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Investigation of fatigue strength and stiffness / weight ratio of knee disarticulation prosthetic socket

A Thesis
Submitted to the Mechanical Engineering Department at College of Engineering /University of Kerbala, in Partial Fulfillment of The Requirements for the Degree of Master of Scienc in Mechanical Engineering
(Applied Mechanics)

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2018

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Ameer Alaa
2018

DEDICATION

To My

Father & Mother …

Beloved Wife …

Lovely Son (Mustafa)

Sincere Sisters

And All People Who Love Me and All That I Love Deeply.

Ameer Alaa
2018

Abstract

Prosthesis is an artificial part that replaces a missing body part, lost due to trauma or congenital defect. People with knee disarticulation (through knee amputation (TKA)) are prone to have their socket failed under different conditions. Various reasons are responsible for that, such as the sort of material used in producing the socket the conditions of load exerted on the prosthetic socket. In this research, composite materials are suggested in order to modify the prosthetic socket for the sake of extending its life and increasing the comfort for its user.

This study included two major parts; experimental and numerical parts. The experimental part included two major types of tests. The first type of tests concerned about performing mechanical tests on the socket materials including available and suggested materials used in socket lamination. The available material consists of laminated (4 perlon 2 fiber glass 4 perlon) layers, and the suggested material consists of (3 perlon 1 carbon fiber 1 perlon 2 carbon fiber 1 perlon 1carbon fiber 3 perlon) layers. Suggested materials provide better physical and mechanical properties, such as improving stability and providing equilibrium for the TKA socket as compared to sockets that are currently available, while nut palm powder does not give any significant effect on the strength of material. The second type of tests dealt with the 27-years old male patient whose weight is 80 kg and height of 180 cm, in order to measure the Ground Reaction Force (GRF), and pressure distribution for both suggested and available prostheses. F-Socket sensor is used to measure the interface pressure between stump and socket. The calculations of the stiffness to weight ratio showed that the suggested socket has a value higher than of the available one for axial (204.5%), bending (189%) and torsion (182.7%).

The second part of this work is the numerical part which involves the analytical and numerical investigations that focus on simulation of through knee prosthesis to calculate the equivalent von-Mises stresses and safety factor by performing finite element analysis conducted through ANSYS 14.5 software. The interface pressure between stump and socket was obtained from, both, analytical and experimental work. F-socket as found experimentally and used in simulation.

For the suggested composite material, the results showed that the maximum equivalent von- Mises stress is equal to 14.43 MPa for the experimental work (F-socket) which is very close to that obtained from the analytical work (15.42 MPa). It was observed that the lamination of the suggested composite material provided better results in terms of high safety factor in experimental and analytically which are 4.83 and 4.93 respectively, low deformation and comfortable feeling gained by the patient during locomotion compared to available materials.

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Nomenclatures

Symbol	Definition	Unit
A	Area supporting surface of skin	m^2
D_l	Lower socket diameters side of the support surface	mm
D_u	Upper socket diameters side of the support surface	mm
d	Vertical displacement	mm
d_n	Normal spring displacement	mm
d_{no}	Pre-compressive displacement of the normal spring	mm
d_s	Parallel spring displacement	mm
E_s	Young's modulus of soft tissue	kPa
E_b	Bending modulus of elasticity	GPa
G	Shear modulus	kPa
H	Vertical distance	mm
K	Vertical interface stiffness	N/mm
K_n	Normal spring constants	N/mm
K_s	Parallel spring constants	N/mm
L	Length limb contacts with socket	mm

ℓ	Length fatigue specimen	mm
m	Mass specimen	g
N	Normal force	N
N_f	Number of cycle	rpm
P_o	Average pre-pressure at skin/prosthetic socket interface	kPa
S	Shear force	kPa
t	Average thickness of soft tissue	mm
t_a	Average thickness for fatigue specimen	mm
V	Volume specimen	m^3
W	Vertical force (weight)	N

Greek Symbols

Symbol	**Definition**	**Unit**
$\boldsymbol{\sigma_{ult}}$	Ultimate strength	MPa
$\boldsymbol{\theta}$	Conical angle	degree
μ	Coefficient of friction	-
$\boldsymbol{\upsilon}$	Poisson's ratio	-
$\boldsymbol{\sigma_e}$	Stress endurance limit	MPa
$\boldsymbol{\delta}$	Deflection	mm
$\boldsymbol{\sigma_a}$	Alternating stress	MPa
$\boldsymbol{\upsilon_f}$	Volume fraction	-
$\boldsymbol{\rho}$	Density	Kg/m^3
$\boldsymbol{\sigma_c}$	Resultant of shear and normal stresses	MPa
$\boldsymbol{\tau}$	Shear stress acting on the extremity surface	kPa
$\boldsymbol{\sigma}$	Average pressure acting on the extremity surface	kPa

Abbreviations

AK	Above Knee
ASTM	American Society for Testing and Materials
BK	Below Knee
BKA	Below Knee Amputee
BRFEC	Bamboo Fiber Reinforced Epoxy Composites
CAD	Computer Aided Design
CAM	Computer Aided Manufacture
CNC	Computer Numeric Control
CFRP	Carbon Fiber Reinforced Plastics
COM	Center Of Mass
COP	Center Of Pressure
FE	Finite Element
FEM	Finite Element Method
FGP	Fiberglass with Polyester
GFRP	Glass Fiber Reinforced Plastics
GRF	Ground Reaction Force
ICRC	International Committee of the Red Cross
ISO	International Standards Organization
MSS	Modular Socket System
O&P	Orthotic and Prosthetic
PE	Poly-Ethylene
PVDF	Polyvinylidene Fluoride

PMMA	Poly-Methyl Meth-Acrylate
PVA	Poly-Vinyl Acetate
PVC	Poly-Vinyl Chloride
RP	Ramie fiber with Polyester
RFER	Rattan Fibers with Epoxy
RE	Ramie fiber with Epoxy
TKA	Through-Knee Amputation
TK	Through-Knee

Chapter One

Introduction

1.1 Preface

Prosthetics of the most important areas to be studied in the Arab world in general and the Iraqi particularly, as long as we are in our problems and political endless wars and committed itself not departed as long as we are suffering from the cycle of terrorism unfortunate, in addition to the thousands of mines scattered along hundreds of kilometers of Arab land. Arabs requires us to prepare for the consequences. And pay attention to the children of our species to melt our science in the crucible of love and feeling with which they suffer the loss of their organs. To serve and assist them. [1]

There is every million people in the world about 390 people lost limbs, and have or may Do not have the prostheses alternative, and in the European Union countries are close to 156 thousand people live prosthetic limbs. Unfortunately, still mill wars increase of this global tragedy [2]. Prostheses are used to take place missing parts from a human body that has trauma, cancers or infections (surgically removed), or people were born with incomplete parts (congenital). The missing part could be an ear, nose or an eye. The term can also be used to describe replacing missing bone or muscle under the skin (an implant).

There is a real need for prosthetics to enable the amputees to do their daily activities normally. Therefore, prosthetics should be developed and improved to help physical challenged people to live their life more easily.

Amputation level refers to the portion of the body where the amputation exists. There are several levels of amputations which are summarized in Figure (1.1) for upper limb amputations.

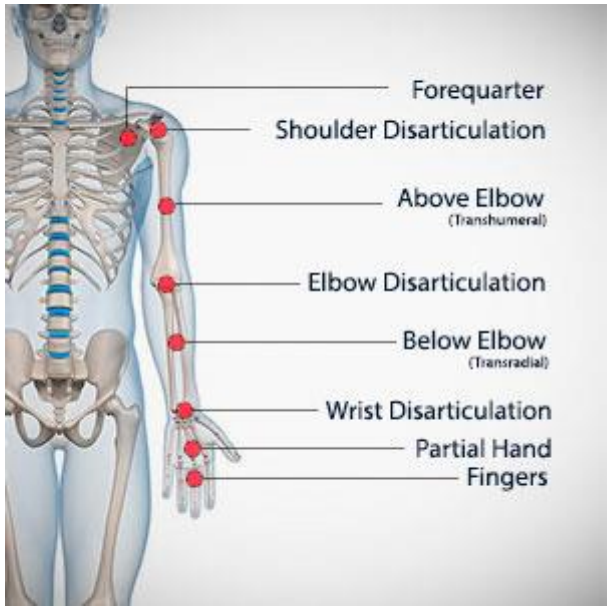


Figure (1.1) Amputation levels for upper limbs[3].

while the lower limb amputations are summarized in figure (1.2). Disarticulation through the knee has several advantages over above-knee amputation. In particular, the long end-bearing stump facilitates balance and control of the prosthesis. Disadvantages are the relative slow healing of the wound, lack of an internal knee mechanism in the artificial limb and the bulky appearance of the limb. The results of rehabilitation could be improved by careful selection of patients and attention to operative details [4].

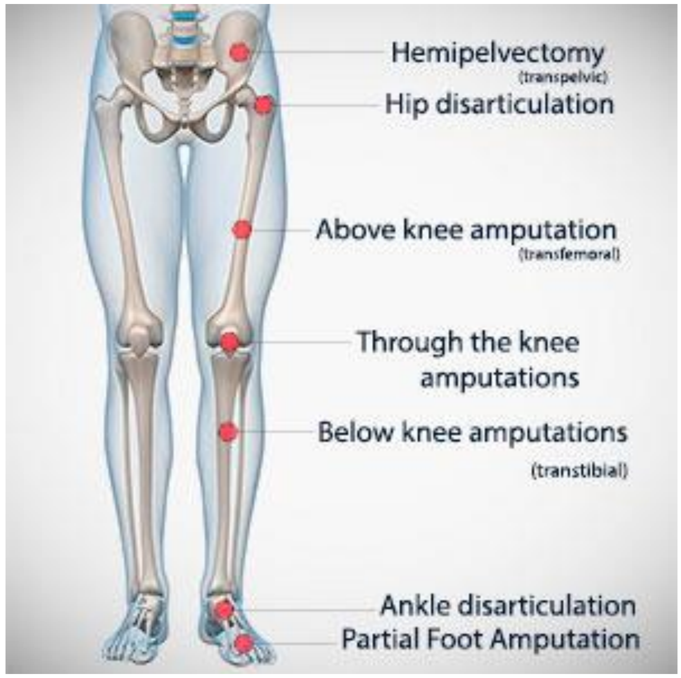


Figure (1.2) Amputation levels for lower limb[5].

1.2 Disarticulation Amputation

In disarticulation amputation (Through knee amputation, TKA), the normal knee function is completely lost by the amputation level. However, the physical differences might be important consideration in the prosthesis procedure. The end of the stump is composed of tissue normally adapted to weight bearing in the kneeling amputated patient. From a biomechanical point of view, this problem is very similar to those of the above knee position. A long lever arm is available for the exertion of control forces by the hip operating in a physiological condition muscles and the muscles themselves are for the most part intact [6].

1.3 Knee Disarticulation Prosthesis

In general, the lower limb function is to help the patient do several functions, including walking and standing. A prosthesis used for patients having through knee amputation generally consists of four essential parts: socket, knee joint, shank and foot, Figure (1.3) [7]. Each part has a special function detailed as follow.

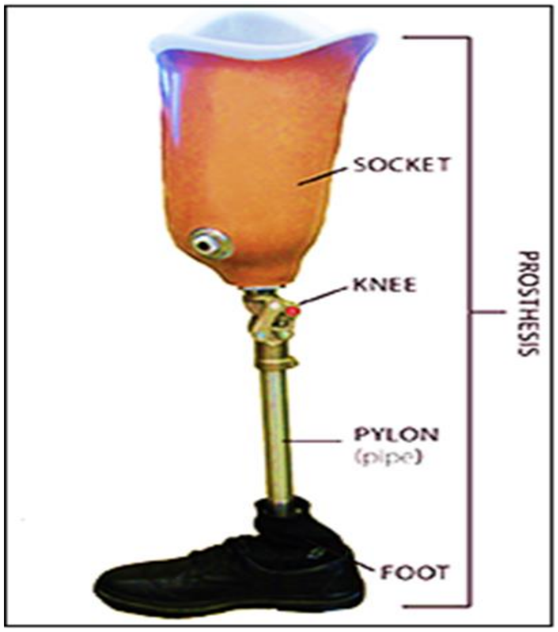


Figure (1.3): Through Knee (TK) prosthesis [7]

1.3.1 Socket

The socket is a part that connects the prosthesis with the patient body. Thus, it must possess a strong resistance to fatigue and creep and should be as light as possible and lack of skin sensitization. That properties of a socket are guaranteed make a patient feel comfortable [8].

Previously, sockets were produced from wood and formed from aluminum. These days, manufacturing technology has progressed allowing them to be made from fiberglass and polyurethane which are much lighter and stronger materials compared to wood and aluminum. Polymer and fiberglass sockets can be easily molded to the leg contours and that leads to less discomfort of the patient and allows a better fitting prosthetic[9].

The socket surrounds the stump, so it should be shaped to distribute forces over the entire surface area of the stump equally. The socket is aligned on the pylon with the specific purpose of increasing the load on the patellar ligament. With minimum contact pressure placed over sensitive areas such as the tibia condyles, fibular head and the enter distal tibia [10]. The socket manufacturing may be achieved in three methods:

1- Sockets made from polypropylene which is adopted by the International Committee of the Red Cross (ICRC). The shortcoming of this material is the change of mechanical properties with time [11].

2- Sockets made from the composite material are used in rehabilitation centers. For example Babylon Center of an Artificial Limb located in Babylon City use the composite material for sockets. The shortcoming of this method is the high stiffness which causes discomfort to the patient.

3- Using the new Modular Socket System (MSS) to produce a prosthetic socket directly on the patient has the potential of being easier and faster to produce, but it is expensive [12].

1.3.2 Knee Joint

The prosthetic knee has several functions. It permits sitting, kneeling, and other similar activities. It also allows a controlled movement of the pylon and foot during swing phase at any gait speed and on any type of surface. In addition, it provides stability during the stance phase. Knee joint can be classified into three types: mechanical, hydraulic or pneumatic, and microprocessor knee. Moreover, knee joint can be subdivided into those that provide control only over stance phase of gait, those that provide control of both swing and stance phase and those that provide control of swing phase [13].

1.3.3 Pylon (shank)

Pylon is a metallic rod that connects the foot to the knee joint. It is usually made of lightweight materials. Some pylons are made of two rods in order to offer additional shock absorption particularly with high-level activity. Pylon allows amputees to run, walk and use stairs easily [13].

1.3.4 Ankle and Foot

The prosthetic foot is the part that acts as a normal foot. Ideally, it should mimic all the activities of a normal foot. It should be able to function in: shock absorbing, muscle activity simulation and a stable base of support [13].

1.4 Gait Cycle

The gait cycle is defined as heel strike to heel strike of the same foot while walking. The stance phase and swing phase are the major two phases of the gait cycle. The time between the heel striking the ground and the toe of the same foot leaving the ground, which includes about 60% of total gait cycle time, is stance phase. While the time of the toe leaves the ground to the time the heel of the same foot strikes the ground, which includes about 40% of the total gait cycle time, is the swing phase [14]. Gait cycle of a person is normally comprised of the following steps [from 15 - 17], as shown in Figure (1.4).

- The **stance phase** events are as follows:

1) Heel strike commences the gait cycle, and it is considered the point at which the body's center of mass is at its lowest position.

2) Foot-flat is when the plantar surface of the foot becomes in contact with the ground.

3) Mid-stance happens when the swinging (contra-lateral) foot passes the stance foot, and the body's center of mass is at its highest position.

4) Heel-off happens when the heel leaves the ground and push off is commenced via the triceps surae muscles, which plantar flexes the ankle.

5) Toe-off terminates the stance phase once the foot is not in contact with the ground

- The swing phase can be summarized as follows:

6) Once the foot loses contact with the ground, acceleration begins, and the subject activates the hip flexor muscles to accelerate the leg forward.

7) Mid-swing happens when the foot passes right beneath the body, coincidental with mid-stance for the other foot.

8) The muscles slow the leg and stabilize the foot which represents the preparation step for the next heel strike.

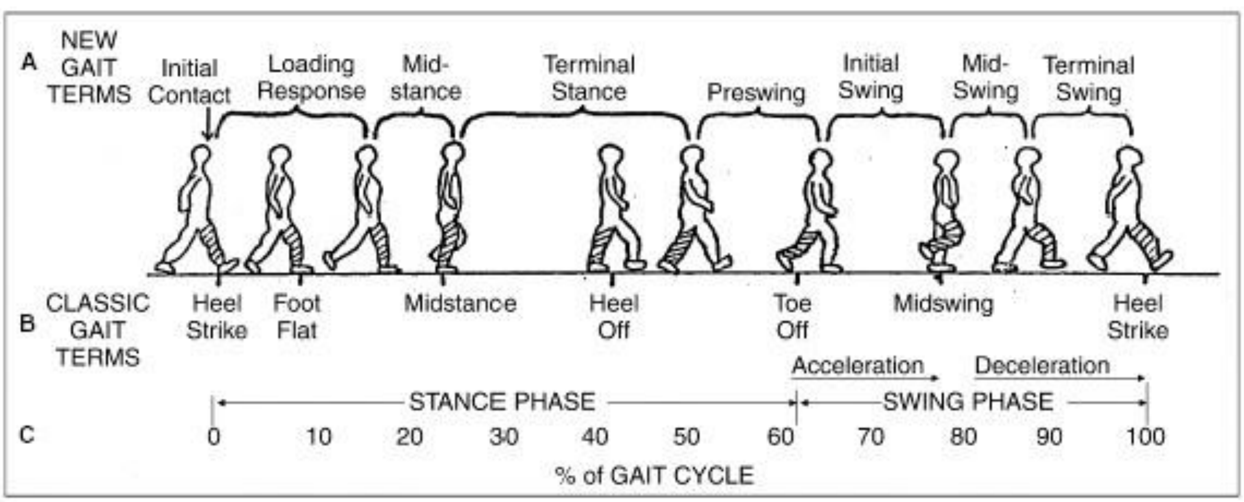


Figure. (1.4): The general steps of human gait cycle[18].

1.5 Outlines

The main objective of the current work is to choose a composite material for a through-knee prosthetic socket. In order to achieve the above objectives, the following steps are followed:

1. An experimental procedure is conducted to cover the material selection, testing and the stacking sequence for the specimen composite materials to be tested.
2. The suggested composite materials in this work are tested using tensile and fatigue tests, for the specimen prepared in step (1).
3. On the basis of the results in step (2), a prosthesis was manufactured for a patient using available socket.
4. To calculate the interface pressure between the stump and the socket, a force plate analysis was conducted for both suggested and available sockets.
5. A numerical analysis, using the FEM, was conducted for both sockets for comparison purposes.
6. Analysis of the results and conclusions were set.

Chapter Two

Literature Review

2.1 Preamble

Prosthetic care goes back to the fifth Egyptian Dynasty (2750-2625 B.C.); archaeologists have unearthed the oldest known splint from that period. The earliest known written reference to an artificial limb was made around 500 B.C., Herodotus wrote of a prisoner who escaped from his chains by cutting off his foot, which he later replaced with a wooden substitute. An artificial limb dating from 300 B.C., was a copper and wood leg unearthed at Capri, Italy in 1858. In 1529, French surgeon, Ambroise Pare (1510-1590) introduced amputation as a life saving measure in medicine. Soon after, Pare started developing prosthetic limbs in a scientific manner. In 1863, Dubois L Parmelee of New York City made an improvement to the attachment of artificial limbs. He fastened a body socket to the limb with atmospheric pressure. He was not the first person to do so, but he was the first person to do so with satisfactory results. In 1898, Dr. Vanghetti invented an artificial limb that could move with through muscle contraction. In 1946, a major advancement was made in the attachment of lower limbs. A suction sock for the above-knee prosthesis was created at University of California (UC) at Berkeley. In 1975, Ysidro M. Martinez' invention of a below-the-knee prosthesis avoided some of the problems associated with conventional artificial limbs. Martinez, an amputee himself, took a theoretical approach in his design. He did not attempt to replicate the natural limb with articulated joints in the ankle or foot which is seen by Martinez as causing poor gait. His prosthesis has a high center of mass and is light in weight to facilitate acceleration and deceleration and reduce friction. The foot is considerably shorter to control acceleration forces, reducing the friction and pressure [19].

2.2 Literature Review on Through-Knee Prosthesis

A prosthetic socket links between the residual limb and the components of prosthetic that enable amputated people's legs to walk. The current selection of accepted materials of the prosthetic socket has been approved historically without extensive evaluation and/or characterization. This is especially true concerning the effects of agreed fabrication processes on this If the materials consideration that made the prosthetic socket of acceptable mechanical properties because these properties assume a vital part in the nature of resulting prosthetic sockets, the loss of this material knowledge becomes significant [20].

Botta P. and Baumgartner R (1983) [21], studied socket design and manufacturing technique for Through-Knee (TK) stumps, which based on 14 years' experience with 290 prostheses. The shape of the stump and full end-bearing quality needs a socket which has very little similarity to above-knee sockets. The requirements of the TK socket are listed, and the manufacturing technique is described and clarified in details. Confirmation is put on the nature of the negative mortar shape. The socket produced by double wall method ordinarily utilized in below-knee gives a maximum of comfort and cosmetic without additional costs.

HUGHES J. (1983) [22], investigated the biomechanics of the through-knee prosthesis in details and comparisons were made with the above-knee (AK) case. Socket shape and suspension are the two parameters that effect on the functionality of knee in both stance and swing stages. The TK amputee is fundamentally the same as biomechanically to the AK. The Socket issues are evidently fairly lessened with great end-bearing and large stabilizing area. The functional replacement, which is required for the loss of the knee joint, presents the same problem as for the AK. That treatment leads to enhance hip control potential, but it has limitations in the provision of mechanical devices in terms of the relative physical relationship of stump and prosthesis.

Jweeg M. and J.S. Jaffar (2017) [23], analyzed through knee amputation prosthesis. Two samples were used in their research. First sample is available prosthetic for a person’s stump through knee joint which consists of 8 layers (3perlon 2 fiberglass 3 perlon). The second sample is a suggested prosthesis which consists of 10 layers (4perlon 2 carbon fiber 4 perlon). Comparison between these two samples was conducted in term of mechanical properties and fatigue life. Results showed that the lamination which was layup from ten layers (4perlon 2 carbon fiber 4 perlon) gives the optimum mechanical properties.

2.3 Literature Review on Prosthetic Socket Design and Manufacturing:

Faulkner, V. et al. (1988) [24], improved a computer-aided manufacture/ computer-aided design (CAM/CAD) system for a light weight plastic prosthesis which is capable of being manufactured remotely from the site of assessment of the amputee. This includes the improvement of a non-contact imaging device that is capable of capturing a three-dimensional topographical is recorded of the amputee's residual limb, the improvement of a computer software system that would permit designing a prosthetic above knee attachment that is biomechanically correct for a generation by a numerically computer controlled processing machine was achieved, the improvement of delivering molding tools for manufacturing of light weight prosthetic plastic components was obtained and finally, the study guided for building up the field of testing the light weight all plastic system.

Faulkner, V.W. and N.E. Walsh (1988) [25], utilized a computed tomography (CT) scan on an amputee's knee residual limb. The scanned area begins around 5cm above the knee joint space and completion at the distal end of the residual limb. These data were put away on a magnetic tape obtained from GE9800 scanner.The CEMAX1000 computer system was used to read the tape to create a three-dimensional picture of the residual limb. Utilizing a standard programming design for the CEMAX1000 system for the three-dimensional picture is reconfigured utilizing normal prosthetic biomechanical contemplations.

Current, T. et al. (1999) [26], measured the structural strength of different trans-tibial composite sockets according to International Standards Organization (ISO) for structural testing of lower limb prostheses (ISO 10328). Ten sockets were assessed. Two resin types and five different reinforcement materials were used to build the sockets. Resin materials were carbon acrylic and resin acrylic resin. Reinforcement materials were carbon-fiberglass stockinet, unidirectional carbon, fiberglass stockinet, fiberglass cloth and carbon fiber cloth. They found that carbon reinforcements were best from fiberglass reinforcements with both resins. Also, they found that the greatest ultimate strength with the unidirectional carbon reinforcement with acrylic resin amount to (3160 N).

Erin Strait (2006) [27], concentrated on the design of the socket. Wooden and leather sockets were used to be extremely prominent with tedious issue and requirement a lot of aptitude. The wooden socket is cut from a square of wood and connected to the pylon by conventional strategies. The metal/leather limb was made of metal bars, a wooden pole for the pylon and a thick bit of cowhide for the socket. The socket was conformed to a positive model of the lingering limb by extending wet leather over a positive cast leather, which is self-alleviating and revising and can be balanced for volume changes. The socket was then joined to the designed metal foundation which itself is appended to the wooden pole. The foot was developed in a similar way and utilizing an indistinguishable material from the bamboo/plaster and the Polyvinyl Chloride (PVC)/ plaster limb utilization.

Neama H. et al. (2007) [27], discussed the comparison between laminated and polypropylene which are used in manufacturing lower limb prosthetic socket. This review was directed to compare the flexural stiffness (K_b) and compressive (K_c) of five prosthetic sockets made of various materials with different thickness polypropylene, polypropylene ,perlon (10-layers) with Orthocryl resin ,Nyglass (8-layers) with Orthocryl resin and (2perlon, 2nyglass, 2fiberglass, 2nyglass, 2fiberglass and 2perlon) with polyester resin. They found that laminate gave preferred the compressive stiffness compared with polypropylene. However, polypropylene gave good the flexural stiffness compared with laminate. As a result, they recommend the use of polypropylene due to its good mechanical properties and cheap when compared with laminates.

Jweeg M.J, et al. (2009) [28], studied the mechanical properties of the above-knee prosthetic socket. The end goal of the study is to set up the premise of the mechanical properties of utilized materials for the fabricate this kind of the prosthetic socket. Tensile test was performed to discover (ultimate tensile stress, yield stress, stiffness and Young's modulus). Furthermore, a bending test was made to discover (maximum bending strength and flexural Young's modulus) on samples for 14 types of materials comprised of various layers of perlon and fiberglass as support materials with resin as matrix material. The ideal mechanical and physical properties were achieved when the laminations were consisted of (3 perlon 2 fiber glasses 3 perlon). The mechanical properties were as per the following ($\boldsymbol{\sigma}_{y}$, $\boldsymbol{\sigma}_{ult}$, **E** , $\boldsymbol{\sigma}_{flex}$, $\boldsymbol{E}_{flex}$) equivalent to (21 MPa, 26 MPa, 2 GPa, 48.65 MPa, 2.89 GPa) respectively.

Shaymaa. H et al. (2010) [29], used two composite materials for fabricating lower-limb prosthetic. The first composite is a thermosetting resin – polyester while a thermoplastic resin poly methyl meth acrylate (PMMA) is used for the second sample. The fiber reinforcement layers which are consisted of 4 perlon layers and 8 fiberglass layers were used for both composite materials. Tensile and fatigue tests were conducted on specimens for two matrix materials. The results showed that the polyester composite and polyester matrix material have better fatigue strength and tensile properties than PMMA composite and matrix respectively. The modulus of elasticity, ultimate tensile strength and the fatigue strength of polyester composite are higher than those of the PMMA composite by 15.6%, 24.3% and 20.54% respectively. Also, results of finite elements method showed that the polyester socket has better fatigue life than PMMA composite socket by 4.65%.

Lenka P. K. and A. R. Choudhury (2011) [30], analyzed the optimum design solution in a parametric study for an above knee socket material by finite element method. A pragmatic three-dimensional finite element model of the above knee socket was improved to discover the stress distribution pattern under physiologically important loading condition during normal walking. The CAD model of the rectified above knee socket was gathered from a CMET 250 non-tactile high exactness (0.06 mm) while light scanner and analyses were carried out utilizing finite element method in ANSYS workbench.

Irawan, A. et al. (2011) [31], investigated the tensile strength of three sockets prosthesis. These sockets were made of fiberglass with polyester (FGP), ramie fiber with polyester (RP) and ramie fiber with epoxy (RE) with thickness of (4 mm). They found that socket made of RE gave better tensile strength and modulus of elasticity 86 Mpa and 9.56 Gpa respectively as compared with RP and FGP.

Irawan, A.P. and I.W. Sukania (2011) [32], focused on the effects volume fraction changing on the mechanical properties for an above knee socket prosthetic laminates manufactured from mixing rattan fibers with epoxy (RFER). These laminates were manufactured by hand layup process with volume fraction (10, 20, 30, 40 and 50 %). Tensile, impact, flexural and compressive tests were performed to acquire the mechanical properties. The optimum mechanical properties were resulted from (RFER with 40 %). It gave (tensile strength 65.25 MPa, compressive strength 35.23 MPa, flexural strength 79.25 MPa and impact strength 46.35 kJ/m^2) compared with the others volume fractions.

Andrew I Campbell, et al. (2012) [33], recommended in their research deliberating and proposing suggestions to prosthetic limb sockets from plant-based composite materials. Tensile strength test for some of the fiber and resin composite samples were obtained as well as finding the average ultimate strengths and Young's moduli for all samples. Ramie and banana fiber composites have the highest ultimate tensile strength among the natural fiber test pieces. The carbon fiber composite samples showed the highest Young's modulus (8.8 GPa) and ultimate tensile strength (127.5 MPa). Bamboo fibers did not appear to have high ultimate tensile strength of the test pieces. For comparison, the tensile strength results were included for Nyglass and 80:20 acrylic resin test pieces. However, the results did not indicate whether the standard materials are stiff or strong. The

Nyglass weave brings about huge numbers of the filaments being typical to the long hub of the test pieces. Therefore, it would be expected both the stiffness and the tensile strength to be higher than the measured values if all the Nyglass fibers were parallel to the long axis.

Irawan, A.P. and I.W. Sukania (2012) [34], used the Bamboo Fiber Reinforced Epoxy Composites (BRFEC), which is strength and in manufacturing laminated socket. Laminated composites were made of from woven bamboo fiber width of 3 ± 0.5 mm and thickness of 0.3 ± 0.05 mm with volume fraction (10%, 20%, 30%, 40%, and 50%). Tensile, compressive and impact tests were conducted on these laminations. The results explained that (BRFEC with 40% vf) gave better results when compared with the others volume fraction. The results of modulus of elasticity, impact strength, tensile strength and compressive strength were 8.96 ± 0.33 GPa, 1.3 kJ/m2, 78.09 ± 1.97 MPa and 87.1 ± 4.3 kN, respectively.

Agarwal, et al. (2014) [35], dealt with the short carbon fiber and bidirectional reinforced epoxy composites. The study reported the effect of fiber loading on mechanical, thermo-mechanical and physical properties. Five different fiber loading of (10wt %, 20wt %, 30wt %, 40wt % and 50wt %) were examined. The mechanical properties and physical properties (e.g. tensile strength, hardness, inter-laminar-shear strength, impact strength and flexural strength) were decided to explain the behavior of composite structures with different fiber loading. The results showed that the increasing in the fiber loading improved the mechanical properties of bidirectional carbon fiber compared to short carbon fiber except for hardness. On the other hand, short carbon fiber showed better results in term of the mechanical properties.

Muhammad .S, et al. (2016)[36], suggested new material to manufacture Above-Knee prosthetic socket using the lamination of monofilament, cotton and perlon fibers. In their research, fiberglass is replaced with monofilament fiber. In the study, two samples were used to conduct a comparison in the mechanical properties and fatigue life between them. The first sample was available which was consisted of (4 perlon+2 fiberglass + 4 perlon) while the second sample was suggested consisting of (4 perlon+1 cotton+1 monofilament + 4 perlon). Results showed that Young's modulus increased by 42% along with decreasing in yield stress and tensile stress by 46% and 10.8%, respectively, but the stress endurance increased by 140%.

2.4 Literature Review on Fatigue Interaction:

Kahtan Al-Khazraji et al. (2012) [37], measured the tensile and fatigue properties for five laminated composite materials which used for manufacturing lower-limb prosthetic sockets. The matrix material was reinforced by using woven carbon fiber, perlon, woven fiberglass, hybrid (woven carbon fiber and woven fiberglass) and hybrid (fiberglass and carbon fiber woven) with micro silica particles (15µm). All samples were reinforced by epoxy. The results appeared that optimum mechanical properties in carbon fiber with epoxy) and fatigue properties reach to ($48.61 J/mm^3$). Also, adding Sio_2 particles to hybrid materials increases the strength and modulus of elastic 1.5 and 1.7 times, respectively.

Adawiya A. H (2013) [38] computed fatigue and mechanical properties of above knee socket prosthetic made of (perlon-fiberglass-perlon) for various four laminations lay-up (6 perlon + 2 fiberglass + 6 perlon, 5 perlon + 2 fiberglass + 5 perlon, 4 perlon + 2 fiberglass + 4 perlon, 3 perlon + 2 fiberglass + 3 perlon) by using fatigue and tensile tests under different temperature (20 ^{o}C to 60 ^{o}C). The results explained that the lamination which was layup from (3 perlon + 2 fiberglass + 3 perlon) record better fatigue and mechanical properties reached to ($E = 0.857$ GPa, $\sigma_{ult} = 73.1$ MPa and $\sigma_e = 15$ MPa). Fatigue and mechanical properties reduced when the temperatures raise up to 60 ℃ for lamination (323) reaches to ($E = 0.794$ GPa, $\sigma_{ult} = 65.4$ MPa and $\sigma_e = 1.6$ MPa). The results showed the effect of low and high temperature upon fatigue and mechanical properties.

Muhammed A. M. (2016) [39], investigated the fatigue properties and tensile behavior of hybrid and components materials such as carbon fibers, E-glass fibers and perlon with epoxy resin, for different thicknesses according to lamination. The outcome demonstrated that with increasing the number of perlon layers to 11, the tensile strength decreases by 22% while the modulus of elasticity (E) is clearly enhanced by 44%. Carbon fiber is stronger than fiberglass matter which is enhanced with fixing layers of perlon for all layers of laminates by 15% for (5 perlon 4 carbon fiber 5 perlon). Reinforcement type has a clear effect upon their fatigue resistance. Carbon fiber reinforcement has the highest fatigue limit 58% as compared with 11 perlon layers only and with (5 perlon 4 fiber glass 5 perlon) 29% appropriate to the maximum Young's modulus (E). The endurance limit stresses are reduced by 27% with the rising number of perlon laminations.

2.5 Concluding Remarks

Few studies have been carried out in the context of through-knee prostheses. All aiming to mimic the normal amputated extremity and help the amputee to walk as normal as possible. Most researchers have concentrated their investigations on the modeling of the human lower limb and design and manufacturing prosthetic socket.

In general, prosthetics is still considered to be a modern trend of a science. Above-knee and below-knee prosthetics are well documented in the literature whether concerning the design of sockets or the manufacturing of them. However, the researchers about through-knee prosthetics are still sparse and there is a general lack in this context.

Therefore this work focuses on deciding the best type of composite materials to be able to withstand the fatigue loading for TK prosthesis as an alternative to the available socket used in the rehabilitation centers in Iraq that meets the special-needs people's requirements. This was achieved experimentally and numerically.

Chapter Three

Analytical and Numerical Work

3.1 Foreword

This chapter represents the analytical and numerical part of this thesis. The stress distribution, directional deformations and safety factor for suggested and available cases in the through knee prosthetic socket were estimated by using a numerical simulation. Analyses were performed in order to calculate the pressure distribution between stump and socket surface.

3.2 Analytical Analysis

One of the main roles of a lower extremity prosthetic socket is to support the amputee's body weight. The weight and the forces produced during walking are conveyed from the bone to the prosthetic socket by limb tissue and skin. A portion of the vertical force is supported by the vertical component of normal force caused by compression of the soft tissues. The effect of a conical angle of the socket shell is very small, so the main component of the normal force operating in the horizontal direction and pushing the skin in the radial direction. Friction has three advantages, such as creating a tangential force to the skin, holding the prosthesis in position and preventing it from being detached during the swing phase of gait and supporting the weight of the body during the support phase[40].

In order to comprehend the frictional action between a stump and a socket, a simplified model has been developed. Two assumptions were taken:

1. The socket would cause the soft tissue of the limb to be compressed into a conical shape.

2. The internal bone was considered of cylindrical shape as shown in Fig. (3.1).

The skeletal forces are conveyed to the prosthesis by compressing and stretching the skin and the soft tissue between socket and bone. In this model, these tissues are considered as the elastic materials and act as a spring support ($\boldsymbol{F = Kd}$). One normal spring is considered to mimic the support force caused by compression of the soft tissues normal to the surface of the skin. Also, a parallel spring represents the force resulting from stretching or shearing in the same direction of the interface. The material is considered to be homogeneous, linear and isotropic [40].

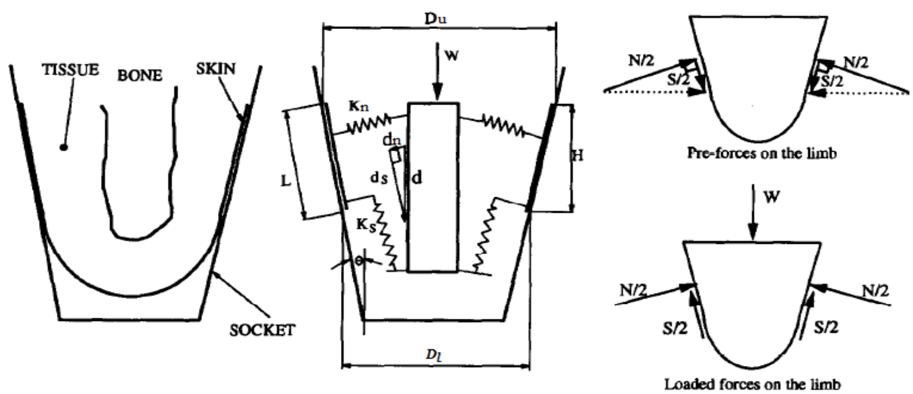


Figure (3.1): Simplified residual limb model [40]

The average pressure and shear stress acting on the extremity surface can be calculated.

$$\sigma = \frac{K_n}{A}\left(d_{no} + \frac{W \sin \Theta}{K_n \ \sin^2\Theta + K_s \ \cos^2\Theta}\right) \quad \textbf{(3.22)}$$

And,

$$\tau = \frac{S}{A} = \frac{K_s}{A}\left(-\frac{K_n}{K_s} d_{No} \tan \Theta \ + \ \frac{W \cos \Theta}{K_n \ \sin^2\Theta + K_s \ \cos^2\Theta}\right) \quad \textbf{(3.24)}$$

It has been shown that the damage of tissues relies on the resultant of shear and normal stresses σ_c. This resultant can be obtained from the **Eqs. (3.22)** and **(3.24)**.

$$\sigma_c = \sqrt{\sigma^2 + \tau^2} \qquad \textbf{(3.25)}$$

3.2.1 MATLAB Coding

A program was constructed and executed using (MATLAB version R2017a), to find the maximum interface pressure between the stump and a prosthetic socket. The output data obtained from MATLAB were implemented by using (ANSYS workbench 14.5) to analyze pressure effect on the socket and to find the shearing stress acting on the limb surface. **Appendix (A)** shows the MATLAB coding used to calculate the analytical stresses.

The following flowchart summarizes the input and output data implemented by MATLAB as shown in Fig. (3.2):

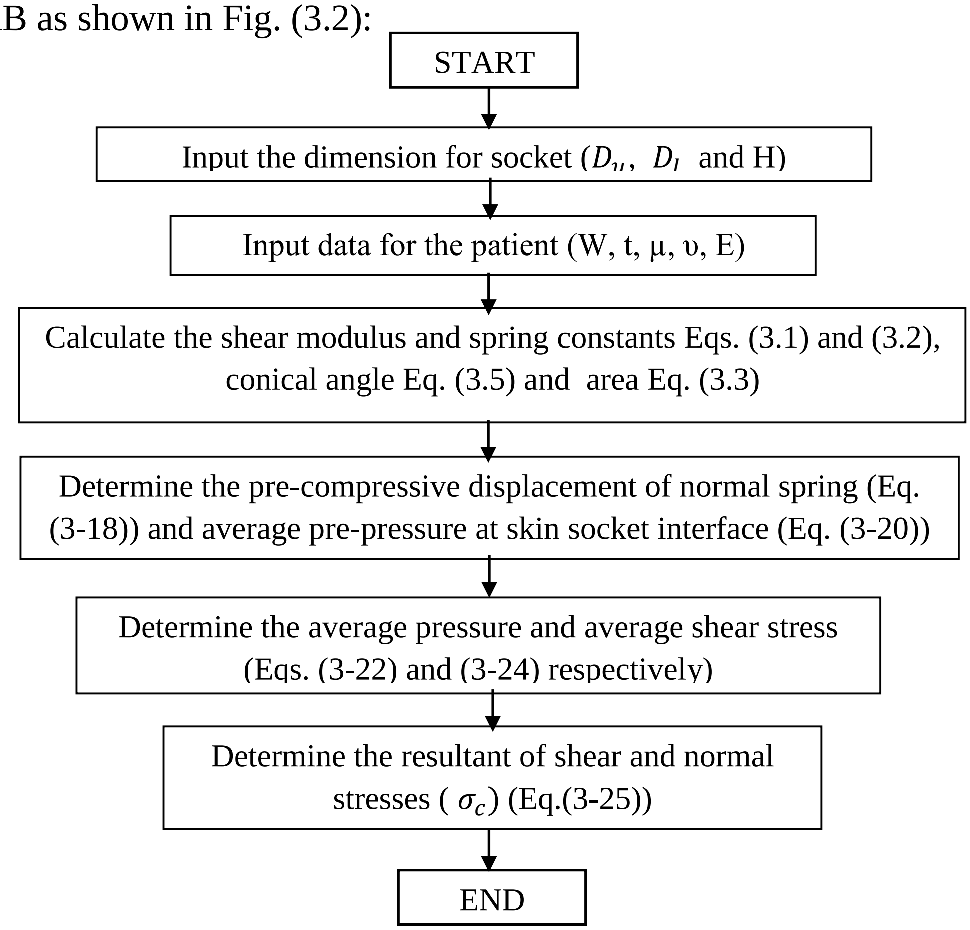


Figure (3.2): Flowchart for MATLAB program

3.2.2 Case Study

The analysis was conducted using some typical values for the geometric and mechanical properties associated with the limb/prosthetic socket interface: , young's modulus of soft tissue E = 100 kPa [41], upper diameter D_u = 145 mm, lower diameter D_l = 135 mm, height H = 40 mm, average tissue thickness t = 20 mm, conical angle $\Theta = 7.12^0$, coefficient of friction μ = 0.4, Poisson's ratio υ = 0.5 and force W= 784.8 N.

In this case, the vertical stiffness is 88.649 N/mm if the weight of 800 N acting on the bone. The minimum pre-compressive displacement of the normal spring $\boldsymbol{d_{no}}$ is 14.065 mm. The average interface normal and shear stresses are 86.6149 kPa and 34.646 kPa, respectively with supporting a vertical load of 800 N. Finally, the resultant of shear and normal stresses is 93.28 kPa.

3.3 Numerical Analysis

The finite element method (FEM) is a powerful tool which can be used in many engineering applications by taking advantage of the continuously developing digital computers with fast computation capability.

This method provides numerical solutions for complicated problems including complex geometrical boundaries and non-linear material properties. In this work, ANSYS workbench 14.5 software was used as a numerical tool to find the effect of the fatigue, the stress distribution contours, directional deformations, and the safety factor.

The analysis done using ANSYS workbench has three unique steps they are:

1. Constructing the geometry and meshing.
2. Applying the boundary conditions and loading.
3. Executing the program.

3.3.1 Modeling

The through-knee amputee is of a male 27 years old and 80 kg. Finite element model of the socket was built according to the real geometrical dimensions of the amputee's stump. The real dimensions of the socket were measured. These dimensions were used later on to draw the model using SolidWorks software (version 2015). The resulted model was then exported to ANSYS workbench 14.5.

3.3.2 Element Type

The ANSYS element library contains various element types. Each element type has a distinct number that identifies the element category. In this work, SHELL181 as shown in Fig. (3.3) is used. SHELL181 is suitable for analyzing thin to moderately-thin shell structures. It is a four-node element with six degrees of freedom at each node: translations in the x, y, and z directions with rotations about the x, y, and z-axes. If the membrane option is used, the element has translational degrees of freedom only. The degenerate triangular option should only be used as filler elements in mesh generation. Fig. (3.3) shows the geometry, node locations, and the element coordinate system for this element. The element is defined by shell section information and by four nodes (I, J, K, and L).

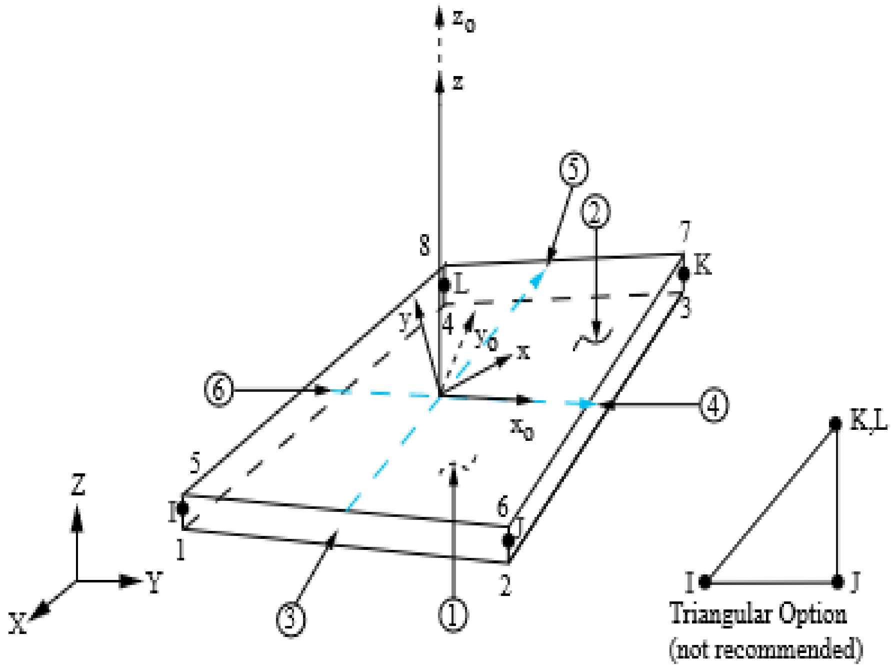


Figure (3.3): Shell181 element geometry.

3.3.3 Meshing the Model

The meshing process has been done by choosing the volume, and then the shape of the element is selected, automatic meshing, as shown in Fig. (3.4).

The improper model formulation has a potential to produce results that are meaningless to the physical system which is intended to be modeled. Number of factors in the problem formulation stages influence the output applicability of the results. These factors are the type, size, and number of the elements.

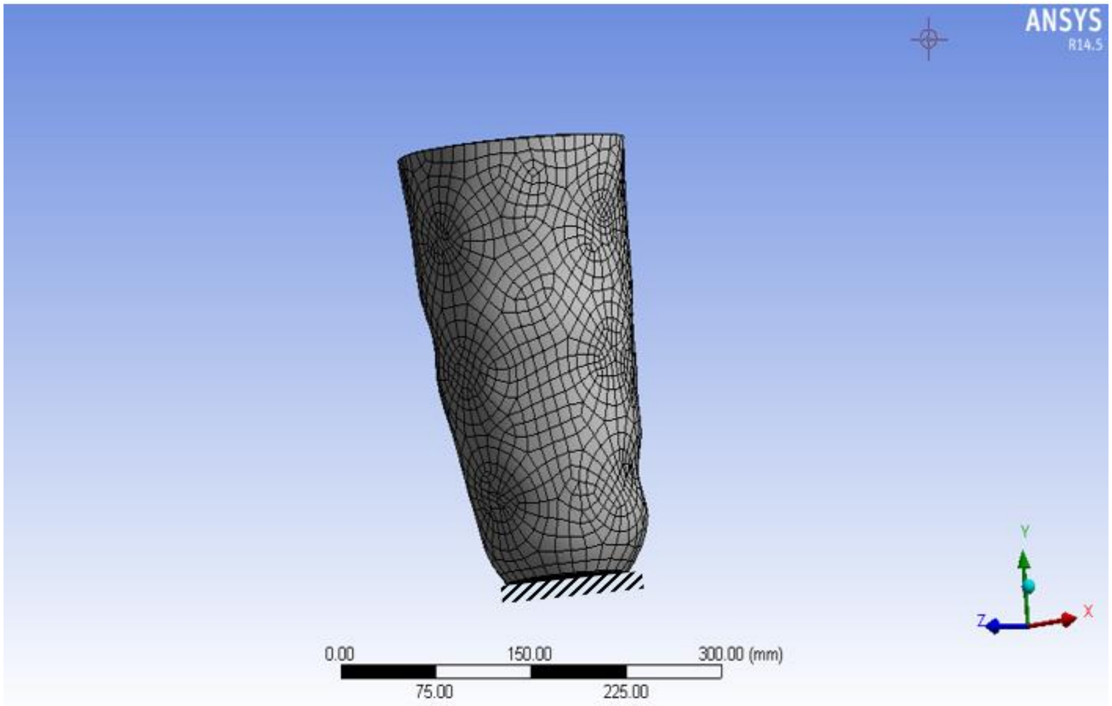


Figure (3.4): The model of through knee socket with meshing.

3.3.5 Analysis Type and Applying Load

The term (load) refers the boundary conditions including constraints, supports and any other external and internal loads. The load which is fed to ANSYS [42] is a fixed support at the adapter of the socket. Simultaneously, the interface pressure is the value produced between the socket and the patient's residual limb during the locomotion. Fig. (3.5) depicts the position of pressure distributions of the present experimental case study. The values of this pressure are obtained from two cases: first case is obtained from the experimental part (F-socket sensor), and the second one is determined from the analytical part. The pressure for the analytical part is represented in Fig. (3.6).

The following steps were followed in ANSYS simulation:

1. Firstly, creating the geometry of socket model,
2. Mesh generation by meshing options,

3. Entering the values for important parameters, e.g. Young's modulus, Poisson ratio, ultimate stress, density and alternating stress.

4. The 10^6 cycle's life is used as the life for the fatigue failure [43].

5. After taking all the previously mentioned steps, Von-Mises theory was considered as the failure criterion.

6. The output of ANSYS is mainly the equivalent stress, maximum shear stress, directional deformations, safety factor, and life at particular loads.

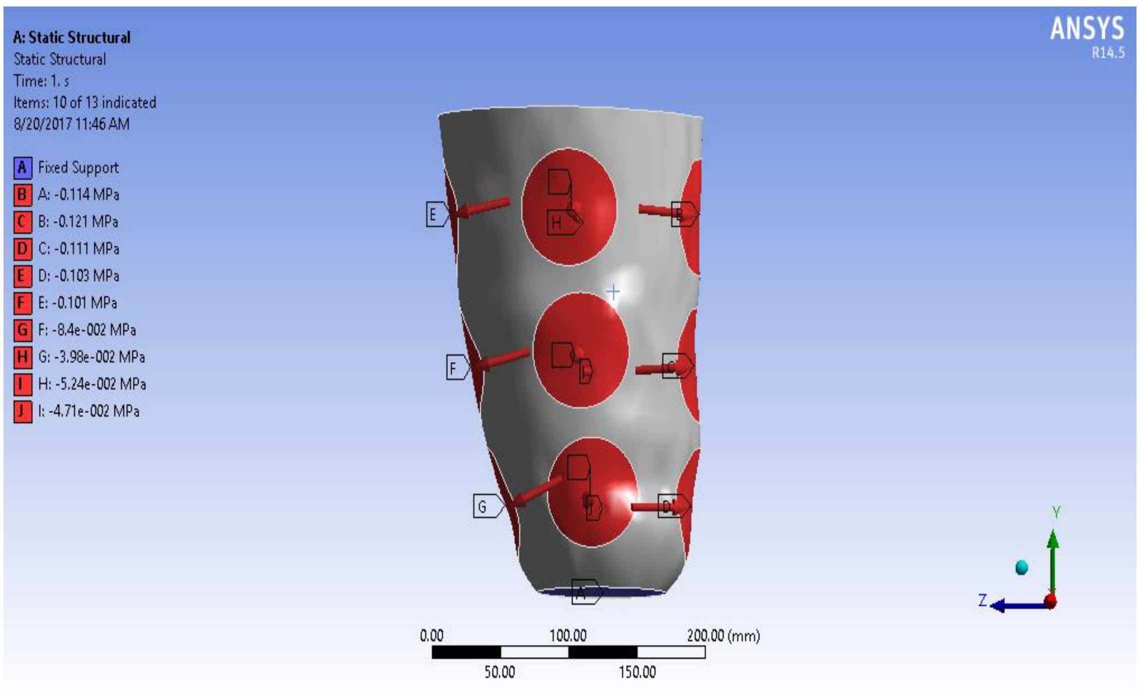


Figure (3.5): Locations of pressure forces on the model for the experimental part (F-Socket)

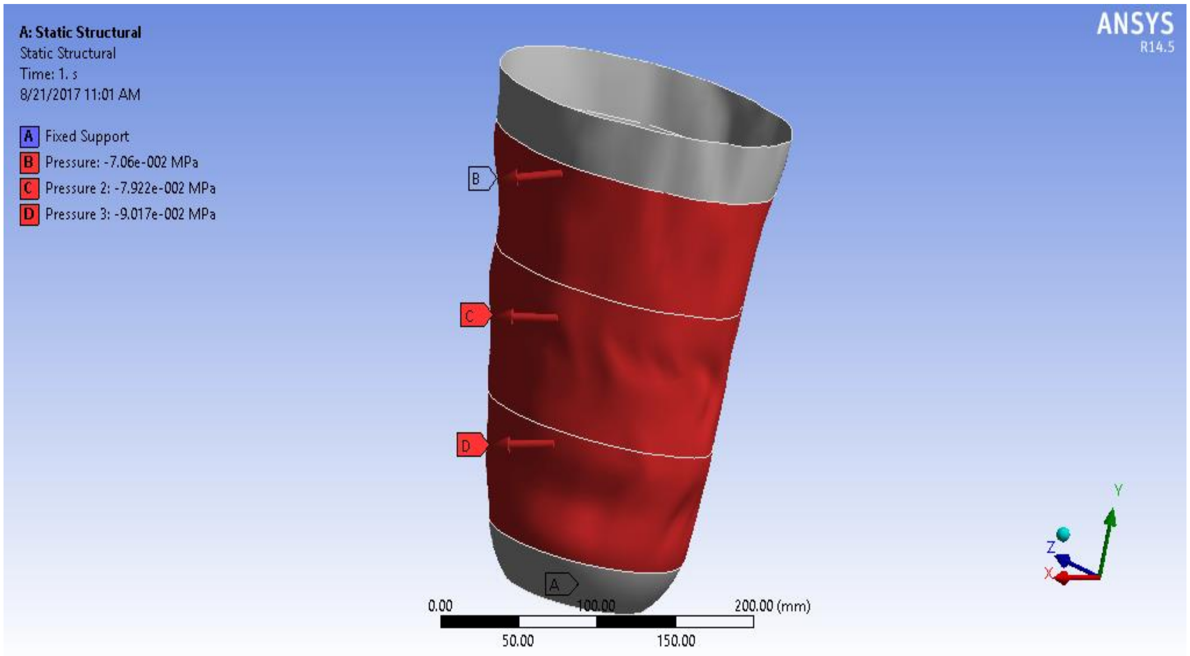


Figure (3.6): Applying the locations of pressure forces on the model for the analytical part

3.3.6 Stiffness to Weight Ratio (Specific Modulus)

The stiffness (***k***) of a body can be defined as a measure of the resistance undergone by an elastic body to deformation. Specific modulus is a materials property consisting of the elastic modulus per mass density of a material. It is also known as the stiffness to weight ratio or specific stiffness. Applying unit loads on sockets in three axis, for axial stiffness (F_y), bending stiffness (F_x) and torsion stiffness (M_y) ($F_y = 1\ N,\ F_x = 1\ N\ and\ M_y = 1\ N.m$) as figures (3.7), (3.8) and (3.9) respectively by using ANSYS for the available and the suggested prosthetic sockets

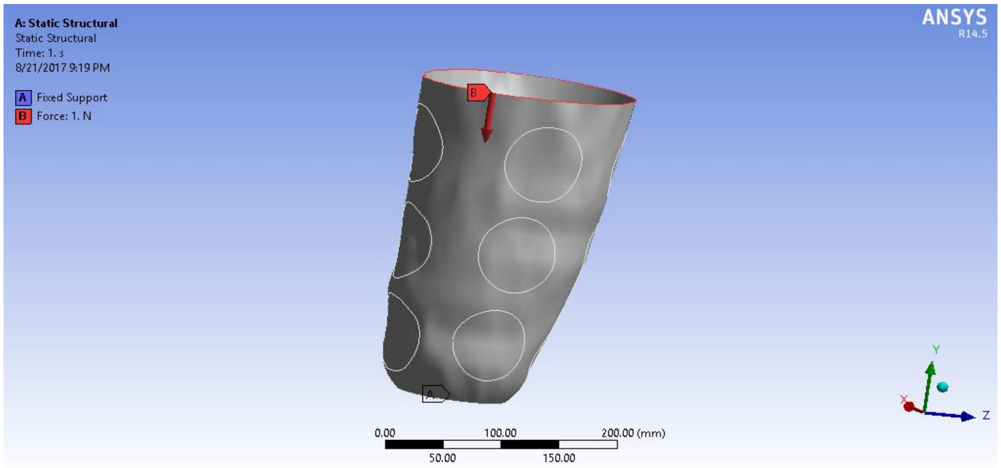


Figure (3.7): Applying the axial unit load force on the model

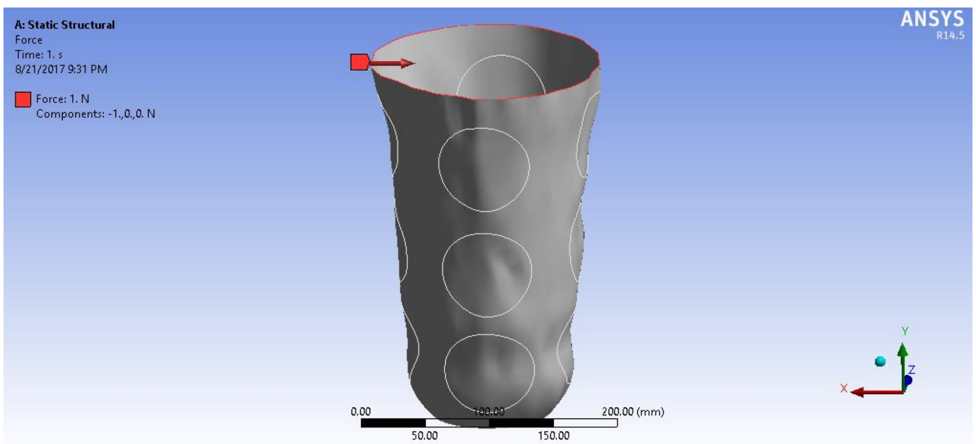


Figure (3.8): Applying the bending unit load force on the model

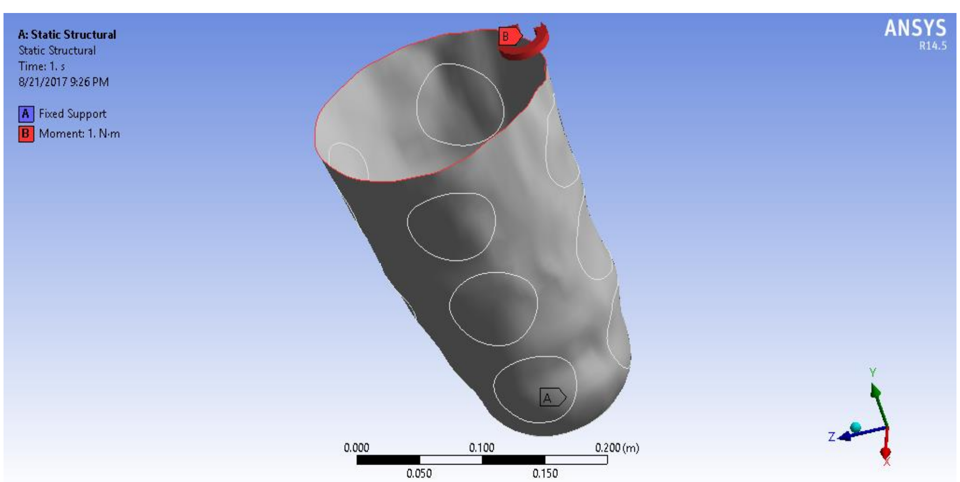


Figure (3.9): Applying the torsion unit load force on the model

3.3.7 Convergence Study

A convergence study was carried out to determine how used mesh is suitable for current model for accurate results and computing time during analysis. The number of degree of freedom used for this model ranged from 10560 to 176712. The results of the convergence study are presented in Fig. (3.10). Based on these results, the model which consisted of elements 8160 provided satisfactory accuracy.

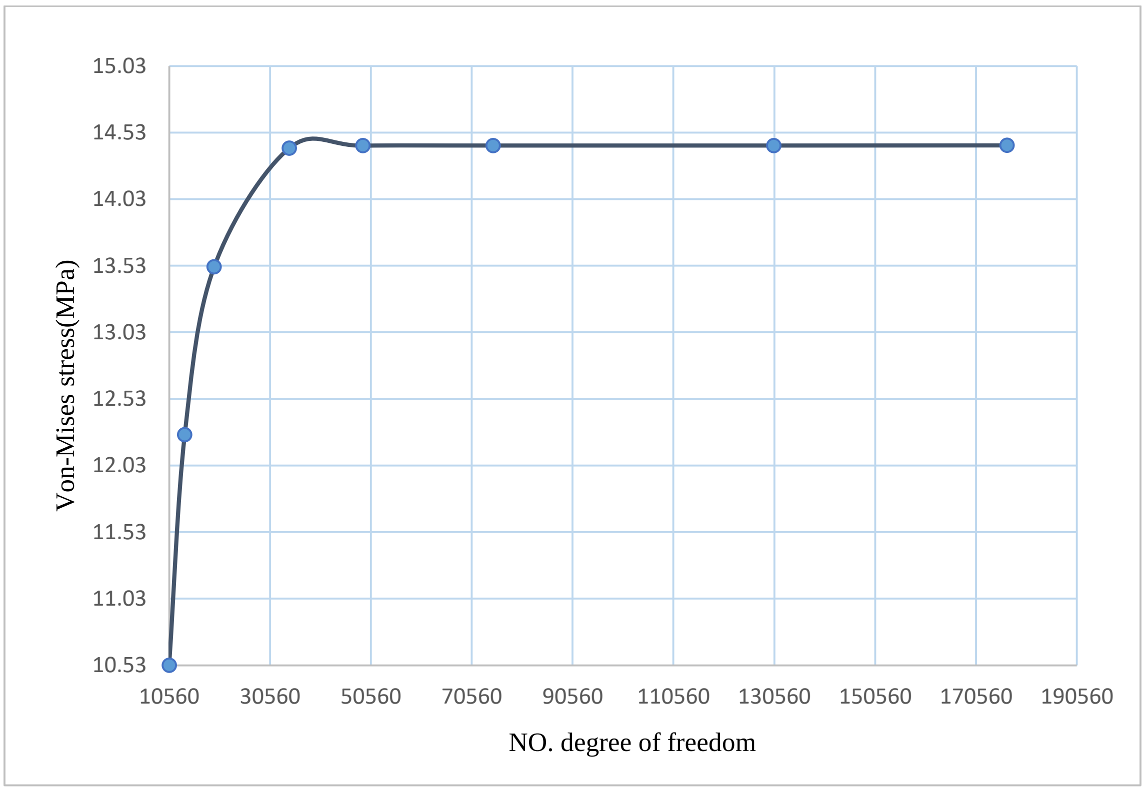


Figure (3.10): Convergence test

Chapter Four

EXPERIMENTAL WORK

4.1 Introduction

The goal of this chapter is to show the preparation steps of socket manufacturing which could have good and useful mechanical properties. Many experiments are presented for designing and manufacturing the through knee (TK) socket prosthesis. The case study is a male patient having a through-knee amputation in his right leg. To achieve the experimental work, different materials were used for socket lamination. The following steps are the main points of interest of this chapter:

a- Mechanical characterization :

i. Tensile test
ii. Fatigue test
iii. Flexural test

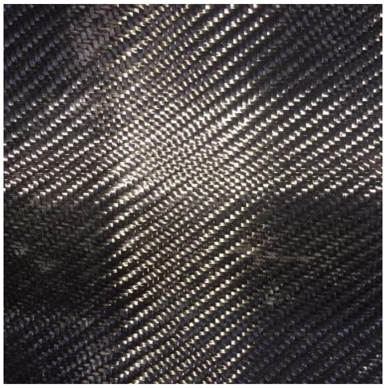
b- Biomechanical characterization :

i. Gait cycle and the ground reaction force testing(GRF) by (force plate)
ii. Measuring the interface pressure between the leg and the socket (F-socket).

4.2. Samples Preparation For Mechanical Characterization:

4.2.1 Raw Materials

Different raw materials are used to manufacture four materials to be used in further steps to manufacture the prosthetic socket. The materials used for the first and second samples (S1 and S2) are knit perlon, woven carbon fiber, orthocryl lamination resin 80:20 and hardener. The same materials were used for the third and fourth samples (S3 and S4) in addition to palm nut powder. Figure (4.1) shows the used materials to manufacture a prosthetic socket for the present study.



Woven carbon fiber

Knit perlon

Orthocryl resin 80:20

Hardener

Palm nut powder

Figure (4.1): Materials used in producing the samples

4.2.2 Calculation of The Physical Properties of the Samples

The mass, density and volume fraction are important parameters to describe the physical properties of samples and improve them in the future. It is possible to know the thickness, mass and volume fraction using Vernier caliper, digital scale and beaker, respectively. Firstly, the mass was measured by the accurate balance and the volume was measured using the beaker in order to calculate the density according to Eq. (4-1). Then the volume fraction was calculated from Eq. (4-2).

$$\rho = \frac{m}{V} \quad \textbf{(4-1)}$$

$$v_f = \frac{V_{fibers}}{V_{matrix}+V_{fibers}} \times 100 \quad \textbf{(4-2)}$$

The physical properties of raw materials and volume fraction for all four samples are summarized in Table (4.1)

Table (4.1): The physical properties of raw materials.

Materials	Density g/cm^3	State	Sample 1 S1 (%)	Sample (2) S2 (%)	Sample (3) S3 (%)	Sample(4) S4 (%)
Fiber carbon	1.2	woven	11.5	21	11.5	11.5
Perlon	1.083	Knit	26.5	25	26.5	26.5
Orthocryl lamination 80:20	1.04	Liquid (resin)	60	52	56	52
Hardener	1.08	Powder	2	2	2	2
Palm nut	1.2	Powder	0	0	4	8

4.2.3 Equipments

- **Gypsum mold**: Gypsum is the material of the mold. The mold has a shape of a cuboid of the dimensions 15×5×30 cm^3. The outer surface of the mold as seen in figure (4.2a) was refined to be smooth and flat for good presenting the inner surface of the samples
- **Polyvinyalcohol (PVA):** Two plies of PVA are used. The first one is used on the outside surface of the mold to prevent adhesion of resin with

the mold, and the second one is applied on the outside surface of the materials to achieve the casting process as shown in Figure (4-2b).

- **Vacuum System**: It includes pipes and vacuum pump (-0.8 bar). It is used for suction the air completely from two possible gaps between the PVA and the gypsum mold and between the inner and outer plies of PVA. There are several benefits of vacuum system shown in Figure (4-2c). evacuation through the first gap gives a perfect shape for gypsum mold and prevents creating bubbles between the mold and PVA. For the second gap vacuum is used to remove the air between the PVA and the part of casting process. Moreover, vacuum process insures no bubbles created during the casting process.
- **Mechanical workshop**, held in the University of Karbala / Prosthetics and Orthotics Engineering Department**,** includes different tools for forming and cutting.

(a) Gyps Mold (b) **PVA** **(c) Vacuum device**

Figure (4.2): The equipment to make the samples

4.2.4 Procedure of Sample's Manufacturing

The sample manufacturing steps are detailed as follow:

1- The gypsum mold was placed on the stand and was connected to the vacuum device with a tube. The inner (PVA) was put on the mold and closed from the top and bottom. Air between them was removed to prevent formation of bubbles and to prevent adhesion between resin and positive mold as shown in the Figure (4.3a).

2- The perlon and carbon fibers were added according to the lamination lay-up as depicted in Table (4.2) as shown in the Figure 4.3 (b, c and e). The outer PVA was put and the top end was left open for resin supply. A hole was made near the lower end to remove air bubbles by using a vacuum device Figure (4.3d).

Table (4.2): The laminations manufactured with different lay-up method.

Number of samples Lamination	Lay–up Samples	Total Number of Layers	Palm nut (%)	Lamination Lay–up Procedures
S1	4-2-4	10	0	4perlon+2carbon fiber+4perlon
S2	3-1-1-2-1-1-3	12	0	3perlon+1carbon fiber+1perlon+2carbon fiber+1perlon+1carbon fiber +3perlon
S3	4-2-4	10	4	4perlon+2carbon fiber+4perlon
S4	4-2-4	10	8	4perlon+2carbon fiber+4perlon
Available[44]	4-2-4	10	0	4 perlon+2fiberglass+4 perlon

a

b

c

d

e

Figure (4.3): Procedure of making the samples

3- Resin was poured into the mold:

i. For the first and second samples (S1and S2), the mixing resin (Orthocryl lamination resin 80:20) with (2-3) % of hardener powder [45] was fed and then poured from the top end along with using vacuum device as shown in Figure 4.4 (a and b).

ii. Same procedure is followed for the third and fourth samples (S3 and S4), but the resin was mixed with hardener powder and different percentages from palm nut powder (4% and 8%) as shown in Figure (4.5c).

a **b** **c**

Figure (4.4): Pouring resin into mold

4- Keeping the vacuum device operating until the composite material warms up and becomes hard. This process takes about 45 minutes. After that, the final product is removed from the mold using a Vibrational Cutter as shown in the Figure (4.5), and it becomes ready to harvest specimens for standard mechanical tests.

(a) **(b)**

Figure (4.5): Procedure for making material prosthetic socket

4.3 Mechanical Tests:

4.3.1 Tensile Test

Tensile testing is very important to know the mechanical properties of the composite materials. Tensile test was done by using the material testing machine as shown in Figure (4.6). The aim of the tests is to estimate the values the modules of elasticity, yield stress and ultimate stress for each material used for both the available and suggested prosthetic socket. Standard tensile test specimens were cut from the final product according to the American Society for Testing and Materials (ASTM) D638 [46]. Three specimens were cut for each lamination at the workshop using a computer numerical control (CNC) machine. The specimens were machined with 50 mm original length and 13 mm width while thickness was varied with kind of the lay-up lamination as shown in Figure (4.7). next, specimens were used in the tensile test which was carried out at the speed of 5 mm/sec.

Figure (4.6): Mechanical testing machine (Tinus Olsen)

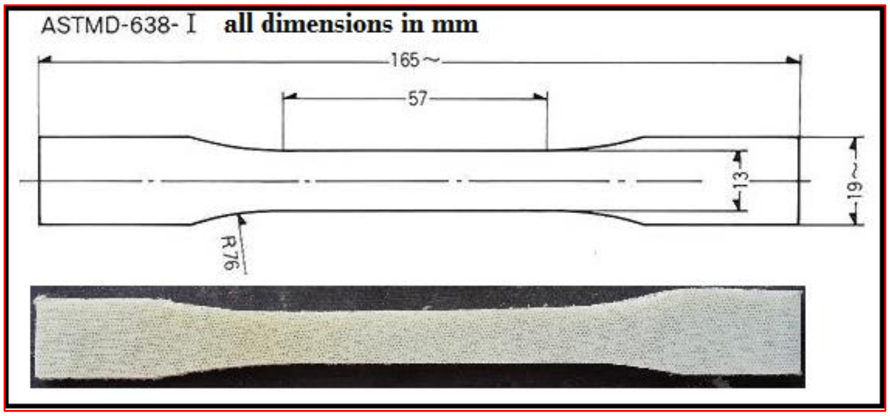
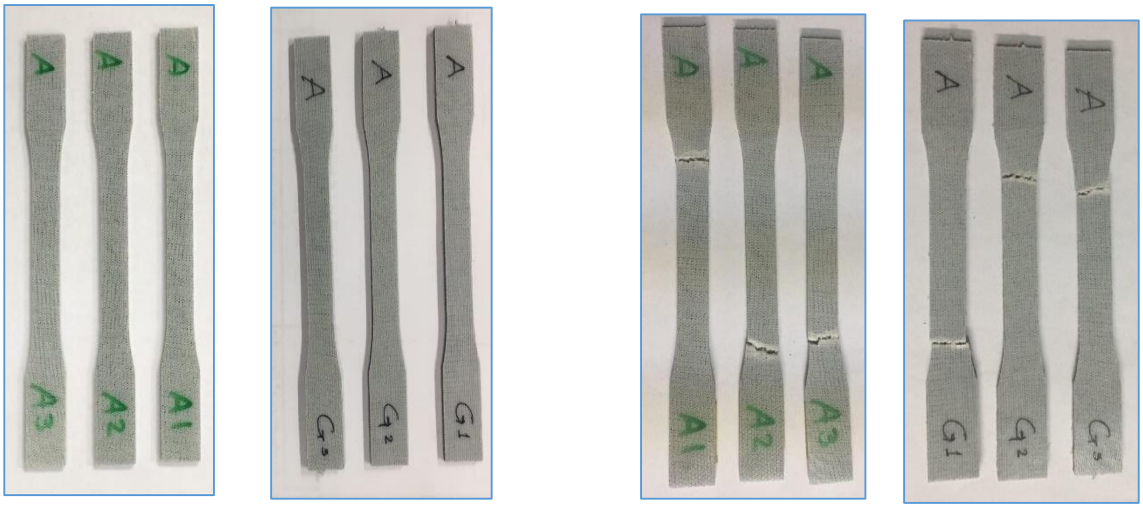


Figure (4.7): Specimen for tensile test (all dimensions in mm)

The tests were achieved for three specimens for each sample in the Material Researches Department (MRD) in the Ministry of Sciences and Technology of Iraq. Figure (4.8a) shows the first and second specimens before and after testing, and Figure (4.8b) shows the third and fourth specimens before and after tensile test.



S1 S2 S1 S2

Before testing After testing

(a)

S3 S4 S3 S4

Before testing After testing

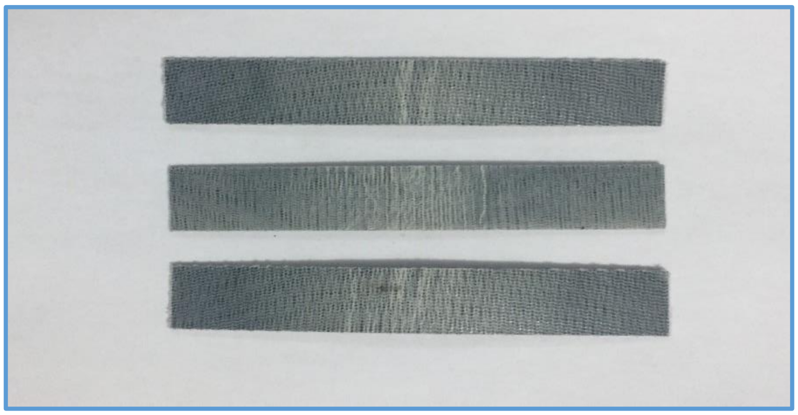
(b)

Figure (4.8): (a) Tensile test for first and second materials before and after test

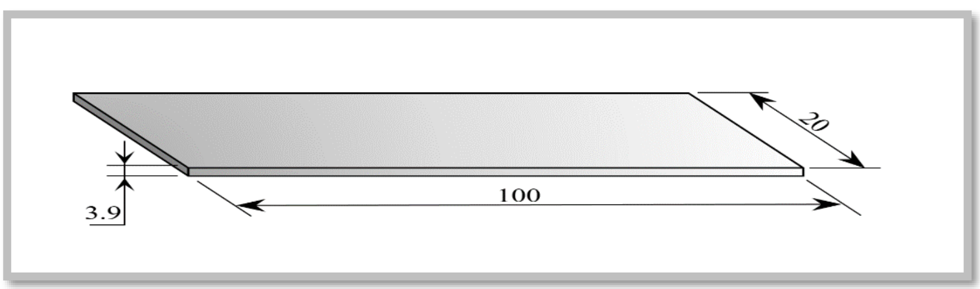
(b) Tensile test for third and fourth materials before and after test

4.3.2 Flexural Test (Three-point bending)

Three specimens from the second suggested material (S2) were tested by flexural test to get the average bending modulus of elasticity (E_b) which is to be used later to find the alternating stress in fatigue test. The test was carried out using testometric machine, and was conducted in ministry of science and technology/material researches center. The size of the samples was selected according to the machine's manual as shown in Figure (4.9b). Figure(4.9a**)** shows the bending specimen after the test.



(a)



(b)

Figure (4.9) (a) Bending specimen after the test.

(b) The dimensions for bending specimen(all dimensions in mm)

4.3.3 Fatigue Test

Fatigue properties of the examined material can be represented by stress-No. of cycle S-N curve. A socket fatigue should be taken into consideration due to dynamic load and fluctuating stress existing in walking.

Fatigue machine (HSM20) as shown in Figure (4.10) was used in this test. The size of the samples was selected according to the machine's manual as shown the Figure (4.11). The test was carried out on eight specimens for lamination (S2). The specimens before and after test are shown in Figure (4.12). Values of the alternating stress (σ_a) are determined form Eq. (4-3) [47].

$$\sigma_a = \frac{1.5\ E_b\ *ta*\delta}{\ell^2} \tag{4-3}$$

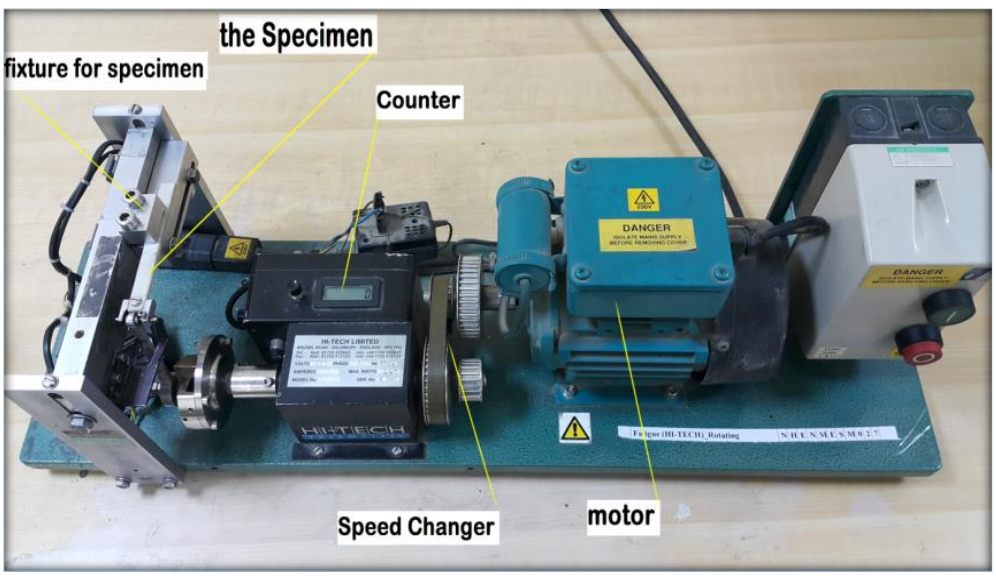


Figure (4.10): Fatigue Machine (HSM20)

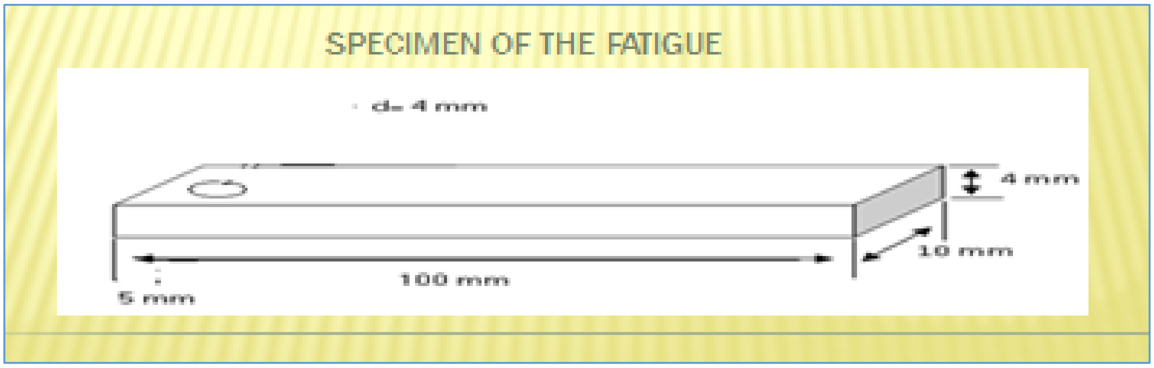
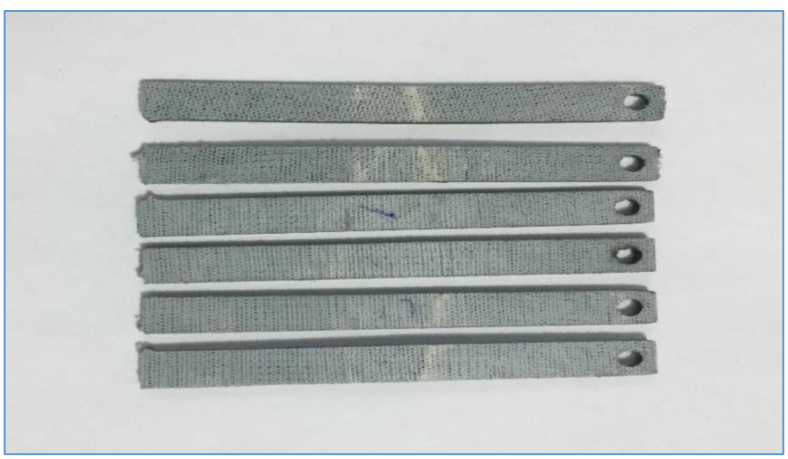


Figure (4.11) Fatigue specimen stander dimentions

(a)



(b)

Figure (4.12) (a) Specimens before test **(b)** Specimens after test.

4.4 Fabrication of Through-Knee socket :

4.4.1 Materials Used

In this work the materials needed in the lamination of through knee socket were chosen to be a sequence laminated. This means that the material is isotropic [48]. The used materials are:

- Perlon
- Woven Fiber carbon
- Lamination resin 80:20
- Hardening powder
- PVA bag

The equipment and tools used for the prosthetic through-knee socket lamination are gypsum mold, vacuum system and mechanical workshop.

4.4.2- Gypsum Mold Preparation:

The gypsum mold (positive mold) was made as follows:

Step 1: Measurement for the stump were taken as shown in Figure (4.13a).

Step 2: A piece of plastic is shaped in the form of a strip between the negative mold and the patient's remaining leg to help remove the negative mold easily as seen in Figure (4.14b)

Step 3: A negative mold from gypsum layer (template) was fabricated for the stump as shown in Figure (4.14c).

Step 4: After the gypsum layer (template) drying and becoming hard, it was cut through the plastic strip, and the negative mold was carefully removed from patient safety as shown in Figure (4.14d).

(a) **(b)** **(c)** **(d)**

Figure (4.13): Steps to fabricate the negative mold

Step 5: The negative mold was now ready to be filled by the gypsum to form the positive mold.

Step 6: The gypsum layer (template) was then removed to have the positive mold as shown in Figure (4.14).

Figure (4.14): Removing the gypsum layer.

Step 7: The positive mold was carved and grinded according to the measurements taken from the patient's stump as shown the Figure (4.15).

Figure (4.15): Carving and grinding the positive mold.

4.4.3 Procedure of Socket Manufacturing

Firstly, soft socket, shown in Figure (4.16), was manufactured to pulls the air from the socket. This soft socket helps to increase the comfort and eliminate pressure against the soft tissue of the residual limb. Put the layer soft socket on the positive mold and shape as the positive mold by using adhesion.

Figure (4.16) Mounting the soft socket on the positive mold.

The soft socket was placed upon the positive mold before starting the casting process. All laminations were done under vacuum device with the following steps:

Step 1: The positive mold was placed on the stand and the pressure tube of vacuum system was connected to the stand. The inner PVA bag was placed on the positive mold and closed from both sides to prevent adhesion between mold and resin as shown Figure (4.17a). The vacuum device was operated at 30 mm Hg at room temperature.

Step 2: The layers of perlon and carbon fibers were arranged according to the laminating lay-up second sample given in Table (4.2) (3Perlon and 1carbon fiber, 1perlon and 2 carbon fiber, 1perlon and 1carbon fiber and 3Perlon) layers and between these layers putted the adapter as shown in Figure 4.17(b and c).

Step 3: The outer PVA bag was placed tightly and closed from the bottom end leaving smaller hole open to remove the air from PVA bag. The other end was left open for lamination resin supply as shown in Figure (4.17d).

(a) **(b)** **(c)** **(d)**

Figure (4.17): Procedure of laminating lay-up.

Step 4: 1.25 liter of lamination resin 80:20 was mixed with 80 gram of hardener according to the standard ratio. Every 100 portions of lamination resin (acrylic resin) are mixed with (2-3) portion of hardener [49]. The matrix resulting from the mixing process was placed inside the upper PVA bag in such a manner that the mixture was consistently distributd over all lamination lay-up under vacuum until the composite material became hardened. Next, the produced socket is removed from the positive mold as shown in Figure (4.18).

Figure (4.18): Positive mold after lamination.

Step 5: After that, the socket was removed from the positive mold and connected to the knee joint as shown Figure (4.19).

Figure (4.19): Connected socket with knee joint

4.5 Biomechanical Tests:

4.5.1 Gait Cycle and Ground Reaction Force Testing (GRF)

The main force acting on the human body during walking is ground reaction force with vertical direction components. These forces are produced during the walking of the patient.

The ground reaction force (GRF) test was conducted for the patient by using force plate shown in Figure (4.20).

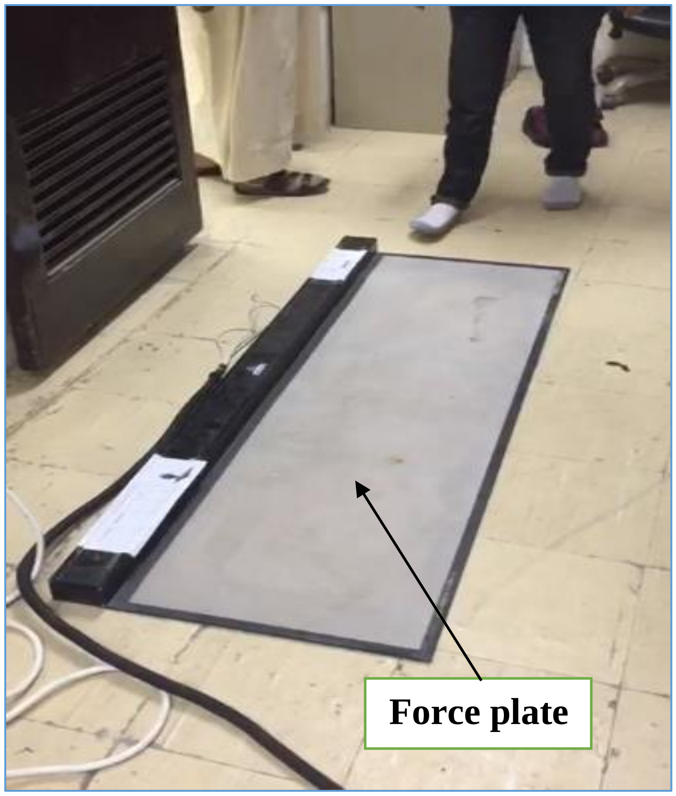


Figure (4.20) Ground reaction force test.

GRF test was done at University of Nahrain / Prosthetics and Orthotics Engineering Department. The first case study was a male having a through knee amputation in his right leg with an available prosthesis as shown in the Figure (4.21a) while the second case study the suggested prosthesis was used as shown in Figure (4.21b). The patient was asked to walk over the device for two minutes in order to assure reaching the state stability. Thus, ground reaction forces of the patient were measured.

(a) **(b)**

Figure (4.21): (a) Patient moving on the device for an available prosthetic.

(b) Patient moving on device for suggested prosthetic

4.5.2 Measuring Interface Pressure

The pressure between the socket and the stump was measured by the F-Socket sensor. This device has sensors to record the magnitude of the pressure resulting from the response of sensor through the stance phase as shown in Figure (4.22). The pressure was measured by placing the sensors onto four regions of the residual limb (Anterior, Lateral, Posterior, and Medial). the output results are displayed on the computer as pressure distribution using a computer program (F-Scan). The resulted data of this test are used later as an input data for ANSYS program.

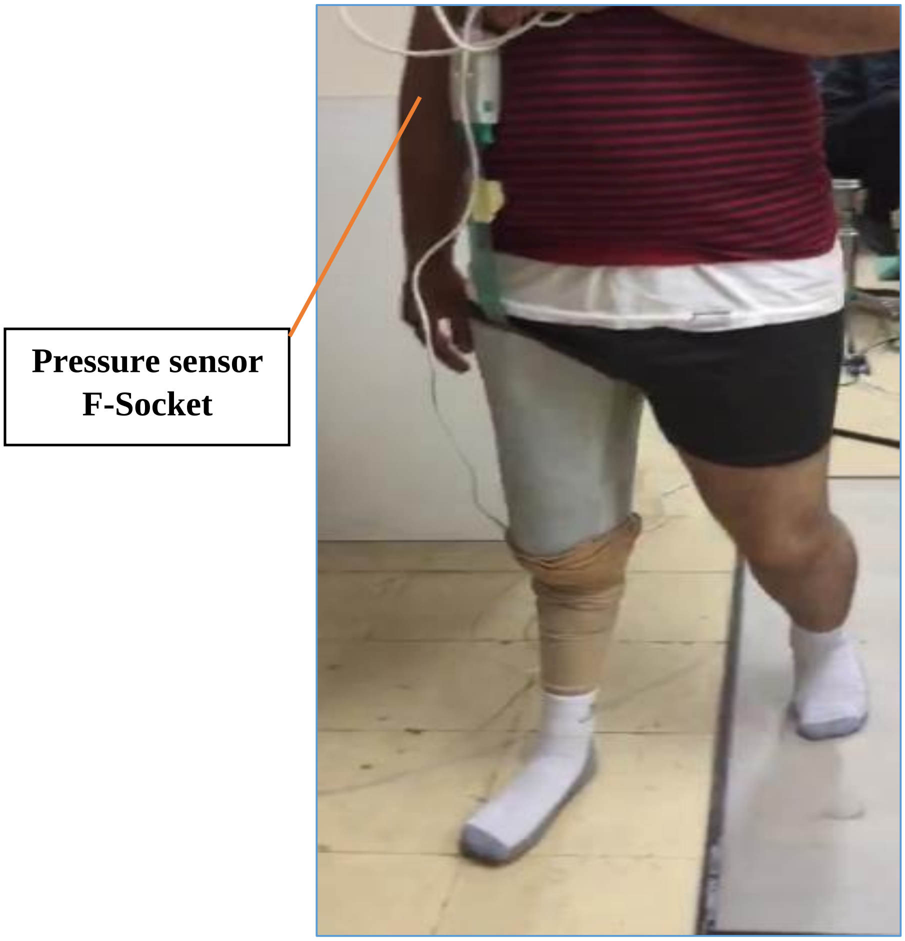


Figure (4.22): Position of sensor of pressure on patient with TK prosthesis

Chapter Five

RESULTS AND DISCUSSIONS

5.1 Introduction

The experimental results included both physical and mechanical properties for all laminations composite materials of prosthetic through knee socket made from perlon and carbon with acrylic resin 80:20 are evaluated and compared. Moreover, the experimental part includes the results stress-cycle (S-N) curves, the results of the interface pressure between the residual limb and inner socket surface for case study and the results of the gait cycle grand reaction force (GRF). The available model [44] and the suggested prosthesis' sockets which made from second sample materials (S2) are analyzed to show the results of stresses distribution, deformation, safety factor as well as the analysis of the (Von-Mises) stress by using simulation software (ANSYS workbench version 14.5).

Finally, depending on the tests conveyed throughout this work, the best lamination which has best mechanical properties was chosen in order to be used for manufacturing the suggested prosthesis' socket.

5.2 Physical Properties of Samples:

Results for physical properties (thickness, density and volume fraction) were measured by using a digital Vernier and a digital sensitive scale device while density and volume fraction are calculated according to Eqs. (4-1) and (4-2). The physical properties results are listed in Table (5.1).

Table (5.1): Physical properties for each lamination samples.

Samples	Thickness (mm)	Density (g / cm^3)	Volume Fraction(%)
S1	3.8	1.235	38
S2	4	1.267	46
S3	4.1	1.535	42
S4	4.2	1.587	46
Available	4.1	1.252	24.8

5.3 Tensile Test

Three specimens for each sample were tested using the mechanical testing machine (Tinus Olsen) at room temperature to obtain stress-strain curve. Due to the similarity of the carves trends of all samples, only one of twelve suggested samples (S2) selected and predict in the Figure (5.1).

The ultimate stress (σ_{ult}) and the modulus of elasticity (E) were obtained from the stress-strain curves for each sample, and the final values, which represent the average value for three specimens, were calculated. The mechanical properties for each lamination were summarized in Table (5.2). These properties were used later to conduct the fatigue test and to be imported to ANSYS 14.5 software to simulate the prosthetic socket. From the simulated model, stress analyzing, deformation finding and safety factor values were obtained.

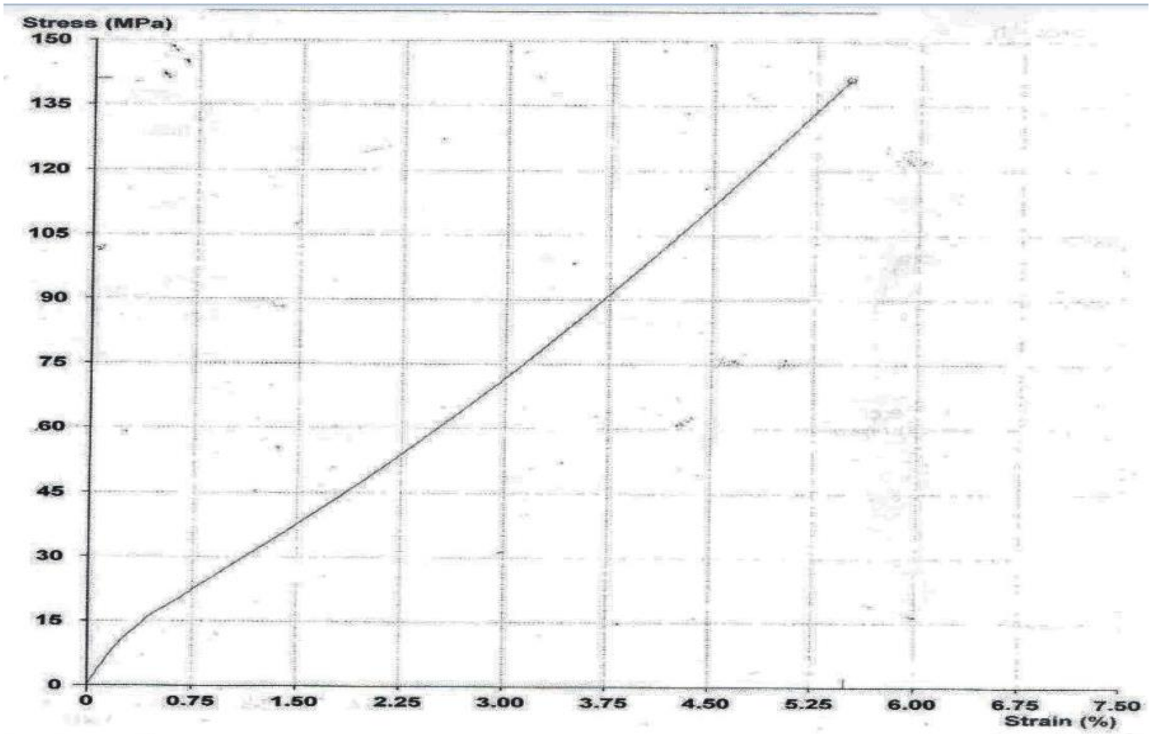


Figure (5.1): The stress-strain curve for suggested material for the second sample (S2)

Table (5.2): Mechanical properties of socket samples

Material	Ultimate stress σ_{ult} (Mpa)	Modulus of elasticity (E)(Gpa)
S1	92.03	1.773
S2	148.2	3.95
S3	96.1	2.113
S4	99.5	2.5
Available model[44]	33.65	0.941

From Table (5.2), it can be noticed that the values of mechanical properties increase by increasing in number of carbon layers, especially when perlon layers (nylon 6) were used. Based on results listed in Table (5.2) and comparisons of the properties between the available materials and the suggested materials, it was found that the suggested material have higher mechanical properties (e.g. σ_{ult} and E). The increasing of the mechanical properties is mainly caused by the number of carbon layers.

The second suggested sample (S2) showed superior results in terms of mechanical properties (σ_{ult} and E) when they were compared to those in the first sample. The reason behind that is the 4 layers of carbon and the arrangement of carbon layers within the matrix. Ultimate stress and modulus of elasticity were increased by 61.03% and 123.16%, respectively. Comparing between the third and fourth samples with the first sample showed a slight increase in ultimate stress about (4.4%) and (8.11%) respectively, the nut palm powder does not give any significant effect on the strength of material, therefore will be canceled.

5.4 Flexural and Fatigue test

The flexural test has revealed the average value of bending modulus of elasticity of 8.25 GPa. This value is used later to calculate the value of alternating stress. Fatigue failure happens when the specimen is fractured under dynamic loading. The readings were recorded by the fatigue testing machine representing the number of cycles until fracture. The relationship between the stress and the number of cycles from fatigue testing is depicted in Figure (5.2) and Figure (5.3). These results were obtained using two materials (suggested and available models) for eight samples. The equation of fatigue curves can be represented in terms of a number of cycles (N), applied stress (σ) and stress endurance limit from (S-N) curve. For material in S2, Eq. (5-1) is used to determine that terms. The calculation of stress endurance limit at ($N = 10^6$) [43] is equal to 71.6 MPa. On the other hand, with available model, the fatigue is given by the Eq. (5-2), and the stress endurance limit

(σ_e) is equal to 7.05 MPa. The (S-N) curves for material in S2 and available material are shown in figures (5.3) and (5.4) respectively. These results indicated that the fatigue strength of suggested model was higher than that of the available one by 915%. In general, the fatigue strength of materials is a function of its stiffness. In the other words, materials with higher modulus of elasticity possess have higher fatigue strength [50]. It can be noticed that lamination for S2 has the best fatigue life.

$$\sigma_e = 360.82(N_f)^{-0.117} \tag{5-1}$$

$$\sigma_e = 451.1(N_f)^{-0.301} \tag{5-2}$$

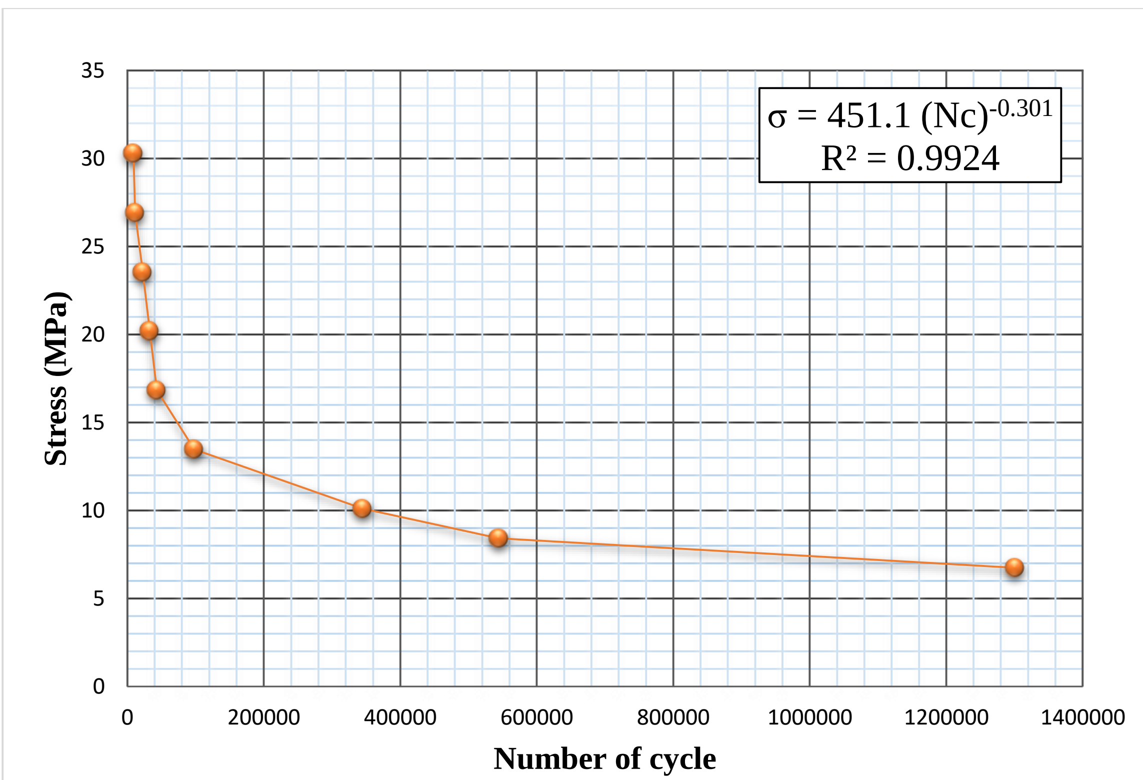


Figure (5.2): S-N fatigue curve for available material

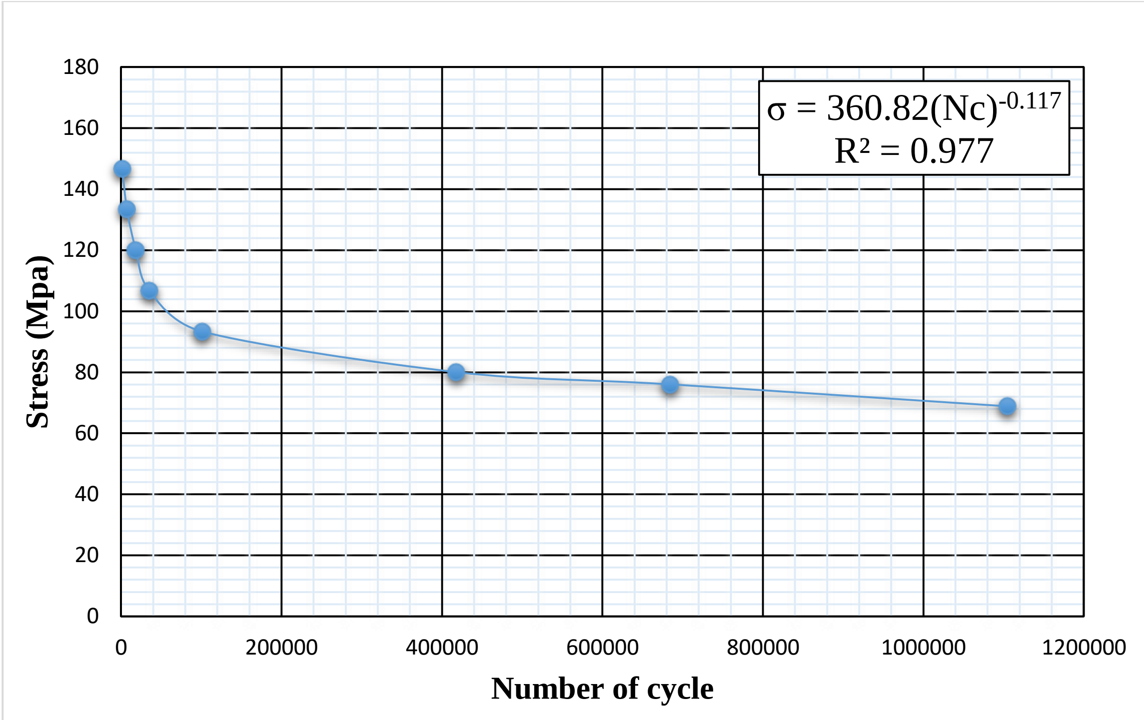


Figure (5.3): S-N fatigue curve for suggested material (S2)

5.5 Gait Cycle and (GRF) Tests

The results of a step for the two cases (available and suggested socket) are summarized in Table (5.3). The results include the prosthetic right leg (injured leg) and left leg (intact leg) for two cases.

The differences can be observed between to the left leg and right leg for the first case (available socket) and the second case (suggested socket). The difference in the step length for the suggested socket is (0.02 m) which was less than the difference in the available socket which was (0.119 m). While the difference for the step width in the second case recorded (0.005 m) which is less than that in the first case which was (0.02 m). Based on these results, it can be concluded that the amount of difference in step length and step width was decreased by (83.2%) and (75%) respectively when the suggested materials were used in socket.

	Available Prosthesis			Suggested Prosthesis		
Step –Strip table	Left Leg	Right Leg	Difference	Left Leg	Right Leg	Difference
Step time (sec)	0.7	0.79	0.09	0.73	0.7	-0.03
Step length (m)	0.313	0.432	0.119	0.45	0.47	0.02
Step velocity(m\sec)	0.285	0.561	0.276	0.631	0.701	0.07
Step width (m)	0.201	0.181	-0.02	0.156	0.151	-0.005

Table (5.3) Step table for the available and suggested prosthesis.

The results of the gait cycle test were detailed in Table (5.4) for available and suggested sockets. They showed that the available socket prosthesis records gait cycle time in the right leg was (1.56 sec) while the left leg records time of (1.69 sec). On the other hand, the suggested socket prosthesis has gait cycle time of (1.50 sec), and the left normal leg has time of (1.59sec). According to the obtained results, it can be concluded that the amount of change in gait cycle time is decreased for prosthetic and normal legs by (3.84 %) and (5.91%) respectively.

	Available Prosthesis			Suggested Prosthesis		
Gait cycle table(sec	Left Leg	Right Leg	Difference	Left leg	Right leg	Difference
Gait cycle Time	1.69	1.56	-0.13	1.59	1.5	-0.07
Stance Time(sec)	1.07	0.91	-0.16	0.93	0.91	-0.02
Swing Time(sec)	0.63	0.54	-0.09	0.5	0.57	0.07
Initial Double time	0.24	0.3	0.06	0.23	0.25	0.02
Total Double time	0.51	0.51	0	0.46	0.46	0
Heel Contact Time	0.66	0.81	0.15	0.57	0.64	0.07
Foot Flat Time	0.66	0.53	-0.13	0.26	0.16	-0.1

Table (5.4) Gait cycle table for the available suggested prosthesis.

The overall gait cycle test results for the available model and suggested one are summarized in Table (5.5). They showed that the value step width between the normal leg and manufactured one made from available materials was 93.2%, but for suggested materials the value was 96.6%. That means the center of mass (COM) was closer to the natural state, and that leads to increase stability and equilibrium in the patient's locomotion [51]. This can be considered as an improvement in the prosthetic field. The suggested socket provides the same degree of comfort according to the person who used it in the present work.

Symmetry table	Available Prosthesis	Suggested Prosthesis
Step Time (%)	90	95.7
Step Length (%)	58.4	93.6
Step Velocity (%)	52.1	89.3
Step Length\Leg Length (%)	46.8	93.1
Step Width (%)	93.2	96.6
Gait Cycle Time (%)	93.5	95.5
Single Support time (%)	83.4	89.1

Table (5.5) Symmetry table for the available and suggested prosthesis.

The results of GRF are shown in Figures (5.4) and (5.5) for the available and suggested S2 sockets, respectively. It is noticed that the GRF for the left leg (intact leg) with available socket was 920 N at initial contact and 900 at toe-off, while with the suggested socket GRF was 900 N and 820 N at initial contact and toe-off. The GRF for the right leg (injured leg) in first case was 780 N and 580 N at initial contact and toe off respectively while for the second case it became 780 N and 650 N at initial contact and toe-off respectively. This shows that the behavior of the gait cycle of the available prosthesis was diverged from the suggested one. This difference is due to the effect of the patient's weight on the prosthetic socket during locomotion.

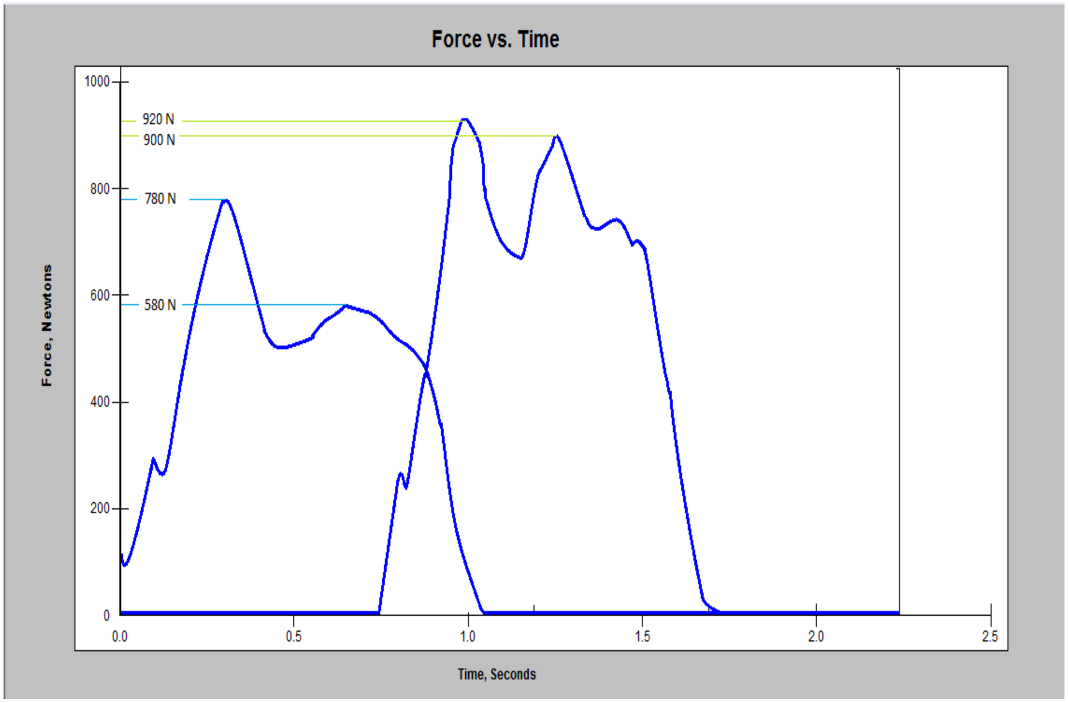
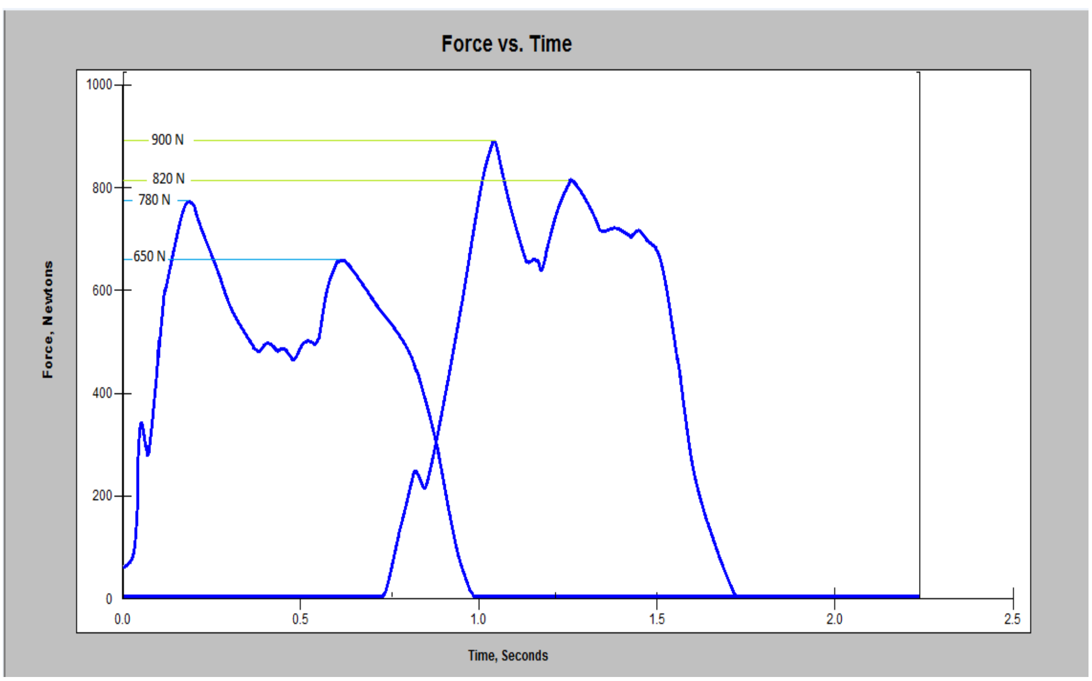


Figure (5.4) The GRF for the prosthesis for available socket



Figure(5.5) The GRF for the prosthesis for suggested socket (S2)

The scanning of the foot on the force plate is shown in figures (5.6) and (5.7) at case one and case two respectively. It can be observed that the difference of pressure distribution between the intact foot and the prosthesis was decreased in case two as compared to case one.

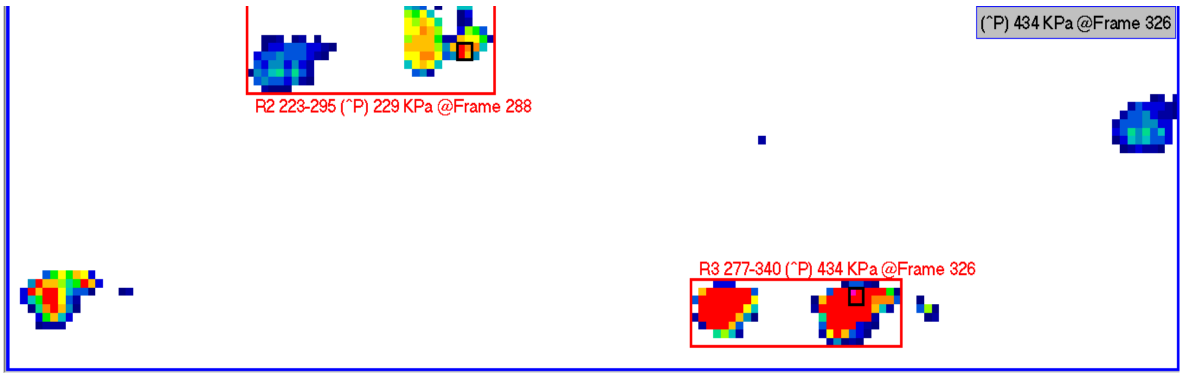


Figure (5.6): The foot scans for the prosthesis for available socket

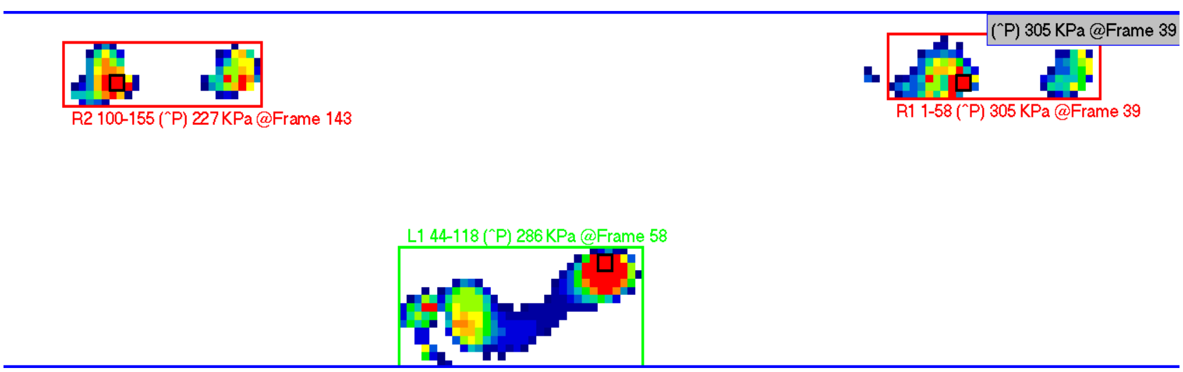


Figure (5.7): The foot scans for the prosthesis for suggested socket (S2)

5.6 Results of Interface Pressure Test

The socket interface designs can be classified into four main classifications based on their respective weight bearing characteristics (Anterior, Lateral, Medial and posterior). After carrying out the tests and obtaining the results of interface pressure for these regions. These regions were divided into three parts: upper, middle and bottom as shown in Figure (5.8) for the posterior region. The normal load exerted on the stump is not distributed uniformly. The reason behind this is because of the fact that some parts of the stump carry a higher load than others. The positions and values of the pressure on the inner sides of the socket are summarized in Table (5.6). Higher pressure presented in the anterior and posterior areas when compared to the lateral and medial areas. That results from the anterior and posterior muscles' contraction and flattened during the movement which cause pressure on the socket. These values of interface pressure are to be imported to ANSYS in a further step in order to be used as boundary conditions.

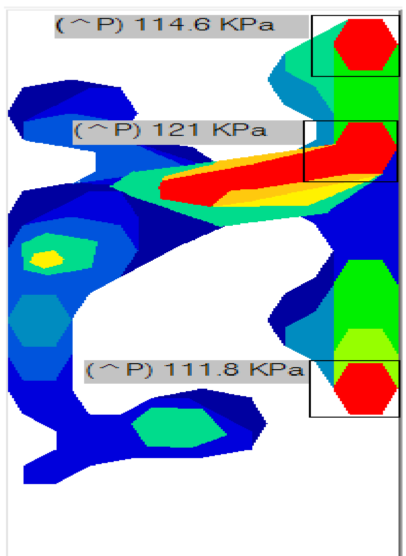


Figure (5.8): Interface pressures distribution in the pressure sensor for the anterior area.

Table (5.6): Values of interface pressure for the prosthetic socket

Socket Regions	**Anterior**			**Lateral**			**Posterior**			**Medial**		
Sensor Positions	AU	AM	AB	LU	LM	LB	PU	PM	PB	MU	MM	MB
Interface Pressure (KPa)	114.6	121	111.8	74.2	56.6	51.5	103	101.5	84	39	52.4	47.1

5.7 Stiffness to Weight Ratio (Specific Modulus)

Table (5.7) shows the results of stiffness to weight ratio (specific modulus) under unit loads, for axial stiffness (F_y), bending stiffness (F_x) and torsion stiffness (M_y) ($F_y = 1\ N,\ F_x = 1\ N\ and\ M_y = 1\ N.m$) by using ANSYS for the available and the suggested prosthetic sockets. Figures (5.9), (5.10) and (5.11) showed that the values of axial stiffness, bending stiffness and torsion stiffness for different socket's materials. From these figures, it can be seen that the stiffness for suggested material (S2) is higher than that of the available material's. That increase comes from the increase in the values of the mechanical properties for suggested materials(S2).

Table (5.7): deflection under unit load, stiffness and stiffness per weight ratio for suggestion and available sockets.

Socket	Unit load	Deflection δ *(mm)*	Stiffness $K = \frac{1}{\delta}(N/m)$	Density (kg /m^3)	Stiffness to weight ratio (K/ρ) *(N.kg/* m^4*)*
Suggested socket	F_y	$1.468 * 10^{-6}$	$6.8 * 10^8$	1267	$5.36 * 10^5$
	F_x	0.00248	$4.03 * 10^5$	1267	$3.18 * 10^2$
	M_y	0.00112	$8.92 * 10^5$	1267	$7.04 * 10^2$
Available socket	F_y	$4.51 * 10^{-6}$	$2.21 * 10^8$	1252	$1.76 * 10^5$
	F_x	0.0072	$1.38 * 10^5$	1252	$1.10 * 10^2$
	M_y	0.0032	$3.12 * 10^5$	1252	$2.49 * 10^2$

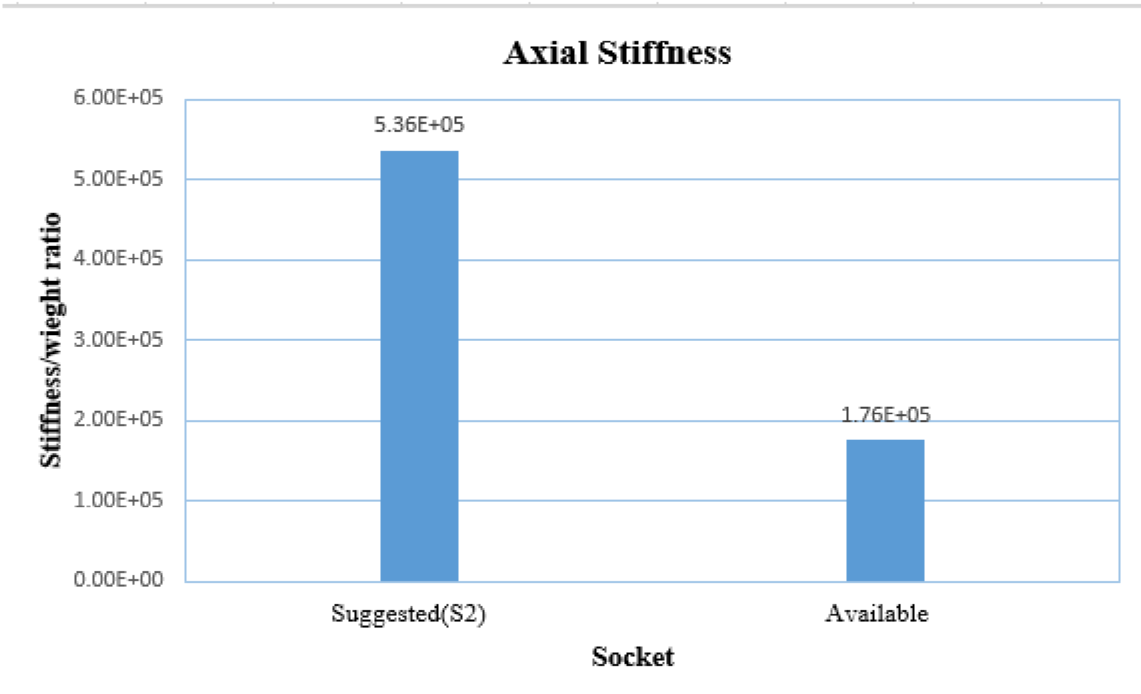


Figure (5.9): Axial stiffness for the prosthetic sockets

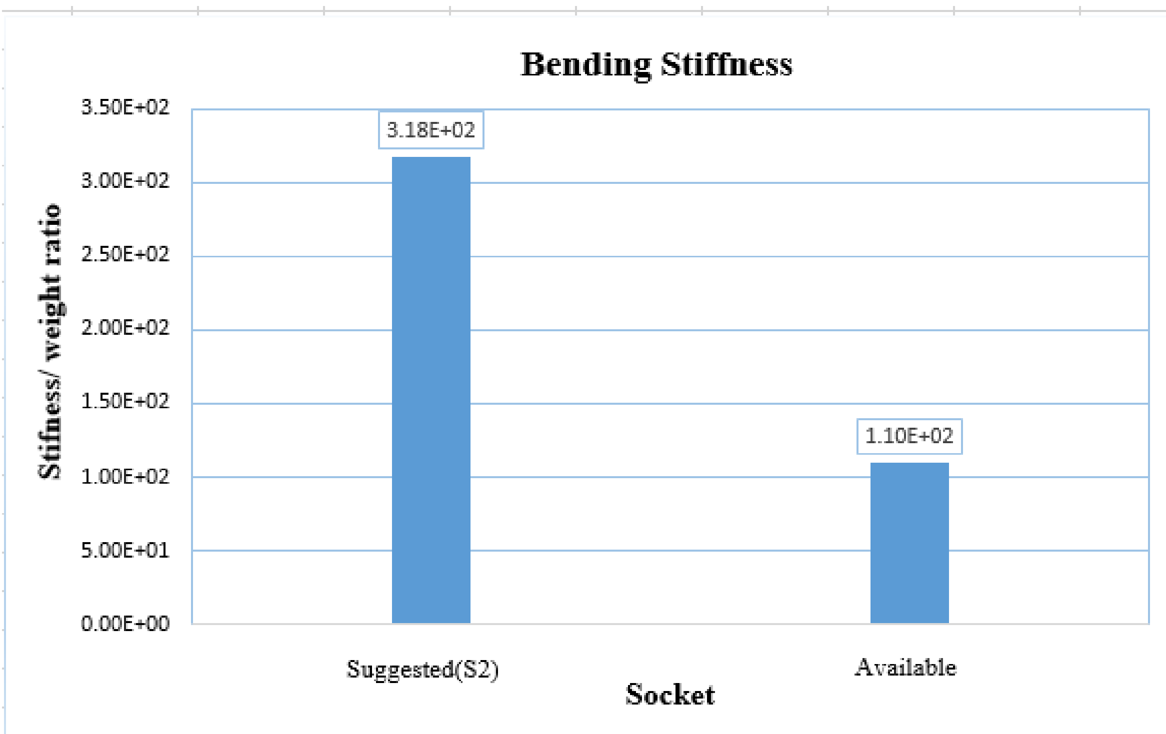


Figure (5.10): Bending stiffness for the prosthetic sockets

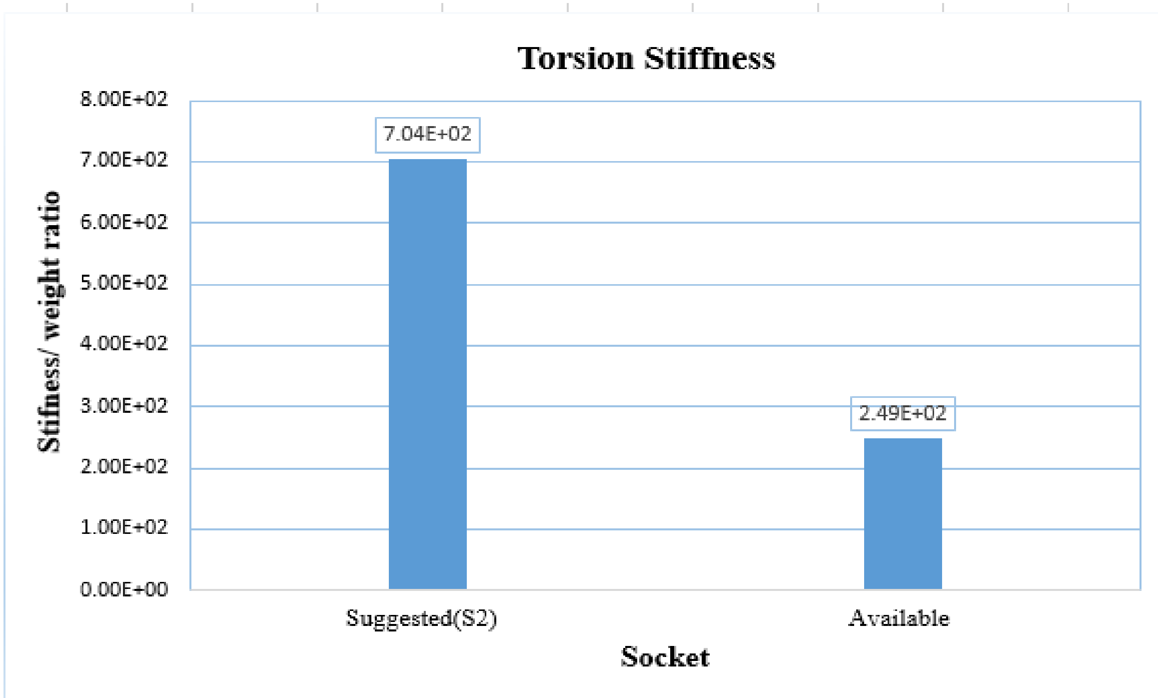


Figure (5.11): Torsion stiffness for the prosthetic sockets

5.8 Results of the Numerical Analyses

The static and fatigue properties of prosthetic socket were investigated using finite element method. Two cases of prosthetic socket were analyzed by simulations conducted by using ANSYS 14.5 software. For the experimental and analytical part, the first case was the available prosthetic socket material (4 perlon + 2 fiberglasses + 4 perlon), and the second one was the suggested prosthetic socket (3perlon+1carbon fiber+1perlon+2carbon fiber+1perlon+1carbon fiber +3perlon).

The physical properties (e.g. density) and mechanical properties (e.g. ultimate stress ($\boldsymbol{\sigma_{ult}}$), Modulus of elasticity ($\boldsymbol{E}$) and fatigue life) for the cases mentioned above were estimated in the experimental part (tensile test) of this work. While the interface pressures between socket and stump was measured for two cases: first case from the experimental interface pressure (F-socket test), and the second case was calculated in the analytical part of this thesis.

The analysis of Von-Mises Stresses allowed to understand the propagation of stress in the socket and gave us the amounts of stress induced. For the experimental and analytical parts, it can be observed that the highest values of stresses for the suggested material S2 are (14.43 MPa) and (15.42 MPa) as shown in Figures (5.12) and (5.13). Figures (5.14) and (5.15) represent the Von-Mises stress contours for the experimental and analytical parts respectively for the available socket.

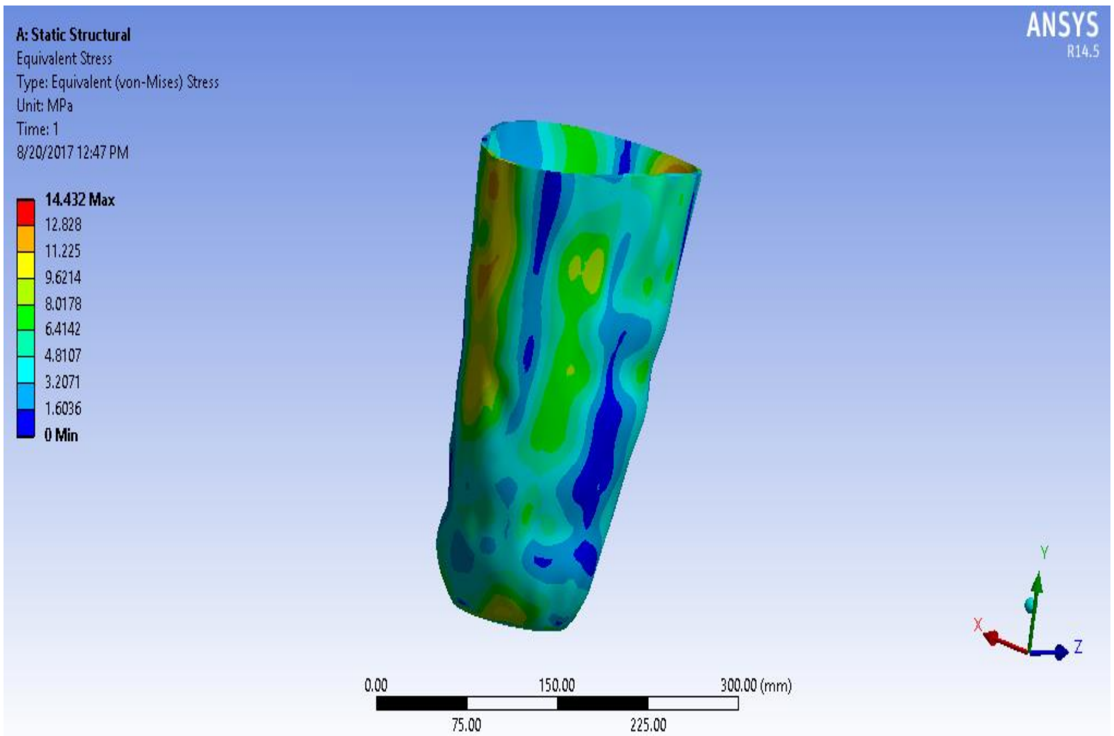


Fig. (5.12): Numerical Von-Mises stress for experimental part for the suggested prosthetic socket (S2)

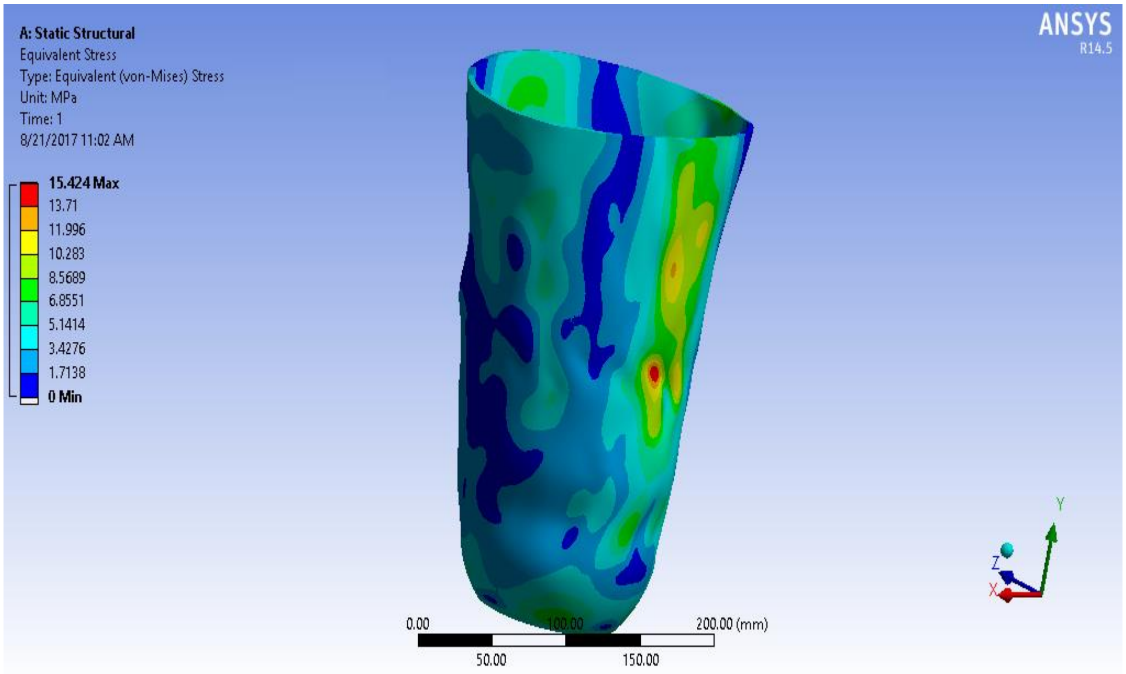


Fig. (5.13): Numerical Von-Mises stress for analytical part for the suggested prosthetic socket (S2)

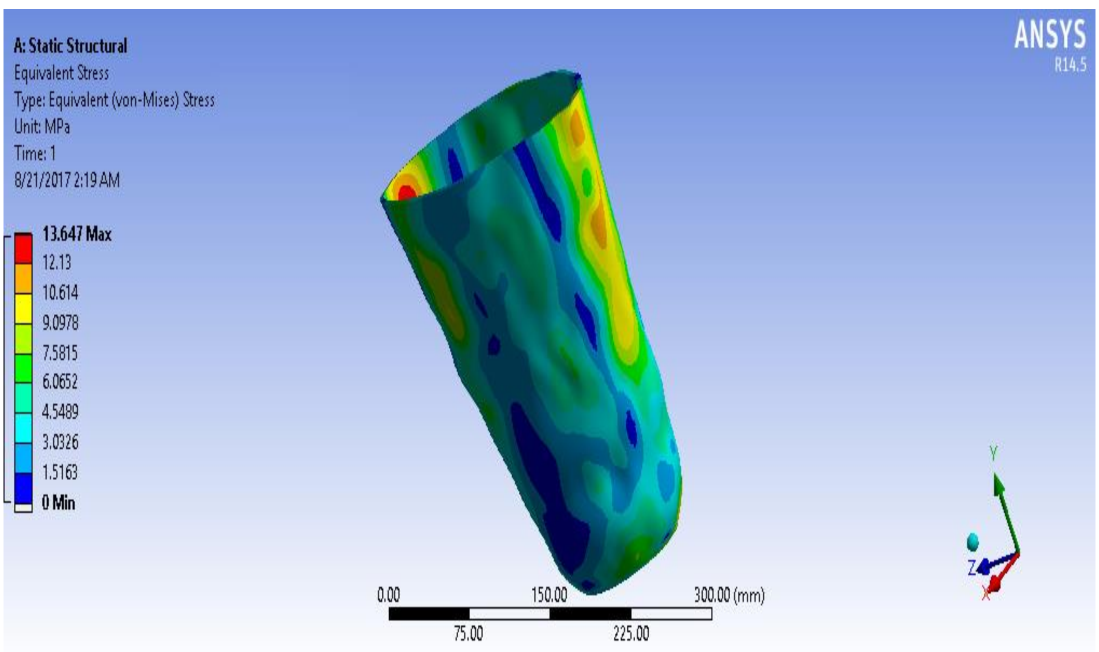


Fig. (5.14): Numerical Von-Mises stress for experimental part for the available socket

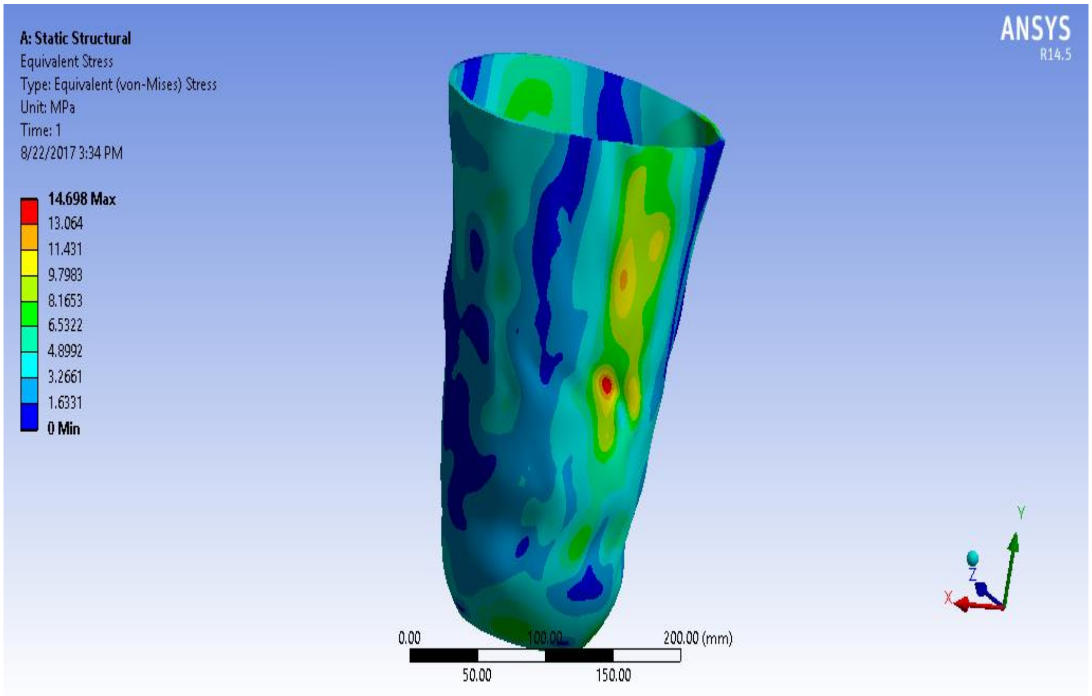


Fig. (5.15): Numerical Von-Mises stress for analytical part for the available socket

The analytical study of deformation helps to know the amounts of directional deformations in (x, y, z) and their positions in the socket. From this study, it can be observed that the highest value of deformation exits in the (x) and (z) directions, while the lowest value takes place in the (y) direction for the suggested and available socket.

For the experimental results, Figures (5.16), (5.17) and (5.18) represent the directional deformations in three axes (x, y, z) respectively for the prosthetic socket made of suggested composite material (S2). The values of maximum deformation for x, y, z directions are 2.937 mm, 0.448 mm and 2.92 mm, respectively. Figures (5.19), (5.20) and (5.21) represent the directional deformation in three axes (x, y, z) respectively for the available prosthetic socket. The values of maximum deformation for x, y, z directions were 8.503 mm, 1.301 mm and 8.43 mm respectively.

For the analytical results, Figures (5.22), (5.23) and (5.24) represent the directional deformation in three axes (x, y, z) respectively, for the prosthetic socket made of suggested sequence of composite material (S2), and the values of the maximum deformation for x, y, z directions were 1.974 mm, 0.382 mm and 2.429 mm respectively. Figures (5.25), (5.26) and (5.27) represent the directional deformation in three axes (x, y, z) respectively, for the prosthetic socket. The values of the maximum deformation for x, y, z directions were 5.729 mm, 1.114 mm and 7.02 mm respectively.

From the previous results, it can be noticed that deformations are at their minimum values for the suggested design in both studies, experimental and analytical. That can be justified by the fact that the suggested design has superior mechanical properties when compared with the available material, and all figures showed the anterior zone has maximum deformation because the maximum value of pressure in interior zone.

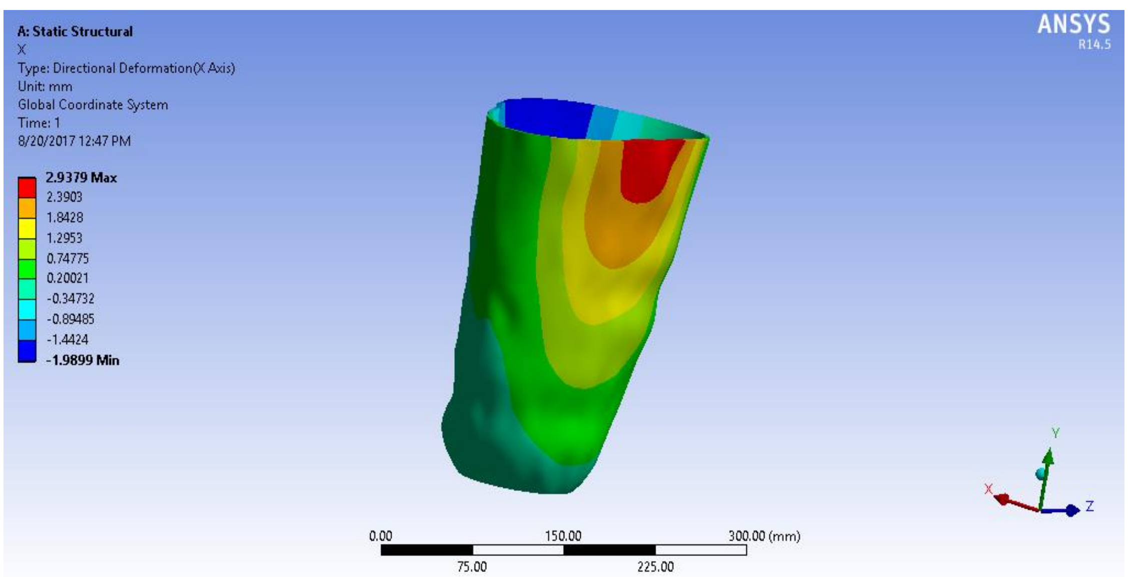


Figure (5.16): Directional deformation (x-axis) for suggested prosthetic socket (S2) for the experimental part

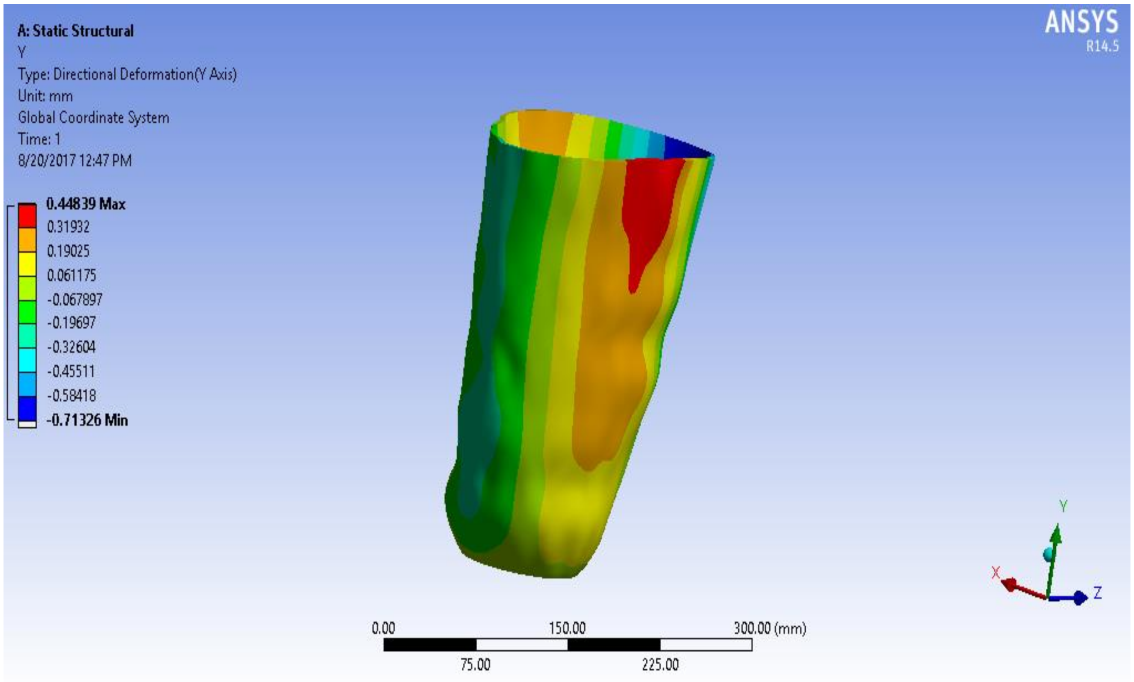


Figure (5.17): Directional deformation (y-axis) for suggested prosthetic socket (S2) for the experimental part

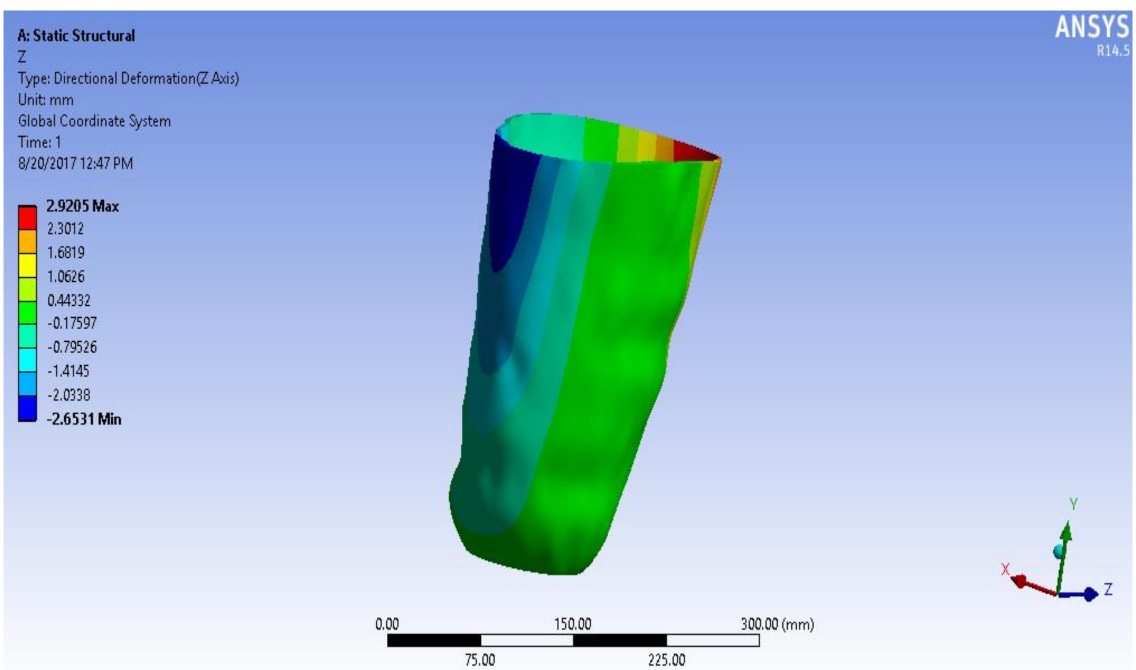


Figure (5.18): Directional deformation (z-axis) for suggested prosthetic socket (S2) for the experimental part

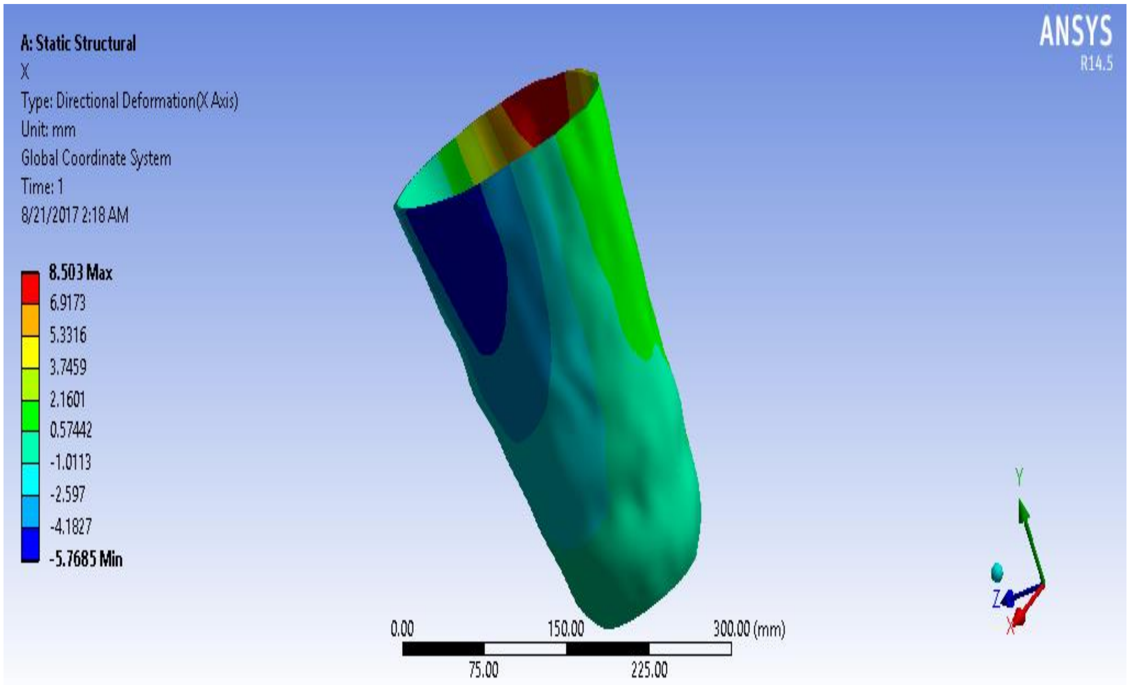


Figure (5.19): Directional deformation (x-axis) for available prosthetic socket for the experimental part

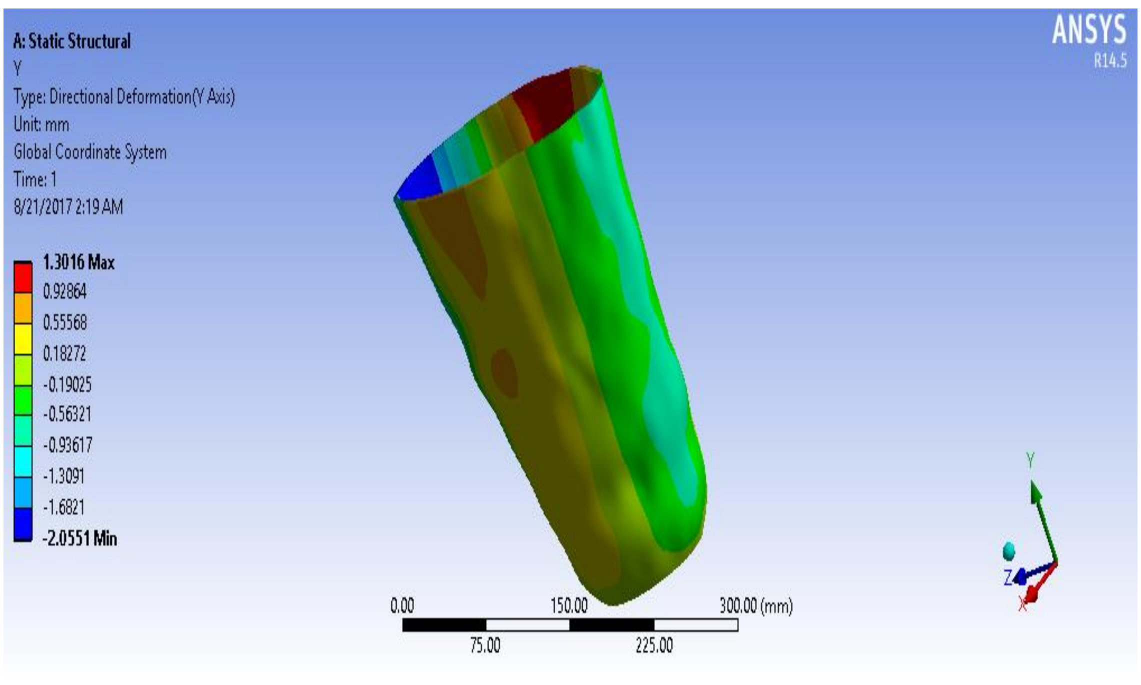


Figure (5.20): Directional deformation (y-axis) for available prosthetic socket for the experimental part

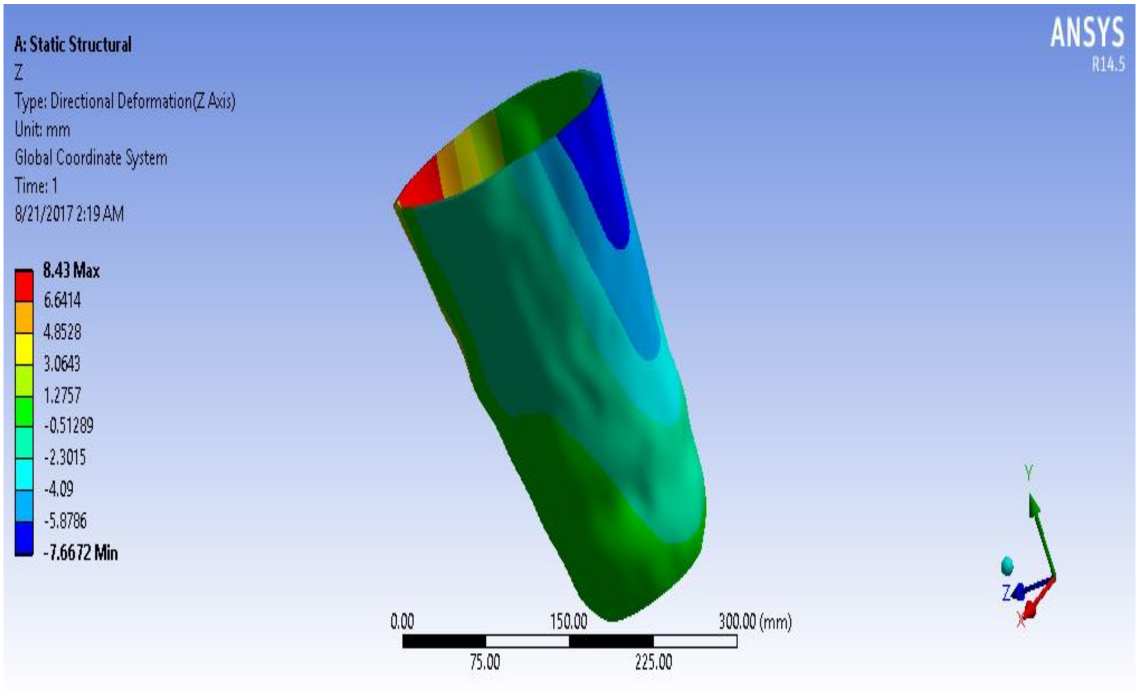


Figure (5.21): Directional deformation (z-axis) for available prosthetic socket for the experimental part

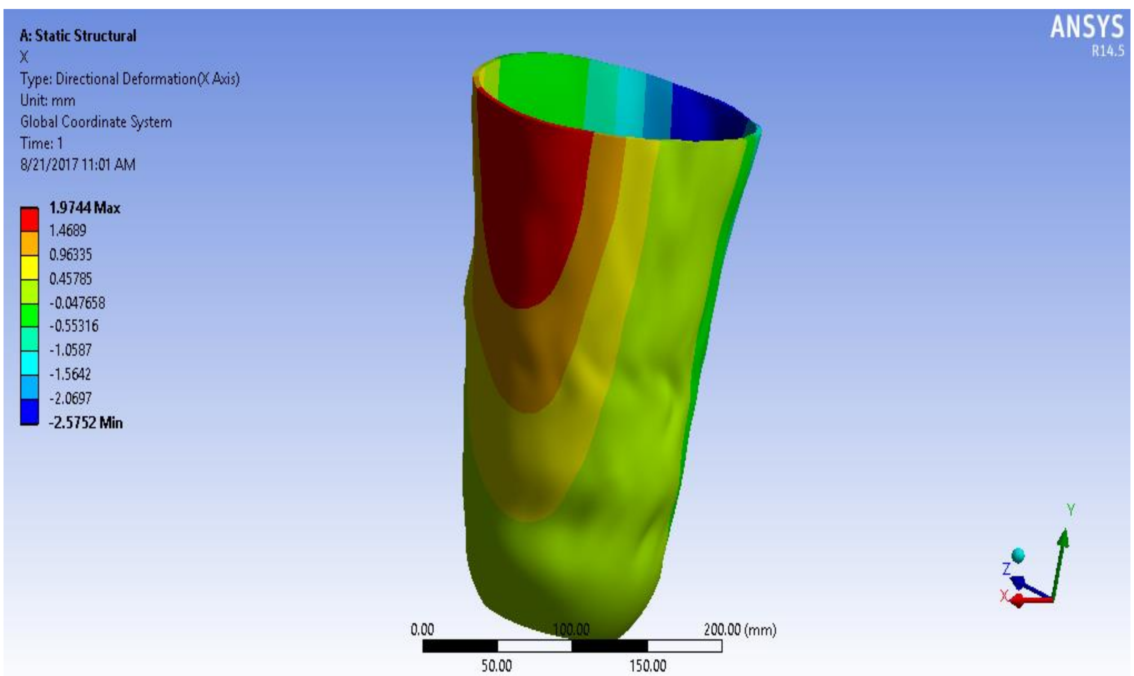


Figure (5.22): Directional deformation (x-axis) for suggested prosthetic socket (S2) for the analytical part

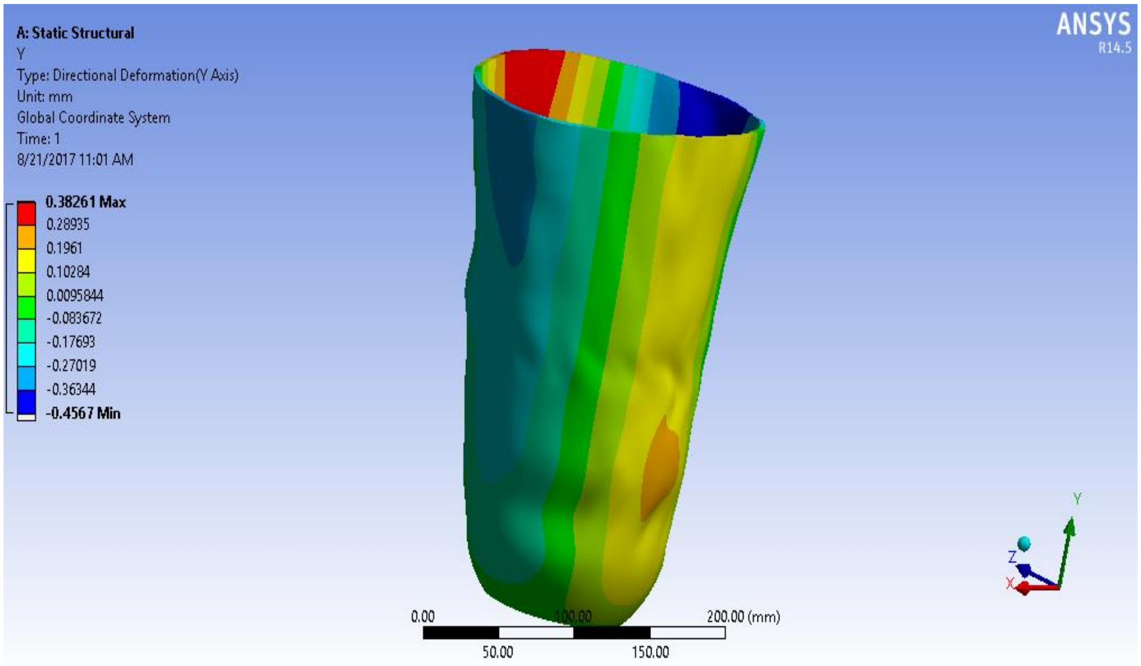


Figure (5.23): Directional deformation (y-axis) for suggested prosthetic socket (S2) for the analytical part

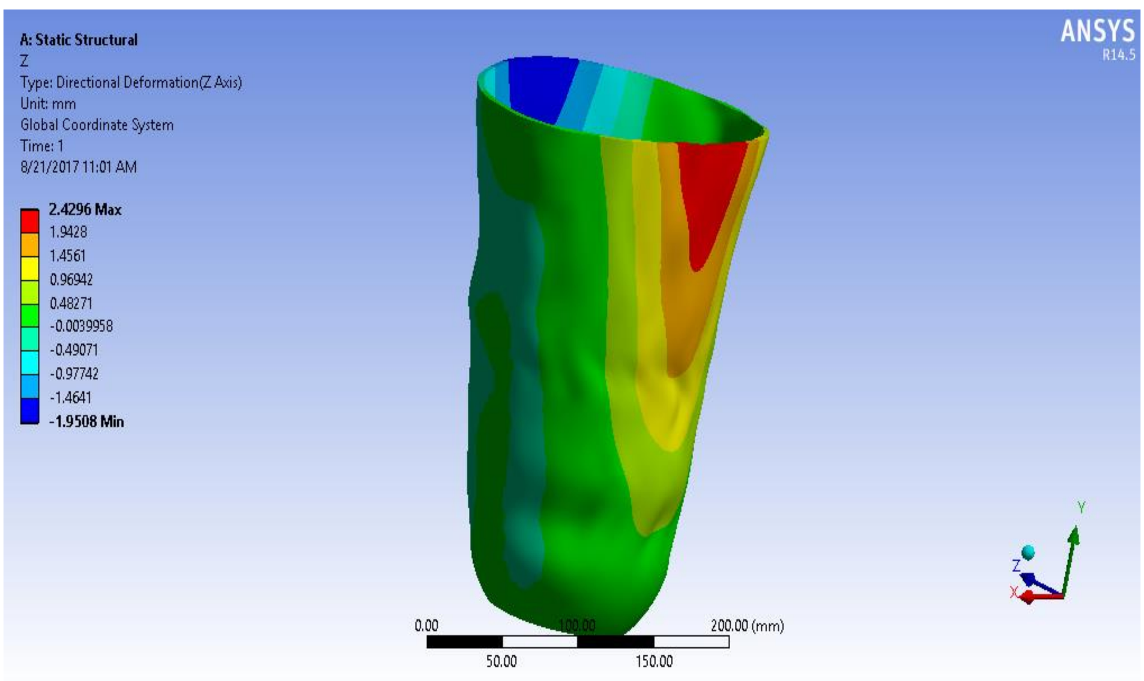


Figure (5.24): Directional deformation (z-axis) for suggested prosthetic socket (S2) for the analytical part

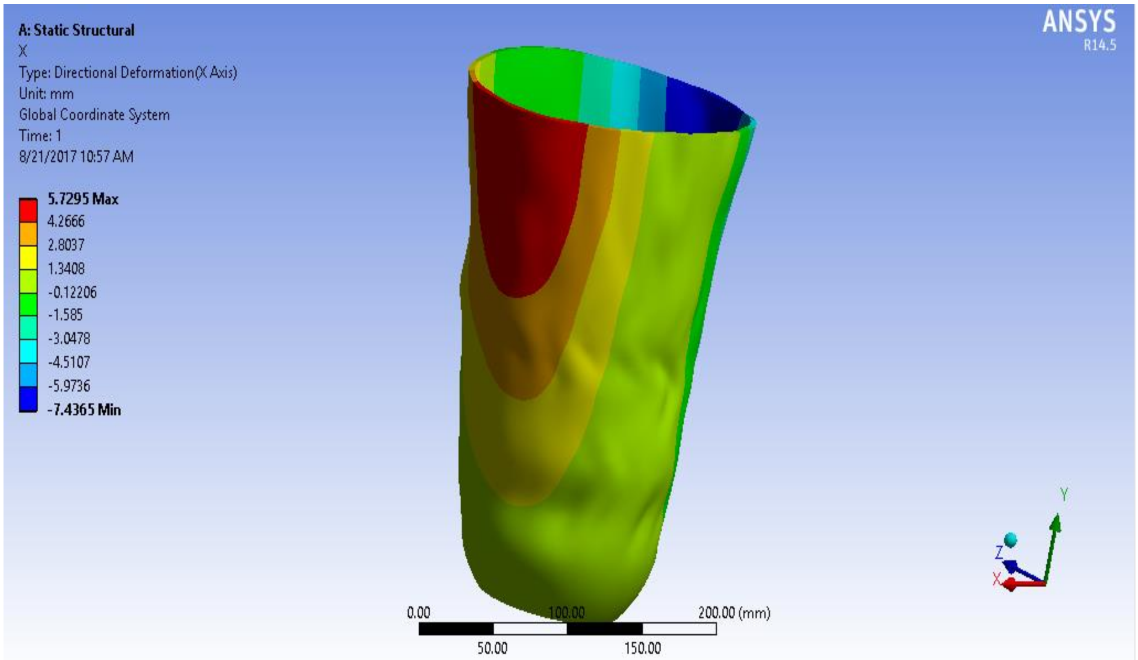


Figure (5.25): Directional deformation (x-axis) for available prosthetic socket for the analytical part

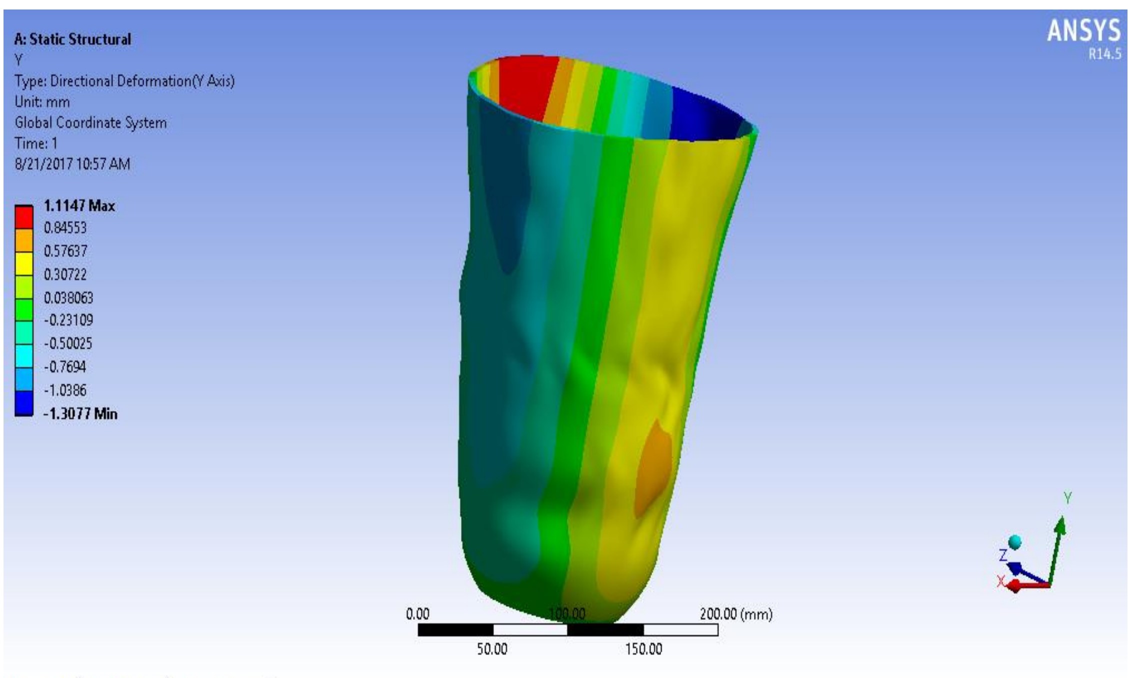


Figure (5.26): Directional deformation (y-axis) for available prosthetic socket for the analytical part

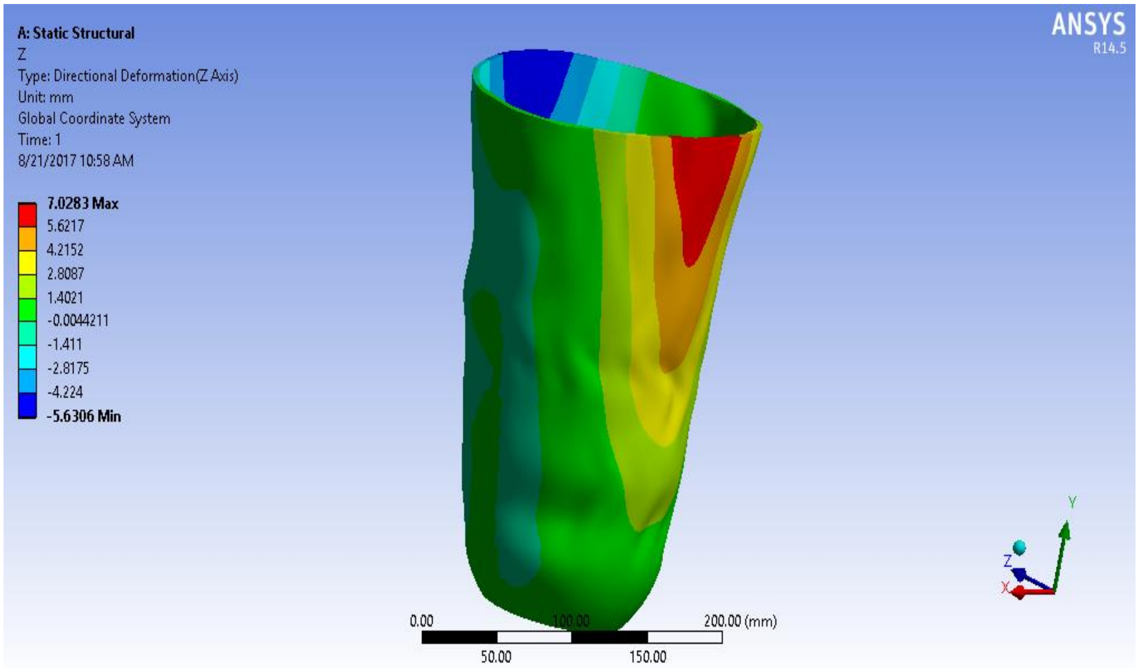


Figure (5.27): Directional deformation (z-axis) for available prosthetic socket for the analytical part

The minimum safety factors obtained experimentally and analytically for the suggested prosthetic socket are 4.832 and 4.932, respectively as shown in Figures (5.28) and (5.29). For the available prosthetic socket, the minimum safety factors were 0.611 and 0.625 as shown in Figs (5.30) and (5.31) respectively.

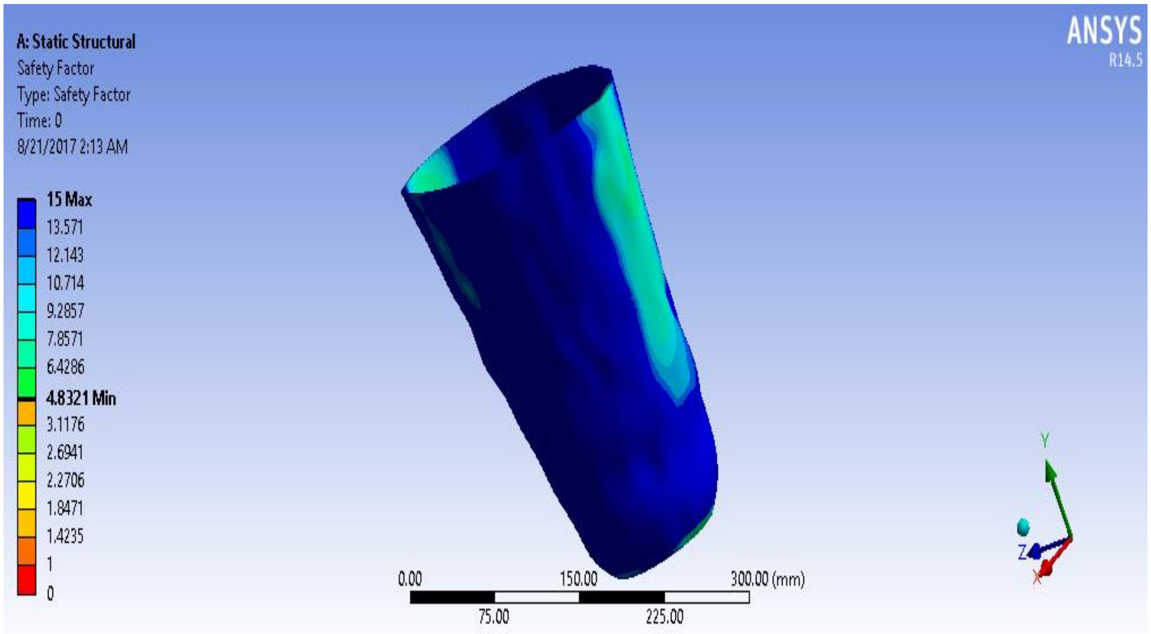


Figure (5.28): The minimum equivalent safety factor for suggested prosthetic socket for the experimental part

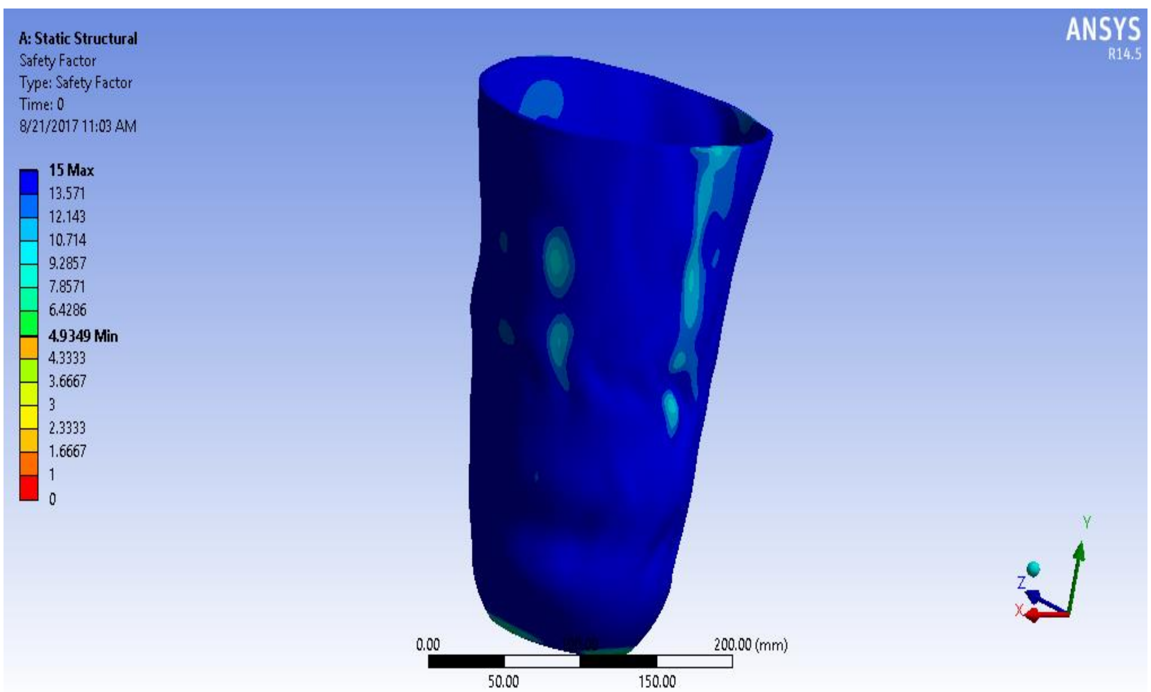


Figure (5.29): The minimum equivalent safety factor for suggested prosthetic socket for the analytical part

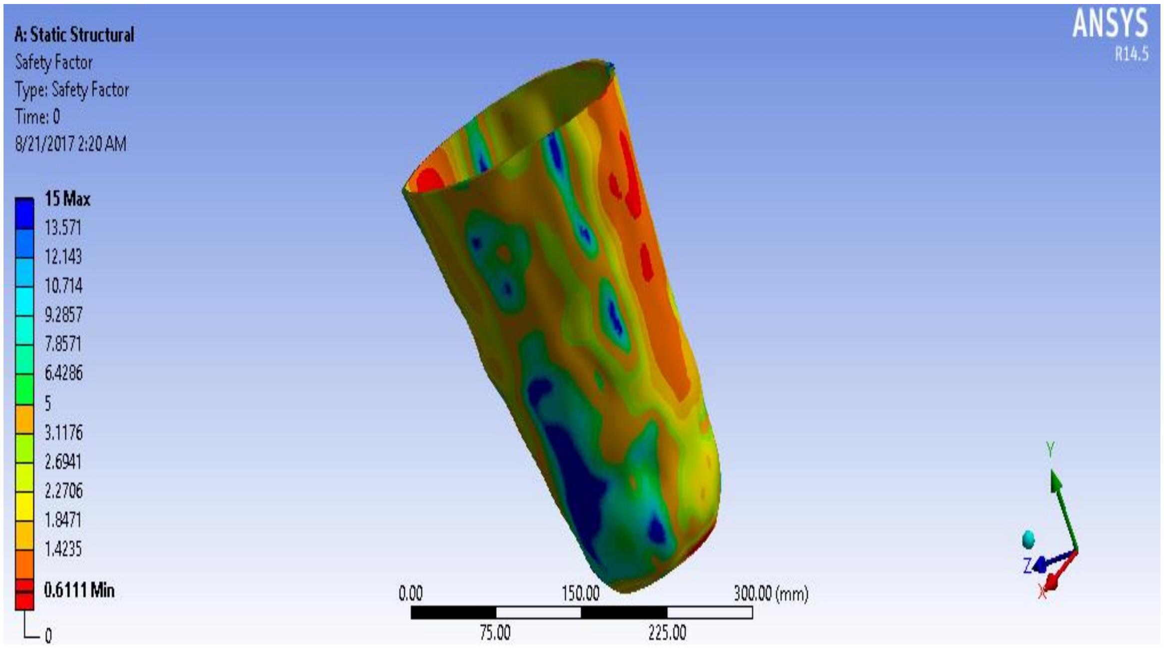


Figure (5.30): The minimum equivalent safety factor for available prosthetic socket for the experimental part

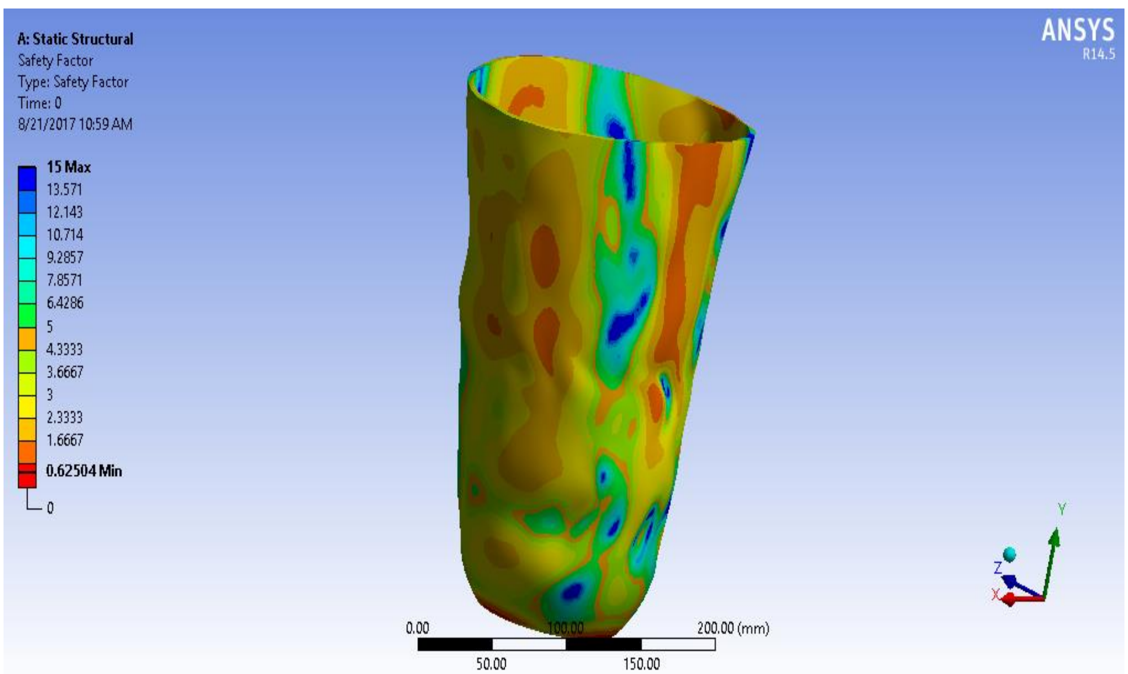


Figure (5.31): The minimum equivalent safety factor for available prosthetic socket for the analytical part

Chapter Six

Conclusions and Recommendations for Future Work

6.1 Conclusions

From the previous discussions, the following conclusions can be summarized as follows:

1. There is a great effect in the strength of the socket with changing its material, it is noted that by using the stacking sequence: (3 perlon +1 carbon fiber +1 perlon + 2 carbon fiber + 1 perlon + 1 carbon fiber + 3 perlon) the strength is increased, according to tensile test results.
2. The use of palm nuts powder did not give a significant effect on the strength of the material, which has a contradiction of the previous suggested work in this respect.
3. The suggested model (S2) has a perfection in both, the ultimate stress and modulus of elasticity, by 61.03% and 123.16%, respectively as it is compared with the available design.
4. As a result of fatigue test the suggested design of composite material sequence has an improvement (915%) in the stress endurance limit when comparing it with the available material.
5. The calculation of the stiffness to weight ratio shows that the suggested socket (S2) has a value higher than of the available one for axial stress (by 204.5%), bending stress (by 189%) and torsion stress (by 182.7%).
6. The safety factor for both sockets showed that the experimental results of the minimum safety factor for the suggested socket (S2) and available socket were 4.83 and 0.6111, respectively. Almost the results of the minimum safety

factor were obtained analytically, for the suggested socket and the available one, the minimum safety factor values were 4.93 and 0.625, respectively

6.2 Recommendations for Future Work

Several recommendations can be suggested for future work as follow:

1. Studying the vibrational characteristics for suggested materials
2. Using nano materials in suggested materials and studying the effect of their presence on the mechanical properties.
3. Taking the creep behavior of the prosthetic socket into account is another interesting scope of research.
4. Capturing the effect of crack growth on the prosthetic socket.
5. Tacking the fatigue behavior for suggested socket.

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Appendix (A)

The MATLAB coding is shown below

```
D_upper=input('Enter a value for D_u:');%['Upper diameter for socket in
mm']
D_lower=input('Enter a value for D_l:');%['Lower diameter for socket in
mm']
H=input('Enter a value for H:');%['the Height of the socket in mm']
t=input('Enter a value for t:');%['the average tissue thickness in mm']
Theta=atand((D_upper-D_lower)/(2*H))%('conical angle in Degree')%('the
conical angle in Degree')
L=H/cosd(Theta)%['Length of the Limb contacts with the socket in mm']
E=input('Enter a value for E:');%['Young Modulus for soft tissue in
kPa']
v=input('Enter a value for v:');%['the poissons ratio']
G=(E)/(2+(2*v))%('Shear Modulus in kPa')
mu=input('Enter a value for u:');%['coefficient friction']
M=input('Enter a value for M:');%['patient Mass in KG']
W=M*9.81%['patient weight in N']
A=pi*L*(D_upper+D_lower)/2%['Area of the supporting surface of skin']
K_n=E*A*1000/(t*1000000)%['the vertival spring constant in N/mm']
K_s=G*A*1000/(t*1000000)%['the parallel spring constant']%['the vertival
spring constant in N/mm']
K=((K_n*(sind(Theta))^2)+(K_s*(cosd(Theta))^2))%['the interface
stiffness in N/mm']
D_n0=(W/K_n)*((cosd(Theta))-
(mu*(K_n/K_s)*sind(Theta)))/((((K_n/K_s)*(sind(Theta)^2))+(cosd(Theta)^2
))*(mu+tand(Theta)))%['pre-compressive displacement of vertcal spring in
mm']
Segma_0=(W*1000/A)*((cosd(Theta))-
(mu*(K_n/K_s)*sind(Theta)))/((((K_n/K_s)*(sind(Theta)^2))+(cosd(Theta)^2
))*(mu+tand(Theta)))%['average pre-pressure at skin socket interface in
kPa']
Segma=(K_n*1000/A)*(D_n0+(W*sind(Theta))/(K))%['average pressure in
kPa']
Taw=(K_s*1000/A)*(((-
K_n/K_s)*D_n0*tand(Theta))+((W*cosd(Theta))/(K)))%['average shear stress
in kPa']
resutant=((Segma)^2+(Taw)^2)^(1/2)%['resultant stress in kPa']
```

الخلاصه

الأطراف الصناعية هي اجهزة تعمل كبديل لعضو مفقود نتيجة اصابة او عيب خلقي. الاشخاص الذين لديهم بتر يمر عبر الركبة هم اكثر عرضة لحدوث فشل في الوقب الخاص بهم لمختلف حالات التحميل. عدة اسباب تقع خلف ذلك من ضمنها نوع المادة المصنوع منها الوقب وحالة الحمل المسلط عليه. في هذا البحث تم اقتراح مادة مركبة لتحسين الوقب لطرف صناعي واطالة عمره و زياده راحة مستخدمه.

يتضمن البحث جزئين: عملي ونظري. الجزء العملي اشتمل على نوعين رئيسيين من الاختبارات: النوع الاول عني بأجراء اختبارات ميكانيكية لمادة الوقب المقترحة والموجودة. المادة الموجودة تكونت من الطبقات (4 برلون 2 الياف الزجاج 4 برلون) بينما شكل الترتيب (3 برلون 1 الياف الكربون 1 برلون 2 الياف الكربون 1 برلون 1 الياف الكربون 3 برلون) الطبقات المستخدمة في المادة المقترحة لاجل الحصول على خواص فيزيائية وميكانيكية افضل اضافه لتحسين الاستقرارية والاتزان (لوقب طرف صناعي لبتر عبر الركبة) عند مقارنته مع الوقب الموجود.بينما لم يتم الحصول على تحسين في الخواص الميكانكيه عند اضافه نسب نوى التمر.

النوع الثاني من الاختبارات عني باجراء اختبارات لقياس قوة رد فعل الارض و توزيع الضغط على طرف صناعي موجود و اخر مقترح لذكر مبتور الطرف ذي 27 عاما و وزن 80 كغم وبطول 180 سم. استخدم حساس F-Socket لقياس الضغط البيني الموجود بين الطرف المبتور و الوقب. اظهرت حسابات الصلابة الى نسبة الوزن (الكثافه) ان الوقب المقترح يمتلك قيم اعلى اذا ما قورنت مع الوقب الموجود لحمل محوري (بنسبة 268.4 %) وحمل انثناء (بنسبة 235%) و حمل التواء (بنسبة 229%).

الجزء النظري هو الثاني من هذا البحث ويتضمن بحثا نظريا و عدديا لغرض محاكاة طرف صناعي عبر الركبة وحساب قيم اجهادات (Von-Mises) ومعامل الامان وذلك عبر اجراء تحليل عددي باستخدام برنامج ANSYS.

تم الحصول على قيم الضغط البيني بين الطرف المبتور والوقب تحليليا و عمليا. استخدمت تلك القيم كمدخلات لبرنامج ANSYS.

اظهرت النتائج الخاصة بالمادة المركبة المقترحة ان اعلى قيمة لاجهاد Von-Mises تساوي 14.43 MPa بالنسبة للجزء العملي بينما حصلت القيمة 15.42 MPa تحليليا وقد لوحظ ان لترتيب الطبقات بالنسبة للمادة المقترحة اثر ايجابي على معامل الامان, تشوه المادة وراحة المريض اثناء الحركة.

وزارة التعليم العالي والبحث العلمي
جـــامــــــعة كربـــــــلاء
كلــــــــية الهندســــــــة
قسم الهندسة الميكانيكيـــة

دراسة مقاومة الكلال و حساب نسبة الصلابة الى الوزن لطرف صناعي عبر الركبة

رسالة مقدمة الى كلية الهندسة – جامعة كربلاء كجزء من متطلبات نيل درجة الماجستير في علوم الهندسة الميكانيكية
(ميكانيك تطبيقي)

من قبل
امير علاء عليوي
بكالوريوس 2012

بإشراف

أ.م.د. محسن عبد الله الشمري
أ.م.د. عماد قاسم حسين

2018