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Effect of Component Mal-rotation on Knee Loading in Total Knee Arthroplasty using Multi-body Dynamics Modeling under a Simulated Walking Gait

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Author Contribution Statement:

Zhenxian Chen, Ling Wang, Yaxiong Liu, Jiankang He, Qin Lian, Dichen Li and Zhongmin Jin conceived and designed the study. Zhenxian Chen performed the simulation and prepared the manuscript. Ling Wang and Zhongmin Jin reviewed and edited the manuscript. All authors read and approved the manuscript.

ABSTRACT

Mal-rotation of the components in total knee arthorplasty (TKA) is a major cause of postoperative complications, with an increased propensity for implant loosening or wear leading to revision. A musculoskeletal multi-body dynamics model was used to perform a parametric study of the effects of the rotational mal-alignments in TKA on the knee loading under a simulated walking gait. The knee contact forces were found to be more sensitive to variations in the varus-valgus rotation of both the tibial and the femoral components and the internal-external rotation of the femoral component in TKA. The varus-valgus mal-rotation of the tibial or femoral component and the internal-external mal-rotation of the femoral component with a 5º variation were found to affect the peak medial contact force by 17.8~53.1%, the peak lateral contact force by 35.0~88.4% and the peak total contact force by 5.2~18.7%. Our findings support the clinical observations that a greater than 3º internal mal-rotation of the femoral component may lead to unsatisfactory pain levels and a greater than 3º varus mal-rotation of the tibial component may lead to medial bone collapse. These findings determined the quantitative effects of the mal-rotation of the components in TKA on the contact load. The effect of such mal-rotation of the components of TKA on the kinematics would be further addressed in future studies.

Keywords: total knee arthroplasty; mal-rotation; multi-body dynamics; musculoskeletal model; contact force.

INTRODUCTION

The fundamental objectives of total knee arthroplasty (TKA) are to restore normal knee joint function and to relieve pain. However, the failure in TKA resulting from clinical error and mal-alignment of the components limits the long-term survivorship of such prostheses. Dalury et al.[1] retrospectively identified 820 consecutive revision TKAs and found that mal-position/mal-alignment was the seventh highest reason for revision. However, mal-position/mal-alignment also affects joint loading, component loosening and wear so that it may have a much larger effect on revision. For example, mal-rotation of the components in TKA may result in overload in the medial or lateral condyle, bone damage and bone cement crack initiation, severe wear of the polyethylene component, component loosening, and ultimately revision surgery.[2, 3] In contrast, good alignment measured against the neutral position (referenced to the mechanical axis to within 2°) after TKA leads to faster rehabilitation and better function.[4]

In previous clinical studies[5], the issue of mal-rotation was the most frequently discussed complication in TKA. Mal-rotation of a measurable degree has been found in approximately 10–30% of patients with TKA, depending on the surgical technique and the anatomical landmarks used.[5] Even in the hands of experienced surgeons, overall coronal mal-alignment (> ±3º from neutral) existed in approximately 28% of the patients.[6] Despite the improvements in surgical instruments and techniques as well as implant designs, a large percentage of the causes for revision are directly associated with the mal-position of the components. Conservatively, surgeons directly influenced at least 27% of the early and 18% of the late causes of revision, if the categories of instability and mal-alignment were purely considered[1]. In particular, variations in the rotational alignment of both the femoral and the tibial component of greater than 3º can occur in clinical surgery[3]. Patients with greater than 3º femoral internal rotation would receive a poor outcome after secondary patella resurfacing[7] and suffer unsatisfactory pain levels[8]. Berend et al.[9] found that a greater than 3º varus alignment of the tibial component was associated with medial bone collapse. Additionally, the wear measurement in retrieved inserts indicated that a varus mal-alignment as low as 3º may result in accelerated wear, even if a nearly ideal overall limb alignment was achieved[10]. Moreover, exceeding ±3º varus/valgus deviation from the mechanical axis has been associated with abnormal stress, deterioration of the prosthesis and aseptic loosening.[11] However, clinical observations could not explain these unsatisfactory results of such patients. Clinical observations can only retrospectively determine the effect of mal-rotation of the components on pain and functional scores. These are generally qualitative in nature and cannot reveal the changes in joint loading with mal-rotation of the component. The clinical observations should be correlated with the mechanical loading environment around the joint directly.

A number of in-vitro studies exist on the effects of component mal-alignment in TKA, e.g., laboratory experiments using cadaver legs in a physiological gait simulator[3], along with finite element analysis (FEA)[2] and multi-body dynamics (MBD) simulation[12]. However, the laboratory experimental costs inhibit the performance of multi-parameter analyses. FEA was previously used to investigate only the effect of components mal-alignment on the stress and strain with given boundary conditions (axial loading and joint motion), such that the effect of the mal-alignment on the overall knee contact force and motion was neglected. Moreover, the majority of FEA studies do not include a more physiological model associated with detailed information of bone, muscles, or ligaments. More recently, a host of musculoskeletal (MSK) MBD models with a deformable joint contact have been developed. Most of these models have been developed specifically for a cadaveric experiment without considering the whole lower limb of the MSK model or the force-producing constraints imposed on the biarticular muscles by neighboring joints.[12, 13] Furthermore, a number of computational simulations of the functional activities can be found on the effects of mal-rotation of the components on the knee contact forces, including the squatting motion[14], the weight-bearing deep knee bend[12] and the seated open-kinetic-chain knee extension[13], but none of these studies have addressed normal gait. In our previous study[15], the developed MSK MBD model was validated during walking gait through comparisons with the measured muscle activations and tibio-femoral (TF) contact forces from the instrumented knee prosthesis. However, only the perfectly aligned condition was simulated. The effects of the mal-alignment of components on knee loading during walking remain unknown.

In this study, by using a validated MSK model under a walking gait, our purpose was to quantify how the variations in femoral or tibial rotational alignment influenced the following: (1) the total TF contact force, (2) the medial and lateral contact forces, and (3) the patellar contact force. In addition, the sensitivity of knee contact forces to different mal-rotation of the components was further determined.

MATERIALS AND METHODS

Publically available data (https://simtk.org/home/kneeloads)[16], collected from an adult female implanted with an instrumented knee replacement (mass 78.4 kg, height 167 cm, left knee), were used for this study. A subject-specific lower extremity MSK model, including the left leg with the total knee replacement, was constructed in the commercially available MBD analysis program AnyBody (version 6.0, Anybody Technology, Aalborg, Denmark) by modifying a generic lower extremity MSK model (AnyBody Managed Model Repository V1.5.1), which was based on the Twente Lower Extremity Model (TLEM)[17] anthropometric database. The subject-specific bone and implant geometries (the femoral component, the tibial insert and the patella button), released in the published database[16], were imported into AnyBody to replace the existing left leg of the generic MSK model. The other segments of the generic MSK model were scaled with respect to the subject's weight and height as well as the relative positions of the ankle, knee, and hip joints as determined from the bone geometries. Six capsular soft tissue structures crossing the tibio-femoral (TF) and patello-femoral (PF) joint were included in the left leg of the modified lower extremity MSK model, including the posterior cruciate ligament (PCL), the medial collateral ligament (MCL), the lateral collateral ligament (LCL), the postero-medial capsule (PMC), and the medial and lateral PF ligaments (MPFL, LPFL). Anterior cruciate ligament (ACL) was removed according to the surgery. The ligament forces exerted by the ligament bundles followed a nonlinear elastic characteristic with a slack region, and a piecewise force-displacement relationship and material parameters for various ligaments were taken from a previous study[18] and are presented in the Supplementary Materials.

The left knee of the developed lower extremity MSK model with the implant was simulated via a force-dependent kinematics (FDK) approach[19]. Two deformable contact models were defined between the tibial insert and the femoral component bearing surfaces and between the patellar button and the femoral component. The contact forces for all contact pairs were calculated using a linear force-penetration volume law[20] with a contact *PressureModule* of 1.24×10^{11} N/m^3 in this study. The *PressureModule* was calculated using the equations derived by Fregly et al.[21], based on the elastic foundation theory; further details can be found in our previous study[15].

According to the patient's surgical report, an instrumented Zimmer NK-II cruciate-retaining prosthesis was implanted into the patient using a standard antero-medial approach. The tibial bone cut was made at 90º to the long axis in the coronal plane (0º varus) and at 90º in the sagital plane (0˚ posterior slope). The distal femoral cut was made at 6º valgus to the anatomic axis of the femur. The posterior femoral cut was made at a 3º external rotation with reference to the posterior surface of the posterior condyles. These were defined as the neutral position of the prosthetic components in the developed lower extremity MSK model. The positions of both the femoral and the tibial components were altered with respect to the neutral position to investigate the following thirteen mal-alignment cases: neutral, 3º and 5º of varus and valgus; 3º and 5º of internal and external rotation; 3º and 5º of anterior tilting and posterior tilting (Fig. 1).

The subject-specific gait pattern from an over-ground gait study obtained from the same adult female and measured at the patient's self-selected speed (approximately 1.0 m/s)[16] was used in this study. The corresponding experimental ground reaction forces (GRFs) and marker trajectories were imported into the developed subject-specific lower extremity MSK model in AnyBody. First, with the model scaling, an inverse kinematics analysis was performed to track the marker trajectories during the subject-specific gait. The pelvis and hip angles as well as the foot spatial locations were calculated. Second, an inverse dynamics analysis with the given muscle recruitment criterion was performed. The muscle recruitment problem was solved by minimizing a cubic polynomial cost function as described by John et al[22]. The calculated pelvis and hip angles as well as the foot spatial locations were taken as the inputs for the MSK model to simulate the kinetics of the patient's gait. The TF and PF contact forces were predicted from the combination of the GRFs, segment mass, muscle and ligament action in the inverse dynamics analysis. Meanwhile, the six degrees of freedom of the TF joint were left free to equilibrate automatically during the calculation under the effect of external loads and the muscle, ligament, and contact forces in the FDK solver. Next, the inverse dynamics analysis was executed for all mal-rotated cases with respect to the neutral position under the same gait. The effects of various configurations of the component mal-rotations on the total TF contact force, the medial and lateral contact forces, and the patellar contact force were predicted using the developed subject-specific lower extremity MSK model.

RESULTS

Knee Contact Forces for the Neutral Position

A general overview of the knee contact forces under the neutral position is shown in Fig. 2. The predicted medial and lateral contact forces, total TF contact force, and patellar contact force all varied during a gait cycle, with the maximum corresponding values of approximately 2.5, 1.0, 3.3, and 0.9 times the body weight, respectively. The effects of component mal-rotation were presented in the following, with respect to the predictions based on the neutral position. The changes were examined at maximum load bearing (approximately, 52% of the gait cycle) and peak knee flexion (approximately, 69% of the gait cycle).

Effects of the Component Mal-rotation on the Total TF Contact Force

Figure 3 shows the effect of the mal-rotation of the femoral and the tibial components in terms of the varus-valgus, internal-external and anterior-posterior tilting cases on the predicted total TF contact force. The peak total TF contact force at the maximum load bearing was increased by 11.0% at a 5º varus alignment of the tibial insert and increased by 17.9% at a 5º varus alignment of the femoral component. The anterior/posterior tilting mal-rotation of the tibial insert influenced only the total TF contact force during the swing phase of a gait cycle (Fig. 3). The total TF contact force at the peak knee flexion was increased by 14.7% at a 5º anterior tilting of the tibial insert and decreased by 12.6% at a 5º posterior tilting. A similar force change in the first half of the stance phase was 7.9% at a 5º anterior tilting of the tibial insert. However, the total TF contact forces were not sensitive to the variations in the internal/external mal-rotation of the femoral or tibial component and anterior/posterior tilting of the femoral component (Fig. 3).

Effects of the Component Mal-rotation on the Medial Contact Force

Figure 4 shows the effect of the mal-rotation of the femoral and the tibial components on the predicted medial contact force. The medial contact force was sensitive to variations in the varus/valgus mal-rotation from the femoral or tibial components and the internal/external mal-rotation from the femoral component. The peak medial contact force was increased by 36.2% at a 5º varus alignment of the tibial insert and increased by 37.9% at a 5º varus alignment of the femoral component. The peak medial contact force was increased by 17.8% at a 5º internal rotation of the femoral component and decreased by 21.3% at a 5º external rotation. The medial contact force at the peak knee flexion was increased by 12.5% at a 5º anterior tilting of the tibial insert and decreased by 12.0% at a 5º posterior tilting.

Effects of the Component Mal-rotation on the Lateral Contact Force

Figure 5 shows the effect of the femoral and the tibial component mal-rotations on the predicted lateral contact force. The lateral contact force at the maximum load bearing was decreased by 68.0% at a 5º varus alignment of the tibial insert and decreased by 40.8% at a 5º varus alignment of the femoral component. In particular, a 5º varus alignment mal-rotation resulted in zero loading on the lateral condyle at 30%, 60% and 90% of the gait cycle. The lateral contact force at the maximum load bearing was decreased by 35.0% at a 5º internal rotation of the femoral component and increased by 38.8% at a 5º external rotation. The lateral contact force at the peak knee flexion was increased by 17.5% at a 5º anterior tilting of the tibial insert and decreased by 13.4% at a 5º posterior tilting.

Effects of the Components Mal-rotation on the Patellar Contact Force

Figure 6 shows the effect of the femoral and the tibial component mal-rotations on the predicted patellar contact force. The maximum change of the patellar contact force at the maximum load bearing was increased by 21.9% at a 5º varus alignment of the tibial insert and increased by 18.5% at a 5º varus alignment of the femoral component. The maximum change of the patellar contact force at the peak knee flexion was decreased by 11.7% at a 5º internal of the femoral component and increased by 31.4% at a 5º external of the femoral component. The patellar contact forces were not sensitive to the variations in the anterior tilting/posterior tilting mal-rotation from the components in TKA (Fig. 6).

The magnitude and percentage changes of each mal-rotation position were examined at the maximum load bearing and the peak knee flexion as described in the Supplementary Materials.

Effect of Varus-Valgus Mal-rotation of the Femoral Component on Muscle Forces

Figure 7 shows the typical changes of the main muscle forces as a result of the varus-valgus mal-rotation of the femoral component. In particular, the peak muscle forces of *vastus medialis*, *vastus lateralis*, *medial gastrocnemius* and *peroneus longus* were increased by 11%, 12%, 19% and 158%, respectively, at a 5º varus alignment of the femoral component. The peak muscle forces of *lateral gastrocnemius*, *tibialis anterior* and *soleus* were increased by 65%, 108% and 98%, respectively, and the peak muscle forces of *medial gastrocnemius* and *peroneus longus* were decreased to zero at a 5º valgus alignment of the femoral component. However, the muscle forces of *biceps femoris long head*, *tensor fasciae latae*, *adductor magnus*, *gluteus maximus*, and *sartorius* were insensitive to the varus-valgus mal-rotation of the femoral component in the TKA.

DISCUSSION

Mal-rotations of the components in TKA have been attributed to several clinical complications. However, the dynamic effects of such variability on joint loading during walking have not been reported in previous studies. This study quantified the effects of the mal-rotation of the components in TKA on the knee contact forces during a walking gait using a lower-extremity MSK MBD model.

Our findings are consistent with the results of a previous study[3], which indicated that a 3º or 5º varus-valgus rotation of the tibial insert greatly changed the TF medial-lateral loading distribution. A steep increase (> 36.2%) of load at a 5º varus of the tibial insert at the medial plateau would support the previous clinical observation[9] that a greater than 3º varus alignment of tibial insert was associated with medial bone collapse. Our results also demonstrated that the effects from a 5º varus mal-rotation of the femoral component were slightly greater than those from the tibial insert, especially on the total TF contact force. Furthermore, the zero loading on the lateral contact force could be associated with the liftoff of the lateral condyle caused by the varus mal-rotation of the tibial or femoral component. Direct comparison of the predicted load distribution change as a result of the mal-alignment of the components in TKR was not possible with the previous studies. Nevertheless, a previous study by Werner et al[3] at a static loading instant when the knee was fully extended indicated a 96% loading shift to the medial compartment at a 5º varus mal-rotation of the tibial insert. This finding was close to the complete shift of the total loading to the medial side of the knee joint at the same instant during a walking cycle predicted from the present dynamics model. Moreover, a previous study [13] found that a 3º internal mal-rotation of the femoral component resulted in a maximum change of the total patellar force of approximately 10% during knee flexion.[12, 13] In contrast, the predicted maximum effect on the patellar contact force from the present study was 9% due to a 3º internal mal-rotation of the femoral component during the swing phase. Furthermore, the prediction of a greater change in knee contact forces may support the tendency to bias the femoral component into external rotation[23], thereby producing reduced TF contact loading and a lower patellar contact force. Although the effect of the tibial internal-external mal-rotation was apparent on only the medial and lateral contact forces during the swing phase (Fig. 4-5), it is still important to avoid excessive mal-rotation of the tibial component to reduce the corresponding effect on the antero-posterior translations[23]. The relative motion of the femoral component with respect to the tibial insert is equally important as far as wear is concerned; this aspect should be investigated in future studies.

Femoral or tibial rotational alignments mainly influenced the muscle/ligaments moment arms as well as the contact position on the tibial insert. This influence, in turn, directly contributed to the change of the muscle force, MCL and LCL forces and the load distribution, eventually leading to the changes in the predicted knee contact forces. Our prediction fully illustrated the change of the muscle force resulting from the varus-valgus mal-rotation of the femoral component. In particular, the changes of *vastus medialis*, *vastus lateralis*, *medial gastrocnemius*, *peroneus longus*, *lateral gastrocnemius*, *tibialis anterior* and *soleus* eventually altered the TF medial-lateral loading distribution. In addition, the external rotation of the femoral component induced a higher LCL force, whereas the internal rotation tightened the quadriceps and the MCL[21]. The tightened quadriceps and MCL might help indicate the reported unsatisfactory pain in clinical observations[8]. Such an imbalanced soft tissue loading resulted in changes in the predicted knee contact forces, especially as the knee moved from flexion into extension. Moreover, the component mal-rotation also led to the variations in the contact position on the tibial insert, which further influenced the predicted knee joint forces and kinematics. This influence can be illustrated with the effect of the anterior-posterior tilting; similar changes occurred in the mid-stance phase and in the swing phase for the medial contact force and the total TF contact force. Such changes resulted from the anterior-posterior contact position variations as a result of the component tilt. Nevertheless, the percentage changes during the swing phase appeared to be larger due to the smaller magnitude of the total contact load.

In a previous publication[15] comparing the predicted and measured force data, the errors of 320 N and 181 N were found for the predicted maximum medial and lateral contact forces, respectively. By contrast, the maximum changes of the medial and lateral contact forces from a 5º varus of the tibial insert were 733 N and 441 N, respectively, and from a 5º valgus of the tibial insert were 1076 N and 574 N, respectively. These values provided further confidence in determining the effect of the variability in component alignment on the predicted joint loading. However, the changes resulted from the tibial internal/external mal-rotation and the femoral or tibial anterior-posterior tilting were smaller than the uncertainties; as a result, these results should be interpreted with caution.

There are some potential limitations to this study, in addition to the uncertainties in the predicted load from the present MSK MBD model. First, the quadriceps and collateral ligaments may be released during operation for the purpose of the installation and stability in TKA, which may influence muscle and ligament function and property. However, the muscle and ligament model of the present MSK model was not adjusted for different mal-rotation conditions. The effects of the uncertainties of muscle and ligament on joint loading predictions were not considered, which could markedly affect the predicted joint loading. Second, according to many previous studies[24-26], joint kinematics and kinetics did not exhibit significant alterations between the pre- and post-operative evaluations, and patients may still retain the pre-surgery gait pattern. Furthermore, all previous musculoskeletal models[12-14, 23], FEA models[2, 27] and experimental studies[3] on the effect of component mal-alignments in TKA on biomechanics have assumed the same input conditions at the neutral position. Therefore, in the present study, small changes in the component alignments were assumed to not affect the motion patterns at the hip or the foot, and the same gait trail was assumed to be used throughout all simulated mal-aligned cases. However, we did find marked changes in the knee joint kinematics, particularly in the anterior-posterior translation, and internal-external rotation, which will be addressed in more detail in a future study. Third, the use of mechanical, anatomical and kinematic alignment for TKA is under debate[28]. Our prediction was limited to the surgical error in the mechanical alignment. The effects of the surgical error in anatomical and kinematic alignments on knee joint loading predictions should be investigated in future work. Furthermore, the mal-alignment may influence the implant failure when combined with other variables, such as the patient's anatomical factors, gait pattern and implant design. Our findings were limited to a single patient with a given implant design. The sensitivity to the patient characteristics and the implant design should be investigated in future work. Despite these limitations, this study demonstrated the advantages of using a computational MSK MBD model to study the effects of variability in component alignment on the predicted knee joint loading.

Surgeons should cautiously avoid the mal-rotation of the components by more than 3˚ variations; the varus-valgus of the tibial or femoral component and the internal-external mal-rotation of the femoral component with as small as a 3˚ variation in angulation changed the medial-lateral force distribution and total TF contact force markedly. Such investigations may be a key step toward understanding the relationship between surgical parameters and knee joint mechanics, thus providing quantitative guidance for the orthopedic surgeons to improve patient satisfaction.

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FIGURE LEGENDS

Figure 1 Diagrams of the varying configurations of mal-rotation showing the varus-valgus rotation, internal–external rotation and anterior–posterior tilting of the femoral and tibial component in left TKA.

Figure 2 Knee contact force under neutral position as a function of gait cycle (unit: contact force in Newtons/Body Weight).

Figure 3 Effects of the mal-rotation of the components in TKA on the total TF contact force for varus/valgus rotation, internal/external rotation and anterior/posterior titling.

Figure 4 Effects of the mal-rotation of the components in TKA on the meidal contact force for varus/valgus rotation, internal/external rotation and anterior/posterior titling.

Figure 5 Effects of the mal-rotation of the components in TKA on the lateral contact force for varus/valgus rotation, internal/external rotation and anterior/posterior titling.

Figure 6 Effects of the mal-rotation of the components in TKA on the patellar contact force for varus/valgus rotation, internal/external rotation and anterior/posterior titling.

Figure 7 Effects of the varus/valgus mal-rotation of the femoral component on the muscle forces.

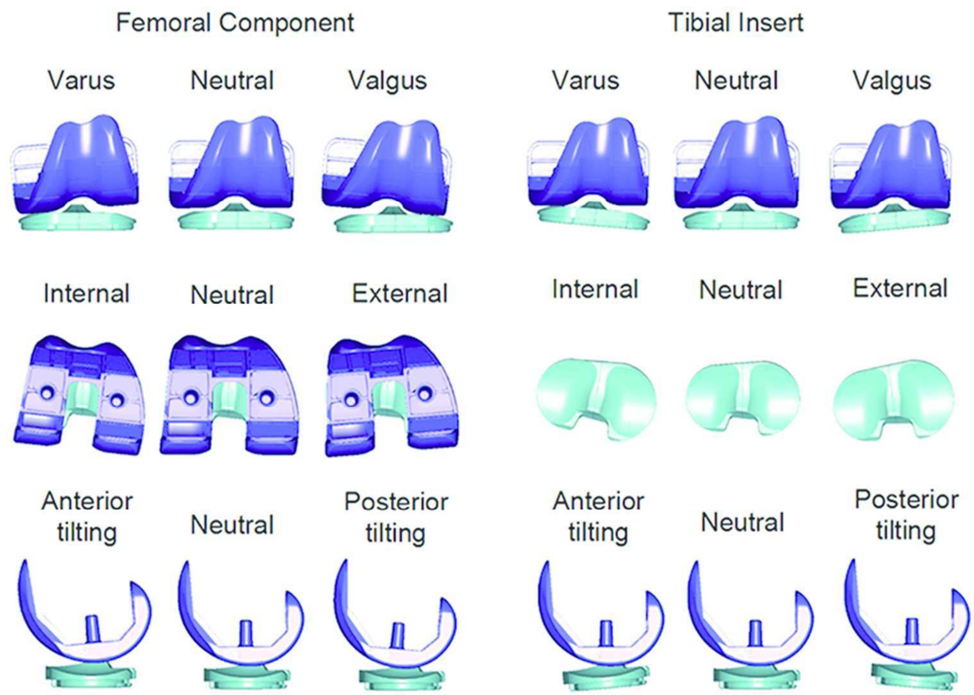

Figure 1 Diagrams of the varying configurations of mal-rotation showing the varus-valgus rotation, internal–external rotation and anterior–posterior tilting of the femoral and tibial component in left TKA.
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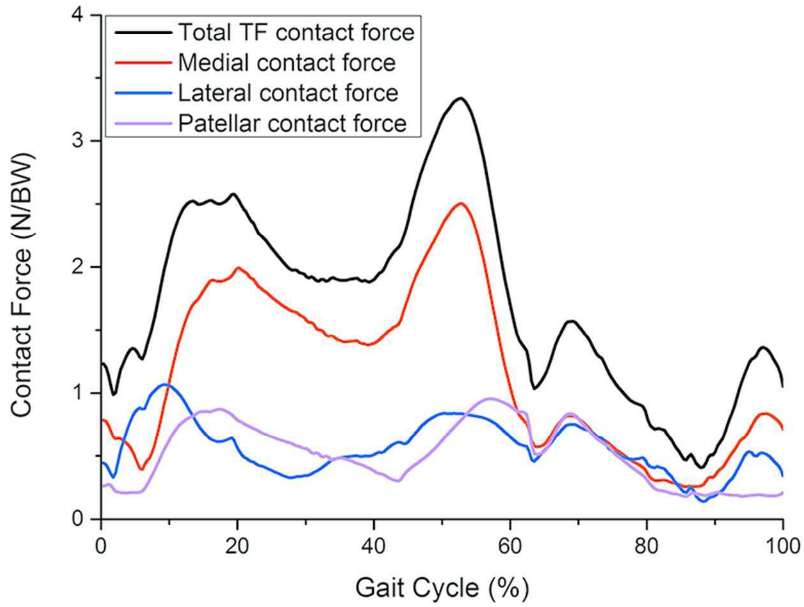

Figure 2 Knee contact force under neutral position as a function of gait cycle (unit: contact force in Newtons/Body Weight).
82x57mm (300 x 300 DPI)

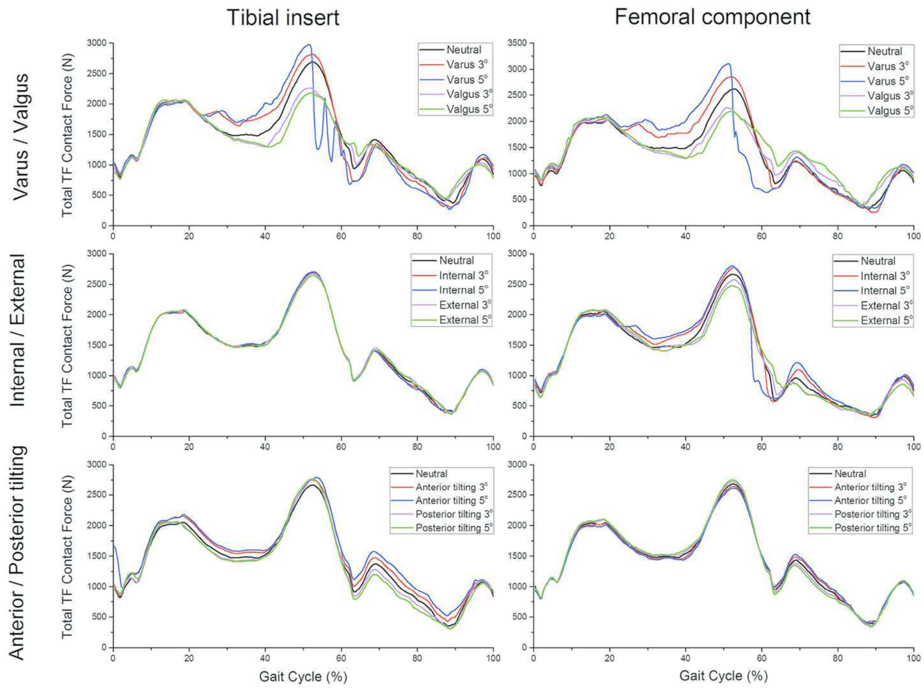

Figure 3 Effects of the mal-rotation of the components in TKA on the total TF contact force for varus/valgus rotation, internal/external rotation and anterior/posterior titling.
168x118mm (300 x 300 DPI)

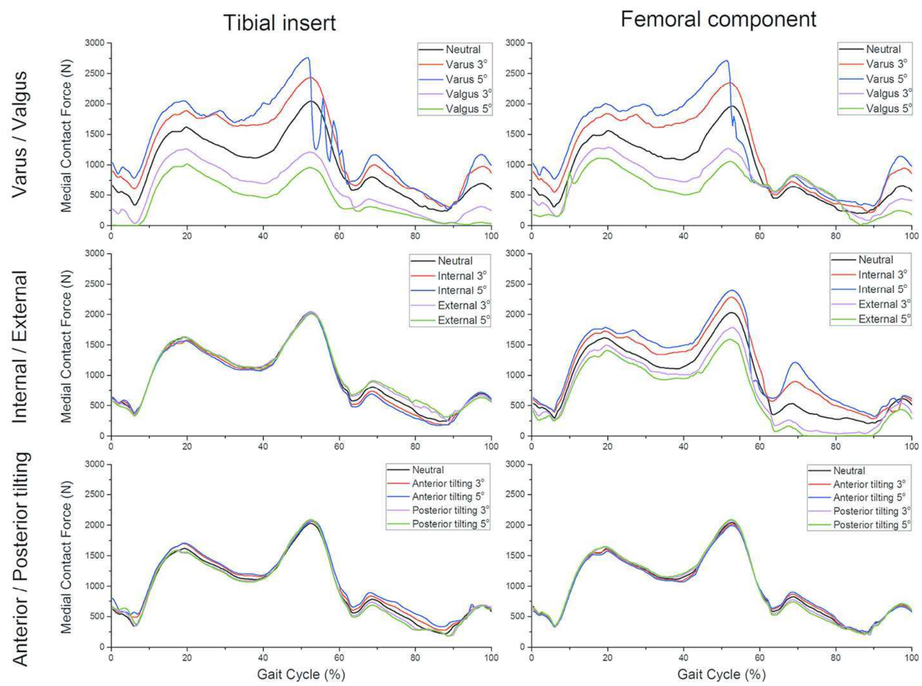

Figure 4 Effects of the mal-rotation of the components in TKA on the meidal contact force for varus/valgus rotation, internal/external rotation and anterior/posterior titling.
168x118mm (300 x 300 DPI)

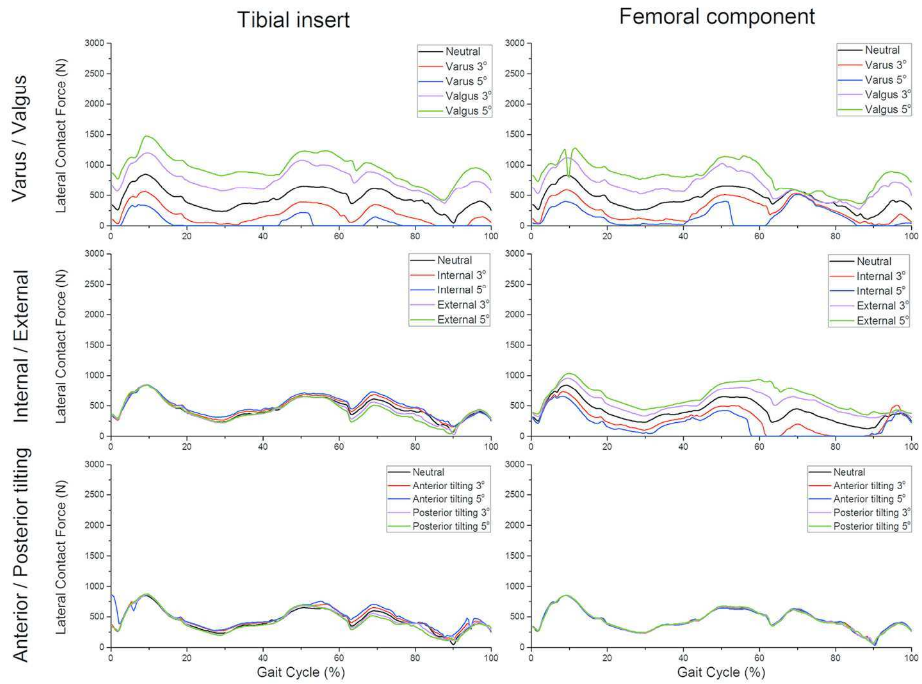

Figure 5 Effects of the mal-rotation of the components in TKA on the lateral contact force for varus/valgus rotation, internal/external rotation and anterior/posterior titling.
168x118mm (300 x 300 DPI)

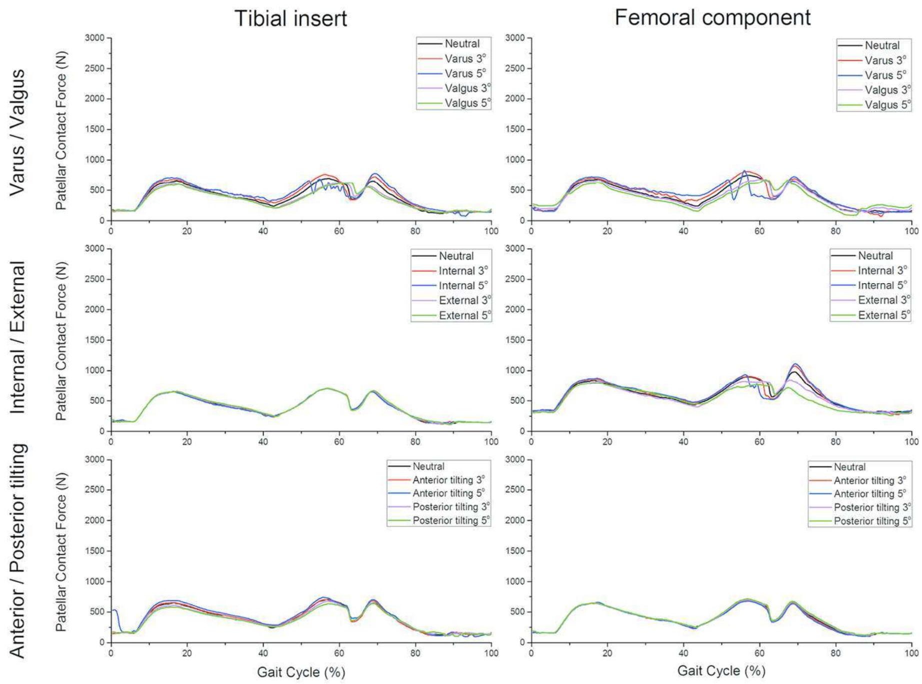

Figure 6 Effects of the mal-rotation of the components in TKA on the patellar contact force for varus/valgus rotation, internal/external rotation and anterior/posterior titling.
168x118mm (300 x 300 DPI)

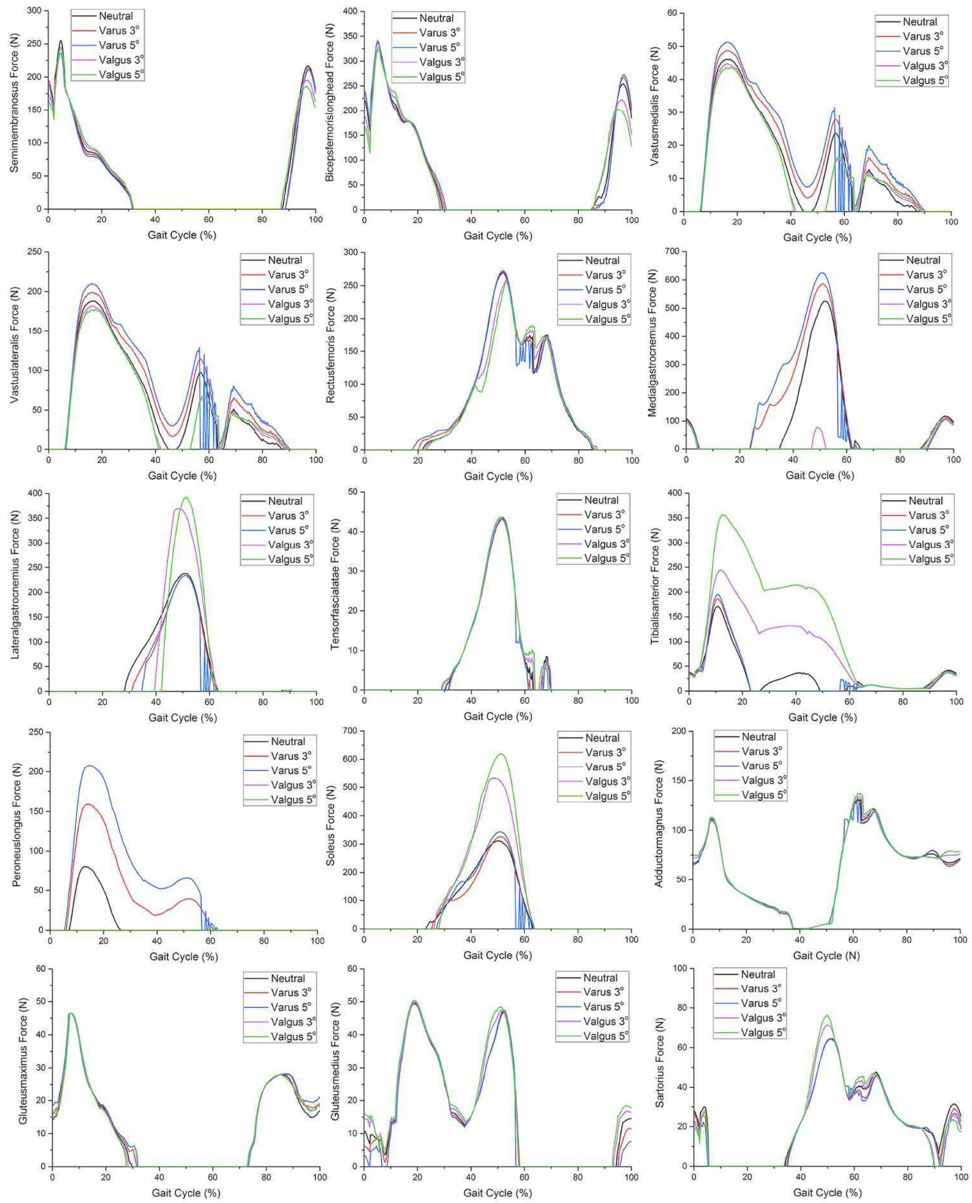

Figure 7 Effects of the varus/valgus mal-rotation of the femoral component on the muscle forces.
168x209mm (300 x 300 DPI)