A PROJECT REPORT

ON

"DESIGN AND ANALYSIS OF BIONIC LEG"

Submitted by

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In partial fulfillment for the award of the Degree

Of

BACHELOR OF ENGINEERING

IN

MECHANICAL ENGINEERING

UNDER THE GUIDANCE

Of

Prof. SHAKIL TADVI

DEPARTMENT OF MECHANICAL ENGINEERING

ANJUMAN-I-ISLAM

KALSEKAR TECHNICAL CAMPUS NEW PANVEL,

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ANJUMAN-I-ISLAM

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CERTIFICATE

This is to certify that the project entitled

"DESIGN AND ANALYSIS OF BIONIC LEG"

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To the Kalsekar Technical Campus, New Panvel is a record of bonafide work carried out by him under our supervision and guidance, for partial fulfillment of the requirements for the award of the Degree of Bachelor of Engineering in Mechanical Engineering as prescribed by **University Of Mumbai**, is approved.

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APPROVAL OF DISSERTATION

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(Internal Examiner) **(External Examiner)**

Date:

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ABSTRACT

The actuated leg prosthesis comprises a knee member, a socket connector provided over the knee member, an elongated trans-tibial member having a bottom end under which is connected an artificial foot, and a linear actuator. At first pivot assembly allows to operatively connect the trans-tibial member to the knee member. A second pivot assembly allows to operative connect an upper end of the actuator to the knee member. A third pivot assembly allows to operatively connect a bottom end of the actuator to the bottom end of the trans tibial member. The prosthesis can be provided as either a front actuator configuration or a rear actuator configuration.

CHAPTER 01

PREAMBLE

CHAPTER 01: PREAMBLE

1.1 Introduction:

Artificial limbs, prosthetic limbs or prostheses are mechanical replacements for missing limbs –arms or legs. When people have the misfortune of losing their arms or legs due to injury, disease or birth defects, artificial limbs help amputees get back to their normal functioning to a certain degree. Cancer, infection and cardiovascular disease brought about by disorders like diabetes are leading illnesses that cause amputation. Serious calamities like the recent earthquake that bring in their wake thousands of seriously injured people who will need artificial limbs as their wounds begin to heal.

Fig.1 Bionic Leg

An upper extremity amputation that might result in the loss of the entire arm or a part of it could leave a person ill equipped to perform day to day activities involving the arm such as washing, writing, typing, lifting objects and so on. Similarly, a lower extremity amputation will have a person missing the whole or a portion of the leg and this could render the person incapable of walking, running and performing other activities involving the lower limbs. Artificial limbs are devised to afford better mobility involving limbs and give people the freedom of functioning without depending on others.

Artificial Limbs:

Ideally Artificial limbs are of two types- Exo-skeletal or Crustacean and Endo-skeletal or modular. The Exo-skeletal variety has a hard and rigid shell because the walls of the artificial limbs are responsible for the shape as well as the weight transmission. In the endo-skeletal type a central shaft covered by a cosmetic covering is used to transmit weight.

Artificial limbs are named according to the level of amputation performed to fit the limbs. The four main names for artificial limbs are:

Trans tibial- the artificial limb that replaces the missing part of leg below the knee

Trans femoral- the artificial limb above the knee

Trans radial- the artificial limb that replaces the arm below the elbow

Trans humeral- the artificial limb above the elbow

Modern day artificial limbs have moved a long way from the peg legs and cumbersome iron and wooden replacements for missing limbs used in olden days. Advanced surgical procedures enable precise amputations to fit appropriately devised artificial limbs. Lighter materials and improved computer aided designs in artificial limbs along with laser assisted measuring and fitting of the limbs, afford greater degree of flexibility and maneuverability for the user these days.

Artificial limbs must be light, flexible and easily adaptable to the user to permit easy movement. It should also be strong enough to support the body's weight if they are artificial legs or feet and manipulate objects if they are artificial arms or hands or parts of the same. Artificial limbs basically need to be functional, comfortable, afford a great degree of stability, be cosmetically acceptable, not too expensive, readily available and serviceable and preferably local for quick repairs and adjustments. Artificial limbs are usually made out of materials like willow wood, metallic alloys, fiber and plastic lamination and complex carbon –fiber substances.

Artificial limbs are not manufactured in bulk but most often custom made to suit individual needs. After initial consultation with the amputee, a prosthetic (healthcare professional skilled in making and fitting artificial limbs) and a physiotherapist, a medical doctor prescribes artificial limbs to the amputee. Following the prescription, the person visits the prostheses to have the artificial limb fitted. Before making the artificial limb the prostheses evaluates the amputee and takes a digital reading or at least an impression of the remaining part of the natural limb. Computers using a CAD/CAM to design artificial limbs and laser guided measuring and fitting have vastly improved the quality of artificial limbs available for amputees.

The cost of fitting artificial limbs depends on the materials used and the degree of sophistication involved in customizing the product to the user. Depending upon wear and tear, and also if the person gains or loses weight, the artificial limb may have to be replaced once in every 3 or 4 years. While there are artificial limbs that cost over Rs.5, 20,000/- a piece there are also well designed, low cost, functional ones like the popular Jaipur foot manufactured in India.

1.2 OBJECTIVE

The Objective for this project are to discuss and analyze the recent developments in the field of prosthetics, their applications and potential future. In order to predict how the future of this field will be shaped, one has to turn to the past.

With the work presented in this, we sought to contribute to the development of integral assistive and rehabilitation technologies that can adapt to the needs of the physically challenged and, thus, improve their quality of life. Consequently, this work has sought to advance the field of Bio-mechatronics, helping lead to a better understanding of human machine integration mechanisms that enhance human capabilities.

CHAPTER 02

THE PROBLEM

CHAPTER 02: THE PROBLEM

2.1 Description:

Presently there are prosthetic legs which act as a composite part. Thus amputee faces problems of adapting to different motion viz walking on ramp, running walking on rough road, climbing up& down slopes etc. They spent more metabolic energy in adapting this various condition by using same ankle joint of composite leg.

On the contrary use of detachable ankle joint in prosthetic leg can save some required energy for adaptation a particular ankle is made for a particular motion condition the attachment / disattachment of ankle joint would be easy & user friendly. The interlocking mechanism is employed for attachment and disattachment.

2.2 History:

Over the years, many kinds of leg prostheses have been devised in effort to replace the leg or legs that amputees have lost. All these leg prostheses have the difficult task of giving to these amputees a life as normal as possible. The complexity of human locomotion, however, is such that conventional leg prostheses have until now only been using passive mechanisms in the most sophisticated available devices. Conventional leg prostheses are very limited compared to a real human leg and some needs were thus not entirely fulfilled by them. According to amputees, specific conditions of use of conventional leg prostheses, such as repetitive movements and continuous loading, typically entail problems such as increases in metabolic energy expenditures, increases of socket pressure, limitations of locomotion speeds, discrepancies in the locomotion movements, disruptions of postural balance, disruptions of the pelvis-spinal column alignment, and increases in the use of postural clinical rehabilitation programs. Another problem is that during the amputees' locomotion, energy used for moving the prosthesis mainly originates from the amputees themselves because conventional leg prostheses do not have self-propulsion capabilities. This has consider able short and long-term negative side effects. Recent developments in the field of energy-saving prosthetic components have partially contributed to improve energy transfer between the amputees and their prosthesis. Nevertheless, the problem of energy expenditure is still not fully resolved and remains a major concern. A further problem is that the dynamic role played by the stump during the amputees' locomotion renders difficult the prolonged Wearing of conventional leg prostheses. This may create, among other things, skin problems such as folliculitis, contact dermatitis, oedema, cysts, skin shearing, scarring and ulcers. Although these skin problems may be partially alleviated by using a silicon sheath, a complete suction socket or powder, minimizing these skin problems remain a concern. Considering this background, it clearly appears that there was a need to develop improved leg prosthesis for above-knee amputees.

Mechanical replacements for legs such as the wooden peg legs attributed to pirate characters in popular fiction like Long John Silver in Treasure Island have been in use since ancient times. Greek and Roman history more than 3000 years old have records of mechanical devices made out of basic materials like wood and metal, and attached to the body with leather straps that served as artificial limbs in battlefields. These were mostly made out of iron with the main purpose of hiding the deformity of a missing limb. Ancient prosthetics could not be controlled by the amputee as modern flexible prosthetics.

Fig.2 Leg

During the 16th century a French military doctor Ambroise Paré, who specialized in amputation techniques invented a hinged mechanical hand and artificial legs that had advanced features locking knees and specially attached harnesses. Towards the end of the 17th century, Pieter Verduyn, a Dutch surgeon, developed an artificial lower limb with specialized hinges and a leather cuff to facilitate better attachment to the body. The contributions made by these two doctors are still the basic features of modern day artificial limbs.

Gaseous anesthesia in the 19th century enabled doctors to perform accurate amputation surgeries in order to prepare the limb stump to fit comfortably with the prosthetic limb. Surgical advancements and sterile, germ free surgeries guaranteed successful amputation procedures, which in turn increased the demand for artificial limbs.

The field of prosthetics vastly improved with suction-based attachment methods and joint technology in the 19th century. Notably, a prosthetic arm that could be controlled by connecting straps in the opposite shoulder was developed in 1812. The surge in the number of World War II veteran amputees saw the establishment of the Artificial Limb Program in 1945 by the American National Academy of Sciences. Throughout the world scientific progress has contributed to rendering artificial limbs more lifelike and functional in terms of materials, design and adaptability.

Latest Research in Artificial / Prosthetic Limbs

Constant research in the field of bio engineering has given improved models of artificial limbs that very nearly replicate the functions of real, biological limbs. Phenomenal advances have been made in measurement techniques, materials science and manufacturing methods to render more realistic prosthetic limbs. Heavy, natural materials like wood, metal, alloy and leather are replaced by lighter, synthetic materials such as polyester resins, epoxy, high density poly ethylene (HDPE), high density polypropylene (HDPP) and other carbon fiber composites that render the artificial limbs lighter and more durable.

From the times of the ancient Egyptians prostheses were created to function, for their appearance, and used as a psych-spiritual sense of wholeness. Amputation was feared more than death in some cultures and it was believed that you would be cursed not only while you were alive, but during the afterlife as well. (Thurston 1114). The earliest that prostheses was recorded was in the fifteenth century B.C. when a mummy had to amputate their toe and replaced it with a prosthetic made out of wood and leather.

Fig.3

The initial true rehabilitation aids were used in Greece and Rome for battle. Loss of limbs became recurrent due to soldiers being wounded in war. During this time, an innovative man named Ambroise Pare who served as a military surgeon for the French began to get a substantial amount of recognition for his creation of upper and lower artificial limbs. His "Le Petit Lorrain", a mechanical hand operated by catches and springs, was worn by a French army captain in battle (Thurston 1114). The prostheses were typically designed as heavy, crude devices made out of materials that were only available at that time such as, wood, leather, metal, etc. Typically, the prostheses would consist of a wooden peg leg, or hook in replacement of a hand. Also, with the lack of technology and medicine during these times, there was a lack in anesthesia for people who got injured. Those that were injured were required to endure the agony and pain and, more so death, in order to receive a prosthetic. Not only was pain tolerance an obstacle in finding treatment, but the amputee also needed to be able to afford the artificial limb. Unfortunately, only the wealthy were able to afford the costly prices of prosthetics, leaving the poor and middle class without limbs.

As well as Pare, other scholarly subjects such as, Alessandro Volta contributed to the advancements in bionic medicine. In about 1797 Volta came to realize that hearing could be restored by electrical stimulation. There was a patient who reported that they could hear a noise after undergoing neurosurgery in response to electrical stimulation of the auditory nerve. This then led to the the first implant for stimulation of the secular nerve. Due to Volta's discovery of electrical stimulation, other scientists named Dijourno and Eyries performed many experiments. They took a coil that connected an electrode to the inner ear to an electrode that connected to the temporal muscle. They then took a second coil and used it to transmit signals generated by a microphone. The subject within the experiment described a sensation of a background noise and could differentiate between changes in signals. With this discovery, led to further findings such as the cochlear implant. Once it was discovered that hearing could be restored due to an electrical stimulation, the creation of a multiple-channel cochlear implant was created. It bypasses the malfunctioning inner ear (cochlea) and provides information to the auditory centers in the brain through electrical stimulation of the hearing (auditory) nerves. (Clark)

Not only were there discoveries made about bionic hearing, but also bionic vision, cardiovascular applications, and much more. The advancements in medical bionic devices over the course of the years have strongly made an impact on the way people who have lost body parts, operate today. In the past, amputees who received a bionic leg would have to walk with a crutch because, although they would have a replaced limb, it was not able to function as a real leg. Now, there are creations in bionics that allow amputees to walk without a crutch and move freely. The discoveries that have been made in the past have led us to today's discoveries and allowed us to build off ideas and concepts that will further the advancements in bionic medicine.

CHAPTER 03

LITERATURE REVIEW

CHAPTER 03: LITERATURE REVIEW

[1]Bragaru M., Dekker R., Geertzen J., & Dijkstra P., ‘’Amputees and Sports’’, *Sports Medicine, 41(9),* 721-740.

Rehabilitating amputee is significant application of prosthetic limbs. This article discuss how loss of a limb through medical amputation, congenital defects, and accidents can decrease quality of life. The use of artificial limb allows patients to become active and repair their health is also explained.

[2]Camporesi S., Oscar Pistorius,’’Enhancement and post humans’’, *Journal of Medical Ethics, 34(9),* 639-639.

This article discusses on study conducted on Pistorius’s physical abilities. It refers various statistics on his running speed and energy consumption. In addition, it presents the morality involved with using prosthesis to enhance athletic performance.

[3]Chin T., Kuroda R., Akisue T., & Kurosaka M., ‘’Energy consumption during prosthetic walking and physical fitness in older hip distraction amputees’’, *The Journal of Rehabilitation Research and Development, 49(8),* 1255-1255.

This article presents a study of prosthetic limb applications among elderly patients. It investigates the exertion experienced by hip-disarticulation suffering unilateral amputees during walking. The study examined the oxygen consumption while walking at a comfortable speed.

[4]Clements I., ‘’the history of prosthetic limbs’’, www.science.howstuffworks.com/prosthetic-limb1 November 19, 2014.

This article includes a moderately detailed description of past prosthetic technologies citing many archeological examples. In addition, it uses example draw from ancient historical texts.

[5]Sawers A., ‘’Microprocessor controlled knees’’ *Journal of Rehabilitation Research & Development, 50(3),* 273-314.

This article examines use of micro controller controlled knees for individuals with transferor limb loss. It uses several criteria including metabolic energy expenditure, activity, gait mechanics, environmental obstacle negotiation, safety, and health and quality of life to evaluate the results of prosthetic knee use.

[6]Jette' M., Sidney K., & Blumchen G., ''Metabolic Equivalents (METS) in exercise testing, exercise prescription, and evaluation of functional capacity'', *Clinical Cardiology, 13*, 555-565.

This source explain metabolic equivalents (METS). It is a unit for measuring athletic activity through oxygen consumption. 1 MET =3.5 mL/kg/min of oxygen consumption.

CHAPTER 04

BIONIC LEG DESIGNING

CHAPTER 04: BIONIC LEG DESIGNING

4.1 Conceptual Details

1] Knee member:

The knee member of bionic leg is the top most part of bionic leg. It is the first part of bionic leg which is attached to the cup, which is directly fitted to the thigh of amputee (patient).This part has two pivot joints, which comprises of a shaft supported in two deep groove ball bearing .These bearings are placed in knee member. The first pivot joint connects the upper part of transtibial part, and allows its operation of turning about pivot, whereas the second pivot connects the upper end of actuator to knee member.

2] Transtibial member:

Fig.4 Transtibial Member

Transtibial member comprises of two parts, viz. upper part connected to pivot joint of knee member and the second part which is ankle part .This two parts of transtibial member are connected together by four aluminum rods ,to form a composite transtibial member. This composite transtibial member is pivoted to pivot joint provided in knee member. The lower part of transtibial member also known as ankle part , has a pivot joint ,which comprises of a shaft supported in ankle joint and its end screwed to fit nut. This pivot joint connects the lower end of actuator to ankle part.

3] Actuator:

Fig.5 Actuator

The actuator comprises of threaded sleeve and threaded nut. This screw pair with single degree of freedom is employed to actuate the leg assembly .A motor provides a turning torque to the threaded sleeve shaft via coupling and relative motion between sleeve shaft and nut causes nut part to move linearly and exert force required to turn the transtibial member about first pivot in knee member and in this way a motion is performed operatively between knee and transtibial member.

4.2 Human Form:

There are large amount of data pertaining to human forms. The bionic leg can be replace as substitute for missing leg for amputee(patient).The bionic leg will be large enough to avoid problems of miniaturization ideas of human control locomotion can be tested.

Weight Distribution:

Body Area	Percentage per part	No. of instances	Total mass percentage
Head,neck,Shoulder,Thorax	31%	1	31%
Arm	5%	2	10%
Pelvic	27%	1	27%
Thigh	10%	2	20%
Shin and Foot	6%	2	12%

Table: Human Mass Distribution

Since kinetic energy is direct function of velocity, the lower extremities have the greatest fluctuation in energy. It is clear that more mass at the feet means more change in kinetic energy. Loss in this energy is very costly

Link Length:

50% of data of male reveals that centre of gravity is above hip at approximately 38''.

4.3Human Motion:

Energetic of human walking:

For a comfortable pace, human body requires 320 Watts of power for a   kg of human weight. Facts reveal that about 25% of this total energy is only mechanical energy.it implies that 80 Watts of mechanical power is required for comfortable pace. This can be a somewhat depressing fact considering one of the electric motors used on the bionic leg can output 90 Watts continuously.

Potential energy (P.E) =m*g*L*cos (ϴ)………….. (a)

Kinetic energy (K.E) =m*g*L*(1-cos ϴ)………….. (b)

It can be seen from above two equations that when kinetic energy is maximum, potential energy is minimum and vice versa.

4.4 Controlling mechanism

Mode of operation

There are three modes of operations in the bionic leg for different applications such as running, walking etc.

Manually operated

In this Mode, movement of the leg is control by adjusting the probes setting for accessing the objects placing at different locations.

Motor operated

The pulse width modulation technique is used to controlling the base unit servo motors and fin gripper motor to function the bionic leg

Power supply

Power supply is used to supply power to motor through SMPS which is an electric convertor, then actuator gets actuated & perform linear motion.

We used Motor operated Actuator, as it has some advantages such as

- Less power losses
- Low power consumption
- Simple in design

4.5.1 Comparison of actuator:

Commercially available prosthesis are mainly made up of passive elastic members. This elastic members stores energy at the beginning of motion step and releases some amount of this stored energy at the end of the step. Average 75 kg human being produces approximately 26 Joules of energy at each stride, and releases 9 Joules of stored energy.

A subject, at a joint torque of 125 Nm produces 250 Watts of power. With servomotors, these would require a number of motors and gear boxes. Research in this field leads to concept of Series Elastic Actuation (SEA). SEA reduces the power requirement, without change in its torque. To reduce the size of motor, human centered actuators are developed. This concept is known as Series Parallel Elastic Actuation (SPEA). SPEEA reduces the torque requirement due to multiple elastic members connected to intermittent mechanisms.

4.5.2 Series Actuator Control and Modelling

Series-elastic actuation of several important benefits to dynamic robots, including high-bandwidth force control and improved safety. While this approach has become common among legged robots, the lack of commercial series-elastic actuators and the unique design requirements of these robots leaves custom-built actuators as the only option. These custom actuators are often designed for nominal behavior rather than an extended performance envelope, and thus lack the capacity for high dynamic behavior and large-scale disturbance rejection outside of controlled operating conditions. This paper details the design and construction of a compact series elastic actuator for use in the knee of a robotic biped built to emulate the dynamic performance of a half-scale human. The actuator specifications are denied by the combined performance envelope of scaled human knee extensor muscles. The design uses two DC motors geared in parallel for improved weight, compactness, and power density. The constructed actuator is designed for a no-load speed of 210 rpm and a nominal torque of 13 Nm. In addition, a unique approach to series-elastic actuator torque control is proposed which uses servo-controlled motor velocity to modulate spring deflection. Several key advantages of our proposed method over traditional torque-embedded control are demonstrated, including increased torque bandwidth, rejection of nonlinear from gear dynamics, and zero steady-state error. Our research group's goal is to develop high-performance legged systems using technology that transfers to rehabilitation robotics. It is our hope that our research will realize improved prostheses and orthotic for humans that greatly surpass the limited performance of presently-available assistive devices. This goal poses many challenges and constraints. First among these challenges is the need to develop devices capable of closely matching the dynamic performance of an unimpaired human. This challenge is twofold: we must understand human dynamics as well as reproduce them in simulation

and hardware. We begin by investigating simplified models for walking and running gaits such as the inverted pendulum and spring-loaded inverted pendulum. We also use our knowledge of the human musculoskeletal system and simplified muscle models to generate a revive neuromuscular model for walking. By taking inspiration and guidance from biology in this way, our goal is to reproduce human-like gaits that exhibit stable and adaptive control over uneven terrain and under large disturbances. With the goal of high-performance dynamic behavior comes a need for precise force feedback and control. Many traditional actuator technologies such as geared electric motors and hydraulics have a high impedance which can make robust force control a difficult problem. It has been shown that an elastic element in between a traditional
" actuator (such as a highly-geared electric motor) and its load can decouple the actuator inertia from the load inertia, allowing for more precise impedance control. If the characteristic force-length curve is known for the compliant element, then the force can be directly inferred by measuring its displacement. Thus, the series-elastic actuator (SEA) force can be controlled by modulating the detection of the compliant element.

The massive inertia of the robot arm will almost instantaneously transfer large quantities of energy to any object that it rigidly collides with. Com-pliant elements, even when completely passive, can improve the outcome of such a collision by decoupling the inertia from the site of impact. A properly-selected compliant element will absorb a large amount of the energy, and by spreading the impulse of the collision across a larger time, the maximum force exerted on the object will be reduced. Lower impact torques within the actuator allows the engineer to choose lighter, more compact drive train components such as gear

4.5.3 Actuator Design

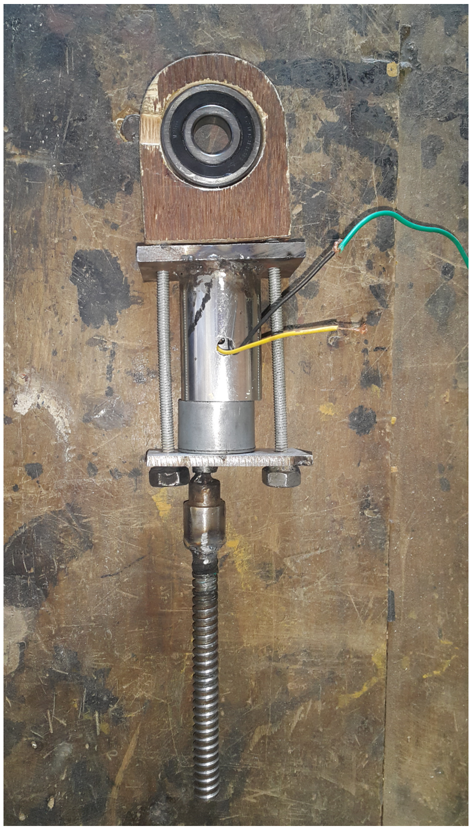


Fig.6 actuator

Fig.7 Actuator

Advantages of series actuator

- It produced high speed.
- Components are readily available
- It gives more power output for same size of motor.

Advantages of parallel actuator

- It produces high torque
- Simple in construction

4.5.4 Motor Power Efficiency

For an iron less core, Dc motors of relative small size, the relationship that govern the behavior of motor in various circumstances can be derived from physical laws and characteristics of the motors themselves.

Kirchhoff’s Voltage rule states that ‘’the sum of potential increases in a circuit loop must equal the sum of potential decreases.’’

The back emf generated by motor Vo = (I*R) + Ve

Where, Vo = Power supply (volts)

I = Current (amperes)

R = resistance in terminal (ohms)

Ve =back emf (volts)

The heat loss (power loss) in armature is given by,

P = (I^2)* R

Chapter 5: Finite Element Analysis & Testing

5.1 Manual Design calculations:

Calculations:

1] Knee member:

Mass of an average human=100kg (approximately)

Force due to weight of human being=100*g =1KN (approximately)

Where=acceleration due to gravity=9.81 m/s^2

Let, factor of safety be 1.5

Therefore, Design force

[F max] =150kgf

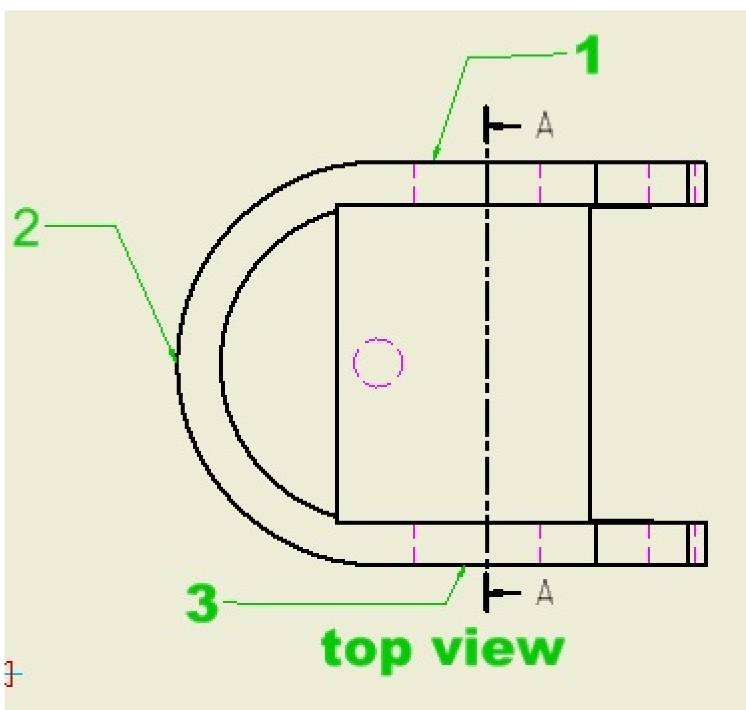


Fig.8 Knee member

On an average, knee part of human measures around 43cm in circumferential perimeter, whose radius is calculated as

$2 * \pi * r = 430$ mm,

Where, $\pi = 3.147$ and

r=radius of knee part

Therefore=68.43mm=70mm (approximately)

Now, 1-2-3 is semicircular section as viewed from top with radius=70mm

Thickness of part as viewed from top=Width of bearing selected margin……………. (1)

For bearing selection, let us first select the rod diameter.

For rotating inner ring loads and for normal and heavy loads,

[7]Shaft diameter should be less than or equal to 18mm

Let, shaft diameter=17mm

Tolerance=j5

[7]Combination of hole and shaft H6j5

Shaft diameter=17+ (5*0.001)

Shaft diameter=$17^{+0.005}$ mm

Let, Fr=radial force on each bearing on knee part

Fr=150/2=75kgf

Assuming axial load to be negligible, i.e.

Fa=0

Equivalent load on bearing, i.e.

[7]Pe =[xvFr+yFa]*s*Kt

=[1*1*75+0]*1.4*1.2

Pe =126kgf

Let, life of bearing in hours, i.e.

Lhrs=87600 hrs (10 years)

Let, rotation of bearings say N=10rpm

Therefore, life in millions of revolutions, i.e

Lmr= (Lhrs*60*N)/ (10^6)

Lmr=52.56mr

Assume 90% probability of survival

Therefore, L10=52.56mr

Now, [7] L10=(C/Pe)^k

52.56=(C/126) ^3

Therefore, C=471.97kgf

Selecting standard bearing

Outer diameter of bearing	ISI No.	Bearing of basic design No.(SKF)	Inner diameter of bearing	Width of bearing	Static load carrying capacity	Dynamic load carrying capacity
D(mm)	--	--	d (mm)	B(mm)	C0(kgf)	C(kgf)
47	17BC03	6303	17	14	630	1060

Now, from equation (1)

Thickness of knee part=14+5=19mm

Thickness of knee part =20mm (approximately)

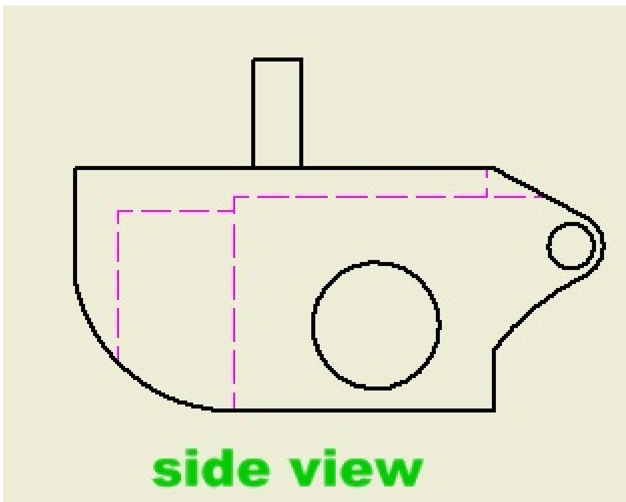


Fig.9 Side view of knee Member

Length of knee part in side view=90+ (47/2) +10

Length of knee part in side view=124mm

By Scaling factor, height of knee part in side view=116mm (approximately)

Length of shaft = (70+20)*2+ (length of threaded part for lock nut)*2…………………………… (2)

[7]For shaft of diameter 17mm, we have M16 (coarse series thread)

For M16 thread,

[7]Nut thickness=10.18mm

Nut width =24mm

From equation (2)

Length of shaft= (70+20)*2+ (10.8)*2+2mm (margin)

Length of shaft=204mm (approximately)

2] Transtibial member:

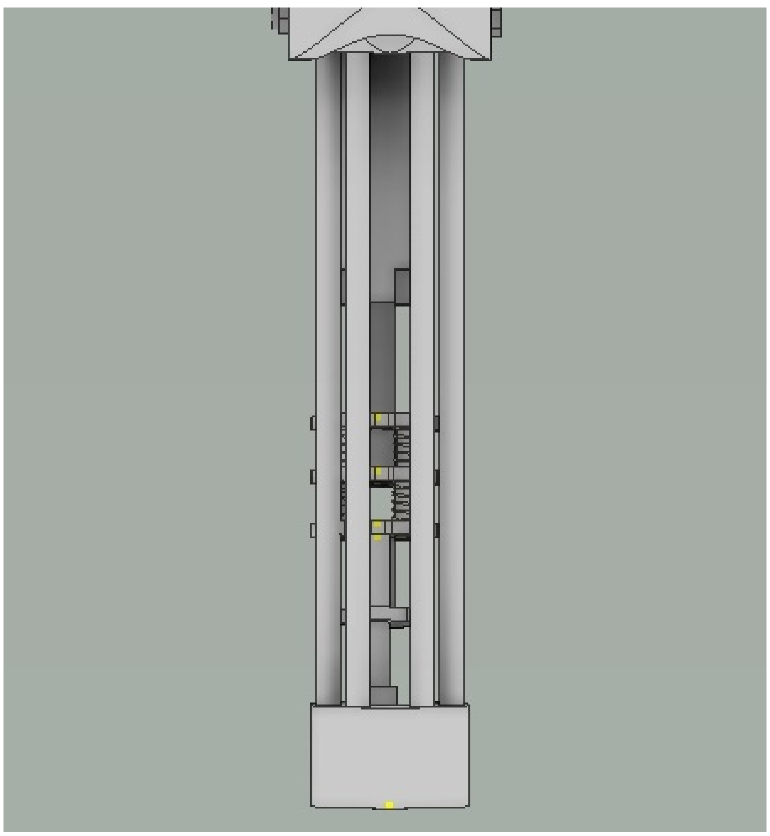


Fig.10 Ttranstibial member

Outer diameter of upper part of transtibial member=inner diameter of knee member

Therefore, outer diameter of upper part of transtibial member=70mm

Refer figure no.top view

From geometry of figure,

2x+y=70mm

Let, x=30mm (assumption)

y=10mm

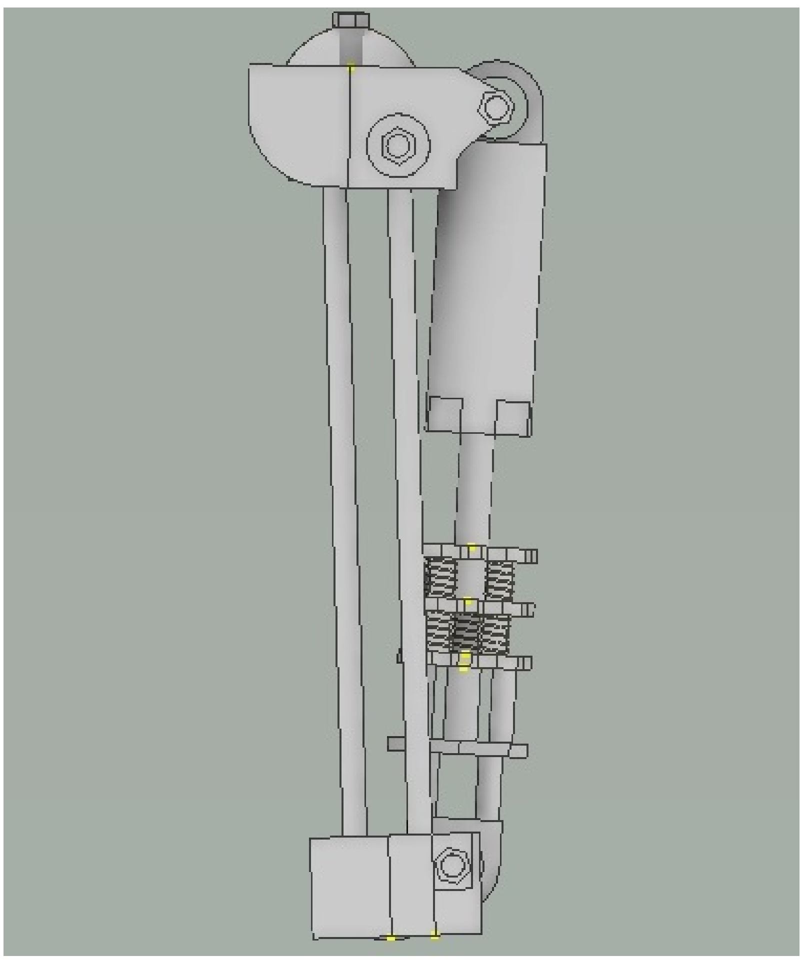


Fig.11 side view

Refer figure no. side view

From geometry of figure

Height of upper part of transtibial member=a+b ………………………………….. (3)

Now, a= (nut width)/2+5mm (margin)

a=24/2+5

a=17mm=20mm (approximately)

Also=a+10mm (margin)

b=20+10

b=30mm

From equation (3)

Height of upper part of transtibial member=20+30=50mm

3] Rod connecting upper part and lower part of transtibial member:

Material: Aluminum

Number of rods: 04 nos.

Diameter of rod=16mm

Length of rod=150mm

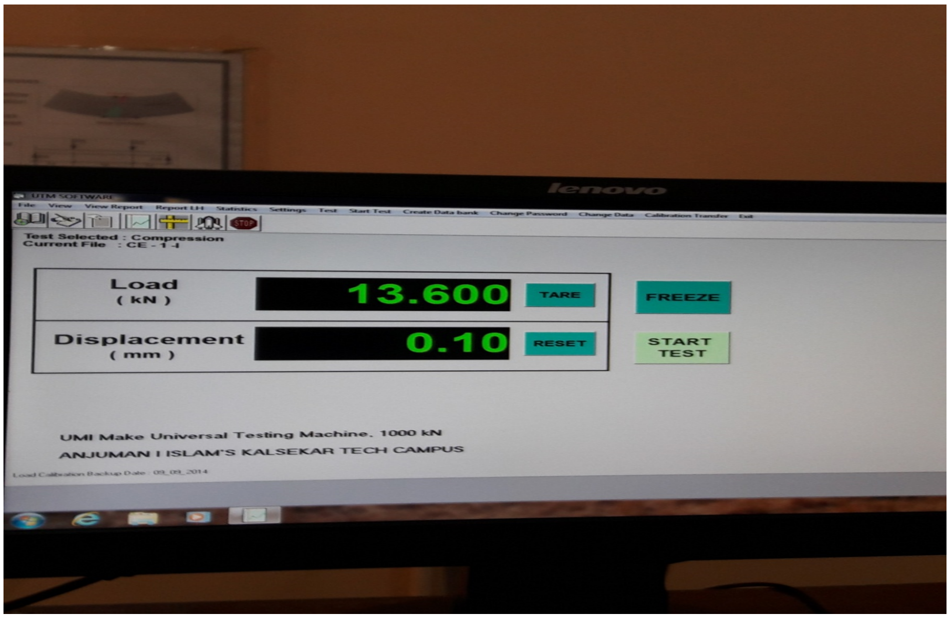
This rods are subjected to direct compressive load, due weight of human.

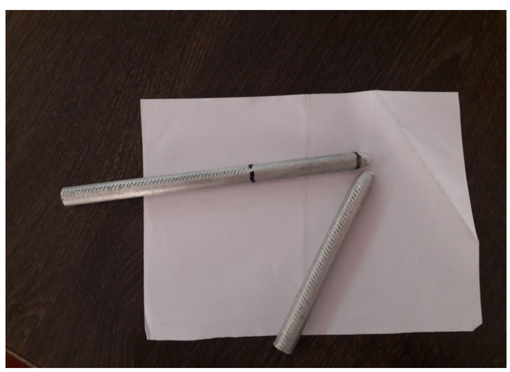
Tensile and compressive test were conducted on Aluminum rod of 12mm diameter and gauge length of 60mm with the help of Universal Testing Machine (UTM).

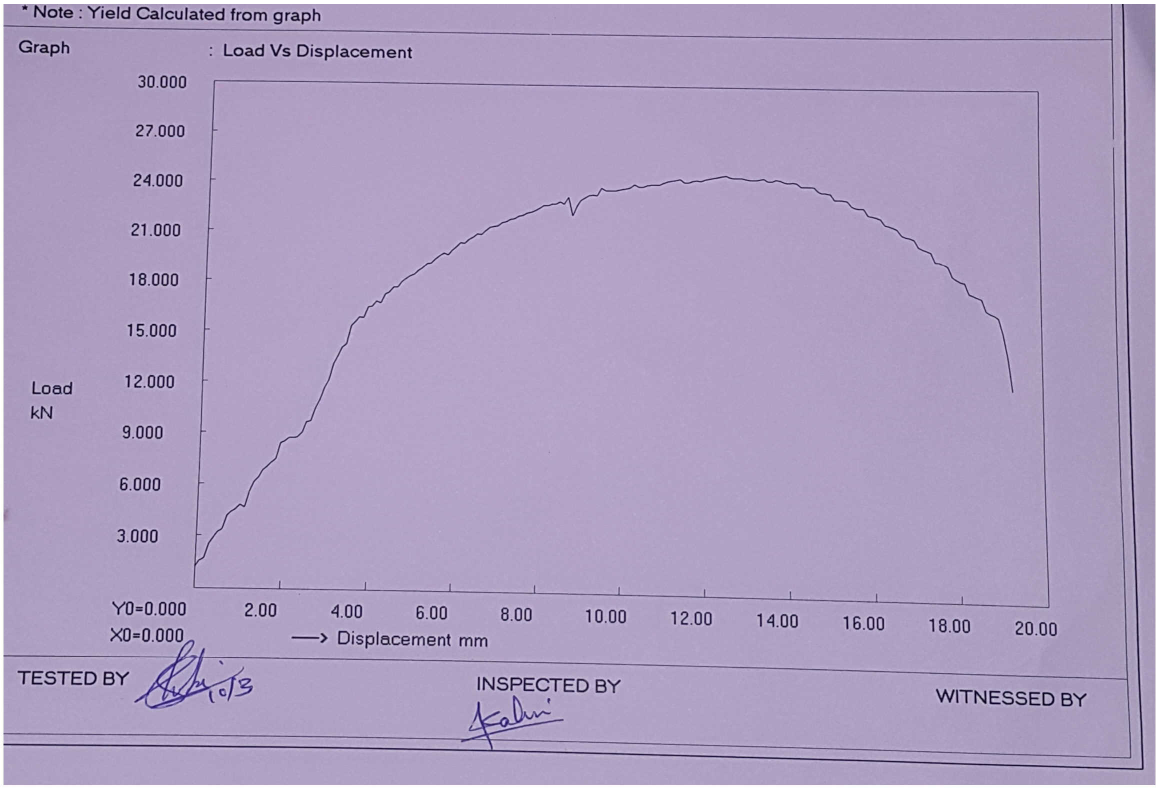
Maximum force (Fmax) =24.65KN…………………….. (Testing certificate on UTM)

The UTM readings indicate that rods selected can be safely employed as design load [Fmax] =1KN

Testing on UTM Machine :







Bar after tensile failure on UTM

Fig.12 load vs. displacement graph

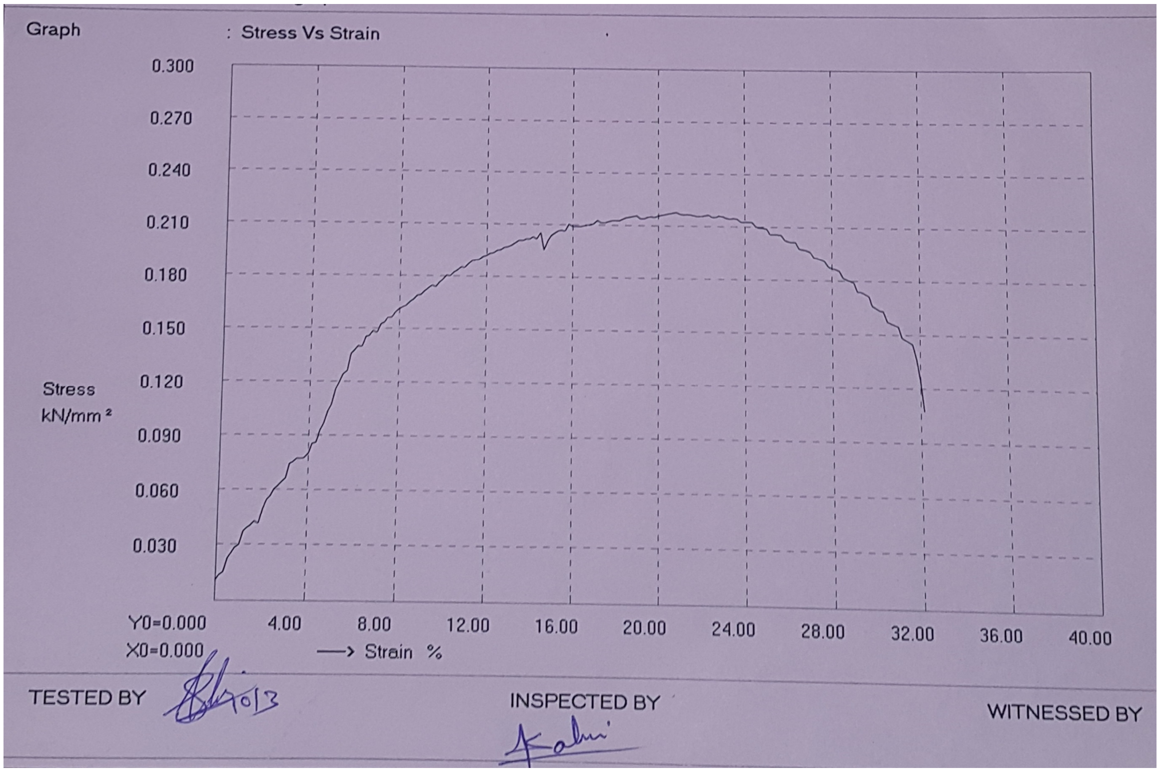


Fig.13 stress vs. strain graph

4] Lower part of transtibial member:

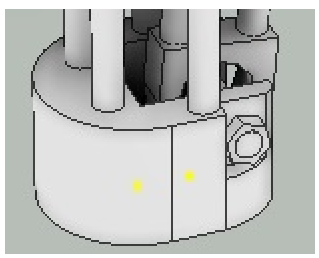


Fig.14 ankle part

On an average, ankle part of human being measures 220mm circumferentially in perimeter

Therefore, 2*pie*R=220mm

Where're=radius of ankle joint

R=35mm

Thickness of part as viewed from top=Width of bearing selected margin……………. (1)

For bearing selection, let us first select the rod diameter.

For rotating inner ring loads and for normal and heavy loads,

[7]Shaft diameter should be less than or equal to 18mm

Let, shaft diameter=17mm

Tolerance=j5

[7]Combination of hole and shaft H6j5

Shaft diameter=17+ (5*0.001)

Shaft diameter=$17^{+0.005}$ mm

Let, Fr=radial force on each bearing on knee part

Fr=150/2=75kgf

Assuming axial load to be negligible, i.e.

Fa=0

Equivalent load on bearing, i.e.

[7]Pe= [xvFr+yFa]*s*Kt

= [1*1*75+0]*1.4*1.2

Pe=126kgf

Let, life of bearing in hours, i.e.

Lhrs=87600 hrs. (10 years)

Let, rotation of bearings say N=10rpm

Therefore, life in millions of revolutions, i.e.

$Lmr= (Lhrs*60*N)/ (10^6)$

$Lmr=52.56mr$

Assume 90% probability of survival

Threfore, L10=52.56mr

Now, [7] $L10=(C/\boxed{Pe})$ ^k

$52.56=(C/126)$ ^3

Therefore=471.97kgf

Selecting standard bearing

Outer diameter of bearing	ISI No.	Bearing of basic design No.(SKF)	Inner diameter of bearing	Width of bearing	Static load carrying capacity	Dynamic load carrying capacity
D(mm)	--	--	d (mm)	B(mm)	C0(kgf)	C(kgf)
47	17BC03	6303	17	14	630	1060

Now, from equation (1)

Thickness of knee part=14+5=19mm

Thickness of knee part =20mm (approximately)

5.2 Analysis of parts with different materials:

Finite Element Analysis (FEA) approach, we have followed for analysis

Structural Analysis (ANSYS)

Structural analysis is the determination of the effects of load on physical structure and their components.

Structures subject to this type of analysis include all that must withstand loads, such as buildings, bridges, vehicles, machinery, furniture, attire, prostheses

Introduction

After finalizing our design on SOLIDWORKS and paper work, we have to analyze our design, for that we have to use a software. Mathematically or on papers it is very difficult to find out different types of forces, deflection, failure, etc. on our designed wheelchair.

There are number of softwares available for analysis of design such like AUTODESK Inventor, ANSYS, etc. But the ANSYS was the software which was the suitable for us to use it also it gives the better results than AUTODESK Inventor.

The purpose for using ANSYS, the software used is:

1. To find static structural analysis.
2. To find various forces acting on each member and to study the failure of frame.
3. To study von-Misses Stresses.
4. To find the total deformation.
5. To evaluate elastic strength intensity.
6. To find the strain energy.

For completing the analysis on the specified software, we have to perform the following steps stated below as our model design is made by using a software as SOLIDWORKS.

Analysis of leg is carried out by following step

1. Geometry cleanup
2. Selection of material and applying material properties in analysis
3. Importing '.igs' file format to ANSYS
4. Definition of contacts and mesh formation
5. Defining forces and supports
6. Solution for given system using solver
7. Plotting the results

To perform the analysis on the given model in ANSYS, first of all we have to import our model into ANSYS, whereas in ANSYS, it requires '.igs' file format. But in SOLIDWORKS we have our model in 'sld.prt' format which we have to convert in '.igs' file format.

1. **Engineering data**

In this, we need to define the following input data such as:

1. Poisson ratio

2. Working temperature

3. Young modulus

4. Bulk modulus

5. Shear modulus

6. Density

Strength Coefficient Pa	Strength Exponent	Ductility Coefficient	Ductility Exponent	Cyclic Strength Coefficient Pa	Cyclic Strain Hardening Exponent
9.2e+008	-0.106	0.213	-0.47	1.e+009	0.2

Temperature C	Young's Modulus Pa	Poisson's Ratio	Bulk Modulus Pa	Shear Modulus Pa
	2.e+011	0.3	1.6667e+011	7.6923e+010

Material Property Table

Tensile Yield Strength Pa	Tensile Ultimate Strength Pa	Reference Temperature C
2.5e+008	4.6e+008	22

Structural Steel > Constants

Density	7850 kg m^-3
Coefficient of Thermal Expansion	1.2e-005 C^-1
Specific Heat	434 J kg^-1 C^-1
Thermal Conductivity	60.5 W m^-1 C^-1
Resistivity	1.7e-007 ohm m

2. Geometry

Geometry clean-up is a task to clarify all the connections and joints of the model given. If we do not go for geometry clean-up, the software used ANSYS will not accept the model made as it will not be able to mesh it.

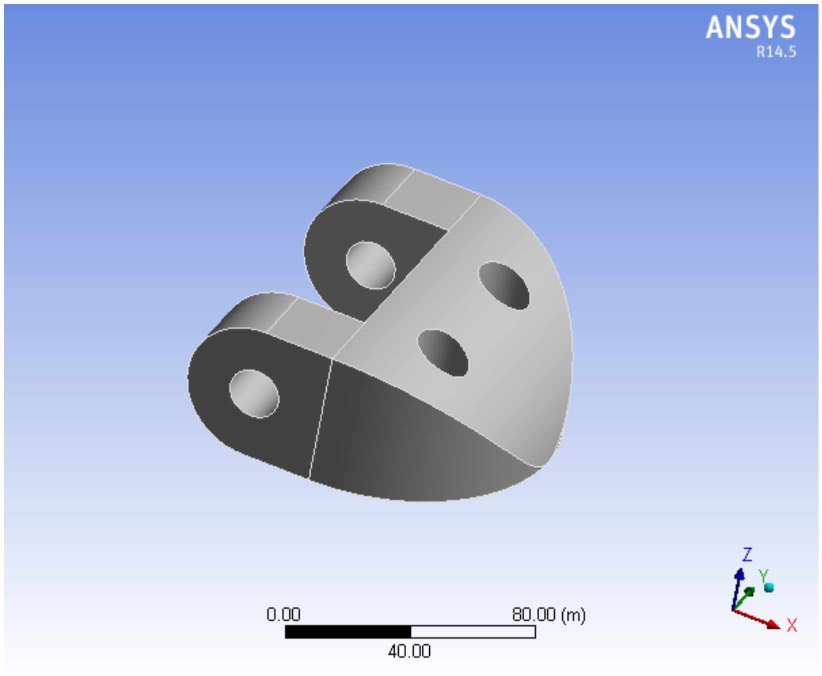


Fig.15 Imported Geometry of knee

When we were importing our SOLIDWORKS.igs file to our analysis software ANSYS and were trying to mesh it, it was not accepted by ANSYS. Because of improper geometry connections software was unable to mesh our model, this was due to undefined contact types. For that we clean up the geometry and defined the every joint so that ANSYS can read it. But the geometry cleaning process we did in SOLIDWORKS.

Model (A4) > Geometry

Object Name	*Geometry*
State	Under defined
Definition	
Source	C:\Users\MAMA\Downloads\bionic leg.igs
Type	Iges
Length Unit	Meters
Element Control	Program Controlled
Display Style	Body Color
Bounding Box	
Length X	0.3773 m
Length Y	0.20799 m
Length Z	0.71931 m

Properties	
Volume	1.5864e-003 m³
Mass	
Scale Factor Value	1.
Statistics	
Bodies	76
Active Bodies	52
Nodes	173960
Elements	85780
Mesh Metric	None
Basic Geometry Options	
Solid Bodies	Yes
Surface Bodies	Yes
Line Bodies	No
Parameters	Yes
Parameter Key	DS
Attributes	No
Named Selections	No
Material Properties	No
Advanced Geometry Options	
Use Associativity	Yes
Coordinate Systems	No
Reader Mode Saves Updated File	No
Use Instances	Yes
Smart CAD Update	No
Attach File Via Temp File	Yes
Temporary Directory	C:\Users\MAMA\AppData\Local\Temp
Analysis Type	3-D
Mixed Import Resolution	None
Decompose Disjoint Geometry	Yes
Enclosure and Symmetry Processing	Yes

3. Model Setup

- **Mesh Generation**
- **Contact**
- **Fixed Support**
- **Forces**

Mesh generation:

Meshing is probably the most important part in any of the computer simulations, because it can show drastic changes in results we get. Meshing means creating a closed geometry of some grid-points called ‘nodes’. The results are calculated by solving the relevant governing equations numerically at each of the nodes of the mesh. The governing equations are almost always a partial differential equations and finite element method (FEA) is used to find solutions to such equations. The pattern and positioning of nodes also affects the solution, good meshing is very essential for a computer simulation to give better results.

In our model we also did the meshing after importing and identifying the connections in ANSYS software. There three types of meshing are available with this software as fine meshing, medium meshing and coarse meshing. We did the coarse meshing as we were doing a student level project and also as the fine meshing takes more time to give us the results.

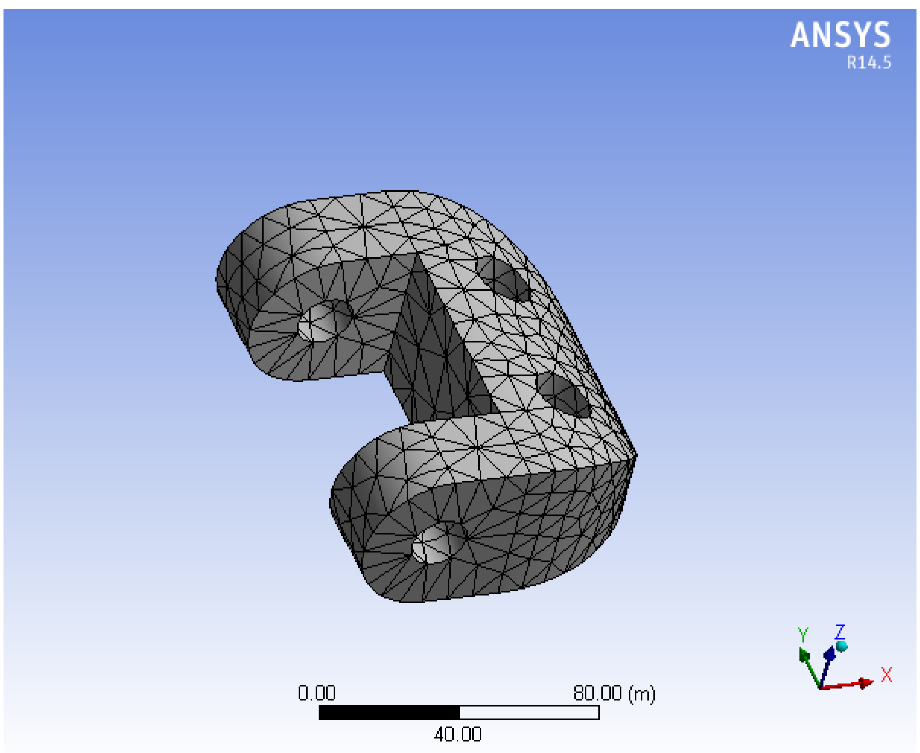


Fig.16 Mesh Geometry

The geometry of mesh can be of any shape as triangular meshing or quadrilateral meshing, etc. but here we did the triangular meshing as it was program controlled and we did not changed it. We can see that in table it is shown as program controlled

Object Name	*Mesh*
State	Solved
Defaults	
Physics Preference	Mechanical
Relevance	0
Sizing	
Use Advanced Size Function	Off
Relevance Center	Coarse
Element Size	Default
Initial Size Seed	Active Assembly
Smoothing	Medium
Transition	Fast
Span Angle Center	Coarse
Minimum Edge Length	3.750 m
Patch Conforming Options	
Triangle Surface Mesher	Program Controlled
Advanced	
Shape Checking	Standard Mechanical
Element Midside Nodes	Program Controlled
Straight Sided Elements	No
Number of Retries	Default (4)
Extra Retries For Assembly	Yes
Rigid Body Behavior	Dimensionally Reduced
Mesh Morphing	Disabled
Statistics	
Nodes	4468
Elements	2482
Mesh Metric	None

Fixed support & Forces:

These are the actual reasons for the failure of any system. There are different types of loads acting on the entire system which are as follows:

1. Inertial loads:

   These loads act on the entire system where density is required for mass calculations and these are only loads which act on defined point masses.

2. Structural loads:

   Forces or moments acting on parts of system.

3. Structural supports:

   Constraints that prevent movements on certain regions.

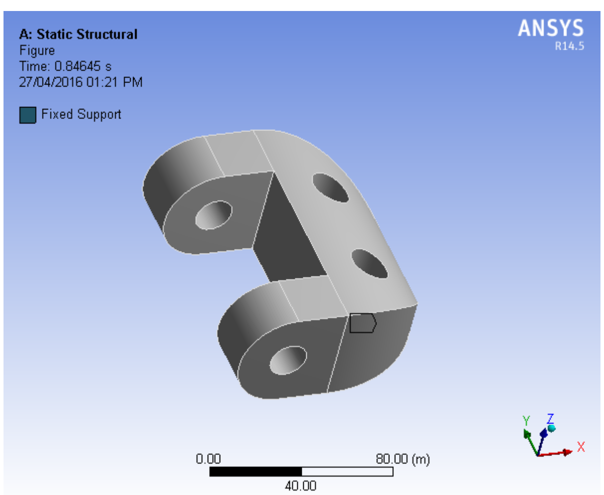
4. Thermal loads:

The thermal loads which result in a temperature field causing thermal expansion or contraction in the model.

The Forces applied on geometry are tabulated in Table given below

Static Structural (A5) > Loads

Object Name	*Fixed Support*	*Force*	*Force 2*
State	Fully Defined		
Scope			
Scoping Method	Geometry Selection		
Geometry	1 Face		
Definition			
Type	Fixed Support	Force	
Suppressed	No		
Define By		Vector	
Magnitude		Tabular Data	500. N (ramped)
Direction		Defined	



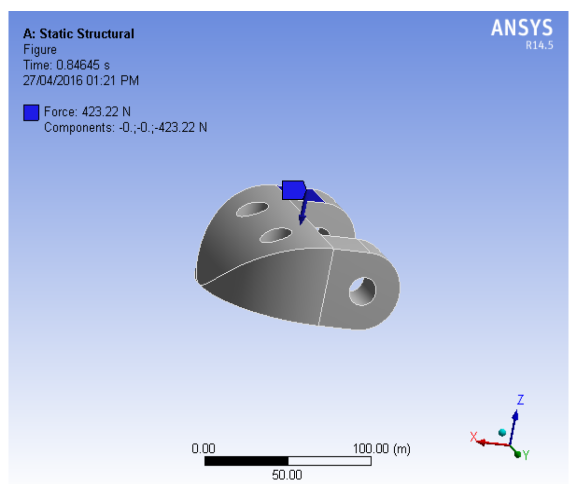


Fig.17 Fixed support 1

Fig.18 Fixed support 2

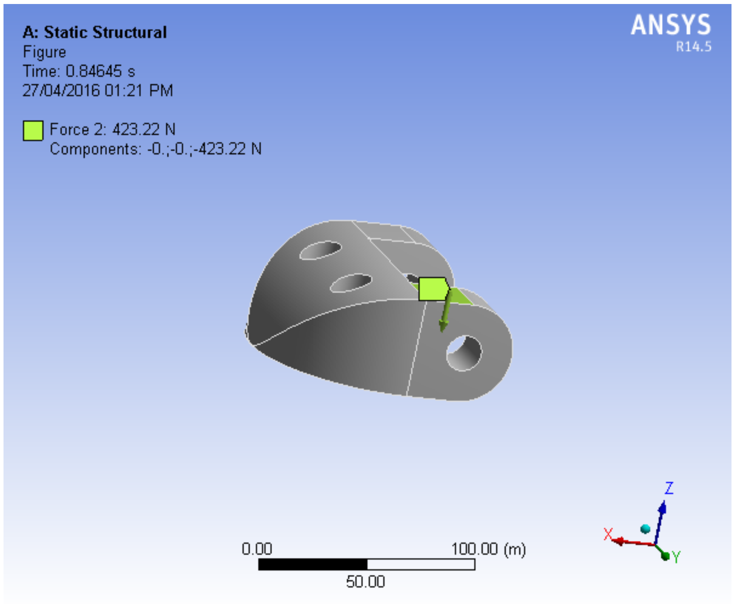


Fig. 19 Force Applied on member

4. Solution information

For solution we have selected results of analysis to display the different stresses, deformations, graphs (linear or non-linear) etc. But we have focused on these:

1. Von-Misses Stresses
2. Total deformation
3. Elastic strength intensity.
4. Strain energy

These are shown in figures given below (Fig.):

Object Name	*Total Deformation*	*Equivalent Stress*	*Strain Energy*	*Equivalent Elastic Strain*
State	Solved			
Scope				
Scoping Method	Geometry Selection			
Geometry	All Bodies			
Definition				
Type	Total Deformation	Equivalent (von-Mises) Stress	Strain Energy	Equivalent Elastic Strain
By	Time			
Display Time	Last			
Calculate Time History	Yes			
Identifier				
Suppressed	No			
Results				
Minimum	0. m	5.641e-003 Pa	2.3753e-016 J	3.9063e-014 m/m
Maximum	2.5708e-010 m	2.6815 Pa	6.58e-010 J	1.7412e-011 m/m
Information				
Time	1. s			
Load Step	1			
Substep	1			
Iteration Number	1			
Integration Point Results				
Display Option		Averaged		Averaged

Fig.20 Summary of Results

1. **Von-Misses Stresses**

The von Mises stresses were used for this study because these stress values allow the most complicated stress situation to be represented by a single quantity In essence, the von Mises stress values are a combination of all of the stress components present in the model into one value. This is a good measure of the overall reaction of the knee member to the loading condition, because it takes all of the stress components and outputs one stress value. This value was compared to the Yield Strength to ensure that the material did not exceed the stress limit presented by the polypropylene copolymer material**.**

The contour plot for the von Mises stresses will be examined for each of the models. These contour plots were used to display the overall distribution of the von Mises stresses throughout the material, as well as determine the approximate location and value of the maximum von Mises stresses. These contour plots were examined and presented in the results section of this report.

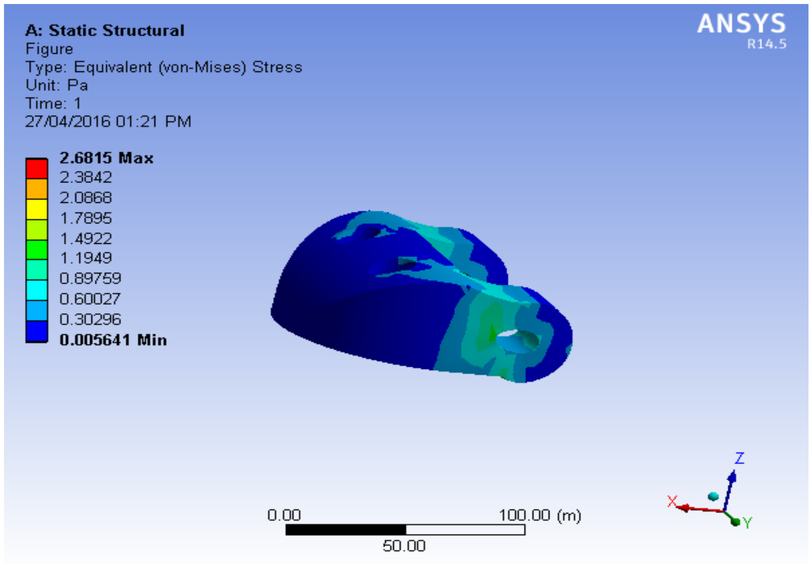


Fig.21 Von-misses stress distribution

2. Total deformation

The next area for study is in the deformation of each model due to the applied loading conditions. The deformations were investigated to determine the overall effect of the parameters and their reaction to the loading conditions in the varying directions. The deformations were examined in component form to determine how the model has deformed from its original shape. These component values were used to find the resultant displacement in each model. These were graphed and analyzed to gain a better understanding of how the combined loading affected the models.

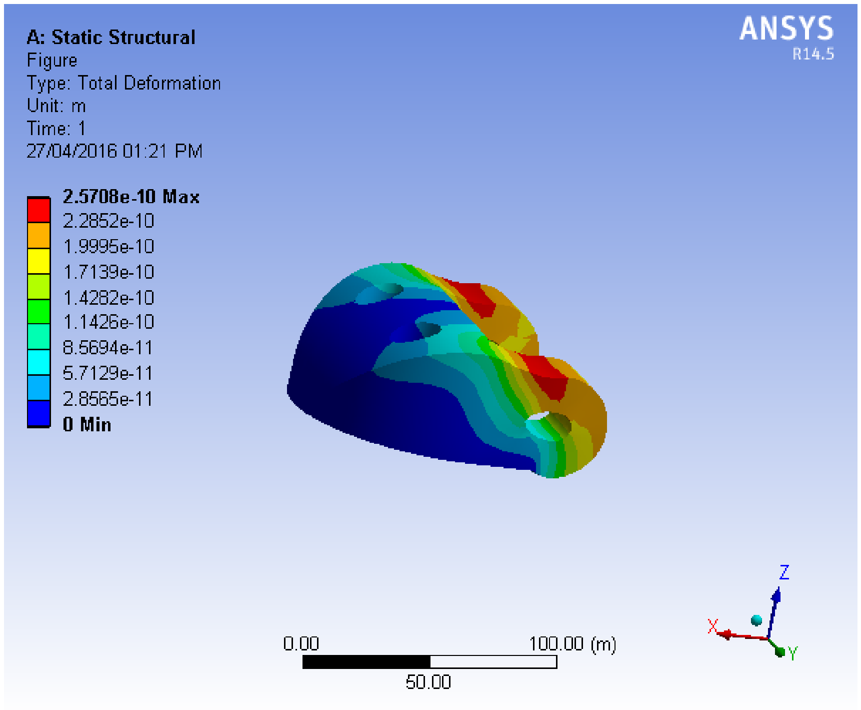


Fig.22 Total Deformation

3. Elastic strength intensity.

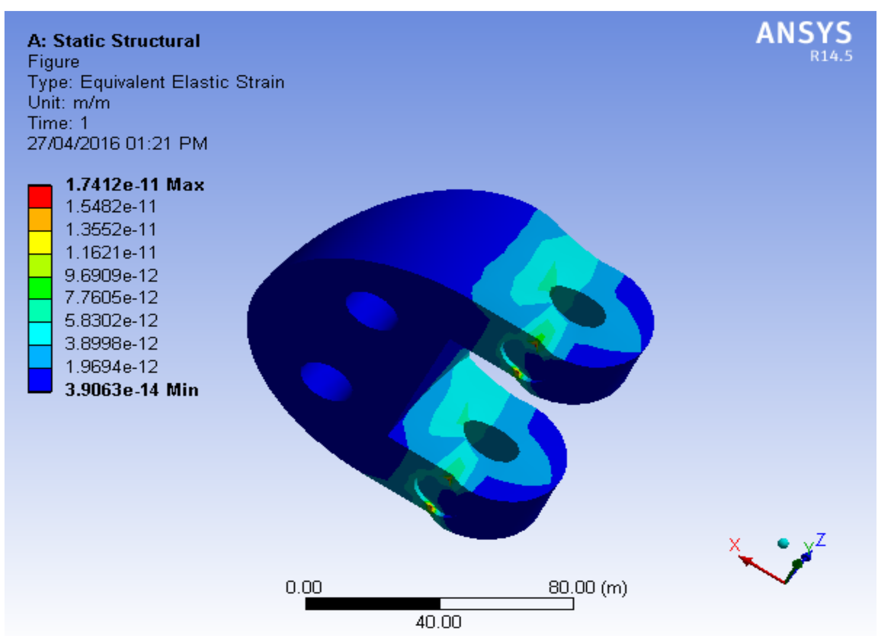


Fig.23 Elastic Strength

4. Strain energy:

It has been observed that a solid under hydro static, external pressure, can be withstand very large stresses. When there is energy of distortion or shear to be stored, as in the tensile stress, the stresses that may be imposed are limited.

The results for the strain energy were observed and plotted on graph so that the maximum and minimum strain energy can be plotted on individual as well as total assembly.

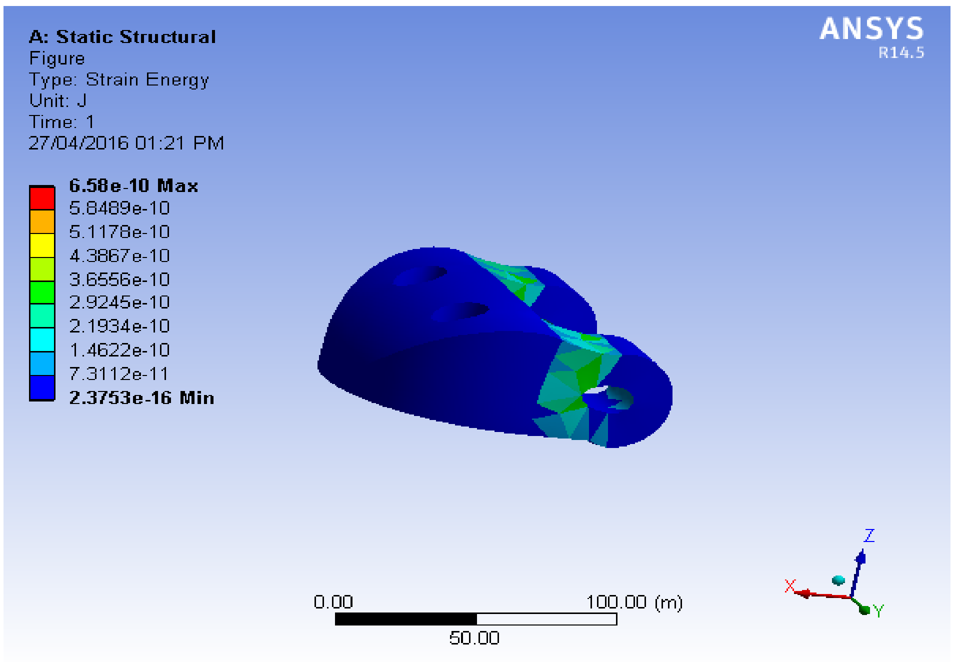
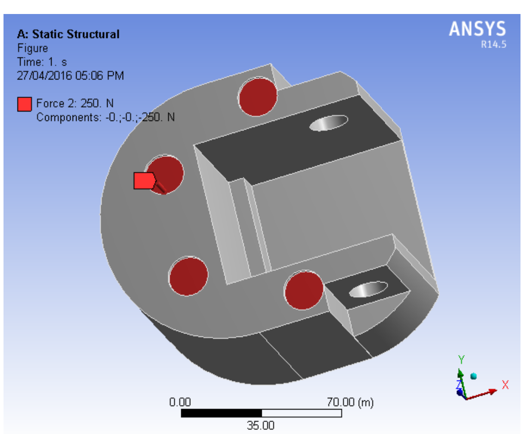
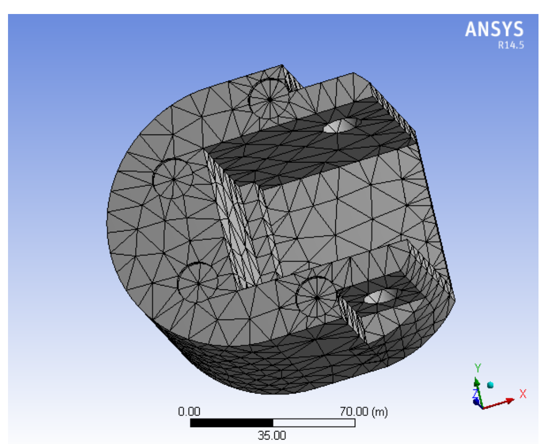
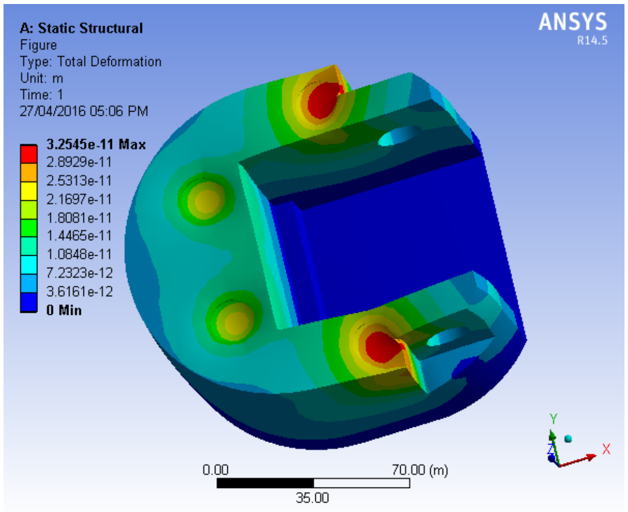


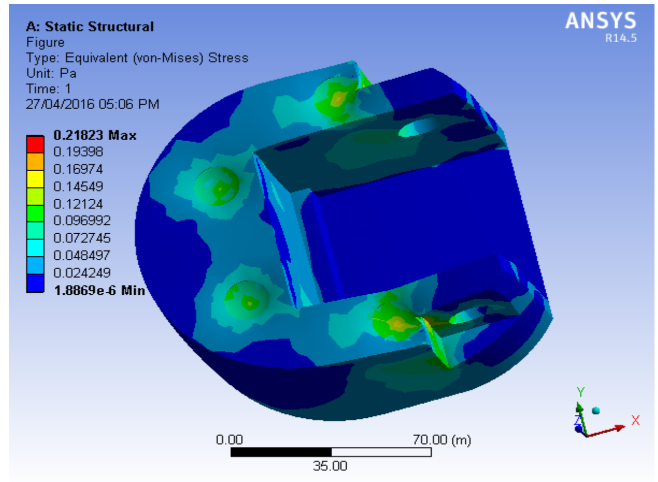
Fig.24 Strain Energy

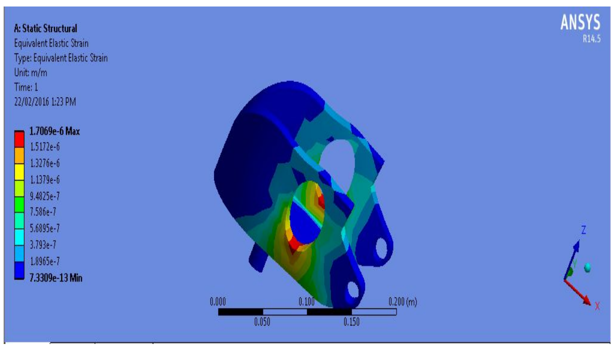
This way analysis has been performed on individual parts as well as whole assembly and results has been plotted as given below

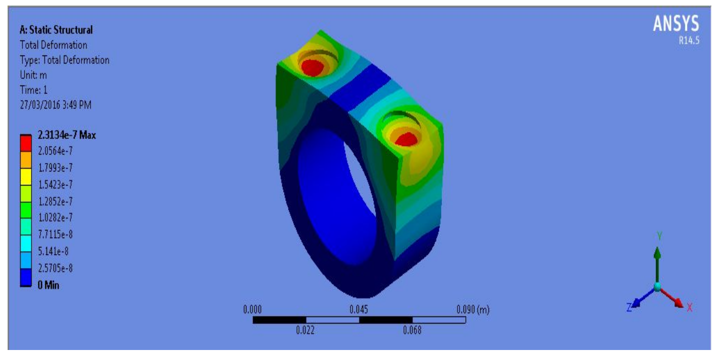


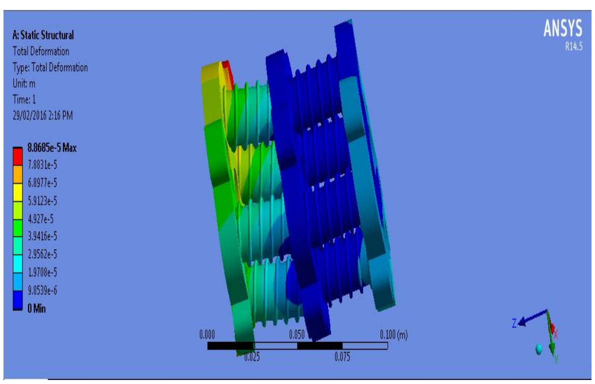


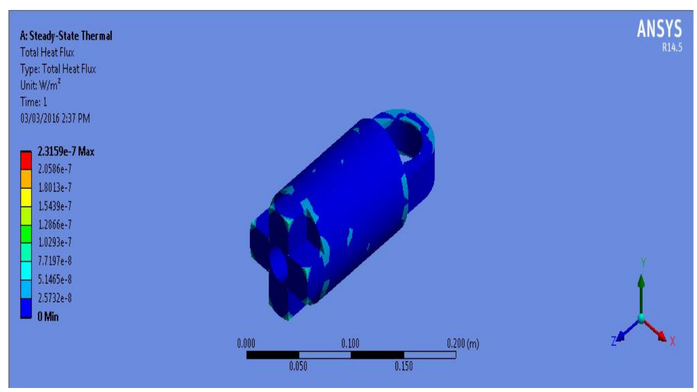

A: Static Structural
Figure
Type: Total Deformation
Unit: m
Time: 1
27/04/2016 05:06 PM
3.2545e-11 Max
2.8929e-11
2.5313e-11
2.1697e-11
1.8081e-11
1.4465e-11
1.0848e-11
7.2323e-12
3.6161e-12
0 Min
ANSYS
R14.5
0.00
70.00 (m)
35.00

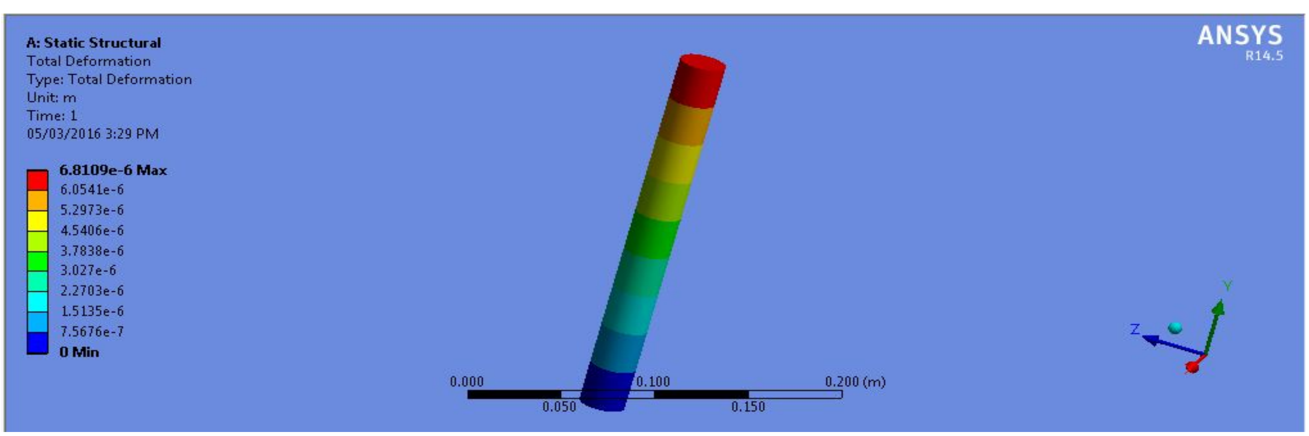

A: Static Structural
Figure
Type: Equivalent (von-Mises) Stress
Unit: Pa
Time: 1
27/04/2016 05:06 PM
0.21823 Max
0.19398
0.16974
0.14549
0.12124
0.096992
0.072745
0.048497
0.024249
1.8869e-6 Min
ANSYS
R14.5
0.00
70.00 (m)
35.00


A: Static Structural
Equivalent Elastic Strain
Type: Equivalent Elastic Strain
Unit: m/m
Time: 1
ANSYS
R14.5
1.7069e-6 Max
7.3309e-13 Min


A: Static Structural
Total Deformation
Type: Total Deformation
Unit: m
Time: 1
ANSYS
R14.5
2.3134e-7 Max
0 Min


A: Static Structural
Total Deformation
Type: Total Deformation
Unit: m
Time: 1
ANSYS
R14.5
8.8685e-5 Max
0 Min


A: Steady-State Thermal
Total Heat Flux
Type: Total Heat Flux
Time: 1
ANSYS
R14.5
2.3159e-7 Max
0 Min


A: Static Structural
Total Deformation
Type: Total Deformation
Unit: m
Time: 1
05/03/2016 3:29 PM
6.8109e-6 Max
6.0541e-6
5.2973e-6
4.5406e-6
3.7838e-6
3.027e-6
2.2703e-6
1.5135e-6
7.5676e-7
0 Min
ANSYS
R14.5
0.000
0.100
0.200 (m)
0.050
0.150

CHAPTER 06

Manufacturing and Cost Analysis

CHAPTER 6: Manufacturing and Cost Analysis

6.1 Cost Estimation:

Introduction

Cost estimation may be defined as the process of forecasting the expenses that must be incurred to manufacture a product. These expenses take into a consideration of all expenditure involved in a design and manufacturing with all related services facilities such as pattern making, tool, making as well as a portion of the general administrative and selling costs.

Purpose of Cost Estimating:

1. To determine the selling price of a product for quotation or contract so as to ensure a reasonable profit to the company.
2. Check the quotation by the vendors.
3. Determine the most economical process or material to manufacture the product.
4. To determine standards of production performances that may be used to control the cost.

Types of Cost Estimation:

1. Material cost

2. Machining cost

Material Cost estimation:

Material cost estimation gives the total amount required to collect the raw material which has to be processed or fabricated to desired size and functioning of the components.

These materials are divided into two categories.

1. Materials for fabrication:

In this the material is obtained in raw condition and is manufactured or processed to finished size for proper functioning of component.

2. Standard purchased parts:

This includes the part which was readily available in market like Allen screws etc. list in for chard by estimation stating the quality, size and standard part and weight of raw materials and the cost per kg for fabricated parts.

Machining Cost estimation:

This cost estimation is an attempt to forecast the total expenses that may include to manufacturer apart from material cost.

Sir no.	Components	Quantity	Cost
1.	Timber wood		4000
2.	Aluminum rod	6	500
3.	Bearing	4	400
4.	Actuator a. C clamp b. Johnson motor c. Dc Battery(12v)	1 1 1 1	 500 800
5.	Shaft	3	150
6.	Bolt and nut	8	100

6.2 Fabrication:

Fabrication is the act of making something from raw materials. Fabrication is carried out in following ways:

WOOD → It is the best suitable material for the prototype.

A) Fabrication of various parts by Timber wood is as follows:

1). Knee part:

The knee part is made up of timber wood as a raw material by carrying out various machining processes, it consist of two bearings attached in part ,opposite to each other .

Fig.15 knee part

2). Ankle part:

The ankle part is also made up of timber wood as a raw material by carrying out various machining processes,

3).Trans-tibial member :

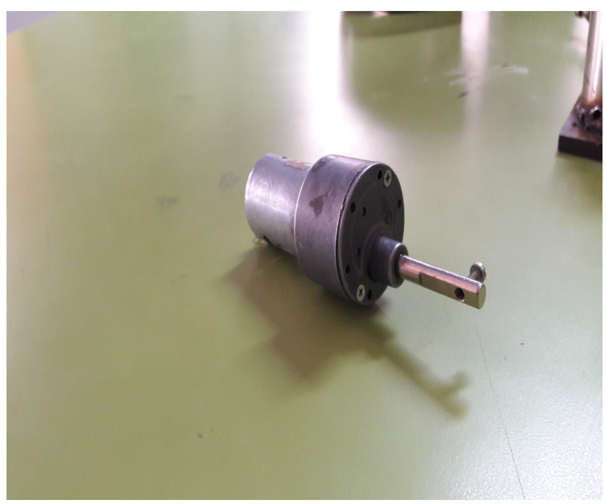
The bars which are connected between the knee part and ankle part are known as trans-tibial member .These bars

Fig.16 trans-tibial member

B) Fabrication of actuating mechanism:

Material Selected: Cast Iron

Components of actuating mechanism:



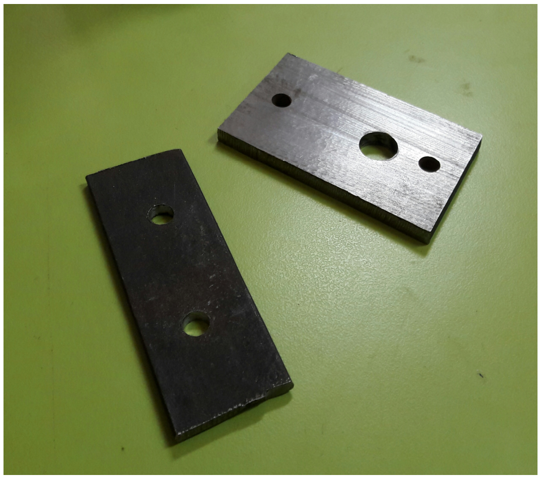


Fig.17 Parts of Actuator

The actuating mechanism consist of

- Motor (Johnson motor)
- Sleeve and sleeve nut
- Coupling (to connect motor shaft to actuator shaft)
- A casing for motor
- External Power Supply

The basic principle of operation is that when sleeve shaft rotates because of motor shaft, and screw pair allows only a relative movement between sleeve and sleeve nut which can expressed by single co-ordinate angle. Thus sleeve shaft and sleeve nut has single degree of freedom.In this way motion is transmitted from motor to transmittal member via linear actuation mechanism of screw pair.

CHAPTER 07

RESULTS AND DISCUSSION

CHAPTER 07: RESULTS AND DISCUSSION

The prosthetic legs available in markets are a single unit composite part. This means damage to even one part destroys the whole prosthetic leg and amputee has to spend extra money to purchase the whole leg, instead of the damaged part. There are no interchangeability with this traditional legs. Also, this composite leg provides negligible flexibility in operations, which in turn effects amputee's motion and also requires extra metabolic energy from patient (amputee).

Bionic leg are not a single unit, but a combination of various member viz. knee member, transtibial member, actuating mechanism, etc., all of which can be replaced individually if damaged. There is no need to spend extra cost to purchase the whole assembly, i.e. parts are interchangeable.

The pivot joints in bionic leg provides flexibility in motion and can adapt itself to various types of motion stride viz. waking on floor, climbing up/down the stairs, running, etc. This saves the metabolic energy of amputee.

In addition to pivots joint, unlike passive elastic prosthetic leg , bionic leg are provided with actuating mechanism(s) which further assist in motion of amputee without wasting metabolic energy.

Though the cost of passive elastic prosthetic leg is less than the cost of bionic leg, but keeping concern about amputee's (patient's) comfort, this cost factor may be avoided. The cost of replacement of a particular part of bionic leg is much less than buying a new passive elastic prosthetic leg completely.

CHAPTER 08

RECOMMENDATIONS

CHAPTER 08: RECOMMENDATIONS

6.1 Scope for Future:

A research team that is working on ways to improve amputees' control over prosthetics with direct help from their own nervous system.

New plastic scaffolds attached to prosthetic devices could enable nerves to feel and control artificial limbs, using electrical signals to bring back real sensations. The research could eventually realize the dream of connecting artificial body extensions to the living nervous system.

Despite major advances in prosthetics, researchers have not been able to fully integrate nerves and prosthetic devices — though several teams, including, have been trying. New research at Sandia National Laboratories, the University of New Mexico and MD Anderson Cancer Center in Houston could make it a reality.

Connecting mechanical instruments to human nerves is complex on several levels because the interface would need to share several special properties between man and machine. It would have to be biocompatible to promote nerve and tissue growth, but mechanically compatible to allow electrodes to connect to external circuits. It would have to be structured to avoid harming surrounding tissue, but it would have to work in concert with that tissue to serve as a real replacement limb.

New biocompatible interface scaffolds designed by Sandia researchers are a step in that direction. Scientist's electro spun liquid polymers to create polymer chains, forming a fiber structure. Multi-walled carbon nanotubes incorporated into the fibers provides electrical conductivity. Using this method, the team created scaffolds with two types of polymer, according to Sandia — PBF, which was developed for tissue engineering, and PDMS, a sort of biocompatible caulk. PBF is biodegradable, so the scaffold would disintegrate once installed, leaving the electrical contacts behind. PDMS is not biodegradable.

The idea is that a scaffold would provide a connection between existing nerves and new electronics, containing enough pores to let new nerves grow. The newly innervated limb would then theoretically have the same sensory characteristics as a real one.

This type of transplant is still year’s away, but recent tests with lab rats show its promise, the Sandia news release says. Robotics engineer Steve Buerger, one of the research leads, said the team is pursuing external funding to continue the research, "so we can bring this technology closer to something that will help our wounded warriors, amputees and victims of peripheral nerve injury."

CHAPTER 09

BIBLIOGRAPHY

CHAPTER 09: BIBLIOGRAPHY

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