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Design of Monolimb Using Finite Element Modelling and Statistics-Based Taguchi Method

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ABSTRACT

Background. Monolimb is a transtibial prosthesis having the socket and shank molded into one piece of thermoplastic material. If properly designed, the shank of a monolimb can have a controlled deflection during walking which simulates the ankle joint motions to some extent. However, there is no clear guidance for the design of monolimb considering the dilemma between shank flexibility and structural integrity.

Methods. Finite element analysis was used to simulate structural tests based on ISO10328 on monolimbs of different configurations. Statistics-based Taguchi method was employed to identify the significance of each design factor in controlling the deformation and stress within monolimbs. The design factors considered were the thickness of the thermoplastics, anteroposterior and medialateral dimensions of the elliptical shank, and depth of the posterior seam line. By progressively fine-tuning the design factors, the monolimb configuration was optimized giving offering appropriate flexibilities of the shank and would not structurally fail in normal uses. Experimental structural test was used to validate the FE model.

Findings. Anteroposterior dimension of the shank was shown to be the most important design factor determining the peak von Mises stress values, deformation and dorsiflexion angles of monolimbs. Depth of seam line appears much less important than the other three factors. A monolimb fulfilling the design requirements was suggested. Experimental test results reasonably matched with the FE results.

Interpretations. FE analysis and Taguchi method was shown to be an effective method in optimizing the structural design of prostheses. Further prosthetic design can be facilitated based on the degree of importance of the design factors on the structural behavior of the prosthesis. Gait analysis of amputees using the suggested monolimb design is needed in the future.

Keywords: Finite element model; Taguchi method; Prosthetics; Optimization; Stress; Deformation.

INTRODUCTION

It is common that transtibial amputees demonstrate some gait abnormalities such as lower walking speed (Molen, 1973), increased energy cost (Waters et al., 1976) and asymmetries between legs in terms of stance phase time, step length and vertical peak force (Robinson et al.,1977). Loss of active dorsiflexion and plantarflexion motions of the ankle joint of amputees is one of the reasons of the gait abnormalities (Bowker and Kazim, 1989). Prostheses have been designed to compensate for the loss of motions at the foot by incorporating energy storing and releasing (ESAR) capabilities using flexible keels or shanks. The Seattle foot^{TM} (Seattle System**,** Poulsbo, WA 98370 USA) and FlexFoot^{TM} are examples of ESAR prosthetic components. Previous research suggested that many amputees subjectively prefer ESAR prosthetic feet to conventional solid ankle cushioned heel (SACH) feet on normal and fast walking (Macfarlane et al., 1991; Menard and Murray, 1989). However, many amputees still utilize the simple conventional SACH designs because of their lower cost.

A "monolimb" prosthesis using a conventional prosthetic foot perhaps is an alternative to ESAR prosthetic feet if properly designed, providing elastic response of the shank (Valenti, 1991), at the same time lower the total prosthetic weight and cost. Monolimb refers to a transtibial prosthesis having the socket and the shank molded into one piece of thermoplastics. Different names have been used for this kind of prosthesis such as endoflex (Valenti, 1991), total thermoplastic prosthesis (Rothschild, 1991) and ultra-light prosthesis (Reed et al., 1979). Due to the flexibility of thermoplastics, the shank of a monolimb can deflect during walking. By optimizing the uses of material and structural design, it is possible that the shank deflection be altered such that the natural ankle joint motions are mimicked so as to enhance the comfort and gait performance. Positive feedbacks were gained including improved gait efficiency and comfort from patients using prostheses with deformable shanks (Beck et al., 2001; Coleman et al., 2001). It has also been shown that increased shank flexibility can have potential in reducing the prosthetic socket- residual limb interface stresses (Lee et al., 2004). At the same time, it should be noted that the structural integrity has to be maintained without permanent deformation of the prosthesis. Until now there is no clear guideline for the shank design of monolimb.

Currently, the structural test specifications of lower limb prostheses are specified in ISO10328. To optimize the design of the shank, monolimbs with different shank designs have to be subjected to tests according to the ISO standards. Performing such test experimentally is expensive and time demanding. Computational analyses such as finite element (FE) modeling, allow parametric study to be performed easily without the need to fabricate prostheses. FE analysis has been widely used in lower limb prosthetics in the past decade. In previous FE models, focus was on investigating the interface contact between the prosthetic socket and residual limb as reviewed by Zhang et al, 1998. The design of a well-controlled deformability of a prosthesis, however, received little attention.

There are a number of methods to find an optimum configuration of a monolimb. "Vary-one-factor-at-a-time" is one popularly used design optimization method in which the effect of one factor is assessed by varying only the factor to be assessed and keeping the other factors fixed at a specific set of conditions. However, this method can sometimes lead to wrong results as the effect of the factor can be changed if other factors are substituted with different conditions (Phadke, 1989). If full factorial is run exploring every possible combination of values of each factor, the total number of simulations required will be very high. A statistical approach developed by Taguchi (Margolis, 1985) utilizes an orthogonal array, which is a form of fractional factorial design containing a well-chosen subset of all possible combination of test conditions. Using Taguchi method, a balanced comparison of levels of any factor and significant reduction in the total number of required simulations can both be achieved.

This paper demonstrates a technique using computational modeling and statistical-based method in optimizing the design of monolimb. FE analysis was used to predict the deformation and stress at monolimbs of different material thickness and shank geometry subjected to loadings based on ISO10328. Taguchi method was used to identify the importance of each design factor and suggest an optimized monolimb design which can resist failure on normal uses and can provide appropriate flexibility.

METHODS

A. Finite element modeling

A plaster cast was taken on a male left-sided transtibial amputee. The cast was digitized using BioSculptor™ system and exported to prosthetic CAD software ShapeMaker™ 4.3. A prosthetist using ShapeMaker™ prepared the geometry of the prosthesis by applying built-in shape rectification template to the digitized limb and aligning a shank blended into the socket. Different designs of the prosthesis, as shown in Table 1 and 2, and Figure 1, were created. Using commercial CAD software SolidWorks™ 2001, the socket together with the shank was given a specified thickness. Foot block, socket filler, extension rod and block were added (Figure 2a), so that the load application points and direction of force can be applied as instructed in ISO10328. The socket filler only extended from the proximal brim of the socket to approximately 10cm above the distal end of the socket to allow the distal part of the socket to deform upon load application. A foot bolt and a screw (modeled as a cylinder) were inserted (Figure 2b). To simulate the screw and foot bolt fixing the monolimb onto the foot block, the screw was rigidly tied with the foot block as well as the foot bolt and contact was defined among the foot bolt, distal end of the shank and foot block so that their surfaces were not allowed to penetrate each other when loading was added. The model in its entirety, as shown in Figure 3b, was exported to ABAQUS version 6.4 (Hibbitt, Karlsson & Sorensen, Inc., Pawtucket, RI, USA). A FE mesh of 3D tetrahedral elements was built using ABAQUS auto-meshing techniques. Tetrahedral element was chosen because of geometrically irregular structures of the monolimb. The number of elements ranged from 27,337 to 39,029 depending on the shape of the shank.

The Young's modulus and Poisson's ratio of the monolimb made of polypropylene homopolyer were 1500MPa and 0.3 respectively (Margolis, 1985). Foot block, adaptor, screw, extension rod and block were assumed to be rigid. The bottom load application point was fixed and loadings were added at the top load application point (Figure 2a). The loadings applied were based on ISO10328. The standard specifies loads for testing prosthesis at normal walking load and occasionally severe load during heel strike (loading condition I) and heel off (loading condition II) of the gait. We observed in our previous structural test experiment and FE analysis (Lee et al., 2004) that the loading condition II caused much more deformation and higher stresses to monolimbs than loading condition I because of the longer moment arm. Facture failure of monolimb was unlikely under the loading condition specified in ISO10328, due to the high ductility of thermoplastic material. However, permanent deformation could occur in some monolimb designs which is undesirable as it permanently changes the alignment of prosthetic foot relative to the socket. Based on the above information, force specified in ISO10328 simulating heel off at normal walking load was used to load the prosthesis in the FE model during the design stage. The selected test load level was A80 (1085N). There are three test load levels specified in ISO10328 which accounts for the different amputee body weights. A80 is for amputees whose weights are between 60kg to 80kg. Geometric nonlinearity resulted from the large deformation was considered in the model. Peak von Mises stress, displacement of the top load application point, dorsiflexion and inversion angles defined as the angle changes between the top and bottom aluminium blocks in sagittal and frontal planes respectively were predicted in the FE model. Through testing different designs, the aim was to design a prosthesis providing high flexibility but without permanent deformation under normal walking.

B. Taguchi method and design optimization

Four design factors namely, the thickness (T) of the thermoplastic material, depth of posterior seam line (S), anteroposterior (AP) and medial-lateral (ML) dimensions of the shank (Figure 1) were selected for evaluation. Each factor was assigned with 4 levels (Table 1). The thickness of the polypropylene material was in the range of 4-7mm. The AP/ ML dimensions and depth of the seam were in the range of 25-55mm and 0-15mm respectively concerning the bulkiness and the structural integrity of the monolimb. To identify the relative significance of the design factors using full factorial approach, a total number of total number 256 analyses (4^4) are required. In this study, a statistics-based-Taguchi method was used to reduce the number of analyses. Sixteen simulations were required. The configurations of the monolimbs were shown in an orthogonal array L_{16} (Phadke, 1989) in Table 2. The mechanical responses namely, von Mises stress, displacement of the top load application point, dorsiflexion and inversion angles were predicted by 16 FE analyses. The mean effect of each level of the four design factors on the mechanical responses was computed. For example, the mean response of thickness at level 1 [$R(T_1)$] on peak von Mises stress is calculated as the mean stress over trial 1 to trial 4. An analysis of variance (ANOVA) was performed calculating the sum of squares of each design factor to determine sensitivity of each design parameter. For example, the sum of squares due to thickness would be equal to $4[R(T_1)-R_m]^2+4[R(T_2)-R_m]^2+4[R(T_3)-R_m]^2+4[R(T_4)-R_m]^2$ where $R(T_1)$, $R(T_2)$, $R(T_3)$ and $R(T_4)$ were mean response of thickness at level 1 to 4 respectively and R_m was overall mean response over 16 trials.

Using a superposition model (Phadke, 1989), the mechanical response of any combination of levels among design factors can be predicted as following:

$$R(T_i, AP_j, ML_k, S_l) = R_m + [R(T_i) - R_m] + [R(AP_j) - R_m] + [R(ML_k) - R_m] + [R(S_l) - R_m] \quad (1)$$

The equation can be read as the response (R), when the design factors which were thickness (T), shank antero-posterior (AP) and medial-lateral (ML) dimensions and depth of posterior seam (S) were set at levels as indicated at their subscripts (i, j, k and l), would be equal to the mean response of all 16 runs (R_m) plus the deviations from R_m caused by setting the four factors at the levels. FE analysis was carried out as a confirmation run to inspect the predictive power of the superposition model.

Using the results of the superposition model and the sensitivity analysis of each factor, levels of each factor were manually adjusted such that the prosthesis can provide high flexibilities without inducing permanent deformation. Using the strength theory, permanent deformation will occur if the peak von Mises stress was equal to or larger than the allowable strength of 26.4 MPa which is equal to the yield strength of the material divided by the factor of safety. The yield strength was 33MPa for polypropylene homopolymer (Margolis, 1985) and the factor of safety was 1.25. Flexibility of the prosthesis was quantified by the computed vertical displacement of the top load application point. Attempt was made to maximize the displacement, at the same time, the deformation of the monolimb should be less than 15 degrees for dorsiflexion angle and 5 degrees for inversion angle.

C. Experimental test

The optimized prosthesis based on Taguchi method was fabricated and structural test was performed to validate the FE model. The monolimb was made of polypropylene thermoplastic material fabricated by drape molding on a foam milled by BioSculptor^{TM} milling machine. The distal 10cm of the prosthetic socket was filled with sponge to allow the deformation at the distal end of the socket to deform after loading is applied. Above the sponge the socket was filled with plaster of Paris embedding a mandrel. The mandrel was aligned along the shank and passed through the estimated knee joint center. Two aluminium blocks were attached to the mandrel and at the distal end of the shank by screws and bolts. Both aluminium blocks had mounting holes with ball joints attached. The aluminium blocks were positioned such that when loadings were applied to the ball joints the position and direction of the load would comply with the specifications in ISO10328. The monolimb was mounted onto a material testing machines (Model 858 Mini Bionix, MTS System Corporation, Eden Prairie, Minnesota).

The monolimb was loaded to 1085N at the loading rate controlled at 100N/s. A mechanical digitizer was used to record the spatial coordinates of the aluminium blocks for the calculation of the "dorsiflexion" and "inversion" angles. Measurement was made immediately after 1085N was applied to reduce the creeping effect. The displacement of the top load application point was recorded real-time during the load-unloading process. Permanent deformation was inspected by studying if the actuator could return to original position upon removal of the load. After load-unloading at 1085N load level, the monolimb was loaded to 2717N which is the occasional severe load specified in ISO10328 A80 load level until the monolimb failed or sustained the test load.

RESULTS AND DISCUSSION

Stress and deformation of the monolimbs under loading were evaluated by finite element analysis which is believed providing a more accurate solution than the analytical method using beam theory when dealing with complicated geometry and boundary condition. Figure 3 shows one typical von Mises stress distribution and the deformation of the monolimb. High von Mises stresses fall over the anterior region of the shank. The stress value is low at the region of the foot bolt. This is because of the stress shielding effect with the foot bolt having much higher stiffness than the thermoplastic material reducing the bending displacement of the thermoplastic material around the rigid bolt. The shank bows towards the back upon load adding. The responses, including peak von Mises stresses, displacement of the top load application point, dorsiflexion and inversion angles of the 16 different monolimb configurations were evaluated by the FE models. Inversion angles were found relatively less sensitive compared with the other three responses when the design factors changed and almost all monolimbs deformed with inversion angles less than 5 degrees.

The mean effect of the design factors at each level can be found in Figure 4. As expected, there is a trend showing the reduction in peak von Mises stress, displacement of the top load application point, dorsiflexion and inversion angles, when the four design factors thickness, cross sectional area and depth of posterior seam line increase. The reduction trend is not obvious when the medialateral is moved from level 2 to level 3. This indicates that the responses are not sensitive to medialateral between level 2 to level 3. Their degrees of importance are different as the slopes of the curves are different. The level of importance of each factor in the structural behavior of the prosthesis can be studied by comparing the sum of squares shown in Table 4. Among the four analyzed design factors controlled at specified levels, anteroposterior dimension of the shank was shown to be the most important design factor determining the peak von Mises stress values, displacement of the top load application point and dorsiflexion angles of monolimbs. As far as the inversion angle of the foot block is concerned, the thickness, anteroposterior and medialateral dimensions of the shank were almost equally important. Depth of seam line appears much less important than the other three factors.

The responses of monolimbs of any combinations of levels among factors can be predicted using equation 1. The displacement of the top load application point would be the highest if the four factors were assigned at level 1. However, the design is deemed inappropriate as the peak von Mises stresses would be much higher than 26.4 MPa and dorsiflexion angles would be much higher than 10 degrees (Table 5). The levels of the factors were adjusted according to their degree of importance and the predicted responses using equation 1. At this stage, the depth of the seam line was fixed at level 2 as it was found less important than the other three factors. Table 5 shows the predicted responses of some selected monolimb designs which were not evaluated in the 16 trials. The six monolimbs listed in Table 5 do not meet the design requirement and require further tuning on the levels of the factors

Some design configurations fulfill the design requirements. One optimized design configuration would be thickness at 5mm, anteroposterior dimension at 25mm, medialateral dimension at 45mm, giving peak von Mises stress 26.1 MPa and displacement of top load application point 26.5 mm predicted by equation 1 when loading simulating heel off was applied. The dorsiflexion and inversion are not deviated significantly from 10 degrees and 5 degrees respectively. FE analysis was performed for the optimized monolimb design and experimental structural test was conducted to validate the FE model. As shown in Table 5, reasonable match was demonstrated among the results measured from experimental structural test, predicted by FE model and using equation 1. Experimental structural tests showed no obvious permanent deformation after the removal of the applied load. The prosthesis is able to sustain 2717N, but demonstrates a permanent deformation of 4.1 mm after removal of the load.

The suggested monolimb is designed for giving high flexibility at push off phase. It could be preferable if the monolimb can provide some degree of flexibility at heel strike to simulate ankle plantarflexion. However, as the line of action of the ground reaction force was relatively close to the shank, there is only a trace of shank deflection at heel strike with the uses of monolimbs. If the shank of a monolimb is designed giving more flexibility at heel strike, it is likely that the monolimb will collapse at heel off phase. To compensate the loss of plantarflexion of the ankle joint, appropriate stiffness of heel cushion giving reasonable deformation at heel strike should be used.

The main objective of this study is to maximize the flexibility of the monolimb shank quantified by the distance traveled by the upper load application point, under the constraint by the peak von Mises stresses which have to be lower than 26.4MPa. In addition to these two parameters, the changes in dorsiflexion and inversion angles were specified for further constraints to the objective. It is not easy to determine the target values of dorsiflexion and inversion. Normal persons dorsiflex and invert the foot at about 10 and 5 degrees respectively at push off phase (Perry, 1992). Different prosthetic feet offer dorsiflexion angles at push off ranging from a few degrees up to 20 degrees (Wagner, 1987). Although previous studies showed that amputees favored with the prosthetic feet with higher flexibility (Beck et al., 2001; Coleman et al., 2001), no consensus has been reached on the dorsiflexion angle that a prosthesis should provide. Inversion angles were seldom mentioned in studies related to prosthetic feet. At this moment, attempt was made to prevent too much dorsiflexion (<15 degrees) and inversion (<5 degrees). Further investigations are required to look into the optimal joint angle for amputees' gait.

It should also be noted that the descriptive method of the "foot" motions used in this study was slightly different from the one used in other gait analysis. Foot block motion was described in this study by the angle changes between the top and bottom aluminium blocks. This measurement method placed emphasis on the motion due to shank deflection which was the primary interest of this study. In gait analysis, on the other hand, ankle motions are commonly measured according to the reflective markers attached to the prosthesis and the shoe. During walking, deformation of the rubber foam at the plantar region of the prosthetic foot and the motion between the shoe and the foot could occur. In addition to the movement of the foot, the motion of the foot-shoe complex and the compression of the rubber foam could both contribute to the foot motion. Further investigation into the relationship between the angles measured by the two methods will be performed.

A factor of safety was assigned to scale down the allowable working stress to account for the uncertainty in design. The factor of safety was relatively low because 1) yielding which is the earliest mode of failure was set as the failure criteria and 2) a high factor of safety would compromise with the monolimb flexibility. The structural integrity was assessed in the FE model by checking if the peak von Mises stress exceeded the yield stress. Experimental results were found resonably matching with the predictions in the FE model. However, in real situation the stress experienced by the monolimb as well as the yield stress of the material could be varied because of the imperfections in materials, flaws in assembly, material degradation and other uncertainties. A larger number of samples have to be experimentally tested to look into the variety among test samples.

To simplify the FE model in this study, viscoelastic property of the thermoplastic monolimb was not considered. The mechanical property was assumed linearly elastic and the strain rate effect on the mechanical property was not considered as the stress applied to the thermoplastic was not high (Ogorkiewicz, 1977). Hysteresis was not simulated in the FE model because attention was paid only to the final deformation state of monolimb upon load application and whether or not the monolimb can return to is original state upon removal of the test load. The force-deformation characteristic for the entire loading and unloading process was of lower interest in this study. Creeping effect is believed minimal as the measurements of dorsiflexion and inversion angles were performed immediately after the loading.

It is expected that the shank length, alignment and the magnitude of loading applied on monolimbs would be different for amputees having different characteristics such as body weight, walking style and residual limb length. Using similar techniques, monolimbs will be optimized considering characteristic differences among different amputees. In future studies, gait analysis will be performed to obtain a clearer picture how flexibility of the shank can benefit gait performance and fatigue life of the optimized monolimbs will be studied.

CONCLUSION

This paper describes methods using FE analysis and statistics-based Taguchi method to design monolimbs. One monolimb design providing high flexibility and resisting permanent deformation on normal uses was suggested. The degree of importance of the design factors which affect the structural behavior of the monolimb is suggested. The information can be used for further optimization of monolimbs to suit amputees with different characteristics.

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Table 1. Design factors and their levels of Taguchi method.

Design factor	Level			
	Level 1	Level 2	Level 3	Level 4
Thickness (mm)	4	5	6	7
AP (mm)	25	35	45	55
ML (mm)	25	35	45	55
Posterior Seam (mm)	0	5	10	15

Table 2. L_{16} orthogonal array table (the numbers under design factors indicate the levels assigned to each design factor)

Trial number	Design factors			
	Thick-ness	AP	ML	Posterior seam
1	1	1	3	4
2	1	2	4	3
3	1	3	1	2
4	1	4	2	1
5	2	1	2	3
6	2	2	1	4
7	2	3	4	1
8	2	4	3	2
9	3	1	1	1
10	3	2	2	2
11	3	3	3	3
12	3	4	4	4
13	4	1	4	2
14	4	2	3	1
15	4	3	2	4
16	4	4	1	3

Table 3. ANOVA for 4-factor, 4-level fractional factorial.

Factor	Sum of squares mm^2 (% of contribution)			
	Peak VM stress	Displacement of top load application point	Dorsiflexion angles	Inversion angles
Thickness	22.8 (24.3%)	226.4 (16.1%)	59.6 (18.5%)	12.0 (31.0%)
ML	249.7 (22.6%)	240.9 (19.1%)	60.9 (18.9%)	13.9 (35.8%)
AP	536.6 (48.5%)	860.0 (61.2%)	184.4 (57.3%)	12.5 (32.3%)
Posterior Seam	50.8 (4.6%)	78.8 (5.6%)	16.9 (5.2%)	0.3 (0.9%)

Table 4. Response and comments for some monolimb designs predicted by equation 1.

	Peak von Mises (VM) stress (MPa)	Displacement of top load application point (mm)	Dorsiflexion angles	Inversion angles	Comments
$T_1AP_1ML_1$	38.2	39.1	19.1	5.9	Dorsiflexion/inversion angle and peak VM stress too high
$T_1AP_2ML_1$	31.2	28.2	14.4	5.0	Peak VM stress too high
$T_2AP_1ML_1$	33.8	34.0	15.9	5.1	Dorsiflexion angle and peak VM stress too high
$T_2AP_2ML_2$	19.7	15.4	6.3	2.9	Flexibility appears too low
$T_3AP_1ML_1$	32.4	32.9	16.1	4.8	Dorsiflexion angle and peak VM stress too high
$T_3AP_2ML_2$	14.4	18.4	6.3	2.7	Flexibility appears low

Table 5. Response of the optimal design predicted by Taguchi method, FE analysis and measured from structural test.

	Displacement of top load application point (mm)	Dorsiflexion angles (degrees)	Inversion angles (degrees)
Predicted by Taguchi method	26.5	12.7	3.7
Confirmation run using FE analysis	27.1	13.6	2.4
Experimental test	28.5	14.4	4.4

CAPTIONS

Figure 1. Four design factors considered in the design of monolimbs.

Figure 2. (a) Geometries of monolimb, foot block, extension block and rod used for FE analysis. (b) Exploded view at the shank distal end showing the use of screw and foot bolt to tighten the shank onto the foot block.

Figure 3. Von Mises stress distribution at the monolimb

Figure 4. Mean effect of the four designs factor at each level on (a) displacement of top load application point, (b) peak von Mises stress, (c) dorsiflexion angles and (d) inversion angles of the monolimb.

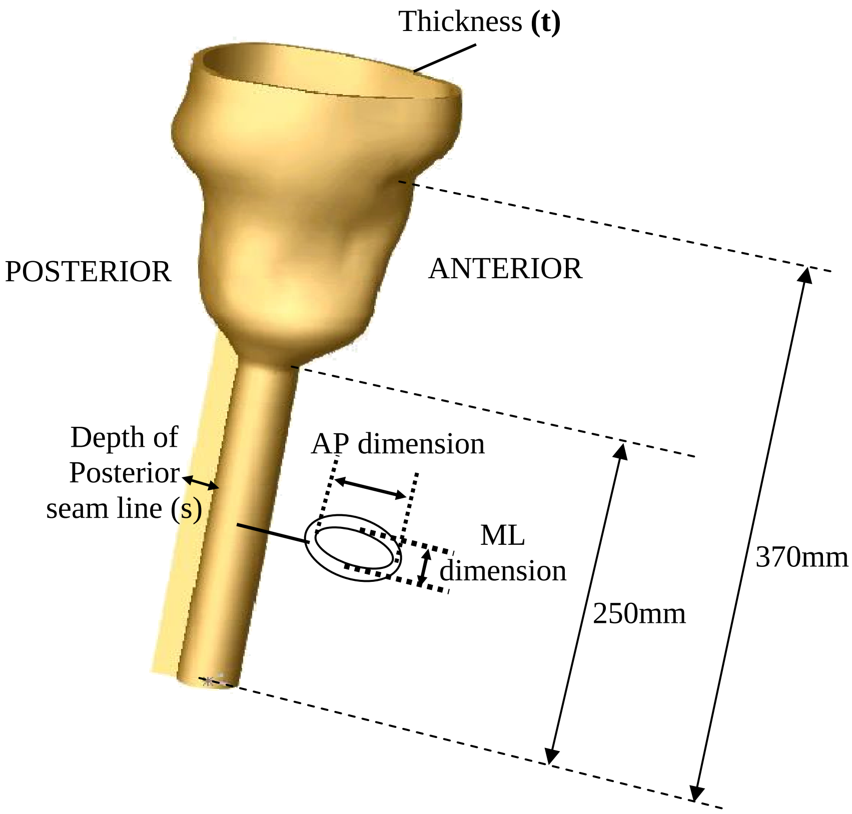

Thickness (t)
POSTERIOR
ANTERIOR
Depth of Posterior seam line (s)
AP dimension
ML dimension
250mm
370mm

Figure 1

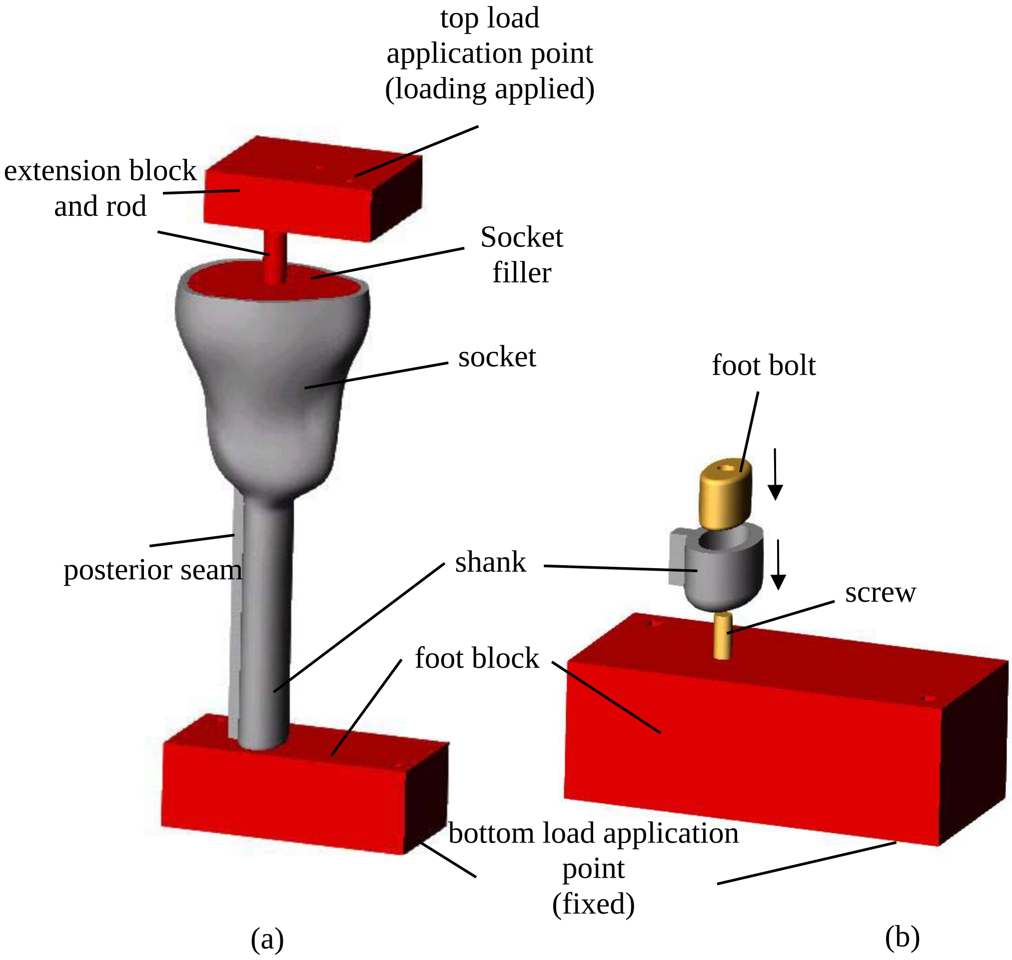
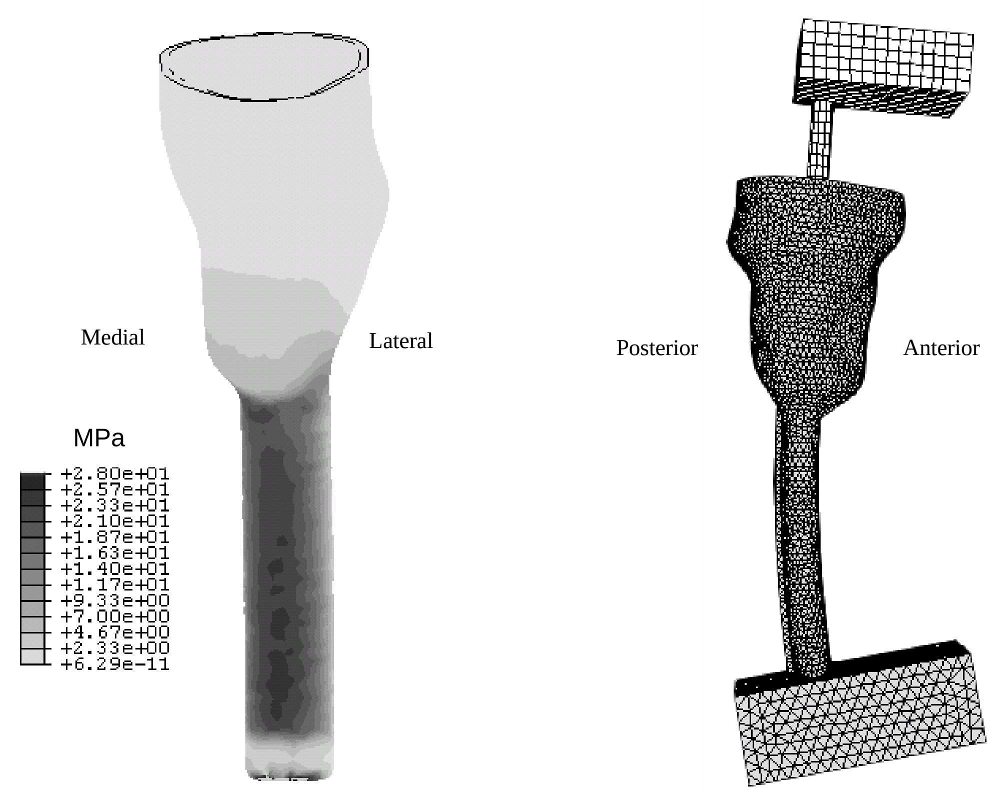

top load
application point
(loading applied)
extension block
and rod
Socket
filler
socket
foot bolt
posterior seam
shank
screw
foot block
bottom load application
point
(fixed)
(a)
(b)

Figure 2


Medial
Lateral
MPa
+2.80e+01
+2.57e+01
+2.33e+01
+2.10e+01
+1.87e+01
+1.63e+01
+1.40e+01
+1.17e+01
+9.33e+00
+7.00e+00
+4.67e+00
+2.33e+00
+6.29e-11
Posterior
Anterior

(a) **(b)**

Figure 3

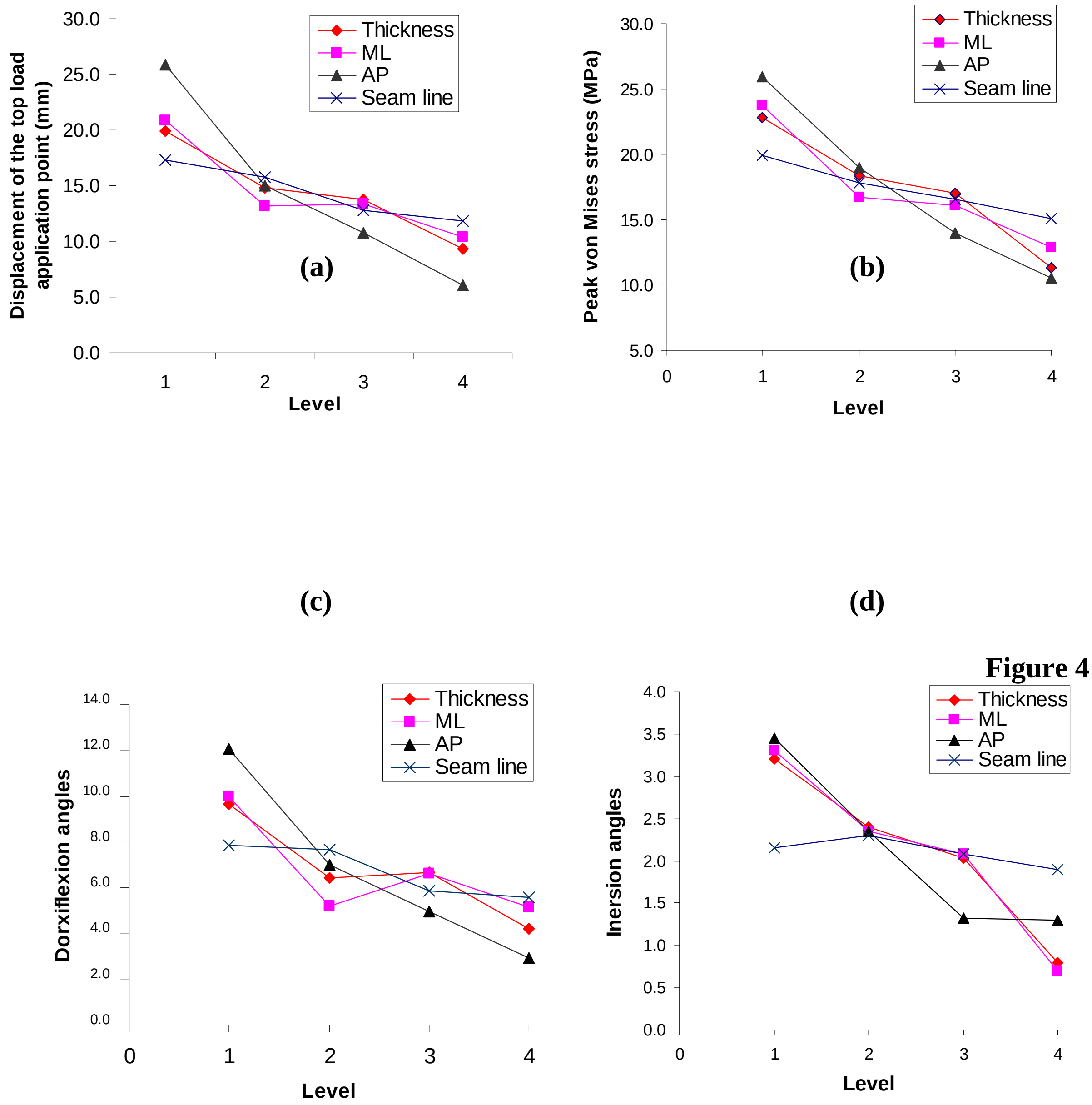

Displacement of the top load application point (mm)
30.0
25.0
20.0
15.0
10.0
5.0
0.0
1
2
3
4
Level
Thickness
ML
AP
Seam line
(a)
Peak von Mises stress (MPa)
30.0
25.0
20.0
15.0
10.0
5.0
0
1
2
3
4
Level
Thickness
ML
AP
Seam line
(b)
(c)
(d)
Dorxiflexion angles
14.0
12.0
10.0
8.0
6.0
4.0
2.0
0.0
0
1
2
3
4
Level
Thickness
ML
AP
Seam line
Inersion angles
4.0
3.5
3.0
2.5
2.0
1.5
1.0
0.5
0.0
0
1
2
3
4
Level
Thickness
ML
AP
Seam line

Figure 4