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Finite Element Analysis of the Contact Interface Between Trans-femoral Stump and Prosthetic Socket*

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Abstract—**Transfemoral amputees need prosthetic devices after amputation surgery, and the interface pressure between the residual limb and prosthetic socket has a significant effect on an amputee's satisfaction and comfort. The purpose of this study was to build a nonlinear finite element model to investigate the interface pressure between the above-knee residual limb and its prosthetic socket. The model was three-dimensional (3D) with consideration of nonlinear boundary conditions. Contact analysis was used to simulate the friction conditions between skin and the socket. The normal stresses up to 80.57 kPa at the distal end of the soft tissue. The longitudinal and circumferential shear stress distributions at the limb–socket interface were also simulated. This study explores the influences of load transfer between trans-femoral residual limb and its prosthetic socket.**

I. Introduction

Transfemoral amputation patients need prosthetic devices after amputation surgery in order to restore the self-esteem of the patient by completing his appearance and regain their functional mobility as much as possible. The prosthetic socket is the primary interface between amputated limb and the prosthesis, which transfers the body loads generated during the gait and plays a significant role in determining the quality of the fit. Appropriate socket fitting can have a significant effect on the patient's comfort, mobility and level of satisfaction with their prosthesis[1].

Some researchers have attempted to evaluate the load transferred at the interface between the residual limb and its prosthetic socket either through completion of clinical assessments that use different types of transducers[2-6]or through simulation techniques[7-9]. Knowledge of the biomechanics of load transfer at this interface would enable objective evaluation of prosthetic fit, and might advance socket design[10]. The load distribution on the stump is an important consideration in prosthetic design.

However, the transducers used in the clinical measurement can produce stress concentrations over the soft tissues, can modify the gait, and the results are valid only at the point where the transducer is located[11, 12]. As a complement to experimentations, numerical method based on finite element (FE) analysis has been identified as a potential method for prediction and evaluation of the load transfer between the stump and a socket. The FE method has been used in many practical engineering problems and medical fields, which can examine the stresses in the entire residual limb including the surface and internal tissues and predict the load transfer prior to socket fabrication. In theory, the accuracy of the FE analysis results depends on model establishment, simplifications and assumptions. In the last decade, FE models for above-knee (AK)[9, 12] and below-knee (BK)[13-15] amputees have been developed to establish the stress-strain state in the interaction between the socket and the stump.

Computed tomography (CT)[16] or magnetic resonance imaging (MRI)[17] was used to obtain the geometry of the residual limb and prosthetic socket to build the FE model. To simplify the simulation and reduce the computational time, some assumptions were made. One simplification is that the data from gait analysis are used to define the magnitude and direction of the loads applied to the residual stump and socket. Another commonly adopted assumption is that the bone and the soft tissue are fully connected as one body assigned with different mechanical properties.

Generally, the finite element analysis includes two separated steps. The first one simulates the residual limb donning into the socket. Zhang[9] introduced a manually radial nodal displacement over specific areas of the stump to simulate the donning process. Zachariah and Sanders[18] used an automated contact method which automatically detect any overlapping of interface nodes and impose a non-penetration condition constraint to the overlapped nodes. However, those two cases do not match with the actual socket donning procedure. Lacroix[16] used an explicit finite element method to apply displacement vector on the proximal part of the socket which is not contacted with the stump before the simulation. It is more reliable than the previous studies. Lacroix[16] obtained the mean maximum pressure is 4 kPa (SD 1.7) and the mean maximum shear stresses are 1.4 kPa (SD 0.6) and 0.6 kPa (SD 0.3) in longitudinal and circumferential directions, respectively. Maintaining the stress-state generated during the first step, the second one starts, when the load obtained from gait analysis is applied over the stump or over the socket to see the distribution of the interface pressures and shear stresses. In this step, the above automated contact method was maintained between the two separate structures of the residual limb and the socket. To offer the friction/slip conditions at the interface, a coefficient of friction was given between the two contact bodies. Slipping was allowed if the shear stress exceeded the frictional limit[17].

The mechanical properties of the bone and the socket are linear elastic isotropic in common. The soft tissue mechanical behavior was used to be the same, but recently proposed viscoelastic[19] or hyperelastic[13, 14] formulations. For the soft

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tissue is elastic than the socket, the deformation of the socket was ignored.

A better understanding of the load transfer between the stump and its socket is necessary for increasing the overall knowledge of prosthetic biomechanics and for developing comprehensive FE model which can simulate the real situation accurately. For most similar studies are focuses on the below-knee model. The objective of this study was to simulate the contact at the residual limb-socket interface of the above-knee amputee to understand the stress-state generated through three loading conditions during walking, which is foot flat, mid-stance and heel off.

II. METHODS AND MATERIALS

A. Geometries

The geometries of the residual limb surface and the internal bones were captured from a male trans-femoral amputee, 41 years old, 171 cm tall and 70 kg in mass who had more than 10 years experience using his prosthesis with a total contact quadrilateral socket, a load-bearing knee joint with locking function and a SACH foot. Computed tomography (CT) scanning data were obtained from the residual limb in supine lying with hip and knee extended position, which were scanned in a multi-slice spiral CT scanner (GE LightSpeed16, USA) in slices 5 mm thick. The bones, the limb surfaces and the socket surfaces were identified and segmented using software Mimics v10.01 (Materialise, Leuven, Belgium). In this model, we assumed that the shapes of the residual limb and rectified socket are the same similar with Zachariah[18].

The surfaces were then imported into Hypermesh version 8.0 (Altair). The soft tissue model was generated by geometrically subtracting the bones from the limb solid. The solid models representing bones, soft tissues and socket were then meshed with 3D tetrahedral elements. The final model (Fig.1) contained 1,808 tetrahedral elements for bones, 16,288 and 1,496 tetrahedral elements for soft tissue and socket, respectively, totaling 19,592 elements to represent the whole model.

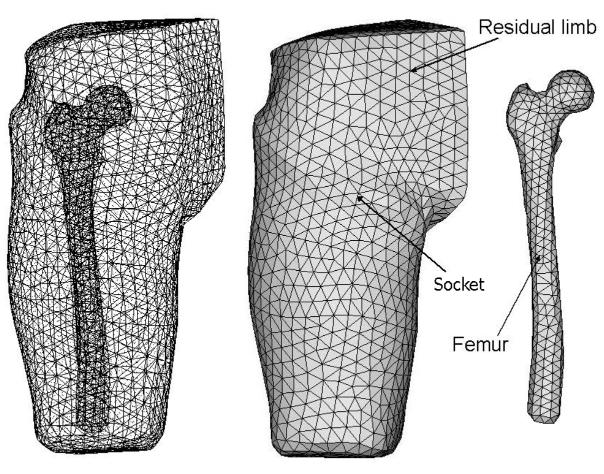

Figure1. FE mesh of the residual limb, prosthetic socket and bones.

B. Material properties

The mechanical properties of the bone and socket were assumed to be linearly elastic, isotropic and homogeneous. The bone and the socket were assigned with Young's modulus of 15 GPa and Poisson's ratio 0.3[9, 20]. After measuring 9 above knee amputees in the relaxed muscle state, Malinausas et al.[21] reported an average modulus from 53.2 to 141.4 kPa, depending on the measured sites. When the amputee donned a socket, the underlying muscles were in a tension state, which would increase the modulus. Therefore, Young's modulus is determined as 200 kPa and Poisson's ratio is assumed to be 0.45 for soft tissue in this model. To consider the effect of large deformation, non-linear hyper-elastic Mooney-Rivlin material model[22] of ABAQUS® was also used for the soft tissue as a comparison (Table 1).

Table I. DETAILS OF MATERIAL PROPERTIES AND ELEMENT NUMBERS FOR THE FE MODEL

	Bone	Socket	Soft tissue
Young's modulus (MPa)	15000	15000	C_{10}＝85.5kpa C_{01}＝21.38kpa
Poisson's ratio	0.3	0.3	0.459
Number of Elements	1,808	1,496	16,288

C. Boundary conditions and analysis steps

The bones and soft tissues were modeled as one body. The residual limb and socket were modeled as two separate parts. To simulate the slip between the socket-limb interface, coefficient of friction of those two contact surfaces ('surface to surface contact algorithm' of ABAQUS®) was assumed as 0.5[23], since the coefficient of friction between the skin and polythene under normal conditions was around 0.5[24]. The inner surface of the socket and the residual limb surface were defined as master and slave surfaces, respectively.

There were two analysis steps. The first step was to establish the pre-stress condition from donning the limb into the socket, which applied a load of 50 N on the upper surface of the soft tissue. In this step, the bottom surface of the socket was fixed. At the second step, the pre-stresses and the deformations calculated in the first step were kept and external loading was applied at the bottom surface of the socket with the upper surface of the soft tissue was fixed. Three load cases were applied separately to simulate the loading conditions at foot flat, mid-stance and heel off during walking[17].

III. RESULTS

The stress distributions can be shown in socket and stump, but focus on the surface of the stump with the socket–stump interaction will be made. Fig. 2(a) shows the normal stress distributions predicted from the first step to simulate the pre-stress, which were more evenly distributed.

Fig. 2, (b), (c) and (d) displays the normal stress distribution obtained from the second step analysis in the bottom surface when loadings simulating the three walking phases were applied with pre-stress considered. The normal stresses up to a maximum of 80.57, 52.41 and 73.37 kPa, respectively. In all this three cases, the highest normal stress was produced at the ischial bearing areas up to 119.3, 89.98 and 104.1 kPa, respectively.

Fig. 3 (a), (b) and (c) shows the longitudinal shear stress distribution and (d), (e), (f) shows the circumferential shear stress distribution at the limb–socket interface when loadings simulating the three walking phases were applied with pre-stress considered. The maximum value is 25.65 kPa and 103.6 kPa over the posterior of the socket brim region for the longitudinal shear stress and circumferential shear stress, respectively.

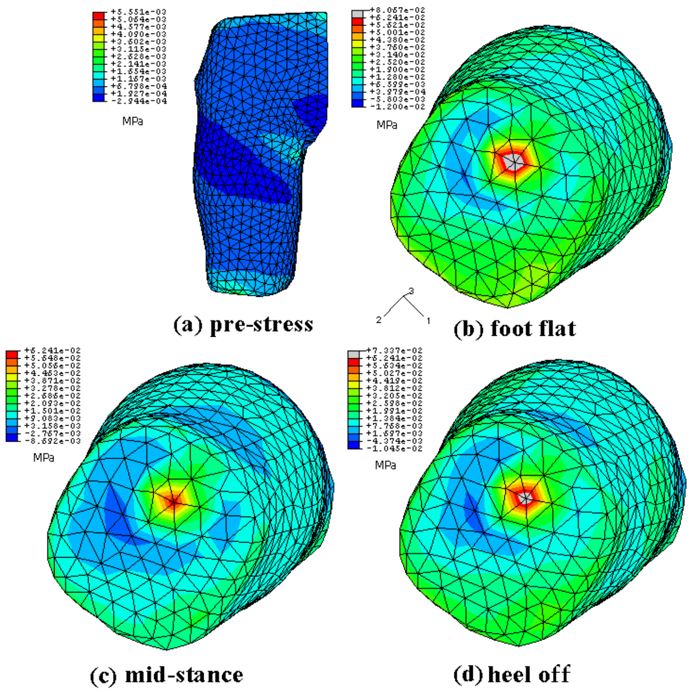

Figure 2. The simulated pre-stress (a) and
the normal stress in the bottom surface of the stump (b), (c), (d).

IV. Discussions

It was assumed in the FE model that the femur position did not change within the soft tissue at different loading cases. The assumption was made because (1) loads were added at the bottom of the socket surface so that the directions of loads were not affected by different femur position, (2) ascertain and prediction the position of femur within the soft tissues could be difficult.

Zhang and Mak[9] obtained a maximum pressure of 65 kPa at the distal end in a full distal-end loading model using a coefficient of friction of 0.5 during stance, which is similar with this study. These results are comparable to this study where the maximum peak pressure of 80.57, 52.41 and 73.37 kPa at the bottom of the residual limb surface in three walking load cases. But, using the Mflex Sensor Distribution System Mu C et al.[23] measured the interface contact pressure on above-knee residual limb at mid stance during walking resulted in a maximum pressure of 258.90 kPa, which is considerably higher than those presented in this study. The

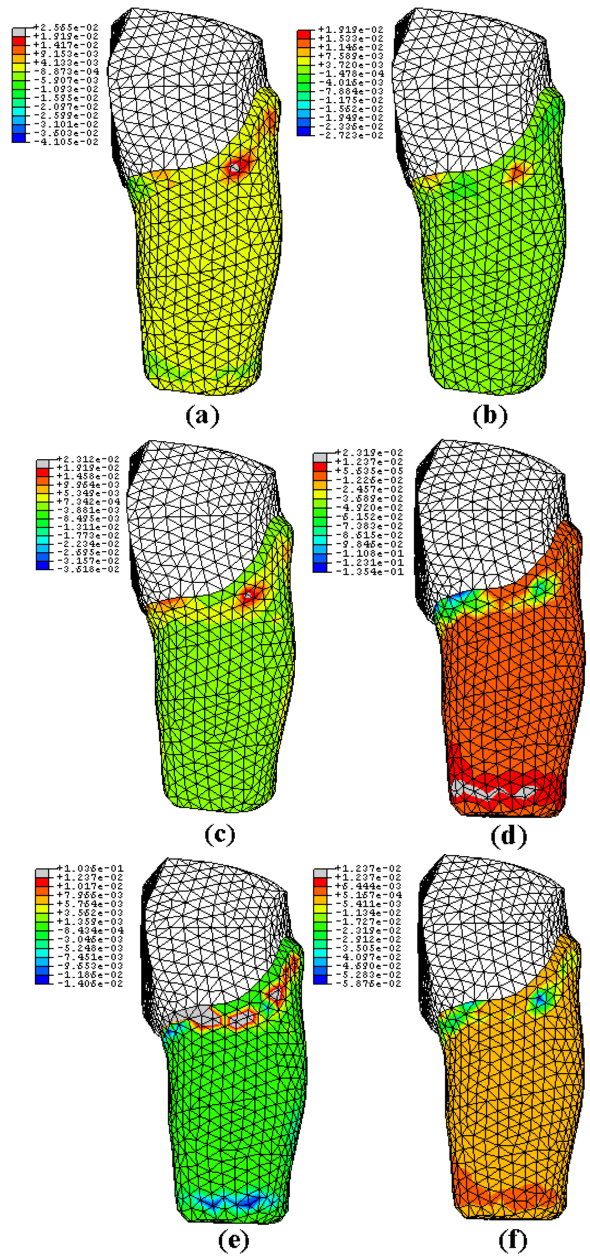

Figure 3. The longitudinal shear stress for loading conditions at foot flat (a), mid-stance (b) and heel off (c) and the circumferential shear stress for loading conditions at foot flat (a), mid-stance (b) and heel off.

differences in the magnitude of the pressures can be associated mainly with that the pressure sensors placed between the stump and the socket inevitably disturb the mechanical condition at the interface and increase concentrated force.

Simulation of donning the residual limb into a rectified socket has been implemented in some models by applying radial displacements to the nodes of the unrectified socket to deform it into the rectified socket shape[10] or axial displacement to the residual limb with the socket fixed[16], which is a challenging task involving large motions and wiggling. The real procedure of socket donning was not simulated in this study. A commonly adopted assumption used in this investigation is that the shapes of the residual limb and rectified socket are the same[18]. The stresses applied to the residual limb after donning into the rectified socket were ignored under the above assumption. The assumption was made because: (1) the socket shape modifications aiming to redistribute the load to load-tolerant regions were not sensitively for above-knee socket than below-knee socket, (2) reduce computational time required for simulating large sliding action. Instead, a pressure of 50 N on the upper surface of the soft tissue was applied to simulate the pre-stress condition from donning the limb into the socket. As a result, the maximum normal stress in this model is 5.55 kPa, which is comparable with Lacroix (the mean maximum pressure is 4 kPa $\pm$ 0.6). The results are informative in a relative sense rather than in an absolute sense, which need to be validated experimentally in the future.

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