FINITE ELEMENT METHOD FOR EVALUATION OF TRANSFEMORAL PROSTHESIS SOCKETS

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ABSTRACT

The socket of a prosthesis is an important part that serves as the interface between the residual limb and the prosthesis. Its aim is to effectively integrate the prosthesis as a functional extension of the body. This goal is limited by the compliance of soft tissue of the residual limb and its local tolerance to externally applied forces. In fact, the soft tissue around a residual limb is not well suited to load bearing and an improper load distribution may cause pain and skin damage. Therefore, correct shaping of the socket for appropriate load distribution is a critical process in the design of lower limb prosthesis sockets.

In this study, a nonlinear finite element model was created and analyzed to determine the pressure distribution between a residual limb and the prosthesis socket of a transfemoral amputee. This analysis was performed in an attempt to develop a process that allows healthcare providers and engineers to simulate the fit and comfort of transfemoral prostheses in order to evaluate the fit of socket shape in the human gait cycle.

Three-dimensional models of the residual limb and socket were created using magnetic resonance imaging data; the models were composed of 17 layers, each separated by 10 mm. The residual limb includes four parts: skin, fat, muscle, and bone. The interaction between them was modeled using a tie condition that simulates bonding between two materials. The load applied to the residual limb at the top of bone was calculated from kinematic parameters. The finite element analysis was carried out in two steps. The first step is donning process, and the second step is in the human gait cycle including stand and swing phases.

The normal stress distribution in the residual limb was observed during the gait cycle. The procedure developed through this work can be used by future researchers and prosthesis designers in understanding how to better design transfemoral prostheses.

1. INTRODUCTION

An amputated limb is one of the most physically and psychologically devastating events that can happen to a person. Not only does lower limb amputation cause major disfigurement, it also renders people less mobile and at risk of loss of independence [1]. There are nearly 2 million people living with limb loss in the United States [2]. For them, the main causes of limb loss are vascular diseases (54%) including diabetes and peripheral arterial disease, trauma (45%), and cancer (less than 2%) [2]. Approximately 185,000 amputations occur in the United States each year [3].

An amputation that occurs through the femur is known as transfemoral prosthesis amputation. A transfemoral prosthesis is used as an artificial limb to restore the amputee's mobility functions for their daily life activities. The transfemoral prosthesis is attached in contact with part of the above-knee residual limb. The uppermost part of the prosthesis is called the prosthesis socket, which surrounds the residual limb and acts as a medium to transfer the load from the residual limb to the prosthesis [4]. The skin and the soft tissue of the residual limb experiences severe stress and excessive distortion during gait positioning such as sitting, standing, taking steps, and walking [5].

The objective of this study was to develop a method to observe the distribution of pressure on the surface of a residual limb. Using this method will allow prosthetic designers to evaluate a patient's transfemoral prosthesis fit analytically and make scientifically sound decisions on how to enhance the prosthesis. It is hoped that this method will reduce the number of refittings needed for the patient. In an effort to achieve this objective, the pressure distribution between the prosthesis socket and residual limb were analyzed using the finite element analysis (FEA) method. The analysis employed linear elastic material properties to model the soft tissue of the limb. Two cases of load application were considered: standing and walking. A finite element (FE) model was developed and solved using the nonlinear dynamic explicit method in LS-DYNA.

2. GEOMETRY MODELING

The subject in this study was a male (age 35) with a left-side transfemoral amputation. He had a height of 169 cm and weighed 63 kg without his prosthesis. The prosthesis incorporated an MCCT (Muanual Compression Casting Technique, IRC) socket [6], a Nabco prosthesis, and an Ottobock foot.

Magnetic resonance imaging (MRI) was used to obtain data of the residual limb with socket prosthesis. The patient wore the socket prosthesis during MRI. The residual limb with socket prosthesis was captured as 17 layers with 10 mm separation perpendicular to the sagittal plane. Subsequently, the three-dimensional (3D) surfaces of bone, muscle, fat, and skin were obtained. The MRI data were loaded into parallel planes and contours manually drawn per slice and lofted into the 3D body by means of a solid modeling software (PTC Creo Parametric) (Fig. 1). The model of socket was offset from the shape of skin. The 3D solid models of residual limb and socket were imported to LS-Prepost for meshing and creating the properties of simulation. Then, the simulation was run by LS-DYNA.

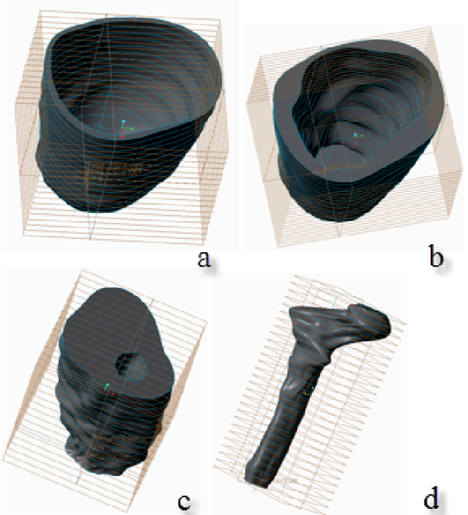

Figure 1. 3D model of parts of residual limb. (a. Skin; b. Fat; c. Muscle; d. Bone)

3. FINITE ELEMENT ANALYSIS PROCEDURE

3.1 Element type

Tetrahedral meshes were generated on the four parts (skin, fat, muscle, and bone). These type of meshes are generally preferred over hexahedral meshes for free-form complex geometries as the former are computationally more cost effective [7] and easier to apply.

3.2 Material models

The mechanical properties of all parts of residual limb and socket were assumed to be linearly elastic, and therefore obey Hooke's law in which strain varies linearly with stresses developed in an elastic body. The materials of the parts were modeled as isotropic, with all uniform elastic properties in all directions. Finally, these volumes were assumed to be homogenous with consistent material properties throughout. The femur bone was modeled with a Young's modulus of 17,700 MPa and a Poisson's ratio of 0.3. The prosthesis socket was modeled with a Young's modulus of 1886 MPa and a Poisson's ratio of 0.39 [8]. The skin and fat were modeled with a Young's modulus of 0.5 MPa and a Poisson's ratio of 0.49. The muscle was modeled with a Young's modulus of 1 MPa and a Poisson's ratio of 0.49. These values are consistent with how bone was modeled by other researchers [9].

3.3 Material models

After determining the element type and material model, the four parts (bone, muscle, fat, and skin) were meshed with tetrahedral elements in LS-Prepost. The total number of nodes and elements for the parts is specified in Table 1.

Table 1. FEM properties

Part	Element	Material	No. of nodes	No. of Element	Element form
Skin	Solid	Elastic	6176	19310	Tetrahedral
Fat			3391	13473	Tetrahedral
Muscle			3739	18544	Tetrahedral
Bone		Rigid	1012	4206	Tetrahedral
Socket	Shell		3053	6008	Triangle
Total			17371	61500	

3.4 Contact conditions

Two contact conditions were defined in the current FE model to perform nonlinear analysis. The first contact definition was a surface-to-surface contact between the residual limb and the socket. Generally, the stiffer and more rigid surface of the contact pair is defined as the master surface, while the deformable surface with softer material is selected as the slave surface. Hence, the outer surface of the residual limb was defined as the slave surface and the socket's inner surface as the master surface. The contact definition requires that the slave surface conforms to the master surface; therefore it is recommended that a finer mesh be applied over the slave surface and a coarser mesh over the master surface. A coefficient of friction of 0.5 was assigned as an interaction property for the contact surfaces, as was justified in Lee's study [10].

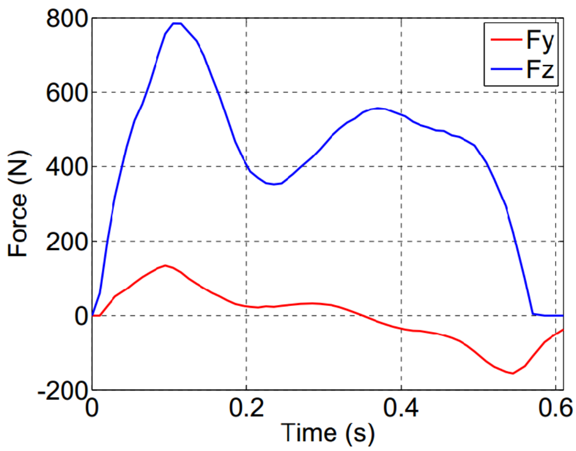

Figure 2. Load in walking phase

The second contact definition applied a tie contact between skin and fat and between fat and muscle. A tie contact is a surface-based constraint using a master-slave formulation. It provides a simple way to bond surfaces together permanently, which prevents slave nodes from separating or sliding relative to the master surface.

The connection between muscle and bone was the constrained_extra_nodes set. The inner face of muscle was constrained with bone to limit all the movement degrees between muscle and bone.

3.5 Loads and boundary condition

The analysis was carried out in two phases. The first phase was to perform standing, and the second phase was to perform walking. The load in the standing phase was half the body weight. Its magnitude was measured from experiments to be 318.5N.

The load in the walking phase is shown on Fig. 2. It was calculated with Simmechanics (MATLAB) from kinematic parameters obtained by experiments [11].

Loads were applied to top of the bone, and the socket prosthesis was constrained in all degrees of freedom in the two phases.

4. RESULT AND DISCUSSION

Standing phase

In the standing phase, the average contact pressure observed was 90 kPa. The location of maximum contact pressure is shown in Fig. 3.

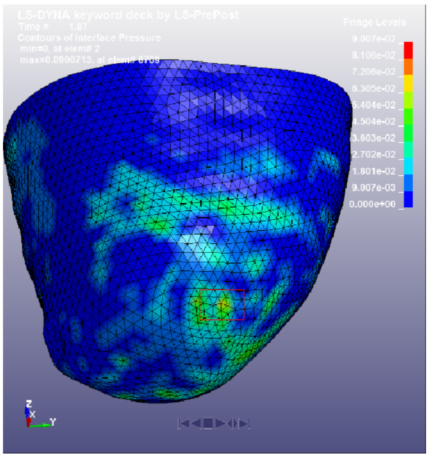

Figure 3. Contact pressure in standing phase (Red rectangular)

Walking phase

Two peaks of contact pressure were observed: 158 kPa at 0.11s and 187 kPa 0.57 s. These values correspond with the peak values of loads applied (Fig. 2). The location where the maximum contact interference region between the residual limb and the socket was observed is shown in Fig. 4 and 5.

The stress levels observed are significantly lower than the pain threshold limit of 690 kPa [12]. The results for the standing phase as well as the walking phase will be helpful to prosthetic designers in improving the prosthetic fit design by understanding the effect of pre-stress developed and the pressure distribution between the residual limb and the socket.

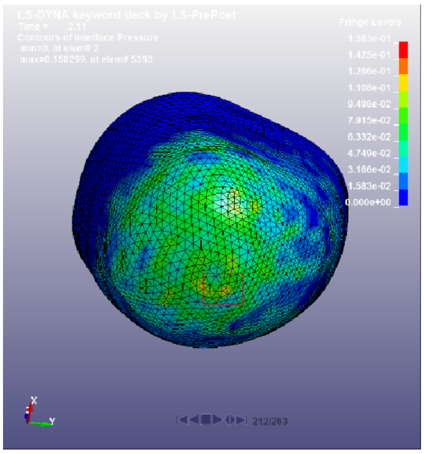

Figure 4. Contact pressure at t = 0.11 s. (Red rectangular)

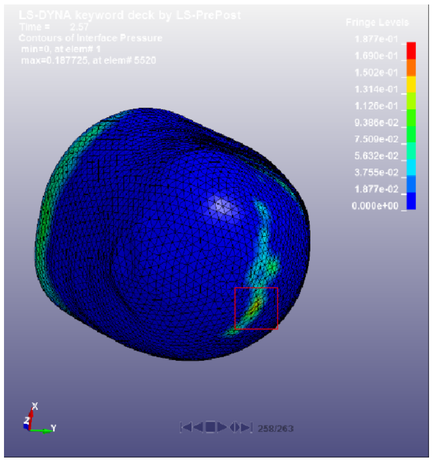

Figure 5. Contact pressure at t = 0.57 s. (Red rectangular)

CONCLUSION

The primary objective of this work was to develop a method that allows healthcare providers and engineers to simulate the fit and comfort of transfemoral prosthetics in order to reduce the number of refittings needed for amputees. Through this work, a process has been developed that can be used by others in modeling and analyzing the transfemoral prosthetic fit. The process starts with scanning of the amputee’s leg and socket, followed by developing separate CAD models for the parts of residual limb, bone, and prosthetic socket. The CAD models are then imported into FE software and assembled properly. Preprocessing operations are completed by meshing the volumes with appropriate element size and element type, assigning correct material properties, and applying contact definitions where appropriate.

In this study, the maximum normal stress observed on the residual limb’s outer surface due to application of half the body weight was 90 kPa, which is comparable to the results predicted by Zhang (65 kPa) [13] and Linlin (80.57 kPa) [14]. The results of the current analysis, along with previous research studies, indicate that finite element modeling of prosthetics must be tailored to the specific individual for whom a prosthetic device is being developed.

In developing more advanced FE models of the transfemoral prosthetic-limb interface, the hyperelastic material properties for soft tissue will be used. Experimental studies on frictional coefficients can provide insight into how to better model the contact analytically. Because of the complexity of the shape of residual limb parts, the accuracy of their 3D CAD model needs to be improved.

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