INTEGRATION OF COMPUTER AIDED DESIGN AND FINITE ELEMENT ANALYSIS FOR LOWER-LIMB PROSTHETIC SOCKETS

K.S.A. Yew[1], J.C.H. Goh[1, 2], P.V.S. Lee[1, 3] and S.L. Toh[1, 4]

[1]Division of Bioengineering, National University of Singapore, Singapore, Singapore
[2]Department of Orthopaedic Surgery, National University of Singapore, Singapore, Singapore
[3]Defense Medical and Environmental Research Institute, DSO, Singapore, Singapore
[4]Department of Mechanical Engineering, National University of Singapore, Singapore

bieyksa@nus.edu.sg

Abstract: Prosthetic socket used by amputees today are mostly fabricated by the traditional artisan method. While these hand made sockets are effective in aiding the amputee in walking, the fabrication process is lengthy, labour intensive and often offers no repeatability. The advancement of CAD technology has made it possible to help the prosthetist reduce the socket delivery time to amputee. FE model was automatically generated via an in-house developed program. Material properties, joint loads during heel strike and relevant boundary conditions were prescribed for the simulation. The model was validated by comparing predicted contact pressure with experimentally measured contact pressure obtained from capacitive pressure sensors. The results demonstrated agreement and showed potential to be developed further as a clinical application to help the prosthetist in fabricating socket.

Introduction

Currently, majority of the prosthetic sockets used in artificial leg assemblies are fabricated via the traditional artisan method. While sockets manufactured this way are comfortable and relatively suitable for ambulation, this process is time consuming and labour demanding. In addition, it also offers little or no repeatability should a replacement socket be required. However, Rapid Prototyping (RP) and Computer Aided Design (CAD) technology advancement over the last decade have allowed prosthetic sockets to be made with minimum time and effort [1]. As such, it is possible to integrate CAD with Finite Element Analysis (FEA) as a socket design optimisation analytical tool and using optimised socket CAD geometry for fabrication via RP technique. Currently, little work has been reported in using such CAD-FEA integration technique. FEA has been suggested as a possible tool for prosthetic socket design [2]. Developing FE model is tedious, and so far no software is readily available to generate FE codes directly from geometric data for socket design optimisation. Although method of hexahedral meshing for FE analysis of residual limb has been reported [3], merging the geometric data sets from different modalities (mechanical digitisation, optical scanner and Computed Tomography {CT}) is still necessary. In this study, automatic generation and analysis of FE model with relevant boundary conditions from geometrical data acquired from a prosthetic CAD system and its preliminary validation of the integrated CAD-FEA process on an amputee subject were investigated.

Materials and Methods

One left unilateral male trans-tibia amputees and a prosthetist participated in this study. The volunteer subject is 168 cm tall, 76.8 kg in mass. The cause of his amputation was due to vascular disease. An active amputee of 11 years, the subject uses his currently prescribed artificial leg for normal walking activities. Rectified and unrectifed castings by prosthetist of subjects' stump were used for surface geometry digitisation using CAPOD's laser scanner and software (Össur Systems, Sweden). A pre-scanned fused tibia and fibula bone was scaled and positioned into the hollow stump using relevant bone markers and anthropometric scaling method. Once the position of the bone and size was confirmed, the bone was removed graphically, leaving a void in the model. The internal surface (of the void) was then offset inwards to form the bone [4]. The CAD geometries of the bone, rectified socket and the stump were then exported as Initial Graphics Exchange Specification (IGES) files for automatic FE model generation [4].

An in-house developed C program was used to generate the necessary FE codes suitable for simulation in ANSYS solver environment [4]. The final FE model consisted of a total of 11200 8-noded hexahedral structural elements (stump) and 4-noded shell elements (socks, socket) and 10400 nodes. Pre-stress due to rectified socket and contact were also taken into

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consideration and automatically modelled. Young's modulus and coefficient of friction for cotton socks are 0.03MPa and 0.1 respectively, while the material properties of bone, soft tissue and polypropylene socket can be referred to Table 1[4, 5].

Table 1: Material Properties of FE model

Material	**Young's Modulus (MPa)**	**Poisson's Ratio**
Bone	Rigid	-
Soft Tissue	0.1	0.49
Socket	1000	0.33

Description of kinetic data of the lower limbs collection can be referred to [4]. Triaxial forces and moments at heel strike of the gait cycle were applied on the distal end of the socket while the nodes of the bone were rigidly held. A linear FE analysis was performed on a computer workstation with a 3.2 Giga hertz processor, and 2 Giga bytes RAM to predict stump/socket contact pressures.

Experimental trials were also performed to compare the contact pressure prediction by the FE model. The scanned rectified casting was sent to a customised system, Rapid Manufacturing Machine (RMM) to fabricate the prosthetic socket. The detailed description of the fabrication process can be referred in [1]. The time taken to fabricate the prosthetic socket was approximately 3.5 hrs. The RMM prosthetic socket was then assembled into an artificial leg with the relevant components. After appropriate alignment by the prosthetist, the artificial leg was worn by the subject for the experiments.

Nine patches of 4×4 Novel Pliance capacitive pressure sensor arrays (Novel GmbH, Germany) as shown in Figure 1 were taped on patella tendon (PT), tibial tubercle (TT), fibula head (FH), distal anterior tibia (DAT), distal medial tibia (DMT), proximal posterior (PP), distal posterior (DP), media tibia flare (MTF), distal lateral fibula (DLF) on the amputee stump before wearing cotton and silicon socks (Figures 2 and 3). A laptop installed with Novel proprietary software was used to acquire the pressure measurement data via Bluetooth transmission from the data logger unit worn by the subject.

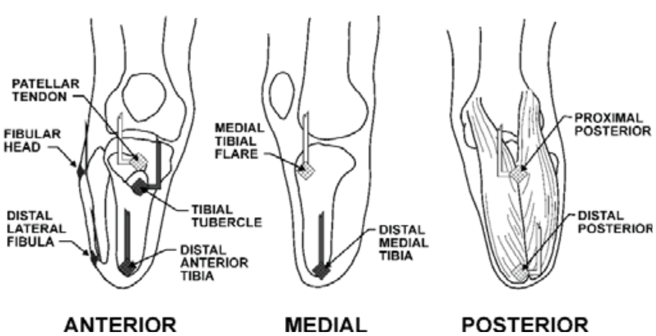

Figure 1: Test locations for contact pressure measurements [6]

Figure 2: Novel capacitive sensors attached to various location of subject's stump

Figure 3: Subject donning artificial leg assembled with RMM prosthetic socket over capacitive sensors

Prior to the experiments, calibration of the pressure sensors was carried out using a customised bladder system (Figure 4). Sandwiched between a flat acrylic plate and a silicon bladder, the sensor pads were subjected to hydraulic pressures between 0 kPa and 240 kPa at increments of 30 kPa (Figure 4). Depending on individual sensor pads, the accuracy of the sensors was estimated to be between -0.1 kPa and -1.4 kPa.

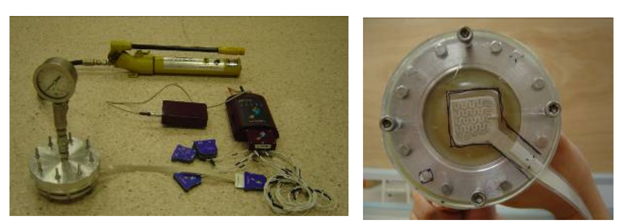

Figure 4: Calibration equipment for capacitive pressure sensors (left); a pressure sensor pad when subjected to hydraulic pressure during loading (right)

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Results

Predicted results from FEA indicated the maximum contact pressure occurred at patella region. Figure 5 shows the predicted contact pressure contour plot. The predicted pressures at the selected locations at heel strike during a gait cycle ranged from 0 kPa to 176.1 kPa (Table 2).

Maximum measured contact pressure was also observed at the patella region. The measured pressures at the selected locations at heel strike during a gait cycle ranged from 0 kPa to 161.8 kPa (Table 2).

Overall, the predicted contact pressures are lower than that of the measured contact pressures apart from PT and DLF locations. Comparing the FEA predicted and experimental contact pressures indicated a percentage difference between 0% and 38% with the exception of 75% difference at the FH test location (Table 2).

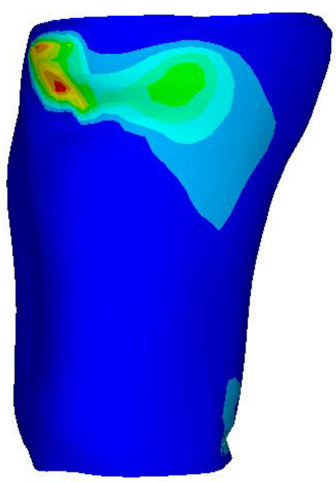

Figure 5: Predicted contact pressure distribution at heel strike

Table 2: Comparison of FEA predicted and experimental measured contact pressures at heel strike

Location	FEA (kPa)	Experimental (kPa)	% difference
PT	176.1	161.8	8.8
DAT	14.5	15.0	3.3
DMT	13.6	22.0	38.2
DP	42.7	43.9	2.7
PP	17.1	27.1	36.9
DLF	61.7	49.0	25.9
FH	21.3	88.7	76.0
TT	0.0	0.0	0.0
MTF	18.8	20.3	7.4

Discussion

The use of capacitive pressure sensors did not necessitate the fabrication of special socket with numerous holes to accommodate pressure transducers as compared with [4]. As such, the actual socket used by the amputee in his normal activities can be readily used. The results in the previous section show relative good agreement of contact pressures between FEA and experimental measurements. From the contour plot of the FEA, the model similarly predicted maximum contact pressure at the patella region under comparable conditions at heel strike. While the results shows a higher percentage difference at DMT, PP and DLF locations, their absolute difference are only approximately between 8 kPa and 12 kPa which are small when compared with maximum contact pressure at the PT region. As for the considerable difference observed at FH location, one possibility could be due to the use of fused bone geometry instead of modelling separate tibia and fibula bone structures. Also, in view of the existence of differences between FE predicted and measured contact pressures, other likely reason for such differences is the non-consideration of non-linear and viscoelastic material properties of soft tissues in the FE models.

A FE model's predictive ability depends on accurate materials prescription, relevant boundary conditions, and correct loading characteristics and to some extent, its mesh density (larger number of elements can lead to better accurate solution), and for FEA in biomechanical studies, the relevant anatomically accurate bone geometries. However, incorporating all these significantly increase computational time and as a result, prolong socket delivery time to amputees. Thus, integration of CAD-FEA to socket design requires optimisation of computational time without sacrificing prediction accuracy. However, with the advancement in computing, it is likely that non-linear, viscoelastic material properties and larger optimised mesh density might also be handled without significant increase to current computational time and should be considered in future work when such possibilities exist.

The employment of FEA in this study has positive implications for both prosthetist and amputee. The quantitative feedback from such application can help the prosthetist improve the chance of successful first fitting, in turn eliminate the need for multiple test sockets, and concurrently reduce socket delivery time. Furthermore, the use of anthropometric scaling method to obtain bone geometry provides a viable alternative to expensive Magnetic Resonance Imagining (MRI) or harmful radiation exposure from CT scans. Whilst the FEA model developed shows potential in aiding a prosthetist, more work needs to be done before this application can be adopted in a clinical environment. Some examples are

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translating contact pressure to clinically relevant pain or discomfort of the amputee [4].

Conclusions

A finite element model utilising an amputee's stump geometry was developed using a CAD-FEA program [4]. The model was validated through the comparison of predicted contact pressure with measured contact pressure obtained from experiments using capacitive pressure sensors. Relative good agreement was observed between FEA predictions and measured contact pressure. The FE model developed demonstrates potential clinical application to help the prosthetist in socket fabrication and fitting.

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