase in the contact stress on the lateral side of the PE insert with internal femoral malrotation. This may be the net effect of the internal rotation of the tibia and lateral femoral roll-back during high flexion.[16] High flexion of the knee joint in the squat position may have further increased the adverse effect of the enhanced contact stress on the medial side of the PE, with internal rotation of the femoral component.

The contact stress on the patellar button associated with femoral malrotation was consistent with the results of previous studies for the gait and squat loading conditions.[14,19,35,36] Our findings indicate that the maximum contact stress on the patellar button increased and decreased with the internal femoral malrotation in the stance-phase gait and squat loading conditions. There is no benchmark regarding the amount of rotation of the femoral component in TKA to reduce the contact stress on the patellar button during movement of the knee.[36] In the present study, the contact stress on the patellar button was influenced to a greater extent by the internal femoral malrotation in the stance-phase gait and squat loading conditions. Furthermore, a previous study found that an internal malrotation of 3° of the femoral component led to a 10% change in the total patellar force during knee flexion.[14]

Femoral malrotation primarily influenced the moment arms of the ligaments as well as the contact position on the PE insert.[37] This influence directly contributed to the change in the stresses on MCL and LCL as well as the load distribution, eventually leading to changes in the prediction of knee stress. We demonstrated that the MCL force increased significantly with internal rotation of the femoral component, and the force on the LCL, PFL, and ALS increased with external rotation in the stance-phase gait and squat loading condition. Our data indicate that internal femoral malrotation was comparatively more sensitive to the force exerted on the MCL ligaments. An increase in force on the MCL and LCL occurred during internal and external femoral malrotations, respectively, in agreement with previous studies.[16,20] In contrast, Kuriyama et al[20] reported that the force on MCL was slightly increased with external femoral malrotation. However, this difference was considered to be due to the loading conditions between the quasistatic and realistic dynamic conditions of daily activities. Excessive internal femoral component rotation may be detrimental to the MCL. Femoral malrotation contributed to an imbalance in the soft-tissue envelope, leading to instability and a reduction in the range of movement.[16,38,39] With an internally rotated femoral component, the tibia is rotated internally relative to the femur, resulting in the posterior movement of the tibial medial condyle. In this situation, the tensile force on the MCL was increased by a change in the distance between the ligamentous attachments. Kuriyama et al[20] reported that the LCL was less likely to be influenced by the malrotation conditions than the MCL because its stiffness value was smaller and there were limitations associated with the modelling methodology.

However, in this study, the force on LCL also increased with external femoral malrotation. The increase in the LCL force with external femoral rotation may be caused by both the external rotation of the tibia with respect to the femur and a combination of rotation and translation.[16] Our results indicated the opposite trend in component alignment with internal and external femoral malrotations. Internally rotated femoral components induce problems in MCL and increase the level of contact stress on the patellar button. In contrast, the contact stress on the patellar button decreased with an externally rotated femoral component. However, the contact stresses on the PE insert and patellar button, as well as the collateral ligament force, exhibited the opposite trends as the femoral implant was rotated internally and externally. From our results, the biomechanical effect of the femoral malrotation was greater during a high-flexion deep squat condition. Therefore, high-flexion rehabilitation exercises should be avoided in malrotation of femoral component.

There are several limitations to this study: only the intact model was validated. The computational model was developed using only data from a young, male subject. Using subjects of various ages would improve the validity of the results because the validity is also dependent upon the geometry of the knee joint. The balance of all of the collateral ligaments was assumed to be appropriate in our FE model. The stance-phase gait and squat simulations were performed, but simulations concerned with more demanding activities (getting up from and sitting on a chair, and stair climbing and descending) would be required for a more reliable investigation in the future.

In conclusion, external malrotation of the femoral component increased the contact stress on the lateral side of the PE insert and the force exerted on LCL, PFL, and ALS. On the other hand, internal femoral malrotation increased the contact stress on the medial PE insert and patellar button as well as the force exerted on the MCL in this study. Therefore, femoral component malrotation could have a negative effect on long-term survivorship in TKA.

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Author Contribution
- K-T. Kang: Co-first author, Designed the study, Evaluated the result using FEA, Wrote the paper.
- Y-G. Koh: Co-first author, Supervisor of study, Data analysis, Surgical simulation.
- J. Son: Developed the 3D model.
- O-R. Kwon: Data analysis.
- C. Baek: Data analysis.
- S. H. Jung: Data analysis.
- K. K. Park: Designed the study, Wrote the paper, Surgical simulation.

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