

**REHAB** *T e c h*

Monash Rehabilitation Technology Research Unit



**Socket  
Stress  
Distribution**

**Ross Stewart**



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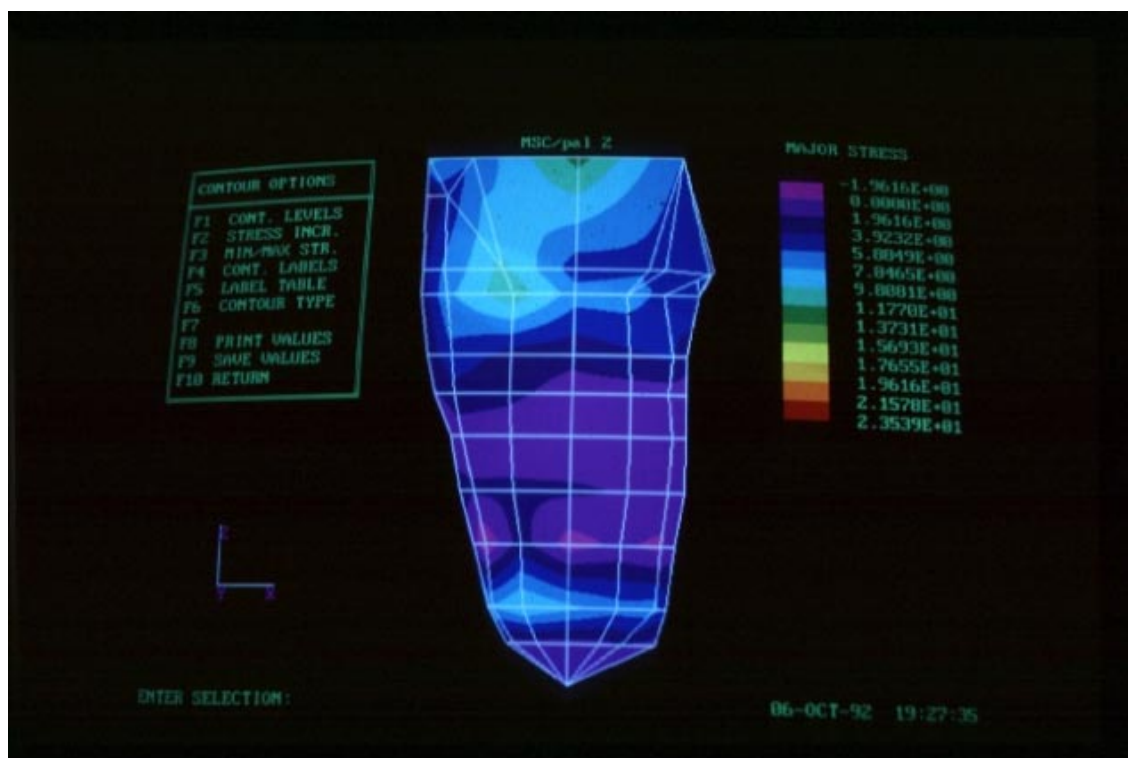
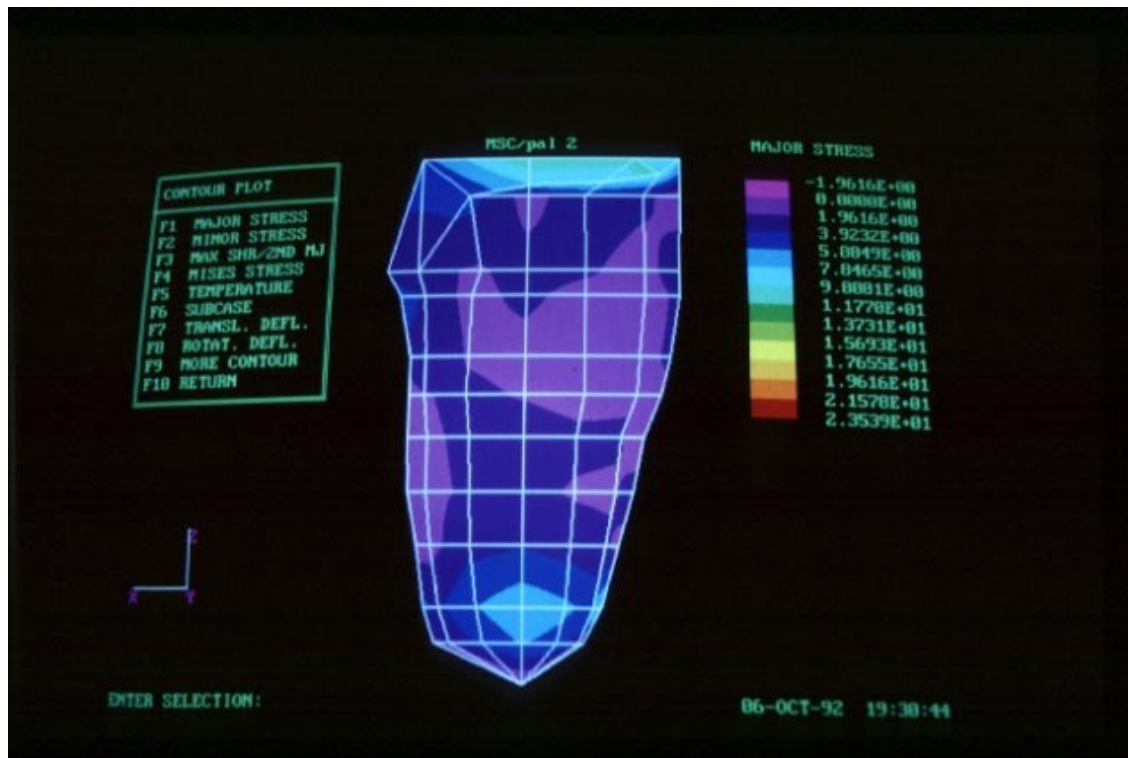
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**PROSTHETIC DEVELOPMENT.**

**ROSS STEWART**

**1991**

**AEROSPACE ENGINEERING.**  
**RMIT**



DISPLAY (UNDEFOR)  
F1 ROTATE  
F2 TRANSLATE  
F3 SCALE  
F4 OPTIONS  
F5 CENTER  
F6 ANIMATION  
F7  
F8  
F9 ACTIVE SET  
F10 RETURN

MBC/pal 2



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## SUMMARY

The aim of this project was to determine the load paths of a prosthetic socket using finite element analysis. This allows for the socket liner and frame geometries to be more accurately chosen. The socket analyzed was a medium length upper left leg socket.

The external forces were found for the leg. While the characteristics for socket loading were assumed. These enabled the socket stress distribution to be found. Allowing differences in socket and socket frame types to be observed.

Fibreglass and polyethylene sockets, double socket frames and four types of fibreglass frames were analysed. The effect of both flexion and extension to alignment was assessed. Finally the materials between the loading and the socket were varied.

### **Acknowledgments**

I like to acknowledge the generous help from Bill Contoyannis together with the Central Development Unit, from the Department of Veterans' Affairs, without which this project would not of have been possible.

## INTRODUCTION.

The artificial leg resembles the normal leg in many ways. It consists of a foot-ankle unit which needs to be attached to the remainder of the amputee's natural leg or stump. For a below knee amputation, the foot-ankle unit is attached directly to the socket frame. The socket frame gives structural support to the socket which is the interface between the stump and the artificial leg. An artificial shank can be attached to the foot ankle unit if the stump fails to extend past the knee. The shank is then attached to a knee unit which in turn, is attached to the socket frame for an above knee amputation.

Through time artificial limbs have been made with various materials. From wood and aluminium of yesterday, to fibreglass and polypropylene of today. Traditionally, the sockets were carved from wood, but with the lighter and stronger materials in use today an appreciation of what happens at the socket-stump interface is required. By finding the load path optimisation of the socket and socket frame can occur. This will result in a more comfortable and durable fitting for the patient.

Appoldt and Bennett, 1967, found the loading on an above knee fibreglass socket by building the socket with the pressure transducers incorporated. Unfortunately their results are only accurate for the single socket used in the experiment. This is due to all modern sockets having different geometries and external loadings due to differences in the amputees.

Similarly Bielefeldt and Schreck, 1979, found the difference in loading of four different sockets, during stance, for the one patient. Their sockets were also built with transducers incorporated. From both of these results a general feel for the load paths can be gauged.

Modern sockets are made with soft liners with stronger frames. One combination is the polyethylene socket. Here polyethylene is used for the inner shell with polypropylene used for the frame. Fibreglass is also used for socket frames in conjunction with a soft liner. This is called a fibreglass socket.

By the use of finite element modelling a selection of sockets and socket frames can be compared quickly.

By taking the subjects physical geometry and weight, the ground reaction forces during walking can be calculated. Using these values all the leg joint forces can be found. The hip reaction forces are then applied to the finite element model of the socket. By varying how the loads are applied to the socket model, through a number of iterative solutions a workable loading scheme can be found.

Using this loading scheme the socket material can be compared, along with various socket frames. Similarly as the knee in an artificial leg can be moved, various alignments can be compared giving an idea of what is the better alignment structurally. Thus the best socket and frame combination can be found for the patient.

## Chapter 1.

### **ABOVE KNEE PROSTHETICS.**

An artificial leg attempts to function like a normal leg.

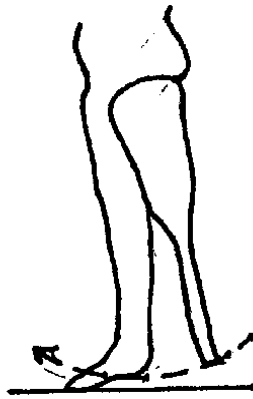
It has to provide a means of support and ambulation. The ideal artificial leg resembles a normal leg closely in how it allows the amputee to walk.

The first artificial leg, the peg leg, caused a pole vaulting effect during walking. This was due to the user having to raise the normal ankle to allow the peg leg to swing clear of the ground through the swing phase. Similarly the user tended to tilt the pelvis for the same reason. These caused bad walking styles, while consuming a lot of energy. By incorporating joints to the artificial leg these problems can be solved.

Introducing a foot and ankle unit, that permits planter flexion (foot leg angle greater than 90 deg) while blocking dorsiflexion (foot leg angle lesser than 90 deg), ambulation can be achieved with a free knee joint. This means the knee will swing like a normal knee during walking. Resulting in a smoother walking motion.

#### **Peg Leg Pole Vaulting**

1.1



### Requirements

Requirements of the artificial leg are specified in the I.S.P.O.'s "Standards for Lower-limb Prosthesis", soon to be formalised to I.S.O standard. These summarise the requirements with an emphasis on environmental conditions and not structural requirements. The structural requirements mainly cover the joint mechanisms.

The only requirement, PH-2 11.2, is for all components to withstand a static axial load of 2080 newtons compression with an axial torque of 24.5 newton metres. They should also be able to withstand a dynamic load of 1060 newtons compression and 12.5 newton metres torque. The loads do not have to be combined for static testing, but they must be for dynamic testing .

However the function of "the lower limb assemblies and components shall approximate the anatomical counterparts in function and provide the types and ranges of motion to produce optimum performance and gait", from PH-2 8.1 in the standards.

### Phases of Gait

The motion of walking are divided into two phases for each step. These are the stance phase and the swing phase. The stance phase takes approximately 63 % of the total time of the step. This value can vary for each subject.

Stance phase starts with heel contact. After heel contact the leg and foot pivot about the heel until the foot is flat. After this the pivot point moves forward along the foot.

The body now starts to move ahead of the foot, the leg pivoting at the toe break. The ankle angle now increases. There has been little knee movement up until now. As the thigh and body increase in forward speed the knee starts to bend lifting the foot and shank. Thus starting the swing phase.

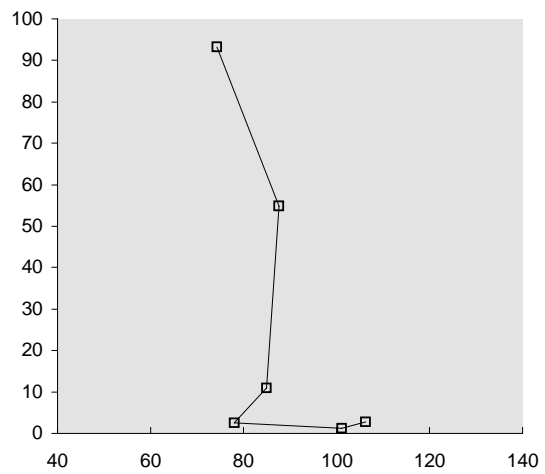
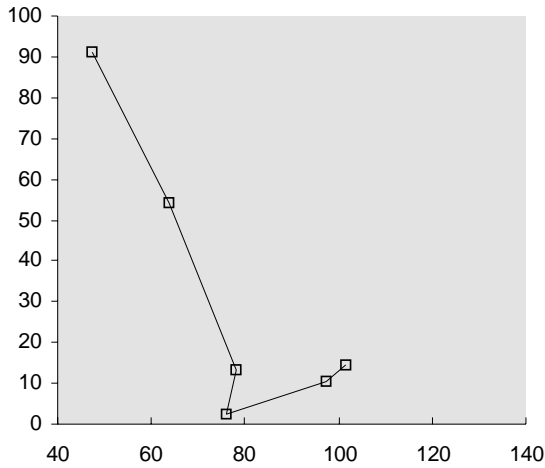
As the thigh continues to move forward gravity counteracts the knee moment until the leg is straight again, in a pendulum type motion. At which point heel strike occurs again.

### Phases of Gait.

1.2

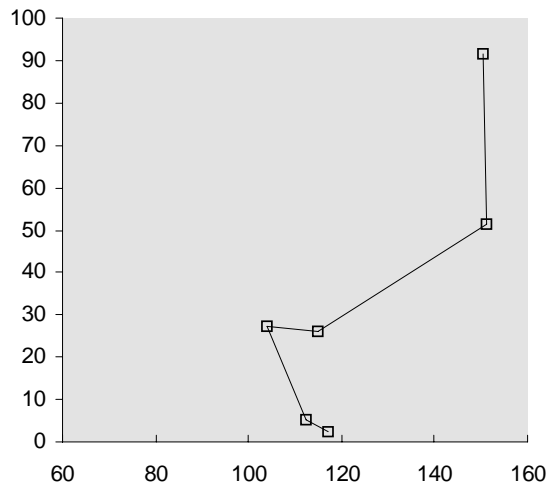
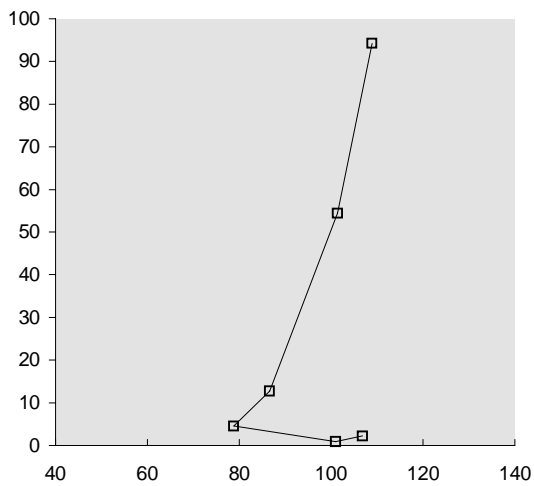
Heel Strike

Toe Strike



Heel Off

Toe Off



**Knee Mechanism.**

As can be seen the knee has to be free to swing yet locked during the stance phase. There are three classes of knee mechanisms with several types in each class. The first class is a free knee where the only limitation on rotation is from the friction of the bolt and bush assembly.

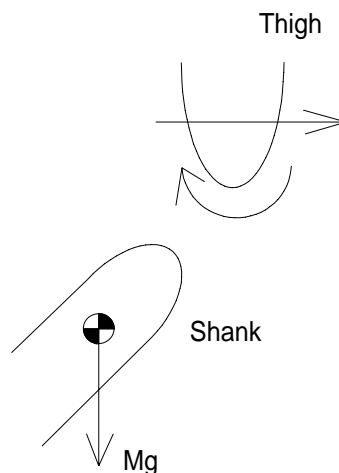
The second class allows for adjustment of resistance to rotation. The resistance is either from mechanical friction or fluid damping. They also include a knee extension aid which

produces moments of different magnitude in flexion to extension. They can also be responsive to cadence (stride length and speed of walking). Class two is thus swing phase adjustable.

Class three is swing and stance phase adjustable. The mechanisms that control knee rotation are mechanical friction, fluid dampers and by shifting the effective centre of knee rotation. The stance phase can be controlled by resistance yielding devices, mechanical or hydraulic locks and friction brakes.

**Forces and Moments on the Knee during Swing Phase**

1.3



Similarly the foot/ankle mechanisms are divided into classes. Class one has movement in the medial-lateral plane. Class two has movement in both the medial-lateral plane and the frontal plane, while class three has movement in all three planes.

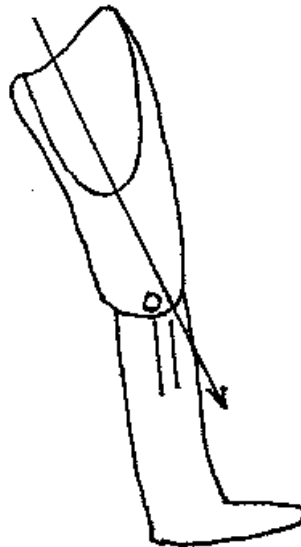
The artificial leg is thus made up of the joint mechanisms with the shank and thigh being pylons between the joints. They

are made up from kits with variations of components and their locations being decided by the prosthetist.

One of the requirements needed in the assembly is for the knee to remain locked during stance phase. The best way to do this is to have the reaction force at the hip acting forward of the knee bending mechanism. To achieve this the mechanism is usually behind the hip ankle line. Thus as the thigh moves forward the reaction forces moves forward at the knee until it is in front of the mechanism. At which point the knee rotates due to the start of swing.

#### Location of Knee joint

1.4



#### Sockets

The earliest sockets made from carved wood, were held in place by straps or semirigid belts. The inside shape was usually conical, requiring woollen socks to be worn between the socket and the stump to soften the interface.

Today the suction socket is used. Roughly quadrilateral in shape they attempt to have total contact between the stump

and the socket. The differential in atmospheric pressure and the air pressure inside the socket hold the socket in place.

The early suction sockets made of wood and fibreglass did not get total contact with the stump. However due to their thicknesses they did not require a frame between the socket and the knee. The socket was attached directly to the knee or thigh mechanism.

Using polypropylene as the socket frame material and Polyethylene for the socket, it is possible to have the stump in total contact with the socket. This is possible as the polypropylene is easily moulded after being heated. As the socket is clear any region where contact does not occur can be seen. Thus the socket shape can be improved at the time of fitting. Polyethylene is also flexible allowing some movement of the socket walls when in use.

The polyethylene socket does require the use of a frame to connect it to the knee. Two main types of frames are being used. These are a larger polypropylene socket, double socket, or a half fibreglass frame. The half fibreglass frame is difficult to make while the polypropylene frame is easily made.

The double socket has the socket sitting inside a larger socket attached to the shank and knee mechanisms.

The polypropylene double socket frame is being used more, due to it's convenience.

### **Socket Frames**

1.5



Double Polypropylene Socket



Fibreglass frame  
Socket ( ISNY Style) with  
hoop open.

## Chapter 2

### **GROUND REACTION FORCES AND SUBJECT GEOMETRY.**

In order to find the external loads on the socket, the loading on the leg and the leg geometry are required. The leg loading was found by finding the ground reaction forces, ground forces acting on the foot, at the required phases of gait. The geometry was measured from the subject, while the segment masses and their inertia were calculated. This allowed for the leg to be broken up into segments, each with its own free body diagram. By working up from the bottom the free body diagram for the thigh was found.

#### **The Geometry**

The following measurements were taken of the subject.

Ischium - Heel 795 mm.

Ischium - Knee 343 mm.

Heel - Ground 18 mm.

Foot length 270 mm.

Heel - Toe break 210 mm.

Mass 60 kg.

A few more measurements were still required, noticeably knee to ankle and ankle to heel. A good approximation for these were found from Drillis and Contini (1966) where the body segment lengths are calculated as a percentage of height.

Similarly ratios can be found from a live subject, such as used by Winter (1979).

**Table 2A Segment Length as Percentage of Height**

Segment	Drilllis	Winter	Subject
Hip - Knee	0.200 H	0.228 H	343 mm
Hip - Ankle	0.446 H	0.469 H	?
Heel - Ankle	0.039 H	0.061 H	?
Knee - Ankle	0.246 H	0.242 H	?
Foot length	0.152 H	0.159 H	270 mm

H = Height of Subject.

Using the Drilllis percentages, H= 1733 mm. These give heel ankle = 67.6 mm, Knee - ankle = 426 mm. Unfortunately adding these segments up give the hip - heel distance as 834 mm, where as the real distance is 795 mm. The reason for the difference lies in two area's. Individual differences and secondly in the determination of joint location. Most noticeably the hip joint were the subjects hip is taken as the Ischium, while Drilllis used the Greater Trochanter.

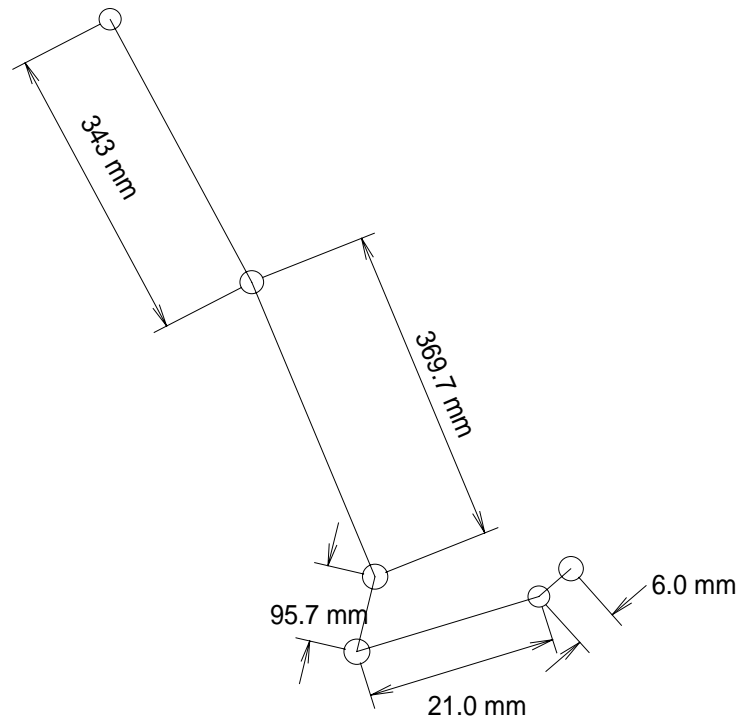
Using the percentages of Winter, gives H = 1698 mm using the foot ratio. While H = 1530 mm using the thigh lengths. By using the foot H for ankle heel distance and thigh H for knee ankle distances, to allow for the foot being larger for height than that of the subject of Winter's, gives;

$$\text{Heel - Ankle} = 369 \text{ mm}$$

$$\text{Ankle - Knee} = 95.7 \text{ mm.}$$

Adding these segments gives an equal length to that measured allowing for the heel being out of line with the hip, knee and ankle. These here the lengths used.

2.1



**Anatomical Location of joints**

Hip - Ischium.

Knee - Lateral femoral epicondyle.

Ankle - Lateral malleolus of fibula.

Heel - 18 mm above ground in line with rear of shoe.

Toe Break- 5th metatarsal phalangeal joint.

### Masses and Inertia

As segment lengths can be calculated as a percentage of height, so too can the segment masses be calculated as a percentage of total mass. This is the only way to find the segment masses, as measurement on a live subject is inaccurate.

One of the first studies done on segment masses and their inertia was by Braune and Fisher (1892). Although an old document, they had access to large numbers of cadavers allowing for an accurate study, and as there has been no change in the human body their data has been accepted and used in subsequent investigations.

Bresler and Frankel (1950) used slightly different coefficients in their study on forces and moments in the leg. As their study is used in many investigations their coefficients are compared.

Finally the coefficients used by Winter (1979) on the single subject as used in the segment lengths are compared.

Bresler and Frankel used the same method as Winter for finding the segment mass moment of inertia, about the segment C.G.. They both used  $\text{Inertia} = \text{mass of segment} * \text{radius of gyration squared}$ .

Braune used a slightly different method where instead of a segment radius of gyration he uses  $0.3 * \text{segment length}$ . The 0.3 is a ratio of the radius of inertia about any axis perpendicular to the longitudinal axis and passing through the center of gravity. Braune found the ratio about 0.3 for all

segments of the body. Thus Braune's equation is  $Inertia = (0.3 * \text{segment length})^2 * \text{segment mass}$ .

Comparing results using these coefficients shows some differences. These differences most notably occur for the foot mass and its inertia. Differences of 40 % occur above the average for foot inertia calculations. Comparing over the whole leg, where one source has a smaller inertia for a segment than the others, the next segment is usually larger than the others. Giving all methods similar inertia for the whole leg.

The external forces are of a magnitude times 100 larger than those due to the foot accelerations and inertia. Thus any error in foot mass or inertia will result in a smaller error for foot segment forces. In other segments the errors were far smaller. There was little difference in centre of gravity location for all methods.

**Table 2B Coefficients for Segment Lengths and Masses**

Segment	Seg Weight/ total body weight	Centre of mass/ Segment Length	Radius of Gyration
Foot	0.021*	0.35*	
	0.0179#	0.35#	0.30#
	0.0145@	0.50@	0.475@
Shank	0.045*	0.433*	
	0.0527#	0.42#	0.25#
	0.0465@	0.433@	0.475#
Thigh	0.110*	0.435*	
	0.1158#	0.44#	0.31#
	0.100@	0.433@	0.323@

\* - Braune and Fischer, # - Bresler and Frankel, @ - Winter

The distances are from the proximal joint (The one closest to the torso).

The average masses and inertias were used as these would allow for any uncertainties in the various methods. Appendix 1 gives the relative masses and errors for the different methods.

**Table 2C Summary of Segment leg Masses and Inertia**

	<b>Mass</b>	<b>Inertia</b>	<b>C.G location</b>
	<b>Kg</b>	<b>kg m*m</b>	<b>% seg length.</b>
<b>Foot</b>	1.068	$0.987 * 10^{-2}$	0.35
<b>Shank</b>	2.884	$3.167 * 10^{-2}$	0.433
<b>Thigh</b>	6.516	$7.402 * 10^{-2}$	0.433

**Ground Reaction Forces**

The ground reaction force is the main force acting on the body during walking. It consists of a vertical component and two horizontal components. These forces are found by having a subject walking across a force plate in the form of a walkway.

Shoes with a force plate sole have been made but results have been inaccurate compared to walkway method.

Two types of force plates are in common use. The first one consists of four triaxial transducers. The magnitude and location of the ground reaction force can be determined by differences in force readings of the transducers.

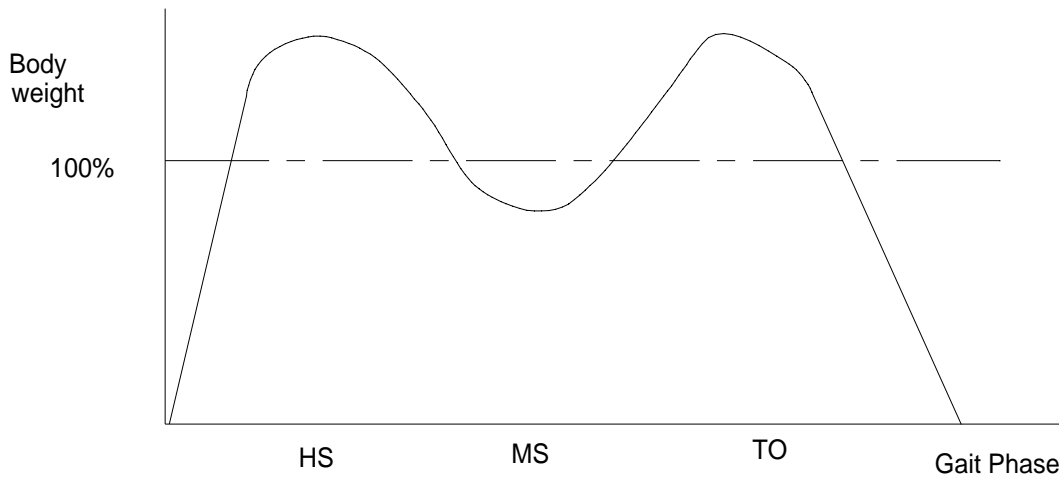
The second type is the central support type. One centrally mounted instrumented pillar supports the plate. This records the reaction force, centre of pressure and the vertical twisting moment. Summing moments and forces gives the relative magnitudes and directions.

Data from both methods are readily available. These are usually in the form of percentage of body weight verses phases

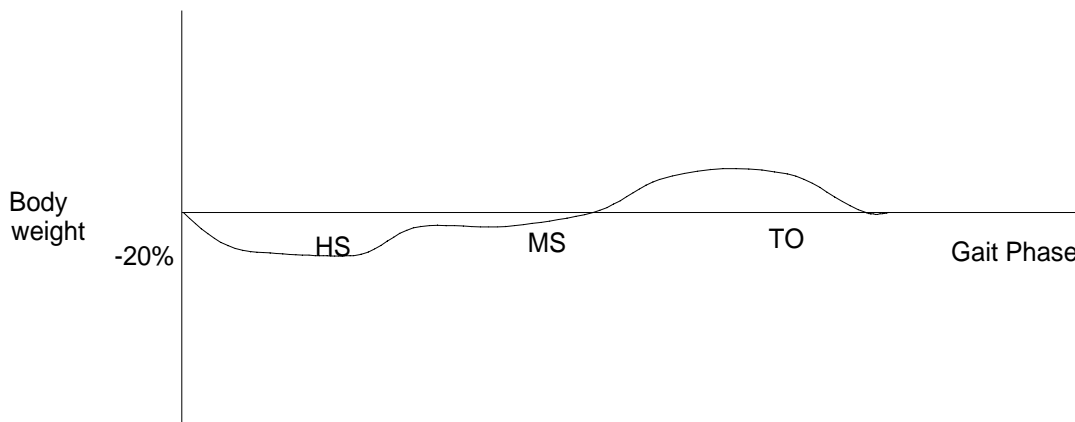
of gait. Correlation between all data is good, giving the following graphs.

**2.2 Ground Reaction Forces as a Percentage of Body Weight.**

Vertical Reaction Forces



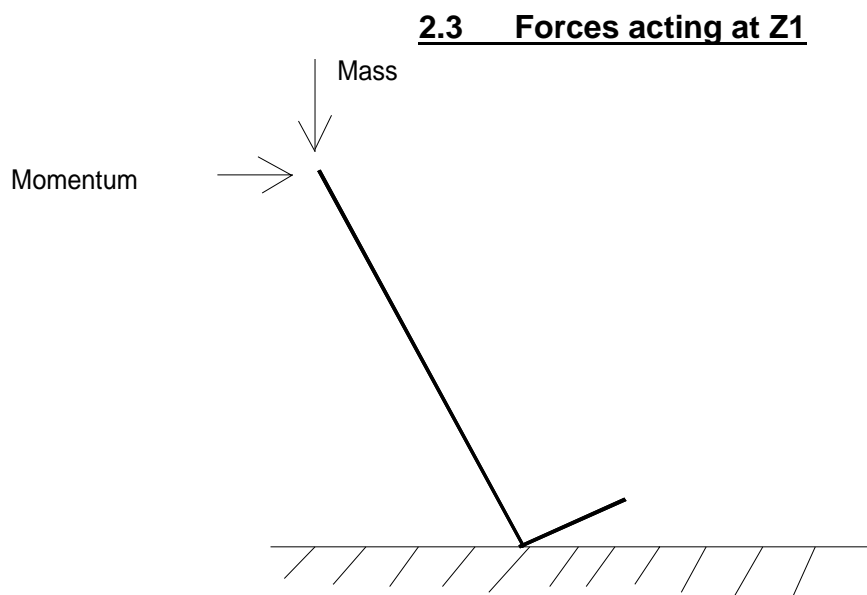
Anterior Posterior Forces



The lateral component for this study has been neglected as it is insignificant compared to the vertical force.

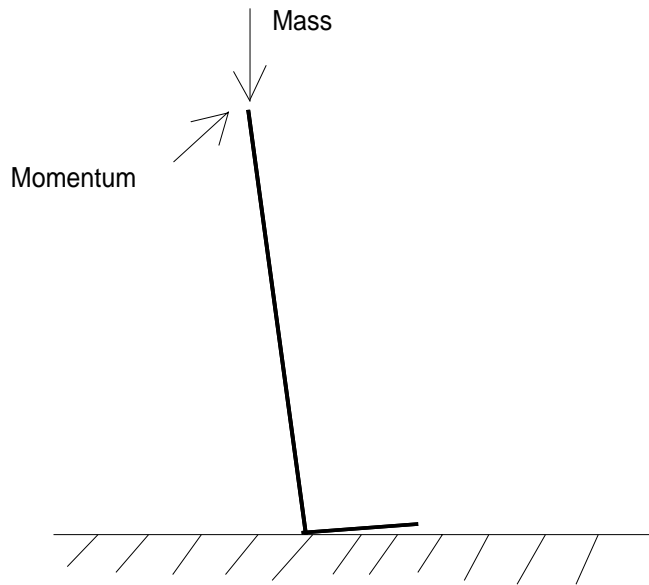
From these graphs three phases of gait were chosen as significant for the socket. They correspond to the peaks and troughs of the vertical force graph. Z1 occurs soon after heel strike, Z3 at mid stance and Z3 just before toe off.

The forces acting at heel strike are shown below. It includes both gravitational forces and momentum of the body. As the leg has to withstand the forward momentum of the body and the body mass, a local maximum occurs when the leg loads up.



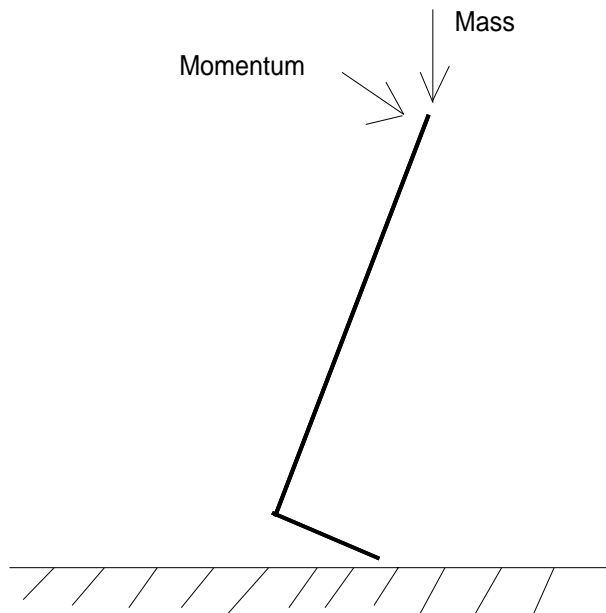
After Z1 the body pivots on the ankle. Giving it an upward momentum. This reduces the ground reaction force until a minimum at Z2 occurs.

**2.4 Forces acting at Z2.**



After mid stance the ground reaction force increases again. This occurs as the vertical momentum acts downwards due to the leg geometry. The momentum comes from the muscle force of the opposite leg.

**2.5 Forces acting at Z3.**



The values of Z1, Z2 and Z3 are available from various sources. Table 2D shows some comparable values. Differences

occur for various reasons including individual walking style, step length and the terrain walked over. Similarly the same subject will give different ground reaction force at different walking speeds. Generally higher walking speeds give higher forces (Andriacchi, 1977, Jansen, 1977, and Soames, 1983).

While acknowledging the subject would probably walk slower than average, the values used for analysis are slightly higher than average to give a conservative stresses. Similar values for the force component in the direction of travel are found.

**Table 2D Values for Z1, Z2 and Z3**

Source	Z1	Z2	Z3
	% Body weight		
Andriacchi, Andriacchi, Ogle, & Galanti, 1977	120	65	120
Bresler & Frankel 1950	Subject 1	105	79
	Subject 2	112	73
	Subject 3	117	52
	Subject 4	123	50
Cavanagh, Williams, & Clarke	104	74	101
Nakhla, King	100	90	100
Payne	130	67	130
Winter, 1979	114	75	106
Standards For Prosthetics	135	(1390 N for 105 kg Heathrow man)	
<b>Values Used</b>	<b>120</b>	<b>70</b>	<b>110</b>

**2E Values of Horizontal Force, corresponding to Z1, Z2, Z3**

Source	Z1	Z2	Z3
	% Body weight		
Andriacchi, Andriacchi, Ogle, & Galanti, 1977	-17	0	10
Bresler & Frankel, 1950	-12.1	-4	20
Nakhla, King	-24	0	16

Payne	-21	-2	20
Winter, 1979	-21	-4	15.6
<b>Values Used</b>	<b>-20</b>	<b>-4</b>	<b>15</b>

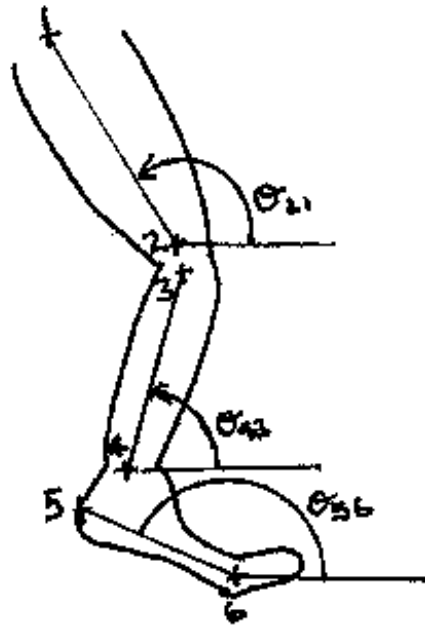
**Geometry for Z1, Z2 & Z3.**

The leg geometry is needed for the above loading cases. The position of the joints of the leg during walking can be found from either Television or photo data. In both cases the joints are highlighted against a dark background, against which the leg is photographed. These photos allow the angles between the joints to be measured data. The photos need to be synchronised with the force plate to get an accurate force displacement correlation. This is due to the force transducers giving impulses where as Television will give continuous data. The positions can also be found using a Goniometer.

A series of photos will give the displacements of the joints over time. This can be used to find segment velocities and accelerations.

2.6

**Joint Angles of the Leg.**



Due to the data having different segment lengths to the subject some modification is necessary. Appendix A from Winter 1979 has data for vertical and horizontal displacements as well as centre of pressure locations for a full cycle of gait. The data is taken at periods of one sixtieth of a second.

As all three phases needed occur during the stance phase when the foot position remains relatively constant, the metatarsal data was used as the reference joint. This joint moves little in the three phases while being close to the centre of pressure. The joint angles were assumed to be constant for each phase, with only the segment lengths varying between the Winter 1977 data and modified data. This was due to the leg in the Winter 1977 data being scaled, rather than a change in geometry.

Using the joint angles found from the Winter data, with the segment lengths of the subject, the leg geometry was built up for each phase. Similarly the centre of pressure was found.

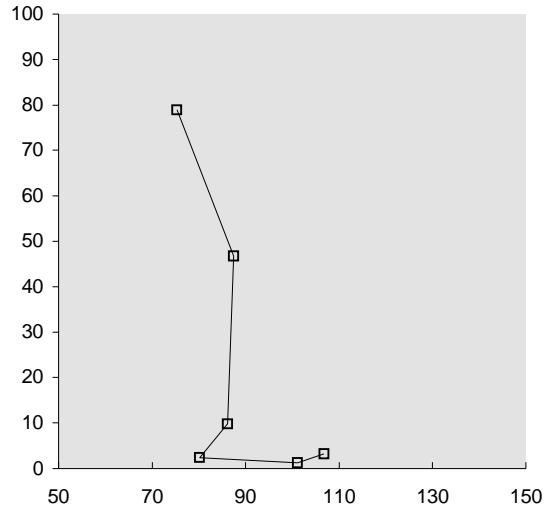
The joint angles and centre of pressure compare favourably with Payne 1977 and Winter 1974.

The angular and segment accelerations were found using finite difference methods for changes in displacements and angles. These compared favourably with Winter for angular acceleration, but not for normal acceleration. Due to a lack of significant figures (3 for displacements, 6 for angles ) in the joint coordinates the error occurred. To find the acceleration the displacement data has to be differentiated twice, compounding the error due to the lack of significant figures. Therefore I used Winters segment accelerations. These will be slightly larger than those of the subject, which should result in higher segment forces, thus giving a conservative stress analysis on the socket.

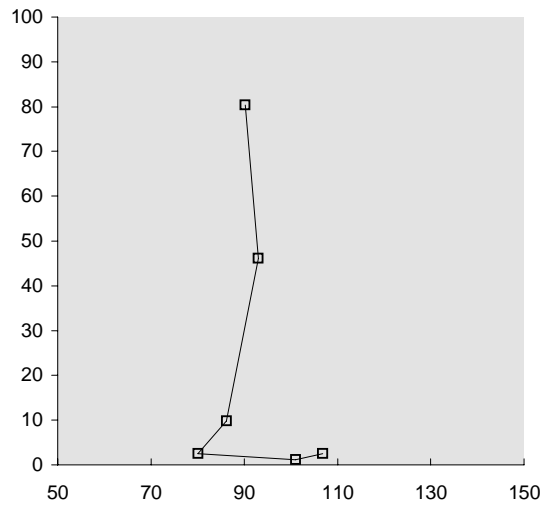
Appendix 2 Shows the calculations for the accelerations, subject geometry and displacements.

**2.7 Geometry for Phases Z1, Z2 & Z3.**

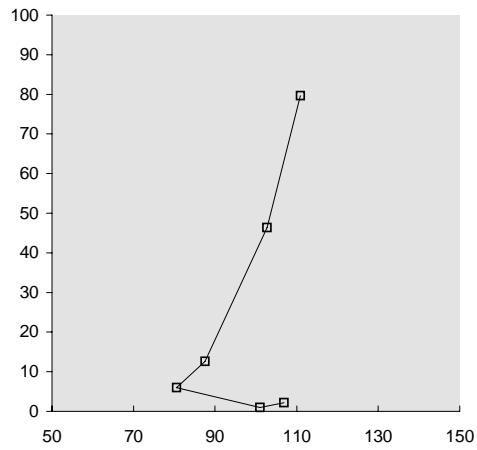
**Z1**



**Z2**



**Z3**



**Anatomical location of joints**

Hip - Ischium.

Knee - Lateral femoral epicondyle.

Ankle - Lateral malleolus of fibula.

Heel - 18 mm above ground in line with rear of shoe.

Toe Break- 5th metatarsal phalangeal joint.

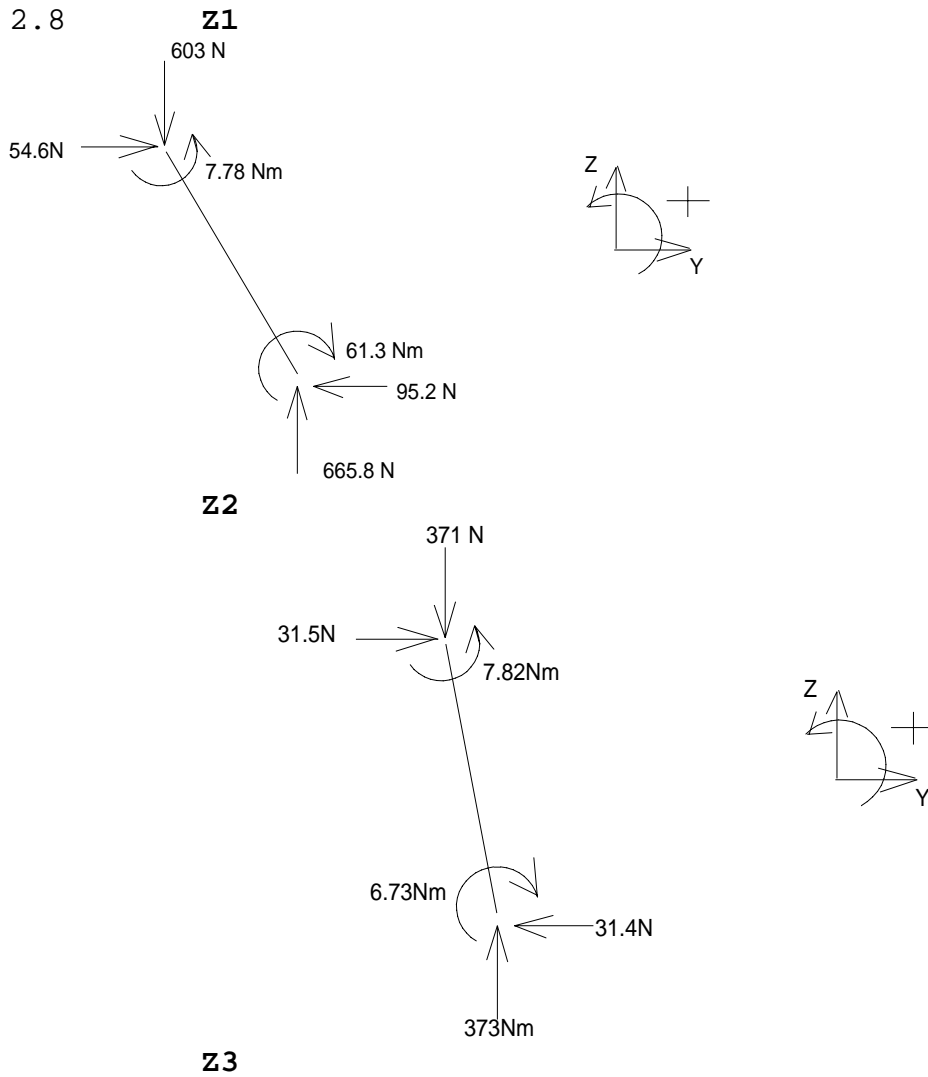
**table 2F Accelerations.**

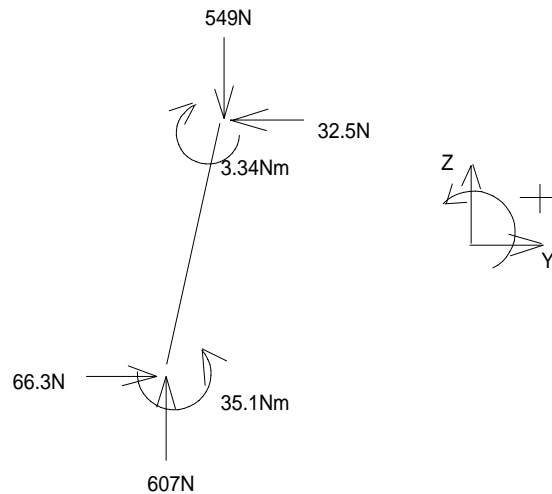
		Ax M/s/s	Az M/s/s	Alpha R/s/s
<b>z1</b>	Foot	-2.17	2.42	19.03
	Shank	-6.86	-0.28	15.53
	Thigh	-6.24	-0.18	-15.71
<b>z2</b>	Foot	0.22	0.11	-2.23
	Shank	0.59	0.15	-0.66
	Thigh	0.01	-1.16	3.44
<b>z3</b>	Foot	2.37	1.16	-23.41
	Shank	6.75	0.11	11.05
	Thigh	5.19	-0.96	13.96

**Thigh Forces.**

Using the geometry and the applied forces found, the forces at the thigh can be found. For calculations see appendix 3.

2.8





### Chapter 3.

#### MODEL AND LOADING.

The finite element package used was PAL 2. This package allowed for many iterations of the same model. As the load files can define the boundary conditions, as well as the applied loads, varying the load files was the quickest method of comparing socket frames and socket types. Various models and loading patterns were used.

#### The Model

The prosthetic socket was not an easy shape to model with finite element modelling. Due to its unique geometry no simplification of the shape or symmetrical simplification was possible.

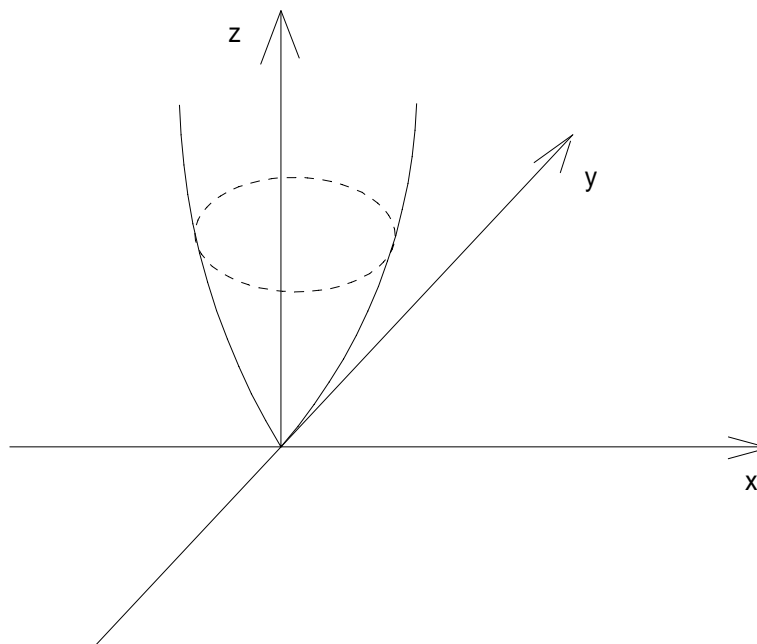
The original socket shape was measured using a flexicurve for correct shape reproduction. Firstly a two dimensional outline of the socket was obtained, in the medial-lateral plane. This gave a height to width relationship. At

approximately equal increase in height, the inside socket shape was obtained with the flexicurve. This built up a three dimensional socket shape.

Using the medial-lateral plane at the bottom of the socket as the x axis, the z axis was then taken as going from the midpoint of the socket brim to the x axis and perpendicular to the x axis. Using the right hand rule the positive y axis acts towards the front of prosthesis.

### Axis System Used

3.1



The dimensions for the socket brim were found for every thirty degrees rotation about the x-y plane. The height of the brim points were then measured relative to the height of one of the cross section measurements. The brim points were also cross checked with the closest cross section point for confirmation of exact position. Other brim points occurred whenever one of the cross section measurements occurred above

the start of the brim. Mapping these gave a more accurate brim reproduction.

The socket was measured using cylindrical coordinates. A measurement for every thirty degrees at each level was taken. These were then imputed by hand for every point into the model file. PAL 2 allows for extra points to be incorporated between the measured values. If the measured values had different radii, the extra points would occur along a helical path between the two measured points. This was accurate enough due to the small changes in angle between measured points. Lastly the socket thickness was measured using a micrometer.

Quadrilateral elements were used for most of the socket. Only at the base and at some parts of the brim were triangular elements used to satisfy the socket geometry. Where possible triangular elements were put one up, one down, to maximise accuracy. Due to the shape of the socket some elements were warped, mainly around the socket brim.

Three models were used. Two were used to compare the effect of grid refinement, one with refinement in the brim only, the other a doubling of grid points. The socket with the least points was used for most iterations due to less computer time per iteration. Appendix 4 contains the three model files.

A full fibreglass socket and frame was used as the standard, against which all iterations could be compared.

The material properties of the socket had to be varied for different iterations. The values used are in table 3.A and were found from the listed sources.



**Table 3A Material Properties**

Material	E	Yield Strength	Poisson's Ratio	Density
	GPa	MPa		g/cm <sup>3</sup>
Fibreglass (E Glass epoxy) *	21.5	570	.28	1.97
Low Density Polypropylene #	11.72	35	.3	.92

\*Table 1.3, Agarwal & Broutman, 1980

#Frank & MacDonald, 1968 and Moore & Kline, 1984

**Loading**

A method of applying the loads from the hip to the socket is required. The socket when loaded by the stump normally would have the load transferred nonuniformly. This is due to parts of the stump carrying higher loads than other sections of the stump. Parts of the stump such as bones and tendons can support more loads than skin and muscle.

Unfortunately during gait the load transfer distribution also varies. At heel strike and mid stance the back of the stump carries a lot of the load. The socket has a lip at the posterior brim to support the ischium. Traditional theory has the ischium transferring most of the load to the socket during stance. However during the swing phase the ischium moves away from the lip so little contact is made, hence little load transfer takes place there at heel strike.

Therefore to accurately predict how the transfer takes place is difficult. To get a reasonable load transfer I used

an iterative process, comparing the results to published data, to assess the method of load transfer.

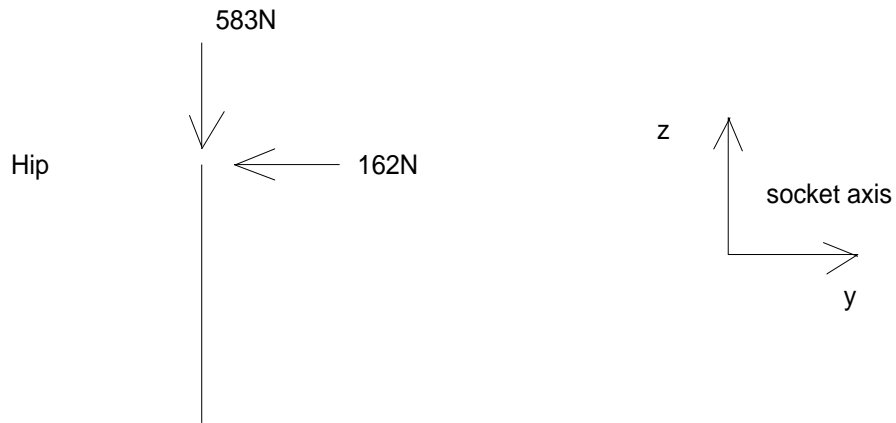
The first iteration used was a distributed loading. This was applied to a square plate at the top of the socket. The plate was located with its centre in line with the centre of the socket brim at a height equal to the top of the socket. The loading was distributed at the node points on the plate. It was then transferred to the socket by extra elements joining the plate to the socket. Any moments at the hip were applied to the centre node of the plate. Steel was the first material used for the plates and transfer elements. Subsequent iterations used an imaginary material with a value of Youngs modulus of 20 MPa. This was used to compare steel with a more flexible material, such as stump tissue. Kenedi, 1974 has some values of the stress strain relationship of skin, unfortunately they are not uniform allowing for a constant modulus.

The z-axis of the socket geometry was assumed to lie along the path of the thigh. Further iterations aligned the socket with a six degrees flexion offset and a ten degree extension offset.

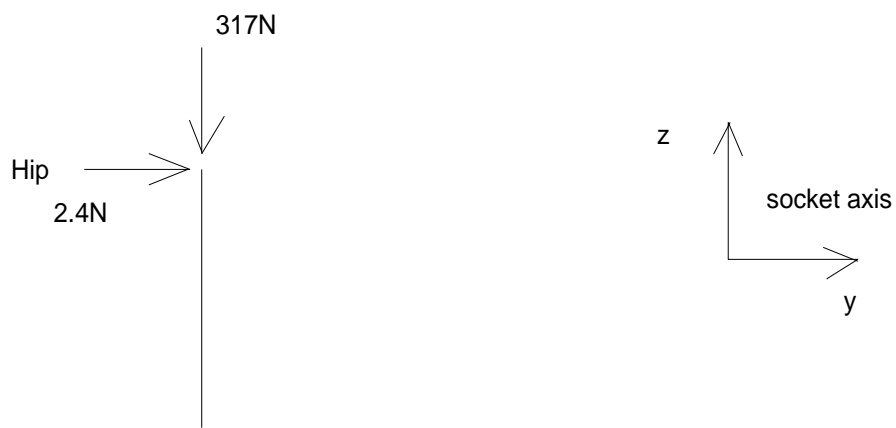
In all cases the vertical loads and horizontal loads of the free body diagrams had to be normalised to the socket geometry. Appendix 5 shows how the loads were normalised.

**Normalised Loads Zero Offset**

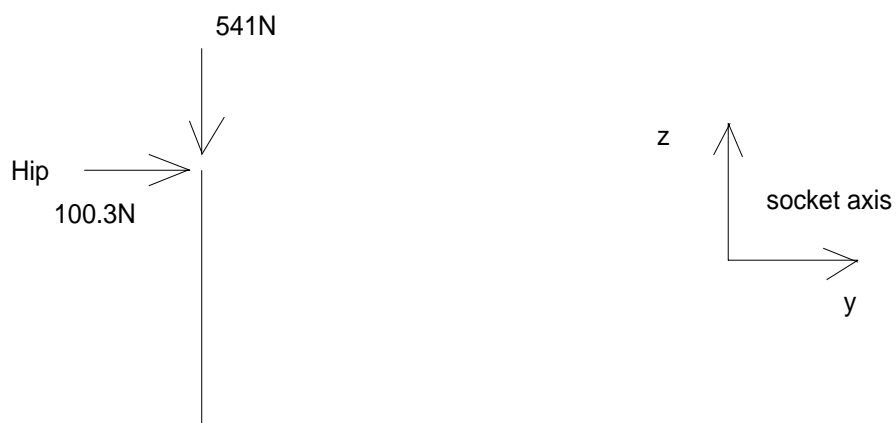
3.2 Heel Strike Z1



Mid Stance Z2

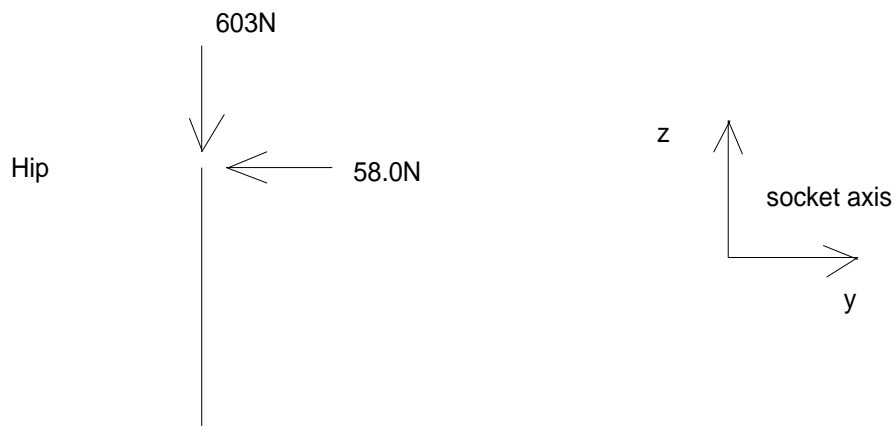


Toe off Z3

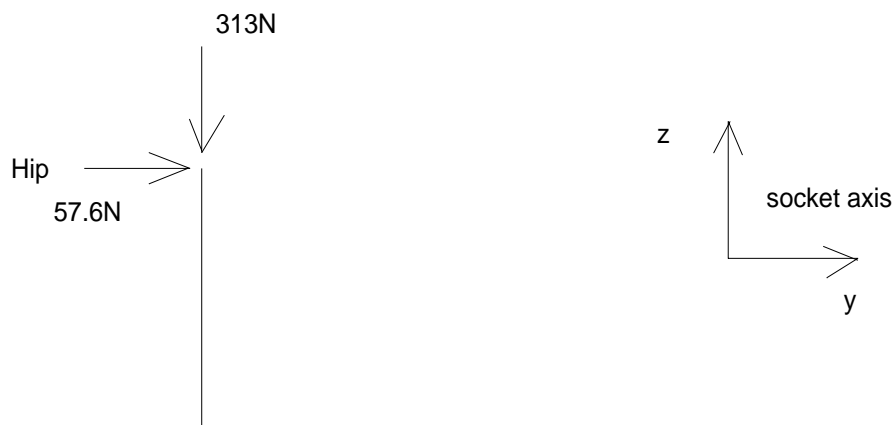


**Normalised Loads 10 Degree**

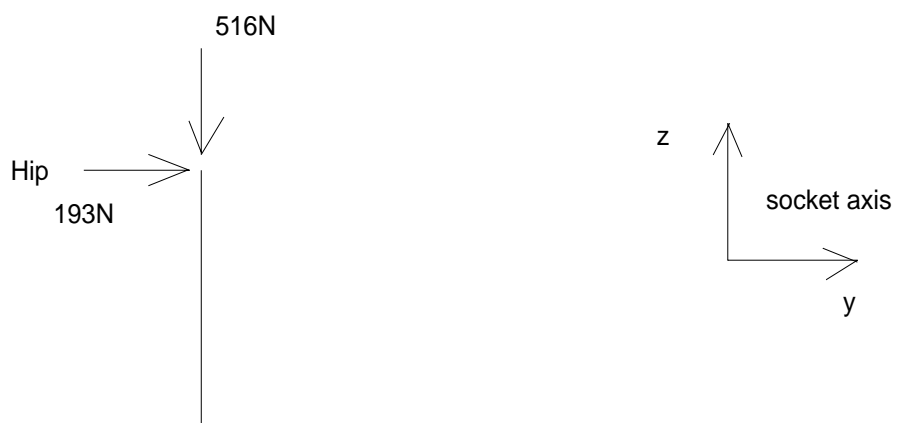
3.3 Heel Strike Z1



Mid Stance Z2

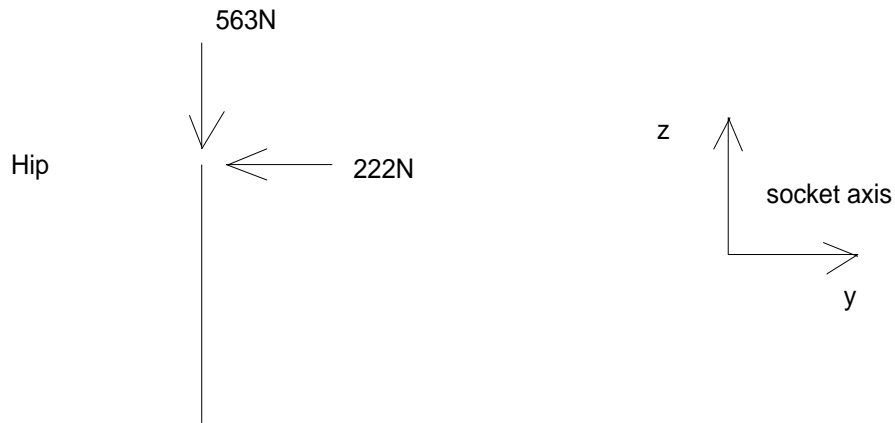


Toe Off Z3

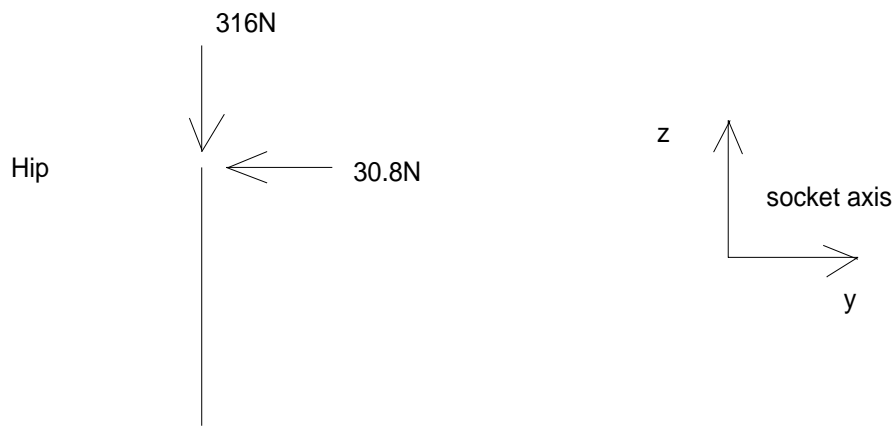


**Normalised Loads 6 Degrees Flexion**

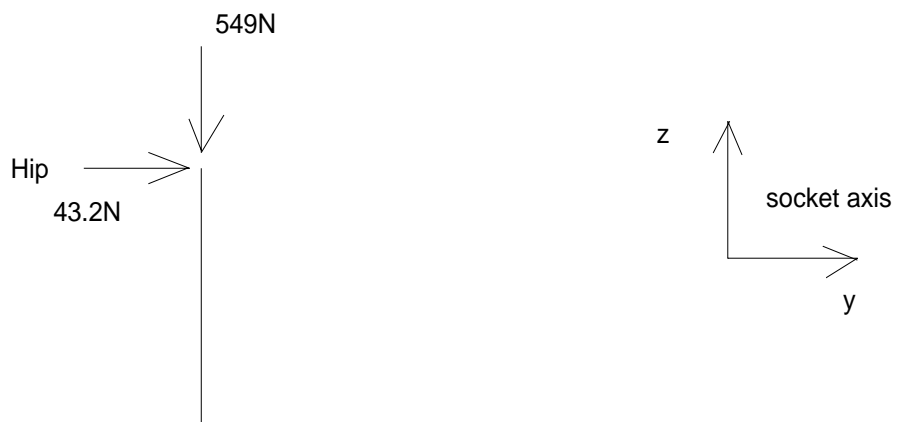
3.4 Heel Strike Z1



Mid Stance Z2



Toe Off Z3



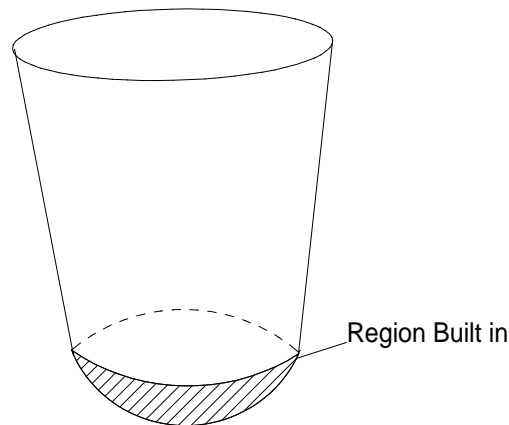
**Boundary Conditions.**

The boundary conditions of the socket varied for each iteration. By restricting the displacements of nodes an artificial socket frame could be made. Thus changing the boundary conditions changed the socket frame.

For the fibreglass socket the boundary conditions consisted of the distal end of the socket attached to a fixed end. This had all the nodes at the distal end with displacements and rotations set to zero. This was assumed as the fibreglass socket also acts as the socket frame. An iteration with a simply supported end was also tried, where only the displacements were set to zero. The fixed region is shown in figure 3.5.

**Fibreglass Socket Boundary Conditions.**

3.5



Four types of polyethylene socket attached to fibreglass frames were tried. These consisted of both closed and open frames mounted either on the medial side with the hoop closed or open on the lateral side, or mounted on the lateral side with the hoop on the lateral side either open or closed. The

nodes at the socket interface this time had their displacements only set to zero, as the socket sits in this frame type.

### Boundary Conditions for Polyethylene Sockets.

3.6



Built up support has zero displacement

Appendix 6 shows all the load files used, with brief explanations.

## Chapter 4

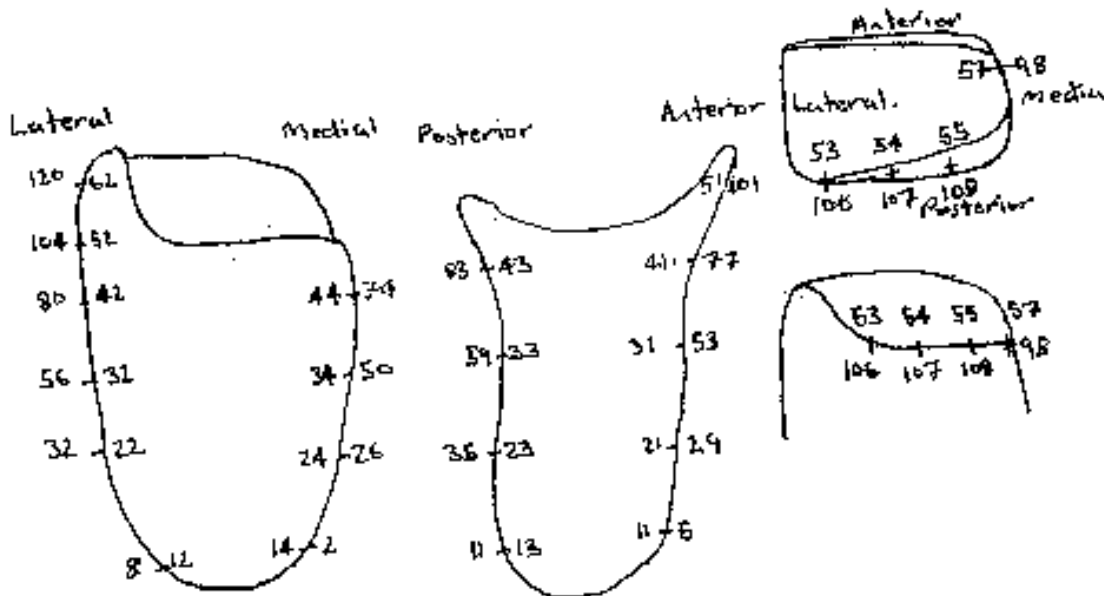
### RESULTS

Socket and socket frame types were seen to vary the socket loading. Similarly, phase of gait and socket alignment also varied socket loading. A series of points on the socket were compared with other iterations and researched results.

The points chosen were used to give a good overall picture while including critical points on the socket. They also compare with the points used by Appoldt and Bennett, 1967 and Bielefeldt and Schreck, 1979. Points were taken at the medial, lateral, anterior and posterior walls at the same height. Points were also taken along the socket lip.

#### Position of Reference Points.

##### 4.1



The equivalent Appoldt and Bennett point number is shown on the inside of the socket while the node number is shown on

the outside. The stresses for the points in figure 4.1 for all iterations are shown in appendix 8. The total stress distributions are available on the accompanying disc.

### Comparison The First Iteration

The first iteration was a fibreglass socket and frame. The loading plate and connecting elements were made of steel. The bottom end of the socket was simply supported.

The first iteration looked at the loadings Z1, Z2 and Z3 which became the respective standards for heel strike ( Z1 ), mid stance ( Z2 ) and toe off ( Z3 ). These compare with the Appoldt and Bennett times of 100 milliseconds, 300 milliseconds and 470 milliseconds.

As expected the socket stresses were less for mid stance than the other two phases. The brim stress was also higher than stresses in the rest of the socket. The stress at node 106 (Appoldt and Bennett, point 54 ) was usually the highest stress point. This compares well with Appoldt and Bennett who found this to be the highest pressure point. This seems reasonable as this point will receive direct loading itself by a large group of tendons acting between the body and the leg.

Node 120 was not a good comparison with point 62. As it lies on the socket brim and as such receives part of the loading from the plate. Appoldt and Bennett, 1967, have little or no pressure reading at this point. The node is located above than the Greater Trochanter. Due to the bone geometry this should be a low loading region. Therefore the stresses

seem to be artificially high at this point, as a result of the assumed load plate.

Comparing the Appoldt and Bennett data to the results from the model can only be used as a rough guide. Appoldt and Bennett recorded the pressure distribution while we determined the stress distribution. Thus the Appoldt and Bennett data will not necessarily mirror the stress distribution, while different sockets and loadings will also cause differences.

To some extent the stress distribution mimics Appoldt and Bennett's data. The lateral side seems to be the exception. This could be due to our two dimensional force geometry. The third force would be acting in the medial lateral direction. There is also a vague similarity with the pressure distributions found by Bielefeldt and Schreck. Their data highlights the large effect different sockets have on the pressure distribution.

#### **Further Iterations**

A series of iterations were processed with a different transfer material between the plate and the socket. This was done to mimic stump tissue. This made little difference in the loading with changes generally less than ten percent. The brim stresses were generally less than those of the steel plate. This was probably due to the new material absorbing some of the load. Overall the rest of the stresses were slightly higher. This was prominent except for the medial side at heel strike and mid stance and everywhere for toe off. Therefore no

definite reason was apparent to justify using the softer material in the model.

A stress rise at the distal end of the socket is evident on all iterations with the full fibreglass frame. This seems to be a result of this region transferring the load to the knee. A larger region than that used would alleviate this, although this rise in stress would occur at the transfer points between the socket and the knee.

An iteration having the end simply supported instead of built in was used. There was still a stress increase with the simply supported socket. The shear stresses stayed approximately the same, while a decrease in major stress resulted in a similar increase in minor stress and visa versa.

Therefore there seemed to be no advantage between either a built in end or a simply supported end.

### **The Effect of Refining the Model**

The difference in the brim refined model and the standard model is minimal. The same general stress increases and decreases occur. The main differences are that the node 106 is more prominently defined as the highest stress region of the brim compared to the refined model. The node 98 which also has high stresses in the standard model, has a decrease in stress in the refined model.

The other difference is that the major and minor stresses at the top of the medial side decrease in the brim refined model. This is probably due to the refined model dissipating

the load more evenly around the socket than in the standard model. However the brim refined model supports the stress distribution of the standard model.

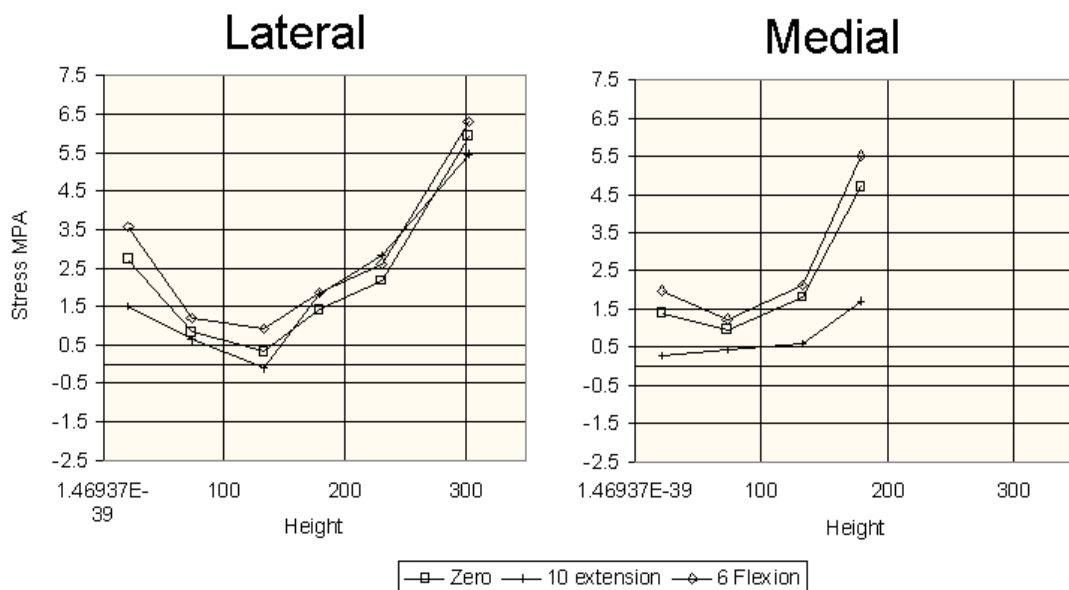
The most refined model has a very similar stress distribution to the standard model. The node at 106 is more prominent, while 98 decreases. These are the only differences between it and the standard model. Thus the standard model is accurate enough for the analysis.

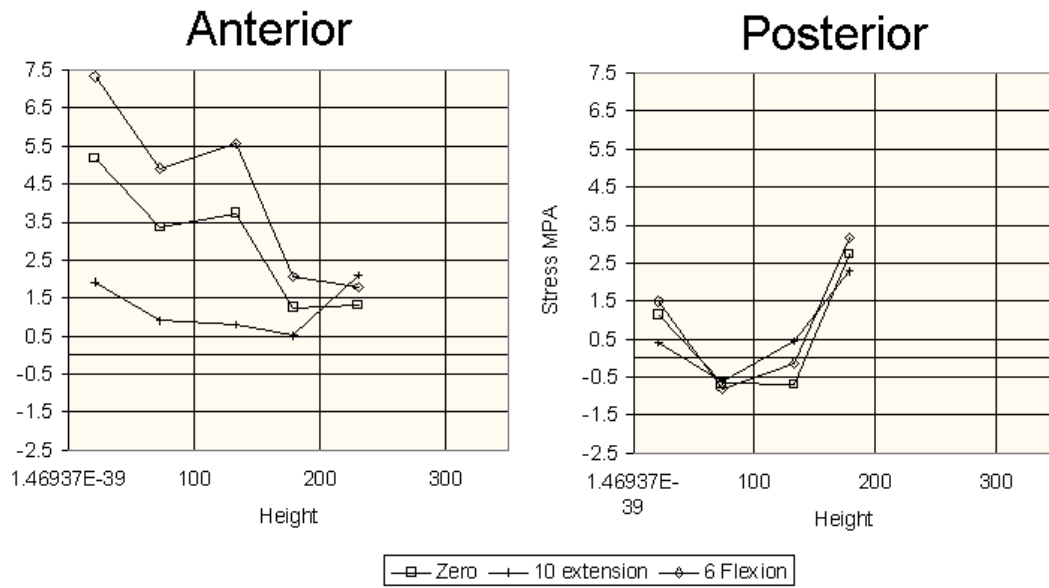
**The Effect of Socket Alignment**

A comparison of the socket loading for all three phases is shown in figures 4.2 to 4.5. It shows a comparison between a six degree flexion alignment, the standard alignment ( zero degrees ), a ten degree extension alignment.

**Heel Strike Stress Distribution**

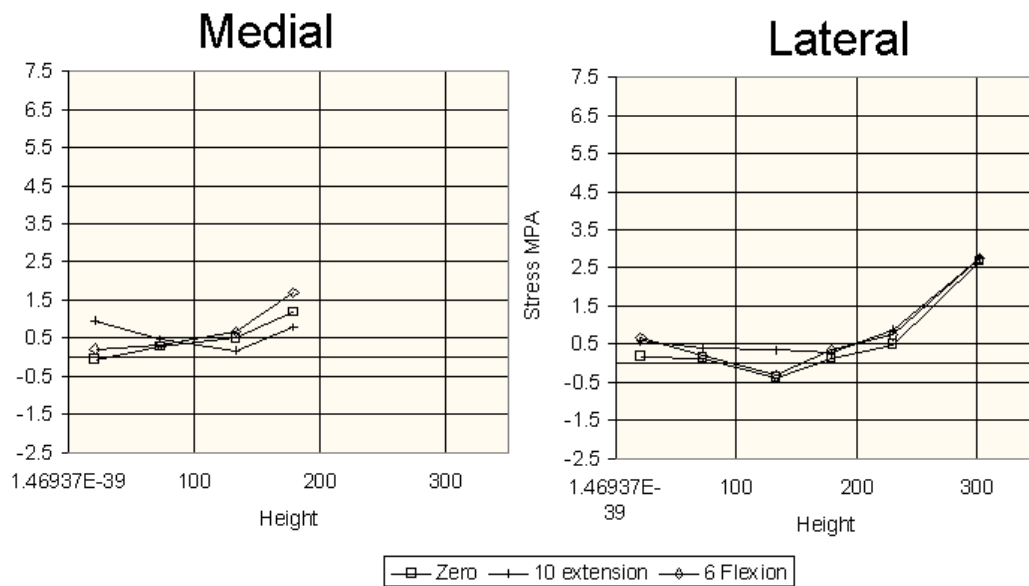
4.2

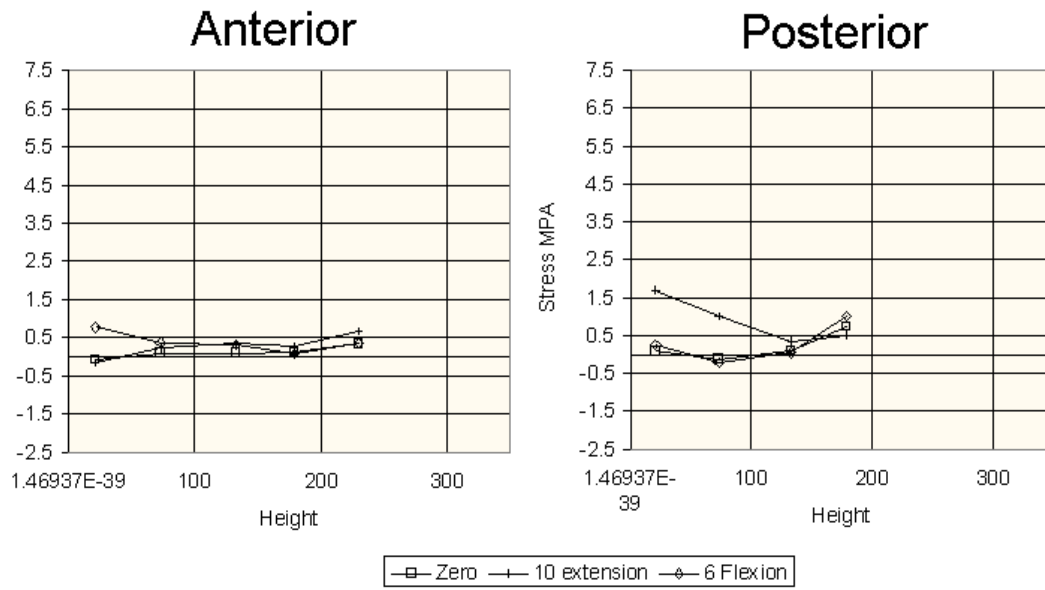




**Mid Stance Stress Distribution**

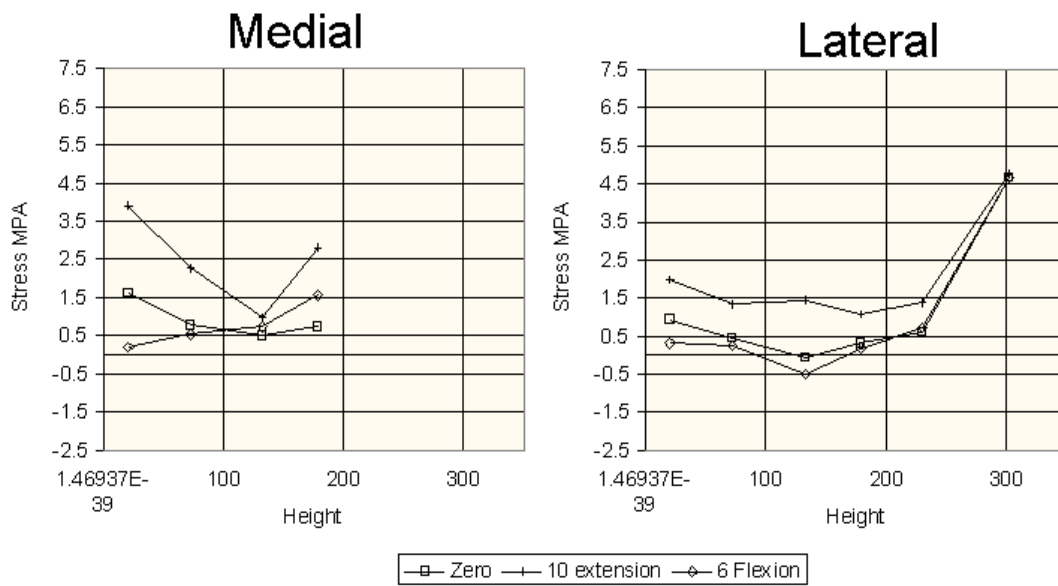
4.3

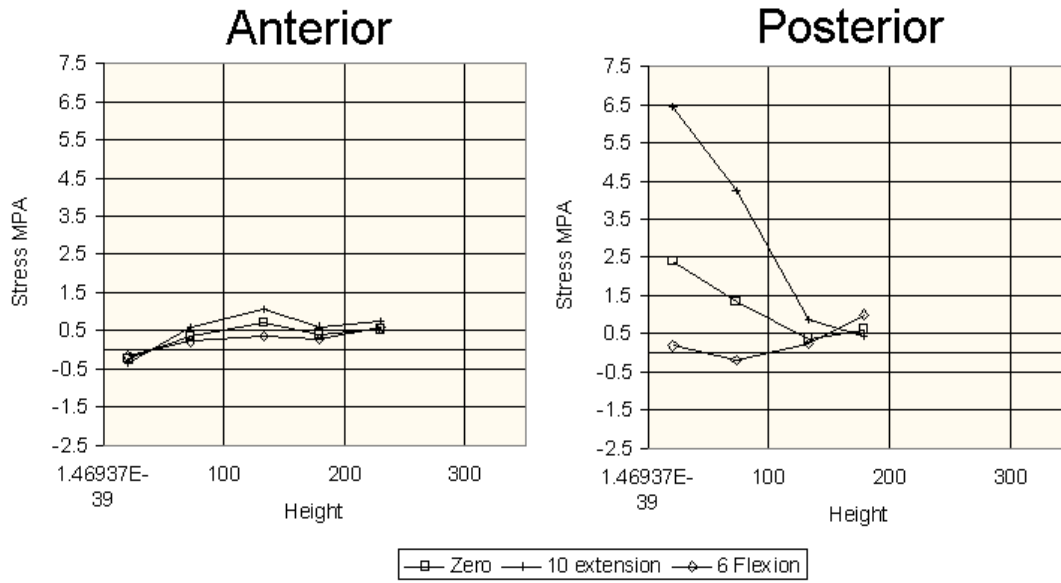




**Toe Off Stress Distribution**

4.4

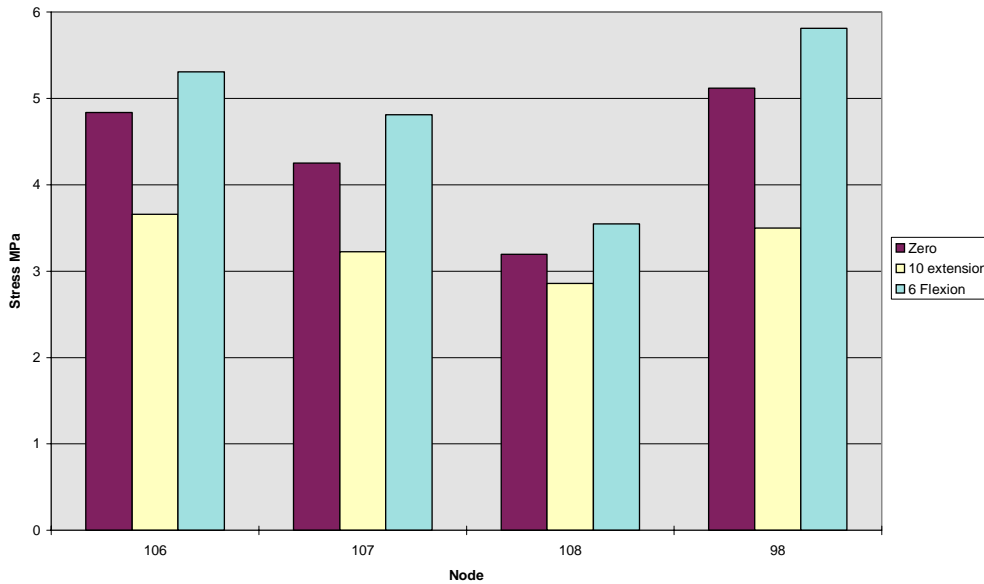




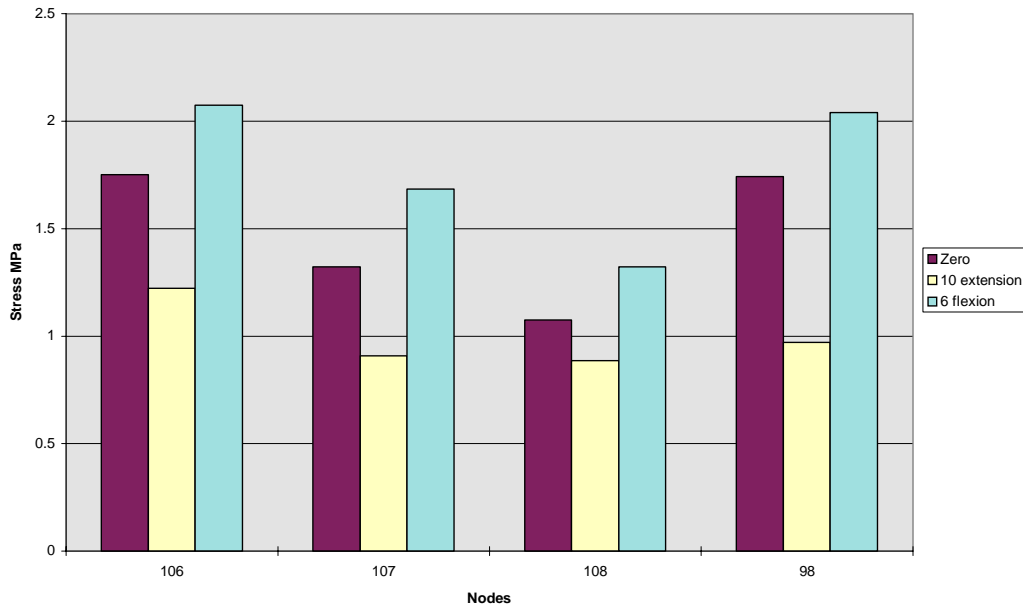
**Brim Stress Distributions**

4.5

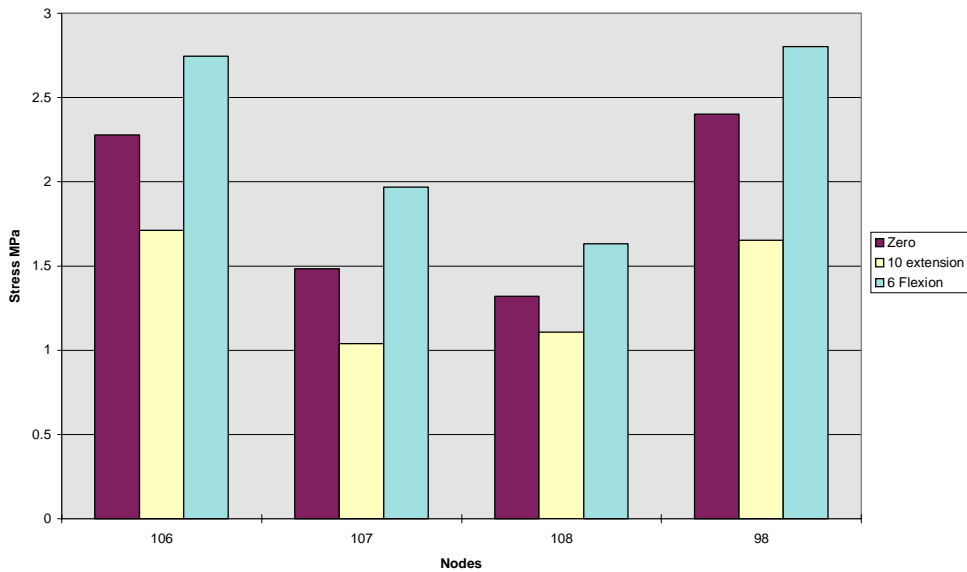
Heel Strike



Mid Stance



Heel Strike



The 10 degree offset had a varying effect on the stress distribution. During heel strike a decrease in stress distribution occurred in the following regions, medial side, bottom of the lateral side and on the front of the socket. This is explained as the socket will be more horizontal than

the standard at this phase so the front of the socket will be facing up, receiving little force.

The medial side remains lower than the standard for mid stance but the bottom rear of the socket has a large increase in stress. This becomes more prominent during toe off. The lateral side also has a large increase in stress at toe off.

The use of a ten degree extension offset has a higher stress distribution for a large part of gait so is not recommended. Most sockets have a five to six degree flexion when fitted so the above case is theoretical only. The only benefit for the extension offset is added stability at the knee.

Six degree flexion would therefore be a more likely case.

The effect of the six degree flexion offset was almost opposite to the ten degree extension. During heel strike an increase in stress occurred on the medial, lateral and anterior walls of the socket.

At mid stance the bottom front had higher stresses on it than the standard model. The medial wall also had an increase in stress. By toe off the bottom of the medial, posterior and lateral sides, together with the entire anterior all show less stress than the zero offset case.

The stress increases in the six degree flexion socket are less than those in the ten degree extension. The maximum increase is 0.5 MPa and 2.1 MPa is the maximum decrease in the six degree flexion offset case. The region with the most decrease in stress is the bottom front and back of the socket

at toe off. The maximum stress increase in the ten degree offset is 4 MPA occurring at the bottom back of the socket for toe off.

The brim stresses are similar for each alignment. The most critical effect of alignment variation is extension increasing stress on the bottom of the socket at toe off, while Flexion decreases the stress, there for toe off. This is the highest stress region apart from the brim.

#### **Effects of Frame Variation**

The effect of frame type also has a bearing on the stress distribution. Four partial fibreglass frames were tried with polyethylene sockets. They had the frame mounted on either the medial or lateral side with the top hoop either open or closed.

In general the frames had a similar effect. The stress was highest at the socket brim, decreasing drastically as you go down the socket. The medial socket with the hoop open had the highest stresses of the four types. The stresses were approximately ten or more times larger than the other socket types. Closing the hoop drastically reduces the stresses.

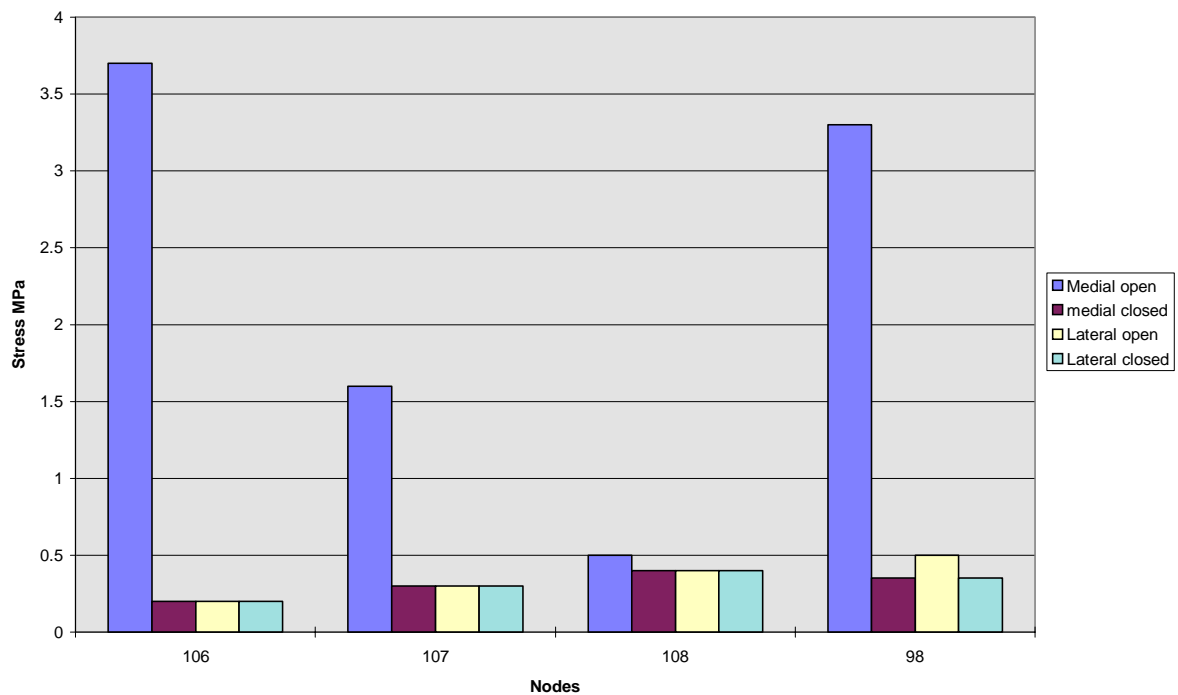
Figure 4.6 shows the differences in brim shear stresses. In practise the open and closed hoop medial frame has had extensive use. One real problem with the medial frame with the open loop has been that the beams of the loop have opened or slipped giving little support to the socket on the top lateral side. Here the sockets tended to split as they had to carry

the excess load. So the high stresses found in the medial frame open loop are justified. For the above hoop failure, the stresses would also increase due to the decrease in effective hoop size.

Prosthetist have been closing the hoop for various reasons. These include preventing the opening described above and using the top lateral position as a support point for hip belts. Results confirm the closed hoop to be the better frame.

**Stress Comparison of the Brim for Different Frames**

4.6



The frame mounted on the lateral side has low stress for both open and closed hoops. The difference in stress is negligible along the brim, while the stresses in the rest of the socket are insignificant compared to the brim stresses. The medial frame with the closed hoop had a very similar stress distribution to the closed hoop lateral frame.

The lateral frame has many practical advantages over the medial frame. It is mounted on the outside of the leg and not inside, reducing interference with the other leg. It will also allow for hip belts to be attached on the frame side instead on the hoop as with the medial frame. The lateral frame with an open hoop seems to be the best option of the four types of frame, due to it's low stress distribution, practicality and lower weight.

### **Double Sockets**

Another frame is the double socket. Two polyethylene sockets one shorter than the other firmly fit one inside the other. The outer one is attached to the knee, while the other supports the stump. Due to the fit, the outer socket receives a loading distribution similar to that of the inside socket. Same as if the stump acts on the outer socket directly. The outside socket is attached to the knee with the same boundary conditions as for the fibreglass socket. It therefore experiences a similar loading to the fibreglass socket. Thus the magnitude of the stresses will be different to those in the fibreglass socket, however the stress distribution will be similar.

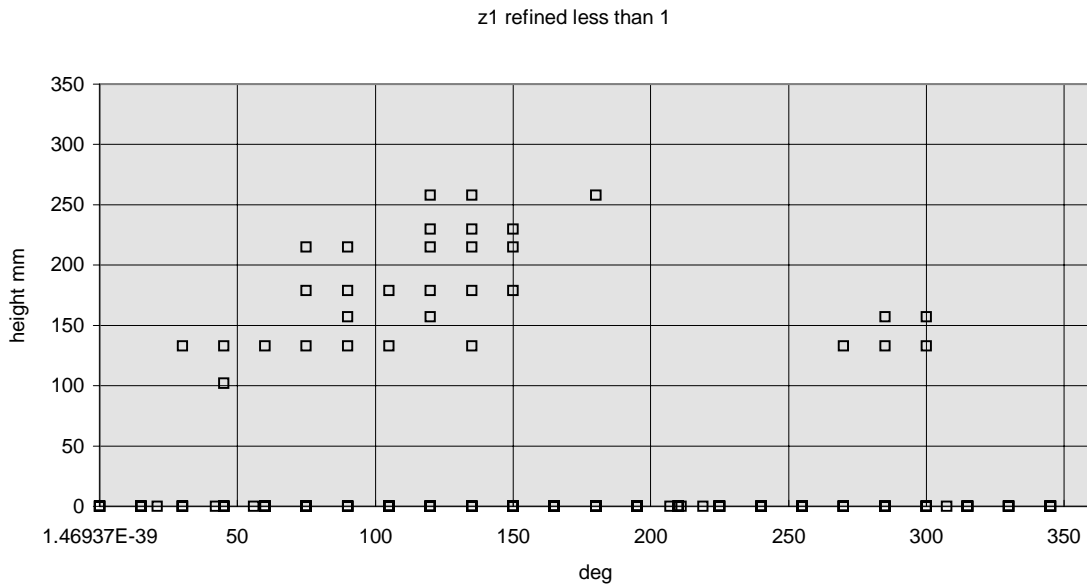
These frames do not have very good heat dissipation, compared to the partial fibreglass sockets. Most are made with cut outs in the outside socket. By mapping the stress distribution of the fibreglass socket, the region with least

stress can be found. These regions are the best regions for cut outs.

From figure 4.7 there are two regions where cut outs could occur. The first is between 75 and 150 degrees and heights of 130 and 240 millimetres on the front of the socket. The second is smaller and is located between 255 and 230 degrees at a height of 130 to 180 millimetres. These regions coincide with the presumed low stress regions which are cut out on double sockets already in use.

**Stress Mapping**

The vertical axis corresponds to the z axis, while the horizontal corresponds to the angular position.  
4.7





The material between the loading and the socket had little effect on the stress distribution. The finite element model used for the iterations had a good comparison to models with a more refined grid.

Finite element modelling is a convenient and accurate way to compare sockets under different conditions.

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## Glossary

- Alignment:** Socket z axis relative to the thigh.
- Anterior:** In front of the body.
- Brim:** Top lip of the socket.
- Distal:** Furthestmost part from the centre of the body.
- Extension:** Angled out from the body
- Fibreglass socket:** Socket comprising of a full fibreglass frame with a socket liner of a softer material.
- Flexion:** Angled inwards to the body.
- Frame:** Supports the socket within the prosthesis.
- Gait:** Manner of walking.
- Greater Trochanter:** Bone on the outside off the hip.
- Heel Strike:** When the foot touches the ground at heel strike.
- Hoop:** The top section of a partial fibreglass frame (ISNY style socket).
- Ischium:** Bone on the inside posterior of the hip.
- Lateral:** Outside of the body on a lateral plane.
- Medial:** Inside of the body on a lateral plane.
- Mid Stance:** Middle of the stance phase during gait.
- Model:** Finite element representation of the socket.
- Moment:** Bending moment.
- Node:** A point on the model representing a point on the socket.
- Phase:** Particular instance of gait.

**Polypropelene socket:** Polyethylene liner with a  
polypropelene frame

**Posterior:** Rear section of the body

**Prosthetic:** Artificial Limb

**Socket:** Interface component between the prosthesis and the  
stump.

**Stance:** Phase of gait when part of the foot is in contact  
with the ground.

**Swing:** Phase of gait when the foot is off the ground.

**Toe off:** When the foot leaves the ground, to start the  
swing phase of gait.