



Knee Loading due to Varus and External Rotation in Gait Supports Medial Compartment Wear in Total Knee Arthroplasty

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Abstract

Malalignment of the lower extremity may be a cause for gonarthrosis and failure of total knee prosthetic components. Neutral lower limb alignment is considered straight alignment of the axes formed from the center of the femoral head to the center of the knee and from the center of the knee to the center of the ankle joint (the mechanical axes of the thigh and shank). Malalignment of the mechanical axis (measured by δ_v , the angle between the femoral and the tibial mechanical axes) occurs when varus moves the line of action farther medially from the knee joint center. Malalignment due to varus results in increases in medial compartment loading and has been attributed to medial compartment articular cartilage degeneration. External rotation of the hip and knee compared to the direction of gait is another form of malalignment. In external rotation, the knee displaces from the midline at stance phase. Knee loading and the consequences of external rotation on cartilage degeneration are not well understood. The purpose of our study was to develop a mathematical model that would calculate forces and moments in knees for gait and study how they vary with varus and external rotation malalignment. An additional objective was to develop a finite element model of total knee replacement to study stress patterns on the polyethylene insert during malaligned gait. We hypothesized that medial compartment loads would be increased by both varus and external rotation alignment of the knee compared to the direction of gait. We also hypothesize that stress patterns on total knee replacement (TKR) inserts under conditions related to malaligned gait will correlate with wear patterns observed in retrieval studies. It was found that for varus and external rotation, there is a shift towards adduction moment, which resulted in an increase in medial compartment loads, supporting our hypothesis. It was also found that the malalignment in gait results in alterations in tibial tray load magnitudes and load distribution that support elevated wear in the medial compartment as observed clinically.

Keywords

Varus, External rotation, Total knee replacement, Wear, Malalignment

Introduction

Total knee replacement (TKR) is a common treatment for gonarthrosis and has seen increasing use over the last 20 years. Malalignment of the lower extremity may be not only a cause for gonarthrosis but a precursor for failure of total knee prosthetic components [1-3]. Polyethylene loss in total knee replacement is evident and commonly reported in numerous retrieval studies as one cause of TKR failure [4-6]. In particular, wear in the anterior medial compartment is frequently observed (Figure 1) [7]. Malalignment of the TKR (e.g. varus or external rotation vs. neutral alignment) may be a contributing factor in accelerating medial compartment wear. Neutral lower limb alignment is considered straight alignment of the axes formed from the center of the femoral head to the knee center and from the center of the knee to the center of the ankle joint (the mechanical axes of the thigh and shank). Malalignment of the mechanical

axis (measured by δ_v , the angle between the femoral and the tibial mechanical axes) occurs when varus moves the line of action farther medially from the knee joint center increasing the medial load [8]. External rotation of the knee compared to the direction of gait exacerbates medial loading by further displacing the knee from the midline at stance [9].

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Figure 1: The tibial component of TKR showing medial polyethylene loss and burnishing in the anterior medial tibial tray [7].

Previous work has shown that in varus malalignment, the line of the force shifts farther medially from the knee joint center increasing the load on the medial compartment of the knee [10-19] resulting in a joint reaction force that is approximately three and a half times the lateral compartment loads [8]. Moyer, et al. [20] showed that the mechanical axis angle accounted for 32%-54% of the explained variation in knee adduction moment, yielding a 3.2 N increase for every 1° increase in varus. Knees that visually demonstrate varus thrust, have been shown to have greater knee adduction moment compared to knees without varus thrust [21-23] and have been reported to be at least four times more likely to have pain during weight bearing activity [24].

The purpose of our study was to develop a mathematical model that would calculate forces and moments in knees for gait and study how they vary with varus and external rotation malalignment. An additional objective was to develop a finite element model of total knee replacement to study stress patterns on the polyethylene insert during malaligned gait. We hypothesized that medial compartment loads would be increased by both varus and external rotation alignment of the knee compared to the direction of gait. We also hypothesize that stress patterns on TKR inserts under conditions related to malaligned gait would correlate with wear patterns observed in retrieval studies.

Methods

Applied knee joint forces

In the approach to define applied loads at the knee, results from clinical gait lab experiments were used [25].

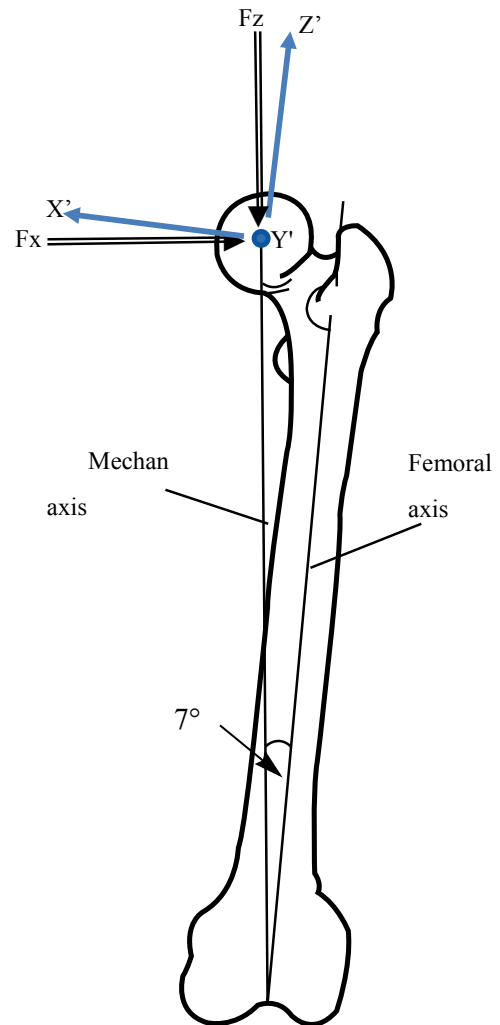
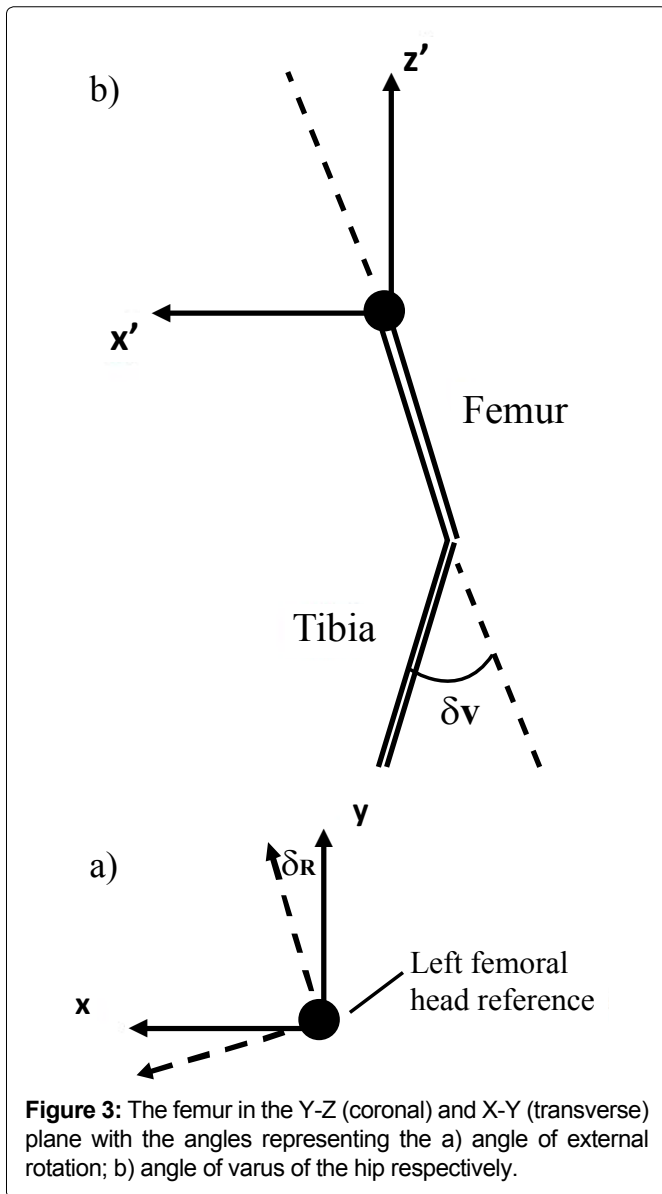


Figure 2: Coordinate system (x',y',z') for the left femur as defined by Bergmann, et al. [25] and contact force consisting of F_x , F_y and F_z as applied in this study along the mechanical axes (x,y,z). F_y is applied anterior to posterior (into the page) at the contact point in the y' direction.

Several full sets of hip joint load data as well as the angle at which these loads are applied is included in the referenced study. The reference includes information on several different loading scenarios including walking, climbing and descending stairs, sitting down, standing up, knee bending, standing on one leg, and stumbling. The current study considered maximum loads while walking. An equilibrium approach was applied to the hip joint loads to find the overall forces acting on the knee. This involved taking “snapshots” of the data at each location of the leg at each increment in gait and imposing equilibrium at each of these positions. The hip forces were imposed on a simplified “stick-figure” model as done previously [26,27]. Our axes were a global anatomical axis system (x',y',z') as well as a local axis system (x,y,z) which places the z -axis along the mechanical axis of the femur (Figure 2). The mechanical axis is the axis from the center of the femoral head to the center of the knee. The data



from Bergman, et al. [25] was offset seven degrees in the x-z plane from the mechanical axis. Therefore, the forces on the femoral head were redefined by a rotation about the y, y' axis of seven degrees. Several articles provided ranges for the angular orientation of the femur in the body to reorient the loads [28-30]. The forces at the knee were then determined using a force balance after positioning the femur and applying experimentally determined femoral loads as described below. The knee loads that define the “neutral” gait case are defined when the femur and the tibia are aligned according to the mechanical axis. This alignment assumes that the hip-knee-ankle angle is 180° , although reports indicate an angle of slightly less than 180° depending on the patient [31,32].

A unit vector, N , which establishes the orientation of the resultant force F from Bergmann, et al. [25] was determined by taking the force vector and dividing it by the magnitude of the force

$$\frac{F}{|F|} = N \quad (1)$$

The unit vector in the form of direction cosines is defined as

$$N = li + mj + nk \quad (2)$$

Where

$$\begin{aligned} l &= \cos \theta_x \\ m &= \cos \theta_y \\ n &= \cos \theta_z \end{aligned} \quad (3)$$

In equation (3) above, $\theta_x, \theta_y, \theta_z$ are the angles between the resultant force and the x, y and z axes, respectively. These direction cosines can be defined by the product of two cosines multiplied together. One direction cosine is defined by the angle between the force vector and the x - y plane (φ_{xy}) or the y - z plane (φ_{yz}). The second direction cosine is defined by the angle between the projected force on the x - y and y - z planes and the x (δ_x), y (δ_y) or z (δ_z) axes. The resulting direction cosines can be rewritten as

$$\begin{aligned} l &= \cos \varphi_{xy} * \cos(\delta_x + \delta_R) \\ m &= \cos \varphi_{xy} * \cos(\delta_x + \delta_R - 90) \\ n &= \cos \varphi_{yz} * \cos \delta_z \end{aligned} \quad (4)$$

External rotation (δ_R) of the knee is defined as a positive rotation of the hip joint such that the hip and knee joint are rotated through the same angle about the z -axis (Figure 3a) similar in orientation to the previously defined δ_x . This results in the knee pointing outward. Taking this into consideration in addition to recognizing the relationship between substituting δ_x and δ_y i.e.

$$\delta_y = 90 - \delta_x \quad (5)$$

we arrive at the direction cosines for external rotation

$$\begin{aligned} l &= \cos \varphi_{xy} * \cos(\delta_x + \delta_R) \\ m &= \cos \varphi_{xy} * \cos(\delta_x + \delta_R - 90) \\ n &= \cos \varphi_{yz} * \cos \delta_z \end{aligned} \quad (6)$$

Equation (6) defines the directions cosines of the resultant force corresponding to an internal rotation of δ_R . For varus of the knee, the direction cosines are determined by adding the desired angle of varus, δ_v (Figure 3b), to the direction cosines defined previously. Recognizing the relationship between δ_z and δ_y similar to eq. (5), and substituting that relation into eq. (4), we arrive at

$$\begin{aligned} l &= \cos \varphi_{xy} * \cos \delta_x \\ m &= \cos \varphi_{yz} * \cos\left(\delta_y + \frac{\delta_v}{2}\right) \\ n &= \cos \varphi_{yz} * \cos\left(\delta_y + \frac{\delta_v}{2} - 90\right) \end{aligned} \quad (7)$$

Equation (7) defines the directions cosines of the resultant force corresponding to an varus rotation of δ_v .

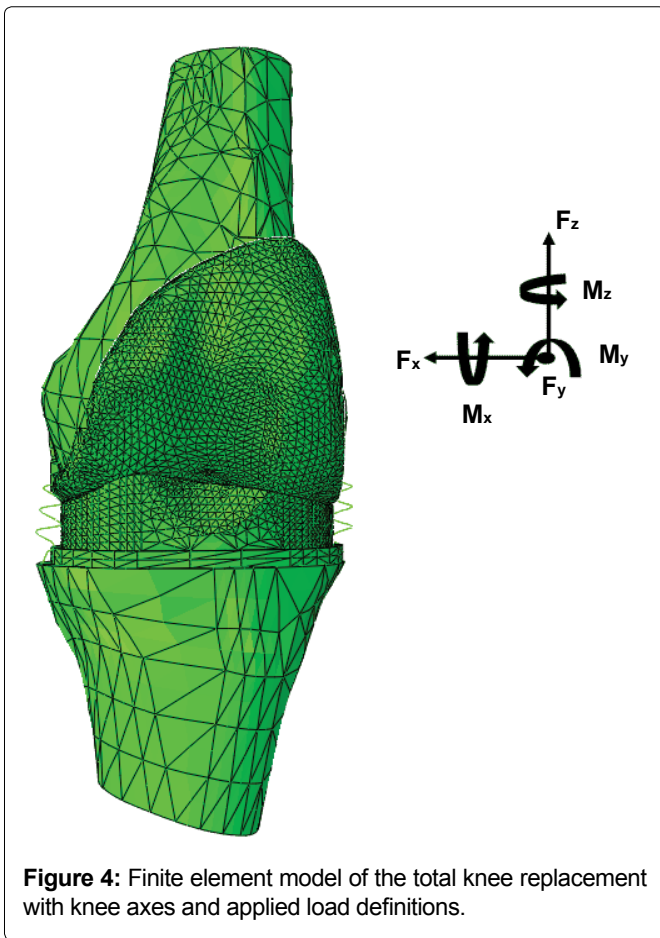


Figure 4: Finite element model of the total knee replacement with knee axes and applied load definitions.

Multiplying unit vectors by the magnitude of the force yields the force vectors for each case (normal gait, external rotation of the femur, and varus of the knee), i.e.

$$\mathbf{F} = |\mathbf{F}| * \mathbf{N} \quad (8)$$

After loading to the femoral head is determined for each case (i.e. neutral, external rotation or varus) in its properly oriented directions, equilibrium is imposed on the femur in order to determine knee loads. For each load case, the sum of forces, F , is expressed as

$$\sum F_i = F_{fi} + F_{ki} = 0 \quad (9)$$

Where f refer to the femoral head, k refers to the knee and i refers to the three mutually perpendicular directions x , y or z . Thus, equilibrium ensures that the knee forces are equal in magnitude but opposite in directions to the applied femoral head forces such that

$$F_{fi} = -K_{ki} \quad (10)$$

The x and y moments at the knee, M_x and M_y , respectively, are determined from the product of the components of the femoral head loads times the length of the femur

$$\begin{aligned} M_x &= -L_{femur} * F_{fy} \\ M_y &= L_{femur} * F_{fx} \\ M_z &= 0 \end{aligned} \quad (11)$$

The moment along the mechanical axis, M_z , is zero consistent with femoral head loads applied to the mechanical axis. The Forces and moments are subsequently geometrically rotated to align with knee axis (Figure 4). This results in modifications to the forces and moments calculated in (10) and (11) resulting in a nonzero torsional moment, M_z . Subsequent calculations using a matlab based program were performed to determine applied knee loadings. To illustrate the extent of the knee loading for moderate to extreme varus and external rotation, both angles were varied for 5, 10 and 25 degrees of malalignment. The algorithm output gave force and moment across the gait cycle from which the critical locations were identified. The analysis showed the maximum force occurred 17% of the gait cycle which is shortly after the opposite leg is leaving the ground and the foot flat on the ground. Forces and moments at the knee were then used to calculate femoral shaft loads using equilibrium. The femoral shaft loads were applied in the finite element analysis described below.

Finite element model

The finite element analysis was modeled after the ADVANCE® Medial-Pivot Knee. A physical model of the implant was scanned using the NextEngine 3D scanner (Santa Monica, CA), which gave the contact surface of both the tibial plastic and the femoral component. RapidWorks (Santa Monica, CA) was used to convert the pointcloud scanned image into a solid surface. RapidWorks was used to solidify the pointcloud into a solid surface by simply using the solidify feature. The two objects were then transferred into SolidWorks (Dassault Systems, Waltham, MA) where the implant model was attached to models of the femur and tibia. The implant was appropriately scaled for the size of the femur and tibia models and was placed in the proper orientation. There were two primary parts that were needed to properly position the knee. The first part is the set of angles that define the knee at a given position for gait. The second part is the contact points between the tibial insert and the femoral condyles in the implant. Gaudreault, et al. [33] was used to obtain flexion-extension, abduction-adduction and internal-external rotation angles of the knee throughout gait. At maximum loading corresponding to 17% gait position, the orientation of the femur to the tibia for normal or neutral gait was found to be 14 degrees of flexion, 0.4 degrees of adduction, and 1 degree of internal rotation. The femur was repositioned for varus and external rotation type gait. When 5 degrees of external rotation was added to the femur, the angle of rotation became 4 degrees. The contact points between the femoral and tibial components were defined at 17% gait [34]. The neutral position was modified for varus and external rotation forms of gait. For varus, an angle of 5 degrees was assumed based on the literature. Varus positioning resulted in condylar liftoff or separation in the

lateral condyle of the knee. For external rotation of the knee, our positioning was based on clinical experience of one of the investigators of this paper (JDB) who is a practicing orthopaedic surgeon. In external rotation the point about which the knee moves in the rotation is located outside of the knee on the lateral side. Therefore, both condyles move anteriorly in the implant, but the medial side moves further anterior than the lateral side of the knee. To find the angle of external rotation, the normal gait model was subjected to an angle of external rotation, keeping the contact points along a prescribed path until the contact was made on the anterior portion of the medial side of the knee. This angle was chosen as 5 degrees. The angle was then used in the rest of the models for the augmented knee replacements. All three models were positioned in SolidWorks and imported into ABAQUS (Dassault Systems, Waltham, MA) finite element software. The model consisted of a tibial tray, tibial plastic, femoral component, and a portion of a left femur (Figure 4). The elastic moduli of the components were 20.0 GPa for the cortical bone, 230 GPa for the cobalt-chromium tibial tray and femoral component and 0.689 GPa for the ultra-high molecular weight polyethylene (UHMWPE). Par-

tial tibia and femur were used to limit the size of the model to approximately 100,000 nodes. The applied loads were calculated as discussed previously and were applied to the femur while the tibia was fixed. The final important task was to define the interactions between the tibial plastic and the femoral component. We used finite sliding, a surface to surface discretization method, and a slave adjustment only to remove overclosure. We applied linear springs to either side to replicate the medial and lateral collateral ligaments. To aid with convergence, a tie constraint was used between the surfaces of the femoral component and the tibial plateau.

Results

The knee force components for 0 (neutral), 5, 10 and 25 degrees are shown in Table 1 and Figure 5 for varus malalignment and in Table 2 and Figure 6 for external rotation malalignment. Results are reported as applied to the tibial plastic. Anterior-posterior (Fy) and proximal-distal (Fz) forces remain nearly unchanged with varus and external rotation. However, medial-lateral directed force (Fx) changes significantly from -403 N laterally directed force to near zero in magnitude as varus and

Table 1: Knee forces and moments for neutral (0°) and 5, 10 and 25 degrees of varus malalignment.

Angle	Fx	Fy	Fz	Mx	My	Mz
0	-403	-249	-1737	107	-82	-13
5	-327	-249	-1752	108	-48	-13
10	-250	-249	-1765	108	-14	-13
25	-18	-249	-1783	108	87	-13

Table 2: Knee forces and moments for neutral (0°) and 5, 10 and 25 degrees of external rotation malalignment.

Angle	Fx	Fy	Fz	Mx	My	Mz
0	-403	-249	-1737	107	-82	-13
5	-374	-263	-1741	114	-69	-14
10	-314	-278	-1751	120	-43	-15
25	18	-306	-1774	132	102	-16

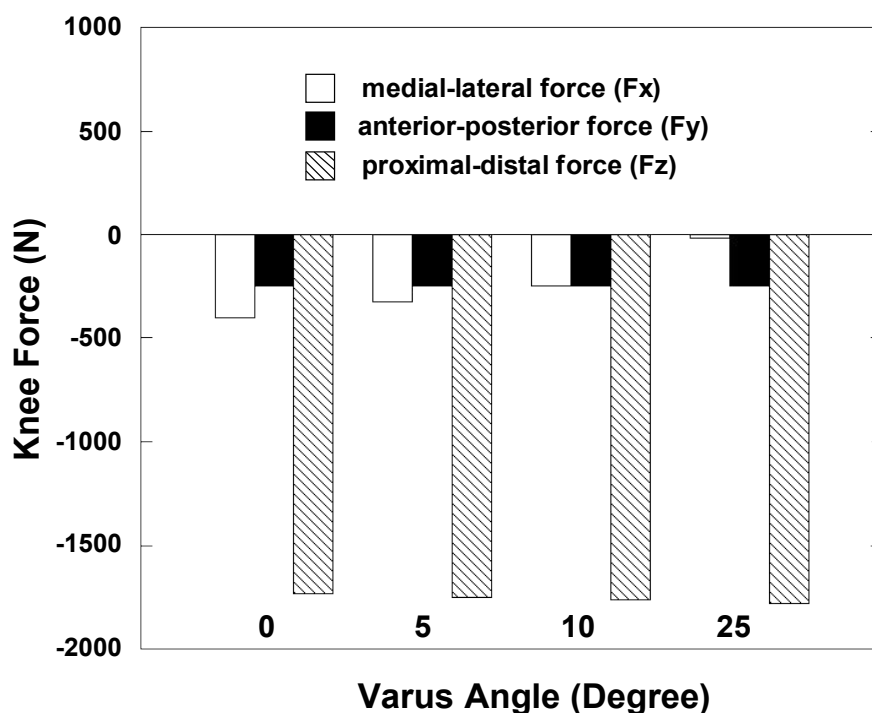


Figure 5: Knee forces for neutral (0°) and varus (5°, 10° and 25°) malalignment. Medially, anteriorly and proximally directed forces are positive.

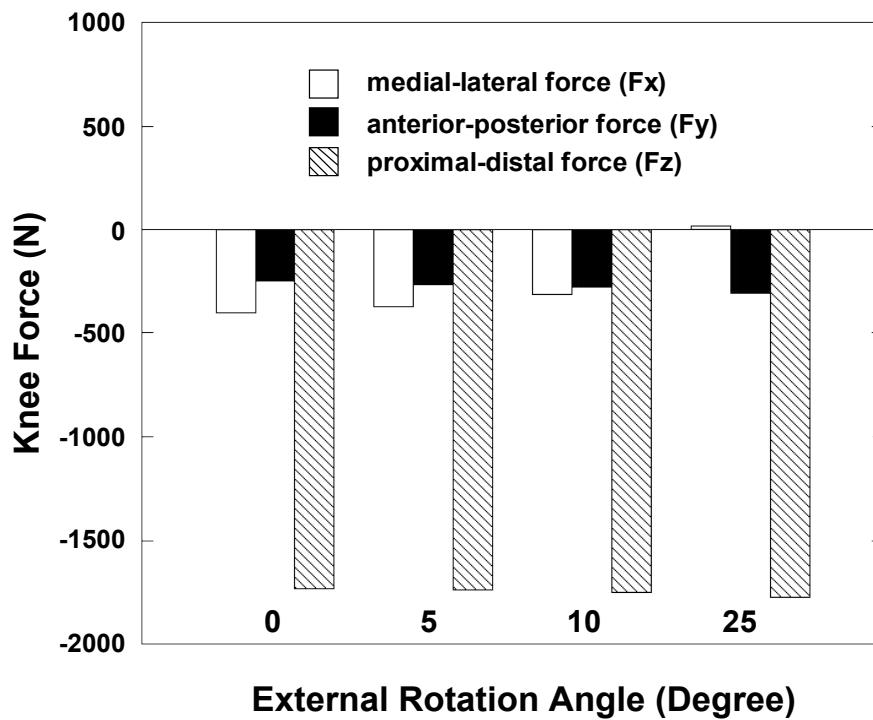


Figure 6: Knee forces for neutral (0°) and external rotation (5°, 10° and 25°) malalignment. Medially, anteriorly and proximally directed forces are positive.

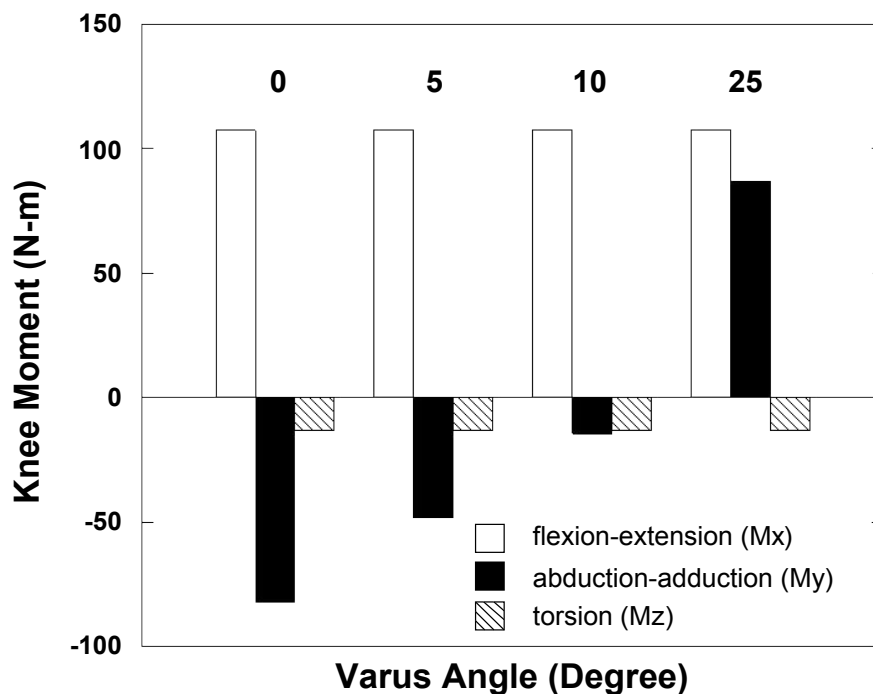


Figure 7: Knee moments for neutral (0°) and varus (5°, 10° and 25°) malalignment. Positive moments correspond to Flexion, adduction and internal tibia rotation.

external rotation angles increase. Thus, malalignment of either form results in forces applied to the knee that shift from medially directed to less medially directed and achieve a small laterally directed force for the case of 25 degrees of external rotation. The knee moment compo-

nents for 0, 5, 10 and 25 degrees are shown in Table 1 and Figure 7 for varus malalignment and in Table 2 and Figure 8 for external rotation malalignment. Results are reported as applied to the tibial plastic. Flexion-extension moment (Mx) and rotation moment (Mz) also ex-

perience small changes with varus and external rotation angle. Abduction-Adduction moment (M_y) however experiences significant changes with increases in varus and external rotation angles. For both types of malalignment, the moment changes from an abduction moment (negative, away from the body center) to adduction moment (positive, towards the body center) as the malalignment angle increase. The adduction moment results in ap-

plied loads to the knee that increase the tibial tray medial compartment loads. These results are reflective in finite element simulations. Finite element results of the tibial tray demonstrate both medial and lateral tibial tray compartment loading near implant center for neutral alignment (Figure 9). For five degrees of varus malalignment (Figure 10), finite element results show concentrated medial compartment loading with elevated stress results

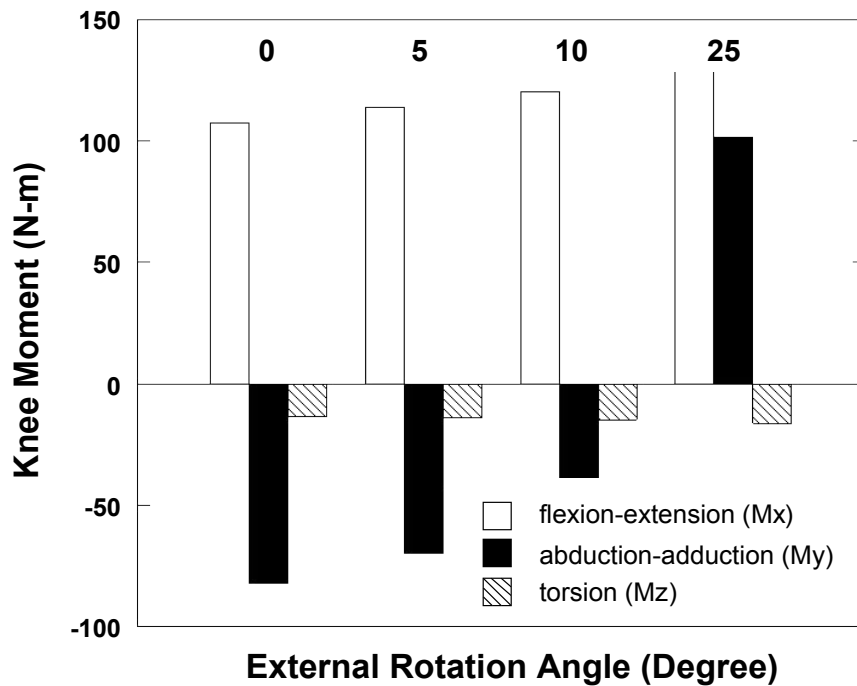


Figure 8: Knee moments for neutral (0°) and external rotation (5° , 10° and 25°) malalignment. Positive moments correspond to flexion, adduction and internal tibial rotation.

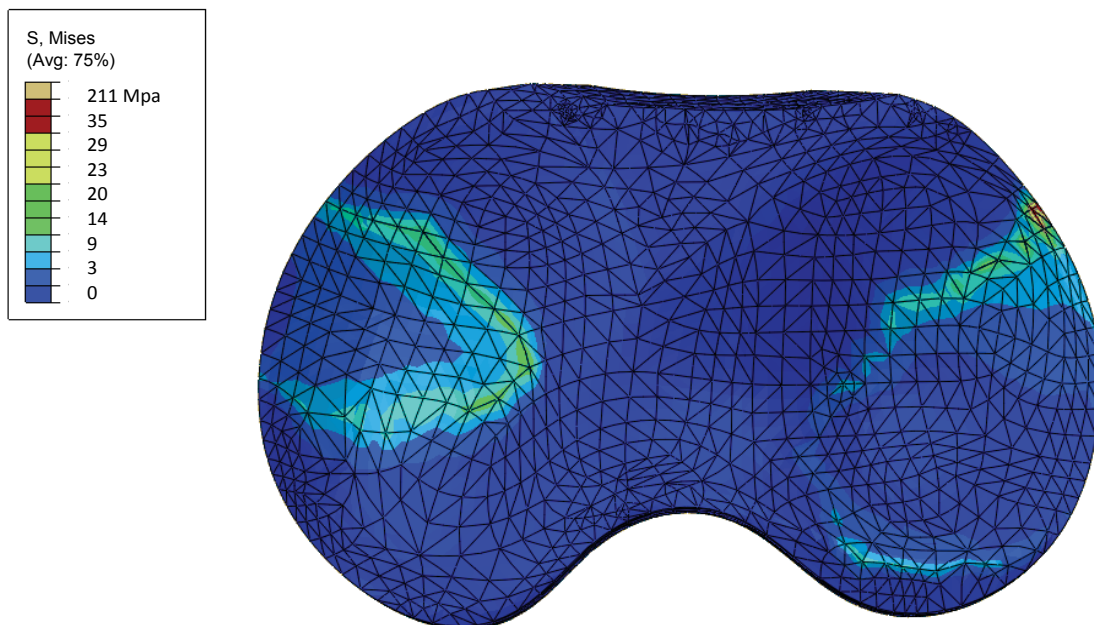
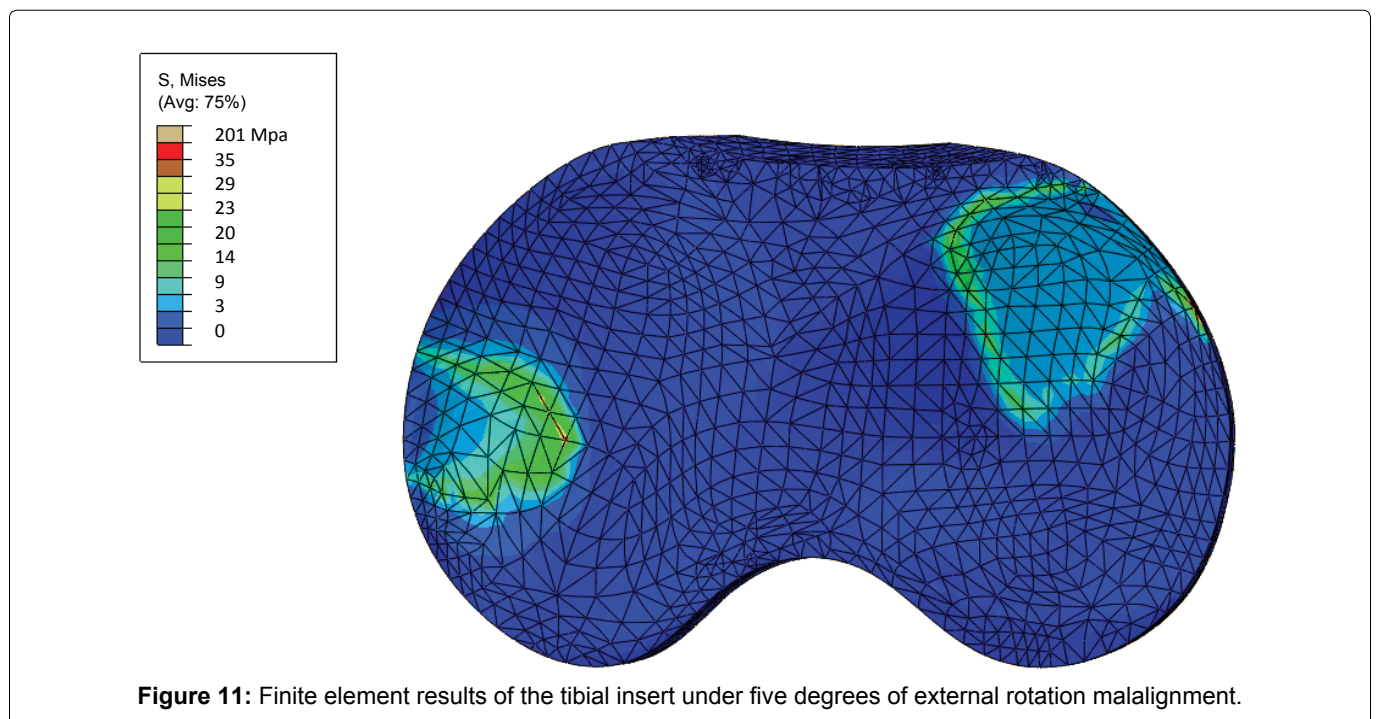
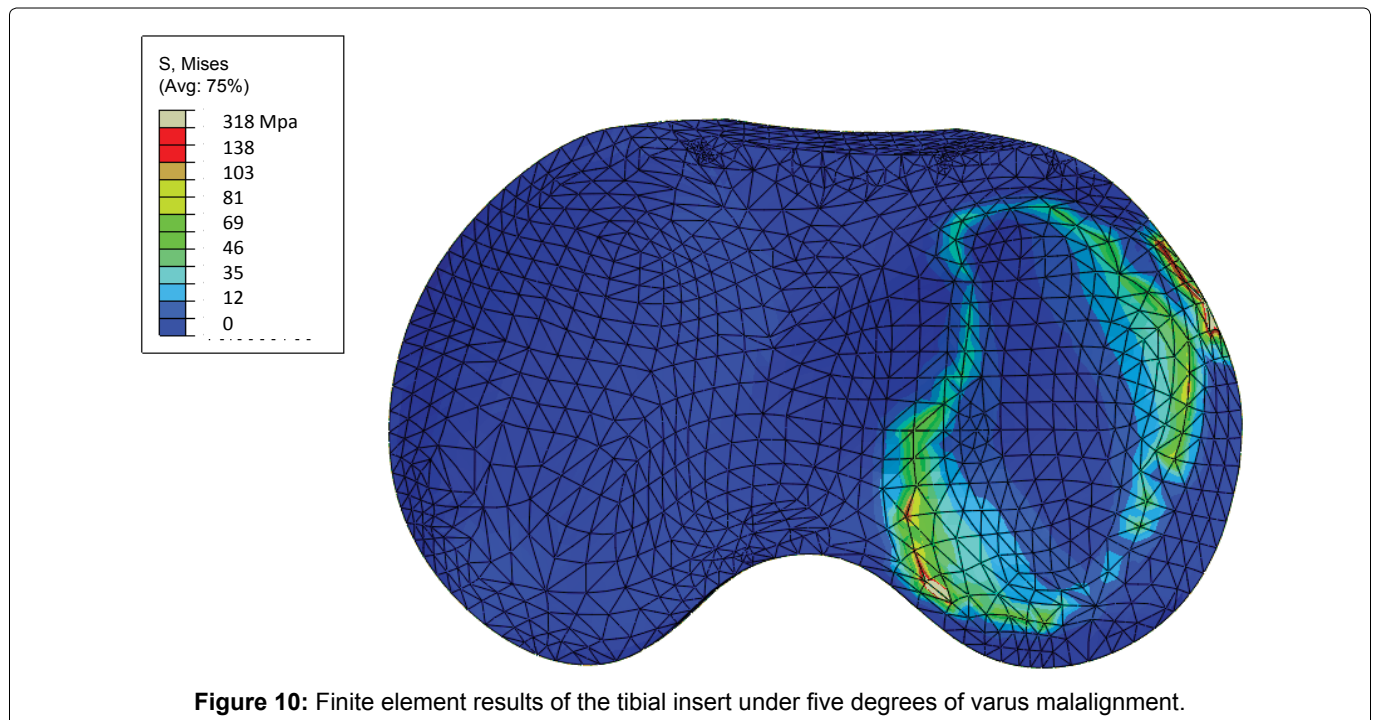


Figure 9: Finite element results of the tibial insert under neutral alignment.

compared to neutral alignment. Five degrees of external rotation (Figure 11) resulted in tibial tray loading in the anterior region of the medial compartment and the lateral-posterior region of the lateral compartment. Stresses in the contact region for neutral and external rotation where both condyles are in contact with the tibial tray result in von Mises stresses of approximately 29 MPa. For varus malalignment where only one condyle is in contact, von Mises stresses in the contact region are significantly higher, on the order of approximately 81-138 MPa.

Discussion

Results of this study show that knee malalignment had a significant effect on the forces and moments at the knee. In particular, the medial-lateral force and the abduction-adduction moment were most significantly affected. Adduction moments are responsible for higher loads in the medial compartment of the knee. Varus and external rotation during gait resulted in a large shift towards adduction moment. The compressive force on the medial compartment due to adduction moment in varus



malalignment combined with the compressive normal force (F_z) along the mechanical axis resulted in an overall greater medial compressive compartment load compared to neutral alignment as observed in finite element results. Results for varus malalignment are consistent with previous studies that show higher adduction moment. The higher stresses concentrated in the medial compartment would result in higher wear rates as observed clinically. For a five degree external rotation, finite element results indicated a change in stress distribution rather than an increase in stress magnitude compared to neutral alignment. It is expected that the stress magnitudes would increase with additional external rotation. In addition, other types of activity not considered here may result in increased stress. This study investigated maximum loading while walking. Other load cases such as running or climbing stairs would result in higher applied loads and consequently higher tibial tray stresses than reported in the current study.

The importance of lower limb alignment in the preservation of femoral-tibial joint health has been shown in many previous studies. Knee adduction moment in subjects with moderate to severe OA ranged from 3.6% x body weight x height to 6.9% x body weight x height, respectively [35]. There is also a higher incidence of Knee Osteoarthritis (OA) in the medial than lateral compartment of the knee [1-3]. Patients having a knee with OA had greater peak knee adduction moment and frontal plane lever arm (perpendicular distance between the ground reaction force and the knee joint center) but less ground reaction force [36]. Bennell, et al. [37] reported that knee loading may be a risk factor for loss of medial tibial cartilage volume. There is also evidence in numerous radiographic, MRI and clinical studies that suggest that knee adduction moment is a strong risk factor for the progression of medial compartment articular cartilage degeneration [8,37-39]. Varus malalignment has been associated with incident of cartilage damage in the medial compartment supporting the idea that varus increases the risk of initial development of osteoarthritis (OA). Increased medial loading due to varus malalignment has been correlated with subsequent disease progression as well [20,36,40-42]. Varus deformity leads to a perpetuating cycle of a decrease in medial joint space and even greater loads on the medial compartment [43].

Knee loading under external rotation is not well understood. Our findings suggest that the Posterior-anterior directed moments also change to more to anteriorly directed moments for external rotation, although not as significantly as varus malalignment. As stated previously, anteriorly directed applied moments to the knee corresponds to adduction moments and are responsible

for higher loads in the medial compartment. External rotation resulted in a shift in the applied loads to the tibial tray, causing more load transfer in the anterior portion of the medial compartment reminiscent of implant wear as described by Knezevich, et al. [7]. Our finding suggests that individuals that externally rotate during gait could experience a higher medial compartment compressive loads and perhaps higher incidence of knee medial compartment articular cartilage degeneration.

The mathematical model developed in this study was written to calculate knee loads during neutral, externally rotated and varus gait cases based on static equilibrium using hip loads at the femoral head. The model is limited by the omission of hip and knee soft tissue loads. The omission of soft tissue loads would result in alterations to forces and moments at the knee. It is also responsible for the loss of a significant torsional moment along the mechanical axis. It can be seen from our result that the abduction-adduction directed moment (M_y) for normal gait is posteriorly directed, which is an abduction moment. From previous work we know that this moment should be anteriorly directed (i.e. Adduction moment), for normal gait. Our simplifying assumptions at the hip including the omission of hip abductor loading are likely responsible for the difference observed here as hip abduction would result in a larger adduction moment at the knee. However, the model is descriptive in the sense that we can see that externally rotated and varus gaits shift the knee moments towards adduction moment, at peak load and throughout the gait cycle.

Results of this study found that for both types of misaligned gait there is a shift towards adduction moment. Increased adduction moment results in an increase in medial condyle loading, supporting our hypothesis. In addition, finite element simulations of misaligned gait demonstrate shifts in load magnitude and load distribution supporting elevated wear in the medial compartment as observed clinically. Therefore, for both types of misaligned gait considered in this study, joint health and post total knee replacement success needs to be more carefully examined.

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