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 mediolateral 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 finite element 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 finite element results. Interpretation. Finite element 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
element, design, method, monolimb, finite, statistics-based, taguchi, modelling

Disciplines
Engineering | Science and Technology Studies

Publication Details

This journal article is available at Research Online: http://ro.uow.edu.au/eispapers/6681
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 mediolateral 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™ (Seattle System, Poulsbo, WA 98370 USA) and FlexFoot™ 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(T1)] 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(T1)-R_m]^2+4[R(T2)-R_m]^2+4[R(T3)-R_m]^2+4[R(T4)-R_m]^2 where R(T1), R(T2), R(T3) and
R(T4) 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:

\[ R(T_i, AP_j, ML_k, S_l) = R_m + [R(T_i) - R_m] + [R(AP_j) - R_m] + [R(ML_k) - R_m] + [R(S_l) - R_m] \] (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™
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 monolims 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 monolims 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 monolims. 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 monolims 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 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 monolims 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 monolims. 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
deflection 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 its 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.

ACKNOWLEDGEMENT

The work described in this paper was supported by The Hong Kong Polytechnic University Research Studentship and a grant from the Research Grant Council of Hong Kong (Project No. PolyU 5200/02E).

REFERENCES


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

<table>
<thead>
<tr>
<th>Design factor</th>
<th>Level 1</th>
<th>Level 2</th>
<th>Level 3</th>
<th>Level 4</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thickness (mm)</td>
<td>4</td>
<td>5</td>
<td>6</td>
<td>7</td>
</tr>
<tr>
<td>AP (mm)</td>
<td>25</td>
<td>35</td>
<td>45</td>
<td>55</td>
</tr>
<tr>
<td>ML (mm)</td>
<td>25</td>
<td>35</td>
<td>45</td>
<td>55</td>
</tr>
<tr>
<td>Posterior Seam (mm)</td>
<td>0</td>
<td>5</td>
<td>10</td>
<td>15</td>
</tr>
</tbody>
</table>

Table 2. L$_{16}$ orthogonal array table (the numbers indicate the levels assigned to each design factor).

<table>
<thead>
<tr>
<th>Trial number</th>
<th>Design factors</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Thickness</td>
</tr>
<tr>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td>2</td>
<td>1</td>
</tr>
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<td>3</td>
<td>1</td>
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<td>15</td>
<td>4</td>
</tr>
<tr>
<td>16</td>
<td>4</td>
</tr>
</tbody>
</table>
Table 3. ANOVA for 4-factor, 4-level fractional factorial.

<table>
<thead>
<tr>
<th>Factor</th>
<th>Sum of squares mm² (% of contribution)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peak VM stress</td>
<td>Displacement of top load application point</td>
</tr>
<tr>
<td>Thickness</td>
<td>22.8 (24.3%)</td>
</tr>
<tr>
<td>ML</td>
<td>249.7 (22.6%)</td>
</tr>
<tr>
<td>AP</td>
<td>536.6 (48.5%)</td>
</tr>
<tr>
<td>Posterior Seam</td>
<td>50.8 (4.6%)</td>
</tr>
</tbody>
</table>

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

<table>
<thead>
<tr>
<th>Peak von Mises (VM) stress (MPa)</th>
<th>Displacement of top load application point (mm)</th>
<th>Dorsiflexion angles</th>
<th>Inversion angles</th>
<th>Comments</th>
</tr>
</thead>
<tbody>
<tr>
<td>T₁AP₁ML₁</td>
<td>38.2</td>
<td>39.1</td>
<td>19.1</td>
<td>5.9</td>
</tr>
<tr>
<td>T₁AP₂ML₁</td>
<td>31.2</td>
<td>28.2</td>
<td>14.4</td>
<td>5.0</td>
</tr>
<tr>
<td>T₂AP₁ML₁</td>
<td>33.8</td>
<td>34.0</td>
<td>15.9</td>
<td>5.1</td>
</tr>
<tr>
<td>T₂AP₂ML₂</td>
<td>19.7</td>
<td>15.4</td>
<td>6.3</td>
<td>2.9</td>
</tr>
<tr>
<td>T₃AP₁ML₁</td>
<td>32.4</td>
<td>32.9</td>
<td>16.1</td>
<td>4.8</td>
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<td>T₃AP₂ML₂</td>
<td>14.4</td>
<td>18.4</td>
<td>6.3</td>
<td>2.7</td>
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</table>
Table 5. Response of the optimal design predicted by Taguchi method, FE analysis and measured from structural test.

<table>
<thead>
<tr>
<th></th>
<th>Displacement of top load application point (mm)</th>
<th>Dorsiflexion angles (degrees)</th>
<th>Inversion angles (degrees)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Predicted by Taguchi method</td>
<td>26.5</td>
<td>12.7</td>
<td>3.7</td>
</tr>
<tr>
<td>Confirmation run using FE analysis</td>
<td>27.1</td>
<td>13.6</td>
<td>2.4</td>
</tr>
<tr>
<td>Experimental test</td>
<td>28.5</td>
<td>14.4</td>
<td>4.4</td>
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CAPTIONS

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

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.
Figure 1

Depth of Posterior seam line (s)

Thickness (t)

AP dimension

ML dimension

370mm

250mm
Figure 2

- **Figure 2a**: Top load application point (loading applied)
  - Extension block and rod
  - Socket filler
  - Socket
  - Posterior seam screw

- **Figure 2b**: Bottom load application point (fixed)
  - Shank
  - Foot block
  - Foot bolt
  - Screw
Figure 3
Figure 4

(a) Displacement of the top load application point (mm)

(b) Peak von Mises stress (MPa)

(c) Dorflextion angles

(d) Inersion angles