Finite Element Analysis and Modelling of Perlon - Fiberglass of a Prosthetic socket

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Abstract
Finite element analysis can be a useful tool in investigating the mechanical interaction between the residual limb and its prosthetic socket, and in computer-aided design and computer-aided manufacturing of prosthetic sockets. The intention of this paper was to analyze prosthetic socket of distinct materials and for different geometry for optimum design solution by finite element analysis. The finite element method (FEM) is a very powerful tool for analysing the behaviour of structures, especially when the geometry and mechanics are too complex to be modelled with analytical methods. A modified three dimensional finite element model of socket was developed in workbench of ANSYS 14.0 to find out the stress distribution and deformation pattern under functionally appropriate loading condition during normal gait cycle. A variety of materials were used for the analysis of the socket like The optimization technique results showed that the best optimal design of the prosthetic above knee socket is in lay-up 4perlon 2fiber glass 4perlon design under temperature 20℃. This study focuses on the analysis of patellar tendon bearing prosthetic sockets with integrated compliant features designed to relieve contact pressure between the residual limb and socket. In this study, the numerical results of stress distribution model of the prosthetic above knee socket showed that the values of maximum stress according to Von-Mises criterion results are increased with the increasing of temperatures. The best designs results of the prosthetic above knee socket are (94.64%) in lamination lay-up 4-2-4 under 20℃ where it took from pervious study. The procedure developed through this work can be used by future researchers and prosthetic designers in understanding how to better design transfemoral prosthesis.
Keywords: Prosthetic; Safety factor; Finite element model; Socket

1. Introduction

In the current socket design, the principles and techniques used by prosthetics in performing such shape modifications are largely based on experience [1].

The socket is a basic component for prosthetic performance. Below-knee amputees generally demonstrate some gait abnormalities such as lower walking speed [1], increased energy cost [2], and asymmetries between legs of unilateral amputees in stance phase cycle, step length and maximum vertical force [3]. Successful fitment of prosthesis may be achieved by understanding the biomechanical structure of socket and its material, weight, thickness in particular to fulfill the desirable load distribution in soft tissues and bone of residual limb. Most commonly used socket design in developing countries is patellar tendon bearing (PTB) socket developed following the World War II at the University of California, Berkeley in the late 1950 s [4, 5]. The Finite Element Method (FEM) has been used widely in biomechanics to obtain stress, strain and deformation in complicated systems and have been identified as an important tool in analyzing load transfer in prosthesis [6]. The finite element analysis (FEA) models have been used to study the effects of the inertial loads and contact conditions on the interface between prosthetic socket and stump of an amputee during gait cycle [7,8].The finite element method has been used as a tool for parametric study and evaluation of prosthetic socket[9,10]. It is common for amputees to experience pain and discomfort in the residual limb while wearing the prosthetic socket [11]. For a lower limb amputee, the comfortableness of wearing prosthesis depends on the distribution of stress at the interface of residual limb and prosthetic socket is either at the pressure-tolerant (PT) or pressure-relief (PR) areas. By employing the technology of computer-aided engineering, the quality uncertainty and labour intensity of traditional process of fabricating a prosthetic socket can be improved. Lower limb prosthesis allows ambulation and improves the performance of daily routine activities. However, poor fitted socket can lead to complications that have adverse effects on the activity level and gait cycle of people with lower limb amputation [12]. The interface between the stump of lower limb amputees and their prostheses is the prosthetic socket.

The contact pressure at the residual limb and prosthetic socket interface is an essential index, and is considered as a promising measure towards good socket design. Therefore, the fundamental concern is to understand pressure distribution at the stump-socket interface. Although the use of pressure sensor is a direct experimental approach towards estimating interface pressure, the analytical approach is an alternating to the experimental one, and finite element modelling of the socket has been used to analyse the contact pressure. Although, the complex features of the soft limb tissues and of their interaction with the socket still remains difficult to model [13]. The variation of interface pressure between the stump and socket is an important factor in socket design and fit. Lower limb prosthetic socket users experience pressure between the stump and socket during daily routine activities. The underlying soft tissues and skin of the stump are not habitual to weight bearing; thus, there is the risk of degenerative tissue ulcer in the stump because of cyclic or constant peak pressure applied by the prosthetic socket [14]. The pressure also can lead to various skin deases such as follicular hyperkeratosis, allergic contact dermatitis, infection and veracious hyperplasia [15-17].

Despite significant scrutiny in the field of prosthetics in the previous decades, still many amputees experience pressure ulcers with the use of prostheses. Sometimes, skin problems lead to chronic infection, which may necessitate re-amputation. This will obviate the long-term use of prosthesis, which indicatively reduces the routine activities of prosthesis users and the quality of life [18]. Many studies have concentrated on interface pressure magnitude between the socket and stump during level walking [19-20]. This paper will address finite element analysis of the socket prosthetic prescription. Specifically, the contribution of Mechanical properties of prostheses mechanical characteristics of socket that include stress, total deformation and safety factor will be discussed.

2. Geometry

To create an FE model, first the geometry of the modelled objects needs to be obtained, including the shapes of the free residual limb, pylon, the socket, foot and the adapter if involved.

Fig**. (1):** Prosthetic hold socket.

Although some models were based on ideal and simplified geometry, an accurate description should be based on an actual geometry as shown in Fig.1. The main object of this study is the prosthetic socket. The simplified geometry of his residual limb was modeled in Pro-engineer and then it is being imported (in IGES formate) and modified in ANSYS 14.0 Workbench.

3. Defining the Analysis Type and Applying Load

The term 'load' includes the boundary conditions (constraints, supports, or boundary field specification), as well as other externally and internally applied loads. The load which is used in the ANSYS workbench version 14 software will be fixed support at the adapter of socket. While, the interface pressure is distributed according to particular positions. Fig. (2) and Fig. (3) Shown values with positions of pressure distributions of present experimental case study. In fatigue solution, the fatigue tool is used to find the equivalent stress, maximum shear stress, total deformation, safety factor, and life at particular loads.

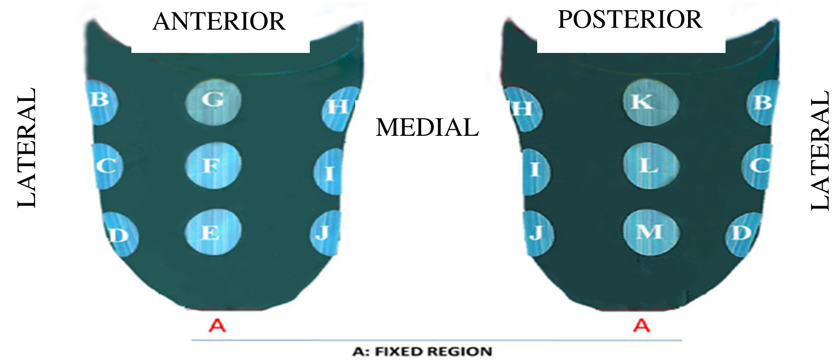

Fig. (2): The distribution of pressure position at socket.

A A:FIXED REGION A

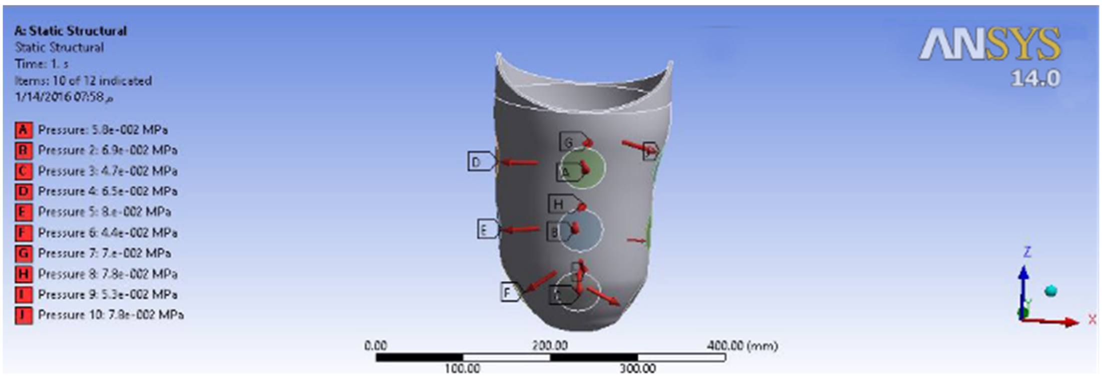

Fig. (3): Applying the pressure forces values and locations to the model in ANSYS workbench version 14.

Also the three dimension finite element model is developed for varying the length of the socket of high, small diameter and bigger diameter 350 mm, 60 mm and 177mm.

Table (1): Voltages versus pressures distribution values of four regions for prosthetic above knee socket [21].

Region	**Voltage (V)**	**Interface Pressure Distribution (KPa)**	**Muscles Group**
Lateral	1.85 2.1 1.25	58 69 47	**B**:gluteus medium. **C**:gluteus maximum. **D**:illiotibial band.
Anterior	2.4 1.95 1.1	42 65 78	**E**:vastus lateraii. **F**: rectus femoris. **G**: rectus mediais.
Medial	1.6 2.07 1.15	70 78 53	**H**:adductor longus. **I**:gracilis. **J**:sartorius.
Posterior	2.25 2.2 1.33	65 80 44	**K**:biceps femoris. **M**:semitendinosus. **L**:semimembranosus.

Also Table [1] shows the maximum recording values of voltages which are calibrated to interface pressures distribution by using calibration [21]. The maximum value appeared on Posterior view of socket.

4. Finite Element and Validations

The predicted resultant shear stresses were less than the experimental values at all measured sites. Best matches were achieved at postero-distal and anteroproximal sites. Consistent mismatches were seen[22,23] measured the interfacial pressure up to 283 kPa, and shear stress up to 44 kPa, at the critical regions of BK sockets using triaxial transducers. The FE analytical pressures up to 226 kPa were, on average, 30% lower than those measured. The difference might result from the fact that the load applied to the prosthetic limb in the FE model was static and equal to the body weight, whereas the actual load during walking was dynamic and larger than the body weight. The comparison suggests that the dynamic analysis is necessary in future FE modelling studies. Owing to the application of the interface elements to simulate the slip/friction conditions at the stump/socket interface, the shear stresses have better agreement, difference in the magnitudes being 10%, on average, and the directions of the resultant shear stresses are identical to the experimental results.

Predicted normal stresses on a BK stump between 0 and 120 kPa, and the experimental measurements were between 0 and 128 kPa. Silver- Thorn [9] predicted that the pressures on a BK socket varied from 0 to 275 kPa, whereas those measured were 0 to 205 kPa [22].

In general, comparison of the stresses predicted by FE analysis and those measured suggested that the results can be in the same range. However, one-to-one correspondence has not been achieved. The validation is limited to a number of the points where stresses can be measured.

5. Results and discussion

5.1Creation of Mesh in the Model

The meshing process has been done by choosing the volume, and then the shape of element is selected as tetrahedron (Automatic meshing), as shown in Fig.(4).

On the meshed model fixed support is being applied at the distal end of the socket, distal end of the socket is further attached with the remaining parts.

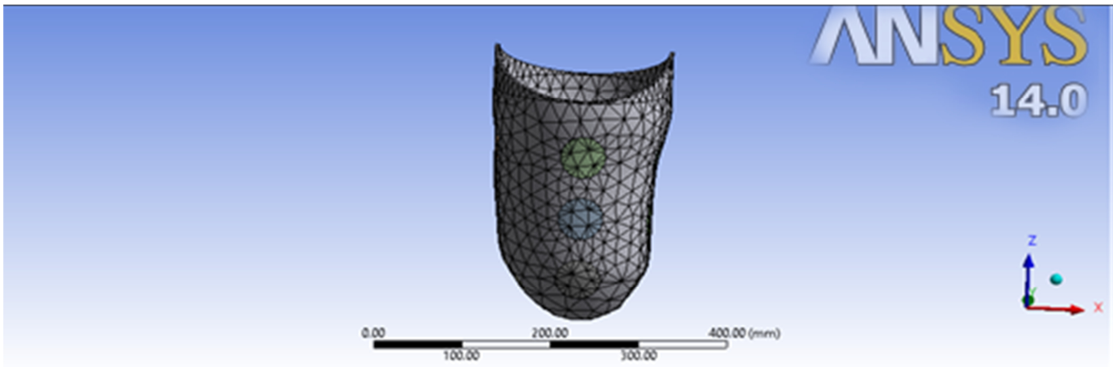

Fig. (4): The model of above knee socket with meshing.

The total number of elements is (3911) with total number of nodes of (8978)).

5.2 Mesh Convergence

One of the important steps in FE modeling is to ensure that the mesh density is fine enough for the mesh to converge. A too coarse mesh utilizing too larger elements will yield inaccurate results while a mesh with unnecessarily small elements will increase the cost of the analysis in terms of time and usage of computer and software resources. Mesh convergence is an iterative process which applies consistent load application and boundary conditions while altering element size until the results of the analysis converge on a similar result for two mesh sizes. When further mesh refinement produces no change in the results, or only changes them below a determined threshold, the mesh is sufficiently converged. This process can be visualized by plotting a curve of element size vs. stress results. The optimum mesh density is defined at the point on the curve at which the percent difference from one analysis to the next is below a specified threshold. The mesh convergence process was applied on the residual limb as it is the volume likely to observe highest deformation and is the volume of most interest.

With a threshold error of 1%, the mesh analysis yielded that the mesh converged with 229340 elements.
Table 2 shows the details of the mesh convergence iterations, including element size, number of elements and nodes, highest von Mises stress, and percent error. Fig. 5 shows how the relationship between the stress and number of elements in a graphical manner.

Table (2): Mesh converge check

Iteration	element size (mm)	number of elements	number of nodes	Von Mises stress (kPa)	Error calculation, % (Von Mises Stresses)
1	6	52021	11834	61	14
2	5	65398	14675	71	6.5
3	4	126141	27696	76	3
4	3	229340	50332	78.4	.75
5	2	268425	58427	79	-

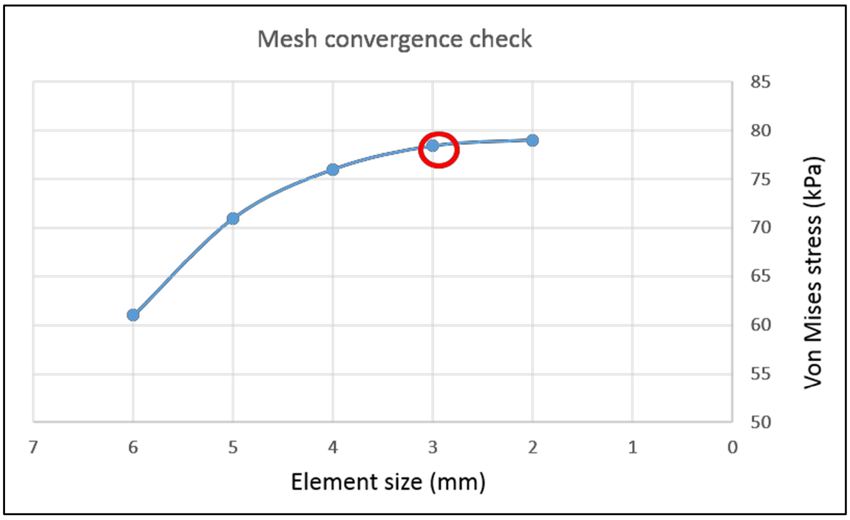

Fig. (5): Mesh convergence plot: Element size vs Von Mises stress

5.3 Material properties

The experimental results are including physical, mechanical and thermal properties results for all laminations composite materials of prosthetic above knee socket which are made from perlon / fiber glass / perlon with acrylic. Table 3 presents the material properties and the main stress is found at 11.11Mpa used in the previous study.

Table (3): The Physical properties for each lamination sample [21].

Lay-up	**Thickness (mm)**	**Cross-Sectional Area (mm^2)**	**Mass (g)**	**Density (g / cm^3)**	**Volume Fraction**
4-2-4	3.92	50.96	8.12	3.186	0.34

Also Fig.6 evidence the maximum value of stress is denoted at 20 ℃.

The figure demonstrate the decreasing in the value of stress according to the temperate degree [21].

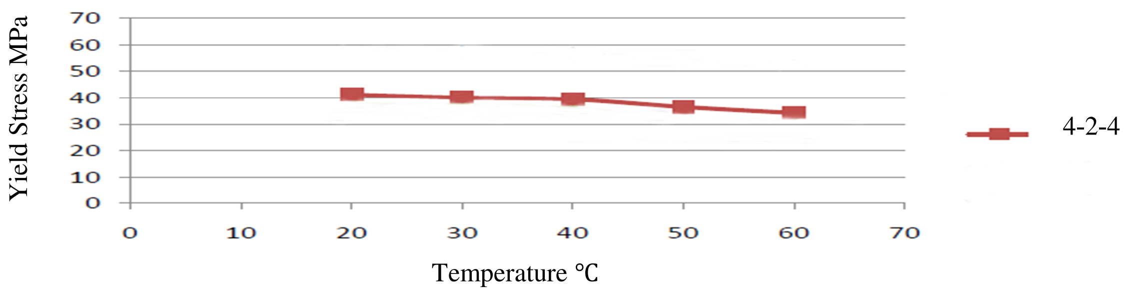

Fig. (6): Yield stress results for each lamination with temperatures [21].

5.4 Numerical Results of safety factor

First of all, to find the safety factor of socket it should use S-N curve where it was getting previously. Fig.7 represents number of cycle with stress. During daily activities of an amputee the total load of knee joint in transtibial prosthesis passes on the prosthetic socket.

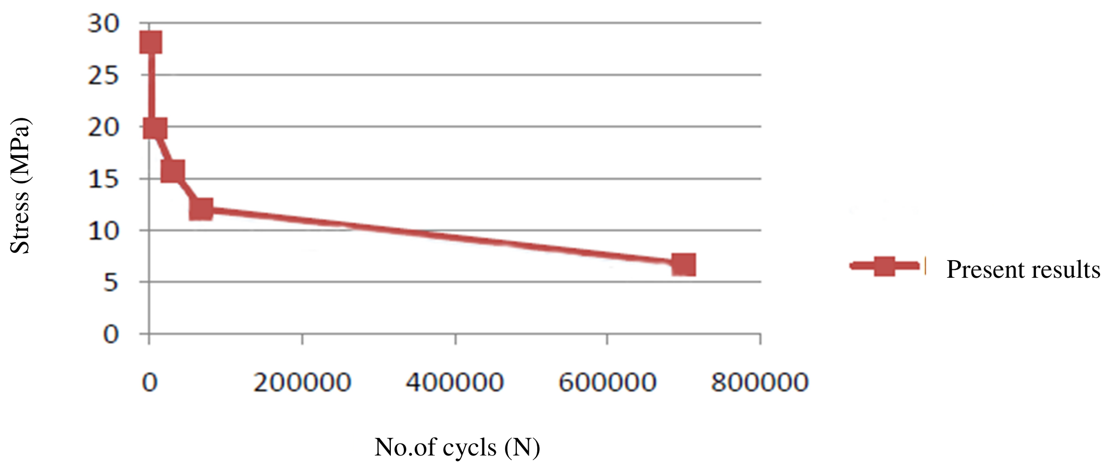

Fig. (7): S-N curves of the results for lay-up (4-2-4) at 20℃ [21].

The model of layers of laminates 4-2-4 showed that the fatigue minimum safety factors results are about 1.2442 without temperature effect or at room temperature 20 ℃) as show in Fig.8.

The maximum value of safety factor is denoted at the top of socket where this region contact to the amputee.

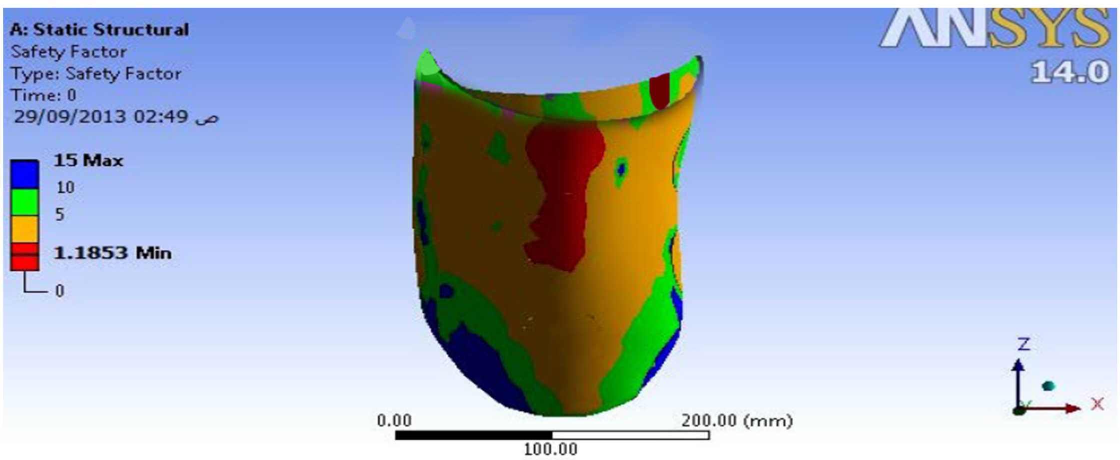

Fig. (8) Safety factor.

5.5 The structural Behavior of total deformation

The residual limb deforms during the donning procedure and therefore stresses develop in the limb. The analysis predicted the maximum interference to occur near the bottom of the residual limb. Not surprisingly, this is also the location of highest contact pressure.

The values of maximum total deformation of the material were analyzed and shown in Fig. 9, and it is found that as the maximum value of total deformation is 5.049 mm in case of 4-2-4 material. The maximum value is found at the top of socket which is refer to the failure of socket.

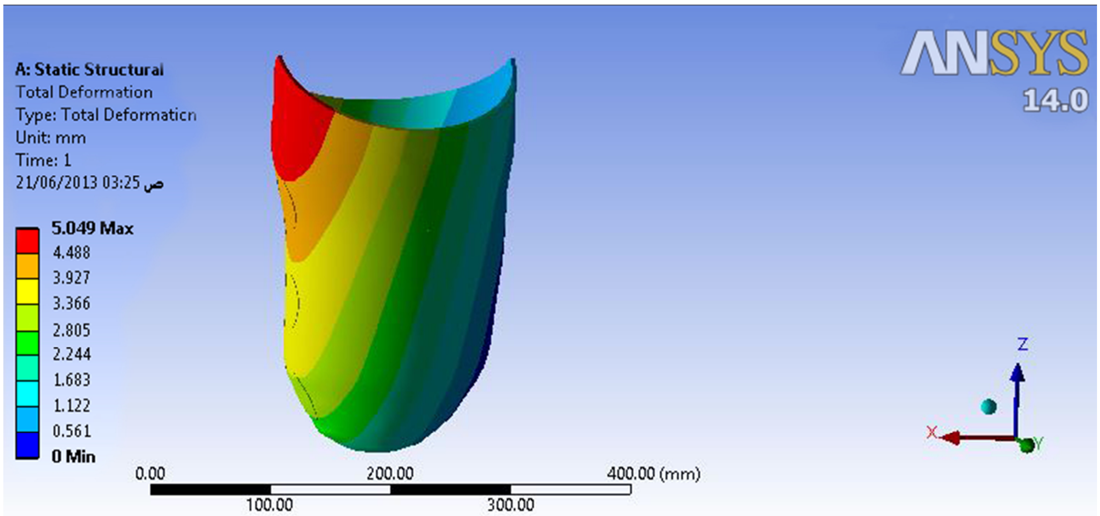

Fig. (9) Total deformation.

5.6 Von-Misses Stresses Distribution

According to the Von-Mises theory the yield stresses are considered as criteria; ($\sigma e < \sigma y$, safe), ($\sigma e = \sigma y$, critical) and ($\sigma e > \sigma y$, failed) where, (σe) is the equivalent stress, and (σy) is the yield stress.

The contour stresses distribution of prosthetic above knee socket for each laminations are obtained through ANSYS workbench. A slight increase in stress appeared under elevated temperature effect. The major increase in stress was located at the pattela tendon. The model of layers of laminates 4-2-4 showed that the maximum stress fatigue (according to Von-Mises criterion) results are about 7.4482 as shown in Fig.10.

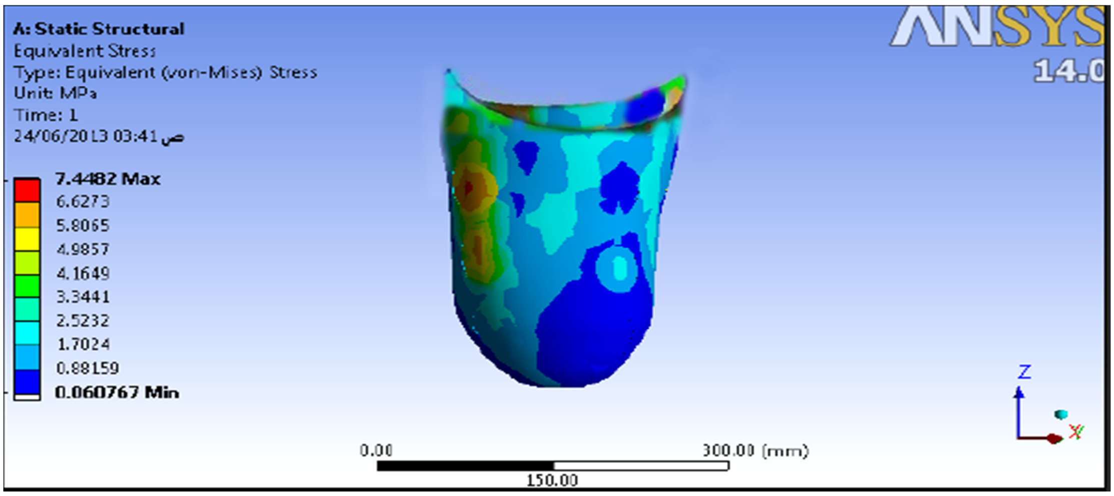

Fig. (10): Von-Mises stresses contour results of laminate 4-2-4 at 20°C.

Conclusion

This study establishes the development of a suitable modeling and analysis of prosthetic sockets using ANSYS Workbench. It have described state of the art mesh generation procedure using finite element method. It obtained in the simulation show good agreement with the measured loads in various regards. In addition, the results summarized that assimilating local submissive properties within socket wall can be an effective methods to distribute maximum stress areas and also to relief contact pressure between the socket and stump. Moreover, Based on the results and the discussion, the composite material is cheap that this study depend on it from previous study is excellent strength, widely available but it has high weight that make it only useful to be used for adult with higher weights. The results obtained from analysis can be used as a reference to choose socket material, thickness and its optimal length for manufacturing of socket in developing countries. The socket buildup of composite material gives the optimal solution for patellar-tendon bearing socket design. Also, the results that obtained gave satisfy safety factor according to the composite materials that used.

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