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"Design and Fabrication of Improved Below-Knee Prosthesis Limb"

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1443 **2021**

بسم الله الرحمن الرحيم

"وَمَا تَوْفِيقي إِلَّا باللَّهِ عَلَيْهِ تَوَكَّلْتُ وَإِلَيْهِ أُنِيبُ"

{سورة هود: 88}

Certification of Supervisor

I certify that this project report entitled **Design and Fabrication of Improved Below- Knee Prosthesis Limb** prepared by **Noor Hisham Abd_Alkhalek, Noor Emad Jassem and Ghasaq Abd_Alhadi Mohammed** was made under my supervision at the Department of Biomedical Engineering /Division of Biomechanics in partial fulfillment of the requirement for the degree of B.SC. in Biomedical Engineering.

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Date: 15 / 7 /2021

Certificate of Examiners

We certify, as an examining committee, that we have read this project report entitled:

"Design and Fabrication of Improved Below- Knee Prosthesis Limb".

Examined the students Noor Hisham Abd_Alkhalek, Noor Emad Jassem and Ghasaq Abd_Alhadi Mohammed in their contents and found the project meets the standard for degree of B.SC. in Biomedical Engineering.

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شكر وتقدير

نتقدم بالشكر الجزيل إلى كافة أعضاء الهيئة التدريسية في قسم هندسة الطب الحياتي الذين أناروا دربنا الطويل وشدوا أزرنا لنكون شعلة للأجيال القادمة الى كل من علمنا حرفاً وكان سبباً في وصولنا الى هذا المستوى العلمي :

" رئيس قسمنا د . وسام كاظم وأساتذتنا الأفاضل "

وخالص الشكر نختص بالذكرالى الدكتور الذي تفضل بالإشراف على هذا العمل ولم يبخل في العلم والمعرفة :

الدكتور:- سعد محمود علي

إهداء

إلى من كافحوا وضحوا في سبيل البلد

إلى من خانتهم الظروف وأجبرتهم على الجلوس قسرًا

إلى المبتورة أقدامهم والسائرة أهدافهم

إلى أصحاب الشأن الذي عسى أن يكون عملنا سلماً ليرتقوا بهِ ويعمل على إعادة تأهيل حياتهم

إلى والدي وأصحاب الهمم نهدي هذا العمل.

Thanks and appreciation

We extend our heartfelt thanks to all the faculty members in the Department of Biomedical Engineering who enlightened our long path and strengthened us to be a torch for future generations to all those who taught us a letter and were the reason for us to reach this scientific level:

"Our department head is Dr. Wissam Kazem and our distinguished professors."

By mentioning the doctor who kindly supervised this work and did not skimp on knowledge and knowledge

Dr.: - Saad Mahmoud Ali

Dedication

To those who struggled and sacrificed for the sake of the country.

To those who were mixed with circumstances and forced them to sit forcibly.

To have their feet amputated and their goals were moving.

To the stakeholders, who may our work be a peace to rise with and work on rehabilitation, their freedom.

To my father and people of determination, we dedicate this work

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ABSTRACT

Intelligent prosthetics are used to replace disabled limbs in the field of biological rehabilitation. The main aims of this project are to design and fabrication of a comfortable prosthesis, light weight, durable, low-cost novel below- knee prosthesis limb to help patients to perform daily activities and to provide convenience for lower limb amputees. The below knee prosthesis limb was accomplished by using various metal cutting machines, in addition to manual works such as filing, polishing, etc. The ANSYS design and simulation were implemented by using the ANSYS 18.0 software mechanical design program by using a large number of parallel and oblique planes and sketches and by using the finite element method (FEM). To determine the mechanical properties of the fabricated prosthesis limb four patients were selected with total weights of 50, 75, 90 and 120 kg. The obtained mechanical properties have been built by using the Response Surfaces Methodology (RSM) and the statistical Expert Systems 11.0 software program. The highest equivalent Von. Misses stresses of the fabricated prosthesis limb occur at the beginning of the step, reached (105.17) MPa at the highest patient weight of 1500 N, which is higher than the position at the end of the step by (19.91 %), and a higher e than when the patient is standing by (54.19 %).The highest total deformations occur at the beginning of the step, reaches (0.271 mm) at the patient weight of 1500 N, which is higher than the position at the end of the step by (110.08 %).The highest equivalent elastic strains occur at the end of the step, reaches (0.00222 m/m). The highest strain energy occur at the end of the step, reaches the highest value as (0.159 mJ). The new fabricated foot is close in terms of all the mechanical properties of the natural foot.

CHAPTER ONE

INTRODUCTION

1.1 Introduction

The lower limb prosthesis is defined as a device that substitutes the function of a missing limb either due to amputation or a congenital defect [1] and an artificial replacement for any or all parts of the lower leg extremity. The loss of lower limb can profoundly influence an individual's quality of life.

However, many amputees reject lower limb prostheses or use them less than needed because of discomfort. The main cause of acquired limb loss is poor circulation in the limb owing to arterial disease, with more than half of all the amputations occurring among people with diabetes mellitus. there are two main subcategories of lower extremity prosthetic devices (lower limb prosthesis), which are [2]:

1- Trans- tibial (any amputation transecting the tibia bone or a congenital anomaly resulting in a tibial deficiency).
2- trans-femoral (any amputation transecting the femur bone or a congenital anomaly resulting in a femoral deficiency.

1.2 The Amputation Sources of the Problem

There are two primary sources of problems that lead to amputations [3]:

1- The number one cause of amputations is diabetes and its related ailments. had a foot or leg amputation due to illnesses caused by diabetes. Diabetes patients have an increased risk of developing peripheral arterial disease.
2- The second leading reason for an amputation is traumatic injury, or a wound resulting from an external source, for example the leading causes of amputations from traumatic injury are car accidents and injuries by machines.

1.3 The Solution of the Problem

Lower limb Prostheses are used frequently on patients who have had amputation of the legs at various levels due to various causes, especially due to accidents and diseases. The Prostheses are made and fitted according to levels of amputations. The considerations when choosing Prosthesis include the following [4]:

- The amputation levels.
- Contour of the residual limb.
- Expected function of the prosthesis.
- Cognitive function of the patient.
- Occupation of the client.
- Hobbies interests of the client.
- Cosmetic importance of the prosthesis (Aesthetics).
- Financial capacity of the client.

1.4 The Project Objectives

This project aims to:

- Design and fabrication of a comfortable prosthesis, to wear, to put on and to remove easy, light weight, durable, cosmetically pleasing mechanically functioning well and easy complete the requires reasonable maintenance.
- Use robotic prosthesis that could potentially enable people with a below-knee amputation to perform different types of motions that require power in lower limb joints. Our initial prototype, smart leg, integrates advanced prosthetic and robotic technology with the state-of-the-art machine learning algorithms capable of adapting the working of the prosthesis to the optimal gait and power consumption patterns, and which there for provide means to customize the device to a particular user.
- Contemporary prosthesis design goals to restore a natural and efficient gait by means of active and inactive components that are optimized to replicate characteristics of an intact limb.
- Use the most suitable material to be used in the making of prostheses so that the low-income wearers could afford to buy them.

CHAPTER TWO
THEORITICAL CONSIDERATIONS

2.1. Introduction

A prosthetic limb is an artificial replacement for a biological limb that may have been lost due to disease, injury, or deformation. For example, someone whose leg has been removed below the knee, a trans -tibial amputee, may choose to use a below-the-knee prosthetic to help them regain some mobility. With a prosthetic leg, he or she may be able to gain independence from a wheelchair. There are many styles of lower limb prostheses, such as running specific or the traditional design for simple walking [2].

2.2. Factors to Design a Prosthetic Limb

1. Material selection plays an important role in meeting the requirements of the prosthesis parts in order to make them effectively functional .
2. The cost of the material chosen has to be relevant (i.e. economical and affordable to low-income amputees, for instance) to be manufactured in mass productions since the material cost itself does contribute a lot in total manufacturing cost for each part [4].

Each design provides its user with the movement they want and can physically handle. Figure (2-1) shows a prosthetic leg made specifically for patients who want an active lifestyle. Around 60% of lower-limb amputees choose to use a prosthetic appendage, meaning that 30 million people worldwide require prosthetic limbs to carry out their daily activities.

Figure (2-1): Running specific prostheses (Courtesy of Ottobock Co.)

The prosthetic leg is often referred to as a "BK" or below the knee prosthesis while the trans femoral prosthetic leg is often referred to as an "AK" or above the knee prosthesis [3].

2.3 Level of Amputation

Figure (2-2) shows the names and heights for lower limb amputations [4].

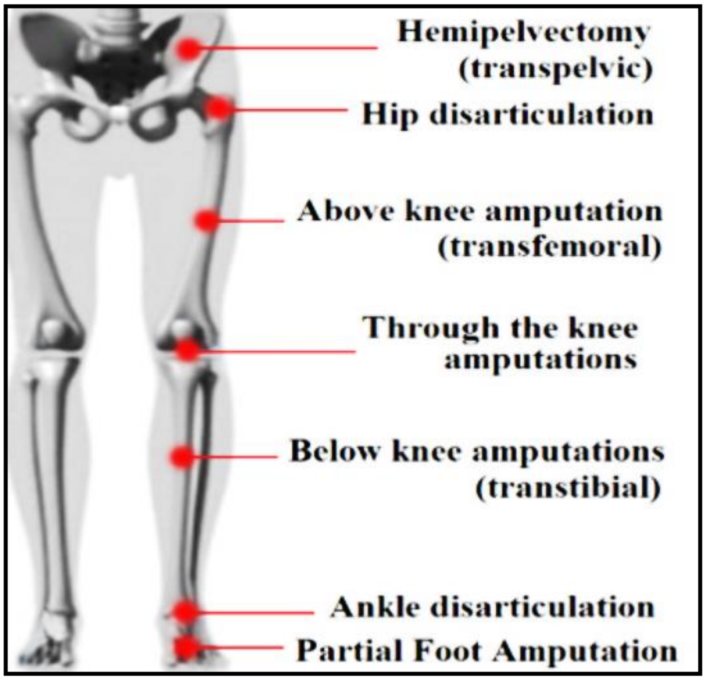


Figure (2-2): Names and heights for lower limb amputations

1.Partial foot Amputation

2.Syme's Amputation

3.Transtibial Amputation

4.Transfemoral Amputation

5.Knee Amputation

6.Hip disarticulation

7.Hemipelvectomy

2.4 Stages to Treatment of below-knee Amputations

The healing process for a lower limb prosthesis has multiple stages to allow gradual healing and adjustments given in figure (2-3). To be fitted for a prosthetic leg a patient must go through physical evaluations and an adjustment period to living without a prosthesis [3]. In general, healthy patients who may have suffered an amputation due to an accident will be fitted for a customized lower limb prosthesis within 4 to 6 weeks after their amputation. On the other hand, amputations caused by diabetes may be able to have a prosthesis after a 6 to 8-week period Furthermore, the process of how a patient is prepared to be fitted for a lower limb prosthesis is explained by the Brigham and Women's Hospital Department of Rehabilitation Services. There are nine stages to an amputee's treatment plan [3].

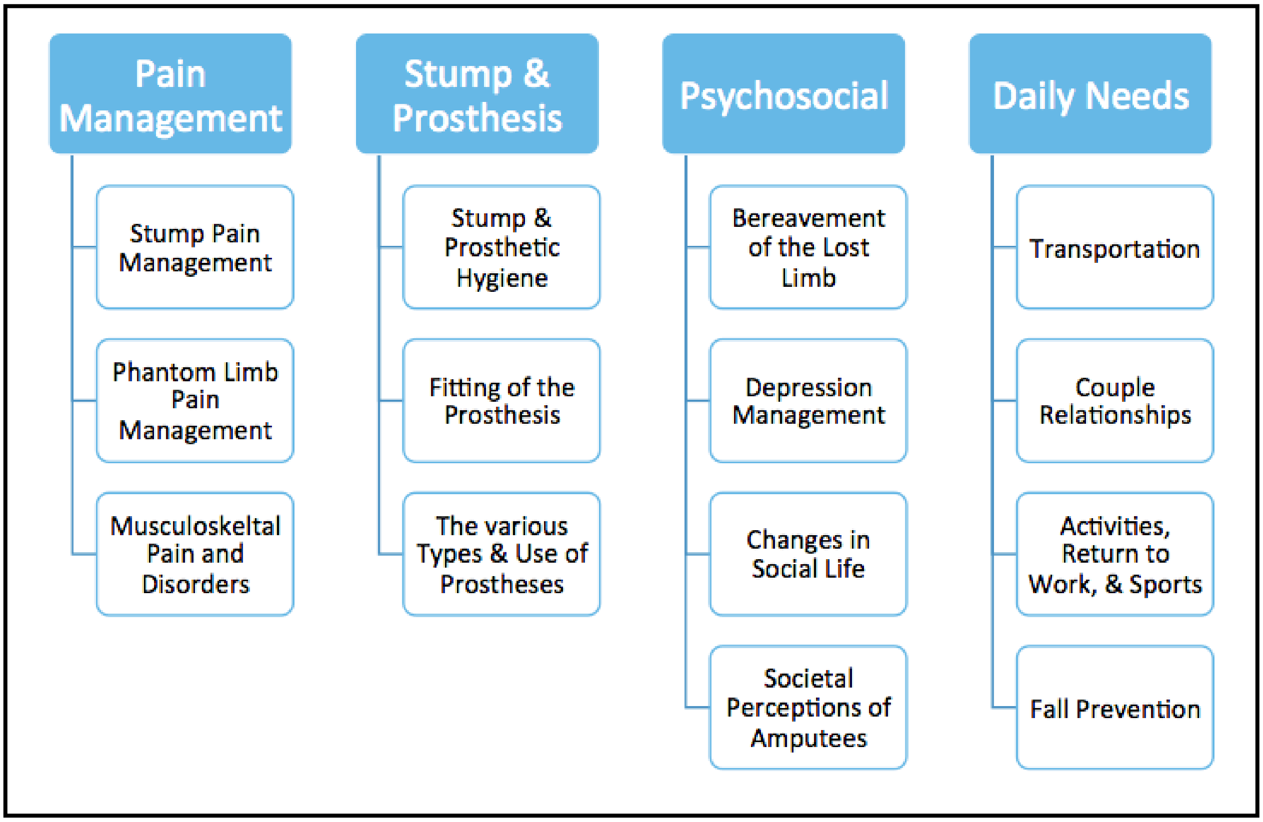


Figure (2-3): Thehealing process for a lower limb prosthesis multiple stages to allow gradual healing and adjustment

2.5 Devices / Accessories Used After Amputation of Both Lower Limbs

2.5.1 Socket

This important part serves as an interface between the residuum (stump) and the prosthesis, allowing comfortable weight-bearing, movement control and proprioception. It is fitting is one of the most challenging aspects of the entire prosthesis. The difficulties accompanied with the socket are that it needs to have a perfect fit, with total surface bearing to prevent painful pressure spots. It needs to be flexible, but sturdy, to allow normal gait movement but not bend under pressure [3].

The socket, which is designed to have properties similar to bone:

1. The socket distributes the person's weight from the body to the ground.
2. The materials used for the socket are composites based on thermoset plastics mixed with other materials.
3. A socket is shaped like a cup to fit around the residual limb.

Furthermore, it is composed of three parts:

1.the seating face, which connects the remaining limb to the prostheses is the which ensures correct motion of the leg and the distal.

2. Controlling socket area
3. Distal socket end, which transfers at most 10% of the person's weight.

Figure (2-4) shows the visual representation of the subdivision of the socket. To keep a healthy stance, the socket must maintain the muscle gluteus medius stretched, and the pelvis balanced. The socket facilitates the comfort and the functionality of the prostheses [3].Figure (2-5) shows the labeled view of a lower limb prosthesis.

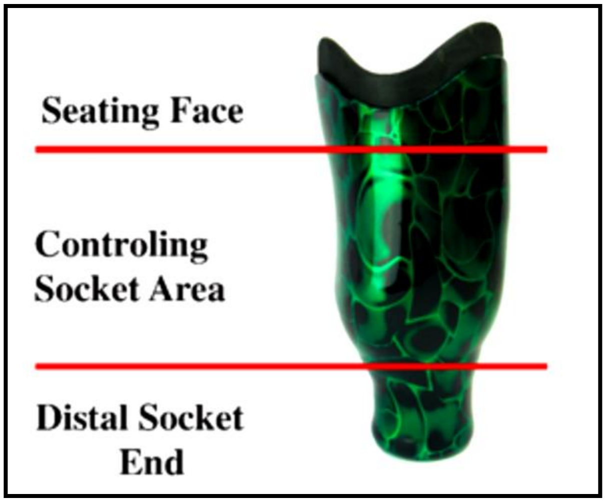


Figure (2-4): Hybrid polymer socket with labeled sections (Courtesy of 5280 Prosthetics LLC).

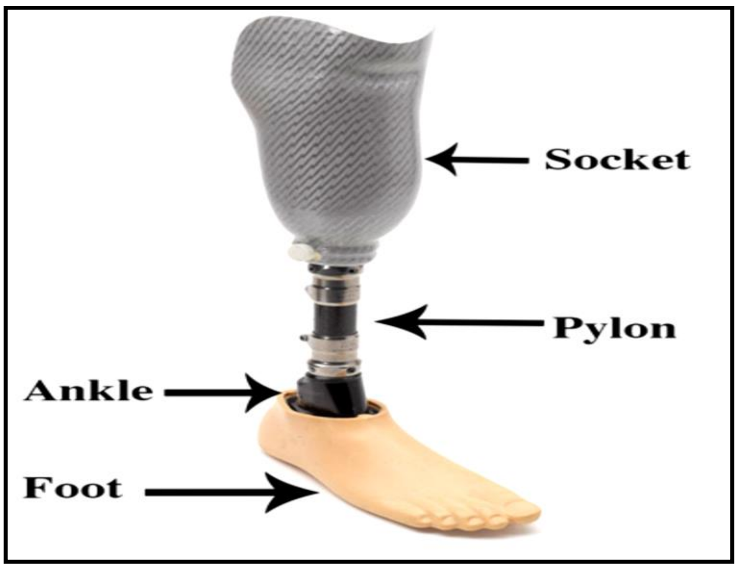


Figure (2-5): Labeled view of a lower limb prosthesis

2.5.2 Shank and Connectors

This part creates distance and support between the knee-joint and the foot (in case of upper-leg prosthesis) or between the socket and the foot. The type of connectors that are used between the shank and the knee/foot determines whether the prosthesis is modular or not. Modular means that the angle and the displacement of the foot in respect to the socket can be changed after fitting [3].

2.5.3 Foot module:

It includes prosthetic foot, foot adapters and heel, called also "virtual heel[5]. The foot connects the prosthesis with the floor and provides energy absorption. The most economical designs for the foot are the solid ankle cushion heel and the stationary attachment flexible endoskeleton foot Patients looking for increased mobility and stability may prefer an articulated single axis foot. The current design for lower limb prosthesis mimics the limb it is replacing [3].

2.5.4 The cosmetic foot

Providing contact to the ground, the foot provides shock absorption and stability during stance. Additionally, it influences gait biomechanics by its shape and stiffness. There are different types of feet, with greatly varying results concerning durability and biomechanics. These results are for adults and will probably be worse for children due to higher activity levels and scale effects. The heel is compressed during initial ground contact, storing energy which is then returned during the latter phase of ground contact to help propel the body forward.

2.4.5 Double adapter:

It includes double male or female pyramid adapters, which connects socket and knee in TF prosthesis as shown in figure (2-6), and can substitute a pylon in both TT and TF prosthesis [5]. The pylon, which is described as a hollow aluminum pipe and serves as the connection between the socket and foot. Additionally, the pylon transfers the weight from the socket to the foot [3].

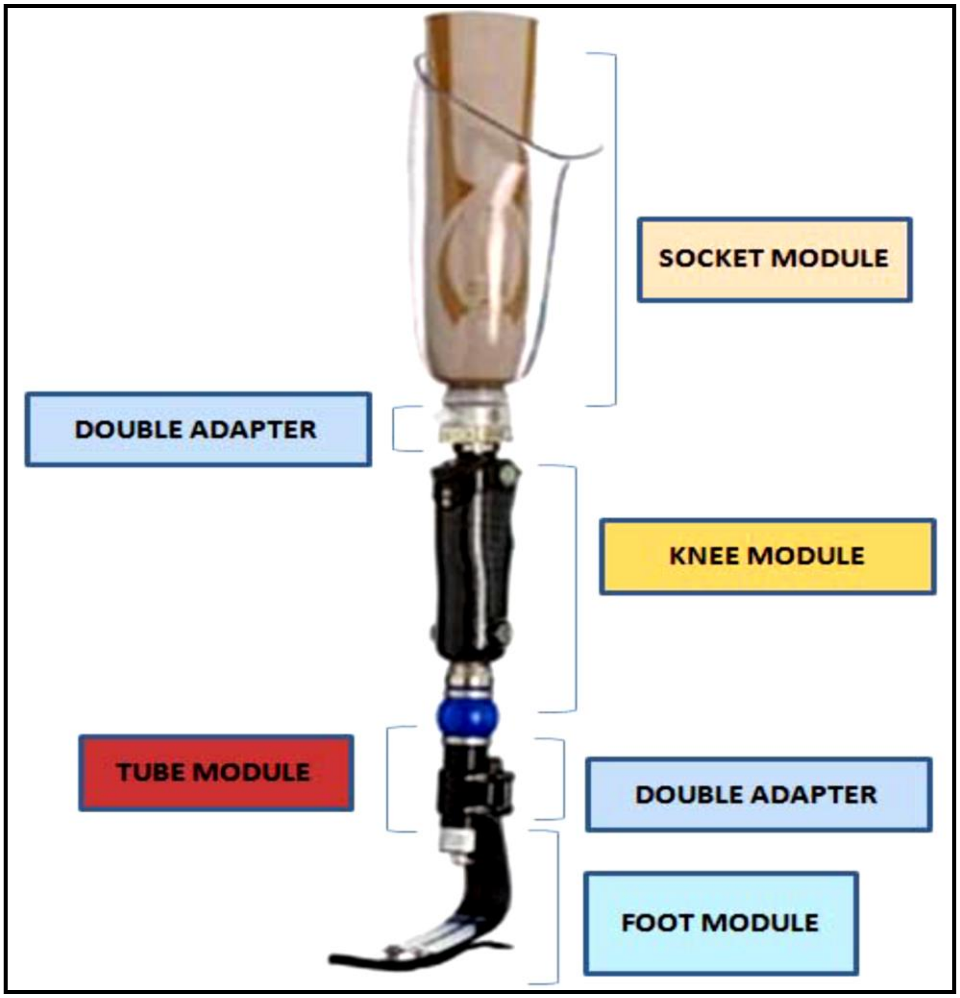


Figure (2-6): Scheme of TF lower limb prosthesis modules

2.6 Design consideration

There are multiple factors to consider when designing a transtibial prosthesis. Manufacturers must make choices about their priorities regarding these factors.

The buyer is also concerned with numerous other factors [4]:

- Cosmetics
- Cost

- Ease of use
- Size availability

2.7 The Cost

2.7.1 High-cost

The cost of an artificial limb does recur because artificial limbs are usually replaced every 3–4 years due to wear and tear. In addition, if the socket has fit issues, the socket must be replaced within several months. If height is an issue component can be changed, such as the pylons [3].

2.7.2 Low-cost

Low cost above knee prostheses often provide only basic structural support with limited function. This function is often achieved with crude, non-articulating, unstable, or manually locking knee joints.

2.8 Manufacturing and Materials of Prosthesis

An important consideration in the design and fabrication of a limb prosthesis is the type of material used for its construction. Interface materials will influence the comfort of the socket. Structural materials will affect the strength and weight of the overall prosthesis. The prosthetist has a vast array of materials to choose from in designing the optimal prosthesis for a particular individual.

Each individual needs to be evaluated with careful consideration given to their lifestyle, expectations and physical characteristics [4].

2.8.1 Plastic Polymer Laminates (Thermosets)

Plastic polymer laminates are widely used for the fabrication of prosthetic sockets. This process is performed under vacuum pressure in order to create a product that is lightweight and strong.

Common types of plastic polymer:

1. laminates used in prosthetics are acrylic.
2. Epoxy
3. Polyester.

The advantage of plastic laminates is that [4]:

- The prosthetist has a great deal of control over the strength, stiffness and thickness of the finished product.
- These variables can be controlled such that the finished product may be strong and thick in certain areas and thin and relatively light in others.

One significant disadvantage of these thermoset resins compared with sheet thermoplastics is that:

- The laminate is difficult and limited in its ability to be remolded after original fabrication. This means that if a specific area of the socket is uncomfortable due to tightness after the prosthesis is in use.
- It will be more difficult for the prosthetist to heat and remold that area.

2.8.2 Reinforcement textiles

Reinforcement textiles are the fabrics used in a laminate to provide strength. These include [4]:

1. Fiberglass
2. Nylon
3. Dacron
4. Carbon.

These materials all have their advantages and disadvantages:

- carbon fiber is used to create thin, lightweight and strong prosthetic sockets. This is one reason why a prosthetist might use several reinforcement materials in combination, a composite, to design a prosthetic socket.
- Carbon fiber is also used in several of the dynamic response (energy storing) feet and to create strong lightweight pylons.

2.8.3 Thermoplastics

Sheet thermoplastics are widely used for prosthetic interfaces as well as structural components [4]. These materials are available in sheet form in various thickness and colors.

The most basic types of sheet thermoplastics are:

1. polypropylene
2. polyethylene.

Characteristics of these materials:

- vary from very stiff to very flexible.
- which is a blend of polypropylene and ethylene creating a material that is fairly rigid yet more flexible.
- crack resistant than straight polypropylene.
- Stiff materials are often used for support so that the forces associated with walking can be transmitted from the amputee to the floor.
- The most common test socket material today is Clear Co-Polyester thermoplastic [4].

2.8.4 Silicone and similar materials

Silicone is used as a padding material in sockets, as a means of suspension in the silicone suction socket (3S Iceross type) and is the material of choice when making high quality cosmetic leg restorations, to name a few.

silicone not only provides excellent padding but also protects the skin from friction (shear). This can be very important because friction often is the cause of skin breakdowns. The silicone suction socket uses what could be called a silicone "sock" worn directly against the skin that incorporates an attachment pin at the bottom, locking the sock, and the amputee, into the prosthesis [4].

2.8.5 Metals

The metal in a prosthetic component include:

1. Knees
2. Pylons
3. Ankles
4. Rotators.

These components can be made of ((aluminum, stainless steel and titanium)).

- **Aluminum:** in general, is considered as a lightweight alternative to steel. It's not as strong but depending on the particular application it is often strong enough to meet the design criteria and pass the necessary testing procedures. Certain knees are fabricated of aluminum, taking advantage of its light weight. Some of these knees are very strong and durable owing much of their strength to the geometry of the knee as well as the material used [4].
- **Steel:** is certainly strong but it is relatively heavy. Because steel is strong, it can be used to create small components that may rely more on the

strength of the material than the geometry of the design. Small knee units used for end skeletal prostheses were originally made of steel. The material is fairly heavy however very little material is needed to construct these knees.

- **Titanium:** is a strong, lightweight alternative. The penalty for using titanium is higher cost. Many of the end skeletal components originally designed of steel are now available in titanium.

2.8.6 Liners and Sleeves

Liners are the materials that are either made during the fabrication process to fit inside the socket or those materials added after the leg is in use in order to accommodate for shrinkage of the residuum. Pelite is probably the most commonly used "soft socket" liner material.

It is a closed-cell polyethylene foam material that is available in various durometers. This material is thermoformable, meaning that it can be heated and formed over the plaster cast. Pelite and materials like it have the advantage that they are easily adjusted by adding additional material when the residual limb shrinks. It is also convenient to use these materials if suspension of a below-knee prosthesis.

Sleeves are worn over the outside of the prosthesis and are used to provide suspension. Some common materials used for sleeves are: neoprene, silicone, latex, and urethane. Sleeves are worn over the prosthesis and extend onto the thigh of the below-knee amputee. The most common materials used in prosthetics today are various plastics, but the more traditional materials such as wood, leather, metal, and cloth still have a role to play [4].

2.8.7 Cloth

Cloth is used for prosthetic socks, waist belts, straps, and harnesses for upper-limb prostheses. Probably the greatest use of cloth is for prosthetic socks which can be keep:

- The skin dry
- Cushion the limb
- Absorb shear forces
- Take up volume to improve the fit.

Prosthetic socks are commonly made:

1. wool
2. cotton, or blends of these natural fibers often combined with nylon, Orion, acrylics, or other man-made materials [4].

Wool is the most common material used for prosthetic socks because of its characteristic elasticity, cushioning, and ability to absorb moisture without feeling damp. Wool also has good resistance to acids from perspiration. The blend of domestic and foreign wool fleece used in prosthetic socks provides greater resistance to shrinkage. Wool prosthetic socks should be dried carefully by first removing the excess water, wrapping them in a towel, and then drying them away from sunlight or any other direct heat. The recent development of machine-washable wool should reduce the need for hand washing in the future.

Cotton is also used for prosthetic socks but is more common in the form of a stockinette used to protect the limb during casting procedures. Cotton is also blended with wool in prosthetic socks, and some 100% cotton prosthetic socks are available [4].

2.8.8 Plastics

Nylonis used for prosthetic sheaths, plastic laminations, bushings, suction valves, and nylon stockings to cover prostheses.

The major advantages of this man-made fiber are:

- Its strength
- Elasticity
- Low coefficient of friction.

Nylon prosthetic sheaths are in common use for transtibial amputees. A thin sheath worn directly over the skin significantly reduces shear stresses and helps to pull moisture away from the skin into the outside prosthetic socks. A nylon stockinette provides inherent strength to nearly all prosthetic laminations.

Acrylicsare thermoplastics that have greater durability and strength than polyester resins do. Acrylic fibers are frequently used in the newer synthetic blends for prosthetic socks since this material is soft, durable, and machine washable. Acrylic resin is increasingly popular for laminations in prosthetics because its high strength permits a thinner, lighter-weight lamination [4].

Polyester *resin* is a thermosetting plastic that is most commonly used for laminations in prosthetics. Thermosetting plastics cannot be heated and reformed after molding without destroying their physical properties. Polyester resins come in a liquid form that can be pigmented to match the patient's natural skin tone [4].

2.8.9 Cosmesis

Cosmetic prosthesis has long been used to disguise injuries and disfigurements. With advances in modern technology, cosmesis, the creation of lifelike limbs made from silicone or PVC has been made possible. Such prosthetics, such as artificial limbs, can now be made to mimic the appearance actual ones.

Custom-made cosmeses are generally more expensive, while standard cosmeses come ready-made in various sizes, although they are often not as realistic as their custom-made counterparts. Another option is the custom-made silicone cover, which can be made to match a person's skin tone but not details such as freckles or wrinkles [4].

2.9 Material for Manufacturers Prosthetic Limb

- **The socket:** is usually made from "polypropylene-resistant to many chemical solvents, bases and acids.
- The pylonis usually made from lightweight metals, such as titanium and aluminum, steel.

The newest development in prosthesis manufacturing has been the use of carbon fiber to form a lightweight pylon.

- **The foot of the prosthesis:** is made from urethane foam with a wooden inner keel construction, but due to uneconomical and hazardous effects on the environment, the use of leather or wood, for instance, has been replaced by polypropylene-based materials, such as polyethylene, polypropylene,acrylics, and polyurethane.

CHAPTER THREE
DESIGN OF BELOW-KNEE PROSTHESIS LIMB

3.1Traditional Prosthetic

Prosthesis is often used to restore appearance and functional activity to persons having lower limb amputation. Below Knee (BK) prostheses are typically comprised of four major components as shown in Figure (3-1), these are [6]:

1- Socket

2- Pylon (shank)

3- Foot prosthetic

4- Cosmesis

Figure (3-1): Below limb prostheses

3.1.1 The Liner, sleeves, socks

Soft interface materials that ensurefit, comfort, and that the prostheses stay attached to residual limb. Certain suspension systems require use of liners. When used properly, they provide a cushioning effect within the socket, help to minimize friction forces, and provide even pressure distribution. Socks can be used to adapt to changes in the volume of the residual limb. It fabricated from ethylene-vinyl acetate (EVA) foam, silicone, gel, urethane, thermoplastic elastomer (TPE), pelite, wool, cotton material that mad it [6].

3.1.2 The Socket

Where the prosthetic device attaches to the residual limb. Because the residual limb is not meant to bear body weight, sockets must be individually moulded and meticulously fitted to ensure pressure is distributed, and to avoid damage to skin and tissue.It fabricated from polypropylene, thermoplastic elastomer (TPE), wood, aluminum, glass-reinforced plastic (GRP), resin, carbon fiber.

3.1.3 The Pylon

Connects the socket to the foot. Lightweight and absorbs shock. It fabricated from wood, titanium, aluminium, steel, carbon fibre, glass-reinforced plastic (GRP).

3.1.4 The Foot

Designed to be the point of contact between prosthesis and contact surface, with different foot designs optimised for different functions or terrains. It fabricated from polypropylene, polyurethane, wood, rubber, carbon-fibre [6].

3.1.5 Cosmesis

Limb covering to mimic appearance of real limb. Can be readymade or custom-designed, or made from locally sourced materials.It fabricated fromsilicone, local fabrics, Ethylene-vinyl acetate (EVA) foam. The fabricating processes shows in figure (3-2).

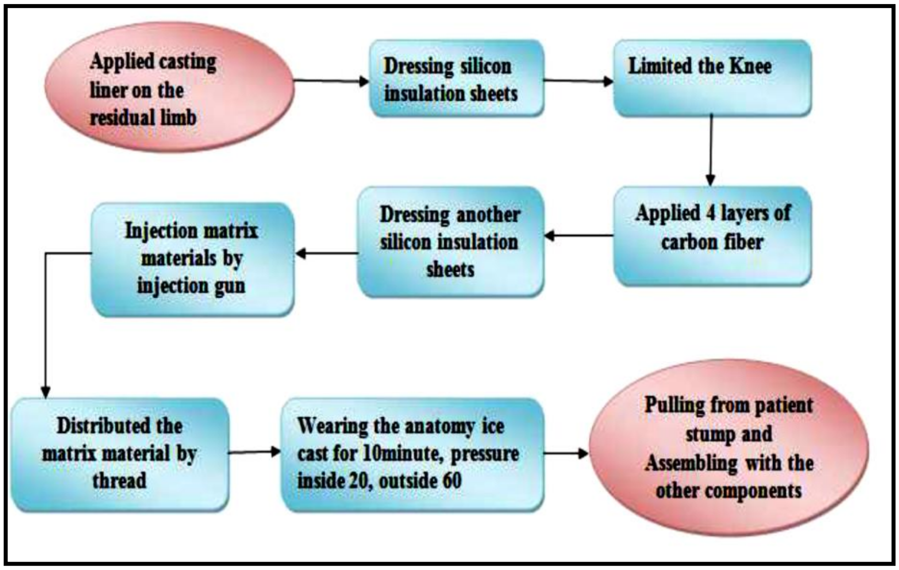


Figure (3-2): The fabricating processes of the mimic appearance of real limb

3.2 Traditional Sockets

3.2.1. Negative mold

Made by wrapping residual limb with a wet plaster-of-Paris bandageas shown in figure (3-3).

Figure (3-3): Thenegative mould

3.2.2. Positive mould

Made by filling the cast with a mixture of plaster-of-Paris and water as shown in figure (3-4).

Figure (3-4): The positive mould

3.2.3. Rectify rectifications

Rectify rectifications are made to the positive mold as shown in figure (3-5).

Figure (3-5): Therectify rectifications made to the positive mold

3.2.4. Socket formed

Socket is formed by draping polypropylene or using laminated resins [6],as shown in figure (3-6).

Figure (3-6): Thesocket formed

3.2.5. Final changes

Final adjustments to the socket made using machinery, suspension attached as shown in figure (3-7).

Figure (3-7): Thefinal changes

3.3 Modern prosthetic

Commercially available below-knee prostheses are completely passive during stance, and consequently, their mechanical properties remain fixed with walking speed and terrain. These prostheses typically comprise elastic bumper springs or carbon composite leaf springs that store and release energy during the stance period, e.g., the Flex-Foot or the Seattle-Lite [7].

Lower extremity amputees using these conventional prostheses experience many problems during locomotion. For example, transtibial amputees expend 20–30% more metabolic power to walk at the same speed as able-bodied individuals, and therefore, they prefer a 30–40% slower walking speed to travel the same distance.

3.4 Engineering Challenges

Two main engineering challenges hinder the development of a powered ankle–foot prosthesis:

1. Mechanical design: With current actuator technology, it is challenging to build an ankle–foot prosthesis that matches the size and weight of the human ankle, but still provides a sufficiently large instantaneous power output and torque to propel an amputee. The shank–ankle–foot complex of a 78 kg person weighs approximately 2 kg, while the peak power and torque at the ankle during walking can be as high as 350 W and 140 N·m, respectively [7].

Current ankle–foot mechanisms for humanoid robots are not appropriate for this application, as they are either too heavy or not sufficiently powerful to meet the human-like specifications required for a prosthesis.

2. Control system design: A powered prosthesis must be position- and impedance-controllable. Often robotic ankle controllers for humanoid robots follow preplanned kinematic trajectories during walking, whereas the human

ankle is believed to operate in impedance control mode during stance and position control mode during swing. Furthermore, for ease of use, only local sensing on the prosthesis is preferable, which adds additional constraints on the control system design. Finally, it is unclear what kind of prosthetic control strategy is effective for the improvement of amputee ambulation

3.5 Design Specifications and Target Ankle Stance Behaviors for the Prothesis

In this section, we first review human ankle biomechanics in walking. Using these biomechanical descriptions, we then define the design specifications for the prosthesis [7].The normal human ankle biomechanics for level-ground walking is shows in figure (3-8).

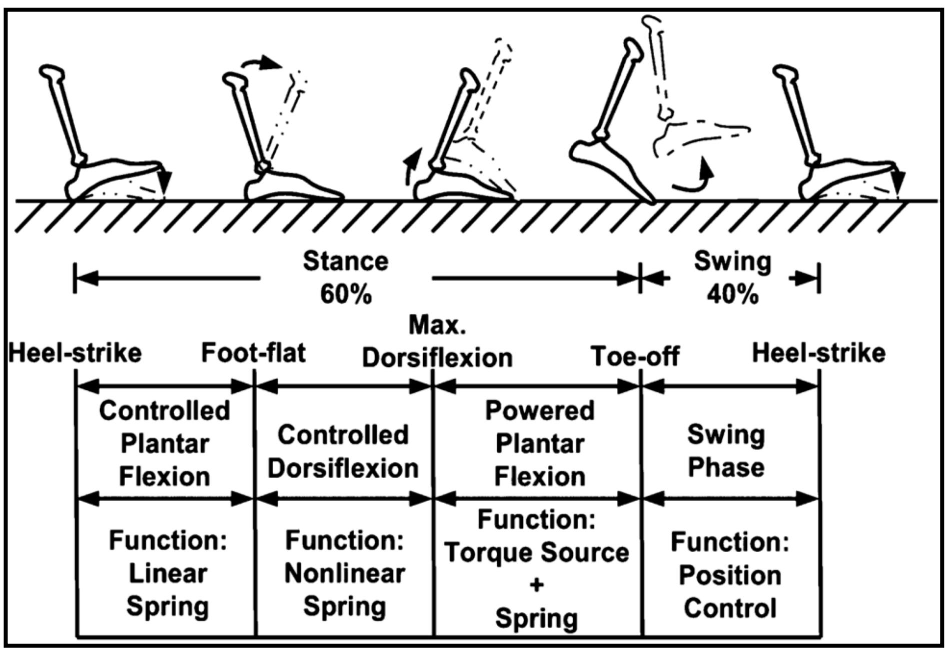


Figure (3-8): The normal human ankle biomechanics for level-ground walking

Figure (3-9) shows theaverage ankle torque is plotted versus ankle angle for N = 10 individuals with intact limbs walking at a moderate gait speed (1.25 m/s).The solid line shows the ankle torque–angle behavior during stance while the dash line shows the ankle behavior during the SW. The points (1), (2), (3), and (4) represent the conditions of the foot at heel-strike, foot-flat, maximum dorsiflexion, and toe-off, respectively.

The segments (1)–(2), (2)–(3), (3)–(4), and (4)–(1) represent the ankle torque–angle behaviors during CP, CD, PP, and SW phases of gait, respectively.Segments (1)–(2) and (2)–(3) reveal different spring behaviors of the human ankle during CP and CD, respectively. The area W enclosed by points (1),(2) ,(3), and (4) is the net work done at the joint per unit body mass during the stance period.

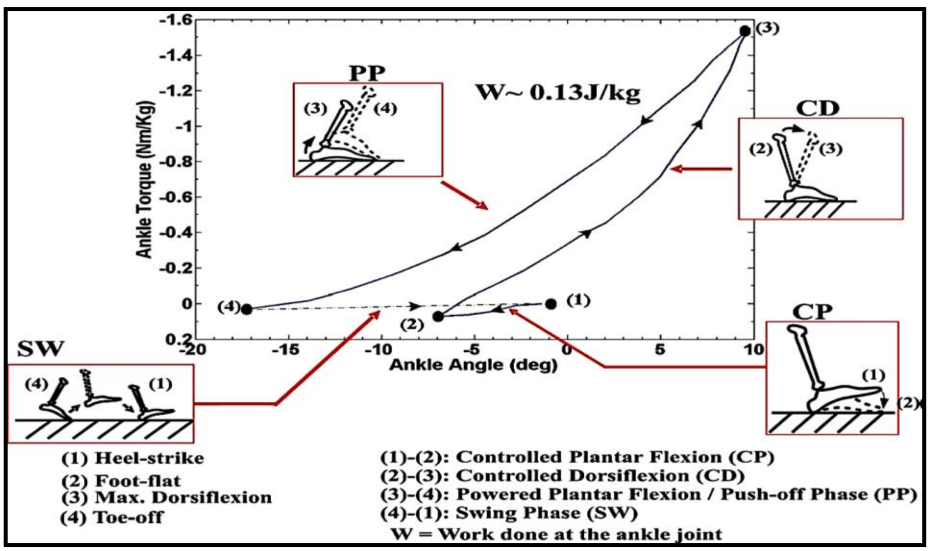


Figure (3-9): The average ankle torque is plotted versus ankle angle for N = 10 individuals with intact limbs walking at a moderate gait speed (1.25 m/s).

3.5.1 Human Ankle Biomechanics in Walking

A level-ground walking cycle is typically defined as beginning with the heel strike of one foot and ending at the next heel strike of the same foot. The main subdivisions of the gait cycle are the stance phase (60% gait cycle) and the swing phase (SW) (40% gait cycle) as shown in figure (3-8). The SW represents the portion of the gait cycle when the foot is off the ground. The stance phase begins at heel-strike when the heel touches the ground and ends at toe-off when the same foot rises from the ground surface [7].

From the stance phase of walking can be divided into three subphases: controlled plantar flexion (CP), controlled dorsiflexion (CD), and powered plantar flexion (PP). These phases of gait are described in as shown in figure (3-8). In addition, figure (3-9) shows average ankle torque–angle characteristics for N = 10 individuals with intact limbs walking at a moderate speed (1.25 m/s). Detailed descriptions for each subphase are provided next.

1. **Controlled plantar flexion**: CP begins at heel-strike and ends at foot-flat. Simply speaking, CP describes the process by which the heel and forefoot initially make contact with the ground. researchers showed that ankle joint behavior during CP was consistent with a linear spring response with joint torque proportional to joint position. As shown in Fig.8, segment (1)–(2) illustrates the linear spring behavior of the ankle.
2. **Controlled dorsiflexion**: CD begins at foot-flat and continues until the ankle reaches a state of maximum dorsiflexion. Ankle torque versus position during the CD period can often be described as a nonlinear spring where stiffness increases with increasing ankle position. The main function of the human ankle during CD is to store elastic energy to propel the body upward and forward during the PP phase. Segment (2)–(3) in Fig. 8reveals the nonlinear spring behavior of the human ankle joint during CD.
3. **Powered plantar flexion**: PP begins after CD and ends at the instant of toe-off. Because the work generated during PP is more than the negative work absorbed during the CP and CD phases for moderate to fast walking speeds, additional energy is supplied along with the spring energy stored during the CD phase to achieve the net ankle work and high plantar flexion power

during late stance. Thus, during PP, the ankle can be modeled as a torque source in parallel to a CD spring. The area *W* enclosed by the points (1), (2a), (3), and (4) shows the amount of net work done at the ankle during the stance period.

4. **Swing phase**: SWbegins at toe-off and ends at heel-strike. It represents the portion of the gait cycle when the foot is off the ground. During SW, the ankle can be modeled as a position source to achieve foot clearance as well as to reset the foot to a desired equilibrium position before the next heel strike.

In summary, for level-ground walking, the human ankle provides three main functions:

1) it behaves as a spring with variable stiffness from CP to CD;

2) it provides additional energy for push-off during PP; and

3) it behaves as a position source to control the foot orientation during SW.

3.5.2 Target Stance Phase Behavior

The key question for the design andcontrol of the prosthesis is to define a target walking behavior. For the SW, the desired ankle behavior is just to reposition the foot to a predefined equilibrium position. Although the equilibrium position of the ankle at heel strike should ideally be modulated between walking cycles based on walking speed and terrain, in this investigation we selected a fixed equilibrium position to simplify the control design. For the stance phase control, instead of simply tracking human ankle kinematics, it is commonly believed that the prosthesis should mimic the human ankle's "quasi-static stiffness," i.e., the slope of the measured ankle torque–angle curve during stance [7]. Mimicking the quasi-static stiffness curve of an intact ankle during walking, as shown in figure (3-9), is the main goal for the stance phase controller of this investigation. As shown in figure (3-10a), a typical quasi-static stiffness curve [from points (1)–(4)] can be decomposed into two main components: spring and torque source.

The first component comprises two springs fitted to the torque versus angle curve during the CP and CD phases [See figure (3-10b)]. The second component comprises a torque source that represents the residual torque between the spring torque and the total human ankle torque. It is noted that the stiffness from points (1)–(2) is approximately equal to the stiffness in the first portion of CD.

Thus, CP and the first portion of CD were modeled with a single spring as shown in figure (3-10b). For the ease of implementation, we simplified these two components (spring and torque sources) to obtain the target stance phase behavior for the prosthesis as depicted in Fig. 9(b). Specifically, we linearized the CD spring and torque source functions, and provided only the spring components during CP and CD since the torque source is negligible during these gait phases [see figure (3-10a)]. Each component is described as follows:

1. The first component comprises a linear torsional spring with a stiffness that varies with the sign of the ankle angle. When the ankle angle is positive, the stiffness value is set to *K*CD. When the ankle angle is negative, the stiffness value is set to *K*CP [See figure (3-10b)].

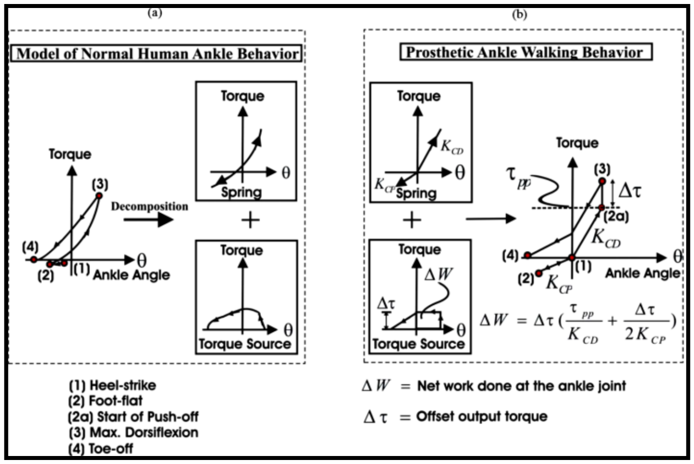


Figure (3-10): The model of human ankle behavior.

2. The second component comprises a constant offset torque $\Delta\tau$that provides the torque source during PP. This offset torque is applied in addition to the linear torsional spring$K$CDduring PP. τppdetermines the moment at which the offset torque is applied, indicated by point (4) in Fig. 9(b). The actual work done by the ankle joint due to the torque source ΔW is:

$$\Delta W = \Delta\tau\left(\frac{\tau_{\mathrm{PP}}}{K_{\mathrm{CD}}} + \frac{\Delta\tau}{2K_{\mathrm{CP}}}\right) \tag{3-1}$$

It is noted here that conventional passive prostheses only provide the spring behavior but fail to supply the torque source function to further enhance propulsion during PP.

3.5.3 Design Specifications

Using the aforesaid biomechanical descriptions and the design goals for the prosthesis are summarized as follows:

1. the prosthesis must have a mass distribution comparable to the missing human limb;
2. the system must deliver a human-like output power and torque during PP.

3. the system must be capable of changing its stiffness as dictated by the quasi-static stiffness of an intact ankle.
4. the system must be capable of controlling joint position during the SW
5. the prosthesis must provide sufficient shock tolerance to prevent any damage to the mechanism at heel-strike. It is important to note that the prosthesis and controller designs are not independent. Rather, they are integrated to ensure that the inherent prosthesis dynamics does not inhibit the controller's ability to specify desired dynamics.

In the remainder of this section, the target parameters for the design goals are outlined [7] as:

1. **Size and weight:**The target height for the prosthesis is specified, based on the nominal height of a conventional high profile below-knee prosthesis, which is about 18 cm from ground to pyramid dome. The desired prosthesis mass should be 2.5% of total body mass, equal to the percent mass of the missing biological limb at a point 18 cm from the ground surface.
2. **Range of joint rotation:**The proposed range of joint rotation for the prosthesis was based upon normal human ankle range of motion during walking. The maximum plantar flexion angle (25◦) occurs just as the foot is lifted off the ground at toe-off, while the maximum dorsiflexion angle (15◦) occurs at terminal CD.
3. **Torque and speed:**the measured peak velocity, torque, and power of the human ankle during the stance period of walking can be as high as 5 rad/s, 1.7 N·m/kg, and 3.5 W/kg, respectively. Both peak torque and power were normalized by body
4. **Mass:** Rather than simply satisfying these peak values, the torque-speed capability of the prosthesis was designed to cover the entire human ankle torque-speed curve of walking.

5. **Torque bandwidth:**The torque bandwidth requirement of the prosthesis was estimated based upon the power spectrum of the human ankle torque data during the stance period of walking. In this paper, the torque bandwidth was defined at that frequency range over which 70% of the total signal power was captured. Analyzing the normal human ankle, the torque bandwidth was found to be ~3.5 Hz in which the ankle torque varies between 50 and 140 N·m. The goal was therefore to design a torque controller capable of outputting any torque level between 50 and 140 N·m at 3.5 Hz. This goal requires that the torque bandwidth of the open-loop system be significantly larger than 3.5 Hz, otherwise the inherent dynamics of the prosthesis may inhibit the controller' stability to specify desired dynamics.
6. **Net positive work:**The prosthesis should also be capable of generating net positive work during stance. The average net positivework done at the ankle joint per unit body mass for self-selected and fast walking speeds is ~0.10 J/kg and ~0.26 J/kg, respectively.
7. **Controlled dorsiflexion stiffness:**The prosthesis should output a human-like quasi-static stiffness during CD, or from point 2 to point 3 in Fig. 2. A target stiffness value was obtained by estimating the slope of the measured human ankle torque-angle curve from the zero torque-angle point to the torque at maximum dorsiflexion, or point 3 in figure (3-9) [7]. The average human stiffness per unit body mass at a self-selected walking speed is ~8 N·m/rad·kg. We design an ankle–foot prosthesis for a nominal male subject, walking at a self-selected speed of 1.25 m/s, whose body mass, height, and foot length are 78 kg, 175 cm, and 27 cm, respectively.

Table (3-1) gives the design specifications for a nominal male subject.

Table (3-1): The design specifications for a nominal male subject.

Total Prosthetic Mass (kg)	2
Max. Allowable Dorsiflexion (deg)	15
Max. Allowable Plantarflexion (deg)	25
Peak Torque (Nm)	133
Peak Velocity (rad/s)	5
Peak Power (W)	274
Torque Bandwidth (Hz)	3.5
Net Work Done (J)	7.8
Controlled Dorsiflexion Stiffness (Nm/rad)	627

CHAPTER FOUR
FABRICATING OF THE BELOW KNEE PROSTHESIS LIMB

4.1 The Selected Material

Different engineering materials were used in this work.Table (4-1)gives the materials list required to fabricating of the below knee prosthesis limb.

Table (4-1): Thematerials list required to fabricating of the below knee prosthesis limb

Item	Part name	Part material	Quantity	Dimensions
1	Tube	Fiber	1	D=72 mm, d=62 mm; L =40 mm
2	Tube	Fiber	1	D=60 mm, d=50 mm; L =40 mm
3	Tube	Aluminum	1	D=66 mm, d=50 mm; L =40 mm
4	Tube	Aluminum	1	D=102 mm, d=72 mm; L =40 mm
5	Teflon sheet	PTFE	1	L= 2 m; W=40mm; T=1mm
6	Shaft bar	Steel	1	D=24 mm; L=15 mm
7	Net (square)	Steel	1	M12 mm
8	Belt	Steel	1	M12 mm; L =40 mm
9	Belt	Steel	2	M12 mm; L =30 mm
10	Belt	Steel	2	M12 mm; L =40 mm
11	Tube	Steel	1	D=32 mm, d=24mm; L =10 mm
12	Belt	Steel	2	M5 mm; L =50 mm
13	Disk	Brass	1	D=24 mm; L=10 mm
14	Sheet	Fiber	1	L= 260 mm; W=40mm; T=5mm
15	Sheet	Fiber	1	L= 130 mm; W=40mm; T=5mm
16	Sheet	Fiber	1	L= 90 mm; W=40mm; T=5mm
17	Tube	Aluminum	1	D=30 mm, d=18 mm; L =250 mm
18	Tube	Aluminum	1	D=40 mm, d=17 mm; L =40 mm
19	Plate	Steel	1	L= 50 mm; W=40mm; T=8mm
20	Strip	Steel	3	L= 35 mm; W=12mm; T=3mm
21	Plate	Aluminum	1	L= 80 mm; W=70mm; T=40 mm

22	Tube	Steel	1	D=80 mm, d=72 mm; L =40 mm

A number of raw materials have been purchased and included; the lihgtweighte aluminum alloy tubes with a dimensions of (D1= 34 mm, D2= 20 mm and L=500 mm), and (D1= 60 mm, D2= 46 mm and L=300 mm) and aluminum alloy shafts with a dimensions of (D= 60 mm and L=120 mm), and (D= 100 mm and L=100 mm)as shown in the figure (4-1).

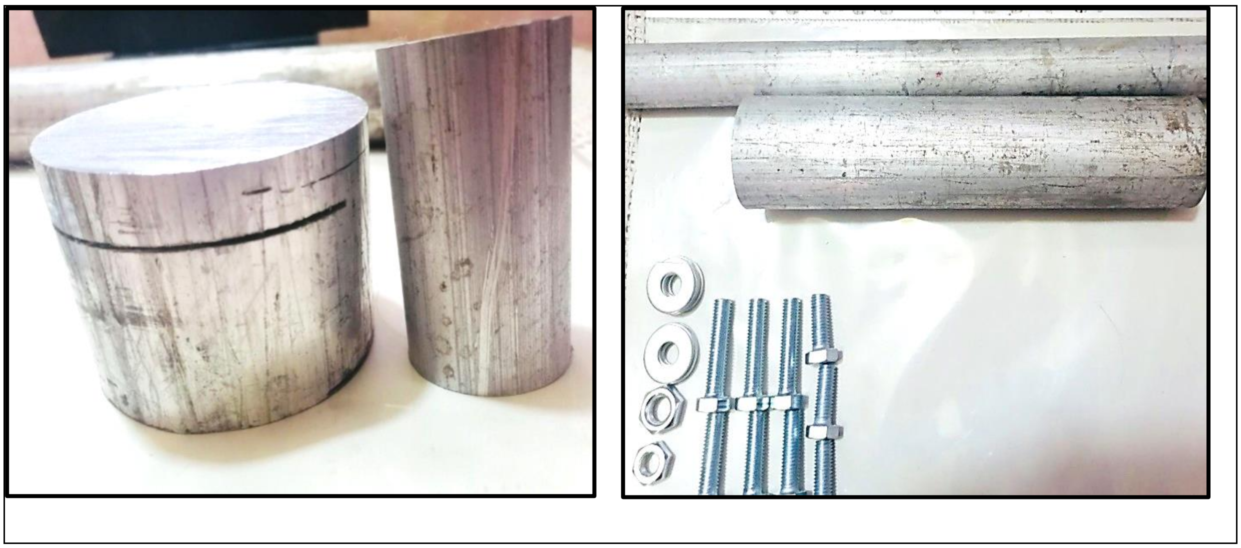


Figure (4-1): The lihgtweighte aluminum alloy tubes and shafts raw materials

Polymeric materials were also purchased included the Polytetrafluoroethylene (PTFE) ribbonwith a dimensions of (L= 2000 mm and W=50 mm) as shown in the figure (4-2). PTFE is a fluorocarbon solid synthetic fluoropolymer of tetrafluoroethylene that has numerous applications. The commonly known brand name of PTFE-based formulas is Teflon . It is a high molecular weight compound consisting wholly of carbon and fluorine. PTFE is hydrophobic: neither water nor water-containing substances wet PTFE, as fluorocarbons demonstrate mitigated London dispersion forces due to the high electronegativity of fluorine. PTFE has one of the lowest coefficients of friction of any solid.

PTFE is non-reactive, partly because of the strength of carbon–fluorine bonds, and so it is often used in containers and pipework for reactive and corrosive chemicals. Where used as a lubricant, PTFE reduces friction, wear, and energy consumption of machinery.In the manufacture of the designed artificial limb, PTFE is use as a shock absorbtion material while walking or running and to increase the flexibility of the fabricated lower limb.

Figure (4-2): ThePolytetrafluoroethylene (PTFE) ribbonused for shock absorbtion and to increase the flexibility of the fabricated lower limb.

Also, two thick black Polyvinyl Chloride (PVC) sheets were purchased, with the dimensions, as shown in the figure (4-3), and the typical physical and mechanicalproperties of PVC sheets is illustrating in table (4-2).

Figure (4-3): The two thick black Polyvinyl Chloride (PVC) raw material

Table (4-2): Thetypical physical and mechanicalproperties of PVC sheets

PROPERTIES	TEST METHOD	UNIT	INTEDUR Type 1
PHYSICAL			
Thickness	ASTIM D1505	in.	1/16″ ~ 1/2″
Density	ASTM D792	g/cm^3	1.36 ~ 1.40
MECHANICAL			
Tensile Strength @ Yield	ASTM D638	psi	8,000 ~ 9,000
Elongation @ Break	ASTM D638	%	25 ~ 45
Flexural Modulus	ASTM D790	psi	380,000 ~ 430,000
Flexural Strength @ Yield	ASTM D790	psi	7,000 ~ 13,000
Izod Impact Strength (Notched)	ASTM D256	ft. lb./in.	0.8 ~ 2.5
Shore Hardness (D scale)	ASTM D2240	D	75 ~ 84
THERMAL			
Heat Deflection Temperature	ASTM D648	F	145 ~ 155
Vertical Burn Test	UL 94	–	V-O

4.2 Fabrication of the Below Knee Prosthesis Limb.

The below knee prosthesis limb fabricating processes were\accomplished by using various metal cutting machines such as turning, milling, sawing, drilling machines as well as the welding process, in addition to manual works such as filing, polishing, etc., as shown in the following paragraphs.

4.2.1 The use of Turning Machine

Engineering drawings were prepared for thebelow knee prosthesis limbparts to be fabricated using the lathe, milling and drilling machines, threads tapping and welding, as shown in the figure (4-6). Figures(4-7 and 8) illustrate the fabricatedprocesses of the prosthesis limb parts.

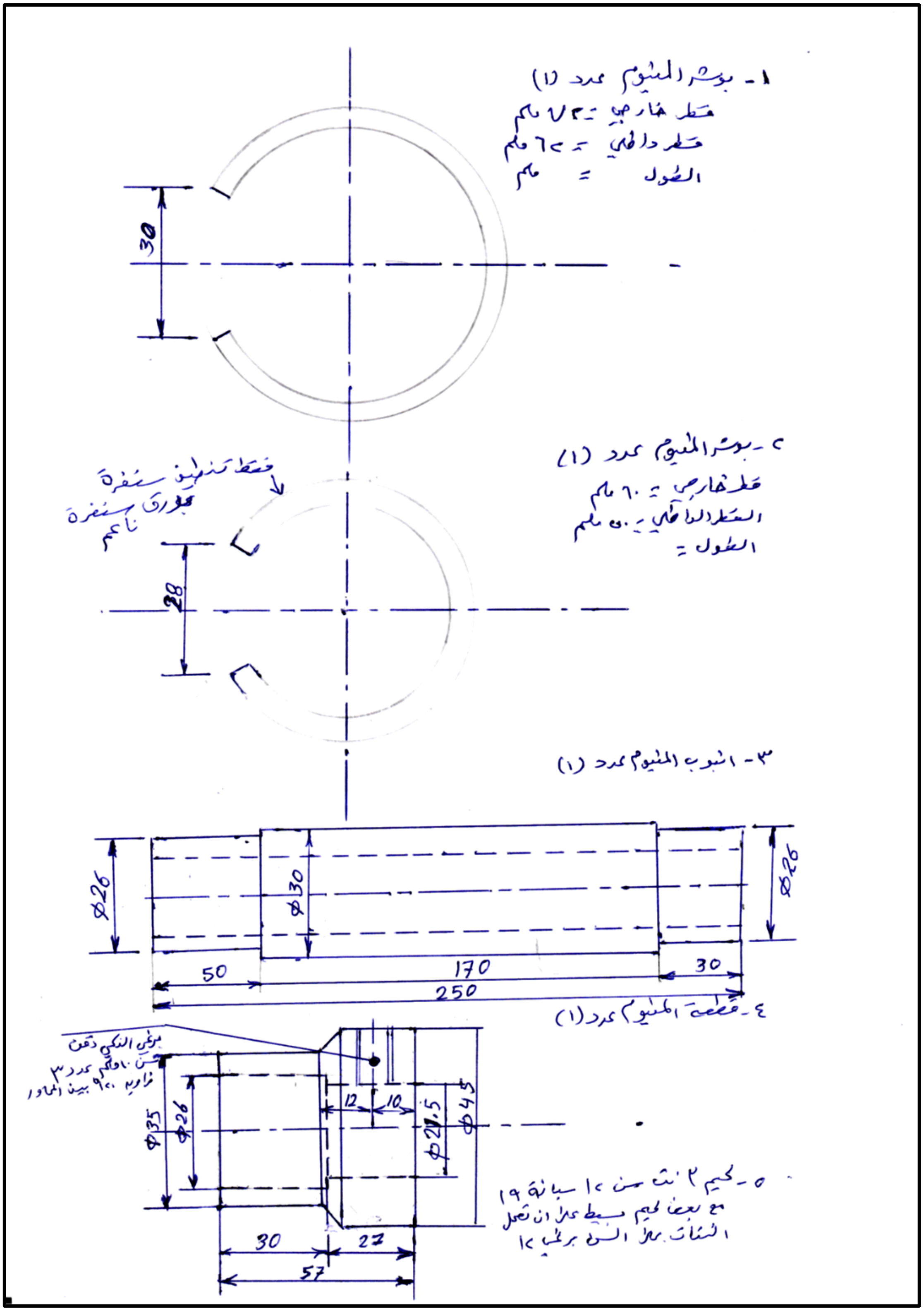


Figure (4-4): Theengineering drawings were prepared for the below knee prosthesis limb parts

Figure (4-5): Thefabricatedprosthesis limb parts

Figure (4-6): Thefabricating of the below knee prosthesis limb parts using the lathemachine

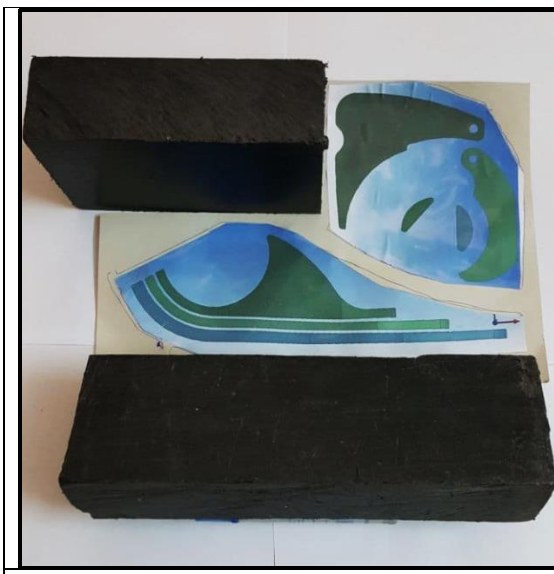
	
(a)	(b)

Figure (4-7): TheANSYS designed drawings for thefabricated prosthesis limb parts

Figure (4-7a) shows theANSYS designed drawings for thefabricated prosthesis limb parts.These drawings were pasted on the purchased Teflon plastic sheets to complete the fabricating of these parts as shown in the figure(4-9b).

(a)

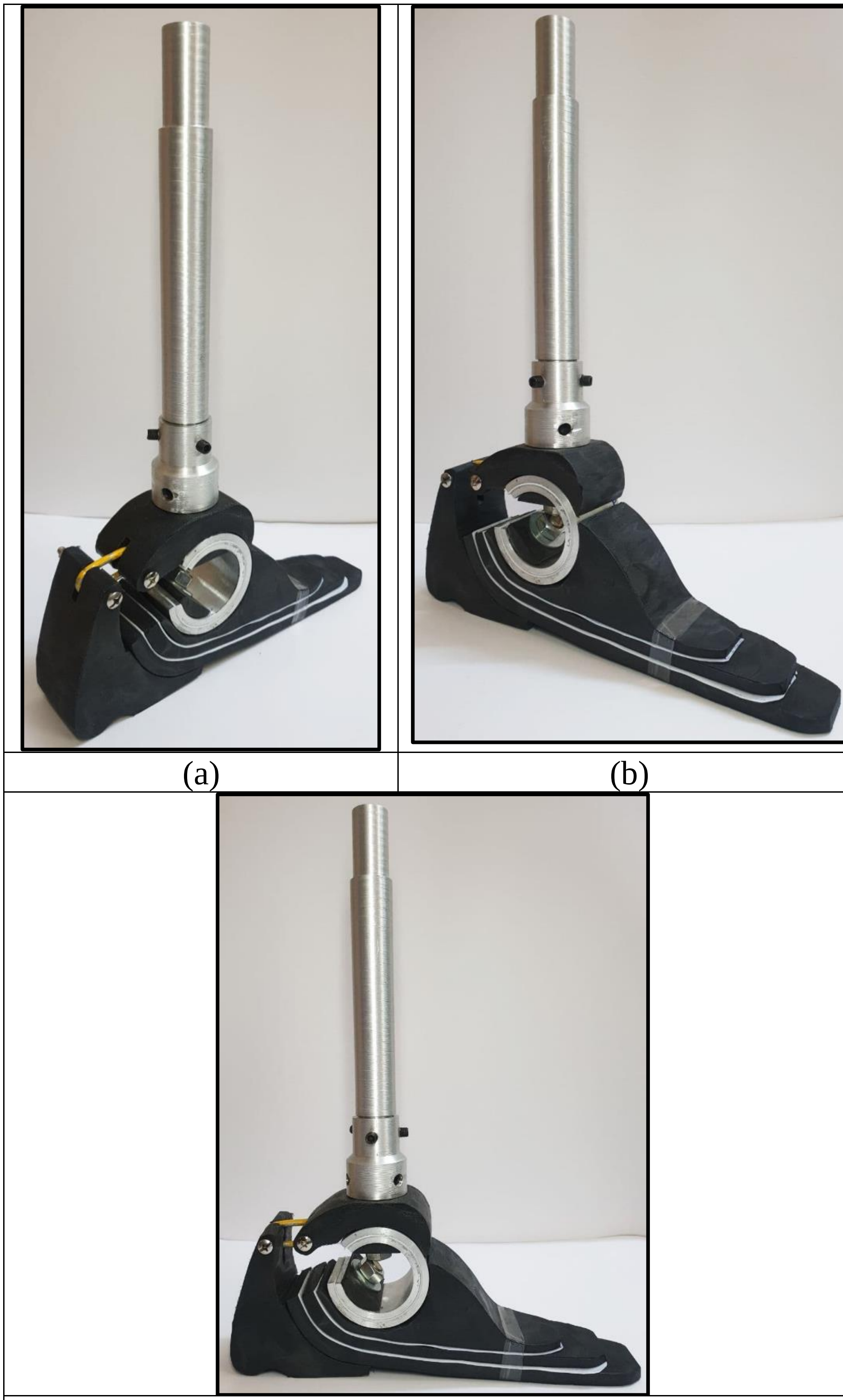
(b)

Figure (4-8): The process of cutting the parts of the artificial foot using a band saw machine.

Figures (4-8a and b) show theprocess of cutting the parts of the artificial foot using a band saw machine.After that, these parts were smoothed and polished using an electric polishing machine. The parts were initially assembled using transparent adhesive tape, and the lower and upper parts of the artificial foot were drilled using the drilling machine as shown in the figures(4-9a and b) respectively.The final assembly of the prosthetic leg parts after the completion of the drilling and threading operations are shown in the figures (4-10a, b and c).

	
(a)	(b)

Figure (4-9): The lower and upper parts of the artificial foot were drilled using the drilling machine



(a) (b)

(c)

Figure (4-10): Thefinal assembly of the prosthetic leg parts after the completion of the drilling and threading operations

CHAPTER FIVE

RESULTS AND DISSCUSSIONS

5.1 Ansys Design of theBelow Knee Prosthesis Limb

The ANSYS design and simulationof the below knee prosthesis limb were implemented by using the ANSYS 18.0 softwaremechanical design program.Ahandsketchwas implemented with the required dimensions was prepared and as shown in figure (5-1).

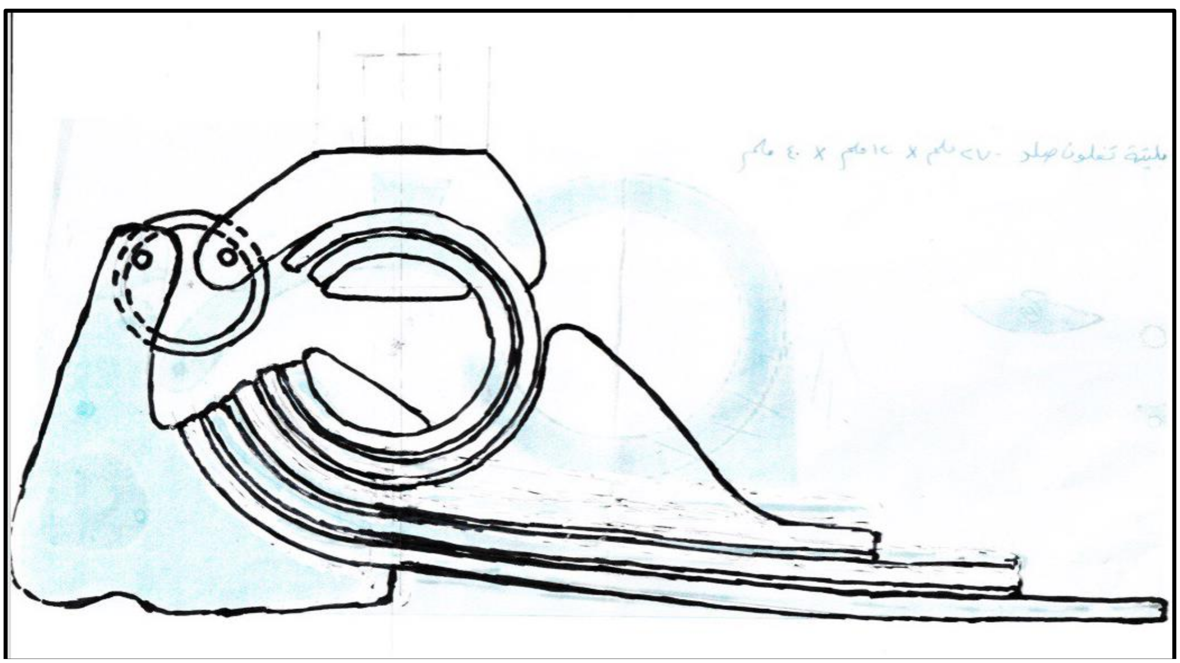


Figure (5-1): The handsketch a design of the below knee prosthesis limb

A large number of parallel and oblique planes and sketches and with a number of stages were used to complete the design of the below knee prosthesis limb using the ANSYS program as shown in the figures (5-2 to 8).

Figure (5-2)shows the below knee prosthesis limb designed using the ANSYS program, and figure (5-3)shows the second stage of the design of below knee prosthesis limb designed using the ANSYS program.

Figures (5-4 and 5)show the third stage of the design of below knee prosthesis limb designed using the ANSYS program, whilefigure (5-6)shows the forth stage of the design of below knee prosthesis limb designed using the ANSYS program.

Figures (5-7 and 8) show the complete isometric design of below knee prosthesis limb designed using the ANSYS program.

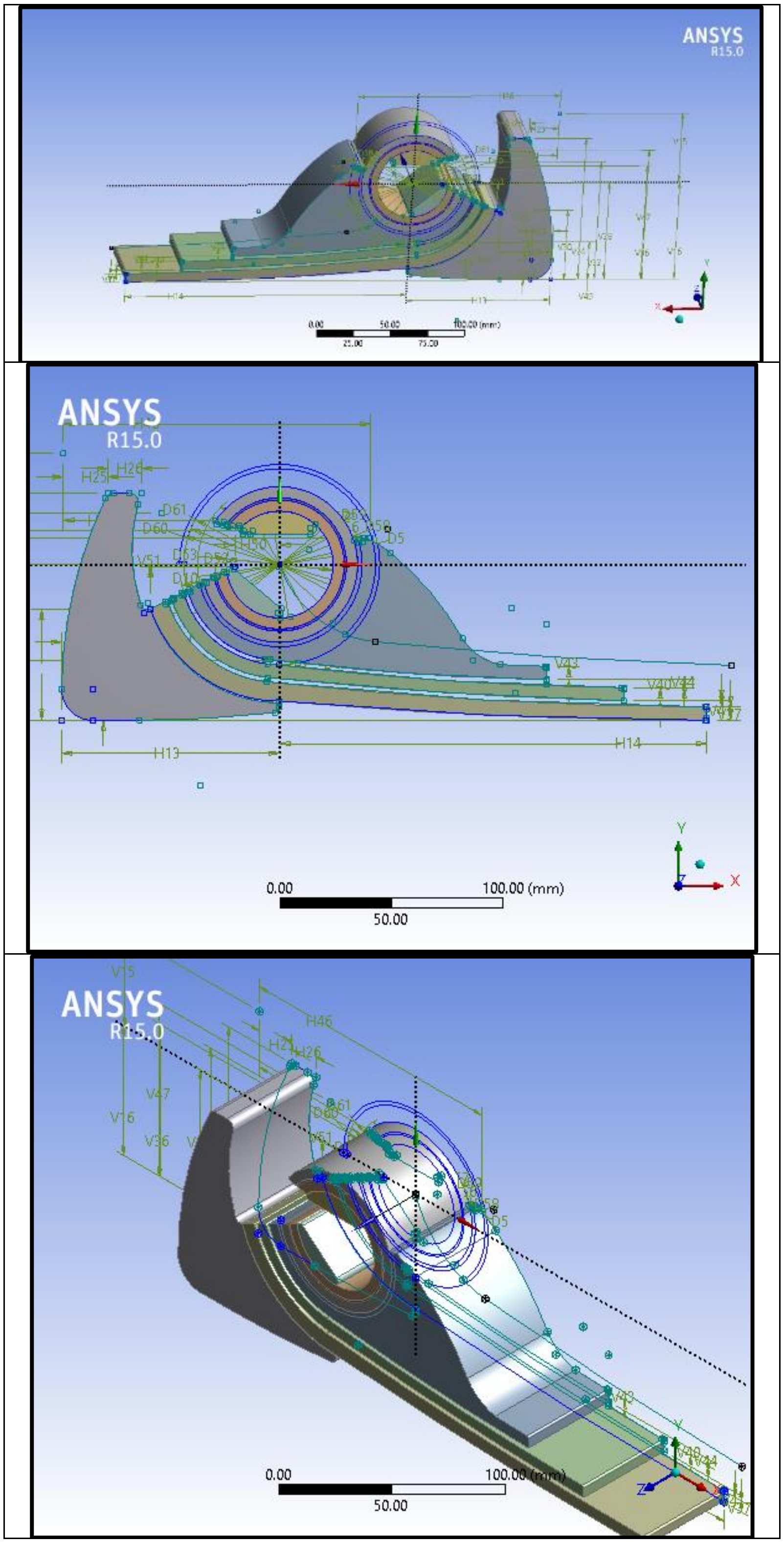


Figure (5-2): The below knee prosthesis limb designed using the ANSYS program

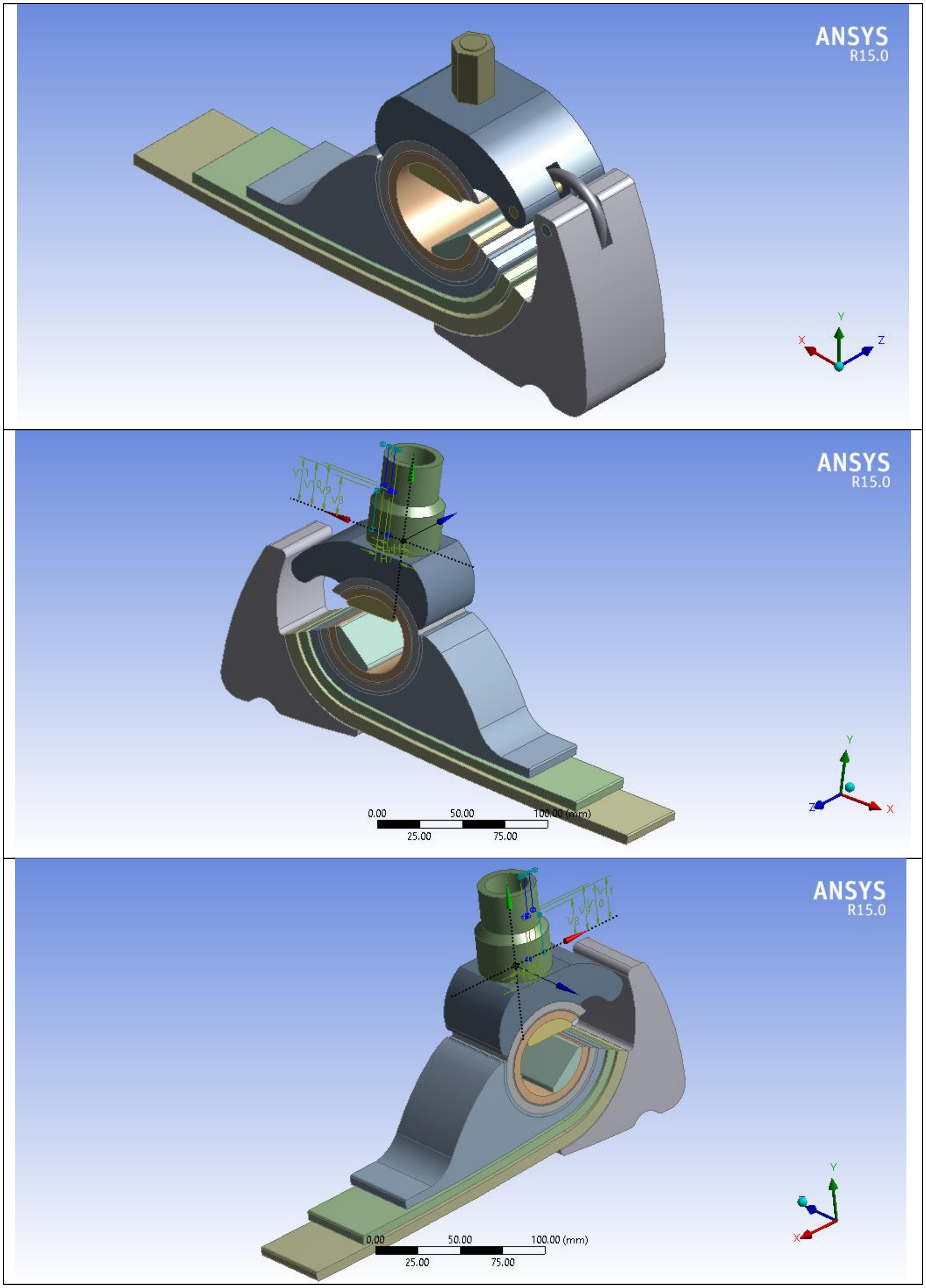


Figure (5-3): The Second stage of the design of below knee prosthesis limb designed using the ANSYS program

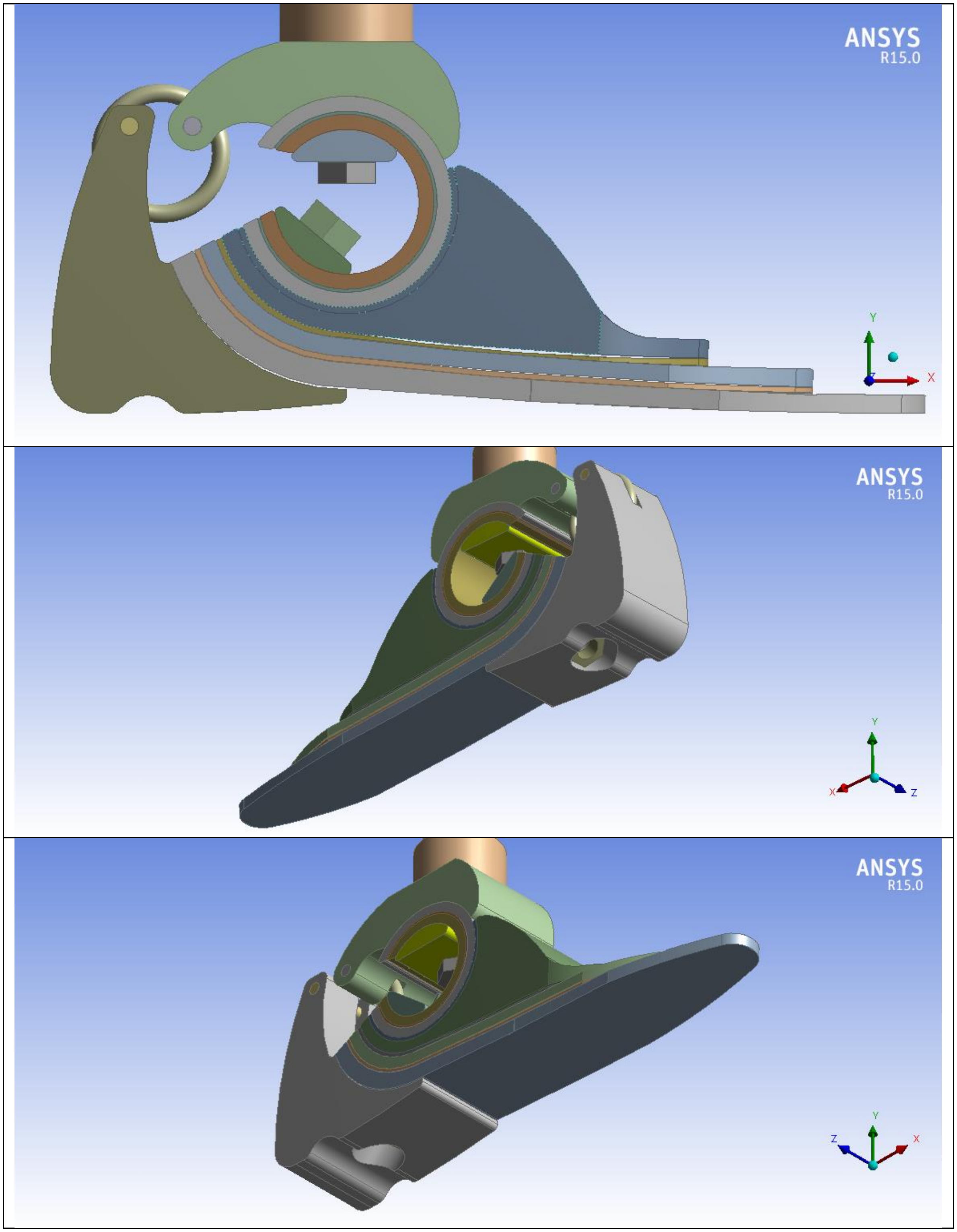


Figure (5-4): The third stage of the design of below knee prosthesis limb designed using the ANSYS program

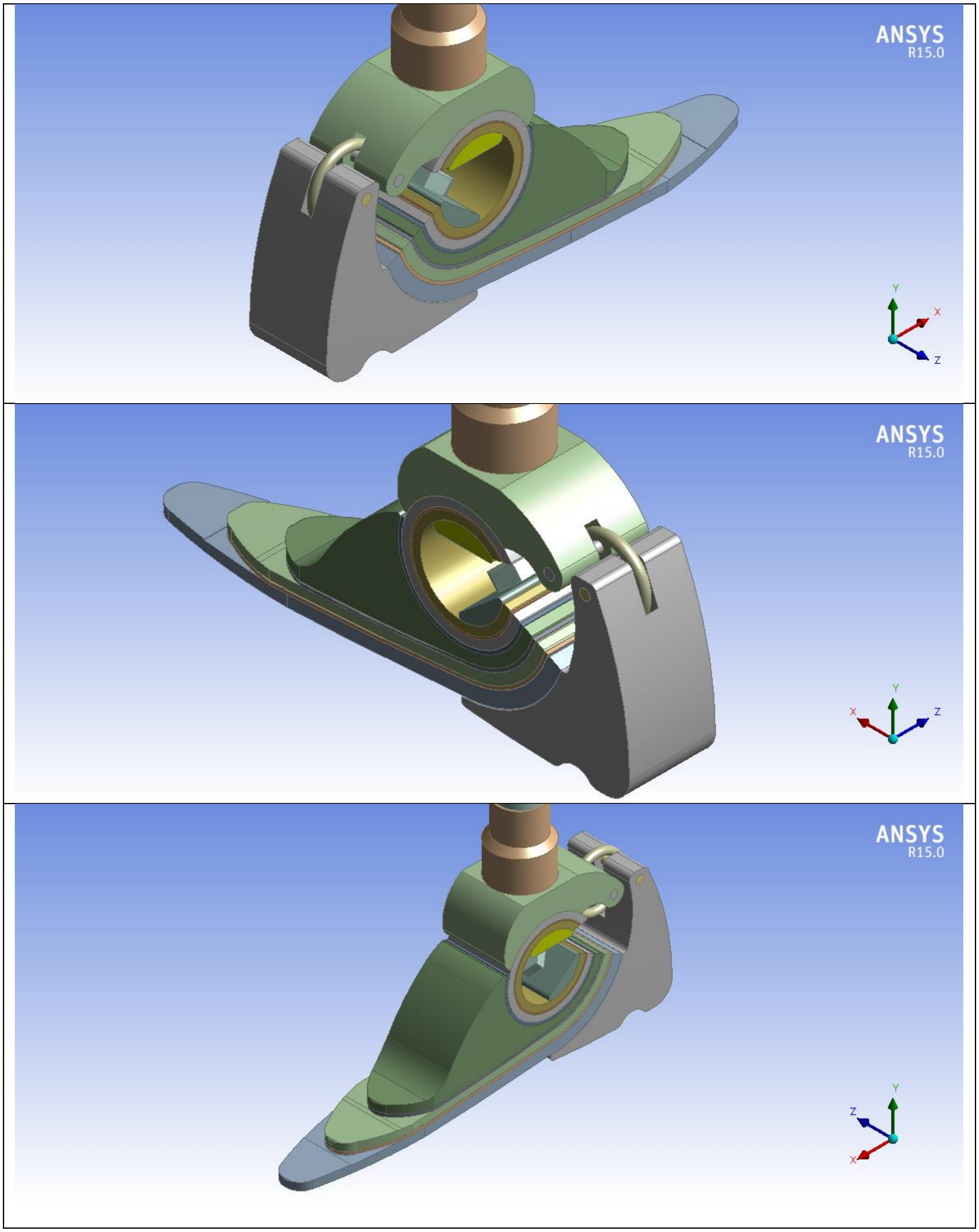


Figure (5-5): The third stage of the design of below knee prosthesis limb designed using the ANSYS program (2)

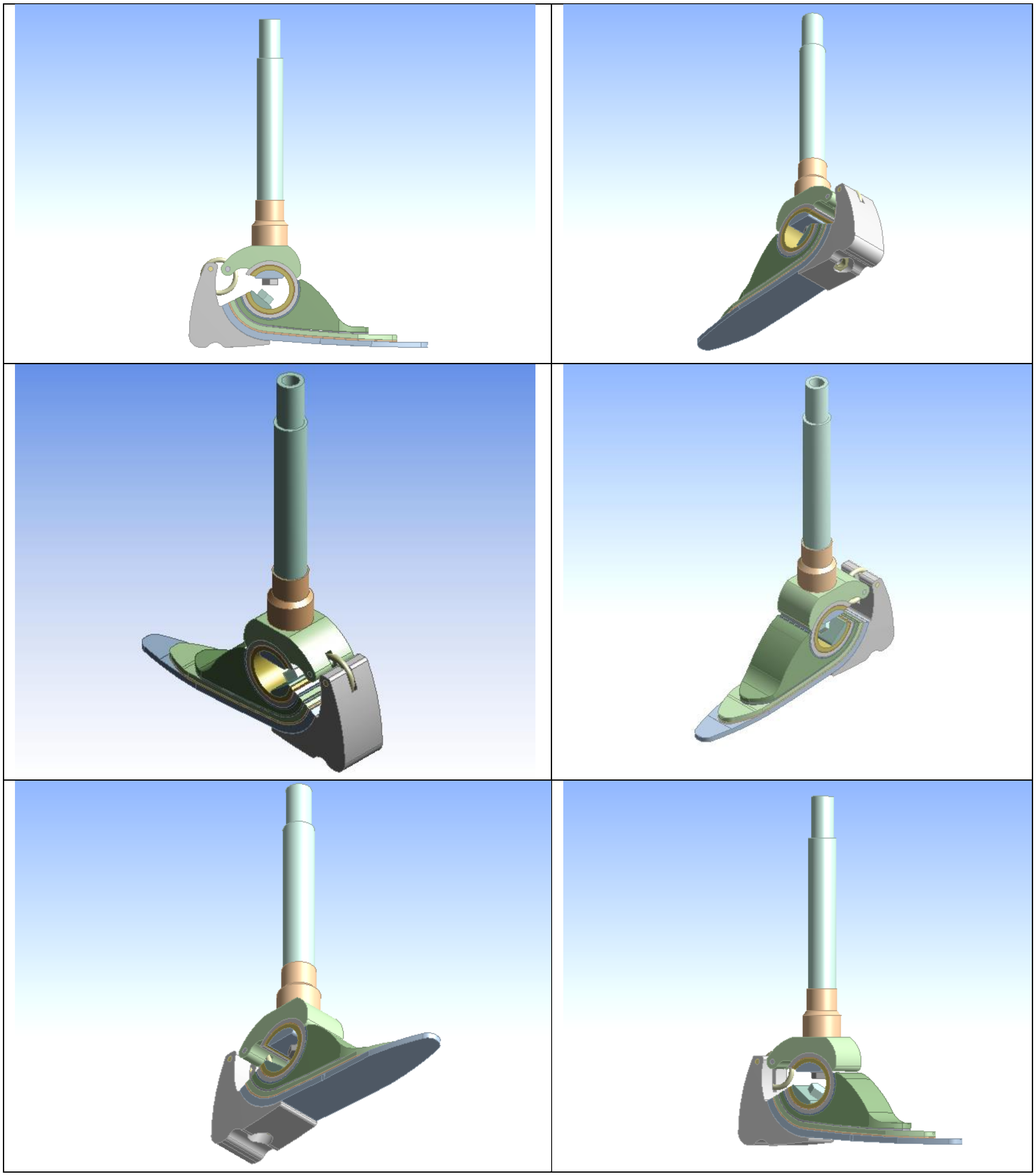


Figure (5-6): The forth stage of the design of below knee prosthesis limb designed using the ANSYS program

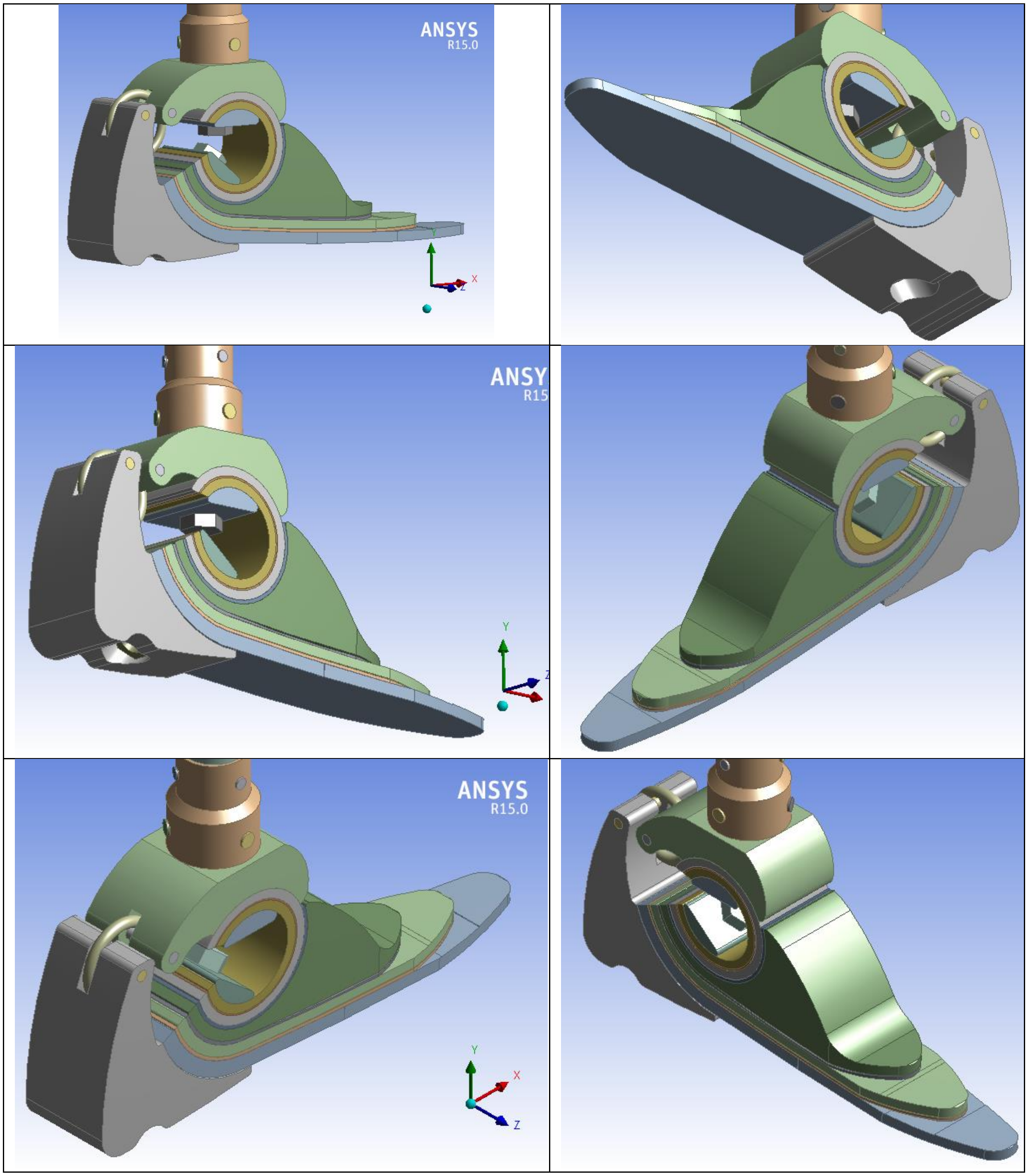


Figure (5-7): The complete isometric design of below knee prosthesis limb designed using the ANSYS program (1)

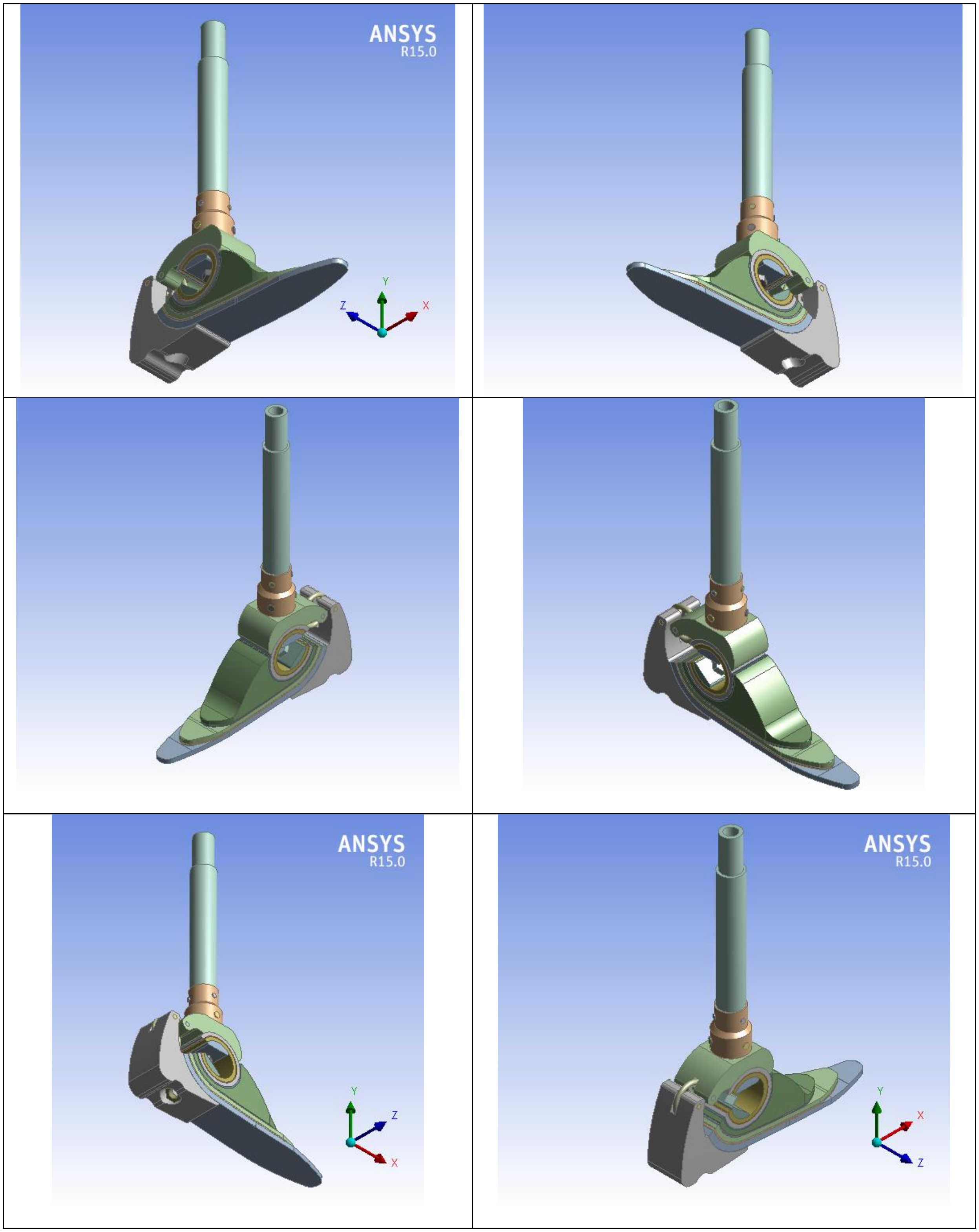


Figure (5-8): The complete isometric design of below knee prosthesis limb designed using the ANSYS program (2)

5.2 Ansys Modeling and Simulation of theBelow Knee Prosthesis Limb

For modeling and simulation the designed below knee prosthesis limb, the finite element method (FEM)was used.The designed prosthesis limb was mished to 55658 nodes and 24597 elements as shown in figure (5-9).

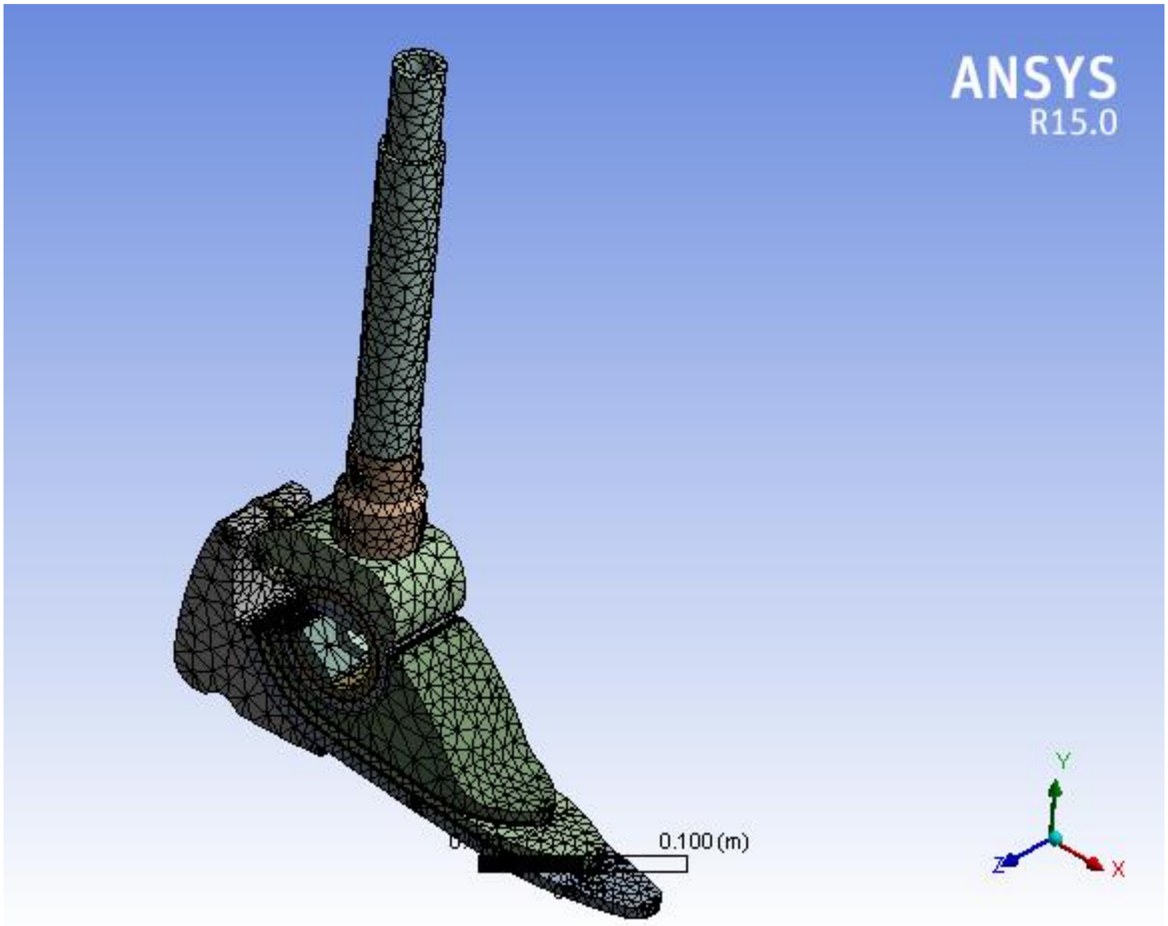
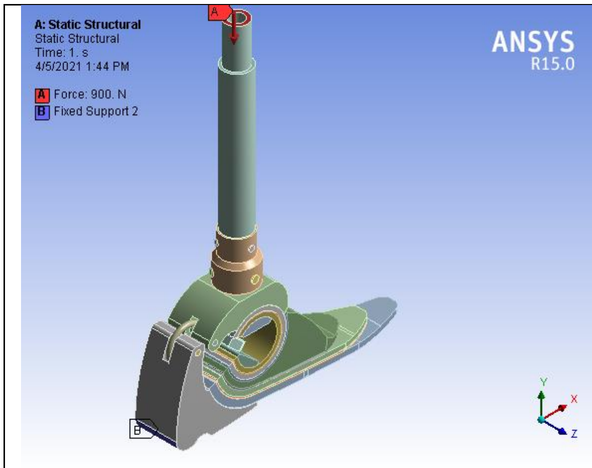
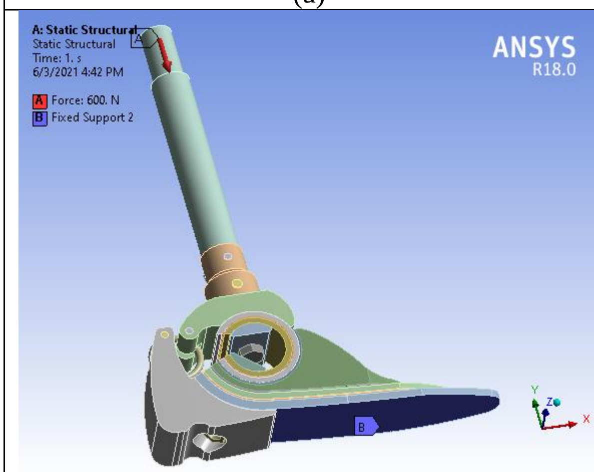


Figure (5-9): The meshed of the designed below knee prosthesis limb by using the finite element method

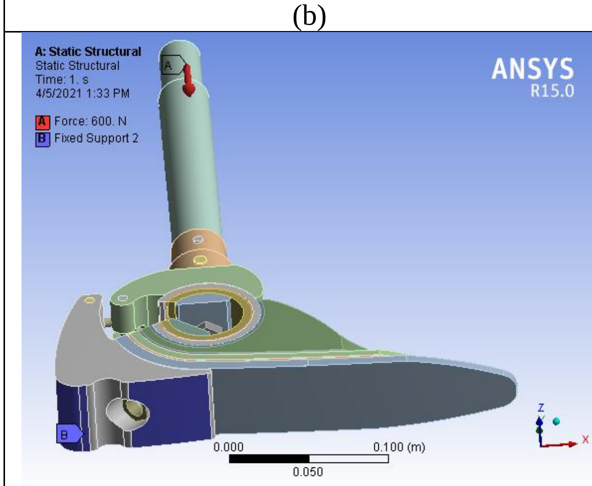
The static structural model was implemented by using threetypes of boundary conditions. The first type used when the patient is standing, where the prosthetic foot is supported on the whole lower surface of the foot, i.e., (on the



(a)



(b)



(c)

Figure (5-10): The boundary conditionsof the designed below knee prosthesis limb; (a) when the patient is standing; (b) when the patient at the beginingstep state; (c) when the patient at the end step state heel and the welt of the foot), as shown in the figure (5-10a).In the case when the patient is at the beginning step state where he is supported only on the sole and instep of the foot, as shown in the figure (5-10b).In the third case when the patient is at the begin state where he is supported only on the sole and instep end of the foot, i.e., (on the heel of the foot), as shown in the figure (5-10c).

5.3The Mechanical Properties of the FabricatedBelow Knee Prosthesis Limb

To determine the mechanical properties of the fabricated prosthesis limb for four patients were selected with total weights of 50, 75, 90 and 1200 kg, and by using a safety factor of 1.2, the patients' total weights applied to the prosthesis limb were: 600, 900, 1200 and 1500 N.The obtained mechanical properties of the fabricated prosthesis limbare given in table (5-1).

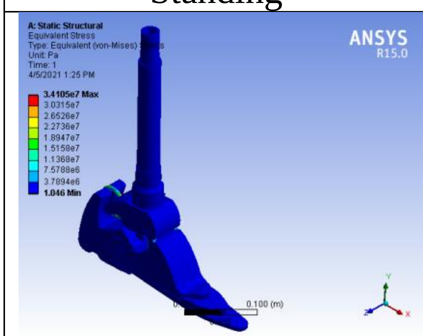
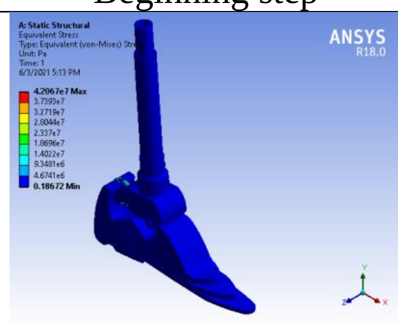
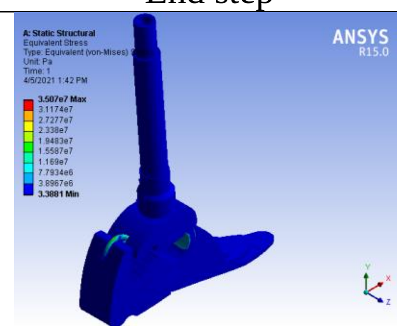
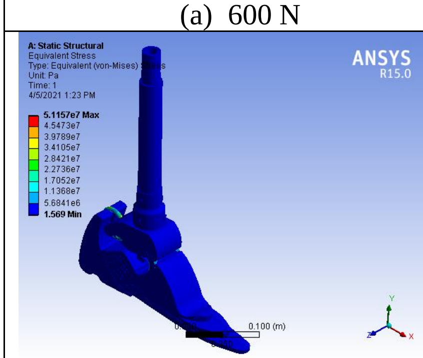
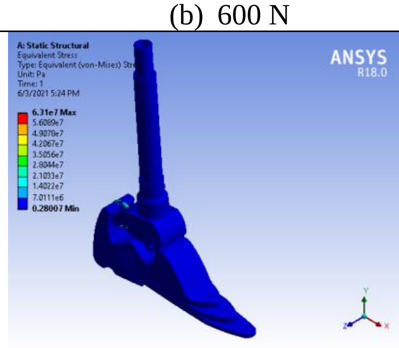
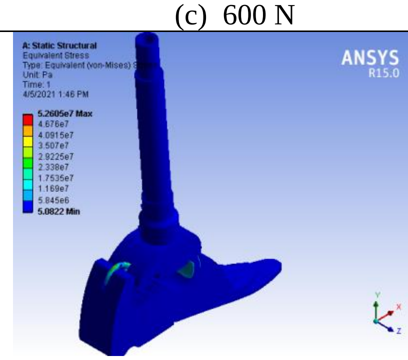
Table(5-1): The mechanical properties of the fabricated prosthesis limbfor 600, 900, 1200 and 1500 Npatients' weights for standing and at the end of walking step

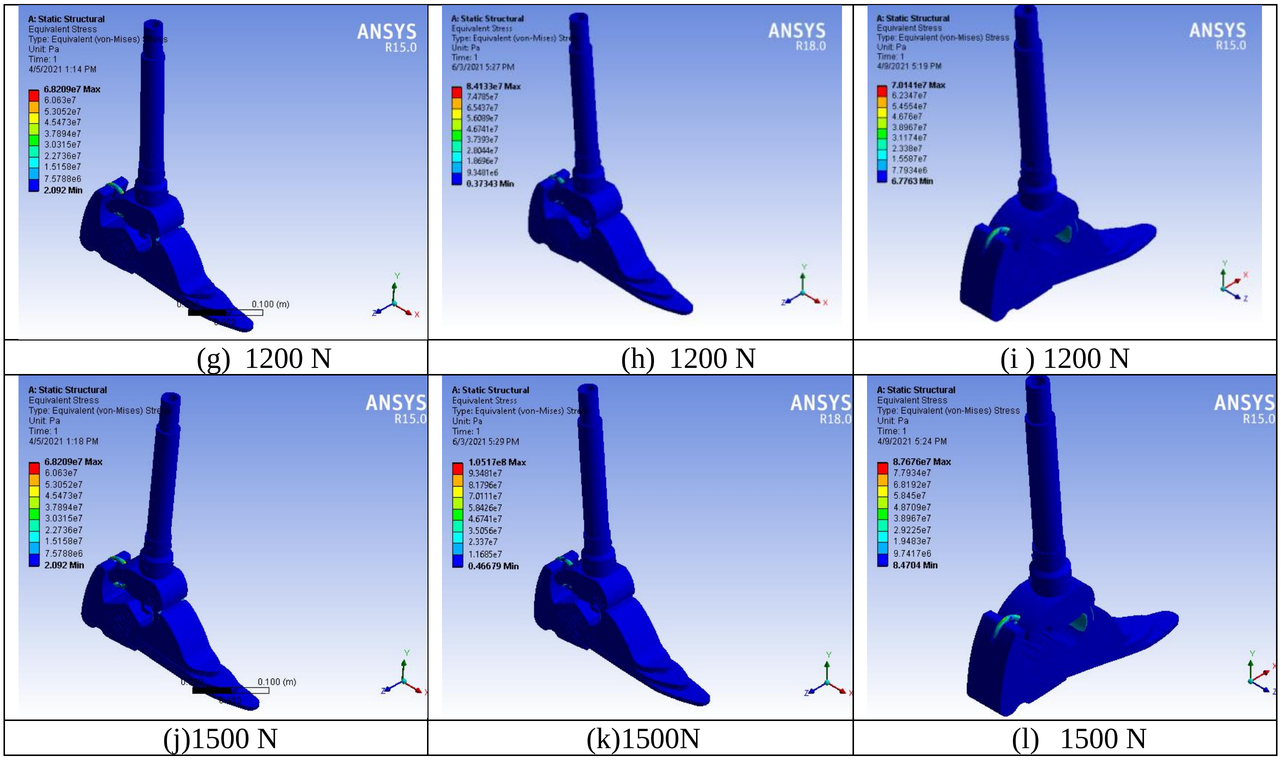
Patients' weight (N)	Patients' condition	Equivalent Von. Misses Stress (MPa)	Total Deformation (m) $X10^{-5}$	Equivalent Elastic Strain (m/m)	Strain Energy (J) $X10^{-5}$
600	Standing	34.11	6.04	0.00066	2.28
	Begin step	42.07	10.86	0.00055	1.99
	End step	35.07	5.15	0.00089	2.28
900	Standing	51.16	9.05	0.00099	5.12
	Begin step	63.10	16.29	0.00082	4.49
	End step	52.61	7.73	0.00100	5.71

1200	Standing	68.21	20.72	0.00132	8.09
	Begin step	84.13	21.72	0.00110	7.97
	End step	70.14	17.76	0.00133	10.15
1500	Standing	68.21	12.07	0.00132	14.22
	Begin step	105.17	27.14	0.00137	12.46
	End step	87.68	12.88	0.00222	15.86

5.3.1 The equivalent Von. Misses stresses

Figure (5-11) shows ANSYS sumulation results for the equivalent Von. Misses stresses of the fabricated prosthesis limbfor 600, 900, 1200 and 1500 Npatients' weightsfor standing and at the end of walking step.

Standing	Beginning step	End step
		
(a) 600 N	(b) 600 N	(c) 600 N
		
(d) 900 N	(e) 900 N	(f) 900 N



(g) 1200 N (h) 1200 N (i) 1200 N

(j)1500 N (k)1500N (l) 1500 N

Figure (5-11): Theequivalent Von. Misses stresses of the fabricated prosthesis limb for 600, 900, 1200 and 1500 N patients' weights for standing and at the end of walking step.

Figure (5-12) showsthe relationship between the equivalent Von. Misses stresses of the fabricated prosthesis limb and patients' weights which has been built using a Response Surfaces Methodology (RSM) and the statistical Expert Systems 11.0 software program.

This figure shows that, the highest equivalent Von. Misses stresses of the fabricated prosthesis limb occurs at the beginning of the step, and it reaches the highest value as (105.17) MPa at the largest patient weight of 1500 N, and this stress value is higher than the position at the end of the step by (19.91 %), and a higher e than when the patient is standing by (54.19 %).

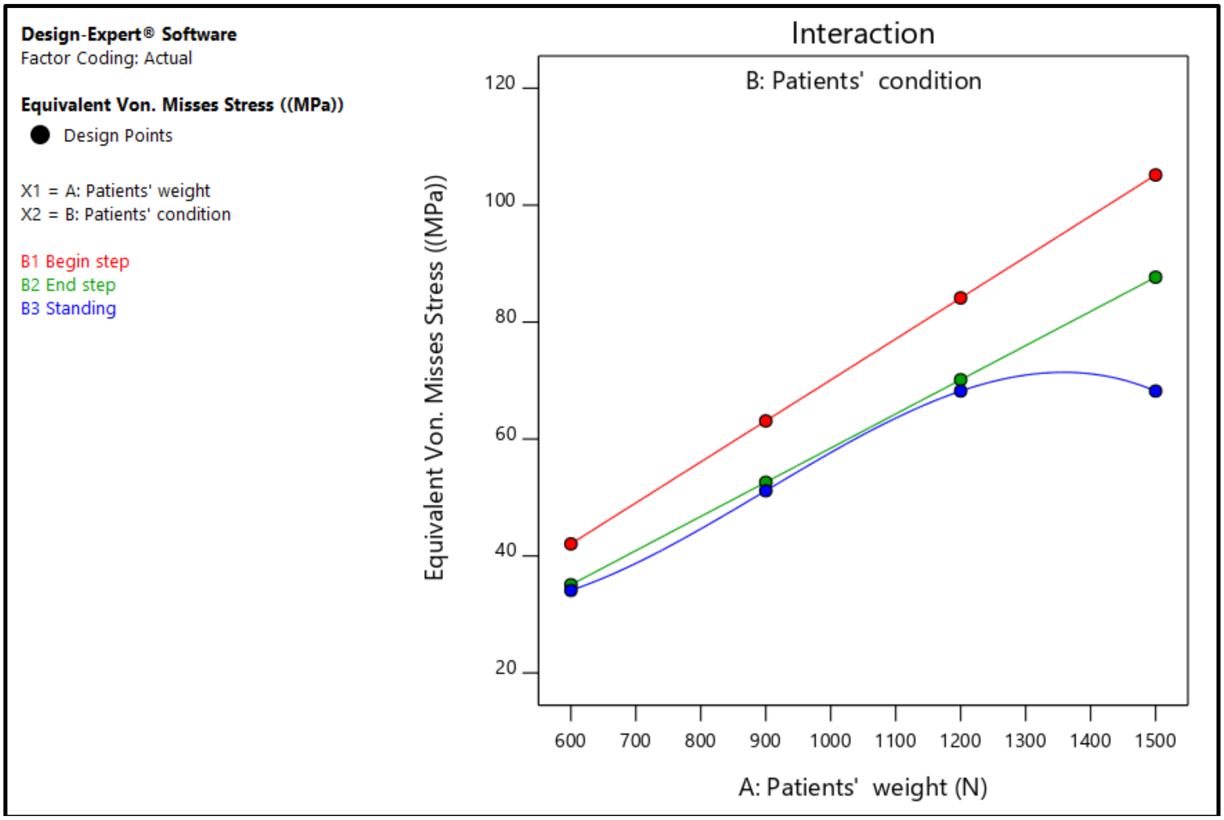
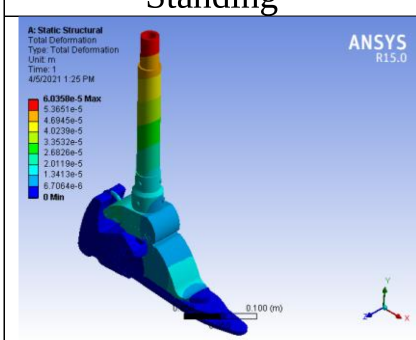
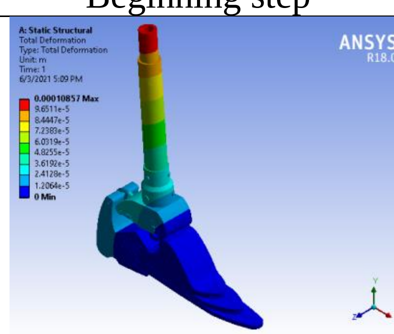
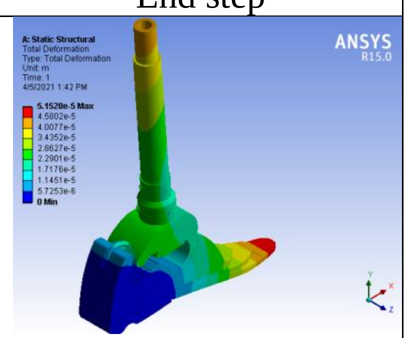
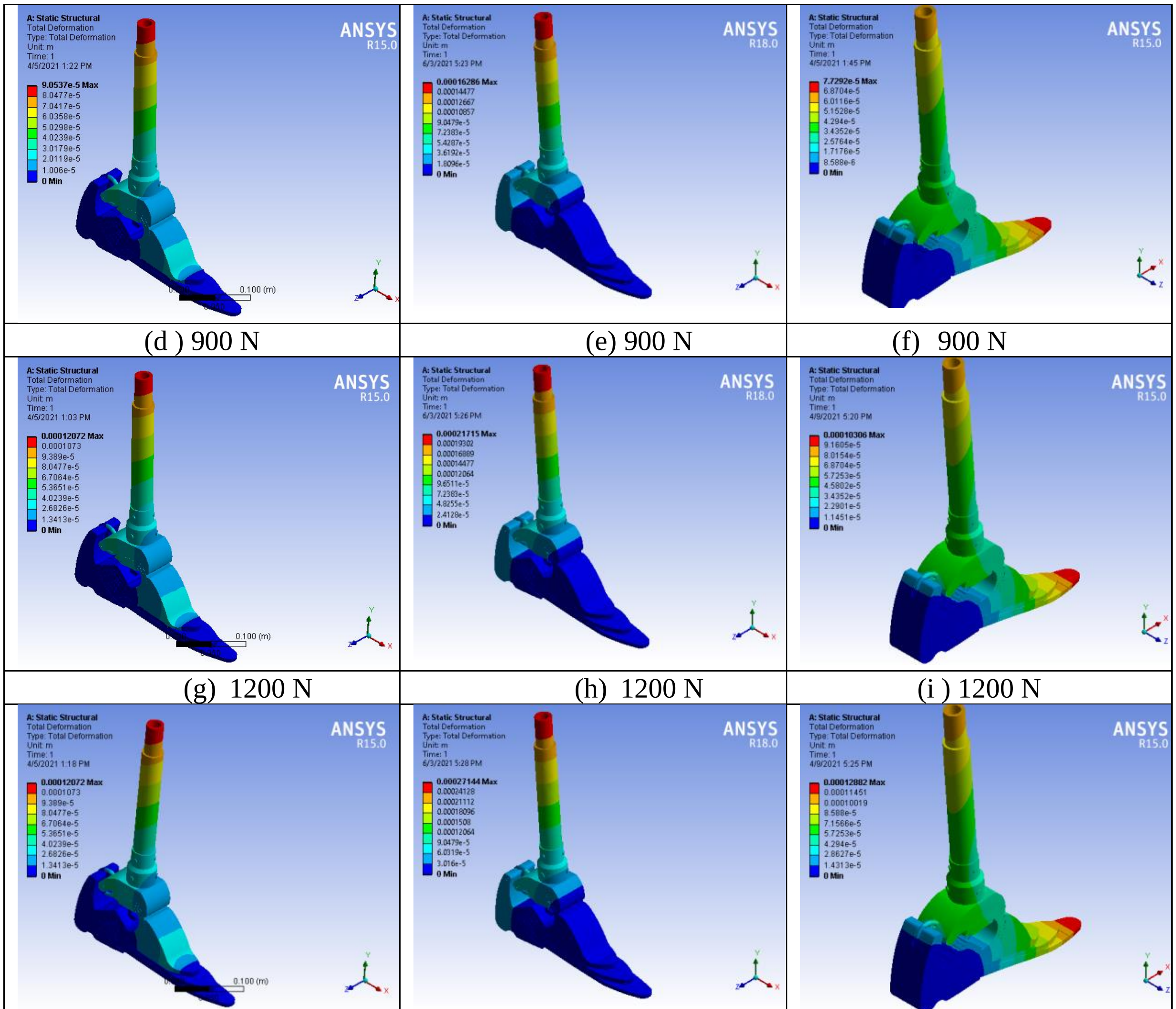


Figure (5-12): Therelationship between the equivalent Von. Misses stresses of the fabricated prosthesis limb and patients' weights

5.3.2 The Total Deformations of the Fabricated Prosthesis Limb

Figure (5-13) shows the total deformations of the fabricated prosthesis limb for 600, 900, 1200 and 1500 Npatients' weights for standing and at the end of walking step.

Standing	Beginning step	End step
		
(a) 600 N	(b) 600 N	(c) 600 N



(d) 900 N | (e) 900 N | (f) 900 N
(g) 1200 N | (h) 1200 N | (i) 1200 N
(j)1500 N | (k)1500N | (i) 1500 N

Figure (5-13): Thetotal deformations of the fabricated prosthesis limb for 600, 900, 1200 and 1500 N patients' weights for standing and at the end of walking step.

Figure (5-14) showsthe relationship between the total deformationsof the fabricated prosthesis limb and patients' weights.This figure shows that, the highest total deformationsoccurs at the beginning of the step, and it reaches the highest value as (0.271 mm) at the largest patient weight of 1500 N, and this stress value is higher than the position at the end of the step by (110.08 %), and a higher e than when the patient is standing by (123.79 %).

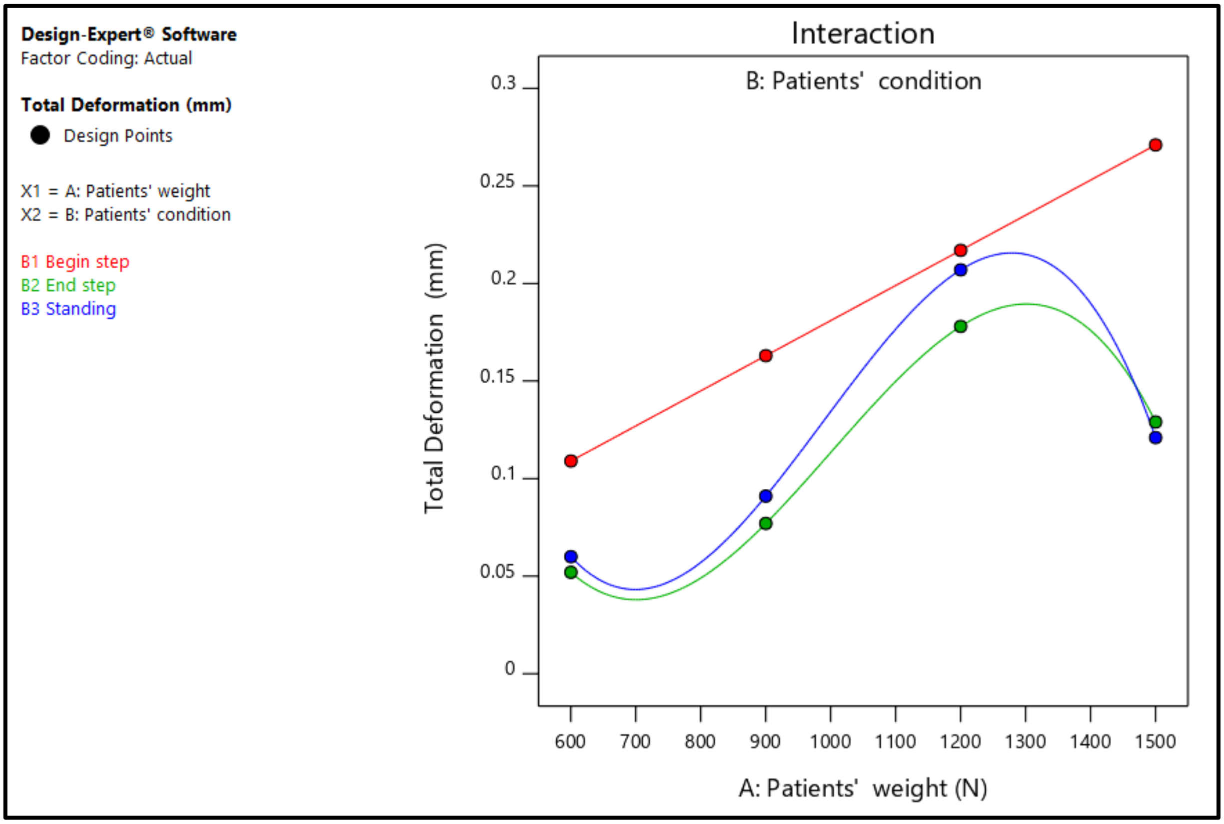


Figure (5-14): Therelationship between the total deformations of the fabricated prosthesis limb and patients' weights

5.3.3 The Equivalent Elastic Strains of the Fabricated Prosthesis Limb

Figure (5-15) shows the equivalent elastic strainsof the fabricated prosthesis limb for 600, 900, 1200 and 1500 Npatients' weights for standing and at the end of walking step.

Figure (5-16) showsthe relationship between the equivalent elastic strains of the fabricated prosthesis limb and patients' weights.This figure shows that, the highest equivalent elastic strains occurs at the end of the step, and it reachesthe highest value as (0.00222 m/m) at the largest patient weight of 1500 N, and

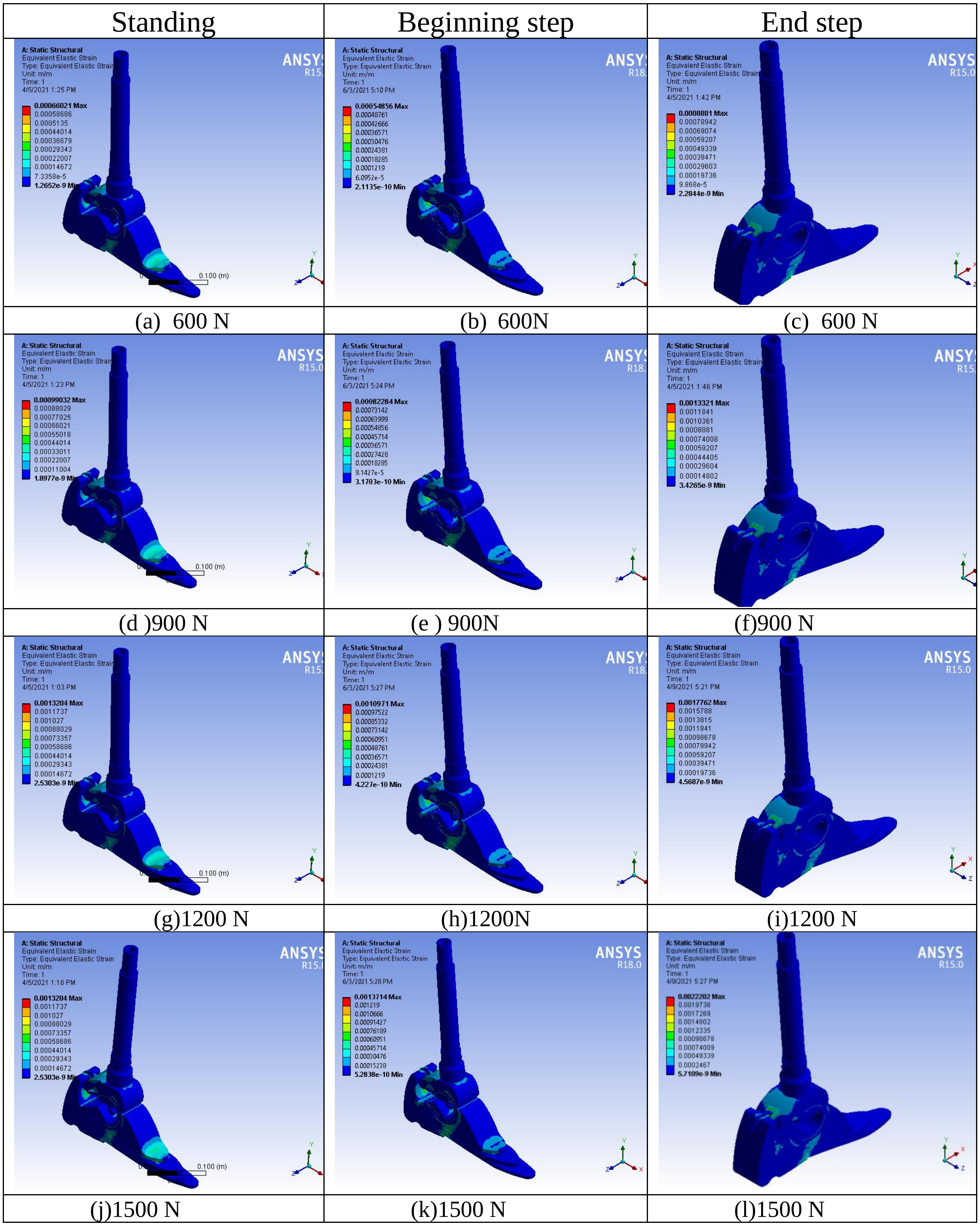


Figure (5-15): Theequivalent elastic strainsof the fabricated prosthesis limb for 600, 900, 1200 and 1500 N patients' weights for standing and at the end of walking step.

this stress value is higher than the position at the beginning of the step by (62.04 %), and a higher e than when the patient is standing by (68.19 %).

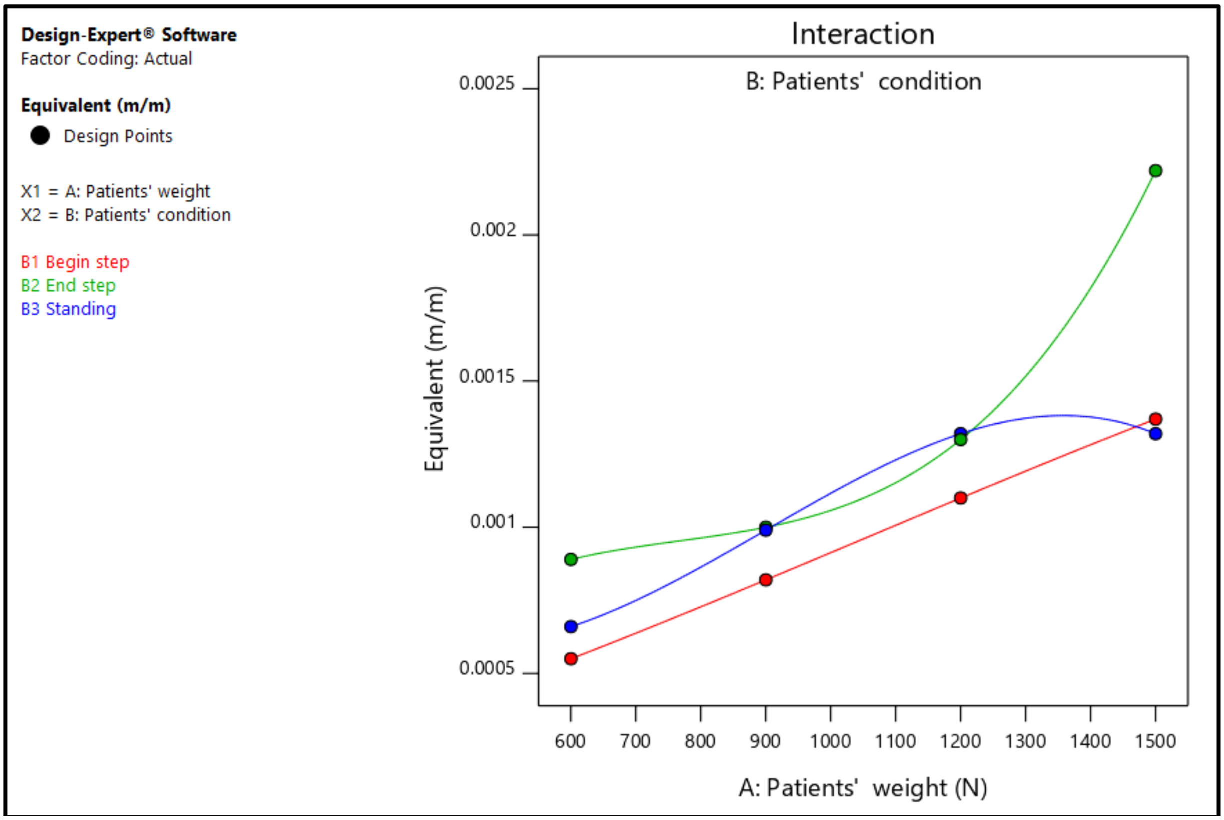


Figure (5-16): The relationship between the equivalent elastic strains of the fabricated prosthesis limb and patients' weights

5.3.4 The Strain Energy of the Fabricated Prosthesis Limb

Figure (5-17) shows the strain energyof the fabricated prosthesis limb for 600, 900, 1200 and 1500 Npatients' weights for standing and at the end of walking step.

Figure (5-18) showsthe relationship between the strain energy of the fabricated prosthesis limb and patients' weights.This figure shows that, the highest strain energy occurs at the end of the step, and it reaches the highest value as (0.159 mJ) at the largest patient weight of 1500 N, and this stress value

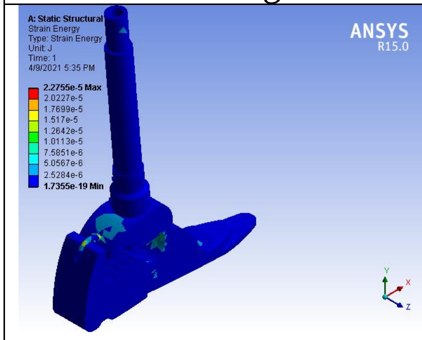
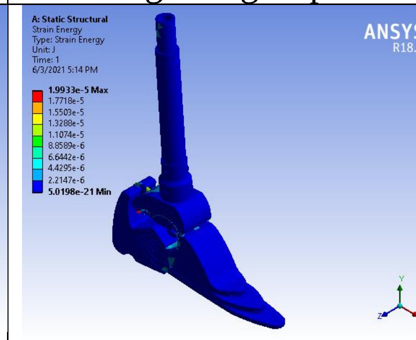
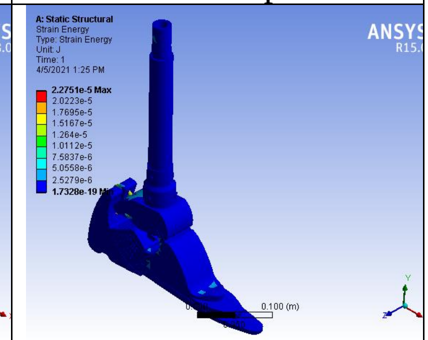
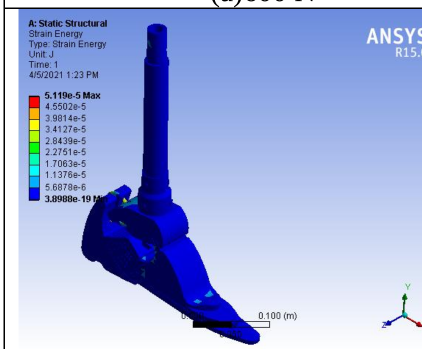
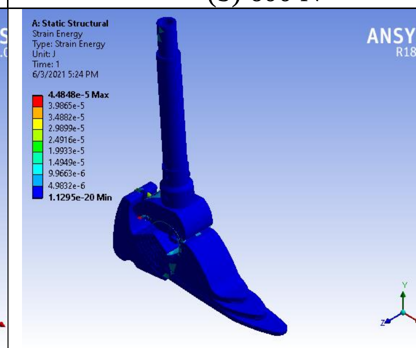
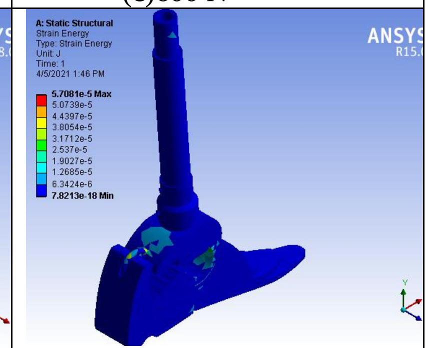
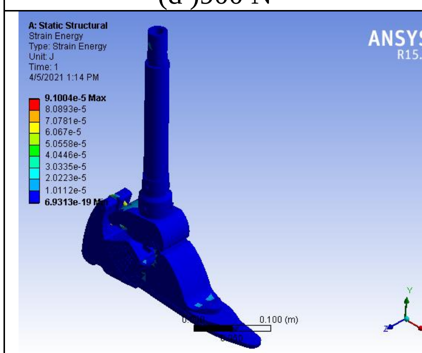
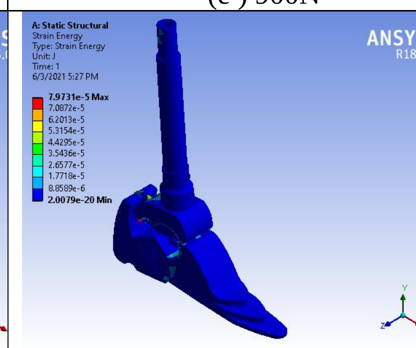
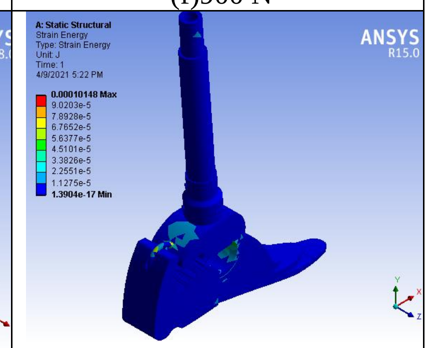
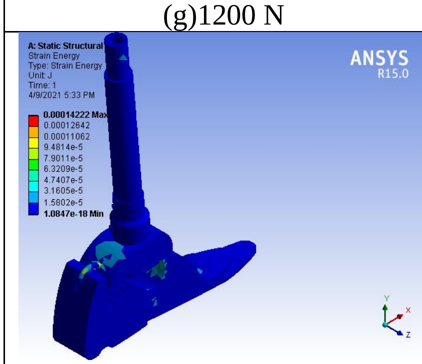
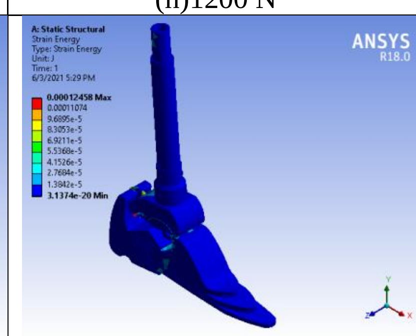
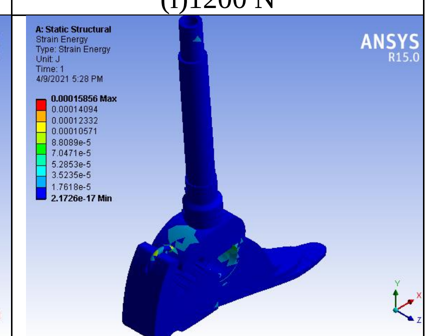
Standing	Beginning step	End step
		
(a)600 N	(b) 600 N	(c)600 N
		
(d)900 N	(e) 900N	(f)900 N
		
(g)1200 N	(h)1200 N	(i)1200 N
		
(j)1500 N	(k)1500 N	(l)1500 N

Figure (5-17): Thestrain energyof the fabricated prosthesis limb for 600, 900, 1200 and 1500 N patients' weights for standing and at the end of walking step.

is higher than the position at the of the standing step by (11.97 %), and a higher e than when the patient is at the beginning step by (27.20 %).

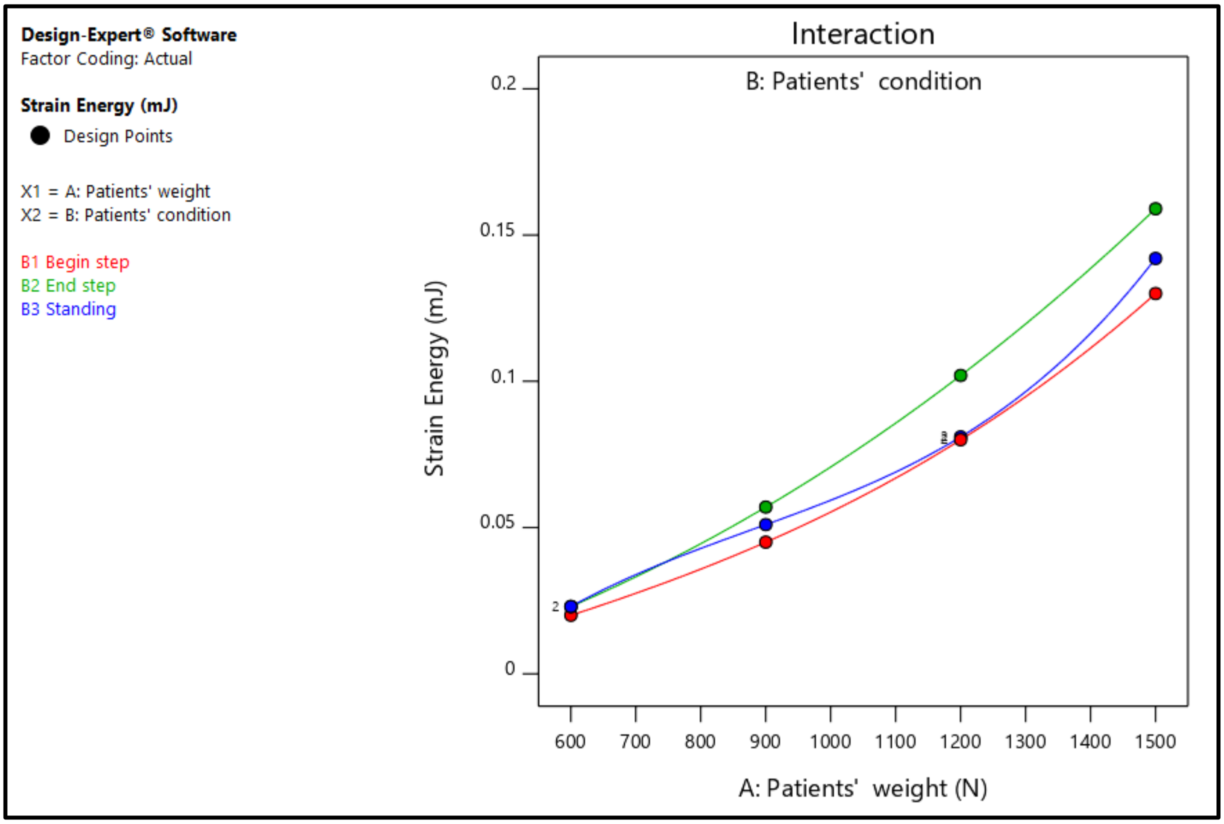


Figure (5-18): The relationship between the strain energy of the fabricated prosthesis limb and patients' weights

CHAPTER SIX

CONCLUSIONS AND SUGGESTINGS FOR FUTURE WORK

6.1 Conclusions

The main conclusions of this project canbe summarized in the followings:

1- Design and fabrication of a comfortable prosthesis, light weight, durable, low-cost novel below- knee prosthesis limb.

2- The below knee prosthesis limb fabricating processes were accomplished by using various metal cutting machines such as turning, milling, sawing, drilling machines as well as the welding process, in addition to manual works such as filing, polishing, etc.

3- The ANSYS design and simulationof the below knee prosthesis limb were implemented by using the ANSYS 18.0 softwaremechanical design program by using a large number of parallel and oblique planes and sketches and by using the finite element method (FEM).

4- The static structural model was implemented by using three types of boundary conditions (when the patient is standing, at the beginning of the step and at the end of the step).

5- To determine the mechanical properties of the fabricated prosthesis limb four patients were selected with total weights of 50, 75, 90 and 1200 kg, and by using a safety factor of 1.2, the patients' total weights applied to the prosthesis limb were: 600, 900, 1200 and 1500 N. The obtained mechanical properties have been built by using the Response Surfaces Methodology (RSM) and the statistical Expert Systems 11.0 software program.

6- The highest equivalent Von. Misses stresses of the fabricated prosthesis limb occur at the beginning of the step, reached (105.17) MPa at the highest patient weight of 1500 N, which is higher than the position at the end of the step by (19.91 %), and a higher e than when the patient is standing by (54.19 %).

7- The highest total deformations occur at the beginning of the step, reaches (0.271 mm) at the patient weight of 1500 N, which is higher than the position at the end of the step by (110.08 %), and a higher e than when the patient is standing by (123.79 %).

8- The highest equivalent elastic strains occur at the end of the step, reaches (0.00222 m/m) at the patient weight of 1500 N, which is higher than the position at the beginning of the step by (62.04 %), and a higher e than when the patient is standing by (68.19 %).

9- The highest strain energy occur at the end of the step, reaches the highest value as (0.159 mJ) at the patient weight of 1500 N, which is higher than the position at the of the standing step by (11.97 %), and a higher e than when the patient is at the beginning step by (27.20 %).

6.2 Suggesting for Future Works

1-Further work is required to design, develop low-cost for other models to suit all cases of lower limb amputation conditions and for all ages, weights and levels of amputation with high sensitivity and specificity.

1- Further work is required to design, develop and a highly controlled and intelligent prosthetic leg that helps restore the overall functions of patients of various professions, with adding a high aesthetic appearance and ensuring the highest levels of patient comfort.

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الخلاصة

تستخدم الأطراف الاصطناعية الذكية لتحل محل الأطراف المبتورة في مجال إعادة التأهيل البيولوجي. تتمثل الأهداف الرئيسية لهذا المشروع في تصميم وتصنيع طرف اصطناعي مريح وخفيف الوزن ودائمي ومنخفض التكلفة تحت الركبة لمساعدة المرضى على أداء الأنشطة اليومية وتوفير الراحة لمبتوري الأطراف السفلية. تم إنجازالطرف الصناعي باستخدام العديد من مكائن وطرق قطع المعادن ، بالإضافة إلى الأعمال اليدوية مثل البرادة والصقل وما إلى ذلك. وتم تنفيذ التصميم والمحاكاة باستخدام برنامج ANSYS 18.0 للتصميم الميكانيكي وباستخدام عدد كبير من المستويات والمخططات المتوازية والمائلة وعن طريق استخدام طريقة العناصر المحدودة (FEM).

ولتحديد الخواص الميكانيكية للطرف الاصطناعي المصطنع تم اختيار أربعة مرضى بأوزان إجمالية 50 و 75 و 90 و 120 كجم. وتم بناء الخواص الميكانيكية التي تم الحصول عليها باستخدام منهجية الاستجابة السطحية (RSM) وبرنامج برنامج الانظمة الخبيرة الإحصائي Expert Systems 11.0. وتم الحصول على اعلى اجهاد مكافئ بطريقة فان مايسز ggg Von Misses طرف الاصطناعي المصنع وعند بداية الخطوة ، وبلغت (105.17) ميجا باسكال عند أعلى وزن للمريض 1500 نيوتن ، وهو أعلى من الموضع في نهاية الخطوة بنسبة (19.91٪)، وأعلى من حالة وقوف المريض بنسبة (54.19٪) ، وحصلت أعلى قيمة للتشوهات الكلية في بداية الخطوة ايضا ،وبلغت (0.271 ملم) فقط عند اقصى وزن للمريض وهو 1500 نيوتن ، وهذه القيمة هي أعلى من الوضع عند نهاية الخطوة بنسبة (110.08٪) . كما تم الحصول على اعلى قيمة لمعامل المرونة المكافئ في نهاية الخطوة وبلغت (0.00222 م / م). وان أعلى طاقة إجهاد مرنة تم الحصول عليها في نهاية الخطوة ، وبلغت (0.159 مللي جول). مما يتيح الاستنتاج بان القدم الصناعية المصنعة المطورة والجديدة قريبة من جميع الخواص الميكانيكية للقدم الطبيعية.

وزارة التعليم العالي والبحث العلمي

الجامعة التكنولوجية
قسم هندسة الطب الحياتي
فرع الميكانيك الاحيائي

"تصميم وتصنيع طرف اصطناعي تعويضي محسن أسفل الركبة "

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