Finite Element Modelling and Analysis of Trans-Tibial Prosthetic Socket

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Abstract

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. A modified three-dimensional finite element model of the patellar tendonbearing (PTB) 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. All essential materials used in the analysis were assumed to be homogeneous, linearly elastic and isotropic. A variety of materials were used for the analysis of the socket like Polypropylene, Composite,90/10 PP/PE, HDPE and LDPE. Analysis was done on a various thickness of socket and of different length along with of different materials commonly applied in developing countries. For boundry condition, fixed support was applied to the distal end of the socket and vertical loads were applied under static condition at pattelar tendonbrim, medial tibia, lateral tibia and popliteal area during stance phase of gate cycle.

***Index terms*—** patellar tendon-bearing (ptb), trans-tibial (tt) prosthesis, finite element (fe) model, socket/stump interface stress.

1 Introduction

he socket is a basic component for prosthetic performance. Below-knee amputees generally demonstrate some gait abnormalities such as lower walking speed [1], incresed 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 pattelar 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 analysing 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 Author ? ? ?: M Tech, Mechanical Engineering Dept., ISM Dhanbad, India. e-mails: imran37708@gmail.com, rameshjaiswal2k8@gmail.com, lyogimee7@gmail.com during gait cycle [7,8].The finite element methode 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, poorfitted 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 modeling 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][16][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].

2 a) Trans-Tibialprosthesis Description

The artificial limb consists of a foot-ankle unit which needs to be attached to the remainder of the amputee's natural leg or stump. The foot ankle unit is attached directly to the socket frame. The artificial shank can be attached to the foot ankle unit and then attached to the socket frame for a below-knee amputation. Today the sockets are roughly quadrilateral in shape. They attempt to have total contact between the stump and the socket.

3 II.

4 Finite Element Model

A frequently used numerical analysis technique in biomechanics is the finite element technique, a computational approach for interface stress or structural deformation calculation evoluted in engineering mechanics. It has been introduces as a useful tool to understand the load transfer mechanics between a residual limb and its prosthetic socket. The finite element technique is a full-field analysis for calculating the state of stress and elestic strain in the specific field. This technique is well suited for parametric analysis in the process of design. The previous finite element analyses showed the significance of considering prestress in predicting interface stresses at loading stage [21][22].

One left unilateral male trans-tibial amputee participated in this study. The volunteer was 45 years of age, 166 cm tall, 70 kg in mass, and the cause of his amputation was an accident. He has been an active amputee for five years, using his prosthetic limb for all his daily chores. 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.

The finite element model was kept as simple as possible in terms of material properties and boundary conditions. Different materials for 2 mm, 3 mm, 4 mm, 5 mm, and 6 mm unit volume layer thickness was used for creating the 3-D FE model. Also the three dimension finite element model is developed for varrying the length of the socket of 16 cm, 17 cm, 18 cm, 19 cm, 20 cm, 21 cm. The model was meshed with brick element solid 185 with fused tibia and fibula bones. A total of 41,073 elements and 20,438 nodes were used. IV.

5 Results

The results for total deformation, shear stress and equivalent von-Mises stress of developed and modified transtibial socket model were obtained by using (ANSYS Workbench v14.0) program. Figure (1) shows the meshed view of the finite element three dimension socket model, and figure (2) shows the maximum von-Mises stress developed for different loading conditions of stance phase like Initial 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 of the prosthesis like shank, ankle foot. The different loading conditions as listed in table 2 were quasi-static approximations using experimentally obtained maximum vertical ground reaction for the prosthetic side of same subject while walking at a given speed using CGD gait cycle analyzer [24][25][26][27].

6 III.

7 Material Properties

In this analysis the different material used are composite, polypropylene, 90/10 PP/PE (90% polypropylene and 10% ethylene), high density polyethylene (HDPE) and low density polyethylene (LDPE). The mechanical properties of the socket material were assumed to be linearly elastic, isotropic and homogeneous. Socket were analyzed for different materials and their valus of Young's modulus, Ultimate strength, Poisson's ratio and density is listed in table 1.

8 Tsai-Hill Criterion, C TH

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Where CTH is the Tsai-Hill failure coefficient, Sv, St and Ssh are the ultimate strengths of composite in the vertical, transverse and shear directions respectively listed in table 5 and ?1, ?2 and ? are the imposed stresses in the longitudinal, transverse, and shear planes. If the value of CTH is less than one than design is safe. The values of maximum von-Mises stress, shear stress and total deformation off all the material in different length were analyzed and shown in figures 11-13, and it is found that as the length of socket increses the values of stress and deformation decreases. The decrease in value of deformation as increase of length is higher in case of LDPE material.

9 d) Structural Behavior vs Thickness

The values of maximum von-Mises stress, shear stress and total deformation off all the material in different thickness were analyzed and shown in figures 8-10, and it is found that as the thickness of socket increses the values of stress and deformation decreases. The decrease in value of deformation as increase of thickness is higher in case of LDPE material.

10 Strength (in MPa)

11 Sv

St Ssh The variation of factor of safety as a function of Weight of the socket for a socket of different thickness is shown in figure **??**, where factore of safety is being calculeted by dividing maximum von-Mises stress at a load of 620 N with the endurance limit (50% ultimate tensile strength value) [28]. During daily activities of an amputee the total load of knee joint in transtibial prosthesis passes on the prosthetic socket. During normal walking , the total joint reaction forces at knee joint is three to four times increases than the total body weight, during jumping and fast running load on knee joint increses more [29]. Therefore, six factore of safety is minimum desirable to withstand the loading of socket. The factor of safety is just below the level of five for LDPE and HDPE so, it can be suggested that LDPE and HDPE are note suitable for prosthetic socket design.

12 b) Case 2 : Analysis of failure

The finite element simulation result of rotation and displacement in different parts of socket validate the biomechanical requirement of structural integrity in patellar tendon bearing socket. Figure **??** shown below describes the variation of Tsai-Hill coefficient with tensile and compressive strength. The value of CTH coefficient in 2mm thick composite for tensile strength is 0.1864wich is only five times factore of safety but thickness between 3 mm (0.0724) to 4 mm (0.031) has a factore of safety more than twenty times. Therefore, the optimum solution of composite material of thickness 3 mm to 4 mm satisfied the Tsai-Hill criterion. In all materials it is found that the von-Mises stress, shear stress and total deformation is inversely proportional to thickness except for LDPE of the socket figure 8-10. However the stress and stress variation were higher in case of 2mm and 3 mm socket and it is relatively low in case of 4mm and 6mm. Thus 3 mm to 4 mm could be a optimal solution in terms of thickness of the socket for all materials where this much thickness is used. The variation of von-Mises stress, shear stress and total deformation for different thickness of prosthetic socket were shown in figure **??**, 9 and 10 respectively. The value of total deformation in case of LDPE of thickness less than 3 mm goes higher and it may loss biomechanical load bearing ability. Thus the result indicates that the LDPE socket length is not suitable fore fabrication of PTB socket of below 4 mm thickness. In all materials it is found that the von-Mises stress, shear stress and total deformation is inversely proportional to length of the socket figure 11-13. However the stress and stress variation were higher in case of 16 and 17 cm length socket and it is relatively low in case of 19cm and 20cm. Thus 19 cm to 20cm could be a viable solution in terms of length of the socket for all materials where this much length is possible. The variation of von-Mises stress, shear stress and total deformation for different length of prosthetic socket were shown in figure **??**1, 12 and 13 respectively. The value of total deformation in case of LDPE of length less than 16 cm goes higher and it may loss biomechanical load bearing ability. Thus the result indicates that the LDPE socket length is not suitable fore fabrication of PTB socket of below 16cm length.

13 Conclusions

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. Based on the results and the discussion, the composite material is cheap, excellent strength, widely available butit 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. The study reconnoitered further future scope for parametric analysis, investigating the effects of liner, socket stiffness, rectification scheme, soft tissues, and materials for the socket/stump interface stress distribution. Review of secondary physical conditions associated with lower-limb amputation and long-term prosthesis use. [1]

Figure 1:

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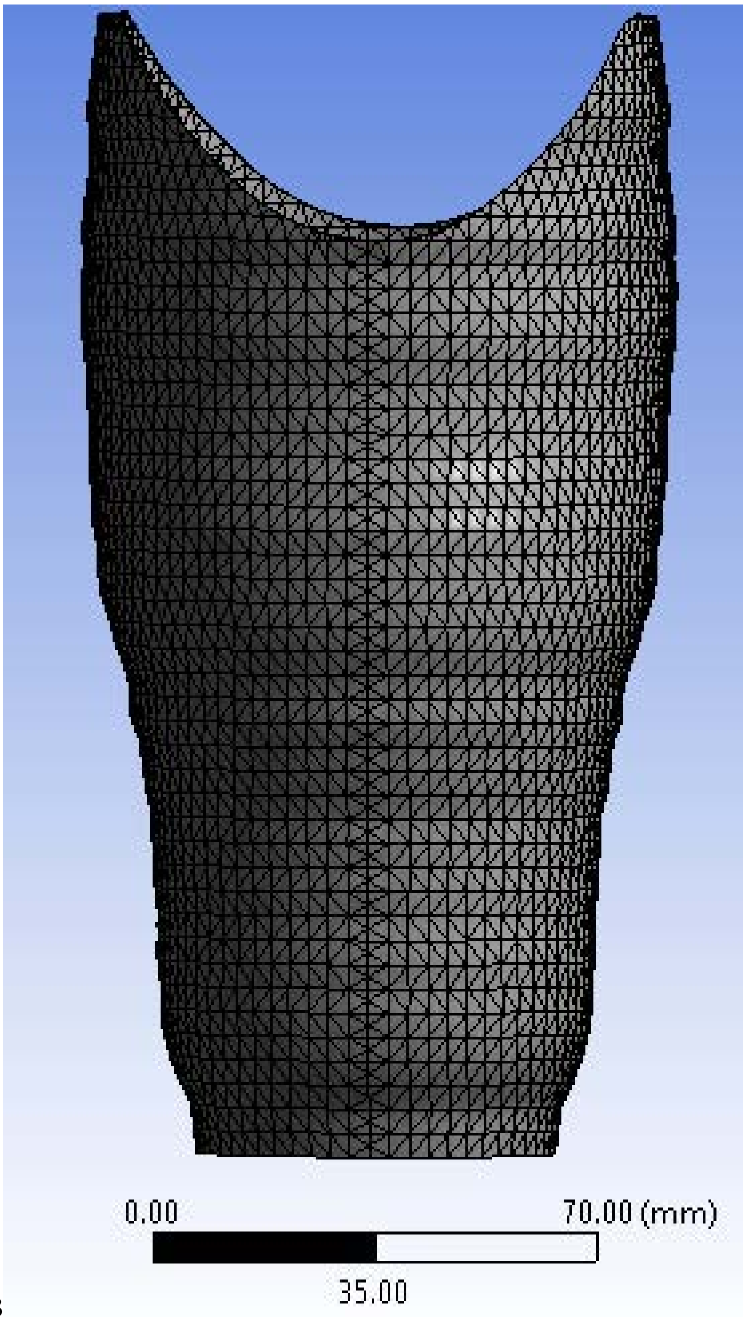


Figure 2: Figure 1 :Figure 3 :

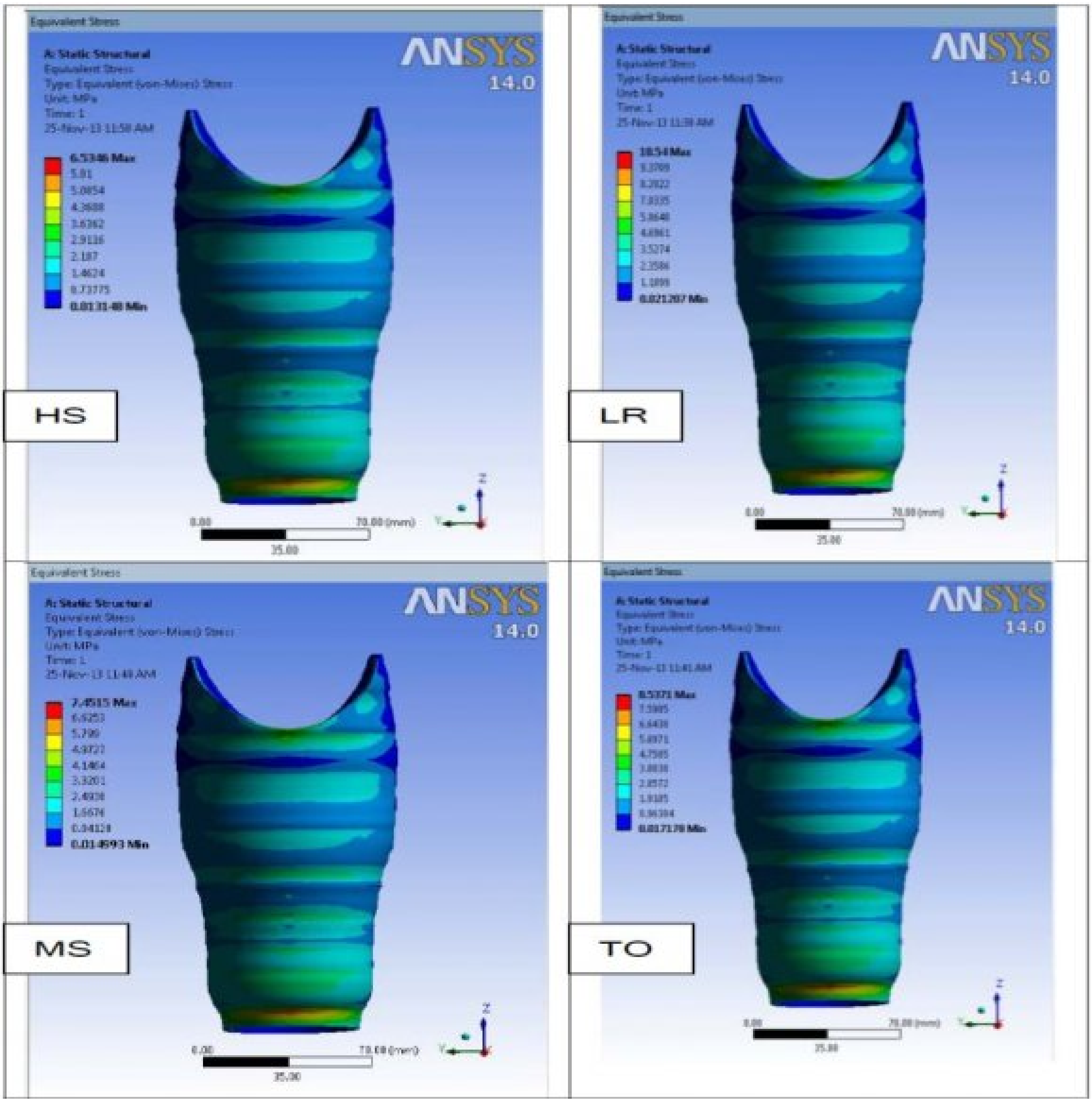
3 

Figure 3: Figure 3 :

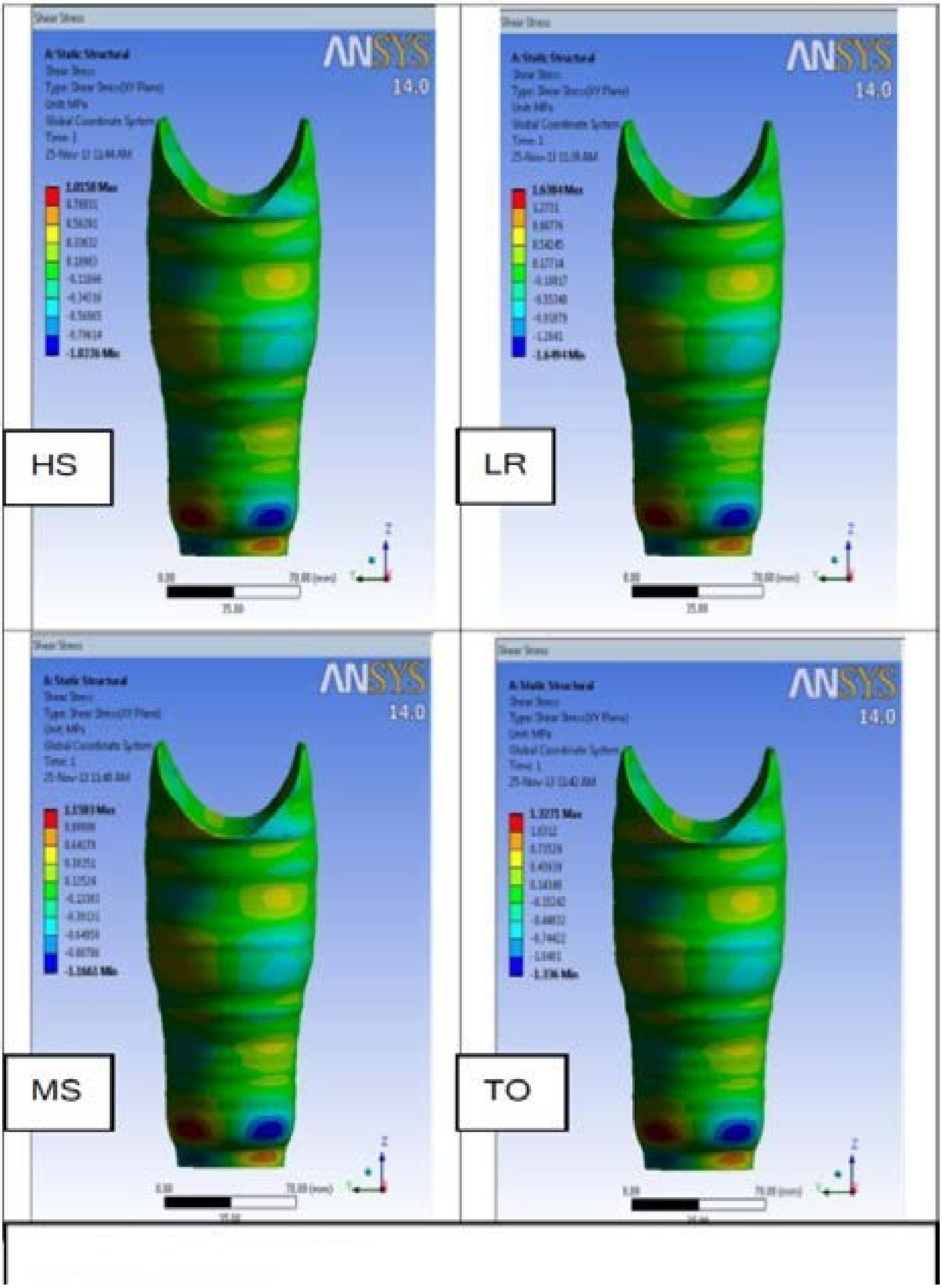


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Figure 4: Figure 4 :

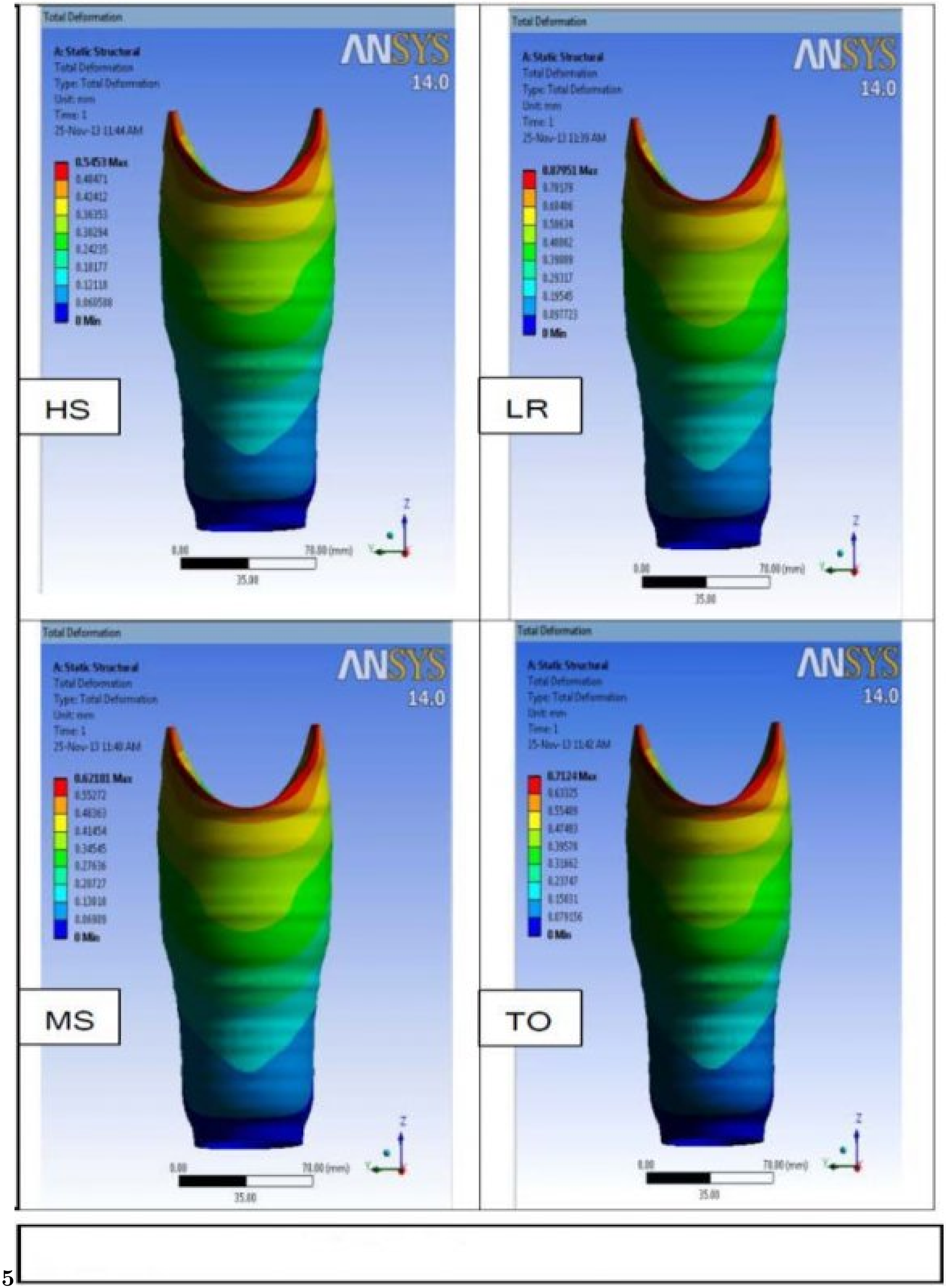

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Figure 5: Figure 5 :

1

Figure 6: Table 1 :

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Figure 7: Table 2 :

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Stance phase	von-Mises stress (MPa)	Shear stress (MPa)	Total deformation (mm)
Initial Contact/Heel Strike (HS)	6.55	1.02	0.55
Foot Flat/Loading Response (LR)	10.54	1.64	0.88
Mid-Stance (MS)	7.45	1.16	0.63
Terminal Stance/Heel Off (HO)	10.54	1.64	0.88
Pre-Swing/Toe Off (TO) of stance phase	8.54	1.33	0.72

Figure 8: Table 3 :

4

Thickness (mm)	Composite	PP	PP/PE	HDPE	LDPE
2	140	106	102	111	107
3	209	160	155	168	174
4	280	212	208	214	222
5	350	266	261	278	269
6	420	320	315	334	323

Table 5 : Tensile and compressive strength of composite [31]

c) Structural Behavior vs Length

Figure 9: Table 4 :

Tension Compression 584 803 43 187 Year 2014 44
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Figure 10:

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