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

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1 **ABSTRACT**

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

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

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

20  
21 *Interpretations.* FE analysis and Taguchi method was shown to be an effective method in  
22 optimizing the structural design of prostheses. Further prosthetic design can be facilitated  
23 based on the degree of importance of the design factors on the structural behavior of the  
24 prosthesis. Gait analysis of amputees using the suggested monolimb design is needed in the  
25 future.

26  
27 *Keywords:* Finite element model; Taguchi method; Prosthetics; Optimization; Stress;  
28 Deformation.

## 1 INTRODUCTION

2  
3 It is common that transtibial amputees demonstrate some gait abnormalities such as lower  
4 walking speed (Molen, 1973), increased energy cost (Waters et al., 1976) and asymmetries  
5 between legs in terms of stance phase time, step length and vertical peak force (Robinson et  
6 al.,1977). Loss of active dorsiflexion and plantarflexion motions of the ankle joint of  
7 amputees is one of the reasons of the gait abnormalities (Bowker and Kazim, 1989).  
8 Prostheses have been designed to compensate for the loss of motions at the foot by  
9 incorporating energy storing and releasing (ESAR) capabilities using flexible keels or shanks.  
10 The Seattle foot<sup>TM</sup> (Seattle System, Poulsbo, WA 98370 USA) and FlexFoot<sup>TM</sup> are examples  
11 of ESAR prosthetic components. Previous research suggested that many amputees  
12 subjectively prefer ESAR prosthetic feet to conventional solid ankle cushioned heel (SACH)  
13 feet on normal and fast walking (Macfarlane et al., 1991; Menard and Murray, 1989).  
14 However, many amputees still utilize the simple conventional SACH designs because of their  
15 lower cost.

16  
17 A “monolimb” prosthesis using a conventional prosthetic foot perhaps is an alternative to  
18 ESAR prosthetic feet if properly designed, providing elastic response of the shank (Valenti,  
19 1991), at the same time lower the total prosthetic weight and cost. Monolimb refers to a  
20 transtibial prosthesis having the socket and the shank molded into one piece of thermoplastics.  
21 Different names have been used for this kind of prosthesis such as endoflex (Valenti, 1991),  
22 total thermoplastic prosthesis (Rothschild, 1991) and ultra-light prosthesis (Reed et al., 1979).  
23 Due to the flexibility of thermoplastics, the shank of a monolimb can deflect during walking.  
24 By optimizing the uses of material and structural design, it is possible that the shank  
25 deflection be altered such that the natural ankle joint motions are mimicked so as to enhance  
26 the comfort and gait performance. Positive feedbacks were gained including improved gait  
27 efficiency and comfort from patients using prostheses with deformable shanks (Beck et al.,  
28 2001; Coleman et al., 2001). It has also been shown that increased shank flexibility can have  
29 potential in reducing the prosthetic socket- residual limb interface stresses (Lee et al., 2004).  
30 At the same time, it should be noted that the structural integrity has to be maintained without  
31 permanent deformation of the prosthesis. Until now there is no clear guideline for the shank  
32 design of monolimb.

33  
34 Currently, the structural test specifications of lower limb prostheses are specified in  
35 ISO10328. To optimize the design of the shank, monolimbs with different shank designs  
36 have to be subjected to tests according to the ISO standards. Performing such test  
37 experimentally is expensive and time demanding. Computational analyses such as finite  
38 element (FE) modeling, allow parametric study to be performed easily without the need to  
39 fabricate prostheses. FE analysis has been widely used in lower limb prosthetics in the past  
40 decade. In previous FE models, focus was on investigating the interface contact between the  
41 prosthetic socket and residual limb as reviewed by Zhang et al, 1998. The design of a well-  
42 controlled deformability of a prosthesis, however, received little attention.

43  
44 There are a number of methods to find an optimum configuration of a monolimb. “Vary-one-  
45 factor-at-a-time” is one popularly used design optimization method in which the effect of one  
46 factor is assessed by varying only the factor to be assessed and keeping the other factors fixed  
47 at a specific set of conditions. However, this method can sometimes lead to wrong results as  
48 the effect of the factor can be changed if other factors are substituted with different conditions  
49 (Phadke, 1989). If full factorial is run exploring every possible combination of values of each

1 factor, the total number of simulations required will be very high. A statistical approach  
2 developed by Taguchi (Margolis, 1985) utilizes an orthogonal array, which is a form of  
3 fractional factorial design containing a well-chosen subset of all possible combination of test  
4 conditions. Using Taguchi method, a balanced comparison of levels of any factor and  
5 significant reduction in the total number of required simulations can both be achieved.

6  
7 This paper demonstrates a technique using computational modeling and statistical-based  
8 method in optimizing the design of monolimb. FE analysis was used to predict the  
9 deformation and stress at monolimbs of different material thickness and shank geometry  
10 subjected to loadings based on ISO10328. Taguchi method was used to identify the  
11 importance of each design factor and suggest an optimized monolimb design which can resist  
12 failure on normal uses and can provide appropriate flexibility.

## 13 **METHODS**

### 14 **A. Finite element modeling**

15  
16 A plaster cast was taken on a male left-sided transtibial amputee. The cast was digitized using  
17 BioSculptor™ system and exported to prosthetic CAD software ShapeMaker™ 4.3. A  
18 prosthetist using ShapeMaker™ prepared the geometry of the prosthesis by applying built-in  
19 shape rectification template to the digitized limb and aligning a shank blended into the socket.  
20 Different designs of the prosthesis, as shown in Table 1 and 2, and Figure 1, were created.  
21 Using commercial CAD software SolidWorks™ 2001, the socket together with the shank was  
22 given a specified thickness. Foot block, socket filler, extension rod and block were added  
23 (Figure 2a), so that the load application points and direction of force can be applied as  
24 instructed in ISO10328. The socket filler only extended from the proximal brim of the socket  
25 to approximately 10cm above the distal end of the socket to allow the distal part of the socket  
26 to deform upon load application. A foot bolt and a screw (modeled as a cylinder) were  
27 inserted (Figure 2b). To simulate the screw and foot bolt fixing the monolimb onto the foot  
28 block, the screw was rigidly tied with the foot block as well as the foot bolt and contact was  
29 defined among the foot bolt, distal end of the shank and foot block so that their surfaces were  
30 not allowed to penetrate each other when loading was added. The model in its entirety, as  
31 shown in Figure 3b, was exported to ABAQUS version 6.4 (Hibbitt, Karlsson & Sorensen,  
32 Inc., Pawtucket, RI, USA). A FE mesh of 3D tetrahedral elements was built using ABAQUS  
33 auto-meshing techniques. Tetrahedral element was chosen because of geometrically irregular  
34 structures of the monolimb. The number of elements ranged from 27,337 to 39,029  
35 depending on the shape of the shank.

36  
37  
38 The Young's modulus and Poisson's ratio of the monolimb made of polypropylene  
39 homopolymer were 1500MPa and 0.3 respectively (Margolis, 1985). Foot block, adaptor,  
40 screw, extension rod and block were assumed to be rigid. The bottom load application point  
41 was fixed and loadings were added at the top load application point (Figure 2a). The loadings  
42 applied were based on ISO10328. The standard specifies loads for testing prosthesis at  
43 normal walking load and occasionally severe load during heel strike (loading condition I) and  
44 heel off (loading condition II) of the gait. We observed in our previous structural test  
45 experiment and FE analysis (Lee et al., 2004) that the loading condition II caused much more  
46 deformation and higher stresses to monolimbs than loading condition I because of the longer  
47 moment arm. Fracture failure of monolimb was unlikely under the loading condition specified  
48 in ISO10328, due to the high ductility of thermoplastic material. However, permanent  
49 deformation could occur in some monolimb designs which is undesirable as it permanently  
50 changes the alignment of prosthetic foot relative to the socket. Based on the above

1 information, force specified in ISO10328 simulating heel off at normal walking load was used  
 2 to load the prosthesis in the FE model during the design stage. The selected test load level  
 3 was A80 (1085N). There are three test load levels specified in ISO10328 which accounts for  
 4 the different amputee body weights. A80 is for amputees whose weights are between 60kg to  
 5 80kg. Geometric nonlinearity resulted from the large deformation was considered in the  
 6 model. Peak von Mises stress, displacement of the top load application point, dorsiflexion  
 7 and inversion angles defined as the angle changes between the top and bottom aluminium  
 8 blocks in sagittal and frontal planes respectively were predicted in the FE model. Through  
 9 testing different designs, the aim was to design a prosthesis providing high flexibility but  
 10 without permanent deformation under normal walking.

## 11 **B. Taguchi method and design optimization**

12 Four design factors namely, the thickness (T) of the thermoplastic material, depth of posterior  
 13 seam line (S), anteroposterior (AP) and medial-lateral (ML) dimensions of the shank (Figure  
 14 1) were selected for evaluation. Each factor was assigned with 4 levels (Table 1). The  
 15 thickness of the polypropylene material was in the range of 4-7mm. The AP/ ML dimensions  
 16 and depth of the seam were in the range of 25-55mm and 0-15mm respectively concerning the  
 17 bulkiness and the structural integrity of the monolimb. To identify the relative significance of  
 18 the design factors using full factorial approach, a total number of total number 256 analyses  
 19 ( $4^4$ ) are required. In this study, a statistics-based-Taguchi method was used to reduce the  
 20 number of analyses. Sixteen simulations were required. The configurations of the  
 21 monolimbs were shown in an orthogonal array  $L_{16}$  (Phadke, 1989) in Table 2. The  
 22 mechanical responses namely, von Mises stress, displacement of the top load application  
 23 point, dorsiflexion and inversion angles were predicted by 16 FE analyses. The mean effect  
 24 of each level of the four design factors on the mechanical responses was computed. For  
 25 example, the mean response of thickness at level 1 [ $R(T_1)$ ] on peak von Mises stress is  
 26 calculated as the mean stress over trial 1 to trial 4. An analysis of variance (ANOVA) was  
 27 performed calculating the sum of squares of each design factor to determine sensitivity of  
 28 each design parameter. For example, the sum of squares due to thickness would be equal to  
 29  $4[R(T_1)-R_m]^2+4[R(T_2)-R_m]^2+4[R(T_3)-R_m]^2+4[R(T_4)-R_m]^2$  where  $R(T_1)$ ,  $R(T_2)$ ,  $R(T_3)$  and  
 30  $R(T_4)$  were mean response of thickness at level 1 to 4 respectively and  $R_m$  was overall mean  
 31 response over 16 trials.

32  
 33  
 34 Using a superposition model (Phadke, 1989), the mechanical response of any combination of  
 35 levels among design factors can be predicted as following:

$$36 R(T_i, AP_j, ML_k, S_l) = R_m + [R(T_i) - R_m] + [R(AP_j) - R_m] + [R(ML_k) - R_m] + [R(S_l) - R_m] \quad (1)$$

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

43  
 44 Using the results of the superposition model and the sensitivity analysis of each factor, levels  
 45 of each factor were manually adjusted such that the prosthesis can provide high flexibilities  
 46 without inducing permanent deformation. Using the strength theory, permanent deformation  
 47 will occur if the peak von Mises stress was equal to or larger than the allowable strength of  
 48 26.4 MPa which is equal to the yield strength of the material divided by the factor of safety.  
 49 The yield strength was 33MPa for polypropylene homopolymer (Margolis, 1985) and the  
 50 factor of safety was 1.25. Flexibility of the prosthesis was quantified by the computed

1 vertical displacement of the top load application point. Attempt was made to maximize the  
2 displacement, at the same time, the deformation of the monolimb should be less than 15  
3 degrees for dorsiflexion angle and 5 degrees for inversion angle.

### 4 5 **C. Experimental test**

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

19  
20 The monolimb was loaded to 1085N at the loading rate controlled at 100N/s. A mechanical  
21 digitizer was used to record the spatial coordinates of the aluminium blocks for the calculation  
22 of the “dorsiflexion” and “inversion” angles. Measurement was made immediately after  
23 1085N was applied to reduce the creeping effect. The displacement of the top load  
24 application point was recorded real-time during the load-unloading process. Permanent  
25 deformation was inspected by studying if the actuator could return to original position upon  
26 removal of the load. After load-unloading at 1085N load level, the monolimb was loaded to  
27 2717N which is the occasional severe load specified in ISO10328 A80 load level until the  
28 monolimb failed or sustained the test load.

## 29 30 **RESULTS AND DISCUSSION**

31  
32 Stress and deformation of the monolimbs under loading were evaluated by finite element  
33 analysis which is believed providing a more accurate solution than the analytical method  
34 using beam theory when dealing with complicated geometry and boundary condition. Figure  
35 3 shows one typical von Mises stress distribution and the deformation of the monolimb.  
36 High von Mises stresses fall over the anterior region of the shank. The stress value is low at  
37 the region of the foot bolt. This is because of the stress shielding effect with the foot bolt  
38 having much higher stiffness than the thermoplastic material reducing the bending  
39 displacement of the thermoplastic material around the rigid bolt. The shank bows towards the  
40 back upon load adding. The responses, including peak von Mises stresses, displacement of  
41 the top load application point, dorsiflexion and inversion angles of the 16 different monolimb  
42 configurations were evaluated by the FE models. Inversion angles were found relatively less  
43 sensitive compared with the other three responses when the design factors changed and almost  
44 all monolimbs deformed with inversion angles less than 5 degrees.

45  
46 The mean effect of the design factors at each level can be found in Figure 4. As expected,  
47 there is a trend showing the reduction in peak von Mises stress, displacement of the top load  
48 application point, dorsiflexion and inversion angles, when the four design factors thickness,  
49 cross sectional area and depth of posterior seam line increase. The reduction trend is not  
50 obvious when the medialateral is moved from level 2 to level 3. This indicates that the

1 responses are not sensitive to medialateral between level 2 to level 3. Their degrees of  
2 importance are different as the slopes of the curves are different. The level of importance of  
3 each factor in the structural behavior of the prosthesis can be studied by comparing the sum of  
4 squares shown in Table 4. Among the four analyzed design factors controlled at specified  
5 levels, anteroposterior dimension of the shank was shown to be the most important design  
6 factor determining the peak von Mises stress values, displacement of the top load application  
7 point and dorsiflexion angles of monolimbs. As far as the inversion angle of the foot block is  
8 concerned, the thickness, anteroposterior and medialateral dimensions of the shank were  
9 almost equally important. Depth of seam line appears much less important than the other  
10 three factors.

11  
12 The responses of monolimbs of any combinations of levels among factors can be predicted  
13 using equation 1. The displacement of the top load application point would be the highest if  
14 the four factors were assigned at level 1. However, the design is deemed inappropriate as the  
15 peak von Mises stresses would be much higher than 26.4 MPa and dorsiflexion angles would  
16 be much higher than 10 degrees (Table 5). The levels of the factors were adjusted according  
17 to their degree of importance and the predicted responses using equation 1. At this stage, the  
18 depth of the seam line was fixed at level 2 as it was found less important than the other three  
19 factors. Table 5 shows the predicted responses of some selected monolimb designs which  
20 were not evaluated in the 16 trials. The six monolimbs listed in Table 5 do not meet the  
21 design requirement and require further tuning on the levels of the factors

22  
23 Some design configurations fulfill the design requirements. One optimized design  
24 configuration would be thickness at 5mm, anteroposterior dimension at 25mm, medialateral  
25 dimension at 45mm, giving peak von Mises stress 26.1 MPa and displacement of top load  
26 application point 26.5 mm predicted by equation 1 when loading simulating heel off was  
27 applied. The dorsiflexion and inversion are not deviated significantly from 10 degrees and 5  
28 degrees respectively. FE analysis was performed for the optimized monolimb design and  
29 experimental structural test was conducted to validate the FE model. As shown in Table 5,  
30 reasonable match was demonstrated among the results measured from experimental structural  
31 test, predicted by FE model and using equation 1. Experimental structural tests showed no  
32 obvious permanent deformation after the removal of the applied load. The prosthesis is able to  
33 sustain 2717N, but demonstrates a permanent deformation of 4.1 mm after removal of the  
34 load.

35  
36 The suggested monolimb is designed for giving high flexibility at push off phase. It could be  
37 preferable if the monolimb can provide some degree of flexibility at heel strike to simulate  
38 ankle plantarflexion. However, as the line of action of the ground reaction force was  
39 relatively close to the shank, there is only a trace of shank deflection at heel strike with the  
40 uses of monolimbs. If the shank of a monolimb is designed giving more flexibility at heel  
41 strike, it is likely that the monolimb will collapse at heel off phase. To compensate the loss of  
42 plantarflexion of the ankle joint, appropriate stiffness of heel cushion giving reasonable  
43 deformation at heel strike should be used.

44  
45 The main objective of this study is to maximize the flexibility of the monolimb shank  
46 quantified by the distance traveled by the upper load application point, under the constraint by  
47 the peak von Mises stresses which have to be lower than 26.4MPa. In addition to these two  
48 parameters, the changes in dorsiflexion and inversion angles were specified for further  
49 constraints to the objective. It is not easy to determine the target values of dorsiflexion and  
50 inversion. Normal persons dorsiflex and invert the foot at about 10 and 5 degrees respectively



1 at push off phase (Perry, 1992). Different prosthetic feet offer dorsiflexion angles at push off  
2 ranging from a few degrees up to 20 degrees (Wagner, 1987). Although previous studies  
3 showed that amputees favored with the prosthetic feet with higher flexibility (Beck et al.,  
4 2001; Coleman et al., 2001), no consensus has been reached on the dorsiflexion angle that a  
5 prosthesis should provide. Inversion angles were seldom mentioned in studies related to  
6 prosthetic feet. At this moment, attempt was made to prevent too much dorsiflexion (<15  
7 degrees) and inversion (<5 degrees). Further investigations are required to look into the  
8 optimal joint angle for amputees' gait.

9  
10 It should also be noted that the descriptive method of the "foot" motions used in this study  
11 was slightly different from the one used in other gait analysis. Foot block motion was  
12 described in this study by the angle changes between the top and bottom aluminium blocks.  
13 This measurement method placed emphasis on the motion due to shank deflection which was  
14 the primary interest of this study. In gait analysis, on the other hand, ankle motions are  
15 commonly measured according to the reflective markers attached to the prosthesis and the  
16 shoe. During walking, deformation of the rubber foam at the plantar region of the prosthetic  
17 foot and the motion between the shoe and the foot could occur. In addition to the movement  
18 of the foot, the motion of the foot-shoe complex and the compression of the rubber foam  
19 could both contribute to the foot motion. Further investigation into the relationship between  
20 the angles measured by the two methods will be performed.

21  
22 A factor of safety was assigned to scale down the allowable working stress to account for the  
23 uncertainty in design. The factor of safety was relatively low because 1) yielding which is the  
24 earliest mode of failure was set as the failure criteria and 2) a high factor of safety would  
25 compromise with the monolimb flexibility. The structural integrity was assessed in the FE  
26 model by checking if the peak von Mises stress exceeded the yield stress. Experimental  
27 results were found reasonably matching with the predictions in the FE model. However, in real  
28 situation the stress experienced by the monolimb as well as the yield stress of the material  
29 could be varied because of the imperfections in materials, flaws in assembly, material  
30 degradation and other uncertainties. A larger number of samples have to be experimentally  
31 tested to look into the variety among test samples.

32  
33 To simplify the FE model in this study, viscoelastic property of the thermoplastic monolimb  
34 was not considered. The mechanical property was assumed linearly elastic and the strain rate  
35 effect on the mechanical property was not considered as the stress applied to the thermoplastic  
36 was not high (Ogorkiewicz, 1977). Hysteresis was not simulated in the FE model because  
37 attention was paid only to the final deformation state of monolimb upon load application and  
38 whether or not the monolimb can return to its original state upon removal of the test load. The  
39 force-deformation characteristic for the entire loading and unloading process was of lower  
40 interest in this study. Creeping effect is believed minimal as the measurements of dorsiflexion  
41 and inversion angles were performed immediately after the loading.

42  
43 It is expected that the shank length, alignment and the magnitude of loading applied on  
44 monolimbs would be different for amputees having different characteristics such as body  
45 weight, walking style and residual limb length. Using similar techniques, monolimbs will be  
46 optimized considering characteristic differences among different amputees. In future studies,  
47 gait analysis will be performed to obtain a clearer picture how flexibility of the shank can  
48 benefit gait performance and fatigue life of the optimized monolimbs will be studied.

49  
50 **CONCLUSION**

1  
2 This paper describes methods using FE analysis and statistics-based Taguchi method to  
3 design monolimbs. One monolimb design providing high flexibility and resisting permanent  
4 deformation on normal uses was suggested. The degree of importance of the design factors  
5 which affect the structural behavior of the monolimb is suggested. The information can be  
6 used for further optimization of monolimbs to suit amputees with different characteristics.  
7

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13

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1 Table 1. Design factors and their levels of Taguchi method.  
 2  
 3

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

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 11 Table 2.  $L_{16}$   
 12 table (the numbers  
 13 indicate the levels  
 14 design factor)

orthogonal array  
 under design factors  
 assigned to each

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 <sup>2</sup> (% 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 <sub>1</sub> AP <sub>1</sub> ML <sub>1</sub>	38.2	39.1	19.1	5.9	Dorsiflexion/inversion angle and peak VM stress too high
T <sub>1</sub> AP <sub>2</sub> ML <sub>1</sub>	31.2	28.2	14.4	5.0	Peak VM stress too high
T <sub>2</sub> AP <sub>1</sub> ML <sub>1</sub>	33.8	34.0	15.9	5.1	Dorsiflexion angle and peak VM stress too high
T <sub>2</sub> AP <sub>2</sub> ML <sub>2</sub>	19.7	15.4	6.3	2.9	Flexibility appears too low
T <sub>3</sub> AP <sub>1</sub> ML <sub>1</sub>	32.4	32.9	16.1	4.8	Dorsiflexion angle and peak VM stress too high
T <sub>3</sub> AP <sub>2</sub> ML <sub>2</sub>	14.4	18.4	6.3	2.7	Flexibility appears low

1 Table 5. Response of the optimal design predicted by Taguchi method, FE analysis and  
 2 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

1 **CAPTIONS**

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

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

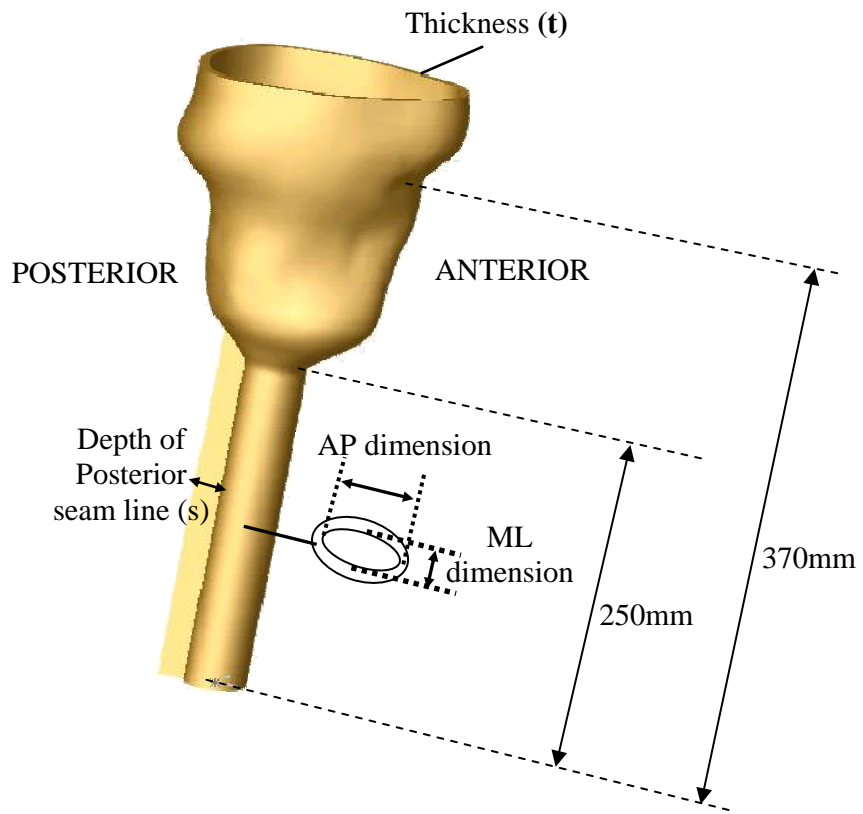
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9 Figure 3. Von Mises stress distribution at the monolimb

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

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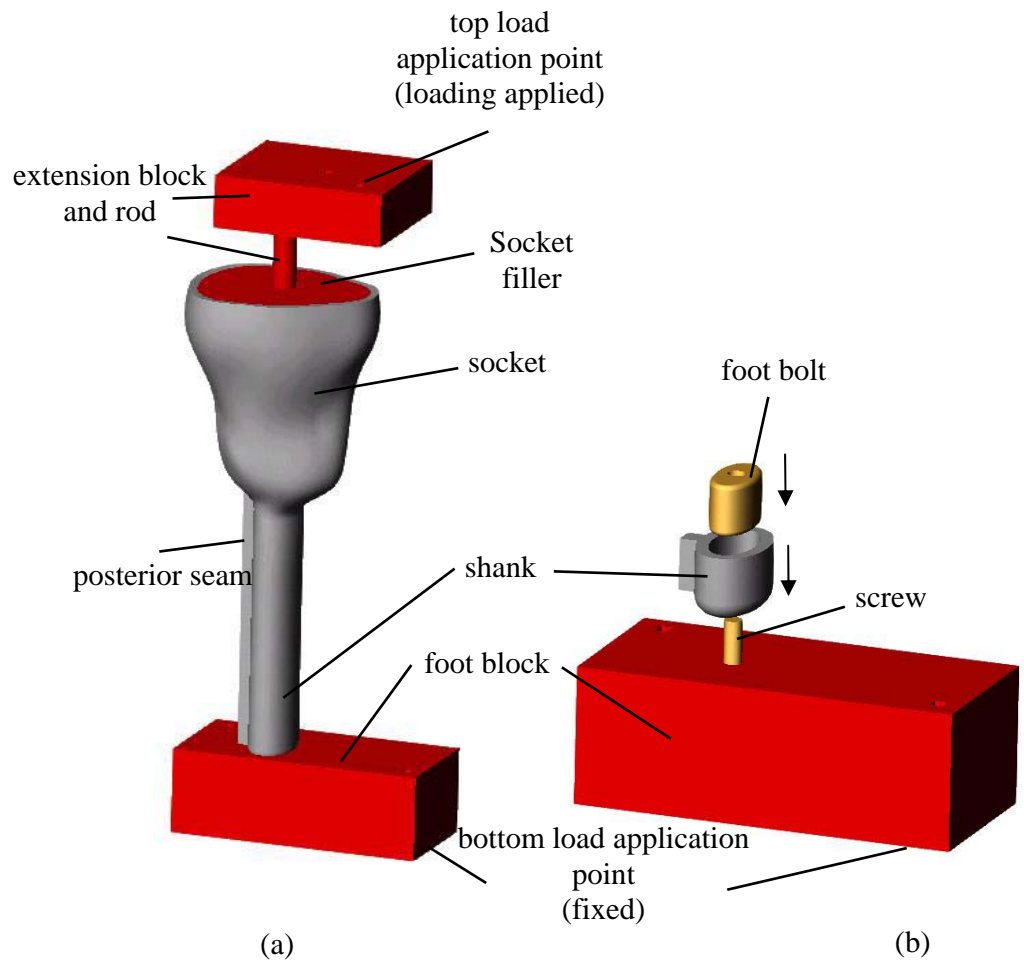


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**Figure 1**



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Figure 2

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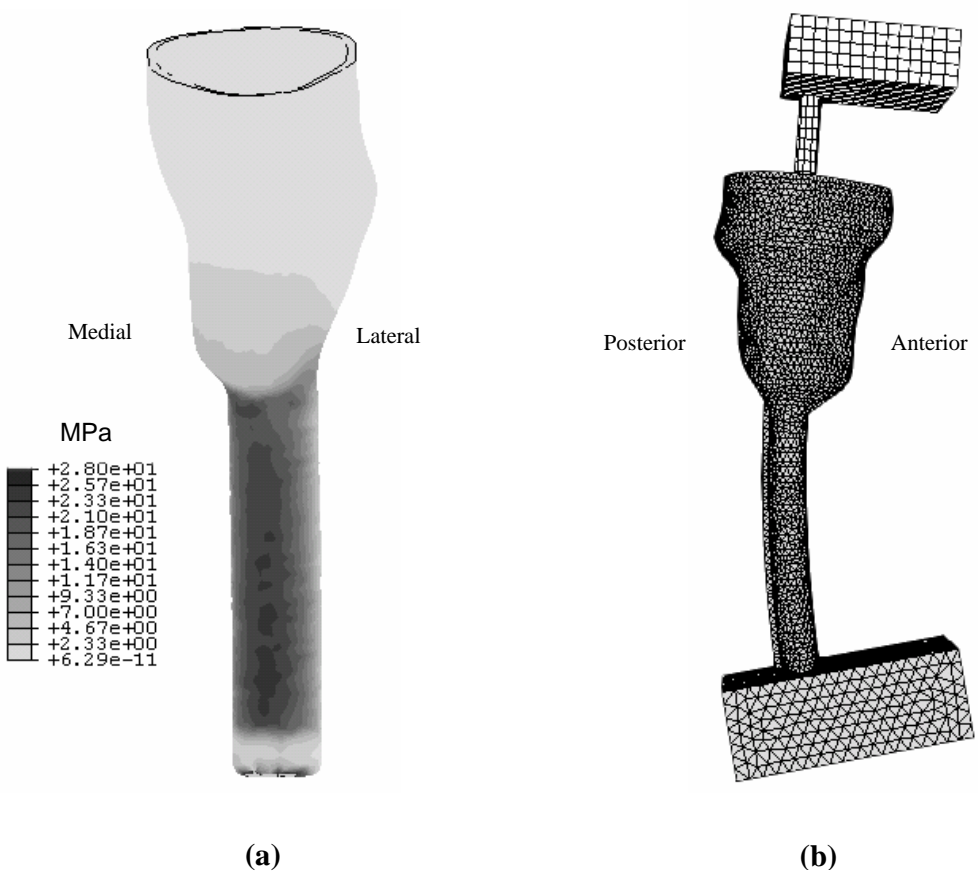
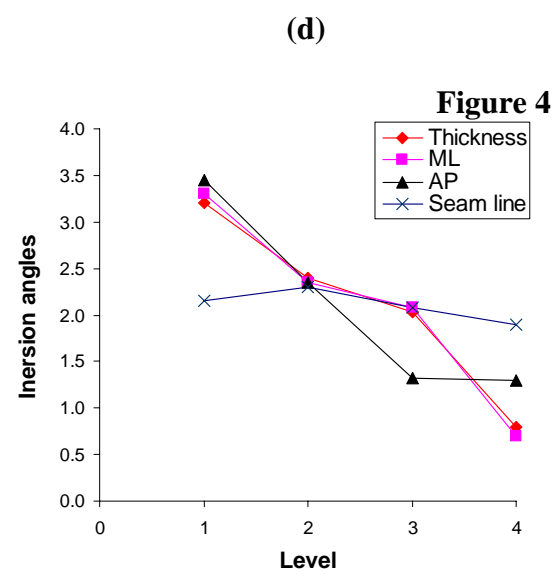
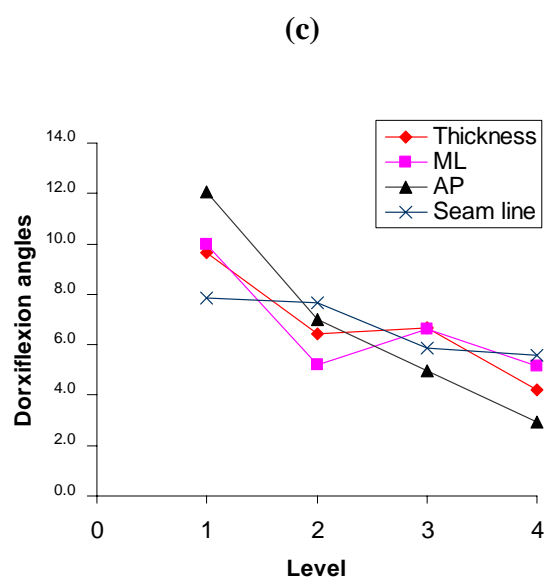
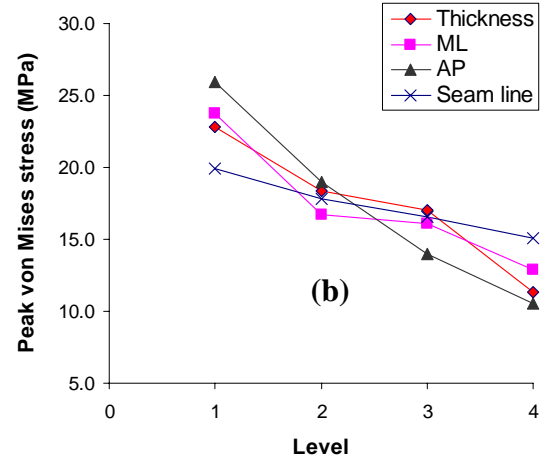
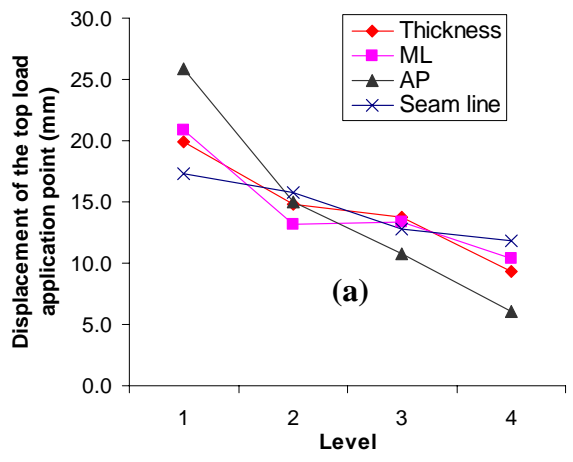


Figure 3

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**Figure 4**