Development of a nonlinear finite element model for analysis of stump/socket interface stresses in below-knee amputee

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ABSTRACT

An nonlinear finite element model was established to predict the stresses distribution at the stump/socket interface in below-knee amputees. The limb tissues and socket liner, based on a measured amputee's geometry, were respectively meshed into 972 and 486 3D 8-node isoparametric brick elements. Bone and the socket shell were assumed to have rigid boundaries. An important feature of this model is the use of 486 interface elements (ABAQUS INTER4) which mimic the interface friction condition. The model was run on a Convex C3840 supercomputer using the ABAQUS finite element package and the FEMVIEW post-processing package. The model can be used to predict normal and shear stresses at the interface with loading; and to analyse the effect of socket rectification and tissue properties, as well as the interface slip situation with loading.

INTRODUCTION

Achieving a good stump/socket interface by assuring an optimal distribution of the interface loads is critical to the successful prosthetic fitting and rehabilitation of the lower limb amputees. A full understanding of the stress distribution is the first step of the optimization.

Besides clinical measurements, finite element analysis (FEA) has been identified as a potential theoretical method to investigate the stress distribution at this interface (Childress[1], Reynolds[2], Quesada[3]). The researches show that finite element methods, when carefully executed, can be used effectively as a design tool for prostheses[4]. An advantage of the modelling analysis is
that it is easy to investigate systematically the effects of many parameters without the need to build prostheses.

In previous FEA studies, several linear models have been established to predict interface pressures between stump tissue and prosthetic socket in below-knee amputees. They are all based on the hypothesis of linear material properties, small deformation of materials, and non-slip at the skin/socket interface.

However, limb tissues have complex characteristics. The local force-displacement curve is often highly nonlinear, the material becoming stiffer with increasing displacement[5]. The tissues may have a large deformation with loading. The another important nonlinear factor is the friction conditions at the skin/socket interface, which was indicated to play a critical role in the determination of interface pressures in Reynolds' ideal model analysis[6]. A more detailed analysis of the interface must take into account the frictional action. Neglecting the frictional aspects may restrict the application of a quantitative analysis. These nonlinear conditions at the stump/socket interface cannot be simulated accurately by the linear models.

A nonlinear finite element model for analysis of stump/socket interface stresses which considers interface friction is created in this investigation.

METHODS

The model was established for analysing the stresses at the residual limb and PTB socket interface with static loading.

The external geometry of the residual limb of a below-knee amputee was obtained from the UCL Computer-Aided Socket Design(CASD) System[2]. The radii of points were generated in a cylindrical coordinate system at axial intervals of 6.35mm, and angular intervals of 10°. The surface dimension of bone was established by digitizing biplanar X-ray views with slight modification. A Pelite liner, which has the same internal dimension as the limb and a uniform thickness of 5mm, was fitted in the residual limb.

The bone and socket wall are assumed as rigid boundaries. The hard socket and pelite liner did not wrap around the distal end of the limb. The limb tissues and liner are meshed into 3-dimension isoparametric eight-node brick elements (ABAQUS C3D8). Figure 1 (a) shows the shrunk mesh. The limb tissues contained two layers of elements (972) with nodal data axially spaced 6.35mm apart at 20° angular intervals and the outer layer of elements (486) represent the liner.

One of the important factors affecting the interface pressure distribution
is the interface friction which results in the surface shear. It is impossible that there is no friction, but the friction force cannot exceed the maximum friction which two surfaces in contact can produce. So the assumption of non-slip or non-friction cannot simulate the condition accurately. Considering the friction condition, 486 interface elements (ABAQUS INTER4), shown in Figure 1 (b), were used between skin and liner in this model. Assuming the skin and liner surfaces are in contact, there will be no slip until the shear stress exceeds the friction limit, which varies proportionally with the normal stress \( F_{\text{max}} = \mu N \), \( \mu \) the coefficient of friction, \( N \) the normal stress). The output parameters for interface elements, \( S_{11}, S_{12}, S_{13}, E_{11}, E_{12}, E_{13} \), are pressure transmitted between the surfaces, two shear stress components, separation of the surfaces in the normal direction, and two relative tangential displacements of the surface, respectively.

In our preliminary study, the properties of limb tissues and liner were still assumed as linear, isotropic and local uniform due to lack of nonlinear material property information. From the published literature, the Young’s modulus assigned in different regions of limb tissues are 145kPa in the patellar tendon, 50 kPa in the popliteal depression, 120kPa in the anterolateral tibia, 50kPa in the anteromedial tibia and 73kPa in other areas. The Poisson’s ratio was 0.45[6]. The Young’s modulus of pelite’s liner was 380kPa and the Possion’s ratio was 0.39[1]. In the further studies, nonlinear tissue properties, such as bi-phasic stress-strain property, will be used.
The analyses were performed in two steps, considering that the deformation of limb tissue from its free state into its shape when loaded in the socket can be thought of in two stages. First, a pre-stress analysis can be carried out by applying a radial deformation to the liner as a result of donning the socket due to difference in limb and socket shape (rectification). The socket rectification was implemented similar to the UCL CASD System, that is the radial modification at frequently rectified local patches. The second analysis can produce the stresses by keeping the rectification shape and applying a vertical downward displacement to the bone occurring as the limb becomes weight bearing. A net resultant force resulting from surface pressures and shear stresses has to counter-balance the downward vertical body load.

This model consists of 1458 solid elements and 486 interface elements with 2520 nodes. Analyses contained 8064 degrees of freedom with a front size of 387.

RESULTS AND DISCUSSION

An analysis was executed under the conditions: a light rectification of maximum radial increment 8mm at the patellar tendon point was applied to the socket; a downward displacement of 8mm was imposed on the bone; the coefficient of friction of skin/liner interface was 0.5.

The model was analysed using ABAQUS finite element package installed on CONVEX C3840 supercomputer in University of London Computer Center(ULCC). Cpu time was 4 minutes for pre-processing and 15 minutes for processing each increment. The results were post-processed using the FEMVIEW package. The resultant vertical force acted on the interface surface was 150N.

Figure 2 shows the displaced mesh. From the upper side of the mesh, the liner (outer layer) has a visible slip upward relative to the tissue. This means that there is not enough friction at the interface, possibly resulting from a lower pre-pressure due to small rectification.
Figure 3 gives a normal stress map at the skin/liner interface in an anterior view. A remarkable high pressure (up to 65.7kPa) acted on the patellar tendon area, and a pressure release (near 10kPa pressure) in the tibial crest area. The higher pressure at the distal end results from the gap removed due to the large interface slip. It is notified that the maximum pressure at the interface was much lower than other researches and clinical measurements because this was the pressure when only 150N force was applied to the interface.

The analysis can also give out the shear stresses in two directions and the interface slip situation. Whether slip exists at the interface depends on the shear stress and friction which relies on the coefficient of friction and normal pressure. If a small pre-pressure is applied to the interface, the remarkable slip will occur. Because of the complex interface geometry and stress distribution, different volume of slip arose in different elements from the calculating results. It may be supposed that a local slip or micro-motion may exist in some element interfaces, even though enough pre-pressure is applied and there is no gross slip or macro-motion. So non-slip assumption will overestimate the shear stresses.

As far as we know, it is very difficult to measure experimentally the relative slip, especially micromotion. In numerical studies of the interface, this is first attempt to consider in detail the effect of interface friction and slip. Further studies will explore the effect of interface stresses and slip.
Some limitations still exist in this model. The slip at the socket wall/liner interface was not considered. A very large displacement of bone may result in a poor convergence of the iteration because of the bulky distortion of elements.

CONCLUSION

This model has proved to be well-behaved in the example analysis. It can mimic the interface conditions accurately. Besides the analyses of stresses, including normal pressures and shear stresses, the model is anticipated to predict the interface slip and to analyse the relationship among slip, pre-pressure (rectification) and friction. This series of investigations is expected to increase our understanding of what factors govern the interface stresses and thus lay better foundations for socket design.

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