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Abstract

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Keywords

between, load, trans-tibial, socket, limb-dynamic, residual, transfer, mechanics, effects, prosthetic

Disciplines

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**Load Transfer Mechanics Between Trans-Tibial Prosthetic Socket and
Residual Limb — Dynamic Effects**

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2611 Words (Introduction through Discussion)

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Abstract

The effects of inertial loads on the interface stresses between residual limb and trans-tibial prosthetic socket were investigated. The motion of the limb and prosthesis was monitored using a Vicon motion analysis system and the ground reaction force was measured by a force platform. Equivalent Loads at the knee joint during walking were calculated in two cases with and without consideration of the material inertia. A 3D nonlinear finite element model based on the actual geometry of residual limb, internal bones and socket liner was developed to study the mechanical interaction between socket and residual limb during walking. To simulate the friction/slip boundary conditions between the skin and liner, automated surface-to-surface contact was used. The prediction results indicated that interface pressure and shear stress had the similar double-peaked waveform shape in stance phase. The average difference in interface stresses between the two cases with and without consideration of inertial forces was 8.4% in stance phase and 20.1% in swing phase. Although the FE model established in this study is not a full dynamic model, the effects of inertial loads on interface stress distribution during walking were investigated.

Keywords: prosthetic socket, finite element analysis, interface pressure, shear stress, inertia

1. Introduction

A lower-limb prosthesis is often used to restore the appearance and the lost functions to individuals with limb amputation. Market requirements for lower-limb prostheses are increasing, not only in number, but also in quality. Prosthetic socket design is most important in determining the quality of fitting, because the socket provides a coupling between the residual limb and prosthesis. The body weight and inertial force have to be carried by the soft tissues, which are not well suitable for load bearing. An improper load application on the residual limb may cause discomfort or skin damage.

The shape of the socket is not an exact replica of the residual limb. An appropriate modification from the residual limb is required to enable the load transfer more effectively. The modification may depend on the residual limb shape, tissue properties and load tolerance of soft tissues. A quantitative understanding of the relationship between a designed socket and load transfer property is fundamental for an optimal design.

Finite element (FE) methods are widely used in biomechanics and bioengineering to determine stress and strain in complicated systems and have been identified as a useful tool in understanding load transfer in prosthetics (Zhang et al., 1998). Several FE models have been developed based on a certain assumptions (Steege et al., 1987; Quesada and Skinner, 1991; Reynolds and Lord, 1992; Silver-Thorn and Childress, 1997; Sanders and Daly, 1993; Zhang et al., 1995, Zachariah and Sanders, 2000). The development started from simple linear elastic models with simplified 2D or symmetric geometry to the nonlinear models with more accurate geometry. The socket modification, varied external loads to simulate walking, nonlinear mechanical properties, and slip/friction boundary conditions have been addressed in different models. These analyses have provided information on load transfer at the residual limb/socket interface and helped to design a better socket.

However, all models reported so far are static by applying static forces/moments to simulate a single or more phases of gait. The significance in developing dynamic models has been reckoned by researchers who developed the static models (Quesada and Skinner, 1991; Sanders and Daly, 1993; Silver-Thorn and Childress, 1997; Zhang et al., 1998). A dynamic model should be developed to consider not only variable external loads, but also material inertial effects during gait (Zhang et al., 1998).

Dynamic analysis requires more computation resource. Before establishing a full dynamic model, it is useful to estimate how much the material inertia will influence the load transfer during walking. The aim of this paper is to study the effects of material inertia on the knee joint load calculation and interface stress distribution between a trans-tibial residual limb and the prosthetic socket during a whole gait cycle.

2. Methods

A male trans-tibial amputee, 56 years old, 158cm in height, and 81Kg in weight participated in this study. He had more than 5 years experience in using an endoskeletal BK prosthesis with PTB socket and SACH foot.

In this study, the kinematic data of the lower-limb and prosthesis and the ground reaction forces applied on prosthetic foot during walking were measured using a Vicon Motion Analysis System (Oxford Metrics, UK) and a force platform (AMTI, USA). In walking trials, the subject was requested to walk along a 12-m long and 1.2-m wide walkway. Data were

recorded during walking at a sampling rate of 60 Hz.

2.1. Calculation of Loads at the Knee Joint

Inverse dynamics based on the Newton's Second Law was used to calculate the equivalent forces and moments applied at the knee joint during walking. To simplify the problem, assumptions were made that there was no relative movement between the residual limb and socket during walking and only inertial effects in the sagittal plane were considered.

Based on the 3D free body diagram shown in Figure 1, both rotational and translational dynamic equations were setup as follows.

$$M_{oz} - m_1gl_1 \sin \alpha - m_2gl_2 \sin \alpha - m_3gl_3 \sin \alpha + F_{gx}y_g + F_{gy}x_g = I_o \varepsilon \quad (1)$$

$$M_{ox} + F_{gy}z_g + F_{gz}y_g = 0 \quad (2)$$

$$M_{oy} + F_{gz}x_g + F_{gx}z_g = 0 \quad (3)$$

$$F_{ox} + F_{gx} = (m_1 + m_2 + m_3)(r\varepsilon \cos \alpha - r\omega^2 \sin \alpha) \quad (4)$$

$$F_{oy} + F_{gy} - (m_1 + m_2 + m_3)g = (m_1 + m_2 + m_3)(r\varepsilon \sin \alpha + r\omega^2 \cos \alpha) \quad (5)$$

$$F_{oz} + F_{gz} = 0 \quad (6)$$

where F_{ox} , F_{oy} , F_{oz} are force components in X, Y, Z axes applied at the knee joint, and M_{ox} , M_{oy} , M_{oz} are moment components about X, Y, Z axes through the knee joint center O; α , ω and ε are angular displacement, angular velocity and angular acceleration of limb and prosthesis in the sagittal plane; m_i ($i=1, 2, 3$) is the segmental mass of residual limb below the knee joint plus the socket, the shank, and the foot plus the shoe, with the center of gravity C_i , respectively; l_i ($i=1, 2, 3$) is the distance from the knee joint center O to the center of gravity C_i ; F_{gx} , F_{gy} and F_{gz} are the ground reaction forces measured on foot; x_g , y_g and z_g are the distances in X, Y and Z axes between the point of application G and knee joint center O; r is the distance from O to the center of mass C of the whole model, determined by $r = \frac{\sum m_i l_i}{\sum m_i}$, I_o is the moment of inertia of the whole model about the Z axis through the knee joint. Based on the anthropometrical data of the subject and prosthesis used, for given parameters, $m_1 = 1.9$ Kg, $m_2 = 0.3$ Kg, $m_3 = 1.5$ Kg, $l_1 = 0.75$ m, $l_2 = 0.25$ m, and $l_3 = 0.387$ m, the calculated r and I_o were 0.216 m and 0.247 Kg·m².

As shown in Figure 1, the angle of the limb related to Y axis α , angular velocity ω and

accelerations ε were calculated according to coordinates (x_k, y_k) and (x_a, y_a) of markers at the knee joint and the ankle joint. x_g , y_g and z_g were obtained from coordinates at point O and G. The point O can be determined by the marker coordinates at knee joint, and the point G can be given by the force platform.

The load calculation was done in two cases with and without consideration of the material inertia. When the angular velocity and acceleration are assumed to be zero, the static transfer without consideration of inertia effect can be obtained.

2.2. Finite element model

As shown in Figure 2, the finite element model was established based on the actual shapes of the socket, the residual limb surface and the internal bones of the same subject as mentioned in load measurements.

The geometry of the residual limb surface and the bony structure was obtained from 3D reconstruction of MR images conducted on the residual limb with axial cross-sectional images at 6mm interval. To reduce the distortion of the soft tissues in a supine lying position, an unmodified cast was wrapped on the residual limb. The bony structures and the soft tissue boundaries in MR images were identified and segmented using MIMICS v7.10 (Materialise, Leuven, Belgium). The boundary surfaces of different components obtained were processed using SolidWorks (SolidWorks Corporation, Massachusetts) to form surface models. The shape of residual limb was further sent to ShapeMaker (Seattle Limb System) to implement socket modification using the rectification template built-in Shapemaker system.

The models were meshed into 3D 4-node tetrahedral elements using ABAQUS v6.3 FE package (Hibbitt, Karlsson & Sorensen, Inc., Pawtucket, RI). The whole FE model consisted of 22,301 elements and 6,030 nodes.

All materials were assumed to be isotropic, homogeneous and linearly elastic. The Young's modulus was 200KPa for soft tissues, 10GPa for bones and 380KPa for prosthetic liner, and Poisson's ratio was assumed to be 0.49 for soft tissues, 0.3 for bones, and 0.39 for liner (Zachariah and Sanders, 2000; Zhang et al., 1995).

A new approach, automated surface-to-surface contact, was developed to simulate the friction/slip contact conditions between the residual limb and prosthesis and at the same time to be able to consider pre-stresses caused by donning the shape-modified socket on the residual limb (Lee et al., 2003). The inner surface of prosthetic liner and the residual limb surface were defined as the master surface and slave surface respectively. Both surfaces were potentially in contact or separated. According to the master-slave contact formulation and hard contact pressure-overclosure relationship used in ABAQUS, no penetration of slave nodes into master surface and no transfer of tensile stress across the interface were allowed (ABAQUS 2002). Load was transferred only when the two surfaces were in contact.

The analysis was performed in two steps corresponding to the two stages of deformation of the soft tissue. In the first step, a pre-stress analysis was carried out to simulate the donning of residual limb into the socket. No external load was applied in this step. All the bones and outer surface of the liner were given fixed boundaries. Because the shape-modified socket had different inner surface shape from the residual limb surface, there was some overlapping

between residual limb and liner over some regions. In this step, the ABAQUS package would detect the nodes on the residual limb surface, which were initially penetrated into master surface, and those nodes were drawn back to the inner surface of liner. As a result, pre-stresses between the contact surfaces were produced.

In the second step, the outer surface of the liner was rigid fixed assuming the hard socket would offer a rigid support. The external forces and moments during walking were applied at the knee joint with keeping the pre-stress and the deformation due to pre-stress calculated in the first step. When the two surfaces were in contact, any contact pressure can be transmitted between them. The surfaces would separate if the contact pressure reduced to zero. Separate surfaces came into contact when the clearance between them reduced to zero. The penalty method for friction was used to determine the shear stress at the interface with a coefficient of friction of 0.5. The relative slip between the limb skin and the inner surface of the liner was allowed.

In order to discuss the effects of the inertia, a variable δ was defined as equation 7 to describe the average difference of interface stresses during a whole gait between two cases with and without consideration of the inertial effect.

$$\delta = \frac{\sum_{i=1}^n |y_{1i} - y_{2i}|}{\sum_{i=1}^n y_{1i}} \times 100\% \quad (7)$$

Here y_{1i} and y_{2i} are the results of stresses predicted by FE model under the load calculation at the knee joint with or without consideration of inertia loads.

3. Results

The experimental results obtained from the motion analysis system and force platform were processed using Matlab 6.5 (The MathWorks, Inc.). According to Equations 1 to 6, the equivalent forces F_{ox} , F_{oy} and moment M_{oz} in the two cases with or without inertia effects were compared in Figure 3. Because the acceleration in frontal plane and transversal plane was not considered, there would not be difference in F_{oz} , M_{ox} and M_{oy} between the two cases.

Comparing the loads with or without consideration of inertial effects, there was an obvious difference in F_{oy} between the two cases in the swing phase (65-100% of gait cycle), but small difference in the stance phase (0-65% of gait cycle). For F_{ox} and M_{oz} , the results in the two cases have large difference not only in the swing phase but also in the stance phase. The maximal difference of magnitude is up to 19% for F_{ox} at the first peak and 27% for M_{oz} in the second peak (Figure 3).

Figure 4 shows the FE predicted interface pressure distribution at the moment of 20% of gait cycle. The pressure is defined as the stress perpendicular to the contact interface. The highest pressure is 292 KPa at the middle patella tendon (PT). The other peak pressure regions include popliteal depression (PD), lateral tibia (LT) and media tibia (MT) and the peak pressures are 267 KPa, 206 KPa and 121 KPa, respectively. These regions are believed to be load-tolerant areas.

Figures 5 and 6 display the comparison of interface pressures and resultant shear stresses

during a gait cycle over PT, PD, LT and MT regions predicted from FE analysis between the two loading cases with and without consideration of inertial effects. The resultant shear stress is the combination of longitudinal and circumferential components of shear stress in the plane of contact interface.

Generally speaking, all the pressure curves are in double-peaked shape, which is similar to the ground reaction force. The effect of bending moment M_{oz} in the sagittal plane can be seen from comparison of peak pressure curves. Around the first peak, the ground reaction force produces a moment to extend the limb, and such a moment increases the pressures over anterior-proximal and posterior-distal sides, and decreases pressures over anterior-distal and posterior-proximal sides. However, around the second peak, the ground reaction force produces a moment to flex the limb, and the effects on pressure are opposite. From Figure 5, it can be seen that the first peak pressure over anterior-proximal side, such as PT, is larger than the second peak one, while over other three regions the first peak pressure is smaller than the second peak one.

Comparing the pressure curves during the stance phase, the pressures predicted in two loading cases do not change a lot, except over the popliteal depression region. At the beginning of swing phase, even though the ground reaction forces disappear, angular acceleration is positive. A force couple is applied at proximal-posterior and distal-anterior regions of the socket to accelerate the prosthesis extension, which induces decreased pressure over PT region and an increased pressures over PD, LT and MT regions. This can be seen from Figure 5 that considering inertial effects will induce a faster deduction in pressure over PT region and slower over the other three regions.

Table 1 gives the differences in predicted peak stresses between two loading cases. The values of δ (Table 2) show the average difference produced by inertial effects in the stance phase, swing phase and whole gait cycle, according to the curves in Figures 5 and 6. The average difference in both pressure and shear stress prediction is 8.4% during stance phase and up to 20.1% during swing phase.

4. Discussion

It is believed that FE analysis, if developed properly, can be strong potential to offer information for the improvement of the prosthesis design. The development of FE models was phased into three generations (Zhang et al., 1998). The third generation, dynamic analysis, is expected to come with consideration of both variable external loads and inertial effects. This study has made a stride forward the goal of the dynamic analysis.

In this study, equivalent forces and moments at the knee joint were calculated during walking. The effects of material inertia during walking were estimated by comparing the results of loading transfer in two cases with and without consideration of the inertia. A 3D nonlinear FE model was established to predict stress at the residual limb/prosthetic liner interface, with considering actual geometry, socket modification, friction/slip boundary condition and large deformation. The effects of inertia on the interface stresses can be estimated by applying two groups of loading cases at the knee joint. The peak pressures predicted over the pressure-tolerant regions are in the range of the clinical measurements (Zhang et al., 1998).

Because the ground reaction forces make main contribution to the equivalent loads applied

at the knee joint in the stance phase, the interface pressures and shear stresses don't change significantly no matter the inertia effects were considered or not. But in the swing phase, there is no ground reaction force and the inertia plays a primary role in the calculation of equivalent loads. As a result, interface pressures and shear stresses are considerably different between two loading cases with and without considering inertial effects.

Although the FE model established in this paper is not a full dynamic model, the inertial effects on the prediction of interface stress distribution were investigated during walking. The findings in this paper will be significant for improving our understanding of interface biomechanics of residual limb/ prosthetic socket system.

In future study, dynamic FE models should be developed, in association with kinematic information of the limb and prosthesis, material inertia and variable ground reaction forces during walking. The model should be further validated by experiments.

Acknowledgements

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Table 1

Comparison of the peak magnitude of pressures and shear stresses without inertial effects to those with inertial effects

Differences of peak magnitude (%)	Pressure				Resultant shear stress			
	PT	PD	LT	MT	PT	PD	LT	MT
Heel contact	-16.5	+17.8	+19	+3.1	-3.7	+14.2	+8.9	+0.8
Foot-flat	-4.6	-1.1	-4.1	-1.8	-3.2	-2.5	-3.7	-3.8
Mid-stance	+2.1	+4.2	+1.3	+0.8	+8.8	-1.3	+9.1	+1.6
Heel off	-3.9	-11.9	-5.2	-3.8	-3.0	-28	+5.9	-6.5
Toe off	+50	-7.3	-25	-17.8	+36	-41	-34	-15.6

Table 2

The average difference of interface pressures and shear stresses predicted from FE model

Average difference δ in different phases (%)	Pressure				Resultant shear stress				Average of four regions
	PT	PD	LT	MT	PT	PD	LT	MT	
Stance phase	11.9	5.4	3.5	3.8	11.4	13.4	13.8	3.9	8.4
Swing phase	33.2	5.4	18.3	16.4	16.7	37.2	19.4	13.4	20.1
Whole gait cycle	19.3	5.4	8.7	8.2	13.2	21.8	15.7	7.2	12.5

Figures:

Figure 1 3D model for calculation of loads at knee joint

Figure 2 Finite element model for residual limb and prosthetic socket

Figure 3 Equivalent dynamic loads applied at knee joint

Figure 4 Pressure distributions obtained from FE analysis at 20% of gait cycle. The numbers express the peak values, and there is 42 KPa between two adjacent lines

Figure 5 Comparison of pressures on residual limb with or without consideration of inertial effects during the whole gait cycle

Figure 6 Comparison of resultant shear stresses on residual limb with or without consideration of inertial effects during the whole gait cycle

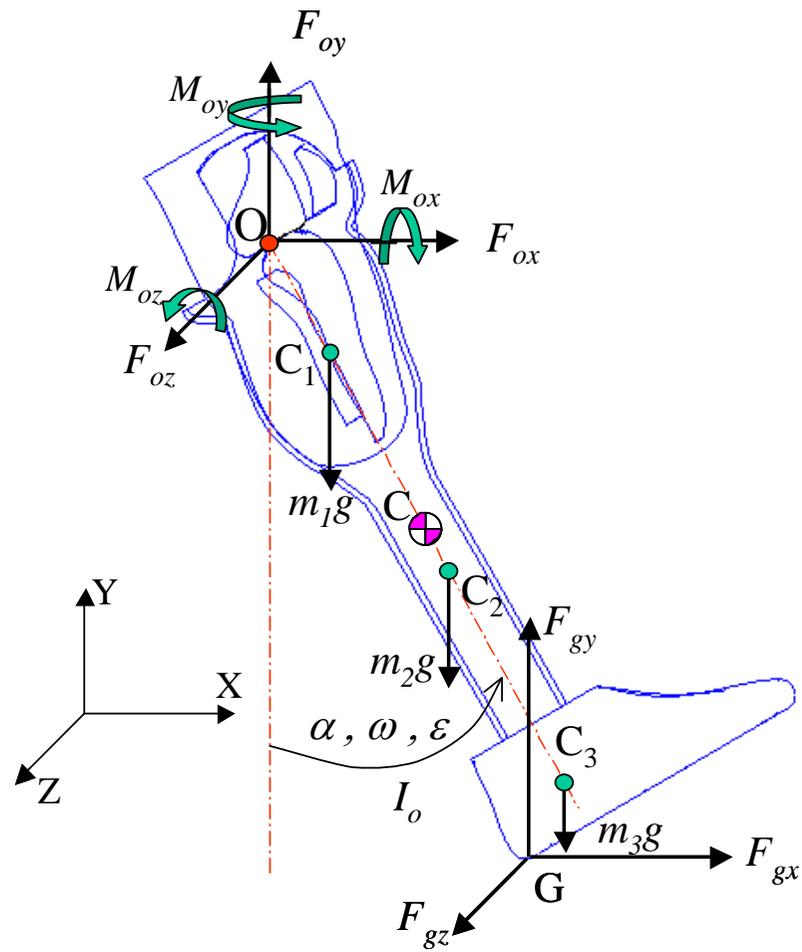
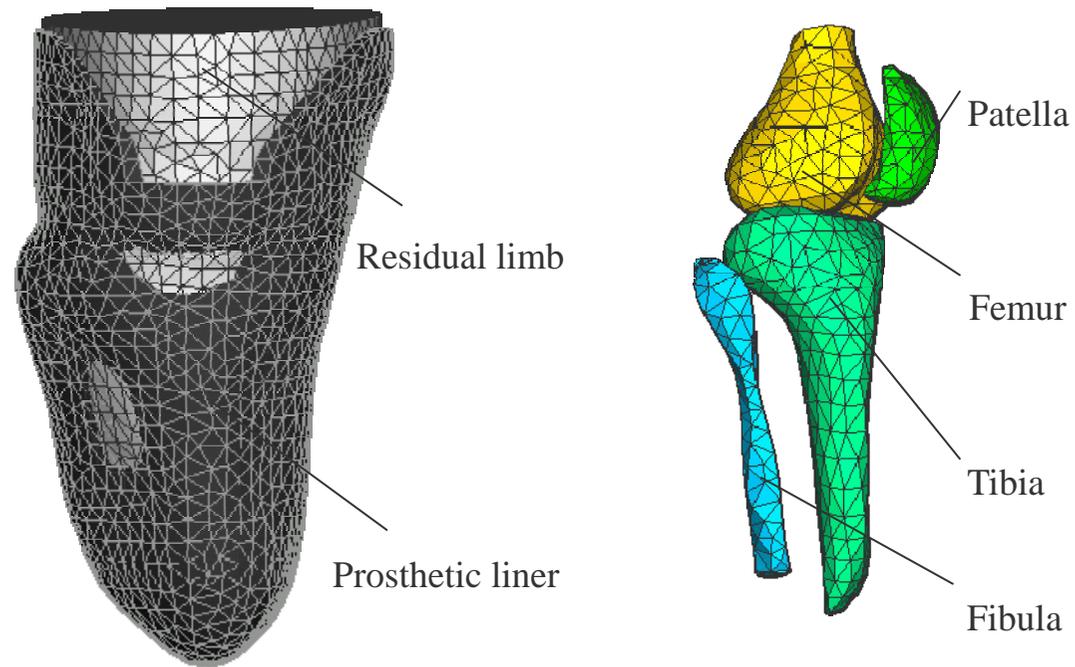


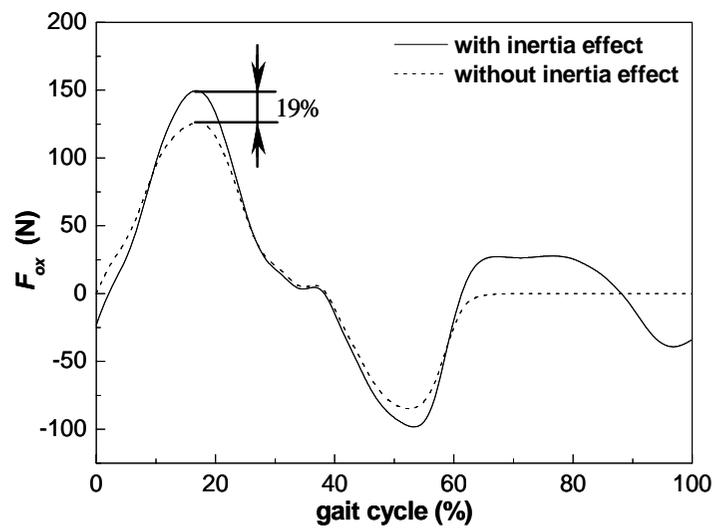
Figure1



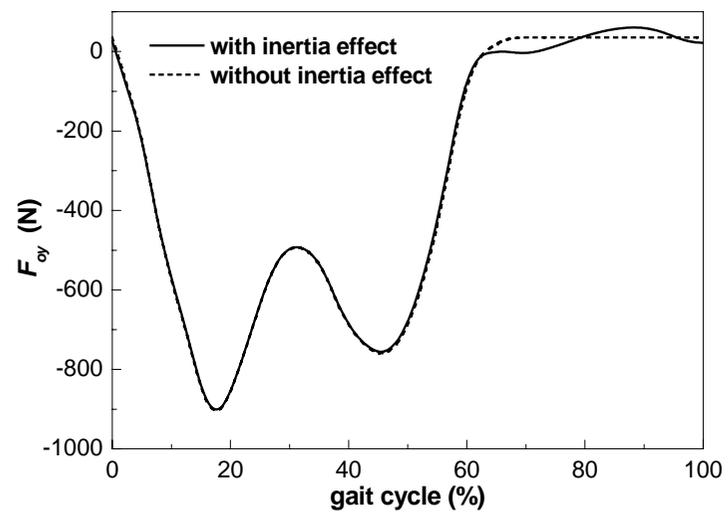
(a) Anterior view of FE model

(b) Lateral view of meshed bones

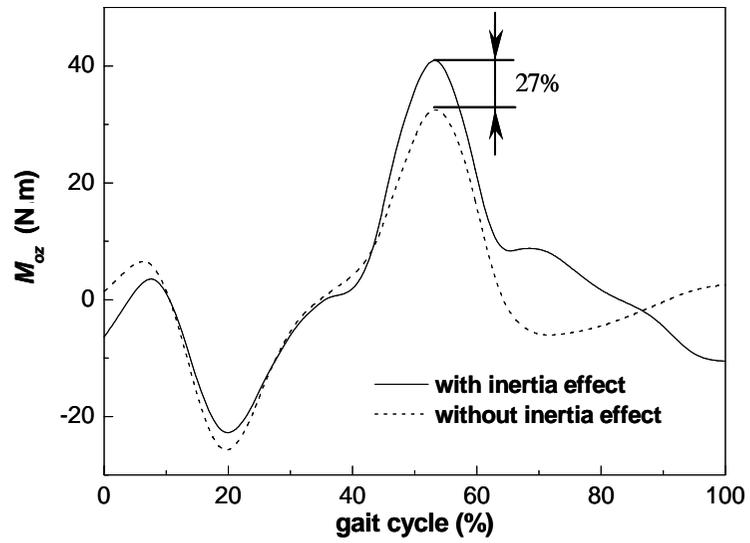
Figure 2



(a) Force along X axis

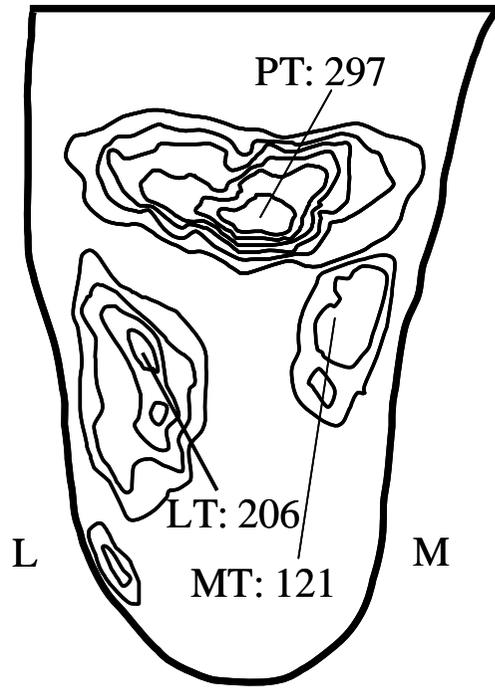


(b) Force along Y axis

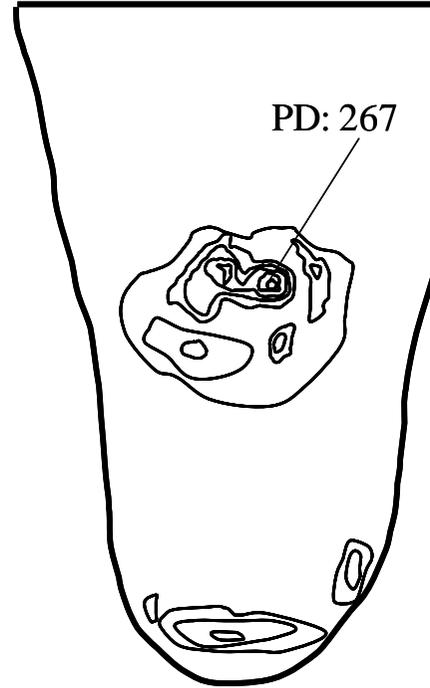


(c) Moment about Z axis through knee joint

Figure 3

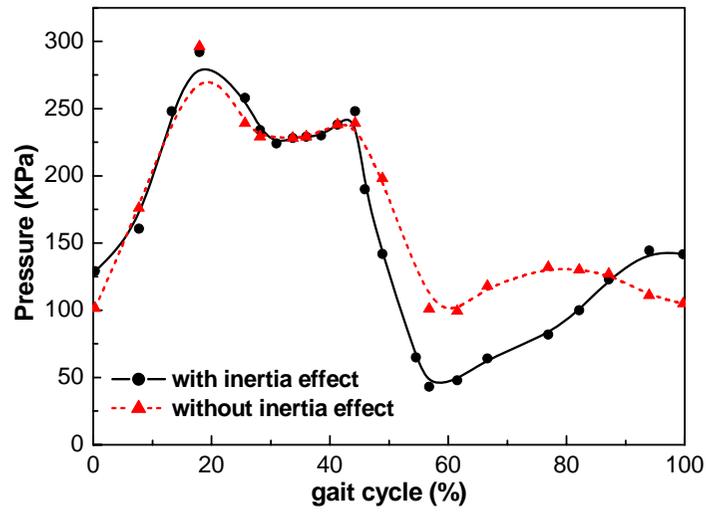


(a) Anterior view

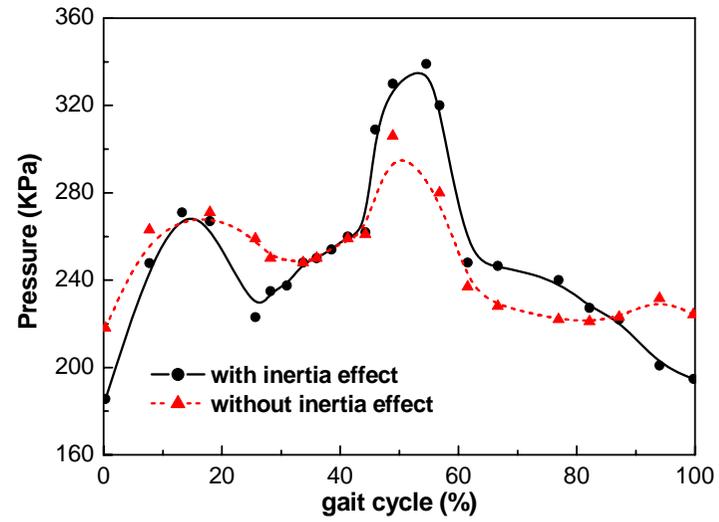


(b) Posterior view

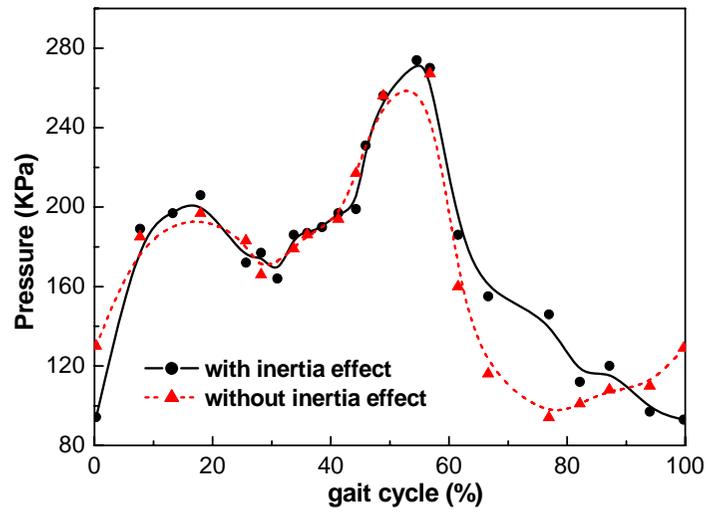
Figure 4



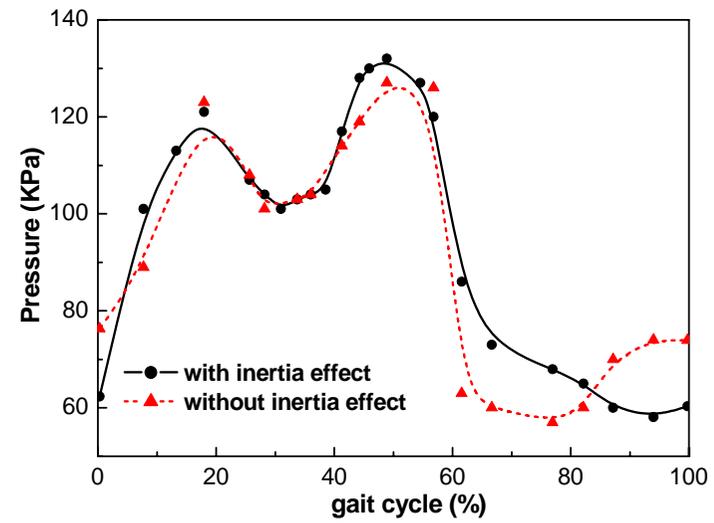
(a) Patella tendon



(b) Popliteal depression

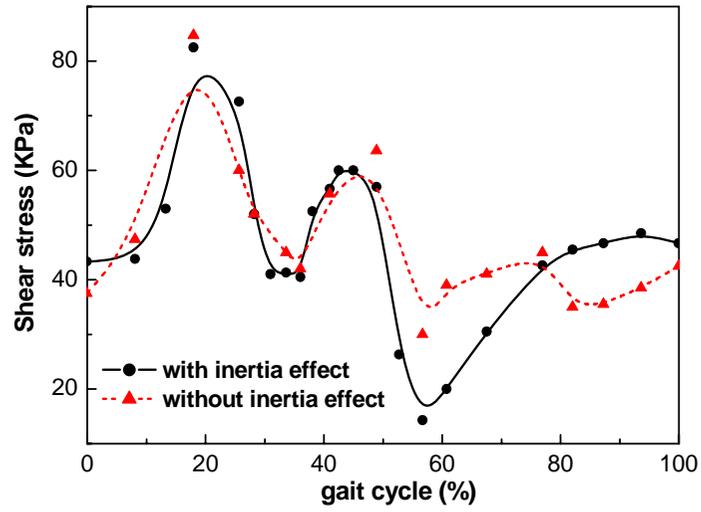


(c) Lateral tibia

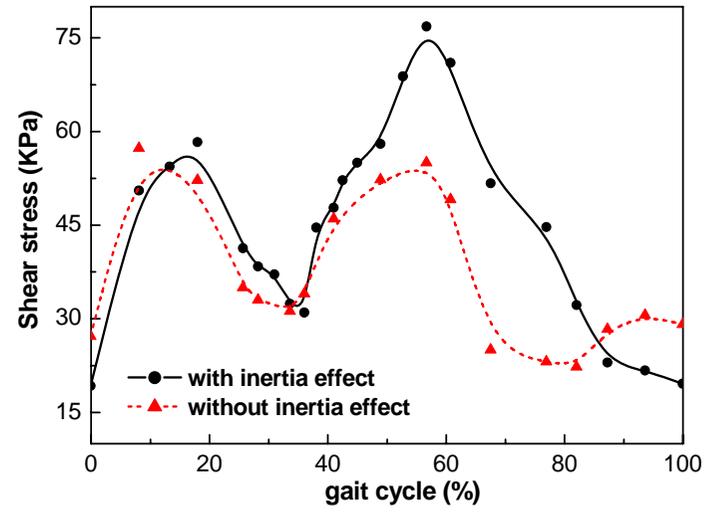


(d) Medial tibia

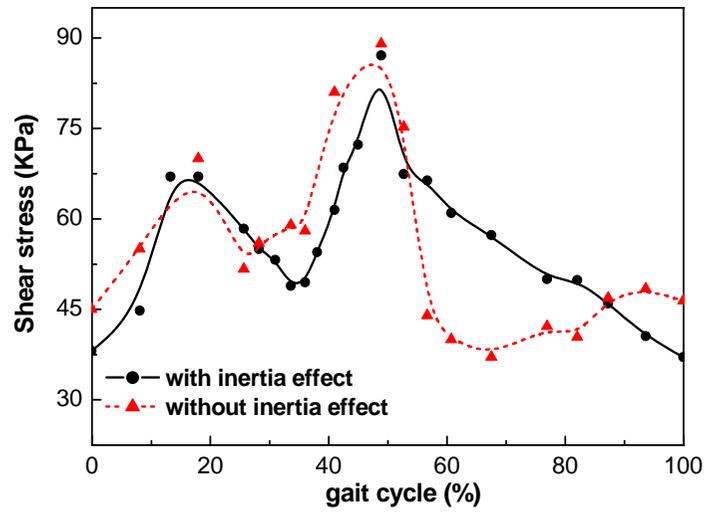
Figure 5



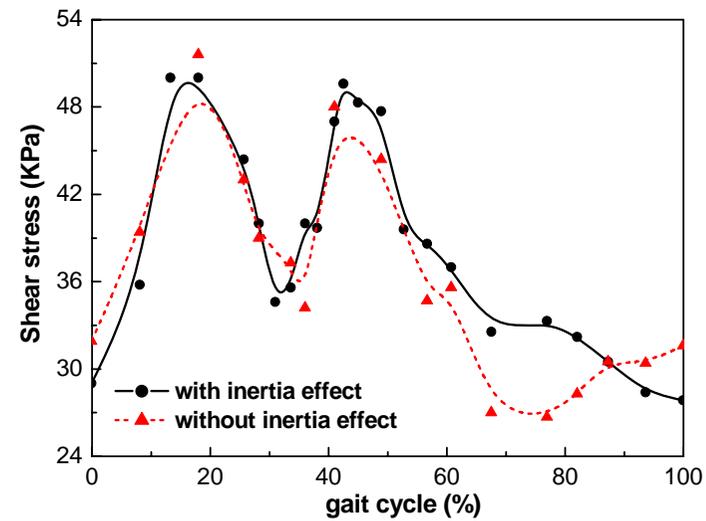
(a) Patella tendon



(b) Popliteal depression



(c) Lateral tibia



(d) Medial tibia

Figure 6