Ministry of Higher Education &
Scientific Research
Al-Nahrain University
College of Engineering

Design and Manufacturing of an Adjustable Below Knee Prosthesis for Children

A Graduation Project is submitted to the Prosthetics & Orthotics Engineering Department in Partial Fulfillment of the Requirements for the Degree of Bachelor of Science in **Prosthetics & Orthotics Engineering**

By

Aaiesha Mohammed Mazhar

Supervised by

Asst. Prof. Dr. Yassr Y. Kahtan

Dhul-Qu'dah 1443
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Supervisor Certification

I certify that this final year project entitled "**Design and Manufacturing of an Adjustable Below Knee Prosthesis for Children**" was prepared under my supervision Assist. Prof. Dr. Yassr Y. Kahtan at the Department of Prosthetics & Orthotics Engineering, Al-Nahrain University as a partial fulfillment of the requirements of Bachelor of Science in Prosthetics & Orthotics Engineering.

Kahtan

Asst. Prof. Dr Yassr Y. Kahtan

Supervisor

5 / 6 /2022

Ahmed

Prof. Dr. Ahmed A. A. Alduroobi

Head of the Prosthetics & Orthotics Department

26 / 6 / 2022

Committee Certificate

We certify, as an Examining Committee, that we have read this project entitled **"Design and Manufacturing of an Adjustable Below Knee Prosthesis for children"** and examined the student **Aaiesha Mohammed Mazhar** in its content and found it meets the standard of final year project for the degree of Bachelor of Science in **Prosthetics & Orthotics Engineering**.

Signature

Name: Asst. Prof. Dr. Wajdi S. Aboud

(Chairman)

26 /6/2022

Signature

Name: Lec. Dr. Hayder A. Sallal

(Member)

26 /6/2022

Signatur

Name: Lec. Dr. Fahad M. Kadhim

(Member)

26 /6/2022

Signature

Name: Lec. Yousra S. Kareem

(Member)

26 /6/2022

Approval of the College of Engineering/ Al-Nahrain University.

ABSTRACT

Number of amputees in Iraq is large and capable to increasing, and this is due to congenital defects, war effects, explosives and accidents. Usually a large number of amputations below the knee (BK) usually represent a large percentage of amputations that affect a person. The purpose of this research is to design , manufacture and test the adjustable below - knee prosthetic limb for a child with 8 years old suffering from transtibil (BK) amputation , which can be adjusted to facilitate the growth of a child. The procedure for designing , manufacturing and testing the prosthetic lower limb was done through three stages . The first stage was to perform a theoretical and numerical design for the adjustable socket and pylon through 3D SolidWorks program to design the parts, then ANSYS software program to analyze this model and calculate the equivalent (Von Mises) stresses , safety factors , and deformation distribution . The second stage was the experimental work , using the composite materials for the socket lamination (The adjustable model) that consisted from 10 layers (10 layers of perlon + 300 g acrylic resin). While the aluminum material (6061-T6) selection for the adjustable pylon model. Tensile test has been carried out to obtain the mechanical properties for the two types of socket models and pylon materials. The third stage is assembling the BK prosthetic parts and checking the alignments that accomplished in Al aibtisam Center for prosthetics after that performed the interface pressure (IP) test and ground reaction force (GRF) test . The tensile test results for (acrylic resin + berlon) were σult= 30.424 MPa, σy =16.089 MPa and E= 1.1027733 GPa for the first proposed design and σult= 26.843 Mpa, σy =15.814 MPa and E= 1.225128 GPa for the second proposed design. The tensile test results for the pylon were σult= 95.049 Mpa , σy=75.609 Mp and E= 6.985216 Gpa. The maximum pressure in the socket for below the knee was 394 KPa in the lateral area of the leg. The

design provides more comfort and lighter weight in addition to the possibility of increasing the size of the socket and pylon during growth, as well as the ease of dislocation and wearing the socket. While the lowest pressure value was in the posterior of the socket, reaching 101 kPa. Walking tests were good. Central pressure in the foot of the prosthesis was 0.18 Mpa.

The design provides increased comfort and lightness as well as the ability to increase bore size and pigmentation during growth, comfort, cell and bore wear.

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Symbols	Meaning
PETG	Polyethylene terephthalate glycol
Bk	Below knee
SACH	Solid Ankle Cushion Heel
RP	Rapid prototyping
ISO	International Organization for Standardization
CAD CAM	Computer aided manufacturing Computer Aided Manufacturing
TFA	Transfemoral amputation
RL	Residual limb
CNC	Computer Numerical Control
EVA	Ethylene-vinyl acetate
ASTM	American Society for Testing and Materials
PVA	Polyvinyl alcohol
GRF	Ground reaction force

List of Nomenclature

Symbols	Meaning	Unit
σult	Ultimate strength	MPa
F	Force	N
A	Area	mm^2
σy	Yield strength	MPa
E	Young's Modulus	GPa

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CHAPTER 1

INTRODUCTION AND LITERATURE REVIEW

1.1 Background

The prosthetic socket is the equipment that joins the residual limb (stump) to the prosthesis. The socket is made just for one patient, according to the condition and shape of the residual limb [1]. The socket is the most important part of the prosthesis, as it is the only part that is in contact with the stump. Therefore, many requirements are taken into account when designing the socket in terms of measurements, appropriate materials, knowledge of the patient's condition and determining what is appropriate for the amputated person. Many researchers and studies related to the study That design of prosthetic limbs is currently moving towards the manufacture of a lightweight, variable-size socket that is characterized by durability and its ability to bear weight applied to the socket from the body and remaining stump, in addition to bearing the environmental and working conditions of the person who wears it. Socket should be suitable with the stump of any type, such as being bony or fleshy end. Studies are lacking in finding solutions and designs for the socket that are suitable for children, changeable, light in weight, can be worn by several people of different ages and different weights. And because the socket is the most expensive part and its prices are high, also difficulty design socket that is suitable for all groups and sizes without needs to replace every period due to the change in size for the stump, and the reasons may also be due to the damage to the socket due to much use. Therefore, in this research, attempts will be presented to find a solution to the problem of a fixed-size socket through the design of an adjustable-sized socket for below knee amputation in 3D design programs for children in different age approximately from 5 to 15 years old. The adjustable socket characterized by light weight, ease of change the size and ease of wearing by the patient. Efforts will be submitted to design the socket from suitable material to be an affordable price and acceptable shape, also soft insert will be design according to each patient in proportion to the stump.

1.2 Definition of Prosthetic Apparatus

Lower limb prosthetics are devices designed to replace the function or appearance of the missing lower limb as much as possible. The basic categories of lower limb prostheses are, by the amputation height, transtibial (TT) and transfemoral (TF) prostheses. Typical transtibial prosthesis consists of a prosthetic foot, tube adaptor, and transtibial socket; a transfemoral prosthesis consists of a prosthetic foot, tube adaptor, prosthetic knee joint, and transfemoral socket [2].

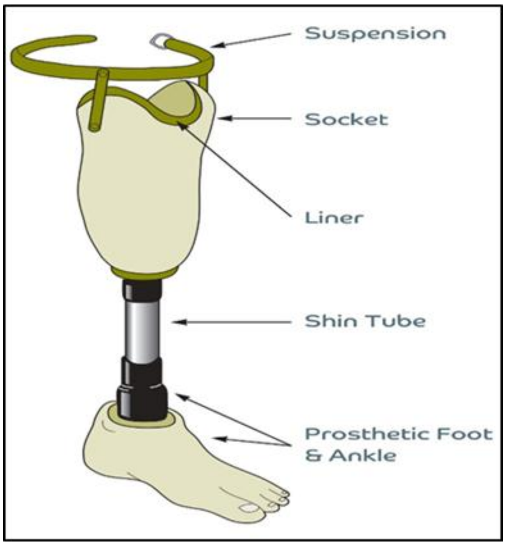


Figure 1-1: Below knee prosthetic device [3]

Prostheses are either placed externally or are implanted within the defective body part. Typically, a team of healthcare professionals chooses the “right” prosthesis and guides their patient through a rehabilitative phase of learning how to live with their new artificial device. Each patient’s reason for needing a prosthetic may vary, but it could involve [1]:

1. Birth defects (congenital deformity)
2. Trauma (i.e. accidents, military combat)
3. Cancer
4. Circulatory problems from diabetes or atherosclerosis that culminate in an amputation.

1.3 Below Knee Amputation (Transtibial)

The term of below knee amputation refers to the level of missing part of the lower extremity involves removing the foot, ankle joint, distal tibia, fibula, and soft tissue structures. Transtibial amputation is one of the most frequently amputations that give better functional outcomes than above knee amputations. The technique of long posterior flap becomes one of the most frequently recommended processes; however the surgeon must be able to dealing with the residual limbs and performing amputations needs to be comfortable with the other techniques as well [4 and 5].

1.4 Below Knee Socket Designs

The field of prosthetics has a long and rich history. It dates back to Ancient Egypt [6 and 7]. The oldest practical prosthesis known to man is a toe made of wood and leather, which archaeologists found attached to the 3000-year-old remains of an Egyptian noblewoman (950-710 B.C.). And one fascinating aspect of this history is that of the check socket in particular. Prosthetic socket materials throughout history nowadays, prosthetic sockets are made of innovative plastic and silicone materials, but this has not always been the case. The first Trans femoral sockets consisted of wood and leather and had a conical shape on the inside. The basic conical socket later advanced into a quadrilateral socket. This type could comfortably bear the amputee's weight on the ischial gluteal area of the residual limb and came with an air valve and an airspace at the bottom. Many patients had good results with these sockets, but many also suffered from edema and skin problems. Later on, prosthetists introduced wooden sockets that had full contact with the stump. This new shape decreased the problems with edema and helped to keep the prosthesis in place. Wood and leather were the materials of preference for a long time. And while they enabled prosthetists to make good prosthetic sockets, they were not ideal. Wood is difficult and time-consuming to shape, and it changes with the humidity levels of the environment; leather is a challenge to keep clean. Thus, prosthetists started experimenting with plastics and resins, which made the prosthetic fabrication process more efficient. Moreover, they allowed prosthetists to check the fit of the socket. The first check socket fitting did not being used until the 1950s [8]. Before that, prosthetists did not yet consider them a useful step in the creation of the perfect prosthesis. One of the reasons for their relatively late application was the controversy that surrounded check sockets. At the time, many prosthetists looked down upon the notion of a check socket .Figure 1-2 shows the check socket.

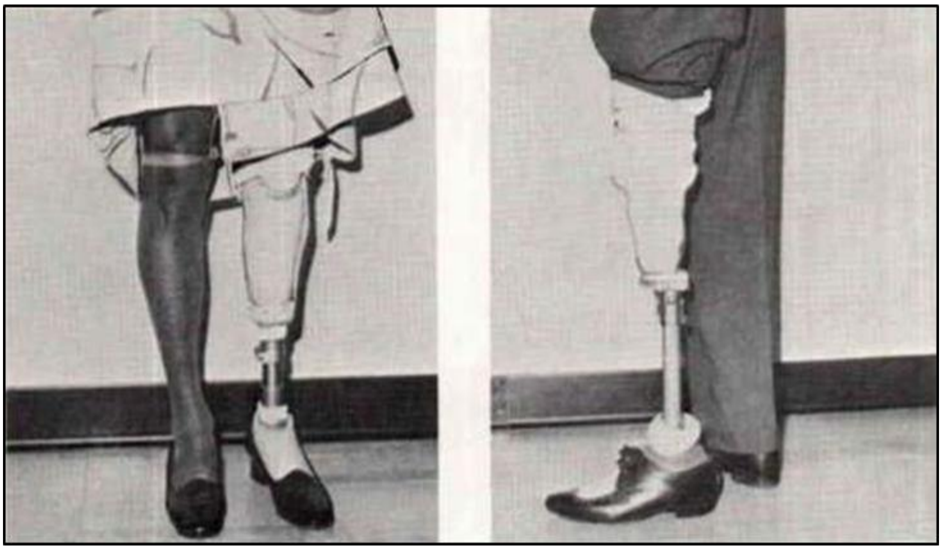


Figure 1-2: Check socket [9]

They believed it to be a redundant step in the creation of a prosthesis. Their thoughts are, if the perfect techniques were used from the beginning, than the prosthetic socket would fit perfectly and therefore, the sockets would not need adjustments. New socket materials in the 1950s and 1960s, this attitude changed. With the introduction of new transparent socket materials, prosthetists did start to see the value of using a test socket. The clear sockets allowed prosthetists to immediately see the areas that needed adjustment in the definitive socket, such as areas of redness or areas with lack of contact. The first clear check sockets consisted of Plexiglas. However, as this material was so tedious and time-consuming to shape, researchers started to look for alternatives. They tried making sockets from acrylic and polyester resins, but these were equally difficult to work with and quite pricey. They soon got replaced with plastic laminates and thermoplastics, which were less expensive and easier to form [6].

1.5 Types of Below Knee Socket

Below knee prosthetic device has many types of socket according to the type of weight-bearing area [10]. Every period, new methods are developed for the manufacture of the socket and how to design it, as well as the trim line and its location in relation to the weight applied to the socket [5]. Below knee sockets types involve:

1.5.1 Socket PTB (Patellar Tendon Bearing)

The weight-bearing takes place below the patella, at the patellar tendon. The suspension is generated by a belt that is tightened around the distal part of the thigh .Figure 1-3 shows the PTB socket with cuff suspension. The tension of that belt limits the blood and lymphatic circulation; moreover, after long term use results in muscle atrophy and other related problems [5 and 11].

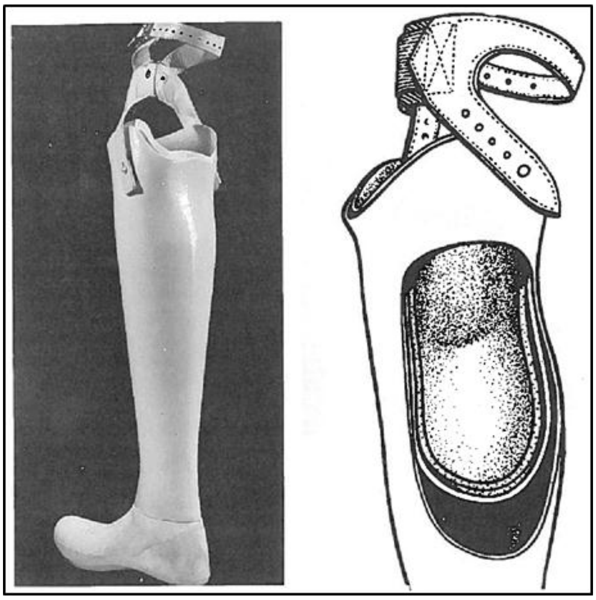


Figure 1-3: PTB socket with cuff suspension [12]

1.5.2 Socket PTB-SC (patellar tendon bearing supracondylar)

The weight-bearing takes place below the patella, at the patellar tendon [12]. The suspension is generated at the medial and lateral areas of the femoral condyles. Compared to the PTB socket with belt suspension, this design does not produce problems of blood circulation or atrophy. For the moment, this type is used worldwide as most basic design for prosthetic fitting of medium and long stumps .Figure 1-4 shows the PTB-SC Socket (patellar tendon bearing supracondylar).

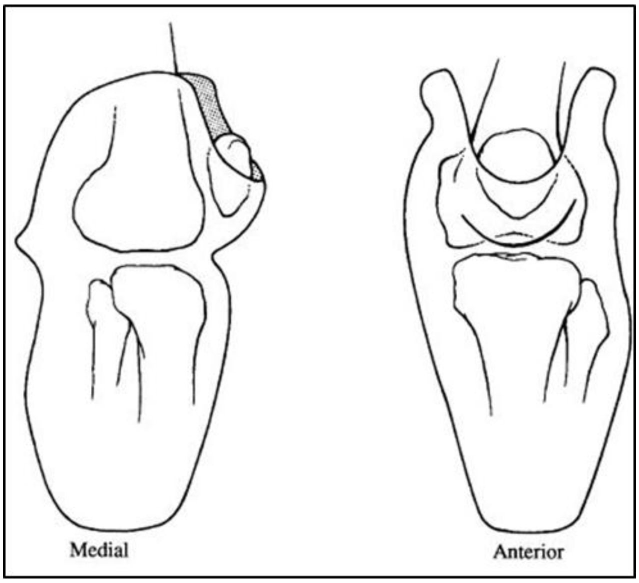


Figure 1-4: PTB-SC Socket (patellar tendon bearing supracondylar) [13]

1.5.2 Socket PTB-SCSP (Patellar Tendon Bearing Supracondylar Suprapatellar)

The weight-bearing takes place below the patella, at the patellar tendon [12]. The suspension is generated at the medial and lateral areas of the femoral condyles and at the suprapatellar area. This type is indicated for short stumps, as well as in cases of antero-posterior instability in the knee .Figure 1-5 shows the PTB-SCSP Socket (patellar tendon bearing supracondylar suprapatellar).

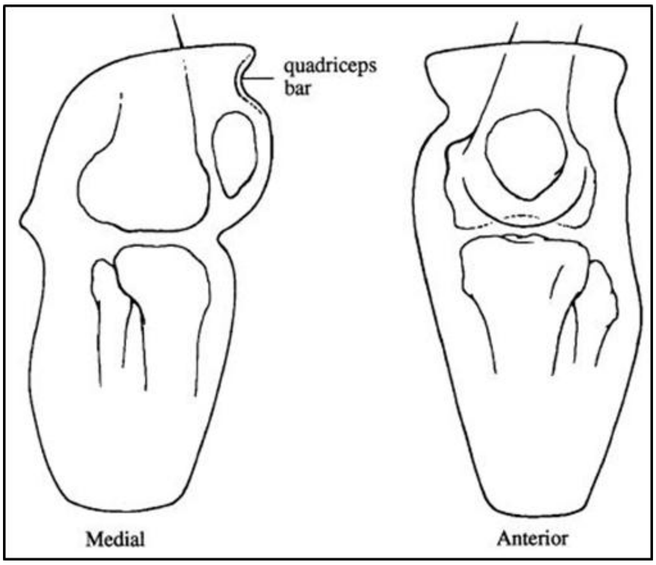


Figure 1-5: PTB-SCSP Socket (patellar tendon bearing supracondylar suprapatellar) [13]

1.6 Literature Review of Prosthetic Socket

Board, W. et al 2001 [14]. A method of preventing this volume loss and maintaining a good fit was developed. A vacuum (-78 kPa) was drawn on the expulsion port of a total surface-bearing suction socket to hold the liner tightly against the socket. Stump volume of 10 trans-tibial amputees was measured prior to and immediately after a 30 minute walk with normal and vacuum socket conditions. Eleven (11) subjects were drawn from a pool of trans-tibial unilateral, traumatic amputees who could walk 30 minutes on a treadmill: 11 were used in the pistoning testing and 10 were used for the volume and gait testing. Subjects were fitted with urethane liners and suspension sleeves, and acrylic copolymer (PETG) test-sockets. Their stumps were vacuum-cast with plaster, and these casts were used to construct sockets and liners. Liners were roughly undersized by 10% and sockets by 4%. With the vacuum, stump volume increased ($p = 0.007$) an average of 3.7% (30ml) during the 30-minute walk. In the normal condition, stump volume decreased ($p = 0.000$) an average of 6.5% (52ml). The liner displaced 0.4cm less (p = 0.000) from the socket when the extraction force was applied under the vacuum condition. The tibia displaced 0.7cm less (p = 0.000) from the socket reference point in the vacuum condition when loaded.

Jia, X., Zhang et al 2004 [15]. In this study, the kinematic data of the lower-limb and prosthesis and the ground reaction forces applied on prosthetic foot during walking were measured using a Vicon Motion Analysis System (Oxford Metrics, UK) and a force platform (AMTI, USAThe motion of the limb and prosthesis was monitored using a Vicon motion analysis system and the ground reaction force was measured by a force platform. Equivalent loads at the knee joint during walking were calculated in two cases with and without consideration of the material inertia. A 3D nonlinear finite element (FE) model based on the actual geometry of residual limb, internal bones and socket liner was developed to study the mechanical interaction between socket and residual limb during walking. To simulate the friction/slip boundary conditions between the skin and liner, automated surface-to-surface contact was used. The prediction results indicated that interface pressure and shear stress had the similar double-peaked waveform shape in stance phase. The average difference in interface stresses between the two cases with and without consideration of inertial forces was 8.4% in stance phase and 20.1% in swing phase. The maximum difference during stance phase is up to 19%.

Lee, W. et al 2004 [16]. Finite element method has been identified as a useful tool to understand the load transfer mechanics between a residual limb and its prosthetic socket. Zachariah and Sanders used an automated contact method to simulate contact between the limb and the socket in previous model. The limb was allowed to slide if the shear stress exceeded the frictional limit. The mechanical properties of the materials were assumed to be linearly elastic, isotropic and homogeneous. Young's modulus was 200 kPa for soft tissues and 10 GPa for bones. Poisson's ratio was assumed to be 0.49 for soft tissues and 0.3 for bones. The socket was assigned with Young's modulus of 1500 MPa and Poisson's ratio 0.3, resembling the mechanical property of polypropylene homopolymer. It was found that peak normal and shear stresses over the regions where socket undercuts were made reduced and the stress values over other regions raised in the model having the simplifying assumption.

Herbert, N. et al 2005 [17]. This paper investigates the use of a cheaper, low end RP technology known as 3D printing. Investigation was an initial approach to using a technology that is normally associated with producing prototypes quickly, some of which could not be manufactured by alternative means. Under normal circumstances, these printed components are weak and relatively fragile. However, comfortable prosthetic sockets manufactured with 3D printing have been used in preliminary fittings with patients.

Shasmin, H et al 2007 [18]. The pylon, which commercially made from Stainless Steel or Titanium is used to connect the socket to the ankle-foot assembly. Today, transtibial amputees have an equal number of options available in the market. However, these high-tech prostheses can cost several thousand dollars in the West and amputees from low-income countries just could not afford to buy them in this study, the commercial pylon (shank) will be replaced by using the bamboo. Setup used for mechanical testing of the prosthesis. Satisfies the ISO 10328 standard. Gigantocloa Ligulata and Bambusa Heterostachya have been selected based on the FRIM mechanical tests on variety species of bamboo. The bamboos have high tensile properties, 341-530 mpa which is appropriate for the pylon characteristics. However, the natural bamboo cannot stand for more than 3 years due to its deformation. This development of new pylon materials is based on the physical properties of the bamboo, tested on the virtual forces and Universal Testing Machine applications.

N A Abu Osman, et al 2010 [19]. The purpose of this investigation was to vary the load on the patellar tendon bar and to study the subsequent effect this has on the pattern of the pressure distribution at the stump–socket interface. Ten male subjects from the Southern General Hospital in Glasgow, UK participated in this study. Measuring systems utilising strain gauge and electrohydraulic technologies were designed, developed and constructed to enable pressure measurements to be conducted. One transducer, the patellar tendon (PT) transducer, was attached to the patellar tendon bar of the socket such that the patellar tendon bar was capable of being translated by ±10 mm towards or away from the tendon. The results of this study showed that the position of the patellar tendon bar had no significant effect on the pressure distribution around the socket indicating that it is an unnecessary feature, which, we propose, may be eliminated during manufacture of a trans-tibial socket.

Lenka, P et al 2011 [20]. The objective of this work was to analyze in a parametric study for optimum design solution of prosthetic socket material by finite element method. A realistic three-dimensional finite element model of the PTB socket was developed to find out the stress distribution pattern under physiologically relevant loading condition during normal walking. All structural materials used in the analysis were assumed to be linearly elastic, homogeneous and isotropic. The results summarized that integrating local compliant features within socket wall can be an effective methods to distribute maximum stress areas and also to relief contact pressure between the stump and socket. The design solution obtained from the results can be used as a reference to choose material for fabrication of socket in developing countries like India, depending on the weight, strength, cost and availability. The study explored further future scope for parametric analysis, investigating the effects of socket stiffness, rectification scheme and materials on the interfacial stress distributions.

Gholizadeh, H et al 2012 [21]. The objective of this study was to compare the pistoning effect of Seal-In X5 and Dermo Liner by using Vicon Motion System. Six transtibial amputees, using both the Iceross Seal In® X5 and the Iceross Dermo Liner, participated in the study The vertical displacement (pistoning) was measured between the liner and socket in single limb support on the prosthetic limb (full-weight bearing), double limb support (semi weight bearing), and non-weight bearing on the prosthetic limb, and also under three static vertical loading conditions (30 N, 60 N, and 90 N). The results demonstrated that the pistoning within the socket when Seal-In X5 was used, decreased by 71% in comparison to the Iceross Dermo Liner. In addition, a significant difference between the two liners under different static conditions was found (pb0.05).

Keith cornell 2014 [22]. This study is a patent that includes design of an adjustable socket for transfemoral prosthesis consisting of a rigid partial stent, a non-flexible stent, and an additional adjusting device. This socket can be used for several types of amputations as Transtibial and in addition to the Trans femoral amputations. The present invention had a number of advantages over conventional sockets. The present invention provided comfortable weight bearing and a suitable stability for the control of the prosthesis in space. Bore size control of the present invention had unprecedented with the use of an adjustable elastic support or clothing.

Jason T. Kahle et al 2016 [23]. Traditional rigid sockets do not accommodate volume fluctuations. Poor fit can cause skin ulcerations and infection and may lead to revision amputation [24]. The objective of this pilot clinical case study is to compare the efficacy of a standard of care (SOC) containing ischial tendon (IRC) with an adjustable Trans femoral interface socket. This design features a high proximal rim that centrally contains the pubic ischial piece which is the SOC. To simulate volume loss (VLOSS), the five-layer RL. In this case study, the care socket criterion was inferior to the comparator adjustable TFA in subjective outcomes and performance. This case study demonstrates the need for a well-designed clinical trial using outcome measures comparing the effectiveness of an adjustable socket prosthesis interface with a standard of care. Clinical trials and evidence comparing socket functional outcomes related to volume variability are lacking.

Samuel terrazas quezada 2017 [25]. The aim of this study redesign of an adjustable socket for transtibial amputation. First phase of this study was the creation of various prototypes to improve the existing mechanism for adjusting the prosthetic socket. The adjustment system was redesigned to improve ease of use and provide a more stable coupling between the socket and the remaining prosthetic components. The second phase consisted of creating a new socket wall design that implements struts with a flexible inner socket. The flexible socket provided a higher probability of producing a more comfort use due to the reduced load applied on the pressure sensitive areas, also the weight analysis saw a reduction for the redesign.

Kay Mitton, et al 2017 [26]. This study described the problems of prescribing a prosthetic socket in a left transfemoral amputee secondary to chronic patella-femoral instability compounded by complex regional pain syndrome [27]; symptoms recurred in the residual limb, presenting mainly with edema. Due to extreme daily volume fluctuations of the residual limb, a conventional, laminated thermoplastic socket fitting was not feasible. An adjustable, modular socket design was trialed. The socket could be worn for a 60 minutes physiotherapy session without any problems of fit, distal end strangulation or skin irritation. She restarted an intensive rehabilitation program focusing on gait re-education, hip and trunk muscle strengthening. The patient was able to wear the prosthesis for 8 h daily and to walk unaided indoors and outdoors. She is able to drive her adapted car with the prosthesis on, attend medical appointments without a wheelchair and complete a food shop. An adjustable, modular socket design accommodated the daily residual limb volume fluctuations and provided a successful outcome in this case.

Timothy Dillingham et al 2019 [28]. This research focused in its study on evaluating the utility of an immediate fit modular prosthetic system (IFIT Prosthetics).this system dealt with a prospective study that includes training for one group for a period of two weeks to evaluate the work of this socket. The design of socket materials for Transtibial amputation consisted of Soft insert, silicone sleeve suspension. The results were satisfactory by the participants wearing this socket and the tip was suitable for them. During the two weeks they were given a stepwise wearing schedule in which outcome measurements were completed during the two weeks and focused on 3 main issues to assess socket functioning through self-reported satisfaction, gait biomechanics, and peak intra-sinus pressure. The number of participants was 26. The results of the participants were somewhat different in terms of self-satisfaction, Intra-socket peak pressures were significantly lower for the IFIT prostheses overall. No falls or limb ischemia were reported. Socket liners and prostheses were adjusted for patients who had redness and skin issues. No mechanical failures occurred the skin issues encountered with the IFIT socket were readily addressed with socket modifications and local dressings to reduce skin friction and pressure.

Khudhyear abbas, h., 2019 [29]. The purpose of this research is to design, manufacture and test the adjustable above-knee prosthetic limb for a child suffering from transfemoral (TF) amputation, which can be adjusted to facilitate the growth of a child for a duration of (1-2 year). The procedure for designing, manufacturing and testing the prosthetic lower limb was done through three stages. The first stage was to perform a theoretical and numerical design for the adjustable socket and pylon through 3D scanning technology, and then using SolidWorks program to modify and add the straps for this design to adjusting, ABAQUS/CAE and ANSYS software program to analyze this model and calculate the equivalent (Von- Mises) stresses, safety factors, and deformation distribution. Using the composite materials for the socket lamination (The adjustable model). While the aluminum material selection for the adjustable pylon model the suggested adjustable socket and pylon model have increased the life of used by 300% and 260% respectively.

Ibarra Aguila, S. et al 2020 [30]. This paper presents a socket interface pressure (SIFP) system to compare the interface pressure differences during gait between two different types of prosthetic sockets for a transtibial amputee. The socket interface pressure (SIFP) device herein proposed is able to measure, both wirelessly and in real time, a pressure profile for the interface between the socket and the residuum of a lower limb amputee during gait testing. The comparative study between the SIFP device and the F-Socket shows that the performance of the two devices in measuring interface pressures is comparable due that both of them show a similar pressure change pattern when loads were applied to an artificial residuum, thereby simulating the walking protocol. While the F-Socket system has many more active sensors, it unfortunately does not conform well to the irregular shape of the residuum as does the SIFP device.

Owen, M. et al 2020 [31]. The goal of the present study is to evaluate the strength at failure and the failure mechanism for 3D-printed transtibial sockets, thermoplastic transtibial check sockets, and carbon-fiber definitive laminated transtibial sockets. Three clinically available materials (carbon fiber, PETG thermoplastic, and 3D print polylactic acid [PLA]) were used to produce identically shaped sockets. Both designs of carbon-fiber sockets had higher US values at failure than thermoplastic sockets and 3D-printed PLA sockets. Thermoplastic sockets had a slightly higher average US than 3D-printed sockets, but 3D-printed sockets exhibited a greater strength-to-weight ratio. Four distinct socket failure mechanisms were determined. Failure mechanisms and Us values differed for all socket types. Carbon-fiber sockets exhibited the highest US, but 3D-printed PLA sockets showed comparable strength to current thermoplastic sockets. Although the ISO 10328 testing standard was sufficient to complete these evaluations, the method lacked some socket-specific measures and loading conditions that could have improved comparisons between socket types.

Muhsin J. Jweeg et al 2021 [32]. The four best lamination types and suitable Nano-material percentages were used in this work to develop a numerical investigation of a below knee prosthesis, facilitating the investigation of the effect of different Nano-material types and weight fractions on the stress and deformation in below-knee socket prosthetic structures manufactured from composite laminated materials with various reinforcement fibres. Numerical analysis using the finite element method was adopted to estimate the Von Mises stresses and deformation behaviours for the below-knee prosthetic structures, with the mechanical properties of the relevant composite materials. A SOLIDWORKS program was used to generate an analysis BK socket. The comparison of different Nano-material types and weight fractions suggested that the best Nano-material was sio2 at a weight fraction of about 2% for the sample 2 Perlon+2 Kevlar+1 Perlon+2 Carbon+1 Perlon+2 Kevlar+2 Perlon, where the stress for the socket was reduced by about 40%, with a reduction in the deformation of about 38%.

1.7 Concluded Remarks

Previous studies focused on the design of sockets that can be adjusted in size to suit the changes of the stump. Weight and cost goals were also set as the focus of the researchers' interest, but no sockets were manufactured specifically for children and were successful in terms of adapting to the stump and the change in size for them, therefore in this research Emphasis will be placed on designing special sockets for children of variable size and suitable for approximately an average age of 5 to 15 years.

1.8 Aims of Research

1. Design and manufacturing an adjustable below knee prosthetic limb for children with changeable size.
2. Light weight.
3. Low Cost.

CHAPTER 2

THEORETICAL CONSIDERATIONS

2.1 Introduction

In this chapter, the traditional Transtibial limb, material, gait cycle and new adjustable prosthetic limb design are prescribed.

2.2 Traditional Below Knee Prosthetic Limb

The prosthetic socket is the main connection between the residual limb and the prosthesis. It is the socket that must distribute the forces through the residual limb. The more intimate the socket the better the function and comfort of the prosthesis [33]. Figure 2-1 shows the Transtibial prosthetic limb.

Figure 2-1: Below knee prosthesis [33]

2.3 Below Knee Prosthetic Limb Components

A prosthesis is composed of a number of components that work together as a single device which is specific to each person. The following is an explanation of each element of the prosthesis [34 and 35]:

2.3.1 Suspension

It is the part which holds the residual limb into the socket. Straps are used as support system to hold the socket into place. There are various types of suspension supracondylar cuff most common type, waist belt, cuff strap, thigh corset, and vacuum suspension.

2.3.2 Socket:

The socket is used to contain the residual limb (amputated limb) and transfer the weight of the body to the rest of the prosthesis, this may also contain liners to act as padding and provide suspension.

2.3.3 Socket Adaptor:

Used to connect the socket to the other components of the prosthesis and used to align the prosthesis. May come in a number of configurations.

2.3.4 Tube Clamp Adaptor:

Connects the socket to the pipe, used to align the prosthesis.

2.3.5 Pylon:

Used to transfer the weight of the body, must be adjusted to obtain the proper height of the prosthesis, and used to align the prosthesis.

2.3.6 Endoskeleton Finish:

Covers entire prosthesis protecting internal components from moisture, dust and dirt.

2.3.7 Ankle:

Attaches foot to prosthesis, allows motion to assist in proper gait pattern.

2.3.8 Foot:

Provides base of support, transfers weight to ground, fits in regular shoes, adapts to ground surfaces.

2.4 Prosthetic Socket Materials

The material on prosthetic socket is very important as to meet the requirements in order to make them effectively functional. Prosthetic sockets were fabricated from materials such as leather, wood, latex and metal before the introduction of nowadays advanced resins such as composites and thermoplastics [36]. In the past, certain types of leather or wood were stretched, soaked, carved and stitched into the form of prosthetics. After dried, lacquered or sealed, it found to be more durable. Polyethylene, polypropylene, acrylics, and polyurethane has replaced the previous leather or wood due to uneconomical and hazardous effects on the environment [37]. Marks and Michael reported that the lightweight pylon which made from carbon fibre was the newest development in prosthesis manufacturing [38]. The revolution of carbon fibre reinforced polymer composites (CFRP) in structural design of artificial limb was enabled the amputee to participate in professional sport. Scholz claimed that it is able to attach an energy return system within lower limb prostheses by using extremely lightweight and high strength CFRP [39]. However, in developing countries, where the disproportionate number of amputee live with the causes by war and landmines, CFRP prosthetic found to be highly expensive and are not always reliable. Generally, polymers such as polyethylene, high-density polyethylene (HDPE), acrylics, polypropylene, and polyurethane are used for fabrication. Carbon and glass fibre composites, with application of acrylic resins, are largely used due to their superior properties. However, these are expensive and will create harmful gasses in manufacturing [40]. Although as glass fibre/high density polyethylene (GF/HDPE) and polypropylene plastic composites offer the less expensive prosthesis for commercial purposes. Currently they are receiving great attention at manufacturing reliability, mechanical performance, and expense of comfort.

2.5 Gait Cycle and Gait Phases

As the body moves forward, one limb act as source of support while the other limb advances itself to a new support site. Then the limbs reverse their roles. This series of events is repeated by each limb with reciprocal timing until the person's destination is reached. A single sequence of these functions by one limb is called a gait cycle (GC). Normal persons initiate floor contact with their heel (i.e., heel strike). Each gait cycle is divided into two periods, stance and swing also known as gait phases [41]. Figure 2-2 shows the Gait phases.

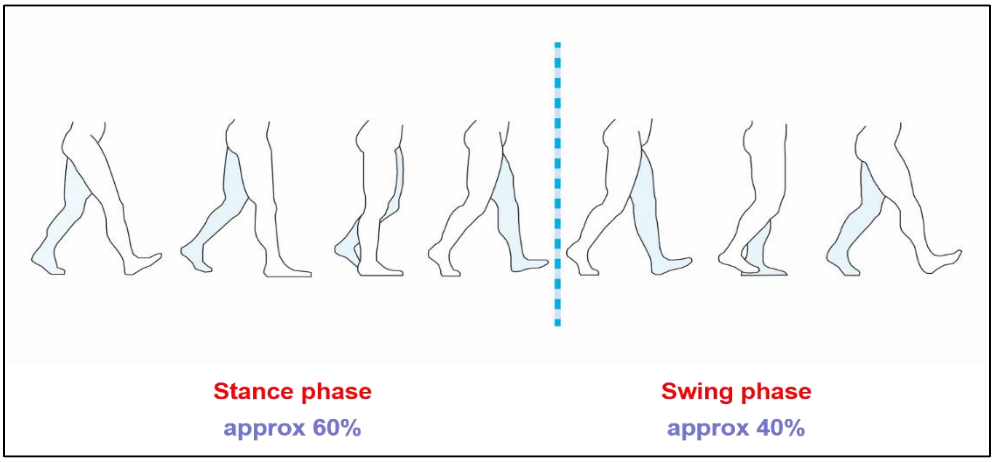


Figure 2-2: Gait phases [42]

Stance is the term used to designate the entire period during which the foot is on the ground. Stance begins with initial contact. The word swing applies to the time the foot is in the air for limb advancement. Swing begins as the foot is lifted from the floor (toe-off). Hence gait cycle can be defined as the time interval between two successive occurrences of one of the repetitive events of walking. The following terms are used to identify major events during the gait cycle. Figure 2-3 shows the events in a gait cycle.

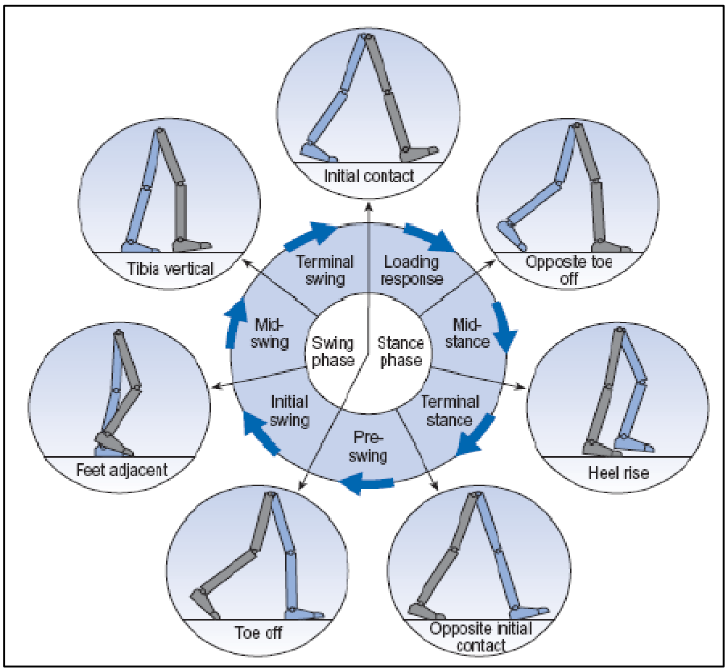


Figure 2-3: Events in a Gait Cycle [43]

In each gait cycle, there are thus two periods of double support and two periods of single support. The stance phase usually lasts about 60% of the cycle, the swing phase about 40% and each period of double support about 10%. However, this varies with the speed of walking, the swing phase becoming proportionately longer and the stance phase and double support phases shorter, as the speed increases [44]. The final disappearance of the double support phase marks the transition from walking to running. Between successive steps in running there is a flight phase, also known as the “float”, “double-float” or “non-support” phase, when neither foot is on the ground. The two main phases of gait include:

2.5.1 The Stance Phase

The stance phase is the period of the gait cycle when the foot is on the ground and bearing body weight. More specifically, it can be described as the period between the moments that the heel of the foot touches the ground (heel strike) until the moment that the toe-off occurs. The stance phase consists of five sub-phases [45]:

2.5.1.1 The Heel Strike

Initial response, contact response, or weight acceptance. In this sub-phase, the heel of the foot makes initial contact with the ground. It requires the body’s weight to be accepted by the leg making contact with the ground.

2.5.1.2 The Foot Flat

Loading response is the second sub-phase when the foot rolls forward until the entire plantar surface is in contact with the ground.

2.5.1.3 The mid-stance

Starts when the weight of the body is propelled forward, directly over the lower extremity, so that the greater trochanter of the femur is directly above the middle of the foot. At this stage, our entire body weight is being balanced over one leg.

2.5.1.4 Heel-off

The next sub-stage and includes lifting the heel off the ground. This is when we start to shift the body weight onto the contralateral leg.

2.5.1.5 Toe-off

Is the final stage of the stance phase and includes pushing the toes into the ground while the ankle plantar-flexes, creating forward propulsion.

2.5.2 The Swing Phase

The swing phase is the second phase of gait when the foot is free to move forward. It is described as the period between toe-off and heel strike. There are three sub-phases of the swing phase:

2.5.2.1 The Early Swing

Acceleration phase is the first sub-phase during which the foot is lifted from the ground. The ankle dorsiflexes and the knee flexes so that the foot and toes can be moved from the ground. The hip flexes to bring the leg forward, moving it directly under the body.

2.5.2.2 The Mid-Swing Phase

Is the second phase when the non-weight-bearing leg passes directly beneath the body and past the weight-bearing leg. At the same time, the trunk is moved forward so that the weight of the body is directly over the weight-bearing leg.

2.5.2.3 The late swing

Deceleration phase is the last sub-phase. The foot is moved to a position in front of the body, the knee extends and momentum decelerates. The lower limb is now ready for heel strike and prepares to accept the transfer of body weight, for the start of the next stance phase.

2.6 The Ground Reaction Force

Ground reaction force (GRF) is an important parameter in the study of kinetics response with respect to gait movement on various surfaces [45]. GRF has been widely used in numerous gait analyses related to running [46] and walking [47] on different runway. GRF response would be able to describe the weight distribution, body stability and kinetics adaptation against the running surface characteristics.

Ground reaction forces (GRF) develops during gait as a result of the force applied to the ground when the foot is in contact with it. GRF is equal and opposite to force that the foot applies on the ground. Since GRF is external force acting on the body during locomotion, it is of great interest in gait analysis. Figure 2-4 shows the ground reaction force in human.

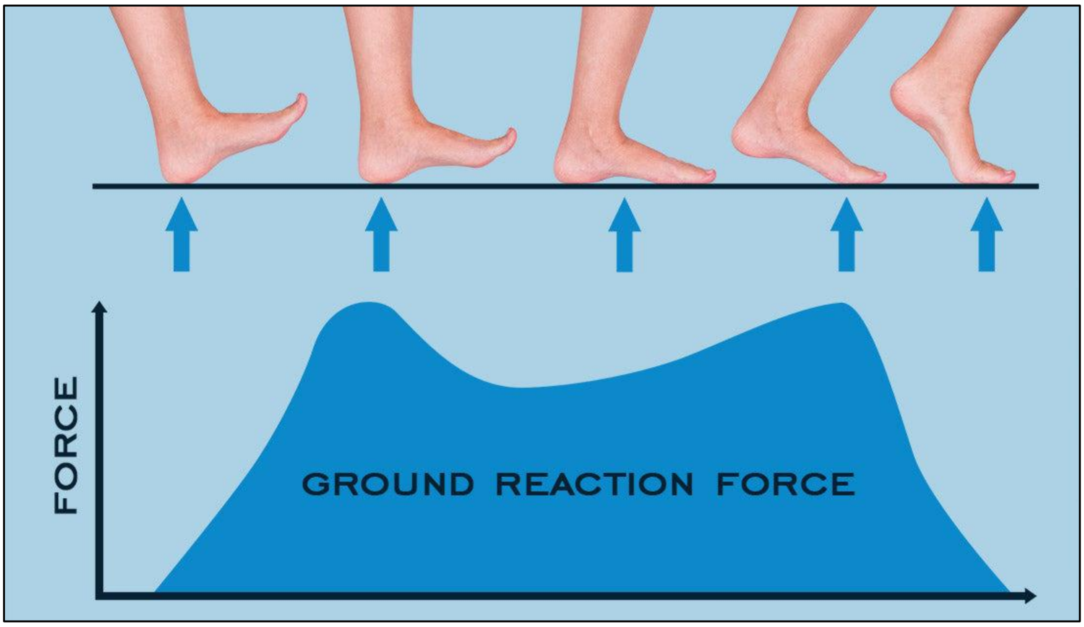


Figure 2-4: Ground reaction force [48]

2.7 Biomechanical Requirement

The magnitude of pressure between the stump and socket imposes major determinants of comfort, stability and any function. Prosthesis biomechanics studies the relationship between the socket and the stump such as the socket shape and alignment [49]. Also, the pressure from the weight bearing of the amputee on the socket or on each component of prosthesis and the reaction forces from the ground at heel strike, foot flat and push off Pressure is directly proportional to the forces applied.

It is expressed in formula ($\sigma = F/A$) which represents the average stress and F, A represents the force and the area over which the force is applied respectively. The comfort stability and function of prosthesis are achieved primarily by the application of certain biomechanical stress [50].

2.8 Prosthesis Alignment

The alignment of a lower-limb prosthesis depends on the spatial relationship of the socket relative to other components of the prosthesis, such as the knee joint and foot unit, and these alignments are critical to the successful utilization by the user. Optimal prosthetic alignment is achieved in three steps: bench, static, and dynamic alignment .Figure 2-5 shows lower limb prosthetic limb.

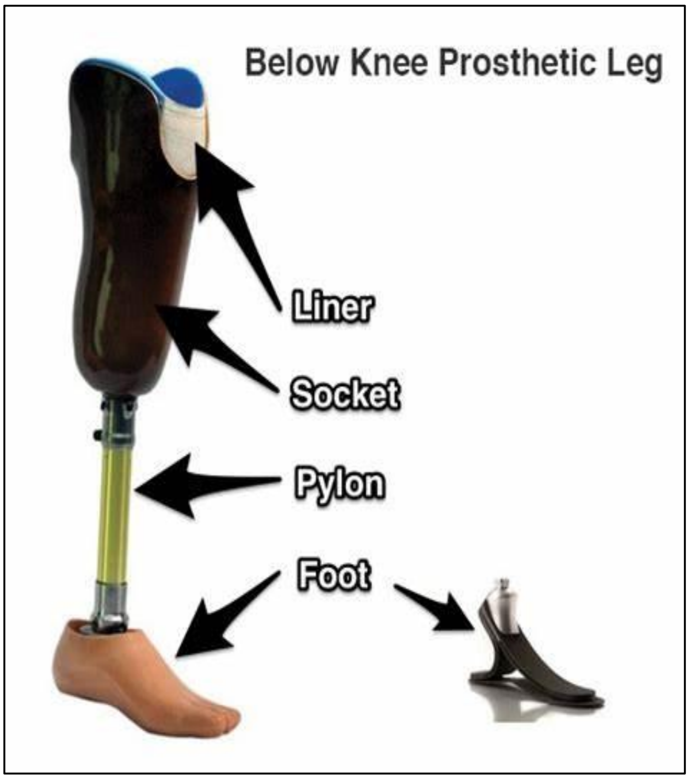


Figure 2-5: Lower limb prosthetic limb [51]

2.8.1 Bench Alignment:

Is the process of configuring the prosthetic components without the patient. Typically, manufacturers of prosthetic components such as knees and feet provide recommendations for how they should be attached together and to the prosthetic socket. The prosthetist will use this information in conjunction with his or her experience and knowledge of the patient to make an initial estimate of a well-aligned prosthesis.

2.8.2 Static Alignment:

Is done with the patient standing in the prosthesis with the body weight distributed equally on both extremities. An alignment Common types of alignment reference line include the Trochanter, Knee, and Ankle (TKA) line and the weight bearing (WB) line. Generally, the more posterior the knee centre is to the reference line, the more stable the prosthesis will be. However, the more posterior the knee is with respect to the reference line, the more difficult the knee will be to flex in terminal stance. A suitable balance between safety, stability, and control is desired.

2.8.3 Dynamic Alignment

Is the process of aligning the prosthesis based on observing the amputee walk while wearing the prosthesis. One main goal of dynamic alignment is to identify how the prosthetic limb is used and controlled by the amputee. Gait deviations that may be a result of misalignment are identified and addressed. The position and orientation of the knee, foot, and pylon are adjusted to achieve a more normal gait pattern [52] Figure 2-6.

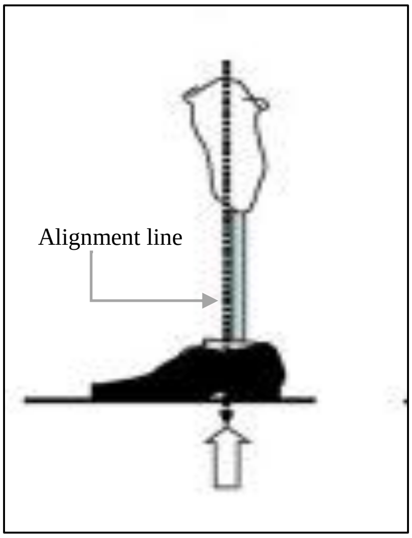


Figure 2-6: Alignment reference line [53]

2.9 The Numerical Designing for Adjustable BK Limb

In this project, new design for socket and shank is proposed. By using 3D Solidwork program, the work will be as below. Figure 2-7 shows the artificial with proposed designs:

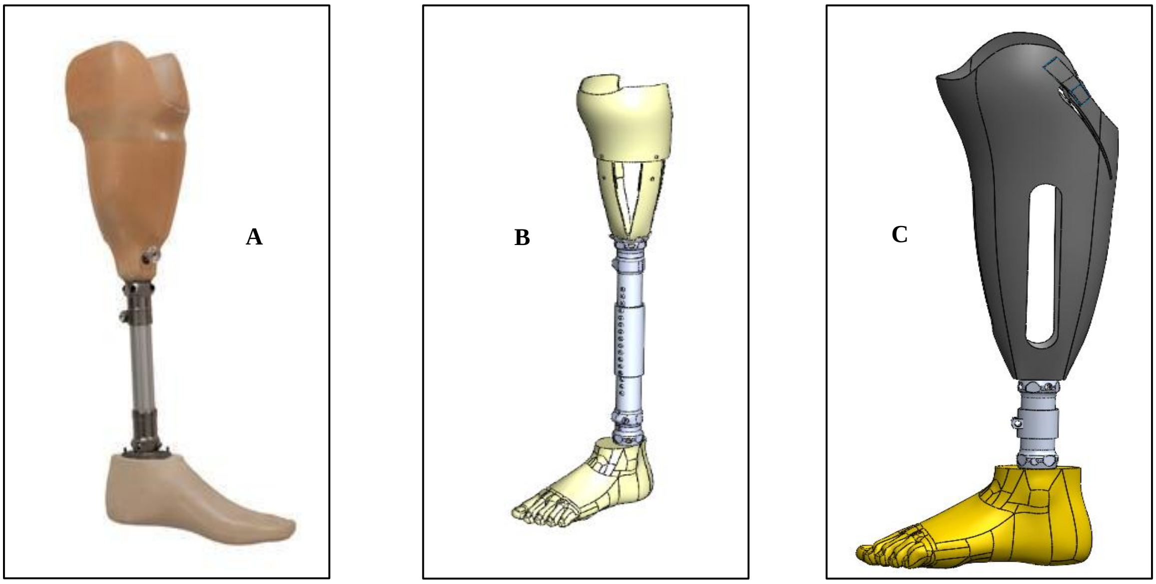


Figure 2-7: Below knee prosthetic limb, A: conventional BK limb [54]. B: first proposed design. C: second proposed design

2.9.1 Below Knee Socket Design

2.9.1.1 First Proposed Design

The first one created as two parts distal and upper to achieve both length changeable and diameter of the socket. The distal part of socket had been cut into four parts assembling together by a pieces of metal to give more length as shown in Figure 2-8.

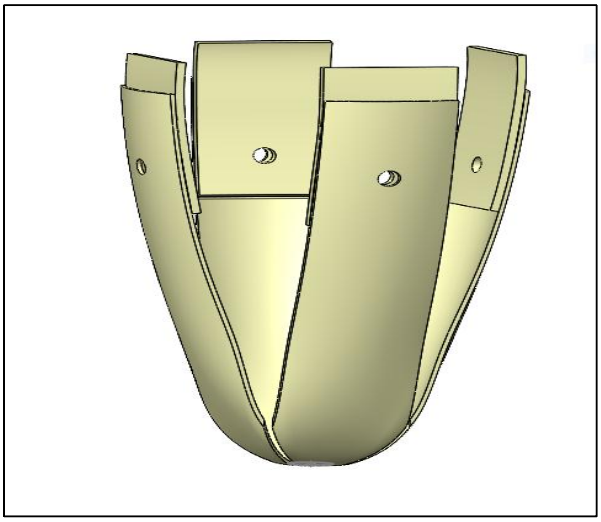


Figure 2-8: Distal part of proposed BK socket

The base of socket open and connecting to the adapter by bolts.

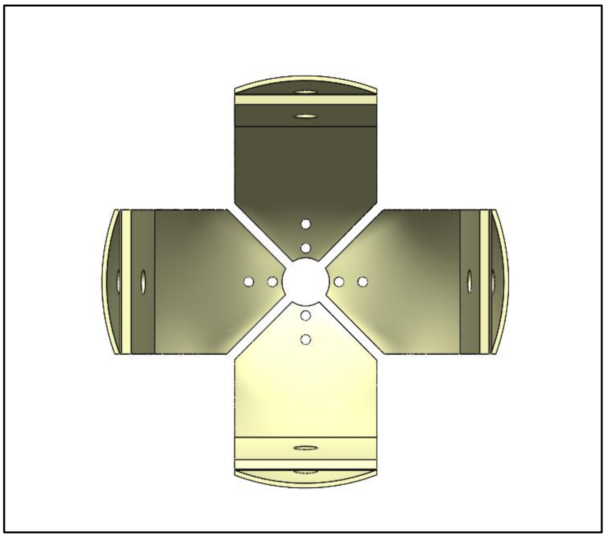


Figure 2-9: Distal base

The upper part will be the same as traditional BK socket as using for suspension and weight bearing as in Figure 2-10.

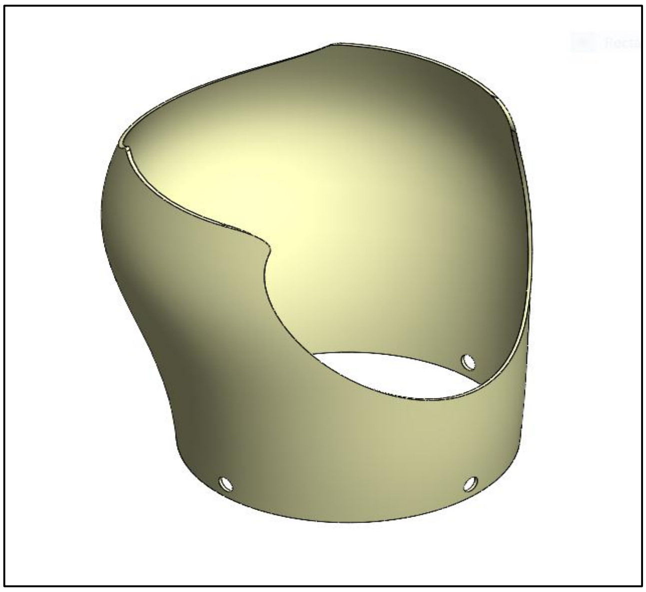


Figure 2-10: Upper part of proposed socket

Holes have been made to connect upper part with distal part as shown in Figure 2-11.

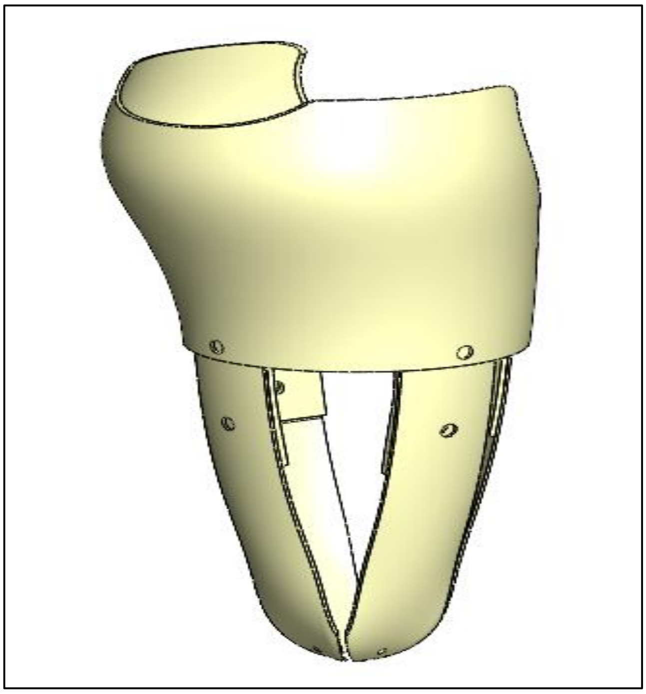


Figure 2-11: Full proposed socket

2.9.1.2 Second BK Socket Design

The socket was left uncut. Instead, 3 holes were made in the front and sides (medial and lateral) as shown in Figure 2-12.

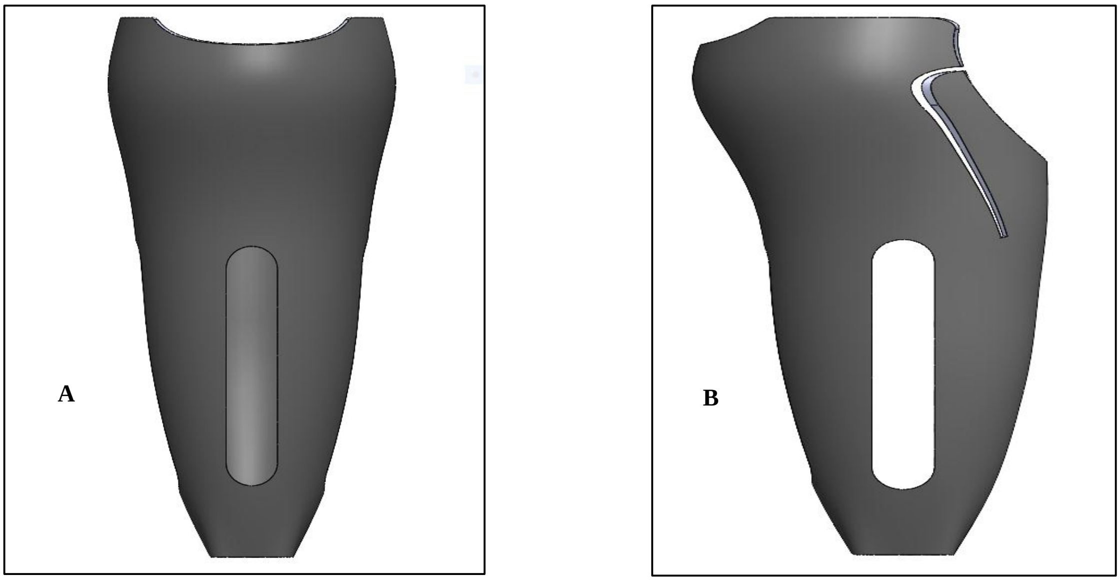


Figure 2-12: Holes in proposed socket A: anterior holes B: sides holes

Each hole has 20mm width and 110mm length as in Figure 2-13.

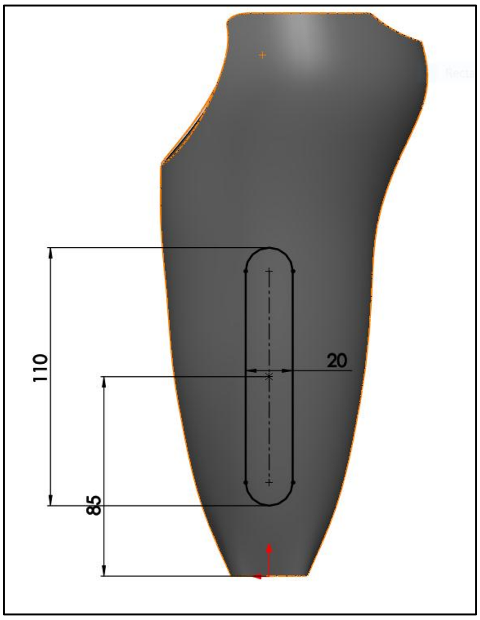


Figure 2-13: Measurements of holes

The upper back has been cut to allow for extra space when the size of the bones in the suspension area changes as in Figure 2-14.

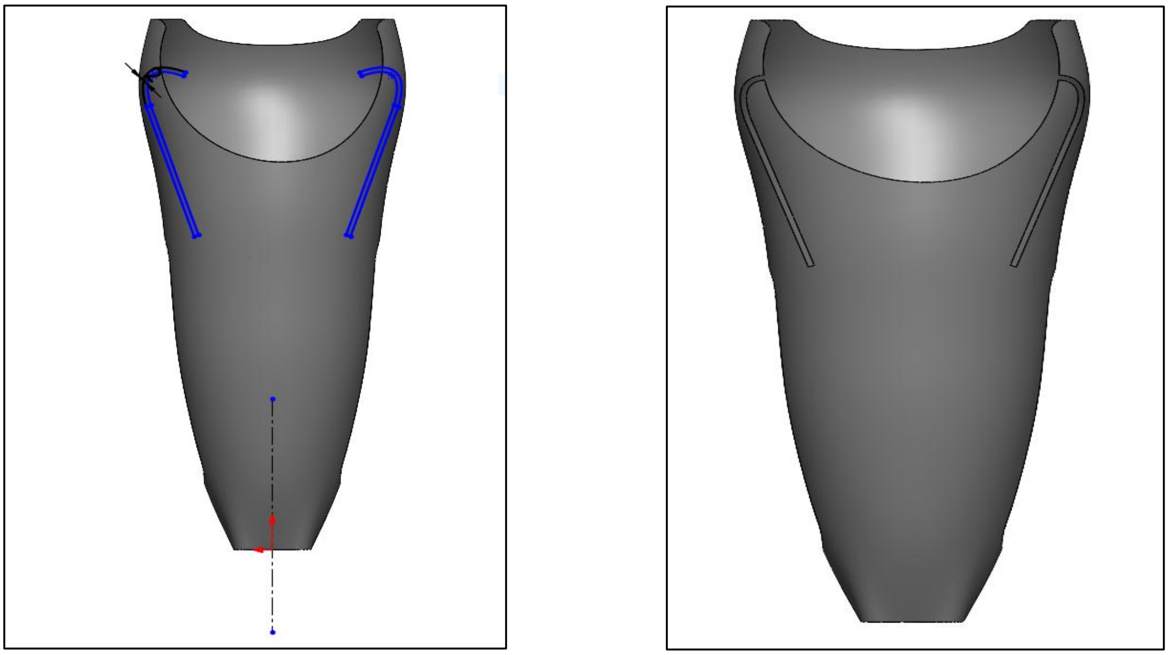


Figure 2-14: Posterior side

2.9.2 Designing the Proposed Shanks

Several designs have been proposed for the Shank, and the following will be explained in detail, as only two design was chosen to be manufactured.

2.9.2.1 First Proposed Design

It is designed in three pieces, the upper and lower pieces have the threads externally while the middle piece has the inner threads. The three pieces are connected by screws for increased fixation Figure 2-15.

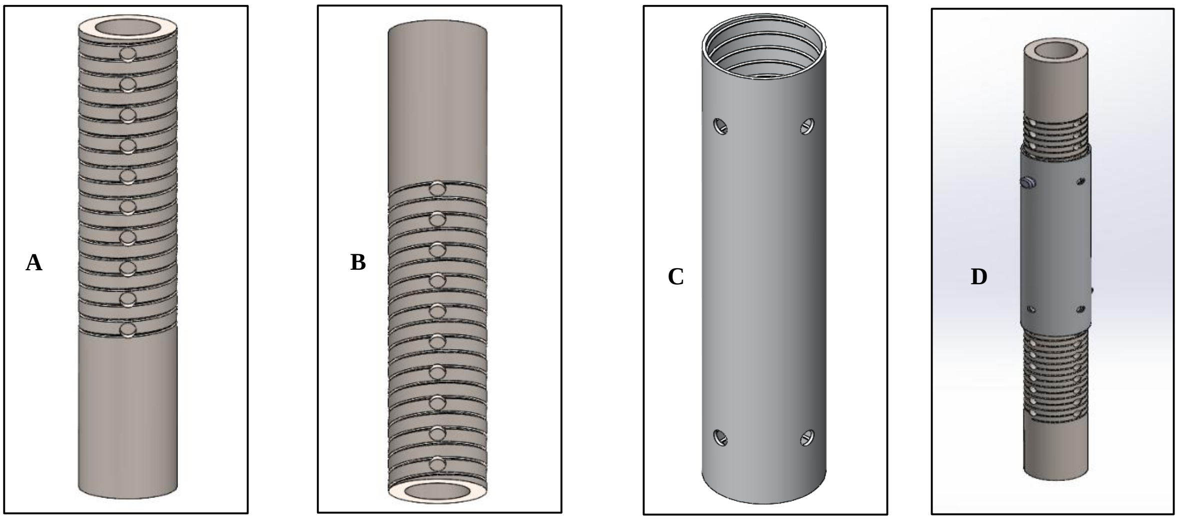


Figure 2-15: First proposed shank design A: upper part B: lower part C: middle part D: full design

This design had difficulty in manufacturing, especially the internal threads, so it was necessary to reconsider another design that is simple and can be manufactured.

2.9.2.2 Second proposed design

This shank is also designed from three parts, but without any thread as shown in Figure 2-16.

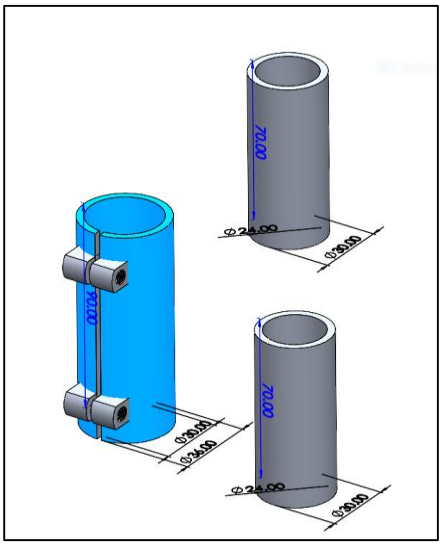


Figure 2-16: Parts of second proposed design

Upper and lower parts is a cylindrical shape, and the measurements of each only the middle has External protrusions for the purpose of controlling the width of this piece by screws as shown in Figure 2-17.

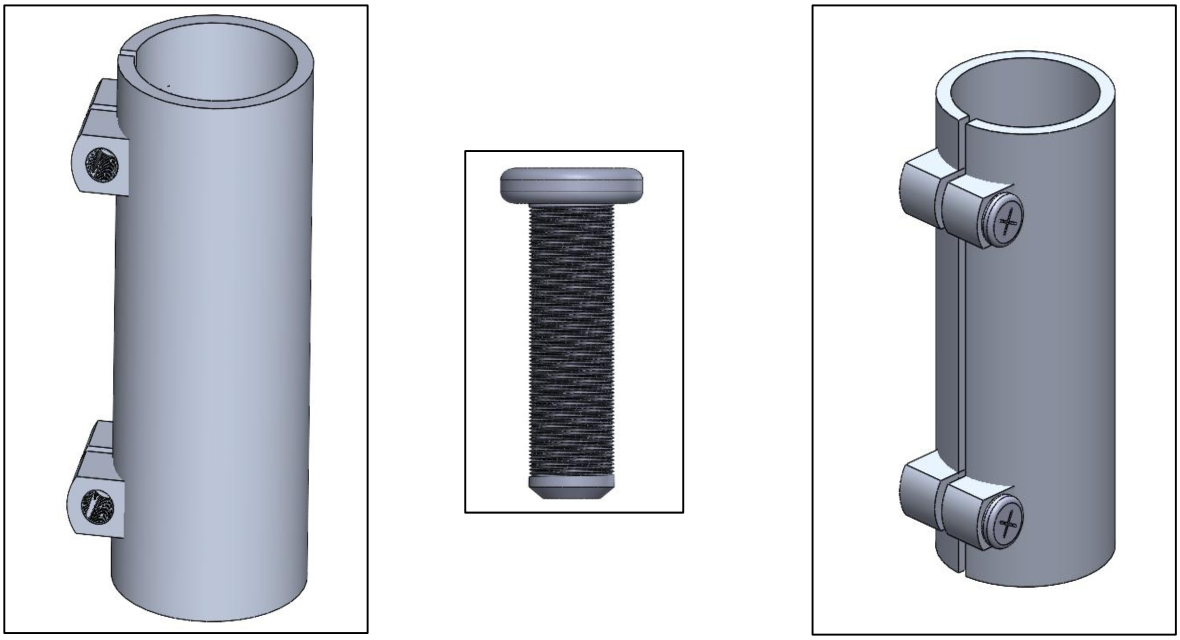


Figure 2-17: Middle part

2.9.2.3 Third propose shank designing

In this design, holes and screws are relied upon to control the length and fixing the pieces together to obtain the final shape of the shank as shown in Figure 2-18.

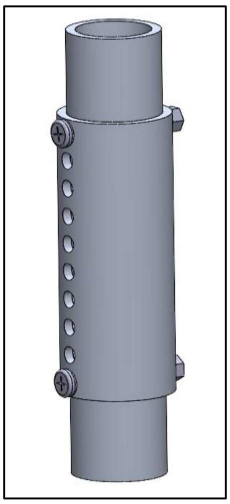
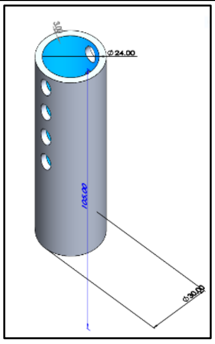
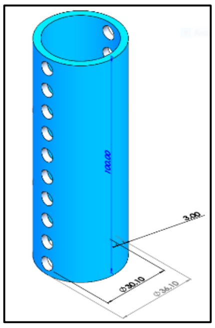


Figure 2-18: Third proposed design

The measurements of the design described in detail in table 2-1.

Table 2-1: Shank measurements

Part	Length mm	Internal diameter mm	External diameter mm	No. of holes	The design
Upper & lower	70	24	30	4	
Middle	100	30.1	36.1	10	

2.9.2.4 Forth proposed shank designing

This design is collecting between the adapter and shank (2&1). Also composes of three pieces, described in table 2-2. This design is fixing the problems of stump length, especially the long as we cannot deal with the stump in comfort. Figure 2-19: Forth proposed design.

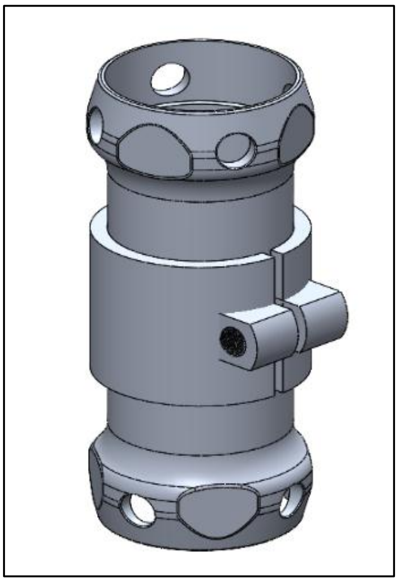
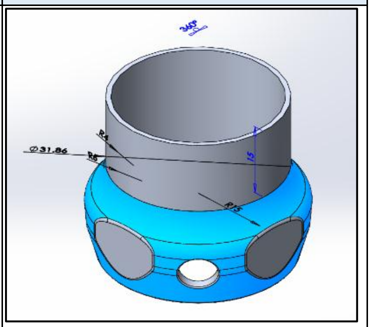
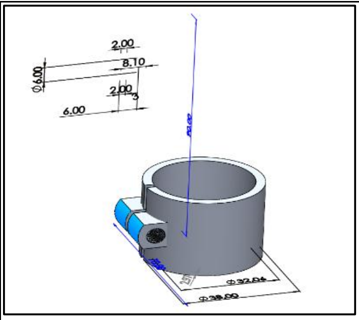


Figure 2-19: Forth proposed design

Table 2-2: Measurements of forth proposed design

Part	Length mm	Internal diameter mm	External diameter mm	The design
Upper & Lower	35	30	31.86	

Middle	30	32.06	38	

2.10 The Numerical Analysis

The finite element method (FEM) is now widely used in a variety of fields of engineering and science. Taking the advantage of the rapid development of digital computers with large memory capacity, as well as, fast computation. The method is recognized as one of the most powerful numerical methods because of its capabilities which include complex geometrical boundaries and non-linear material properties. In this work, FEM with aid of ANSYS Workbench 2019 R3 software to determine the behaviour of maximum stress, total deformation and safety factor [55]. The general analysis used by ANSYS has three distinct steps which are building the geometry as a model, applying the boundary conditions load and obtaining the solution and reviewing the results.

2.10.1 Creation of Mesh in the Socket (Second Proposed Design) and Shank

The meshing process has been done by choosing the size as fine shown in Figure 2-20 for socket.

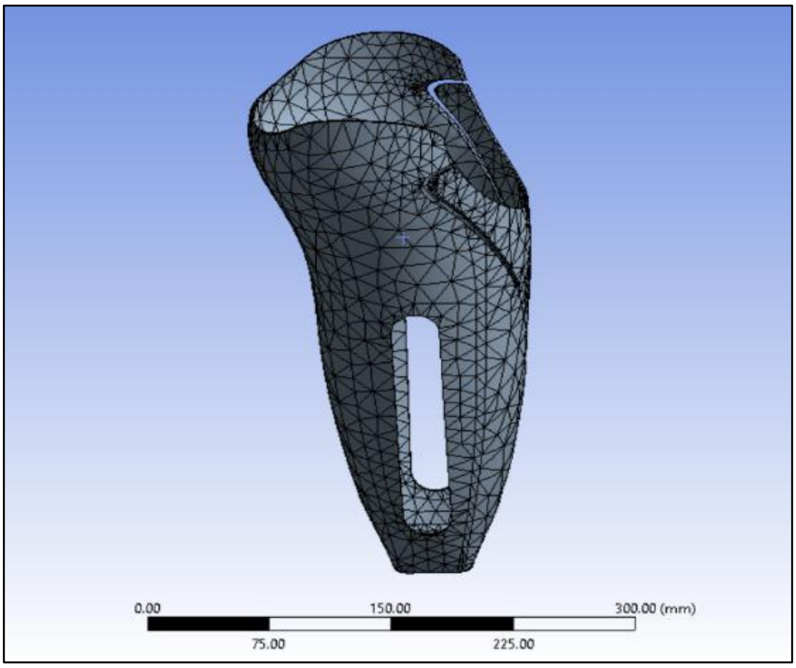


Figure 2-20: The model of BK with meshing.

2.10.2 Defining the Analysis and Applying Load.

The boundary conditions for the modal analysis was defined as the fixed support. The lower surface of the socket was chosen as a fixed support. While, the interface pressure was distributed according to the particular positions [56], as shown in Figure 2-21 for socket.

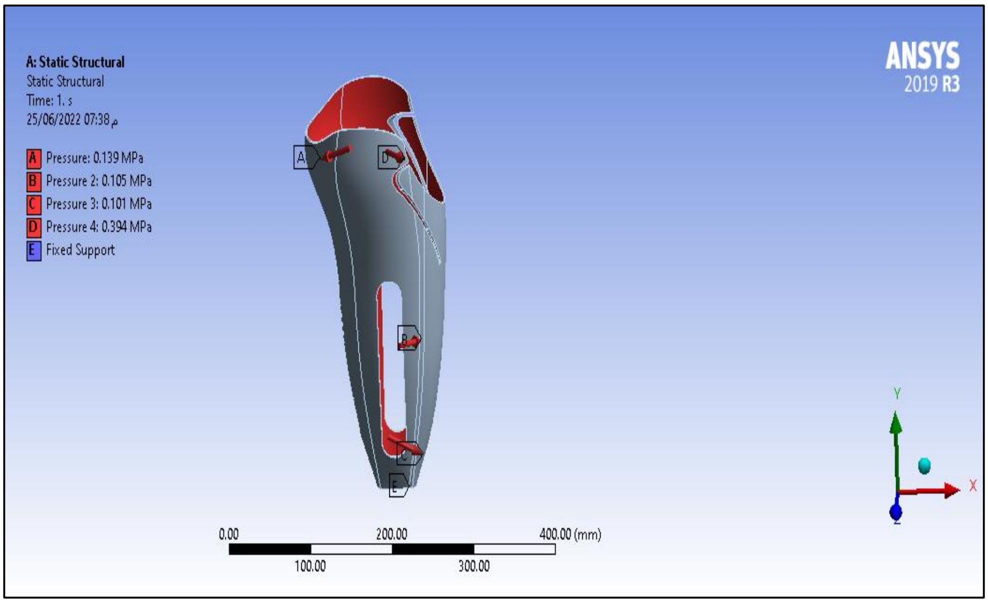


Figure 2-21: Fixed support and interface pressure for BK socket

The meshing process for the shank done as shown in Figure 2-22.

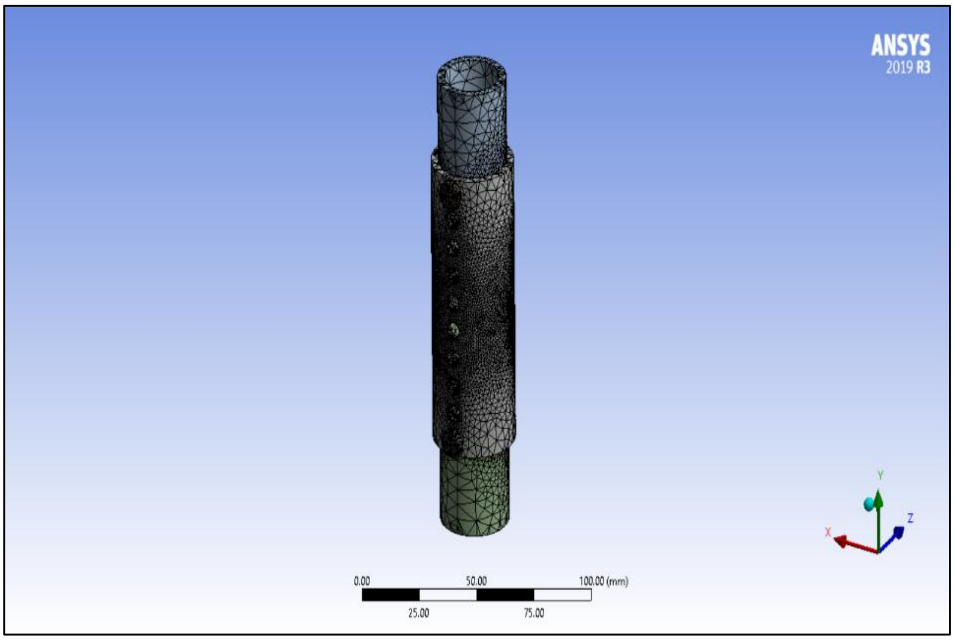


Figure 2-22: Shank mesh

Fixed support done at the upper surface of the model, while the load applied at the lower surface as shown in Figure 2-23.

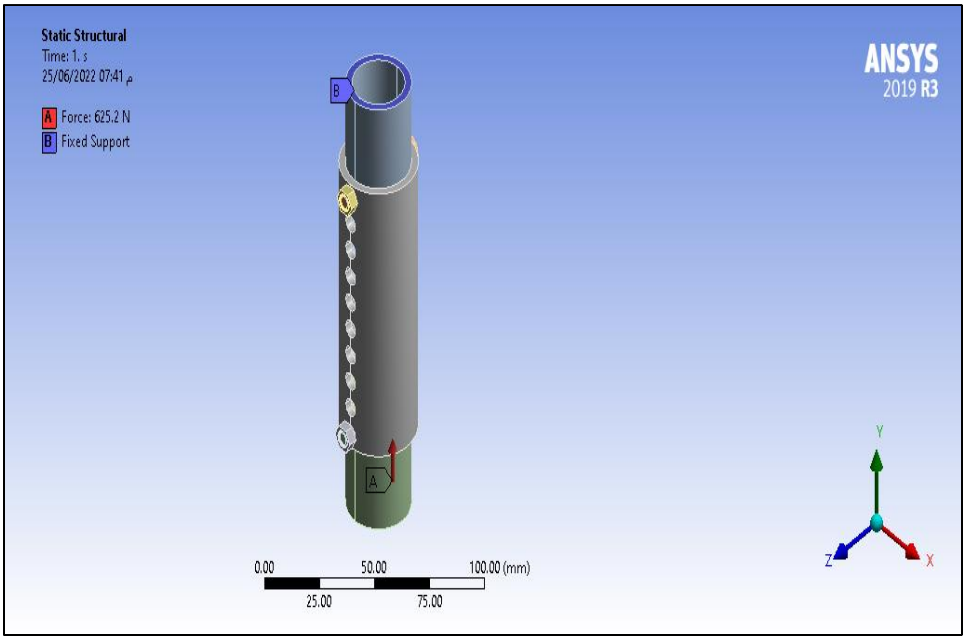


Figure 2-23: Fixed support and load in shank

To calculate the Poisson's ratio, program has been used to construct the values that used in ANSYS to analyse the socket model [57]. The program is included in Appendix A.

CHAPTER 3

EXPERIMENTAL WORK

3.1 Introduction

This chapter deals with materials and apparatus used in this research work as well as the experimental procedure.

3.2 The Materials Used in This Study

Composite materials has been used as BK socket materials described below in table 3-1:

Table 3-1: Components of socket manufacturing

Name of materials	Descriptions and details	Shape of products
Poly methyl methacrylate Acrylic resin	The amount that used in manufacturing Bk socket was 300 g*	
Perlon	10 layers were used for manufacturing The measurement of each layer was 8cm. Here 2 layers with 10 cm of black perlon were used for the purpose of fixing the colour (for second design)	 

Polyvinyl alcohol PVA	2 layers 10 size	
Hardener & color	10 g used in mixture	

The mechanical properties for both perlon and PMMA resin described in following table3-2 [58].

Table 3-2: Mechanical properties of materials used in socket manufacturing [58]

Materials	σ_{ult} MPa	Elongation percentage	Elastic modulus GPa	Poisson's ratio
Perlon	78	1-30	2.6-3	0.39
Poly methyl methacrylate (PMMA resin)	48.3-72.4	2-5.5	2.24-3.24	0.35

3.3 Case Study

Measurement were collected for male patient is about (8) years old, his weight 126 kg with 120 length, suffered from left side below knee amputation due to accident as shown in Figure 3-1.

Figure 3-1: Patient with below knee amputation

3.4 Manufacturing Procedure of Below Knee Prosthetic Device

The manufacturing process for BK will be explained in the following steps:

3.4.1 Taking Measurements

1. Length of stump has been taken, then recorded in the chart as in Figure 3-2.
2. Circumferences of calf and stump end as well as the middle of stump have been recorded.

3. Foot size has been taken

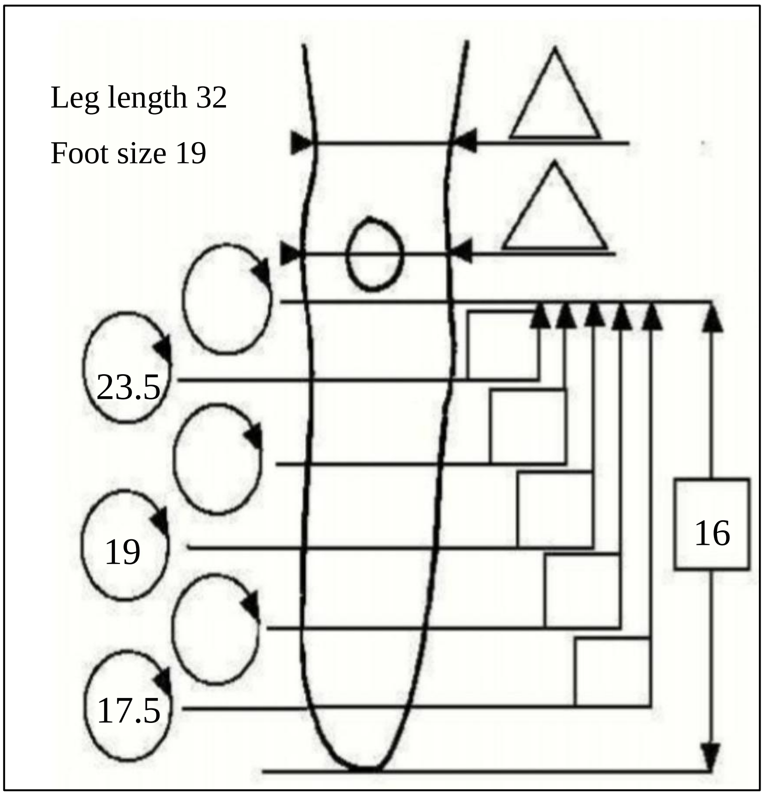


Figure 3-2: Stump measurements [59]

3.4.2 Handing Casting

1. Four layers of nylon were added above the patella, then twisted distally to the end of stump.
2. Elastic plaster of paris has been wetted and start wrapping with one turn above the knee, continue downwards wrapping diagonally then stretched the bandage to increase the pressure on stump at suspension regions.
3. Thumb was marked below the patella and position of hands behind the knee.
4. On both sides of the condyles and tibia has been modeled. Advantage was taken of all weight bearing areas. Mold around and below the patella and in the dorsal region.

5. After the suspension regions were marked, the mold left to dry for period of time as shown in Figure 3-3.

Figure 3-3: Mould taking

3.4.3 Preparing for Rectification

The top section of the cast Wrapped with plaster bandage, so that the bandage extended about 2 cm proximal to the lateral trim line.

The inside of the cast Coated with a liquid soap, so removing the cast from plaster was easier. The cast Placed in sandbox and filled with plaster. The mold was left to dry completely as shown in Figure 3-4.

Figure 3-4: Mould with plaster

3.4.4 Cast Rectification

1. Plaster with water Mixed until the water is saturated. The negative Filled with the vacuum tube inside.

2. The negative has been Cut and removed. Special contours marked, sensitive areas and the plumb lines.

3. The measurements have been checked and the cast has been smoothed as shown in Figure 3-5.

Figure 3-5: Rectification steps

3.4.5 Fabrication of the Soft Insert Liner

1. Eva was used as BK soft socket with 3mm thickness with length 25cm, proximal width 30 and distal width 19 has been take from the mold.
2. The opposite edges was Grinded on the grinding drum until the edge is beveled.
3. Glue material was placed on the edges of the piece and left to dry for 10-15 minutes.
4. The piece has been inserted inside the furniture 5-10 minutes with 200^{o}-250^{o}C.
5. Piece was wrapped onto the mold. Cap also made as the same procedure to cover distal end of soft socket.
6. Four pieces with rectangular shaped has been glued to the outer face of soft socket for the purpose of making grooves shape within the sockets during casting for placing metal pieces described in Figure 3-6.

Figure 3-6: Steps of soft insert fabrications

3.4.6 Lamination of Socket

For first design and by using composite materials (polymethylmethacrylate+ perlon) with PVA socket was fabricated by the following:

1. PVA was wrapped onto the mold then the 10 layers of perlon, finally the second layer of PVA.
2. Adapter was aligned and fixed to the mold after the sixth layer of perlon
3. 300 g acrylic resin with hardener & color mixed for 3 min then the mixture has been added to the mold
4. Vacuum turned on during the casting. After 30 min socket was ready to modification procedures. Figure 3-7 shows lamination of socket.

Figure 3-7: Lamination of first proposed socket

For second design, the lamination procedure was the same accept, two layer of black perlon 10cm size was added for the purpose of clarifying and fixing the color as shown in Figure 3-8.

Figure 3-8: Lamination of second proposed design

3.5 Socket Modifications

The modifications for converting the socket into an adjustable device will be described for both designs.

3.5.1 First Proposed Design

For first proposed socket, the modification explained in details as below:

1. By using a metal plates for giving an adjustable length to the socket.
2. The socket has been cut into two parts then the distal part into four
3. The four parts have been grinding, then assembled with adapter by the bolts. Figure 3-9 described modification procedures.

Figure 3-9: Modification of first proposed design

3.5.2 Second Proposed Design

1. Three holes have been made in socket for minimize the weight.
2. Length has been increased at the distal end by 2 cm
3. The posterior section at suspension part has been cut to give wide circumference when the size of bones changed.
4. In suspension section small metal plates connected between them work as straps as shown in Figure 3-10.

Figure 3-10: Modification of second proposed design

3.6 Shank manufacturing

First proposed design has been fabricated by CNC machine with aluminium material 6061-T6 as shown in Figure 3-11.

Figure 3-11: Proposed shank manufacturing

Second proposed design fabricated by 3D printing just for only presenting the design as in Figure 3-12.

Figure 3-12: Proposed 3D shank printing

3.7 Artificial limb Assembly

The parts for assembling the limb are:

1. Socket
2. Adapter Socket
3. Double adapter
4. Adapter foot
5. Foot

The parts connected together, then it was worn by the patient. Double adapter has been used in assembling because the stump was long and the aluminum shank was not appropriate as in Figure 3-13.

Figure 3-13: Artificial limb assembly

3.8 Mechanical Properties Tests

Testing the samples is implemented by using the Testometric machine (Al-Nahrain University / mechanical Engineering). The tensile was carried out by Hi-TECH device.

3.8.1 Tensile Test for Composite Materials (Acrylic Resin + Perlon)

Fibers are used in orthopedic technology as stockinet [58]. Layers as reinforcing materials in PMMA resin. Table 3-3 illustrate the Type of lamentations.

Table 3-3: Table of lamination materials for both proposed socket design

No. of samples	Thickness	Type of materials	Arrangement of layers
Sample 1	3	PMMA + Perlon Layers	10 perlon layers
Sample 2	3	PMMA + Perlon Layers	10 perlon layers
Sample 3	3	PMMA + Perlon Layers	10 perlon layers

For shank, table 3-4 shows the details of materials that used in manufacturing the samples.

Table 3-4: First proposed shank materials

No. of samples	Thickness	Type of materials
Sample 1	3	Aluminium 6061-T6
Sample 2	3	Aluminium 6061-T6
Sample 3	3	Aluminium 6061-T6

All tensile test samples tested using the universal testing instrument (testometric). For the composite material the cross head speed is 5mm/min. Figure3-14 shows samples under tensile test.

Figure 3-14: Tensile test device

Therefore, the specifications of the tensile test have been restricted according to the American Society for Testing and Materials specifications (ASTM); the tensile specimen's geometry and dimensions for standard (D638) [60] was specified for composite material used in socket fabrication, as shown in Figure 3-15.

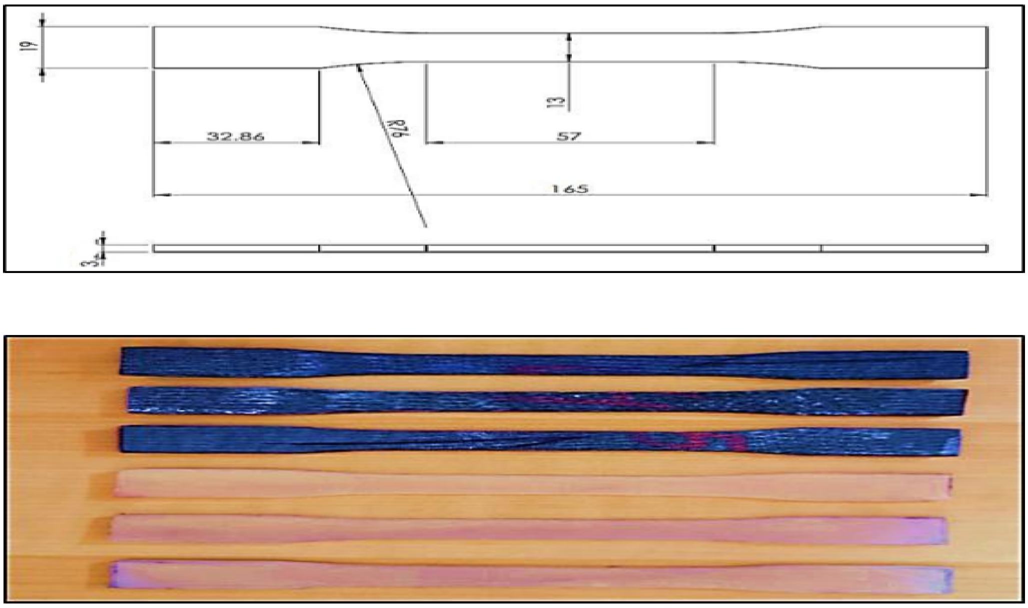


Figure 3-15: The general shape and dimensions of tensile specimens D638.

3.8.2 Tensile Test for Aluminum Material

By using the American Society for Testing and Materials specifications (ASTM); the tensile specimen's geometry and dimensions for standard (E8) [61] was specified for aluminum material used in shank manufacturing, as shown in Figure 3-16.

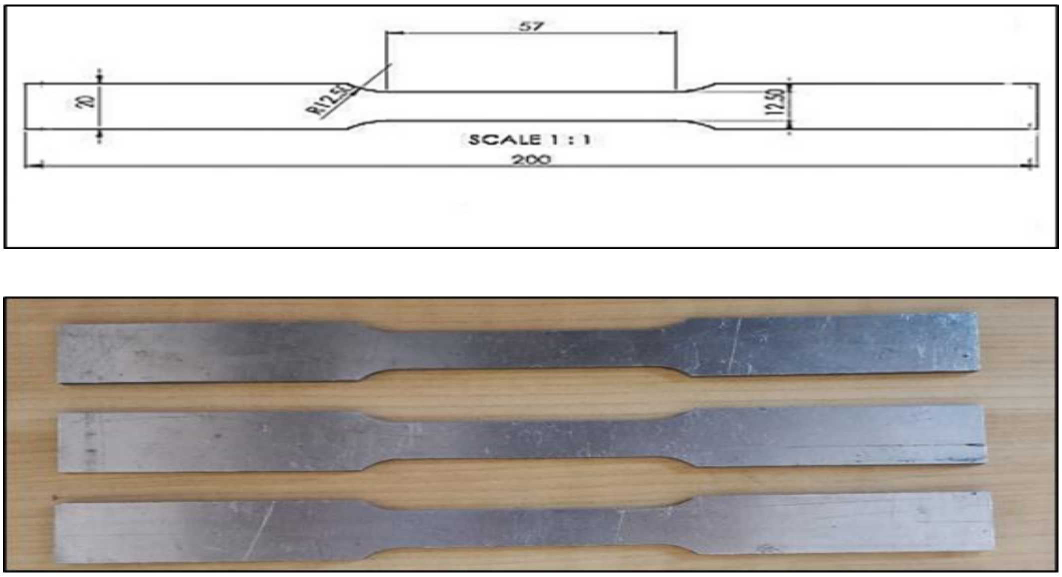


Figure 3-16: The general shape and dimensions of tensile specimens E8.

3.8.3 Testing Procedure

1. The sample was dragged into tensile grips.
2. The both end of grips closed well.
3. The test started for each sample at speed rate 2 mm/min for socket materials and 5 mm/min for shank material.
4. The test stopped when the sample was break as in Figure 3-17.

Figure 3-17: The samples after testing

3.9 F-Socket Device

By using F-socket device that consist of sensors which carried out in Prosthetic and Orthotic Department/ Al-Nahrain University, the interface pressure between the residual limb and socket were measured.

The steps of measured the interface pressure start by applied the sensors of the F-socket software at the anterior of the residual limb and then the socket was put on, the patient begin to move and the software start to record the movement and draw a curve between the pressure and time, then the same procedure repeated for the posterior, lateral and medial side of residual limb as in Figure 3-18.

Figure 3-18: Sensor inside the socket

3.10 Gait Cycle Testing

Gait cycle analysis was done in the laboratories of the P&O department of Al-Nahrain University on force plate (Tekscan’s Walkway Pressure Assessment Systems) .The Walkway system provides static and dynamic gait data and barefoot pressure and force measurements over several steps using a low profile floor walkway. It is ideal for capturing multiple foot strikes during natural or perturbed. Figure 3-19 shows the force plate and the child walking over the plate with below knee prosthetic limb during testing.

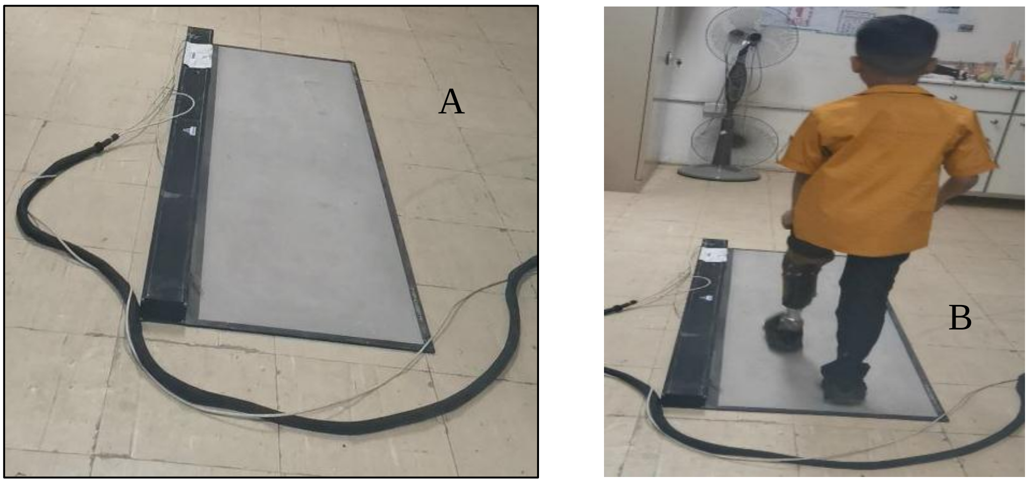


Figure 3-19: (A) Force plate (walkway) (B) patient with BK on gait cycle table.

CHAPTER 4

RESULTS AND DISCUSSIONS

4.1 Introduction

In this chapter, the results of the experimental work for the adjustable socket design will be shown and explained. Results of tensile test and the gait cycle (GRF, Pressure distribution, force distribution, peak contact area, peak contact pressure, Centre of Pressure (COP), Gait analysis, step-stride, gait cycle tables and foot print analysis). The analysis of the kinematics data will indicate the differences of the gait cycle between the normal and pathological subject. Results of F-socket device measured the interface pressure by sensor between the stump and the socket of BK prosthetic will be shown in this chapter.

4.2 Tensile Properties Results

The results of the mechanical properties (tensile test) from testometer for the socket and shank materials shown in tables 4-1 to 4-3. The stress-strain curve for shank material shown in Figure 4-1 a, b and c. the stress-strain curve for socket material (black socket) shown in Figure 4-2 a, b and c and (yellow socket) design in Figure 4-3 a, b and c. The results from the laboratory examination have been attached in Appendix B.

Table 4-1: Mechanical properties for composite material (PMMA + Perlon) (first proposed design)

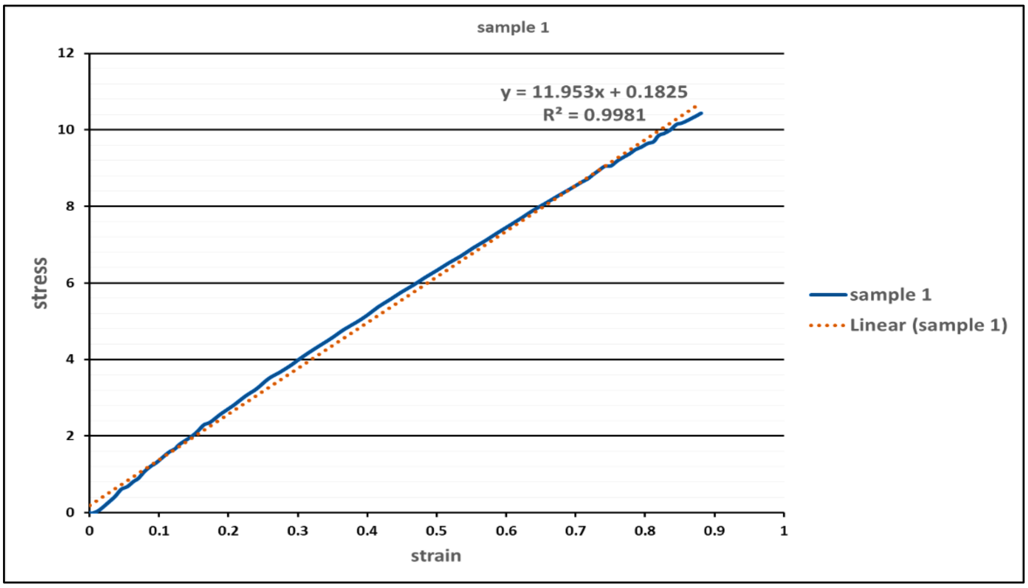
No. of sample	Thickness mm	σy (MPa)	σult (MPa)	E (GPa)
Sample 1	3	20.813	30.821	1.127431
Sample 2	3	12.705	30.115	1.032685
Sample 3	3	14.754	30.336	1.148082
Averages	-----------	16.089	30.424	1.102733

Table 4-2: Mechanical properties of composite material for socket (PMMA+ perlon) (second proposed design)

No. of sample	Thickness mm	σy (MPa)	σult (MPa)	E (GPa)
Sample 1	3	32.631	32.631	1.634602
Sample 2	3	9.367	23.715	1.044292
Sample 3	3	5.444	24.182	9.96490
Averages	-----------	15.814	26.843	1.225128

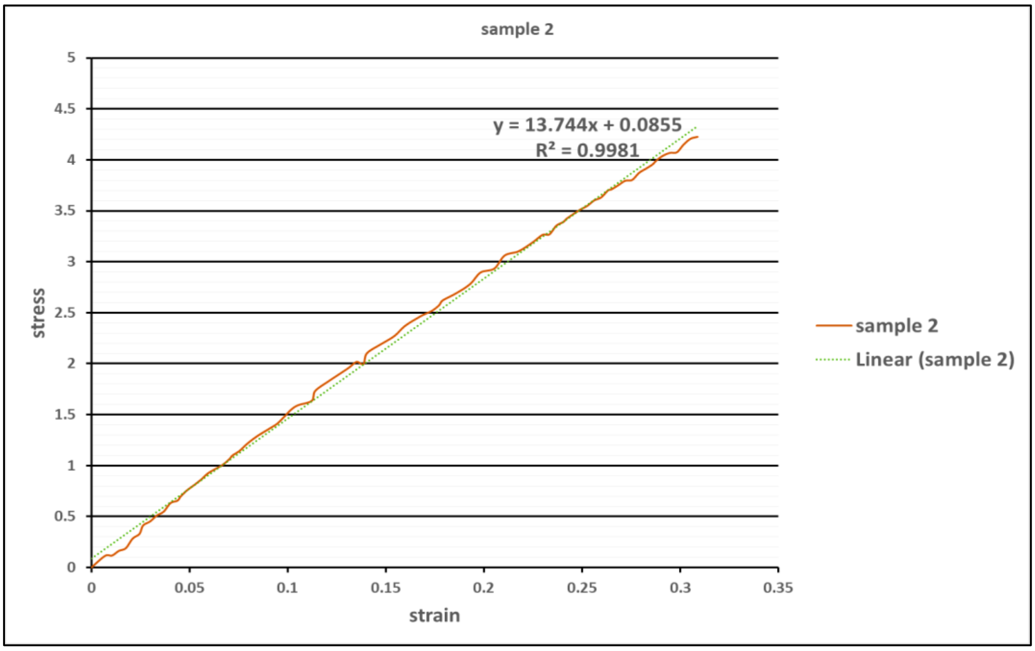
Table 4-3: Mechanical properties of shank material (aluminum 6061-T6)

No. of sample	Thickness mm	σy (MPa)	σult (MPa)	E (GPa)
Sample 1	3	79.520	96.613	6.642485
Sample 2	3	69.893	94.293	8.537214
Sample 3	3	77.413	94.240	5.775950
Averages	----------	75.609	95.049	6.985216



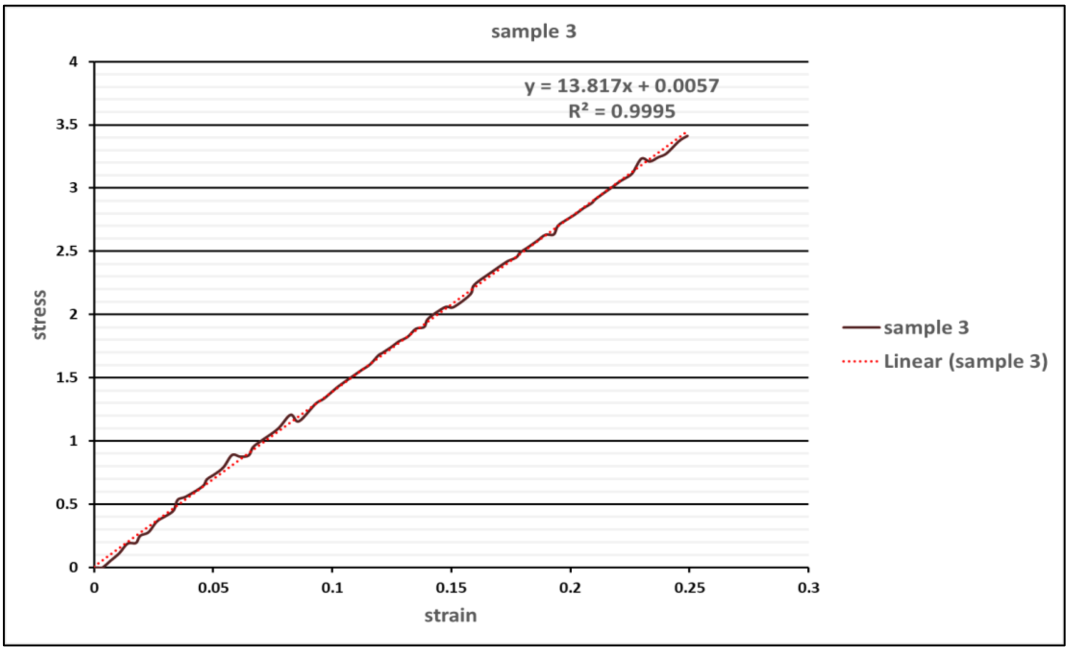
sample 1
y = 11.953x + 0.1825
R² = 0.9981
stress
strain
0
2
4
6
8
10
12
0
0.1
0.2
0.3
0.4
0.5
0.6
0.7
0.8
0.9
1
sample 1
Linear (sample 1)

A



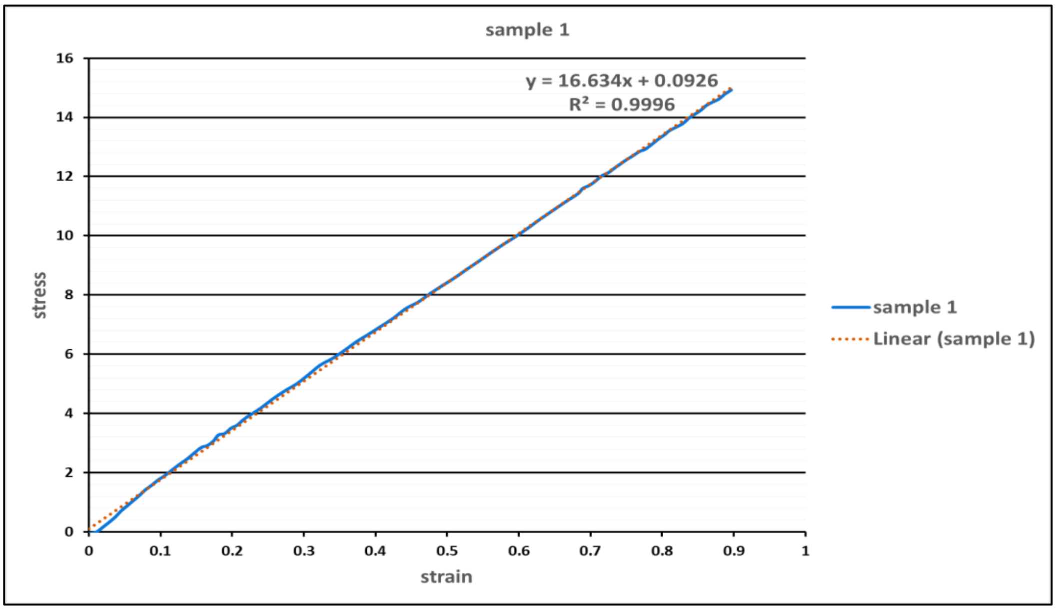
sample 2
y = 13.744x + 0.0855
R² = 0.9981
stress
strain
0
0.5
1
1.5
2
2.5
3
3.5
4
4.5
5
0
0.05
0.1
0.15
0.2
0.25
0.3
0.35
sample 2
Linear (sample 2)

B

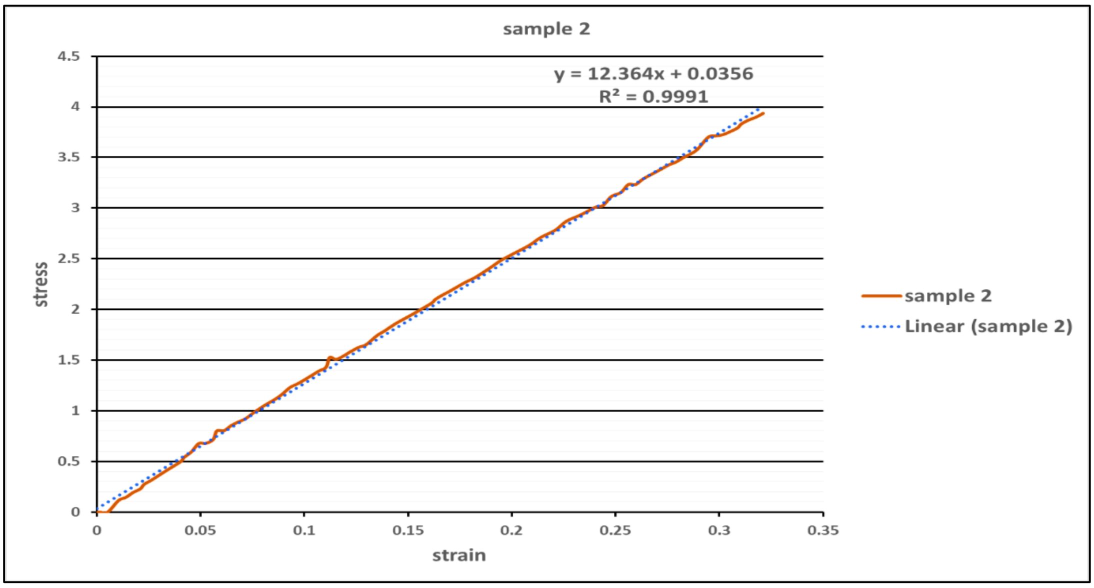


C

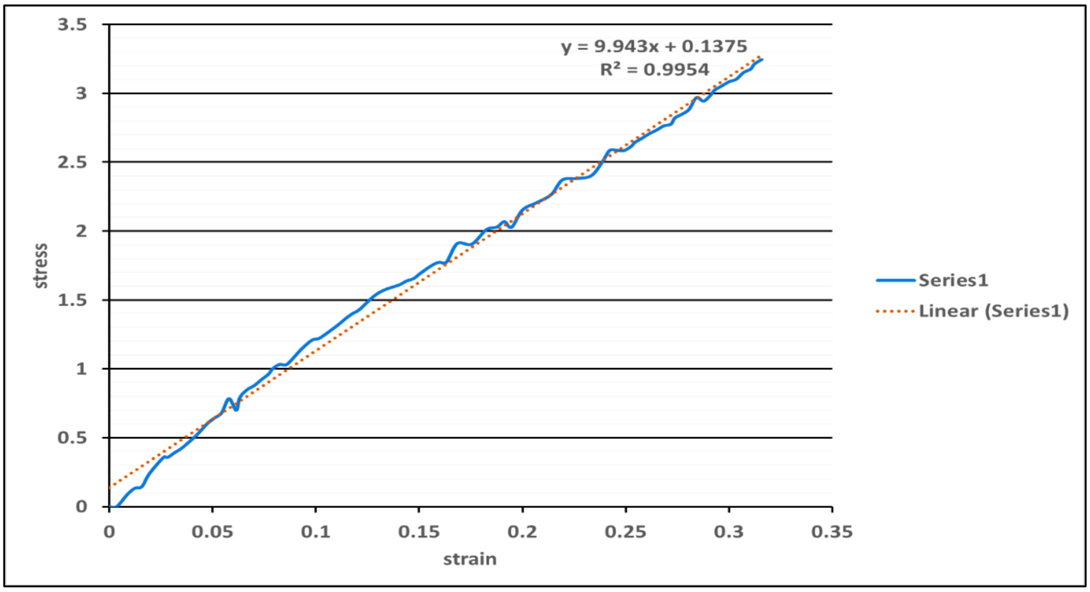
Figure 4-1: First proposed socket samples a, b and



A

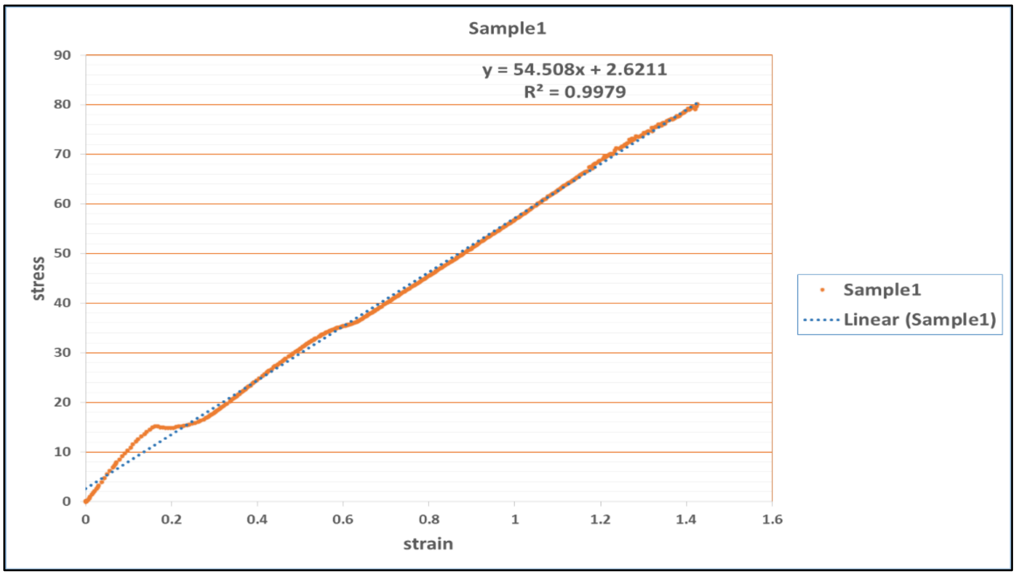


B



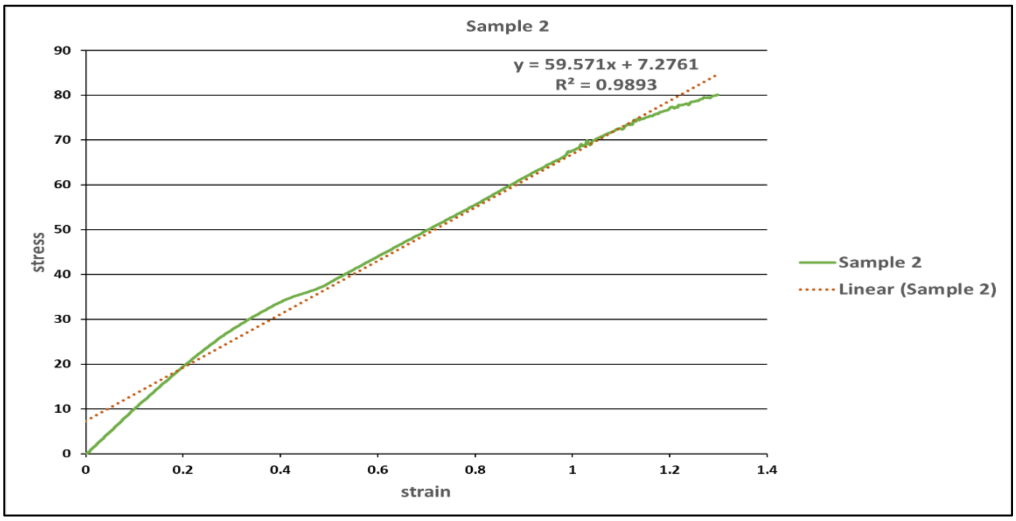
C

Figure 4-2: Second proposed socket samples a, b and c



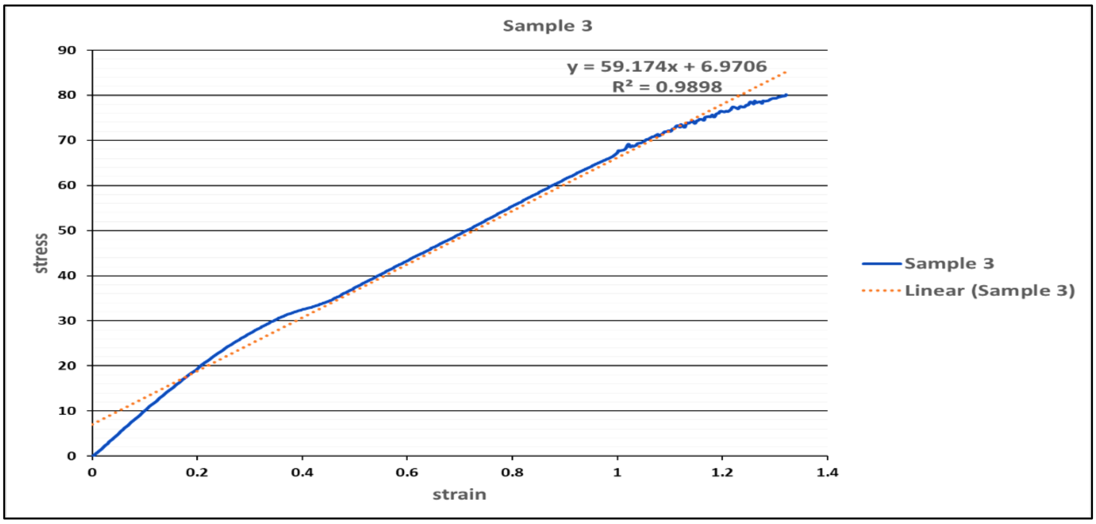
Sample1
y = 54.508x + 2.6211
R² = 0.9979
stress
strain
90
80
70
60
50
40
30
20
10
0
0
0.2
0.4
0.6
0.8
1
1.2
1.4
1.6
Sample1
Linear (Sample1)

A



Sample 2
y = 59.571x + 7.2761
R² = 0.9893
stress
strain
90
80
70
60
50
40
30
20
10
0
0
0.2
0.4
0.6
0.8
1
1.2
1.4
Sample 2
Linear (Sample 2)

B



C

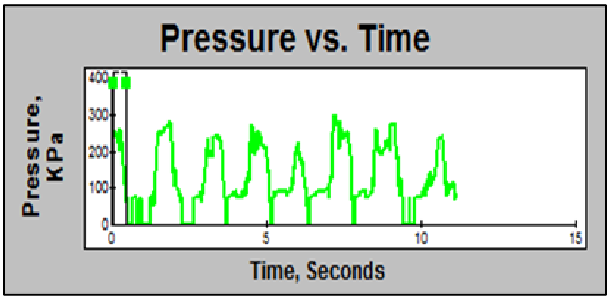
Figure 4-3: Proposed shank material samples a, b and c

4.3 F-Socket Results

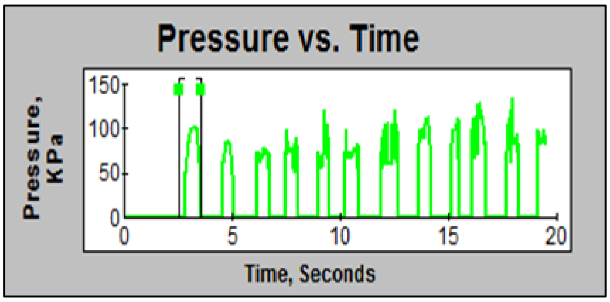
The pressure test results indicated that the highest value was on the lateral region of the socket and recorded in table 4-4, Figures 4-4 a, b, c and d showing the interface pressure (IP) distribution at four sides of socket.

Table 4-4: Readings and results for IP

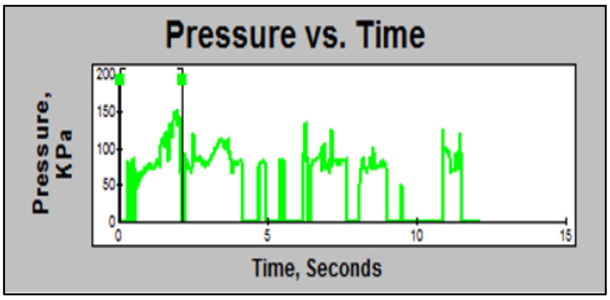
Socket regions	Anterior	Lateral	Posterior	Medial
IP(KPa)	139	394	101	105



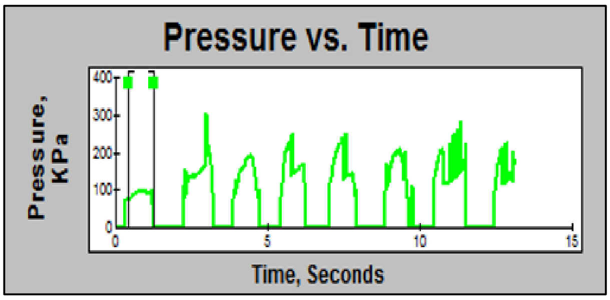
A) Anterior



B) Medial



C) Lateral



D) Posterior

Figure 4-4: Pressure vs. time

4.4 The Results and Discussion of the Gait Cycle Parameters.

The ground reaction force (GRF) introduced under sole, due to biomechanical effects on leg during swing and stance phases, can be done for patient with BK amputation in left leg by using force plate. Walk over force plate where the force distribution is developed under sole due to patient gait cycle. The obtained data from the gait cycle test recognize the major differences for the parameters of the left leg. The results will be discussed in detailed for each case as follows: The results of gait as shown in table 4-5.

Table 4-5: Gait table

Gait Table	patient
Number of Strikes	6
Cadence (steps/min)	28.6
Gait Time (sec)	6.29
Gait Distance (cm)	94.7
Gait Velocity (cm/sec)	15.1
Gait Velocity/Leg Length (LL/sec)	0.5

The main parameters which are shown in Tables 4-6 & 4-7 describe the behaviour of the gait cycle for patient wearing BK socket.

Table 4-6: Gait cycle table

Gait Cycle Table (sec)	patient		
	Left	Right	Difference
Gait Cycle Time	n/a	n/a	n/a
Stance Time	0.89	1.22	0.33
Swing Time	n/a	n/a	n/a
Single Support Time	n/a	n/a	n/a
Initial Double Support Time	0.00	0.11	0.11
Terminal Double Support Time	0.11	0.00	-0.11
Total Double Support Time	0.11	0.11	0.00
Heel Contact Time	0.63	0.82	0.20
Foot Flat Time	0.17	0.53	0.36
Midstance Time	3.66	0.56	-3.11
Propulsion Time	0.26	0.40	0.14
Active Propulsion Time	0.26	0.29	0.03
Passive Propulsion Time	0.00	0.11	0.11

Table 4-7: Step stride table

Step-Stride Table	patient		
	Left	Right	Difference
Step Time (sec)	3.87	0.58	-3.30
Step Length (cm)	58.7	34.4	-24.3
Step Velocity (cm/sec)	15.2	59.7	44.5
Step Length/Leg Length (ratio)	1.83	1.07	-0.76
Step Width (cm)	n/a	n/a	n/a
Stride Time (sec)	n/a	n/a	n/a
Stride Length (cm)	n/a	n/a	n/a
Stride Velocity (cm/sec)	n/a	n/a	n/a
Maximum Force (%BW)	625.2	486.8	-138.4
Maximum Force (N)	162.54	126.57	-35.97
Impulse (%BW*sec)	375.6	370.3	-5.4
Impulse (N*sec)	97.67	96.27	-1.40
Maximum Peak Pressure (MPa)	0.20	0.08	-0.12
Foot Angle (degree)	n/a	n/a	n/a

Centre of pressure for the patient with BK prosthetic device. The left leg as the one inside the blue box and right in green box as shown in Figure 4-5

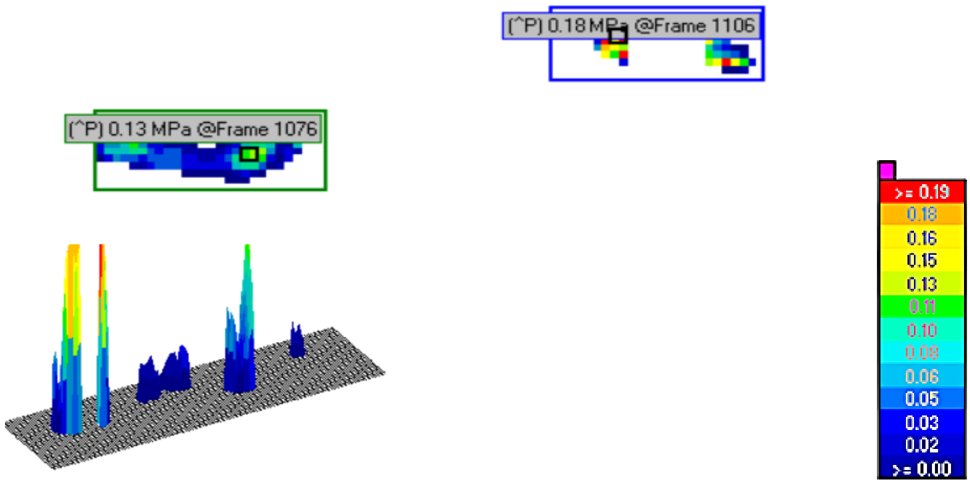


Figure 4-5: Centre of pressure in both feet

The force distribution under sole due to patient gait for two feet is shown in Figure 4-6.

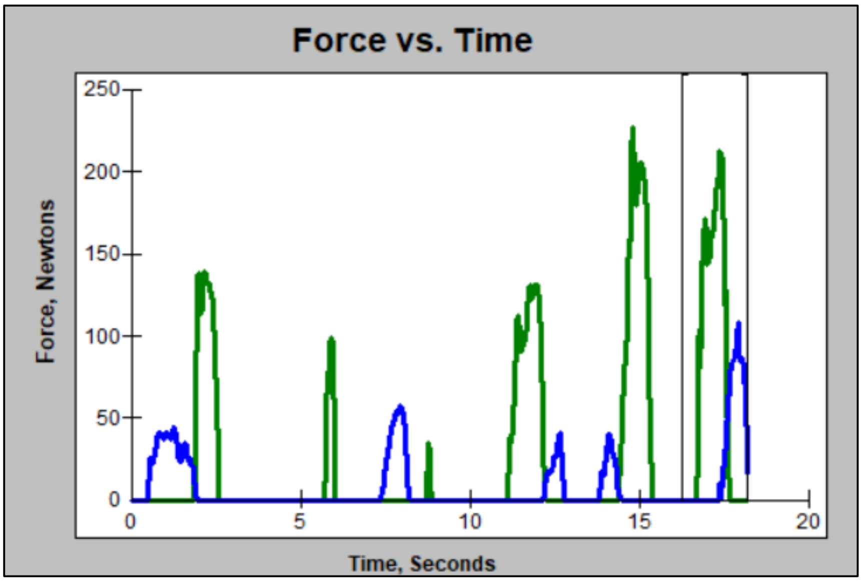


Figure 4-6: Force vs. time

4.5 Ansys Results

The model of prosthetic shank was analyzed by using Workbench software 19 R3 to calculate the total deformation, equivalent stress (Von-Mises) and safety factor of the socket and shank when the force applied vertically on the distal surface of the shank taken form force plate test, while interface pressure distributed in four region anterior, posterior, lateral and medial taken from F-socket test. According to the Von-Mises theory that considers the yield stress as criteria; ($\sigma e < \sigma y$, safe), ($\sigma e = \sigma y$, critical) and ($\sigma e > \sigma y$, failed), Where, (σe) is the equivalent stress, and (σy) is the yield stress, so according to Von-Mises theory the shank are safe. The safety factor will be safe in design if the safety factor about or more than (1.25) [62 and 63], so according to safety factor the socket is unsafe, while shank was good. The result of models are shown in Figure 4-7 to 4-14.

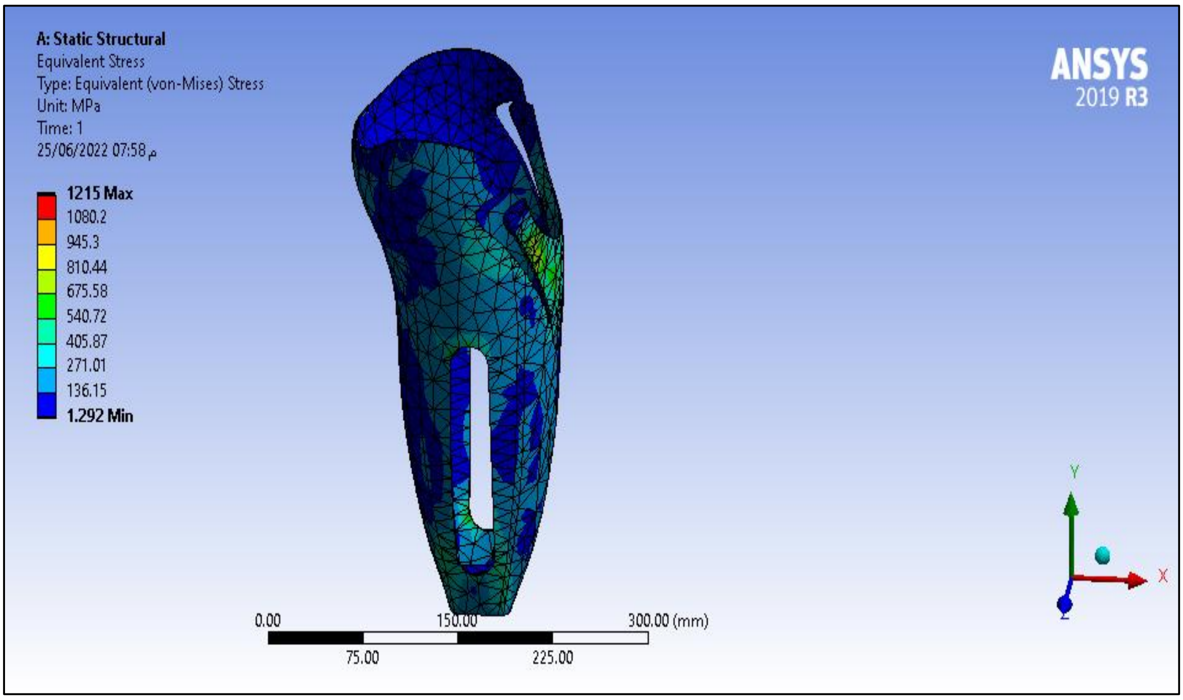


Figure 4-7: Equivalent stress of socket

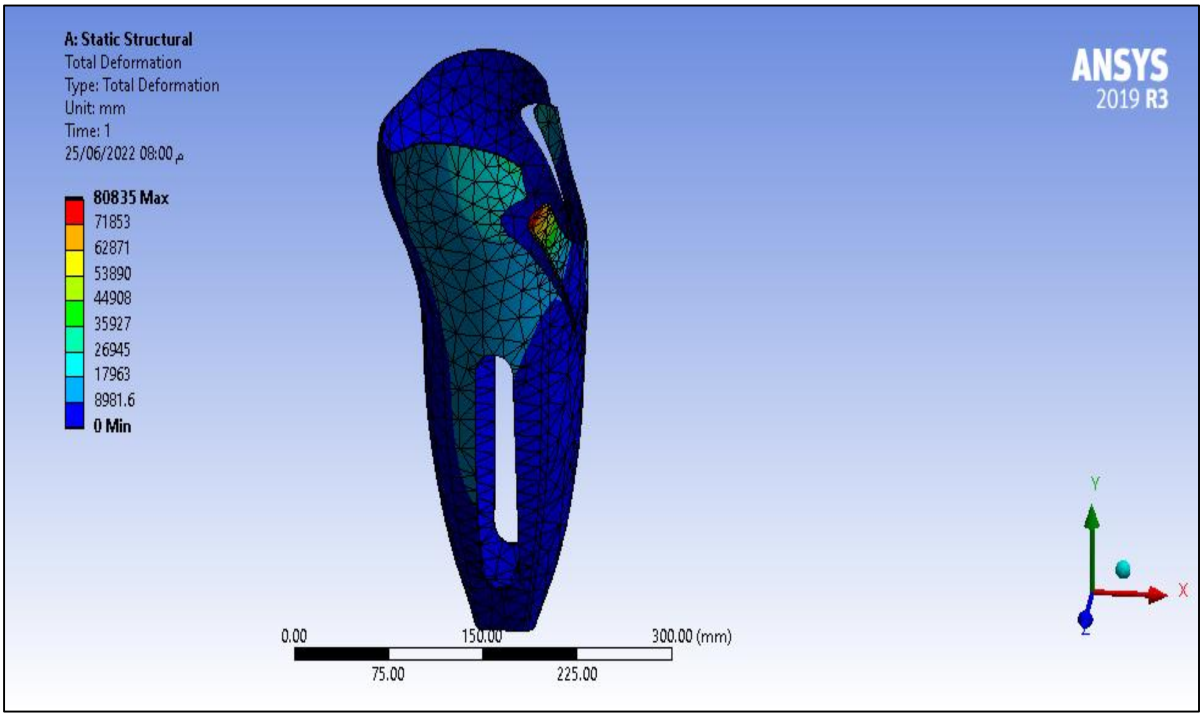


Figure 4-8: Total deformation of socket

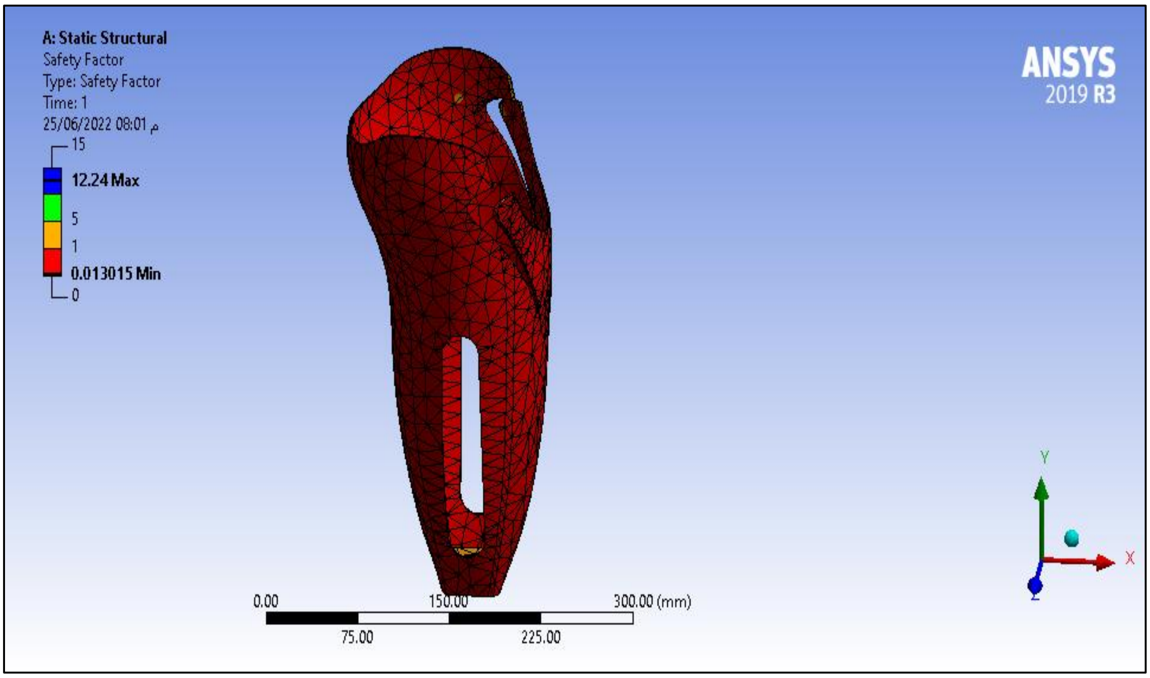


Figure 4-9: Factor of safety of socket

Through the above results, it is clear that the equivalent stress is higher than the yield stress and this is due to the sharp edges of the design where high stress is concentrated in the edges in addition to the number and size of the meshes must be small relative to the edge lines, therefore the stress concentration increases. Together, these causes lead to rapid analysis failure and a high rate of equivalent stress. The design in real has curvatures at edges and no sharp ends appear, therefore the values at sharp edges can be ignored and it is possible to choose another area that is exposed to load, the stress and safety factor are calculated of them.

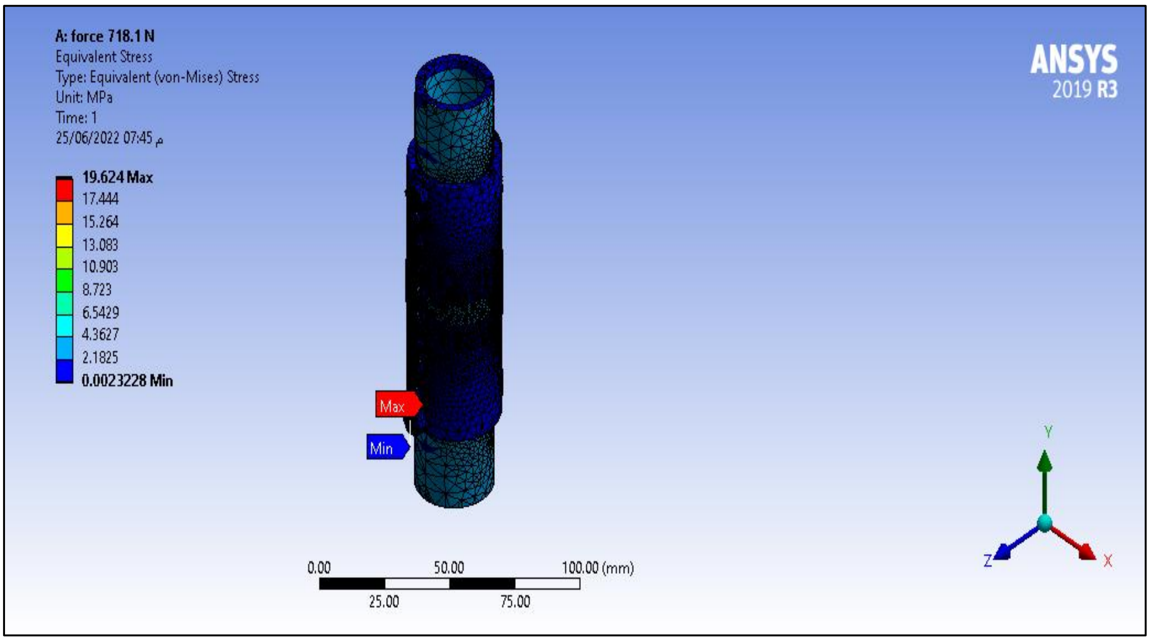


Figure 4-10: Equivalent stress of shank

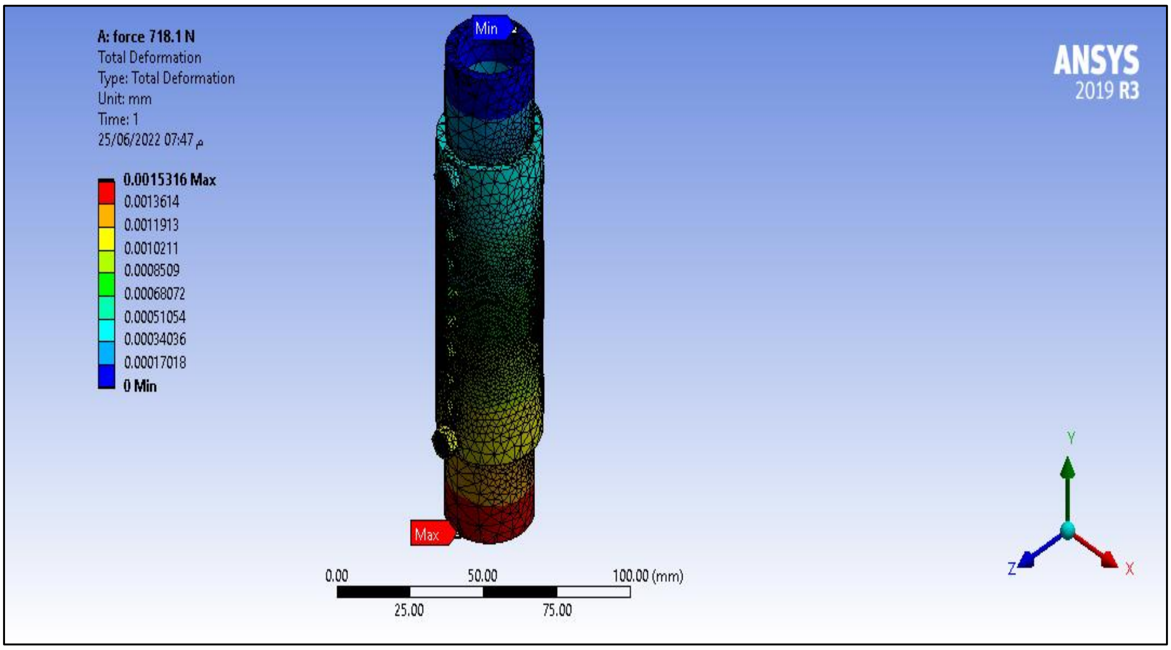


Figure 4-11: Total deformation of shank

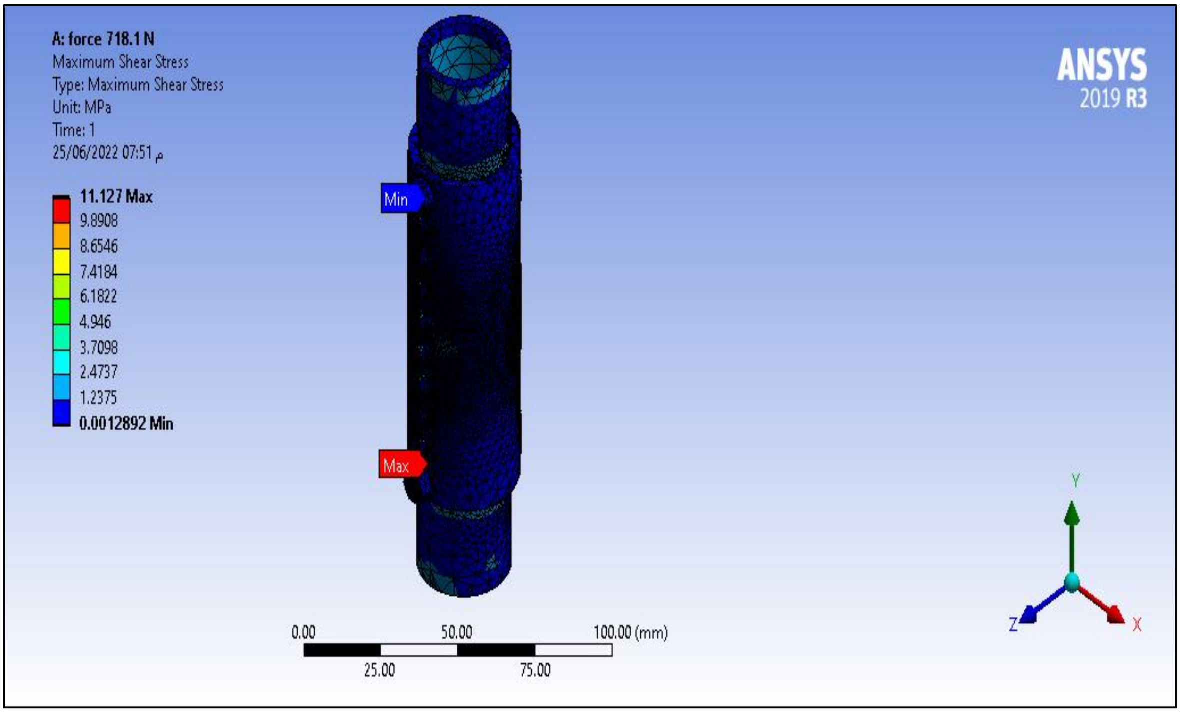


Figure 4-12: Shear stresses of shank

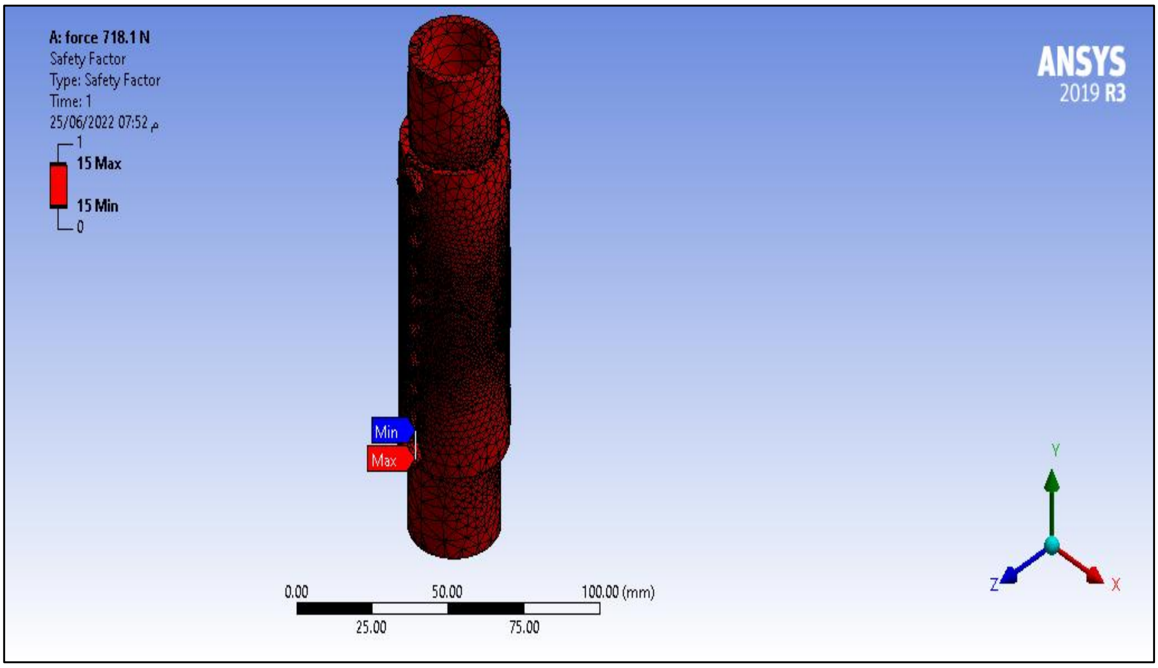


Figure 4-13: Factor of safety of shank

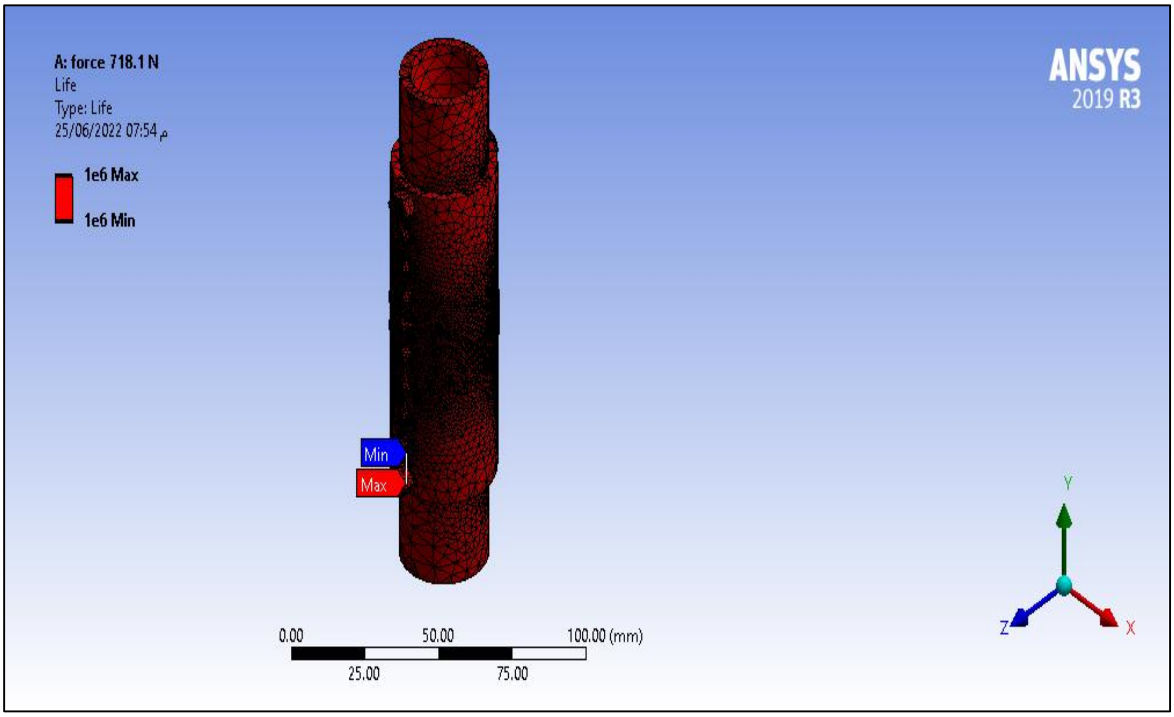


Figure 4-14: Fatigue life of shank

CHAPTER 5

CONCLUSIONS AND RECOMMENDATIONS

5.1 Introduction

In this chapter conclusions that concluded from the previous calculations, as well as the recommendations that can be applied in the future to develop this project.

5.2 Conclusions

1. The term of adjustable limb is possible to implement for a below knee prosthetic device
2. The sockets designs in this thesis achieved the adjustable purpose
3. The shank designs in this thesis achieved the term of an adjustable purpose
4. The second proposed design provide a wide range of circumference approximately 1 cm for each side, in this case it will provide comfort for the patient.
5. The holes make sockets lightweight as 346 g whole socket with adapter.
6. According to the Von-Mises theory the foot is safe, but according to safety factor the socket is unsafe because the mechanical properties of the selected materials were weak.
7. The maximum pressure of F-socket for patient with below knee amputation is (394Kpa) at the lateral region because of the patient has an

abduction in leg, there for the fibula head press on socket lateral wall higher than other regions.

8. The perlon fibers that choice need to reinforced with another materials that have good mechnical properties such as layers of carbon fibers.

5.3 Recommendations

The following suggestions are recommended for future work:

1. Study fatigue properties using a special device to know the endurance limit of socket life and the safe load that can be applied without failure.
2. The use of other materials that are lighter in weight and have better mechanical specifications, and it is preferable that they have a percentage of flexibility for socket.
3. Manufacturing the forth design of shank from metal materials and testing its tolerance when adding it to the artificial limb and wearing it by the patient.
4. Replacing the metal pieces in the first design with lighter pieces that are easy to manufacture and use in terms of changing the length of the socket.

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APPENDICES

Appendix A:

Formulas for the Stiffness of Composites with Periodic Microstructure

```
% Elastic micromechanics, isotropic fibers [Luciano & Barbero 1995].
%  Copyright: Ever J. Barbero
%             Department of Mechanical and Aerospace Engineering
%             West Virginia University, Morgantown, WV 26505-6106
%  27 Feb 2006, MATLAB 7
%  IF YOU PUBLISH WORK BENEFITING FROM THIS M-FILE, PLEASE CITE IT AS:
%  R. Luciano and E. J. Barbero, "Formulas for the Stiffness of
Composites
%  with Periodic Microstructure," Int. J. of Solids Structures, 31(21),
%  2933-2944, (1995).
%  http://www.mae.wvu.edu/barbero/feacm/Examples/Ch6ex/PMMIE.m
clear all; close all;format short g;
time = [.001;.01;.1;1;2;3;5;10;20;30;50;100;200;300];
syms s complex;
disp('Perodic Microstructure, isotropic fibers and isotropic matrix');
disp('============================================================');
%% Matrix
% elastic solution
E_0=0.592; nu_m = 0.311; E_m = E_0;

lm_0=E_m*nu_m/((1+nu_m)*(1-2*nu_m));
mu_0=E_m/(2*(1+nu_m));
nu_0=nu_m;

%%%Fiber:
isotropic


E_f    = 24.42; nu_f  = 0.443;

lm_1=nu_f*E_f/((1+nu_f)*(1-2*nu_f));
mu_1=E_f/(2*(1+nu_f));
nu_1=nu_f;



%% a    b       c       g

a=mu_1-mu_0-2*mu_1*nu_m+2*mu_0*nu_f;
b=-mu_0*nu_m+mu_1*nu_f+2*mu_0*nu_m*nu_f-2*mu_1*nu_m*nu_f;
c=(mu_0-mu_1)*(-mu_0+mu_1-mu_0*nu_m-
2*mu_1*nu_m+2*mu_0*nu_f+mu_1*nu_f+2*mu_0*nu_m*nu_f-
2*mu_1*nu_m*nu_f);
g=2-2*nu_m;

V_f = 0.6;
```

```
%%S_3   S_6     S_7
S_3=0.49247-0.47603*V_f-0.02748*V_f^2;
S_6=0.36844-0.14944*V_f-0.27152*V_f^2;
S_7=0.12346-0.32035*V_f+0.23517*V_f^2;

%%D_    C_11*   C_12*   C_23*   C_22*   C_44*   C_66*
D_=(a*S_3^2)/(2*mu_0^2*c)-(a*S_6*S_3)/(mu_0^2*g*c)+a*(S_6^2-
S_7^2)/(2*mu_0^2*g^2*c)+S_3*(b^2-a^2)/(2*mu_0*c^2)+(S_6*(a^2-
b^2)+S_7*(a*b+b^2))/(2*mu_0*g*c^2)+(a^3-2*b^3-3*a*b^2)/(8*c^3);
C_11=lm_0+2*mu_0-V_f*((S_3^2)/(mu_0^2)-2*S_6*S_3/(mu_0^2*g)-
a*S_3/(mu_0*c)+(S_6^2-S_7^2)/(mu_0^2*g^2)+(a*S_6+S_7*b)/(mu_0*g*c)+(a^2-
b^2)/(4*c^2))/D_;
C_12=lm_0+((S_3/(2*c*mu_0))-((S_6-S_7)/(2*mu_0*c*g))-
(a+b)/(4*c^2))*b*V_f/D_;
C_23=lm_0+V_f*((a*S_7)/(2*mu_0*g*c)-(b*a+b^2)/(4*c^2))/D_;
C_22=lm_0+2*mu_0-V_f*((-a*S_3)/(2*mu_0*c)+(a*S_6)/(2*mu_0*g*c)+(a^2-
b^2)/(4*c^2))/D_;
C_44=mu_0-V_f/((-2*S_3)/mu_0+1/(mu_0-mu_1)+4*S_7/(mu_0*(2-2*nu_0)));
C_66=mu_0-V_f/((-S_3/mu_0)+1/(mu_0-mu_1));

%%% Engineering Constants
%%E_1   E_2     n_12    n_23    G_12    G_23
E_1=C_11-(2*C_12^2)/(C_22+C_23);
E_2=((2*C_11*C_22+2*C_11*C_23-4*C_12^2)*(C_22-
C_23+2*C_44))/(3*C_11*C_22+C_11*C_23+2*C_11*C_44-4*C_12^2);
n_12=C_12/(C_22+C_23);
n_23=(C_11*C_22+3*C_11*C_23-2*C_11*C_44-
4*C_12^2)/(3*C_11*C_22+C_11*C_23+2*C_11*C_44-4*C_12^2);
G_12=C_66;
G_23=E_2/(2*(1+n_23));

% disp('[E_1,E_2,n_12,n_23,G_12,G_23]');
disp([E_1,E_2,n_12,n_23,G_12,G_23]);
fprintf(' E_1 = %g\n E_2 = %g\n n_12 = %g\n n_23 = %g\n G_12 = %g\n G_23
= %g\n',E_1,E_2,n_12,n_23,G_12,G_23)

%%% Restrictions on elastic constants
%%1-n12n21>0   1-n13n31>0     1-n23n32>0     1-n12n21-n23n32-n13n31-
2n21n32n13>0
%=1-(n_12.*n_12.*E_2./E_1)
%=1-(n_12.*n_12.*E_2./E_1)
%=1-(n_23.*n_23)
%=1-(n_12.*n_12.*E_2./E_1)-(n_23.*n_23)-(n_12.*n_12.*E_2./E_1)-
(2.*(n_12.*E_2./E_1).*(n_23).*(n_12))

% end
```

Appendix B:

Experimental tensile test results

Shank material (6061-T6)

First sample

Testometric
materials testing machines

winTest™
Analysis

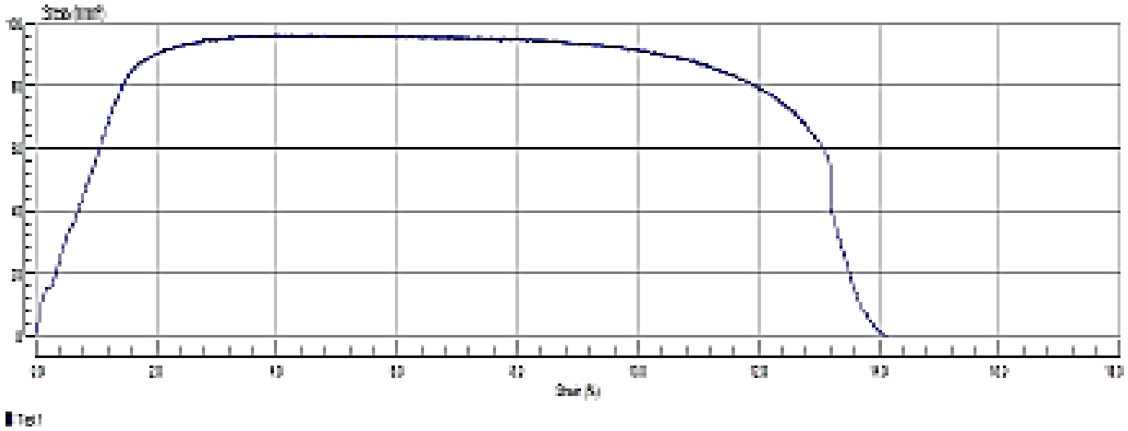
STRESS VIA STAIN
W

Ref 1 : T4
Ref 2 :
Ref 3 :

Test Name : Default
Test Type : Tensile
Test Date : 28/04/2022 08:36 ص
Test Speed : 2.000 mm/min
Pretension : Off
Width : 12.500 mm
Thickness : 3.000 mm
Sample Length : 57.000 mm

Test No	Force @ 0.000 mm (N)	Force @ Break (N)	Force @ Peak (N)	Strain @ Upper Yield (%)	Strain @ Lower Yield (%)	Youngs Modulus (N/mm²)	Stress @ Yield (N/mm²)	Stress @ Upper Yield (N/mm²)	Stress @ Lower Yield (N/mm²)
1		-160.400	3623.000	1.414	1.433	6642.465	79.520	79.520	80.053

Test No	Stress @ Break (N/mm²)	Stress @ Peak (N/mm²)	Strain @ Peak (%)	Strain @ Break (%)	Strain @ Yield (%)	Elong. @ Break (mm)	Elong. @ Peak (mm)
1	-4.277	96.613	4.233	17.414	1.414	9.926	2.413



Page 1

Unit 1 Lincoln Business Park Lincoln Close, Rochdale, Lancashire, England OL11 1NR

Tel: (44) (0)1706 654039 Fax: (44) (0)1706 646089
Email: info@testometric.co.uk website: www.testometric.co.uk

Second sample

Testometric
materials testing machines

winTest™
Analysis

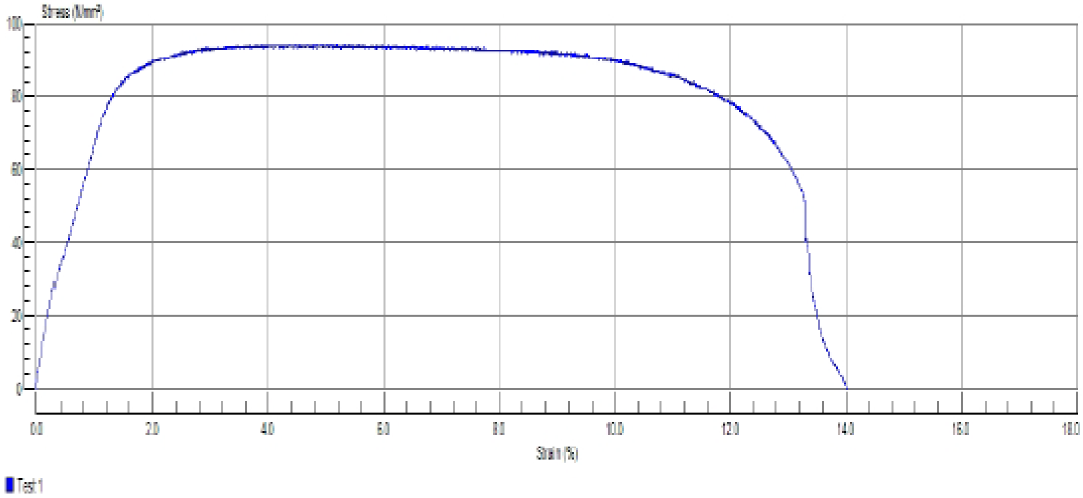
STRESS VIA STAIN
W

Ref 1 : T4
Ref 2 :
Ref 3 :

Test Name : Default
Test Type : Tensile
Test Date : 28/04/2022 08:46 ص
Test Speed : 2.000 mm/min
Pretension : Off
Width : 12.500 mm
Thickness : 3.000 mm
Sample Length : 57.000 mm

Test No	Force @ 0.000 mm (N)	Force @ Break (N)	Force @ Peak (N)	Strain @ Upper Yield (%)	Strain @ Lower Yield (%)	Youngs Modulus (N/mm²)	Stress @ Yield (N/mm²)	Stress @ Upper Yield (N/mm²)	Stress @ Lower Yield (N/mm²)
1		-217.900	3536.000	1.030	1.102	8537.214	69.893	69.893	72.373

Test No	Stress @ Break (N/mm²)	Stress @ Peak (N/mm²)	Strain @ Peak (%)	Strain @ Break (%)	Strain @ Yield (%)	Elong. @ Break (mm)	Elong. @ Peak (mm)
1	-5.811	94.293	4.667	16.216	1.030	9.243	2.660



Page 1

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Third sample

Testometric
materials testing machines

winTest™ Analysis

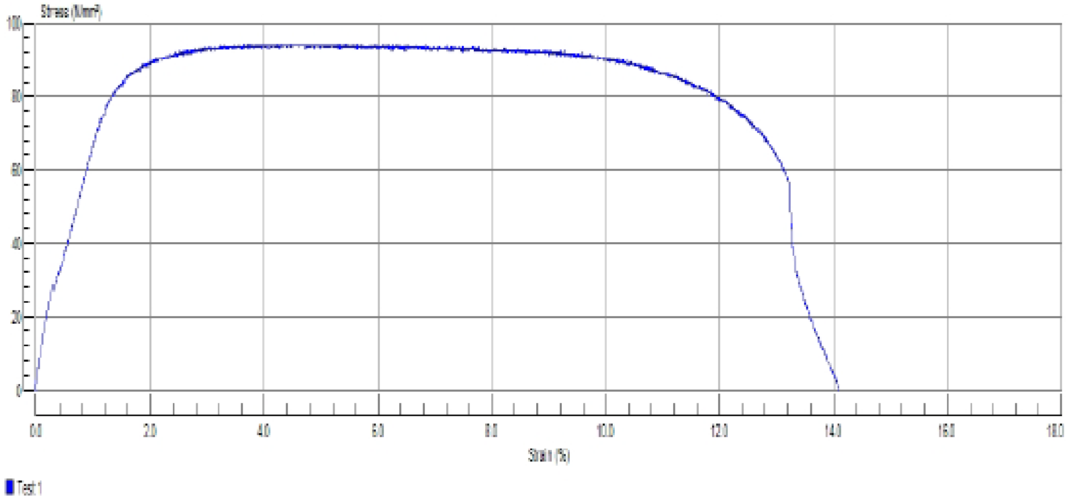
STRESS VIA STAIN
W

Ref 1 : T4
Ref 2 :
Ref 3 :

Test Name : Default
Test Type : Tensile
Test Date : 28/04/2022 08:53 ص
Test Speed : 2.000 mm/min
Pretension : Off
Width : 12.500 mm
Thickness : 3.000 mm
Sample Length : 57.000 mm

Test No	Force @ 0.000 mm (N)	Force @ Break (N)	Force @ Peak (N)	Strain @ Upper Yield (%)	Strain @ Lower Yield (%)	Youngs Modulus (N/mm²)	Stress @ Yield (N/mm²)	Stress @ Upper Yield (N/mm²)	Stress @ Lower Yield (N/mm²)
1		-228.000	3534.000	1.219	1.239	5775.950	77.413	77.413	77.387

Test No	Stress @ Break (N/mm²)	Stress @ Peak (N/mm²)	Strain @ Peak (%)	Strain @ Break (%)	Strain @ Yield (%)	Elong. @ Break (mm)	Elong. @ Peak (mm)
1	-6.080	94.240	4.218	16.711	1.219	9.525	2.404



Page 1

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Socket material (First proposed design)

First sample

Testometric
materials testing machines

winTest™
Analysis

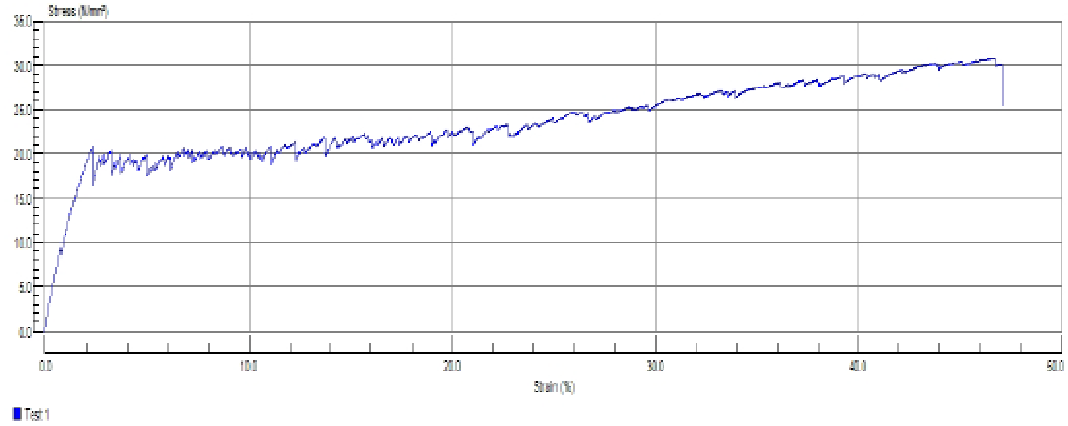
STRESS VIA STRAIN
W

Ref 1 : T4
Ref 2 :
Ref 3 :

Test Name : Default
Test Type : Tensile
Test Date : 28/04/2022 09:02 ص
Test Speed : 5.000 mm/min
Pretension : Off
Width : 13.000 mm
Thickness : 3.000 mm
Sample Length : 57.000 mm

Test No	Force @ 0.000 mm (N)	Force @ Break (N)	Force @ Peak (N)	Strain @ Upper Yield (%)	Strain @ Lower Yield (%)	Youngs Modulus (N/mm²)	Stress @ Yield (N/mm²)	Stress @ Upper Yield (N/mm²)	Stress @ Lower Yield (N/mm²)
1		991.100	1202.000	2.258	3.211	1127.431	20.813	20.813	17.433

Test No	Stress @ Break (N/mm²)	Stress @ Peak (N/mm²)	Strain @ Peak (%)	Strain @ Break (%)	Strain @ Yield (%)	Elong. @ Break (mm)	Elong. @ Peak (mm)
1	25.413	30.821	46.575	47.107	2.258	26.851	26.548



Page 1

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Email: info@testometric.co.uk website: www.testometric.co.uk

Second sample

Testometric
materials testing machines

winTest™
Analysis

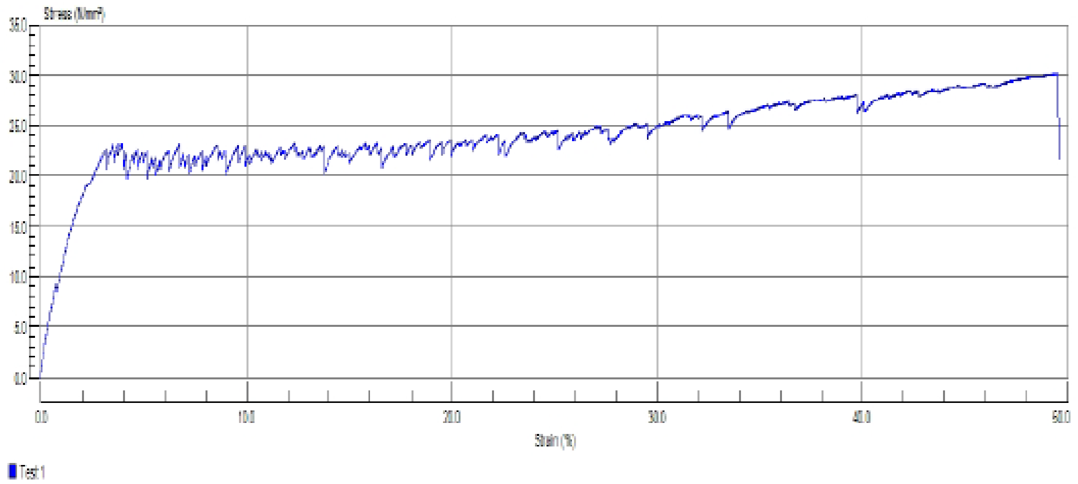
STRESS VIA STAIN
W

Ref 1 : T4
Ref 2 :
Ref 3 :

Test Name : Default
Test Type : Tensile
Test Date : 28/04/2022 10:46 ص
Test Speed : 2.000 mm/min
Pretension : Off
Width : 13.000 mm
Thickness : 3.000 mm
Sample Length : 57.000 mm

Test No	Force @ 0.000 mm (N)	Force @ Break (N)	Force @ Peak (N)	Strain @ Upper Yield (%)	Strain @ Lower Yield (%)	Youngs Modulus (N/mm²)	Stress @ Yield (N/mm²)	Stress @ Upper Yield (N/mm²)	Stress @ Lower Yield (N/mm²)
1		846.300	1174.500	1.156	1.184	1032.685	12.705	12.705	12.874

Test No	Stress @ Break (N/mm²)	Stress @ Peak (N/mm²)	Strain @ Peak (%)	Strain @ Break (%)	Strain @ Yield (%)	Elong. @ Break (mm)	Elong. @ Peak (mm)
1	21.700	30.115	49.430	49.575	1.156	28.258	28.175



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Unit 1 Lincoln Business Park Lincoln Close,
Rochdale, Lancashire, England OL11 1NR
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Email: info@testometric.co.uk website: www.testometric.co.uk

Third sample

Testometric
materials testing machines

winTest™
Analysis

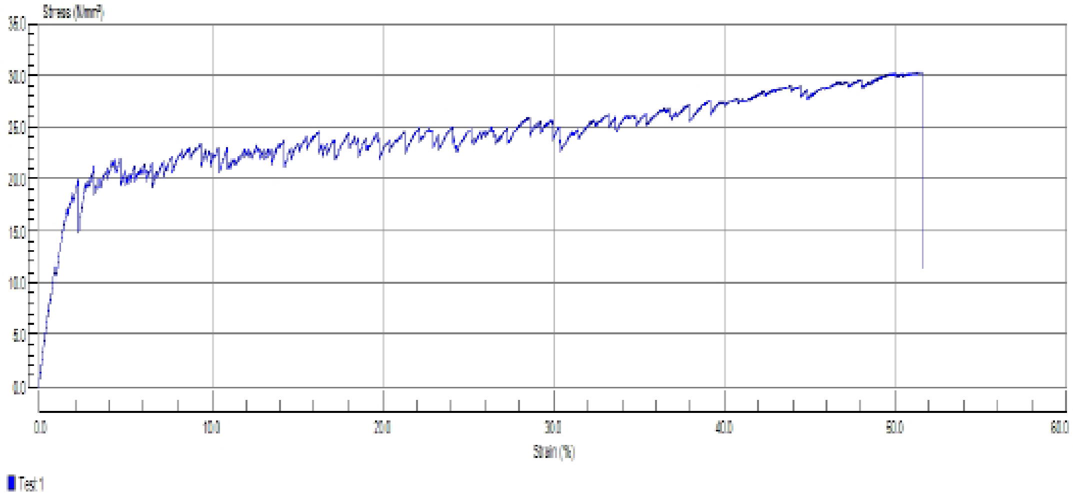
STRESS VIA STAIN
W

Ref 1 : T4
Ref 2 :
Ref 3 :

Test Name : Default
Test Type : Tensile
Test Date : 28/04/2022 11:06 ص
Test Speed : 2.000 mm/min
Pretension : Off
Width : 13.000 mm
Thickness : 3.000 mm
Sample Length : 57.000 mm

Test No	Force @ 0.000 mm (N)	Force @ Break (N)	Force @ Peak (N)	Strain @ Upper Yield (%)	Strain @ Lower Yield (%)	Youngs Modulus (N/mm²)	Stress @ Yield (N/mm²)	Stress @ Upper Yield (N/mm²)	Stress @ Lower Yield (N/mm²)
1		447.600	1183.100	1.263	1.281	1148.082	14.754	14.754	14.805

Test No	Stress @ Break (N/mm²)	Stress @ Peak (N/mm²)	Strain @ Peak (%)	Strain @ Break (%)	Strain @ Yield (%)	Elong. @ Break (mm)	Elong. @ Peak (mm)
1	11.477	30.336	51.481	51.605	1.263	29.415	29.344



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Socket material (second proposed design)

First sample

Testometric
materials testing machines

winTest™ Analysis

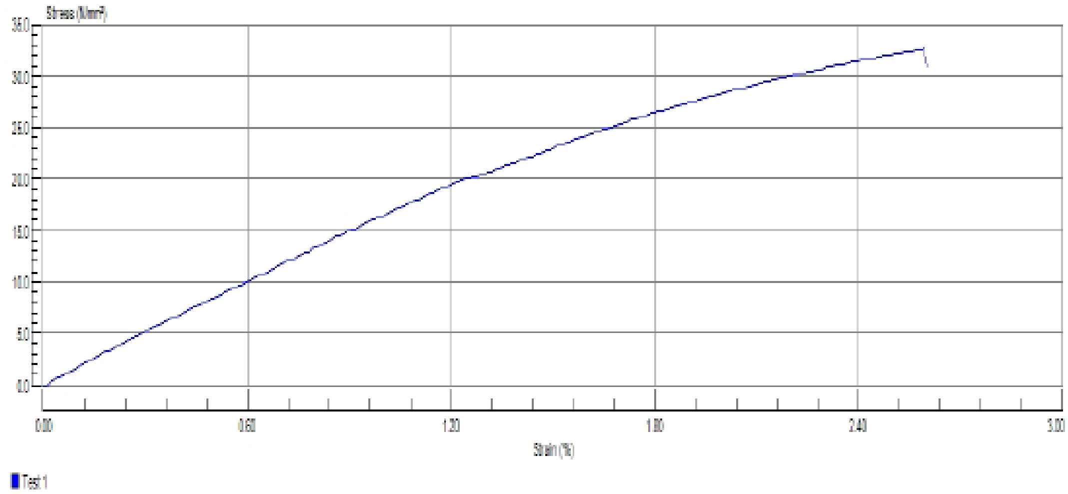
STRESS VIA STRAIN
W

Ref 1 : T4
Ref 2 :
Ref 3 :

Test Name : Default
Test Type : Tensile
Test Date : 28/04/2022 09:12 ص
Test Speed : 5.000 mm/min
Pretension : Off
Width : 13.000 mm
Thickness : 3.000 mm
Sample Length : 57.000 mm

Test No	Force @ 0.000 mm (N)	Force @ Break (N)	Force @ Peak (N)	Strain @ Upper Yield (%)	Strain @ Lower Yield (%)	Youngs Modulus (N/mm²)	Stress @ Yield (N/mm²)	Stress @ Upper Yield (N/mm²)	Stress @ Lower Yield (N/mm²)
1	-1.900	1208.900	1272.600	2.593		1634.602	32.631	32.631	

Test No	Stress @ Break (N/mm²)	Stress @ Peak (N/mm²)	Strain @ Peak (%)	Strain @ Break (%)	Strain @ Yield (%)	Elong. @ Break (mm)	Elong. @ Peak (mm)
1	30.997	32.631	2.593	2.602	2.593	1.483	1.478



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Second sample

Testometric
materials testing machines

winTest™ Analysis

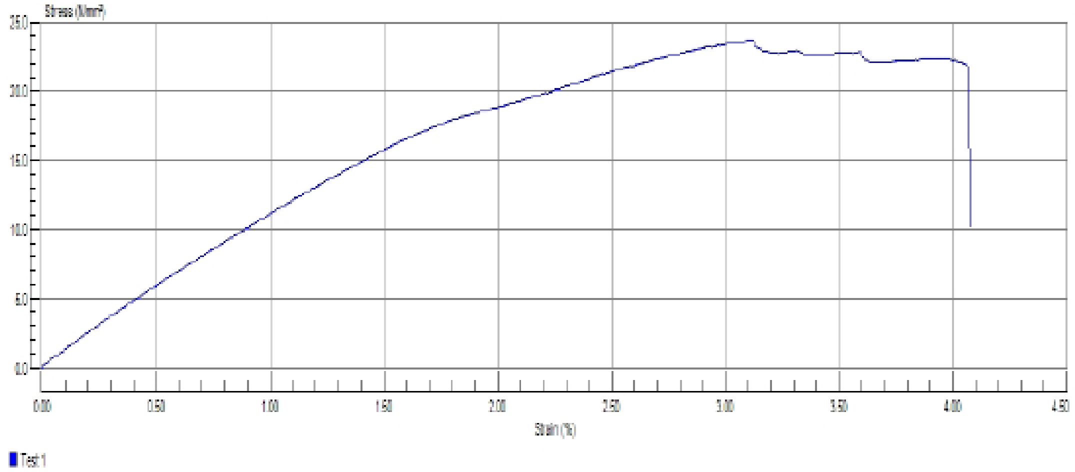
STRESS VIA STAIN
W

Ref 1 : T4
Ref 2 :
Ref 3 :

Test Name : Default
Test Type : Tensile
Test Date : 28/04/2022 09:17 ص
Test Speed : 2.000 mm/min
Pretension : Off
Width : 13.000 mm
Thickness : 3.000 mm
Sample Length : 57.000 mm

Test No	Force @ 0.000 mm (N)	Force @ Break (N)	Force @ Peak (N)	Strain @ Upper Yield (%)	Strain @ Lower Yield (%)	Youngs Modulus (N/mm²)	Stress @ Yield (N/mm²)	Stress @ Upper Yield (N/mm²)	Stress @ Lower Yield (N/mm²)
1		397.400	924.900	0.823	0.895	1044.292	9.367	9.367	10.054

Test No	Stress @ Break (N/mm²)	Stress @ Peak (N/mm²)	Strain @ Peak (%)	Strain @ Break (%)	Strain @ Yield (%)	Elong. @ Break (mm)	Elong. @ Peak (mm)
1	10.190	23.715	3.125	4.075	0.823	2.323	1.781



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Third sample

Testometric materials testing machines

winTest™ Analysis

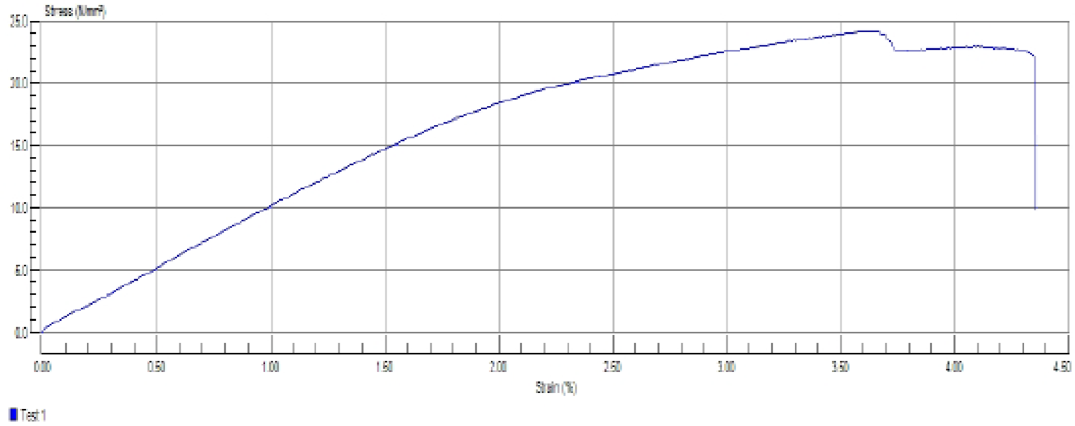
STRESS VIA STAIN
W

Ref 1 : T4
Ref 2 :
Ref 3 :

Test Name : Default
Test Type : Tensile
Test Date : 28/04/2022 09:21 ص
Test Speed : 2.000 mm/min
Pretension : Off
Width : 13.000 mm
Thickness : 3.000 mm
Sample Length : 57.000 mm

Test No	Force @ 0.000 mm (N)	Force @ Break (N)	Force @ Peak (N)	Strain @ Upper Yield (%)	Strain @ Lower Yield (%)	Youngs Modulus (N/mm²)	Stress @ Yield (N/mm²)	Stress @ Upper Yield (N/mm²)	Stress @ Lower Yield (N/mm²)
1		384.400	943.100	0.525	0.570	996.490	5.444	5.444	5.800

Test No	Stress @ Break (N/mm²)	Stress @ Peak (N/mm²)	Strain @ Peak (%)	Strain @ Break (%)	Strain @ Yield (%)	Elong. @ Break (mm)	Elong. @ Peak (mm)
1	9.856	24.182	3.656	4.356	0.525	2.483	2.084



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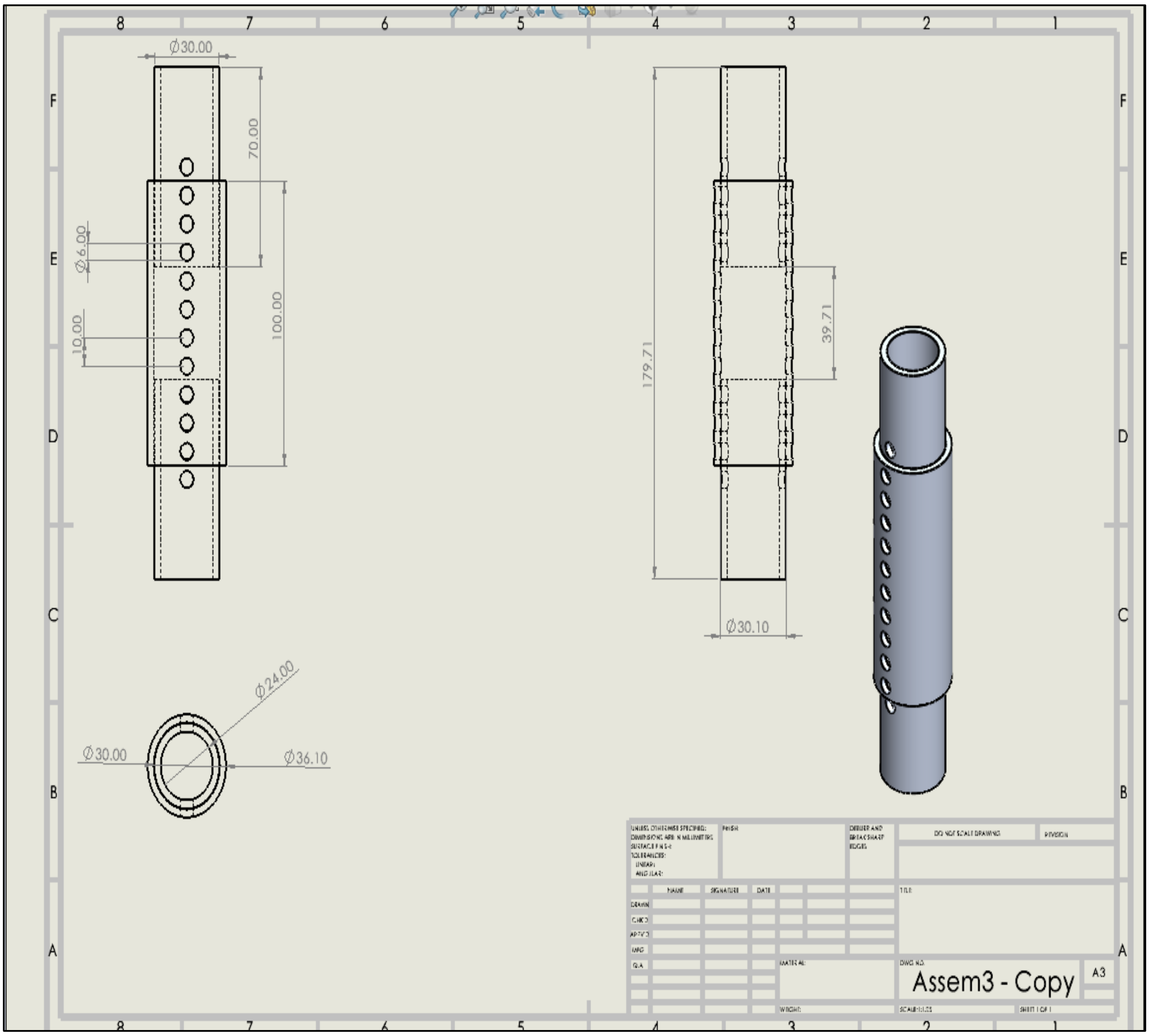
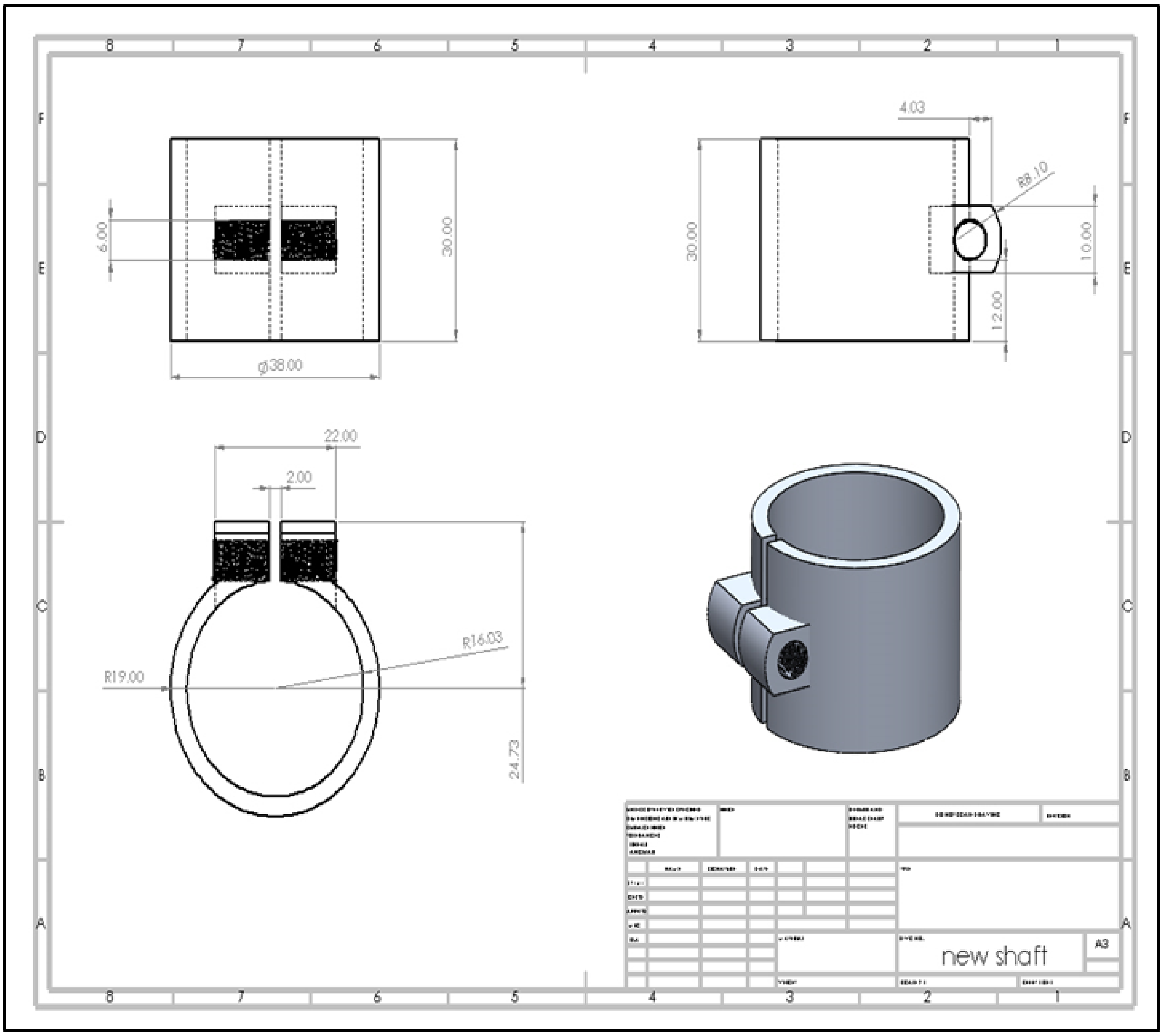
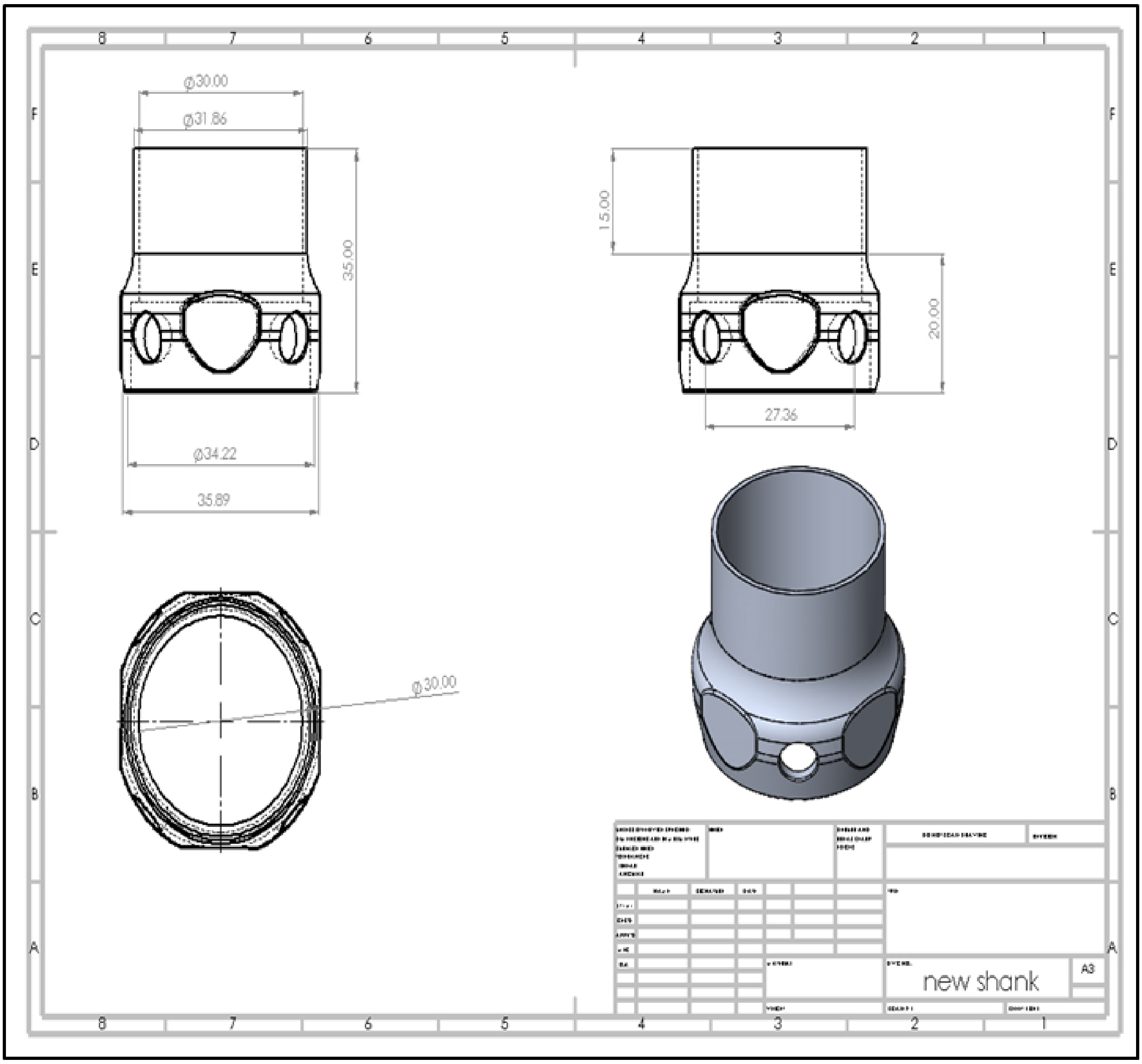
Shanks measurement





8
7
6
5
4
3
2
1
F
E
D
C
B
A
6.00
30.00
⌀38.00
4.03
R8.10
30.00
10.00
12.00
22.00
2.00
R16.03
R19.00
24.73
new shaft
A3


ø30.00
ø31.86
35.00
ø34.22
35.89
15.00
20.00
27.36
ø30.00
new shank
A3

إهداء

﴿وَمَنْ أَحْيَاهَا فَكَأَنَّمَا أَحْيَا النَّاسَ جَمِيعًا ۚ﴾

{المائدة :٣٢}

إلى الذين قست عليهم الحرب وأخذت جزءً غالياً منهم ....

إلى أبي سيادة النقيب الطيار ....

إلى أمي التي كانت ولا تزال معلمتي الأولى ....

إلى مشرفي د. ياسر يعرب لإختياره هذا المشروع ....

إلى أساتذتي الكرام ....

إلى الفرص الثانية ... الى الاصرار والنهوض بعد كل سقوط ؛ إلى المواقف واللحظات التي تجعلنا نتمسك باحلامنا ونجتهد لتحقيقها بأفضل الطرق.

إلى جميع من ذكرت أُهدي هذا البحث

الخلاصة

اعداد مبتوري الأطراف في العراق كبير وقابل للزيادة و هذا يعود الى اسباب كثيرة منها العيوب الخلقية وآثار الحرب والمتفجرات والحوادث. وعادة ما يمثل عدد كبير من عمليات البتر تحت الركبة نسبة كبيرة من عمليات البتر التي تصيب الانسان.

الغرض من هذا البحث هو تصميم وتصنيع واختبار الطرف الاصطناعي القابل للتعديل تحت الركبة لطفل يبلغ من العمر ٨ سنوات ويعاني من بتر تحت الركبة (BK) ، والذي يمكن تعديله لتسهيل نمو الطفل. تم إجراء تصميم وتصنيع واختبار الطرف السفلي الاصطناعي من خلال ثلاث مراحل. كانت المرحلة الأولى هي إجراء تصميم نظري وعددي للوقب القابل للتعديل والساق من خلال برنامج 3D SolidWorks لتصميم الأجزاء ، ثم برنامج ANSYS لتحليل النموذج وحساب الضغوط المكافئة (Von Mises) وعوامل الأمان وتوزيع التشوه . المرحلة الثانية تمثلت بالعمل التجريبي بإختيار المواد. لصنع نموذجين من الوقب (وقب قابل للتعديل) تم اختيار المواد المركبة لصنع النموذجين (١٠ طبقات من البيرلون + ٣٠٠ جم من راتنج الأكريليك). بينما تم اختيار مادة الألومنيوم (T6-٦٠٦١) لنموذج الساق القابل للتعديل. تم إجراء اختبار الشد للحصول على الخواص الميكانيكية لمواد الوقب والساق. بعد تجميع الأجزاء الاصطناعية من BK التي تم تصنيعها في مركز الأبتسام و عمل المحاذاة للطرف، تبدأ المرحلة الثالثة بإجراء اختبار ضغط السطح البيني (IP) واختبار قوة رد فعل الأرض (GRF). كانت نتائج اختبار الشد لـ (راتنج أكريليك + بيرلون) $\sigma ult = 30.424$ ميجا باسكال ، $\sigma y = 16.089$ ميجا باسكال و $E = 1.1027733$ جيجا باسكال للتصميم الأول المقترح و $ult = 26.843$ ميجا باسكال ، $\sigma y = 15.814$ ميجا باسكال و $E = 1.225128$ جيجا باسكال للتصميم الثاني المقترح. كانت نتائج اختبار الشد للساق $\sigma ult = 95.049$ Mpa، $\sigma y = 75.609$ Mp و $E = 6.985216$ Gpa. كان أقصى ضغط في التجويف أسفل الركبة 394 كيلو باسكال في المنطقة الجانبية الخارجية من الساق. بينما كانت اقل نسبة ضغط في المنطقة الخلفية للوقب حيث بلغت مقدار 101 كيلو باسكال. كانت اختبارات تحليل المشي جيدة. الضغط المركزي في القدم للطرف الصناعي بلغ 0.18 ميكاباسكاليوفر التصميم مزيدًا من الراحة وخفة الوزن بالإضافة إلى إمكانية زيادة حجم التجويف والصبغة أثناء النمو ، فضلاً عن سهولة الخلع وارتداء التجويف.

وزارة التعليم العالي والبحث العلمي
جامعة النهـــــرين/ كلية الهندســـة

تصميم وتصنيع طرف صناعي قابل للتعديل أسفل الركبة للأطفال

رسالة مقدمة

الى كليـــة الهندســـة في جامعة النهرين

وهي جزء من متطلبات نيل شهادة بكلوريوس علوم

في

هندسة **الأطراف و المساند الصناعية**

من قبل

عائشة محمد مظهر

اشراف

أ.م.د. ياسر يعرب قحطان

ذو القعدة ١٤٤٣هـ

حزيران ٢٠٢٢مـ