STUDY OF FE MODELS FOR A SMART SENSOR BASED PROSTHETIC SOCKET DESIGN SYSTEM

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Abstract— This paper reports the research work related to establishing biomechanical models for socket design. Through numerical simulation using finite element method, comprehensive relationships between pressure on residual limb and socket geometry were established. This allowed local topological adjustment for the socket so as to optimize the pressure distributions across the residual limb.

Index Terms— Above knee amputee, Finite element simulation, Hyperplastic model, Prosthetic socket.

I. INTRODUCTION

Prosthetic socket is a component that connects the residual limb of an amputee with an artificial prosthesis. It is widely recognized as the most critical component that determines the comfort of a patient when wearing the prosthesis in his/her daily activities. Through the socket, the body weight and its associated dynamic load are distributed and transmitted to the prosthesis during walking, running or climbing. The current clinical practice of making a prosthetic socket is achieved by a touch-and-feel approach which relies on the well-trained prosthetist and is a highly subjective process.

Although there have been significant advancements in prosthetic technologies such as microprocessor controlled knee and ankle joints in the last decade [1], the progress in designing a comfortable socket has been rather limited. This means that the current process of socket design is still very time-consuming, and highly dependent on the expertise of the prosthetist. Supported by the state-of-the-art sensor technologies and numerical simulations, a new socket design system is developed to help prosthetists achieve rapid design of comfortable sockets for above the knee amputees [2].

In order to optimize the design of a prosthetic socket for the above knee amputees, it is essential to understand the biomechanical properties of the soft tissue of the residual limb. In the current clinical practice, the material properties are obtained by a prosthetist using a touch-and-feel approach. Apparently this approach is subjective, and can only provide qualitative information about the residual limb. To achieve a quantitative information of the soft tissue, an indentation device is used to record the dynamic response of the residual limb. By analyzing these dynamic data, more quantitative biomechanical characteristics can be obtained.

Indentation method has been a widely used for material characterization [3,4]. In our study, a tissue indentation device was developed using a force and a displacement transducers to record the load-displacement curve during an indentation test.

The force – displacement measured using the indentation device can be used to derive key material property, i.e. local elastic modulus of the residual limb. With position specific elastic modulus of the residual limb, FE method can be carried to simulate the pressure on the residual limb, which depends on the geometrical dimensions and shape, that can provide key information for the socket design.

However, unlike the conventional engineering material human soft tissue behaves hyperelasticly. To achieve the best simulation result, in this study, different hyperelastic FE models are evaluated to identify the best model for our project.

II. METHODS AND MATERIALS

To validate the simulation results of using different hyperelastic models, we compared the simulations with experiment result on a range of silicone models.

A. Silicone Models and Its Material Property

To mimic different elastic modulus of muscle, fat and skin, three elastic modulus of the cylindrical silicone samples were made close to the values shown in table 1. They were subjected tensile tests to obtain the exact elastic modulus and were used as the elastic modulus for the FE simulation.

Table 1

Soft Tissue	Elastic Modulus (kPa)
Muscle	90
Fat	20
Skin	120

The tensile tests were carried out using an Instron 6025 mechanical testing machine to obtain the force and displacement relationships of specific silicone sample. The force and displacement relationship was used to determine the elastic modulus.

The silicone acquired for the sample was EcoFlex 10 and EcoFlex 30 from the EcoFlex® Series (Smooth-On Inc., Texas). Silicone production requires the mixture of two parts (Part A and Part B), by altering the ratios of the parts the resulting elastic modulus of the silicone produced is altered.

Three tensile testing samples were created in different ratio of the silicone parts. After being degassed, the silicone was cured in cylindrical molds all with the same dimensions. Once cured, the removed silicone tensile test samples were measured and displayed a diameter of 15mm and a testing region of 52mm which is highlighted by the marked black lines indicating the top and bottom clamping regions during tensile testing. Figure 1 shows the tensile test carried out by this study.



Figure 1 Instron 6025 setup for tensile testing

The results of the tensile tests on the cylindrical silicone samples gave three elastic modulus: 25.6KPa, 102.2KPa and 115.6KPa for this study. Figure 2 shows the force – displacement curve of the 115.6KPa cylindrical silicone sample.

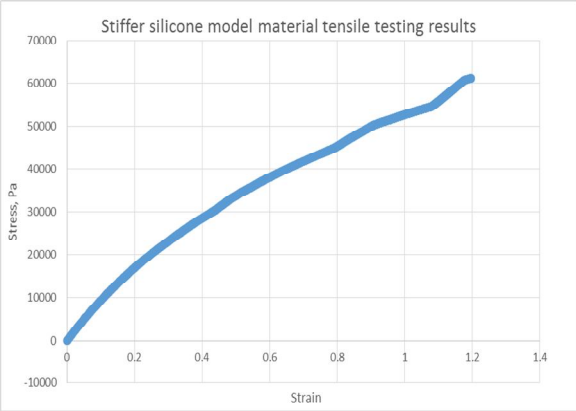


Figure 2 Tensile test result of cylindrical silicone sample

B. Indentation Tests

At the same time of the silicone sample were made, silicone bulk models were also made. These models were firstly used to validate the indentation device, where the results were compared with those obtained from an Instron mechanical testing machine. Then these models were reused for indentation tests to obtain the force-displacement curves for different silicone bulk models. Figure 3 shows the indentation test on a silicone bulk model.

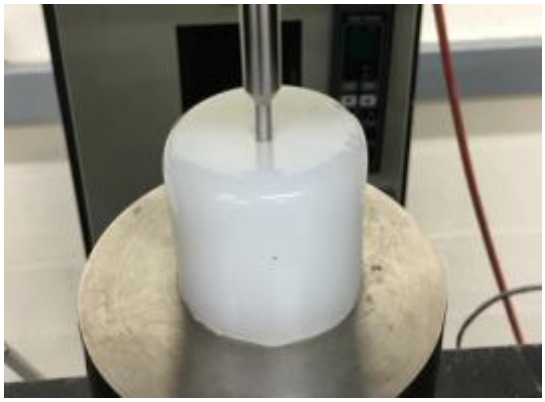


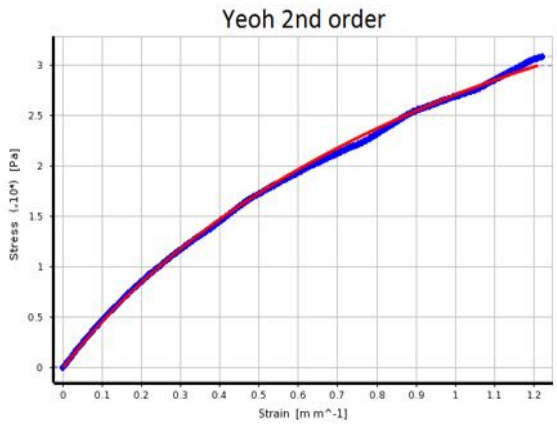
Figure 3 Indentation test on a silicone bulk model

C. FE Simulation Using Different Models

In the FE modelling, hyperelastic algorithms were applied to the tensile test material data by a curve-fit algorithm function of the FE package ANSYS. The least squares method was used for the data process. Figure 4 (a), (b). (c) and (d) show the Ogden 2nd order, Yeoh 2nd order, Mooney-Rivlin parameter 3 and 5 order hyperelastic models.



(a)



(b)



(c)



(d)

Figure 4 curve-fit of different hyperelastic model

From the Figure 4(b), (c) and (d), it can be observed that for the Mooney-Rivlin parameter 3 and 5 hyperelastic models give near-perfect representation of the input data and Yeoh 2nd order algorithm also shows very good agreement of input data to output curve-fitted profile.

However, it can be seen from the Figure 4 (a) that the Ogden model exhibits a very poor curve-fit to the stress-strain data, and the resulting stress-strain relationship does not represent the soft tissue behavior in any way. For this reason, no subsequent analysis using the Ogden model was undertaken.

The FE simulations were carried out using FE package ANSYS 18 and axisymmetric models as shown in Figure 5.

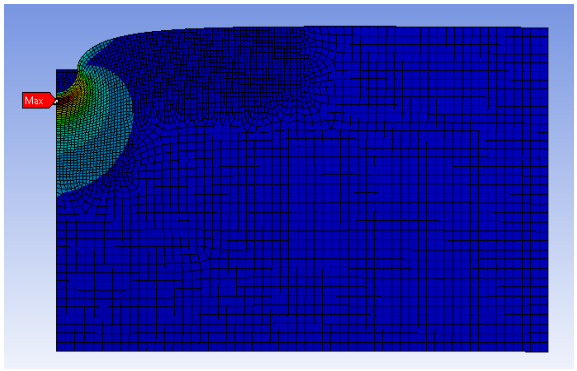


Figure 5 FE Silicone bulk models

Convergence testing was performed on the models until a satisfactory mesh was achieved. Figure 6 shows a comparison between a coarse mesh (left) and a fine mesh (right).

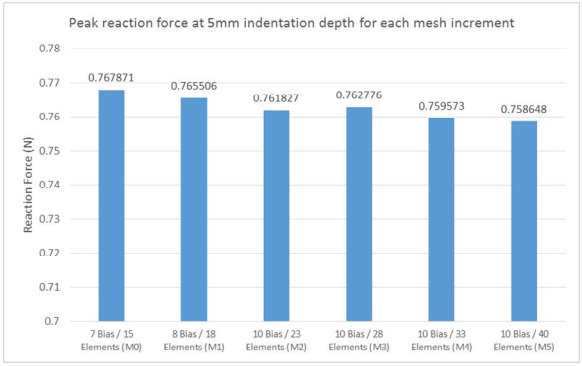


Figure 6 Peak reaction force at 5mm indentation depth for each mesh increment

III. RESULTS

In this section, the result form the FE simulation and indentation tests are compared. FE models dictate how much of the above curve-fits are utilized in FE analysis. The table 2 below gives the maximum strain experienced in indentation simulation of the soft tissue with different applied hyperelastic models.

Table 2 Maximum equivalent strain of the FE models

Sample mix	Model	Maximum equivalent strain
1/1	Yeoh 2nd order	1.347
1/1	Mooney-Rivlin 3rd order	1.103
1/1	Mooney-Rivlin 5th order	0.847

Analysis of the stresses normal to the path on the surface of the soft tissue model was performed. The stresses normal to the Y-axis path are shown in Figure 7. These results were compared for each of the models used for analysis, and are plotted below on the same axes.

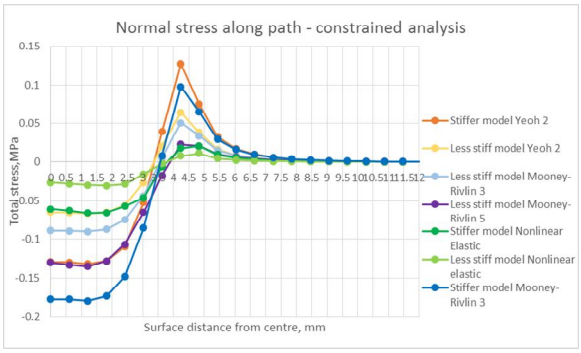


Figure 7 Normal stress along the path

The shear stress along the path was also investigated. Figure 8 below shows the shear stress on the surface of the different FE soft tissue models on the same axes.

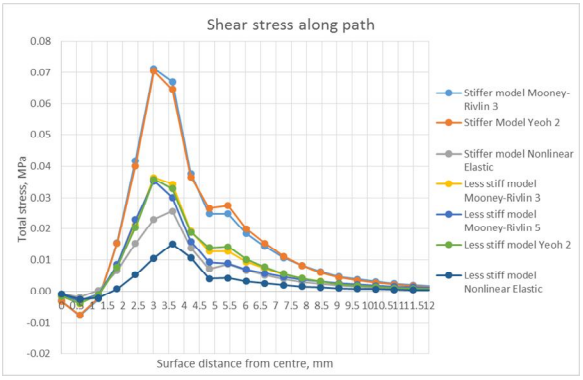
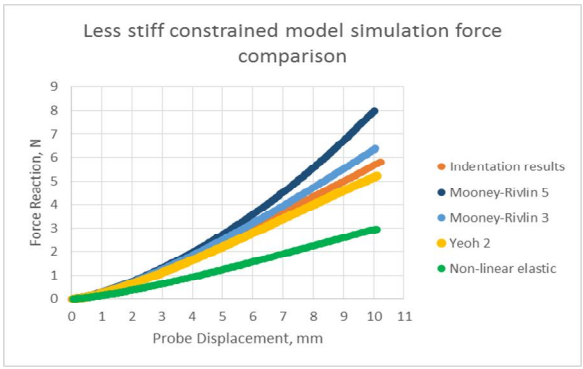
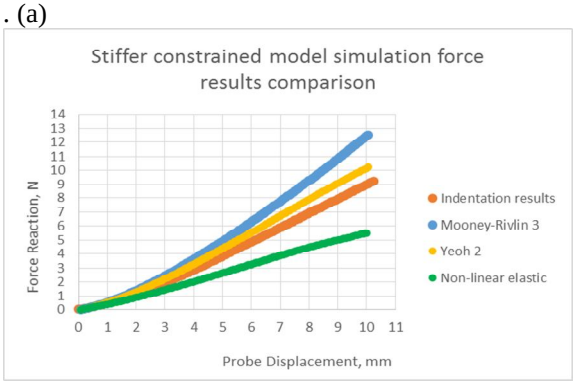


Figure 8 The shear stress along the path

The Yeoh and Mooney-Rivlin 3rd order hyperelastic models exhibit the highest shear stresses, with the peak shear stress experienced at 3 mm, corresponding to regions underneath the indenter probe. The nonlinear elastic models (using as a reference) give the lowest values of shear stress among all the models utilized, with their peak values of shear stress at 3.5 mm, just at the edge of the indenter probe. All of the models, however, show an increase in shear stress from 5 mm along the path to 5.5 mm, before finally reducing to zero. The different models were applied to the indentation simulation model and the load-displacement data was measured. These results were plot on the same axes, and are shown below in the figures 9 and 10. In these graphs, the experimental indentation results are included (orange line) for comparison. But the Mooney-Rivlin 5th order algorithm could not be run with the stiffer simulation model, therefore there is no dark blue line in Figure 9 (b)) and 10 (b).

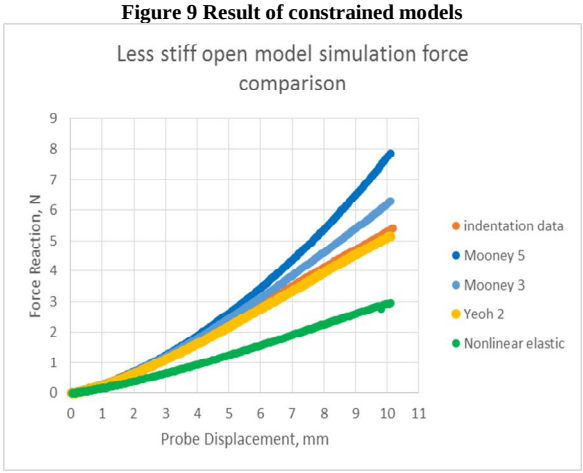


. (a)

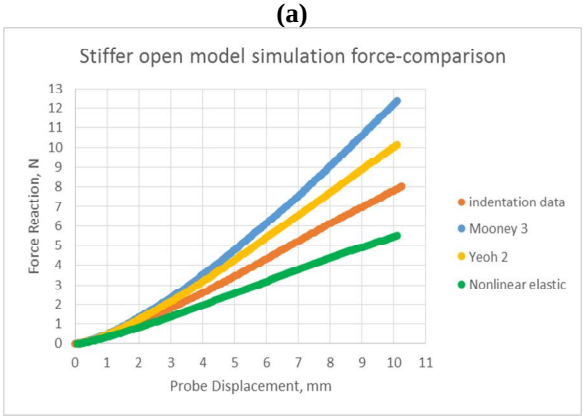


(b)

Figure 9 Result of constrained models



(a)



(b)

Figure 10 Results of Unconstrained models

IV. DISCUSSION

Based on the experiments we first obtained material

properties of the silicone models. The tensile testing provided the elastic modulus used to create the silicone bulk model for both indentation test and FE simulation. Before the FE simulation is carried out, the curve-fit graphs in figure 4 show that the Ogden hyperelastic model is poorly fitted with tensile test result of the force-displacement. We found this is contrary to the study of Hassan et al. [5]. In his study, there is a good curve-fit for the Ogden hyperelastic model to tensile data achieved using the tests of papillary muscle of a Guinea pig heart, when the ex vivo with different loading conditions were investigated. It is understandable that different the hyperelastic model may not be the same to be used to represent different soft tissue.

The comparison between the indentation test and different hyperelastic models show that there isn't any hyperelastic model can be used to produce an exactly the same result of the indentation test.

As shown in Figure 9 (a) and (b), for the constrained FE models, the result of Mooney-Rivlin models are higher than the indentation test results of the silicone models. For the Yeoh 2nd order models, the results are higher than the indentation test when the elastic modulus is relatively higher. However, when the elastic modulus of the silicone model is relatively low, the simulation results of the Yeoh 2nd order model is slightly lower than the indentation tests, but it is very close to the indentation tests results.

As shown in Figure 10 (a) and (b), for the unconstrained FE models, the result of Mooney-Rivlin models are always higher than the indentation test results of the silicone models no matter the silicone models have the higher or lower elastic modulus. For the Yeoh 2nd order model, the results are very close to the indentation tests when the elastic modulus is relatively lower. However, when the elastic modulus of the material is relatively high, the simulation results for the Yeoh 2nd order model become higher than the indentation tests.

The normal stress and shear stress from the FE simulations produced certain difference between three different hyperelastic FE models. This indicated that there were inaccuracies produced when all hyperelastic models provided by the Ansys were used. In general, the higher the strain involved, the higher the inaccuracy could be produced.

We believe that the inaccuracy may be produced with the two reasons. One reason could be the hyperelastic model itself and the other one could be introduced during the indentation tests. For example, in the real test, there was a friction taken place between the head of inventor and the surface of the silicone. However in the simulation, although the friction force was applied, it would not be exactly the same as the real tests. This shows a limitation of using the hyperelastic model of FE package to simulate the indentation tests.

CONCLUSION

Comparisons of results obtained from FEA simulations and experimental tests showed that the Ogden model did not fit well with the silicone indentation data, while the Yeoh model gave the best representation of a lower elastic modulus silicone models. Compared with hyperelastic model, the result showed that nonlinier elastic model also had significant errors (green line in Figure 9 and 10).

In addition, normal and shear stress distributions on the surface of the silicon models were obtained. The effect of friction in compression testing, and the influence of material stiffness and testing boundary conditions were also assessed. These are believed the main reason that led the difference between the indentation test and simulation results. However, when the strain is relatively low the simulation can produce a reasonable accuracy and be used for our SocketMaster project to get a good-fit socket for the above knee amputees.

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