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Analysis of A below Knee Prosthetic Socket

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Abstract

This paper presents the analysis of a below knee prosthetic socket for successful ambutation, comfort and stability. Seven different design models are created by changing the socket wall thickness, enlarging pressure relief areas of the socket and changing the material.

The model designs are implemented by finite element program, ANSYS. Useful information on stresses and deflections are obtained and a final design of socket is selected.

الخلاصـــــة

تم في هذا البحث تحليل سلوك الوقب الصناعي تحت الركبة للتأمين راحة واستقرار المريض. درست سبعة نماذج تصميم وذلك بتغيير سمك جدار الوقب و تغيير المادة في المناطق الحساسة للضغط.

استخدم برنامج ANSYS لحساب الاجهادات والتشوهات لكل تصميم ومن خلال النتائج تم اختيار التصميم المناسب.

1. Introduction

A prosthetic socket is the interface between the human and the mechanical support system. Design and fit are what ultimately determine the energy expenditure, comfort, support and patient acceptance of the socket. For optimal prosthetic performances, the socket must facilitate motion. Forces, generated by the residual limb through gait motion, must be efficiently transmitted for the limb to the prosthesis; thus, any relative motion exhibited between the residual limb and the socket will challenge successful ambulation, thereby increasing fatigue and discomfort.

Patients expect to receive prosthesis with proper fit, ultimate comfort and functionality. Additionally, the prosthesis must be lightweight and cosmetically appealing. Despite the best efforts of the prosthesis, the prosthesis is rejected due to inability of the patient's residual limb to tolerate normal and appropriate socket pressures associated with traditional below the knee (BK) prosthetic design. Therefore, tissue tolerance remains an issue of concern that most of the below-knee amputees undergo dissatisfaction of socket discomfort due to tissue breakdown of the skin resulting in residual limb pain. The most prominent areas of sensitivity include the fibular head and the distal end of the tibia.

Several innovations have been introduced to relieve tissue tolerance, these include double wall socket consisting of a rigid outer shell and a semi-flexible inner shell. In Iraq, the double well socket is used [1].

The behaviour of the prosthetic socket was studied by many investigators. Ross [2], developed finite element model for above-knee prosthetic socket using PAL2 program. Daniel [3], analyzed the characteristics of different types of prosthetic feet, he explained the role of materials used in construction of the prosthetic feet. Tara et. al. [4], performed a research concerning design of a mechanical cycle fatigue to assess prosthetic feet as a predictor of field service life. Michael et. al. [5], designed dynamic response prosthesis for natural and fast walking using a visual basic program.

2. Design Specifications

The design of the prostheses is based on the following design specifications:

1. Materials: The main different materials used in the analysis are:

- **a) Polypropylene:** Polypropylene is quite similar in properties to polyethylene. It is however, the lightest thermoplastic known and has a specific gravity of 0.9. It has a high melting point of 150ºC. The tensile strength, rigidity, and crack resistance are greater than polyethylene. Polypropylene has good abrasion resistance, excellent electrical properties, and high creep resistance. Being the lightest thermoplastic at temperature 0ºC, it is brittle compared with high density polyethylene and is more vulnerable to oxidation [6].
- **b) Duraform PA6 (Nylon):** Duraform PA6 delivers the impact strength and durability required for practice. Tensile and flexural strength combine to make tough prototypes

with the flex associated with many production thermoplastics. Duraform PA6 meets the challenge through its broad range of chemical and thermal resistance with a deflection temperature of (180ºC) and resistance to alkaline, hydrocarbons, fuels and solvent. The young modulus, yield strength and poisson ratio of duraform PA6 are 1600, 44 MPa and 0.31 respectively [7].

c) 10% Glass Reinforced Polypropylene: Tensile strength of 10% glass reinforced polypropylene at 135ºC is similar to that of unfilled polypropylene at 20ºC [8].

2. Physical and Mechanical Constraints:

1. Thickness range of socket wall is (1-5mm).
2. Designs based on the weight of a (75kg) human male.
3. Static loading limited to normal physical activity (i.e. standing, walking and sitting).
4. The maximum loading is at heel strike and calculations of the reaction forces at the socket base during heel strike are of the same values as in the socket stress distribution mentioned previously.
5. Maximum wall displacement is 2mm [1].

3. Force Analysis and Finite Element Modeling

One of the objectives in this analysis is to provide flexibility in specific locations of the socket wall to eliminate tissue breakdown and provide pressure relief region areas. The surface model is divided into desired areas representing the pressure relief areas. The element used to mesh the socket was SHELL 63. The socket is analyzed at the most extreme load conditions, which exist at heel strike. The pressure load on the socket base is evaluated as follows:

Gait data is readily available, usually in the form of percentage of body weight and phases of gait. The horizontal and vertical loads are based upon a 75.5 kg subject [2]. The normal force (Fy) and horizontal force (Fx) are evaluated as 130% and 18% of the body weight, respectively [2]. The force calculations are as follows:

$$\mathbf{Fy = (1.3)\,(75.5)\,(9.81) = 926.851\ N} \quad \ldots \mathbf{(1)}$$

$$\mathbf{Fx = -(0.2)\,(75.5)\,(9.81) = -148.131\ N} \quad \ldots \mathbf{(2)}$$

The maxim angle at heel strike is 30º [1], assuming that the length of the shank is 250 mm. **Figure (1)** shows the geometry of the foot-shank and the location of GRFs.

The force components are transferred to the socket base as a vertical load and a moment about the center of the socket base. The horizontal force affects the bolt mainly which fastens the socket to the shank.

The moment at the end of the shank is:

$$\circlearrowright + \Sigma M = Fx\,(L'\cos 30^{o}) + Fy\,(L'\sin 30^{o}) \quad \ldots \mathbf{(3)}$$

This moment is applied to the socket base as a distributed pressure along the base area. From Equation (3), the (ΣM=-83785.072 N.mm).

At the end of the shank, the resultant force which is parallel to the shank produced by Fx and Fy is:

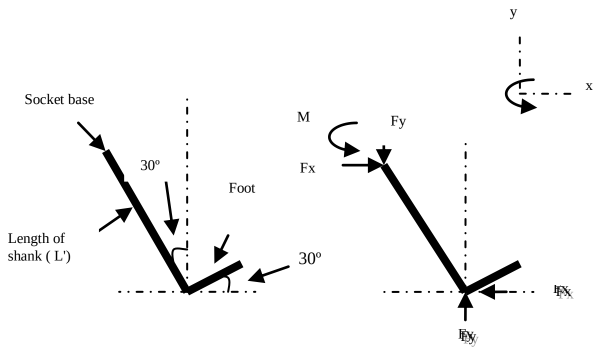
$Fy' = Fy\cos 30 + Fx \sin 60$

$= (962.851) (\cos 30^{o}) + (148.131) (\sin 30^{o})$

$= 907.918$ N

$PFy' = (Fy') (4) / \pi(Do2-Di2)$

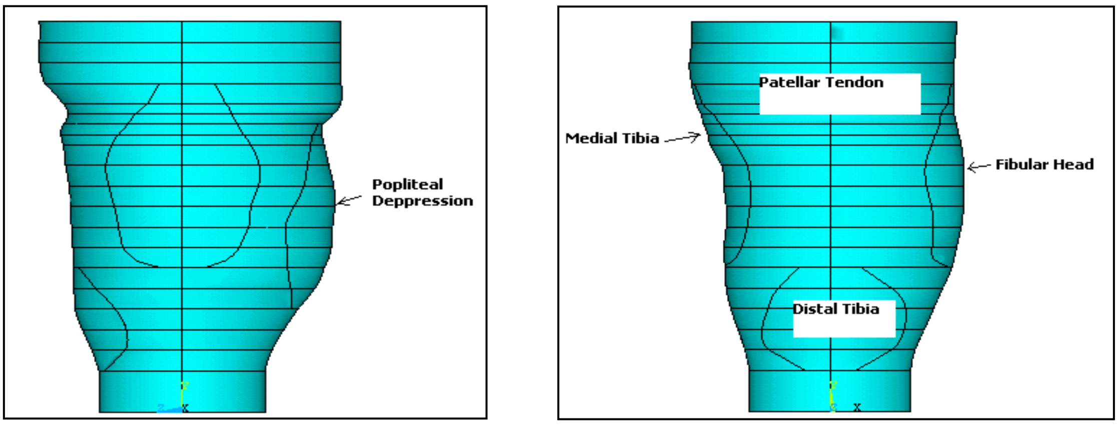
$= (907.918) (4)/\pi(702-302) = 0.288$ N/mm2



(a)Foot-shank system **GRFs & reaction forces**

Figure (1) The foot-shank system at heel-strike

In addition to the loads applied to the socket base, various interfacial pressures from previous experimental analyses are applied to the inner surface of the socket wall. These pressures are associated with the areas of the below knee socket: patellar tendon (PT, 8.4 N/cm^2), medial tibia (MT, 4.0 N/cm^2), popliteal depression (PD, 7.4 N/cm^2), distal tibia (DT, 9.9 N/cm^2) and the fibular head (FH, 4.8N/cm^2) [1]. **Figure (2)** shows the pressure relief regions of the socket.



(a) Lateral view of the socket **(b) Anterior view of the socket**

Figure (2) The pressure relief regions of the socket by ANSYS

The residual limb consists of several types of tissues (i.e. muscle, adipose tissue, skin and bone). Each tissue is comprised of its own material properties including different modulus of elasticity. The elasticity of tissues is modeled in ANSYS by assigning an elastic foundation stiffness value to the defined pressure relief areas of the socket in contact with the residual limb. Elastic foundation stiffness (EFS) is defined as the pressure required producing a unit normal deflection of the foundation. The EFS values are determined for the following areas of the socket: fibular head, medial tibia, distal tibia and popliteal depression.

4. Results and Discussion

Table (1) lists the EFS values for each area and associates soft tissue elastic modulus. In ANSYS, the EFS value can be input as a real constant for SHELL 63.

Table (1) EFS & E Values for Pressure Relief Regions [1]

EFS & E	Fibular Head	Medial Tibia	Tibial End	Popliteal depression
E(N/cm^2)	40.5	24.7	249	30
EFS(N/cm^2)	11.571	7.057	24.900	2000

The socket is constrained at the inner ring of the socket base. It was constrained in all degrees of freedom.

4-1 Alternate Designs

Different design models are created by changing the socket wall thickness and the material for the socket fabrication. **Table (2)** shows the selected model for design specifications.

Table (2) The selected models for design

Analysis No.	Socket Thickness	Material
1	5mm for all model	PP for all model
2	1mm for pressure relief areas while 5mm for the rest of model	PP for all model
3	5mm for all model	PA for all model
4	1mm for pressure relief areas while 5mm for the rest of model	PA for all model
5	5mm for all model	PP for pressure relief areas while PA for the rest of the model
6	1mm for pressure relief areas while 5mm for the rest of model	PP for pressure relief areas while PA for the rest of the model
7	5mm for all model	PP with modulus of elasticity of 330MPa for pressure relief areas while reinforced PP with E=3500MPa for the rest of the model

The main factor for designing the prosthetic socket is the deflection of the socket and the yield stress. From the socket stress distribution, it is found that heel strike phase is the critical phase and the stresses produced in the socket are highly lower than yield stresses for polypropylene and laminated sockets. Therefore, the deflection is the controller element of the design, especially for thermoplastic materials.

Figures (3) to (9) show the Von Mises stresses and deflection plots at critical phase and heel strike for seven analyses. **Table (3)** shows the maximum deflections and stresses for these analyses.

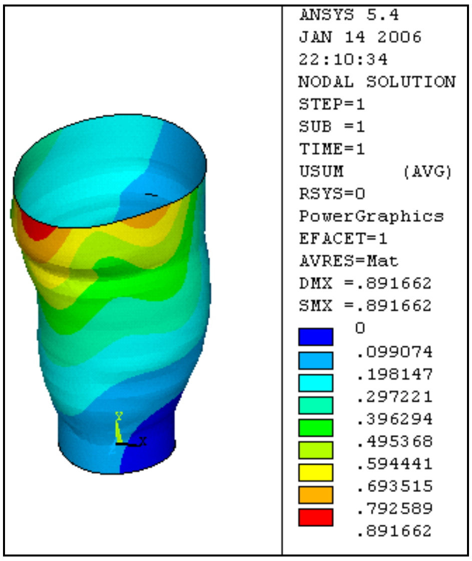


Figure (3) Deflection shape of analysis 1

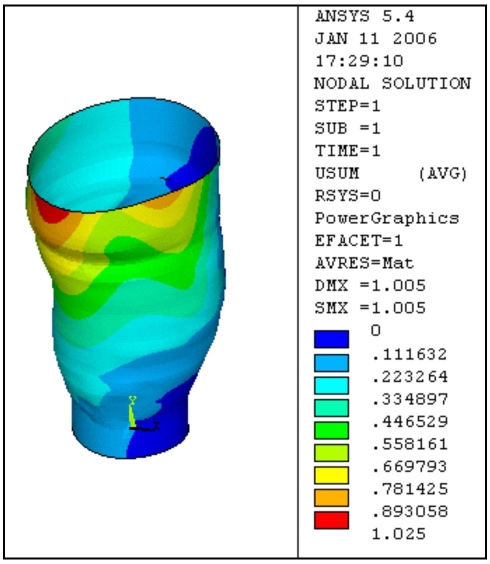


Figure (4) Deflection shape of analysis 2

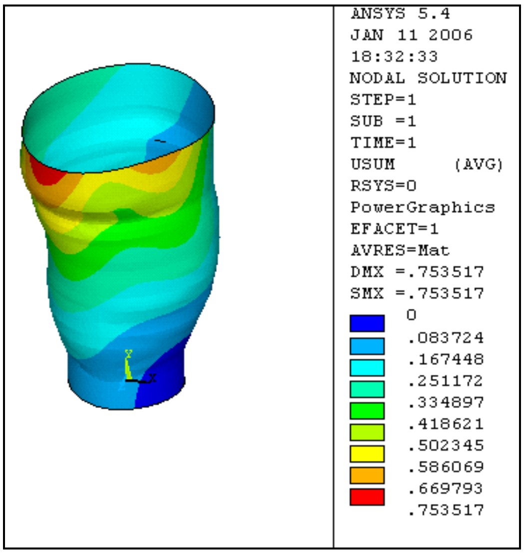


Figure (5) Deflection shape of analysis 3

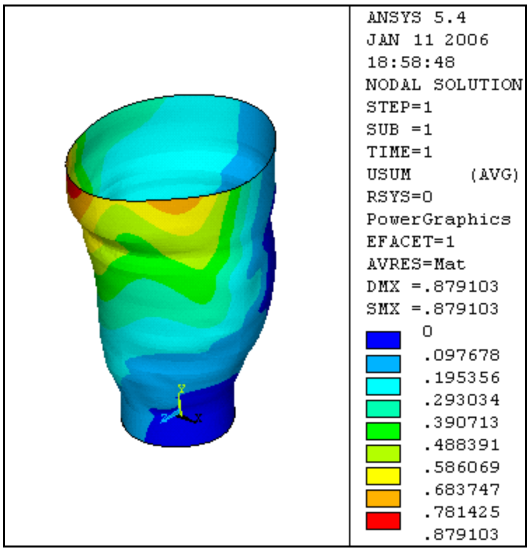


Figure (6) Deflection shape of analysis 4

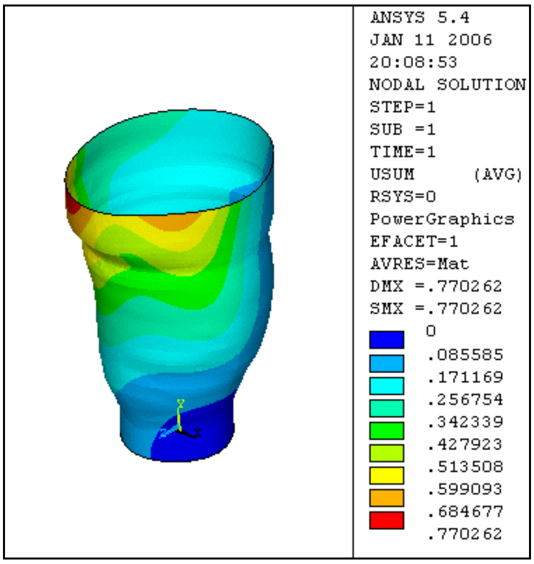


Figure (7) Deflection shape of analysis 5

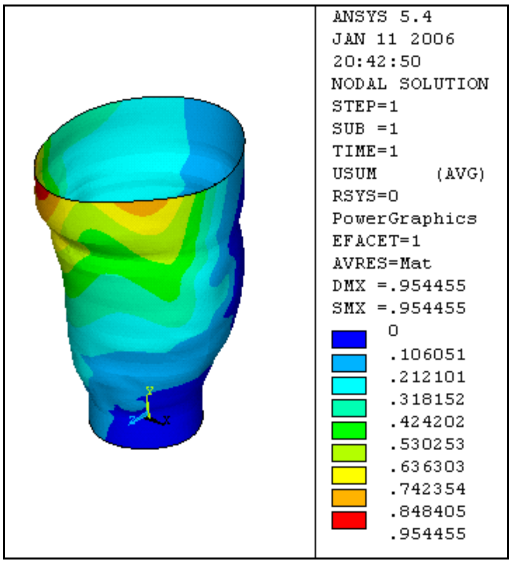


Figure (8) Deflection shape of analysis 6

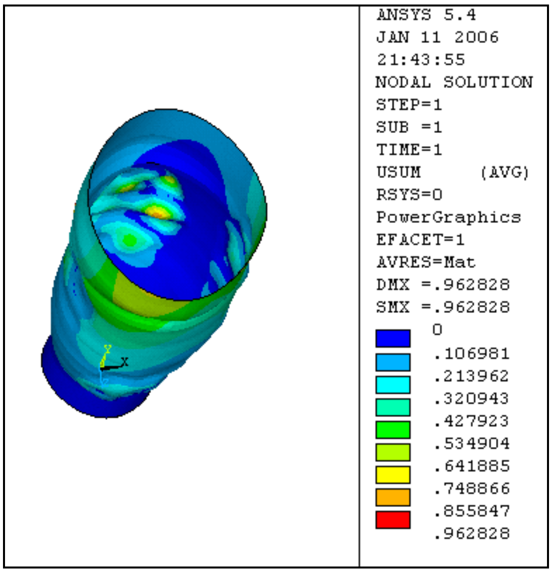


Figure (9) Deflection shape of analysis 7

Table (3) The maximum deflections and stresses for design analyses

Analysis No.	Description	Maximum Deflection (mm)	Maximum Stresses (MPa)
1	PP(5mm)	0.891662	7.351
2	PP(5-1mm)	1.025	7.511
3	PA(5mm)	0.753517	7.626
4	PA(5-1mm)	0.879103	7.820
5	PP & PA (5mm)	0.770262	7.408
6	PP & PA(5-1mm)	0.95455	7.684
7	Reinforced PP & PP	0.962828	8.128

The polypropylene socket gives results less better than Duraform PA through the first four analyses with regard to the highest deflections (socket deflection should not exceed 2mm [1]).

To get a better socket design that combines between the flexibility at the pressure relief regions and the good stiffness of the rest socket frame, No.5 and No.6 analyses are performed. The deflections in these analyses are less high than the No.3 and No.4 analyses.

By manufacturing the pressure relief regions from polypropylene (E=330 MPa), and the rest of the socket with reinforced polypropylene with 10% fiberglass, the result gives the highest deflection with regard to the No.3, No.4, No.5 and No.6 analyses. The maximum produced stresses in all the analyses are under the yield stresses.

From the model designs, it can be concluded that polypropylene gives the best results. But, considering the different conditions accompanied the actual analysis such that the repeated loading during walking and the changing environment which affects the thermoplastic materials, No.7 model is preferred.

It can be concluded that the PTB region has the maximum deflection; therefore, it is very important to reinforce this region.

5. Conclusions

1. For the design procedure of below the knee prosthetic socket, a reduction in wall thickness increases the deflection in the pressure relief regions to approximately 1mm or over 1mm. Changing the material gives approximate results.
2. The final design of the prosthetic socket is based on manufacturing the socket frame from reinforced polypropylene with flexible material such as polyethylene to give the best results. It is suggested to make slots at the pressure relief areas of the socket and use a soft inner socket.
3. The patellar tendon region should be reinforced because the maximum deformation occurs at this region.

6. References

1. Diane Higgs, et. al., ***"Design Specifications of Lower Limb Prosthetic Socket"***, Proceedings of the 2002 ASEE Gulf, Southwest Annual Conference.

2. Ross Stewart, ***"Stresses Distribution of AK Prosthetic Socket"***, Monash Rehabilitation Technology Research Unit; Research; 1991.

3. Daniel, ***"Prosthetic Foot Design"***, Monash Rehabilitation Technology Research Unit; 2001.

4. Tara, et. al., ***"The NPO Fatigue Tester: The Design and Development of A New Device for Testing Prsthetic Feet"***, Human Mobility Research Center (AMRC); 2001.

5. Michael et. al., ***"Dynamic Response of Lower Limb Prostheses"***, University of limeric; 2001.

6. E. A. A., Crane, and J. A., Charles, ***"Selection and Use of Engineering Materials"***, Butterworths; 1987.

7. Francis, J. Trost, ***"Duraform Functionality for Diverse Applications"***, Accelerated Technologies, Inc.; 2002.

8. D. P., Hanley, ***"Introduction to the Selection of Engineering Materials"***, Van Nostrand Reinhold Company Ltd.; 1980.

Symbols

M: **moment at the end of the shank**

F_x: **orizontal force of GRF_S**

F_y: **vertical force of GRF_S**

L : **length of the shank**

D_O: outer diameter of the base

D_i: inner diameter of the base