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To those who taught me that science is a message, and that the pursuit of knowledge is endless…

To my family who have always been my source of support and inspiration…

To my teachers who instilled in me a love of research and discovery…

And to every researcher who seeks to enrich knowledge and serve humanity, I dedicate this work. I extend my sincere thanks and appreciation to everyone who contributed to the completion of this scientific work, whether through academic support or continuous encouragement.

I would like to thank my professors and colleagues in the academic field, who did not skimp on their advice and guidance, and I also express my deep gratitude to the scientific research institutions that provided me with the resources and capabilities necessary to complete this project.

I will not forget my family and friends, who supported me during this scientific journey, without their continuous support, this work would not have seen the light.

Contents

Abstract

Certain individuals or animals may necessitate the fitting of a prosthetic limb above the knee or the implantation of an artificial hip joint due to amputation, which can arise from various factors such as congenital anomalies, chronic illnesses, or severe injuries that lead to the necrosis of leg tissues and damage to adjacent arteries. The healing process typically requires a duration of 2 to 3 months following the amputation, owing to the sensitivity of the residual limb.

Lower limb amputations are predominantly the result of accidents or medical conditions. In our research, we have identified PETG as a suitable material for the production of prosthetics. This material is particularly advantageous due to its affordability, unique mechanical properties, and favorable safety factor. It is lighter and more convenient compared to other high-pressure materials, which tend to be heavier. In this study, we concentrated on developing the optimal design for the lower limb (shank) prosthesis by examining various cross-sectional shapes using carbon fiber. We designed samples of this material and conducted tensile tests to ascertain its mechanical properties. Additionally, we utilized the Ansys program to model and analyze the Finite Element Method, comparing the different cross-sections. By applying two weights, we calculated the effects of the imposed forces in terms of deformation, stress, and safety factor. The results indicated that the circular and elliptical sections exhibited similar performance across all metrics, with only a negligible difference. Conversely, the hexagonal section demonstrated the least resistance to stress and experienced significant deformation compared to the other shapes, suggesting that circular or elliptical designs are preferable for prosthetic applications.

Keywords (Prosthetic shank. Carbon fiber material. Lightweight and durability. Biomechanics. Human gait analysis).

Chapter 1 Introduction

Chapter One
1.1 Introduction

Prosthetics are artificial devices designed to replace lost body parts, such as hands and legs. The selection of a specific type of prosthesis is influenced by the method of amputation and the characteristics of the missing limb. The motivations for physicians to recommend prosthetic installation for patients are diverse; they may stem from medical conditions that necessitate amputation, congenital defects, or injuries resulting from accidents. Research conducted by Swiss scientists has indicated that the ancient Egyptians were pioneers in the development of prosthetic limbs that were appropriately sized and comfortable for users. Presently, researchers are focused on advancing smart prosthetic limbs that are equipped with artificial skin capable of conveying sensations such as temperature, thereby mimicking the functionality of natural limbs and providing users with a sense of compensation for their loss **{ 1 } .**

While joints and prosthetics enhance mobility for individuals who may experience restlessness following the loss of a limb due to an accident or medical complications, they are not without their limitations. In particular, prosthetic limbs are often associated with significant costs, and some patients may find it challenging to adapt to their use. Additionally, these devices require regular maintenance and repairs, and certain prosthetic joints necessitate extensive and invasive surgical procedures for implantation, often resulting in restricted movement, as seen with prosthetic knees.

The invention of the prosthetic leg dates back to November 1846. Prior to Palmer's innovation, individuals utilized wedge legs, which can be observed in historical images. This advancement has enabled a more natural range of motion compared to the earlier wedge designs. Prosthetic legs empower many individuals to maintain an active lifestyle, with some even participating in competitive sports such as track and field and marathons. It is noteworthy that Benjamin Franklin Palmer not only invented but also patented this significant development, which continues to evolve today. **{ 2 } .**

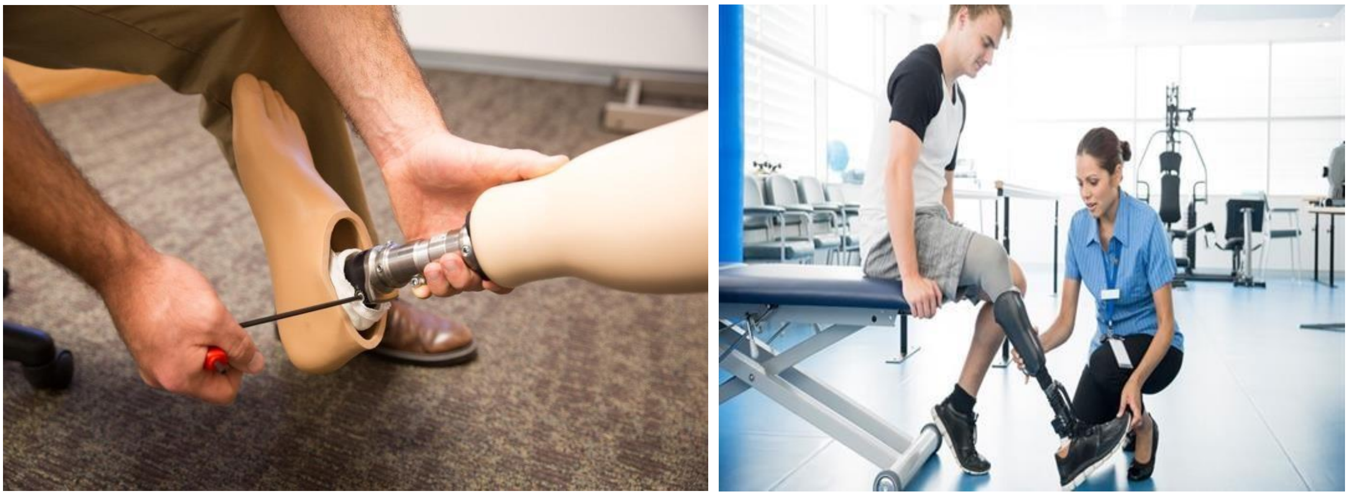


Fig (1.1) illustrates lower prostheses

1.2 A Brief History Of The Industrial Parties: -

The history of humanity, marked by numerous conflicts, has seen countless individuals sustain injuries resulting in the loss of limbs. Over the centuries, efforts have been made to address these disabilities in various ways. The use of prosthetic limbs can be traced back to ancient times, with one of the earliest documented instances involving Queen Vishpala, as referenced in the Rigveda. This ancient Indian text recounts how the queen lost her foot during battle and subsequently had iron rods fitted to enable her to walk and return to combat. Additionally, the ancient Egyptians were pioneers in the application of prosthetic devices, as archaeological discoveries have revealed wooden fingers that date back to the period of the modern Egyptian kingdom **{3 } .**

Fig (1.2) shows artificial fingers made in ancient Egypt

1.3 Amputation: -

Amputation is often regarded as the most significant failure in surgical practice; however, it is, in fact, a carefully planned and performed reconstructive intervention. The procedure involves the removal of a portion or the entirety of a limb. Amputations are undertaken only after all other treatment options have been thoroughly assessed and deemed unsuitable, with limb removal identified as the most appropriate course of action for the patient. Various factors can necessitate amputation, including traumatic injuries, vascular issues, infections, malignant tumors, and congenital limb deficiencies **{ 4 } .**

1.3.1 Amputation types and levels: -

There are two primary categories of amputation: upper and lower. The procedure is intended to remove a limb or a portion of a limb that is causing significant pain and dysfunction, while also facilitating the formation of a stump suitable for the attachment of a prosthesis. This prosthesis is designed to restore functional capabilities and promote pain-free rehabilitation. The specific level of amputation is typically identified by the joint or major bone at which the procedure is performed. Amputation of the lower extremity can significantly impact an individual's ability to stand and walk, often necessitating the use of prosthetic devices and, in many cases, additional mobility aids **{ 5 } .**

There are seven main categories of lower extremity prosthetic devices (lower limb prosthesis), which are: -

1- Partial-Foot Amputations: any amputation through foot.
2- Symes: this is an ankle disarticulation while preserving the heel pad.
3- Trans-tibial: (any amputation transecting the tibia bone or a congenital anomaly resulting deficiency) **.**

4- Trans-femoral: any amputation transecting the femur bone or a congenital anomaly resulting in a femoral deficiency.

5- Knee disarticulations: this usually refers to an amputation through the knee disarticulating the femur from the tibia.

6- Hip disarticulations: this usually refers to when an amputee or a congenitally challenged patient has either an amputation or anomaly at or in close proximity to the hip joint.

7- Hemipelvectomy: Although the anatomic differences between hip disarticulation and trans pelvic (hemipelvectomy) amputations are considerable, prosthetic component selection and alignment for both levels are quite similar **{ 6 } .**

1.4 Prosthetic Components: -

Prosthetics have undergone substantial advancements in recent years, particularly regarding their weight, lightness, functionality, and variety. This progress can be attributed to technological and scientific innovations in the utilization of carbon fiber materials, which contribute to the prosthesis's reduced weight, enhanced strength, and realistic appearance. Additionally, improvements in electronic components have facilitated better control of prosthetic devices, enabling users to perform various tasks while manipulating different objects or walking **{ 7 } .**

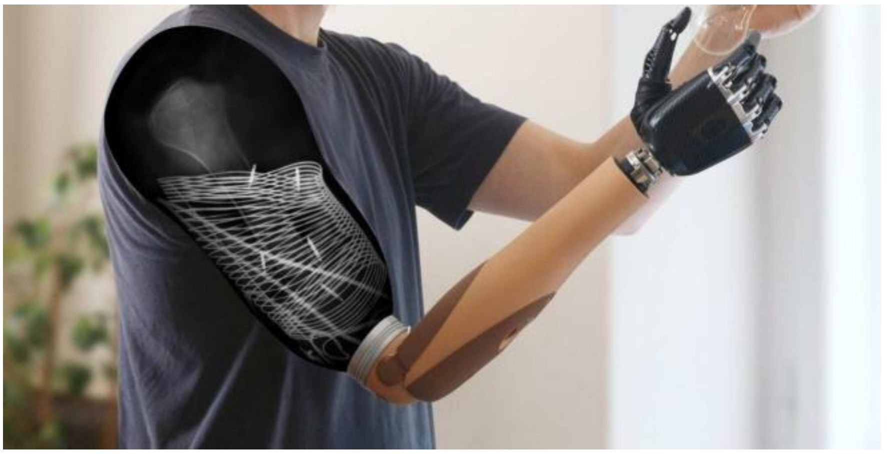


Fig (1.3) advanced prostheses

1.5 The components of the prosthetic limbs: -

The materials used in the manufacture of prosthetics have developed, but the main sections of the prosthesis have remained the same until now, which are :

1. The first piece, which is called the pylon: -

The internal framework of the prosthesis serves as its support structure. Recently, advancements have been made in the design of the prosthesis. This framework primarily consists of metal bars, with carbon fiber incorporated into its construction due to its advantageous properties. Additionally, this component is coated with a material that can be matched to the skin tone of the prosthetic user, ensuring a seamless appearance while providing lightness, strength, durability, and resilience [8] **.**

2. The second section, which is the cavity: -

The prosthesis is affixed to the patient's body in a manner that ensures secure attachment to the residual limb of the amputation. It is essential that the installation minimizes friction against the skin to prevent potential damage or injury, thereby reducing the risk of skin infections. To mitigate these issues, patients may wear specialized socks beneath the prosthesis, which is securely anchored in place [8] **.**

3. The last section is the comment system: -

The various components that aid in securing the prosthesis to the body are essential. These mechanisms vary based on the type of prosthesis, with different belts and sleeves employed to connect the body to the prosthetic device **{ 8 } .**

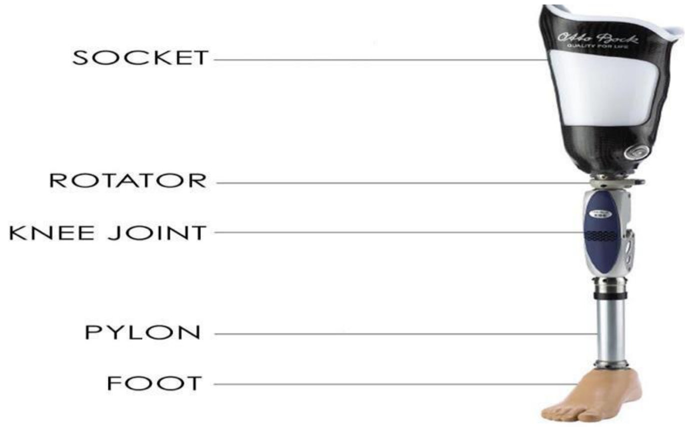


Fig (1.4) artificial lower limb parts

1.6 Prosthetics are designed into two types: -

1.internal prosthetics: -

Modular prostheses, characterized by their internal supporting structure, represent the most commonly utilized category of prosthetics. These internal devices leverage the human skeleton as a model, featuring a tubular frame that serves a weight-bearing function, complemented by a foam covering that enhances the prosthesis's natural appearance. The central component of this design is known as the pylon, which is constructed from materials such as aluminum, titanium, or stainless steel. The pylon serves to connect the proximal socket to the distal prosthetic foot, while also integrating joint components tailored to meet the specific requirements of the amputee **{ 8 } .**

The advantages of internal prosthetics are: - :-

a) Changes can be made at any time.
b) Lightweight and convenient to carry.
c) It is cosmetically acceptable and gives a semi-natural appearance suitable for all levels of amputation.
d) It gives proper tuning and dynamic alignment.

e) Good minuses Less resistance to external corrosion.

f) Foam cap does not last much longer and needs to be changed. Frequently **{ 8 } .**

internal prosthetics

2. External Skeletal Prosthetics: -

An exoskeleton prosthesis refers to a category of prosthetic devices characterized by an external supporting framework. Commonly referred to as traditional prosthetics or crustaceans, these devices feature a rigid outer shell that serves as a structural support, facilitating the distribution of shape and weight. The outer shell is responsible for bearing the weight, and it is typically constructed from materials such as wood or solid polyurethane, which is then encased in a durable plastic coating [9] **.**

<u>Advantages of external skeletal prostheses: -</u>

a) It lasted longer **.**

b) More resistant to external corrosion **{ 8 } .**

Disadvantages of external skeletal prostheses: -

a) It is cost effective.

b) Heavy and uncomfortable to use. Manufacturing time is longer.

c) Not suitable for knee amputation.

d) The alignment cannot be changed and cannot be modified **{ 8 } .**

Fig (1.5) external skeletal prosthetics

1.7 GAIT CYCLE : -

The gait cycle is a process that repeats the steps of walking. The step time is calculated from the point at which the first foot makes contact with the ground to the point at which the second foot makes contact with the ground. The distance between two separate feet is described as the stride length [9,10]. The gait cycle is divided into two primary phases: the stance phase, which includes heel strike, midstance, and toe-off, and the swing phase. Approximately 60% of the walking duration is allocated to the stance phase, while the remaining 40% occurs during the swing phase at different intervals throughout the cycle. A critical aspect of the gait cycle is the ground reaction force exerted on the foot at heel contact, where this force reaches 1.25 times the individual's body weight, closely matching the force applied.

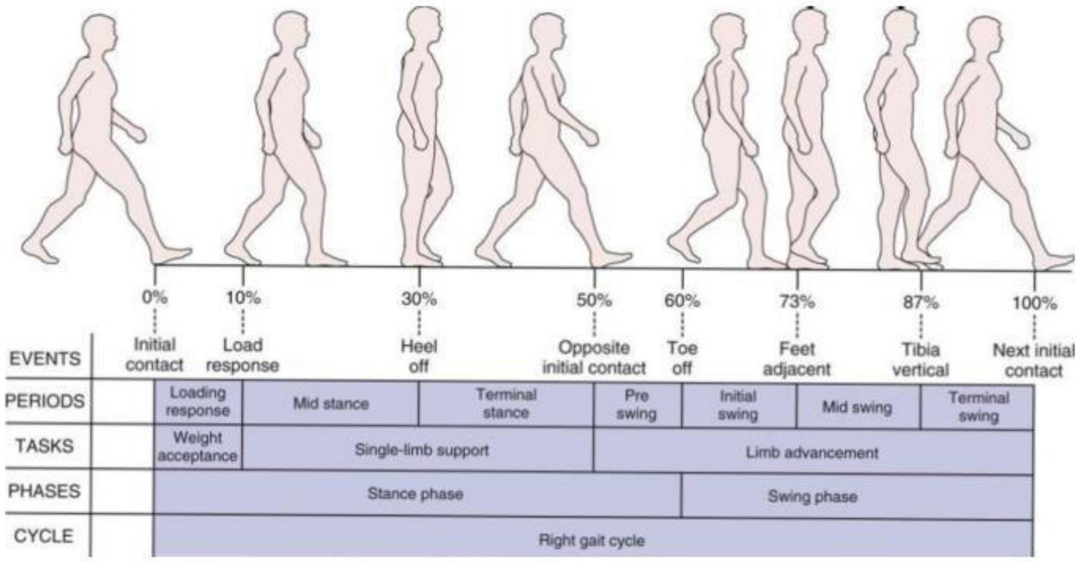


Fig (1.6) gate cycle process

1.8 Materials with Lower limp: -

A diverse range of materials is employed in the production of the actual tip, such as acrylic resins, carbon fibers, thermoplastics, silicone, aluminum, and titanium..

To achieve a realistic appearance, a foam cap can be molded and adjusted to conform to the actual party. A pliable, leather-like material will then be applied over the foam to enhance its lifelike quality. Nevertheless, this aspect is typically not acknowledged as a medical necessity by the majority of insurance providers, resulting in the patient bearing the financial burden of the extra expense associated with a "lifelike" party. Once the party has been set up, installed, and delivered, periodic adjustments are usually required to ensure that the fit remains appropriate.

The primary objective is to develop a prosthetic tip that aligns seamlessly with the patient's lifestyle and desired activity level. Currently, a diverse range of both natural and synthetic materials is employed in the field of prosthetics. Regardless of their origin, these materials must meet specific professional standards, including biocompatibility, strength, durability, lightweight properties, and ease of manufacturing. Presently, various forms of plastic are the most commonly utilized materials in prosthetics; however, traditional materials such as wood, leather, metal, and fabric continue to be relevant **{ 10 } .**

1.8.1 Carbon Fibers: -

1. It has a very high tensile strength; it is very strong relative to its size.
2. It is characterised by its high chemical resistance.
3. Withstands high temperatures with low thermal expansion.
4. About 90% of it is made from polyacrylonitrile and 10% of it is rayon silk or petroleum process.
5. It is characterised by high flexibility, and has the most flexible efficient coefficient for demanding applications.
6. It is characterised by the fact that its strongest types are five times stronger than steel. It has the ability to bend
7. It is characterised by high hardness and light weight [11]

1.9 Objectives of the Research: -

1. application on easy to obtain and low-cost material of lower limb

2.choosing cross section of shank it will be more resistance of stress and more safety factor

3.it is possible to manufacturing lower limp by the person that subjected to amputation by using 3D printer, and this does not require rehabilitation

1.10 Methodology: -

- Using computational methods (Solid works and ANSYS), Parameters:
  - Three deferent cross section area of shank.
  - Mechanical test: stress, Strain, Deformation
  - Select final Design

Chapter 2 Literature Review

Chapter 2 Literature Review

2.1 Introduction: -

This section aims to provide a concise overview of the historical development of lower limb prosthetic devices, along with a review of prior research in this field. It will explore the evolution of studies and investigations related to prosthetics, summarize key research findings, and present a historical account of the advancement of industrial prosthetic limbs and their current state.

2.2 Brief History of Prosthetics: -

Prosthetics that are thoughtfully designed not only offer practical functionality and aesthetic appeal but also contribute to the amputee's perception of completeness. A prosthesis enhances mobility and provides emotional reassurance, making the narrative of prosthetics not merely a scientific account, but a reflection of human experiences throughout history, encompassing those who have lost a limb due to congenital conditions, injuries, or accidents.

The earliest known example of a prosthetic device is not a leg, arm, or even an eye, but rather a toe. This particular big toe, which belonged to a noblewoman, was discovered in Egypt and is believed to date from 950 to 710 BC. While the significance of toes is widely recognized, it is intriguing that the oldest tangible representation of prosthetic history is a toe, rather than a more prominent limb like a leg or arm. The big toe held particular significance for the Egyptians, as it was essential for wearing their traditional sandals. This artifact has been in existence for nearly 3,000 years, illustrating that the history of prosthetics encompasses both functionality and identity.

To highlight the limited advancements in prosthetics throughout much of history, one can examine the prosthetic hands and legs used during the Dark Ages—almost 2,000 years later. Knights of this era frequently utilized iron prosthetics, crafted from the same metal as their armor. These substantial limbs were not particularly functional; rather, they served primarily to mask the absence of a limb, which was often viewed as a source of shame. As the use of prosthetics became more widespread, innovations in joint technology and suction-based attachment methods emerged, marking progress in the field. A notable development occurred in 1812, when a prosthetic arm was created that could be operated by the opposite shoulder through the use of tie belts, akin to the braking mechanism on a bicycle.

The National Academy of Sciences, a governmental body in the United States, established the Prosthetics Program in 1945. This initiative was a direct response to the significant number of veteran amputees resulting from World War II and aimed at promoting scientific progress in prosthetic development. Since its inception, advancements in materials, computer-aided design, and surgical techniques have contributed to the increasing realism and functionality of contemporary prosthetics. The world wars highlighted the urgent need for innovations in prosthetic technology. Following World War I, the Surgeon General of the US Army initiated efforts that would eventually lead to the formation of the American Society of Prosthetics and Orthotics. However, substantial progress in prosthetics did not occur until the post-World War II era, when the US government allocated funds to military contractors to enhance the design and performance of prosthetic devices. This funding facilitated the introduction of various modern materials, including plastics, aluminum, and other composite substances.

The invention of the absorbent sock for above-the-knee prostheses at UC Berkeley in 1946 is noteworthy. In 1975, Ysidro Martinez, a Mexican-American inventor, developed a below-the-knee prosthesis aimed at addressing the walking difficulties associated with prosthetics of that era. This design featured a high, lightweight center of mass, which minimized friction and pressure, thereby facilitating smoother acceleration and deceleration.

Thanks to the dedication of prosthetic pioneers, we are now closer than ever to achieving the full functionality of a biological limb. Blade prosthetics enable amputees to run, while microprocessor-controlled knees allow prostheses to adjust their flexion and extension according to varying environments. With the progress in neural prosthetics and the development of fully brain-controlled devices, we are still striving toward the aspiration of completely replacing a lost limb. [13].

Fig. (2-1) The world's first prosthetic toe (left), artificial leg for a below - knee amputee right

2.3 Literature Review on Prosthetic: -

In 2012, Rosalam Che Me and colleagues conducted a study on artificial limbs, specifically examining the cavity section, which is modified and substituted with natural biocomposites. The researchers posit that biocomposites derived from natural fibers, such as reinforced natural plastics, possess comparable properties to existing materials suitable for diverse applications. The study's outcomes highlight the compatibility between the characteristics of current materials and those proposed, thereby offering alternative solutions that are both cost-effective and environmentally sustainable while preserving the essential features required for prosthetics. The findings aim to assist young patients or users who struggle with the necessity of prosthetic devices, enabling them to achieve greater independence. The research advocates for a new material that is not only economically viable but also retains the desired attributes of prosthetic devices. Additional advantages of this material include its biodegradability, recyclability, and renewability. Furthermore, it is anticipated that this innovation will benefit individuals in impoverished regions who are unable to afford high-cost prosthetics [14].

M. Praveen Kumar , In 2012, a study was conducted to investigate the influence of sample dimensions on the bending modulus of isotropic and fiber-reinforced polymer (FRP) composite materials through three-dimensional finite element analysis. The modeling of the problem was executed using ANSYS software, where the bending modulus, derived from the tangential deflection, was compared with the material properties provided as inputs for the analysis. The study also addressed the impact of the physical dimensions of the sample on the percentage error associated with Young's modulus. A three-dimensional finite element simulation of a three-point bending test was carried out to assess how geometric dimensions affect the bending modulus calculated using the Euler bending formula. The analysis focused on isotropic FRP materials and carbon, evaluating the percentage deviation in the transverse modulus resulting from changes in geometry. The following conclusions were reached: the span-to-depth ratio should exceed 50 for a fixed width of 25 mm, and the sample width should be limited to less than 5 mm to maintain a constant depth ratio of 70/3. In the case of composite materials, where the percentage deflection is more significant, the design of the four-point bending test, despite its complexity, may yield more accurate results for the bending modulus due to the predominance of pure bending in the structure [15].

Liai Pan, et.al at 2015, The principles of finite element analysis are introduced, laying the groundwork for the theory. Utilizing ANSYS Workbench, an analysis of the static characteristics was conducted while simulating the normal walking posture of the human body. The findings indicate that the design of the exoskeleton's strength and rigidity is adequate to support patients weighing up to 100 kg, while also achieving a larger deformation area for the exoskeleton. This provides essential theoretical data to ensure safety for the subsequent development of the exoskeleton. The finite element analysis, executed through the ANSYS program, focused on the exoskeleton during typical human gait. By performing a static analysis, the study assessed whether the exoskeleton's strength and rigidity align with the design specifications, identifying the maximum stress and deformation to avert potential damage to the structure [16].

Catalina Quintero - Quiroz, et.al 2017, A comprehensive review was conducted on the various polymers utilized in the creation of lower limb cavities and external interfaces for prosthetics and orthotics, focusing on their functional requirements and the potential dermatological issues that may arise from their application. The literature review involved an extensive search through databases including EBSCO, Embase, LILACS, SciELO, ScienceDirect, and Scopus, resulting in the identification of 47 articles and papers that fulfilled the inclusion criteria. The polymers employed in the production of prosthetic and orthotic interfaces and sockets include thermoplastics, thermosets, foams, gels, and elastomers. Nevertheless, research indicates that between 32% and 90.9% of individuals using these devices have encountered skin-related issues on the affected stump or limb, such as excessive perspiration, wounds, and irritation [17].

Hasan Saad,et.al at 2017 , The 3D modeling of prosthetic lower limbs was conducted, leading to the development of a modified finite element socket model within the ANSYS 14.0 workbench. This model aimed to analyze stress distribution and deformation patterns under functionally relevant loading conditions during a typical gait cycle. The study focused on enhancing bore loading deflection, as well as the foot and leg components of the pylon. The optimal results for above-knee prosthetic designs reached 94.64%. Additionally, a simplified stiffener was incorporated to enhance the moment of inertia of the primary component, thereby improving deflection. This case study adhered to international socket specification standards to clarify the objectives of the research. Moreover, a certified design was meticulously chosen to ensure it met the criteria for high strength and minimal weight. The findings indicated that employing a stable method allowed for the acquisition of stable and precise numerical approximations that align with the physical configuration of the problem on a coarse network, utilizing ANSYS Workbench 14.0, AutoCAD, and COMSOL 5.2. High Density Polyethylene (HDPE) was utilized for the photographic documentation [18].

Szykiedans ,et.al at2017 , The study concentrated on the mechanical characteristics, specifically the fundamental tensile strength and elastic modulus, of components produced through the Fused Deposition Modeling (FDM) technique, utilizing two distinct materials: unmodified polyethylene terephthalate glycol

(PETG) and glass-fiber reinforced PETG. This paper delineates the advantages and disadvantages of the materials examined and contrasts the properties of PETG with and without the incorporation of glass fiber. The data collected serves to quantify the mechanical properties of PETG components and may also be applicable for modeling the characteristics of 3D printed elements [19].

Nurhanisah Mohd et al. In 2017, a study was conducted to investigate patient satisfaction regarding the quality of prosthetic leg sockets designed for individuals with lower limb amputations. This qualitative research involved in-depth interviews, which were preceded by a questionnaire, with patients from a Rehabilitation Center and Hospital in Malaysia. A total of twelve amputees, specifically those with below-knee amputations, were randomly selected from both outpatient and inpatient groups. The analysis of the participants' narratives focused on identifying the functional and aesthetic attributes of the prosthetic leg sockets currently in use, as well as any associated issues. The findings revealed that 41.7% of the twelve participants expressed satisfaction, while 25% reported being somewhat satisfied with their existing prosthetic sockets. Participants identified durability and comfort as the most critical characteristics of these sockets, with 83.3% emphasizing their importance. Regarding the aesthetic aspect, 66.7% of respondents indicated that the material used in the socket's construction was the most significant feature. Consequently, it can be concluded that the current levels of satisfaction with the quality of prosthetic sockets among amputees in Malaysia are generally favorable, with many preferring their prostheses. These findings may inform future research on the aesthetics and design of prosthetic sockets [20].

Arianna Menciassi and Linda paterno (2018) The purpose of this paper is to examine the primary parameters—namely displacements, stress, volume changes, and temperature—that influence the stump-socket interface and compromise the comfort and stability of limb prostheses. This review presents a classification of various socket types found in the literature, along with an evaluation of the advantages and disadvantages of each solution from multiple perspectives. Furthermore, the paper outlines the technological approaches available to address the altered stress distribution on the residual limb tissues, the volume changes of the stump over time, and the temperature variations impacting the residual tissues within the socket. The ongoing challenges in this research area are emphasized, and potential future directions are explored, aiming towards the ambitious goal of developing an advanced socket that can adapt in real-time to the intricate interplay of factors affecting the stump during both static and dynamic activities [21].

Sofiane Guessasma, et.al at 2019, The printability of PETG for fused deposition modeling (FDM) was examined by utilizing an infrared camera to monitor the filament's temperature. The structural characteristics of 3D-printed PETG were assessed through X-ray micro-tomography, while the tensile properties were evaluated across a range of printing temperatures from 210°C to 255°C. A finite element model, grounded in the 3D microstructure of the printed material, was employed to elucidate the deformation mechanisms and the impact of microstructural defects on mechanical performance. The findings indicate that PETG can be effectively printed within a constrained temperature range. Furthermore, the results demonstrate a notable decline in mechanical performance attributable to the FDM process, particularly highlighting a significant reduction in elongation at fracture. This reduction is attributed to the heterogeneous deformation of the PETG filaments. X-ray micro-computed tomography findings indicate a minimal level of porosity resulting from the process, which is confined to the thickness of the sample [22].

Ameer A. Kadhim et.al at 2021, The stem was designed and analyzed using SolidWorks software, followed by fabrication with a 3D printer to ensure optimal load distribution across the stem walls. The new stems are constructed from various materials, including ABS and PLA. During gait cycle testing with a force plate, two distinct phases were identified: the swing phase and the stance phase. The ground reaction force generated during the stance phase was measured at 1.3 times the weight of the amputee at the heel of the toe. Subsequently, a specialized device was developed to assess the durability of the new leg under alternating loads. Both practical and numerical analyses revealed that the mechanical properties of the new leg met the necessary requirements for prosthetic characteristics. The findings from this research indicate that the new stems are not only easy to manufacture and cost-effective but also lightweight. Additionally, 3D printing technology can be employed to create essential and precise prosthetic components, such as the leg with adapters. Furthermore, testing with a fatigue foot shank tester demonstrated that the SACH foot experienced failure prior to the leg. Under identical temperature conditions, the hot climate results confirmed that the new stems are well-suited for use in warmer environments [23].

Ming - Hsien Hsueh ,et.al 2021, The investigation focused on the characterization of FDM polyethylene terephthalate glycol (PETG) materials under four distinct loading conditions: tension, compression, bending, and thermal deformation. The objective was to gather data pertaining to various printing temperatures and speeds. The findings revealed a notable asymmetry in tensile and compressive properties between PLA and PETG materials. It was observed that the mechanical properties—

namely tension, compression, and bending—of both PLA and PETG improve with increased printing temperatures, while the impact of printing speed yields differing outcomes for the two materials. Furthermore, PLA exhibited superior mechanical properties compared to PETG; however, the latter demonstrated greater thermal deformation. These insights are expected to significantly benefit researchers engaged in polymer studies and FDM technology, particularly in the pursuit of sustainability. The research outcomes highlighted the following points: 1. PLA and PETG materials exhibit clear tensile and compressive asymmetry, with compressive stress surpassing tensile stress. 2. An increase in printing temperature correlates with enhanced mechanical properties for both PLA and PETG. 3. While the mechanical properties of PLA improve with higher printing speeds, those of PETG decline. 4. PLA consistently outperforms PETG in terms of Young's modulus and strength, although the reverse is true for thermal deformation. 5. Under conditions of less than 20% filler density, compressive strength remains higher than tensile strength, yet lower than bending strength, with Young's tensile modulus exceeding that of compressive modulus [24]

Zainab H. Zaier, et.al at 2022, The study involved the design and analysis of stems using solid modeling software, which were subsequently manufactured with a 3D printer. Two types of materials, PLA and ABS, were utilized in the construction of these stems. A specialized device was then created to evaluate the lifespan of the new stalk under alternating loads. Following both practical and numerical assessments, the mechanical properties of the new leg were found to be satisfactory, aligning with the required specifications for prosthetic features. The analysis indicated that the new shank is not only easy to manufacture and cost-effective but also lightweight. The 3D printing technique proves to be effective for producing critical and precise prosthetic components, including the shank and its adapters. Furthermore, the new shank demonstrates a favorable lifespan when compared to the SACH foot [25].

Ali Noori Kareem et. Al at 2021, Lower limb amputations are predominantly caused by accidents or medical conditions affecting the limbs. The essential functions of the socket, shank, and prosthetic foot are to restore skeletal function, compensate for the absence of structural components, and replicate the muscular functions of the ankle, pylon, and foot. The conventional prosthetic shank is crafted from a titanium-aluminum alloy, which is favored for its lightweight properties. This study employs ANSYS software to design and analyze the shank. The newly developed shank incorporates various materials (A. As A, da, and ka), which are noted for their lightness and cost-effectiveness compared to those typically used in lower limb prosthetics. In addition to material modifications, the research explored changes to the internal cross-sectional shape of the shank, replacing the original circular design with elliptical and hexagonal alternatives. Numerical simulations conducted with ANSYS 2020 R2 revealed that the elliptical cross-section enhanced the shank's performance in terms of stress, deformation, and safety factor, whereas the hexagonal design led to a notable reduction in the safety factor and an increase in stress levels. The findings indicate that the optimal shank design, considering both internal cross-sectional shape and material selection, is the elliptical configuration, which is recommended for implementation [26].

Chapter 3

Theoretical &Methodology

Chapter 3 Theoretical &Methodology

3-1 Analyzing the Pylon Mathematically and Simulating Human Movement: -

The primary aim of biomechanical analysis is to determine the specific actions of muscles, including the timing of their contractions and the forces generated. Two independent techniques, models and analytes, are employed for the modeling and understanding of human gait through analysis and statistical methods. A computational approach is utilized to calculate the center of gravity, ankle moment, and dorsiflexion angle for the lower limb above the ankle joint, based on the free body diagram of the lower limb, while assuming the body is represented as a straight line with joints located at the knee and hip. Refer to illustration (3) for further clarification. When the equilibrium equation is applied, the resulting equations of motion in the X and Y directions are derived [16].

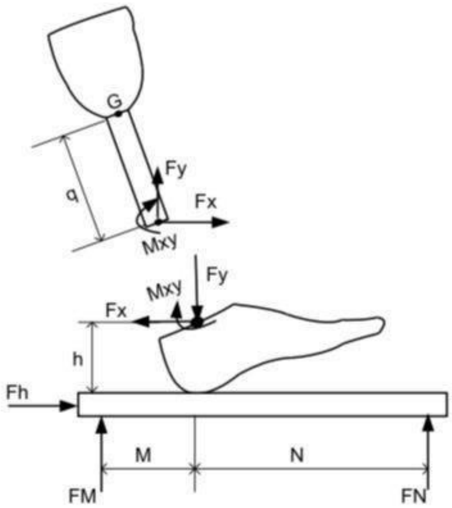


Fig (3-1) Force and moment on the prosthetic below knee.

Provided the VPP is above the CoM, then hip torques tend to decrease deviations between the trunk and hip--CoP axes (figure 1a, negative feedback). This negative feedback stabilizes the VPP model for upright [17] and for pronograde gait [18]. If however, the VPP is below the CoM, then hip torques tend to increase deviations between the trunk and hip–CoP axes

Stresses

Stress is a physical quantity which expresses the internal forces exerted on each other by neighboring particles of a continuous material, while strain is the measure of the material's deformation

Strain

The calculation of geometric deformation pertains to the relative displacement of a material body among its constituent particles. Various processes, such as the stress applied by external forces on the bulk material (including gravitational forces) or on its surface, can induce strain within the material. This strain may arise from contact forces, external pressure, or friction. The application of pressure on a rigid material generates an internal elastic stress, analogous to the reaction force of a spring, which facilitates the return of the material to its original, undeformed condition.

$$\mathbf{F_x = M_t\, Xc}$$

$$\mathbf{F_Y = M_t\,(\, g + \ddot{y}_C\,)}$$

The angle of inclined with vertical axes is

$$\boldsymbol{\theta} = \mathbf{2tan}\left(\frac{-K_1 \pm \sqrt{(k_1^2 + k_2^2 - k_3^2}}{(k_3 - k_2)}\right)$$

The location of the COM can be obtained

as $\mathbf{y_C = b_i \cos\theta\ x_C\ = b_i \sin\theta}$

The equation can be used to determine the ankle moment

$$\mathbf{M_{xy} = F_N N - F_M + F_h\, h - mga}$$

Where:

$$\mathbf{K_1 = (M_t + m)I - (M,\, b)g - I(F_N + F_M)}$$

$$\mathbf{K_2 = F_h\, I}$$

$$\mathbf{K_3 = M_t\, bi\,(\, F_N\, N - F_M\, M + F_h\, h\,)}$$

Where Fx and Fy : are the perpendicular ground response forces to the force plate, as measured by front and rear transducers, respectively. F, is the horizontally measured **ground response force parallel to the force plate using a transducer.[23]**

3-2 Tensile test Area=width*thickness: -

Strain = $\frac{\Delta l}{l}$

Stress = F/A

Where Δ L = chang in lenth of sample L = orignal length of sample F = force, load applied upon sample A = Area of sample

Where were the dimensions of sample?

L = 150.8mm Thickness = 6mm 3-3 Case study

(Stress analysis of carbon fabber) The vertical and horizontal load were applied using the information of a person whose mass is (75,100,) assuming that the length of the pylon is (300 mm) as shown in Figure (3-2).

Fz= (1.25) (human weight in z direction) Fx = (-0.18) (human weight in x direction)
F`z=Fz cos30° +Fx sin30° = (the load that applied in static analysis)

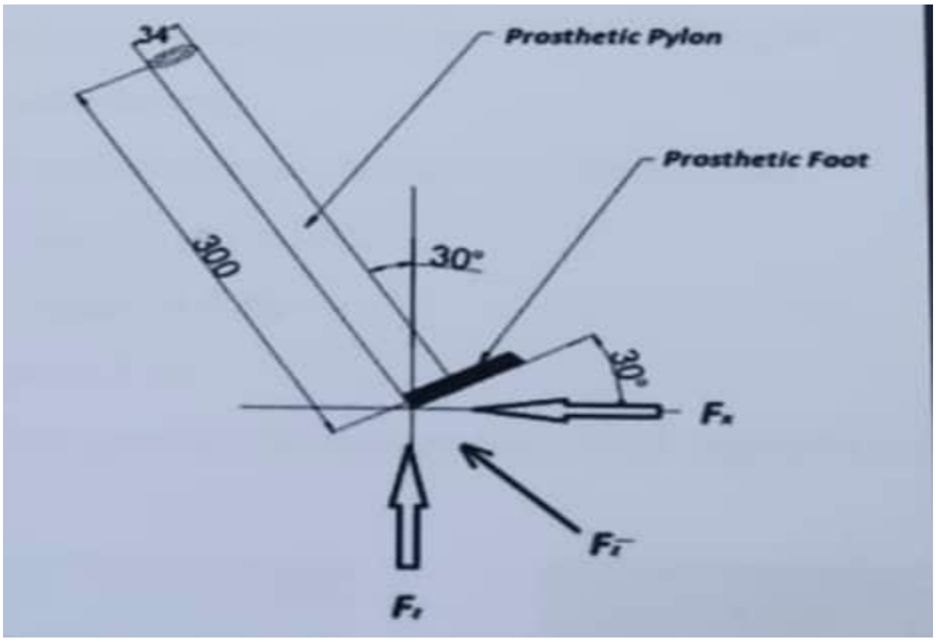


Fig (3-2). Stress Analysis of prosthetic pylon (case study).

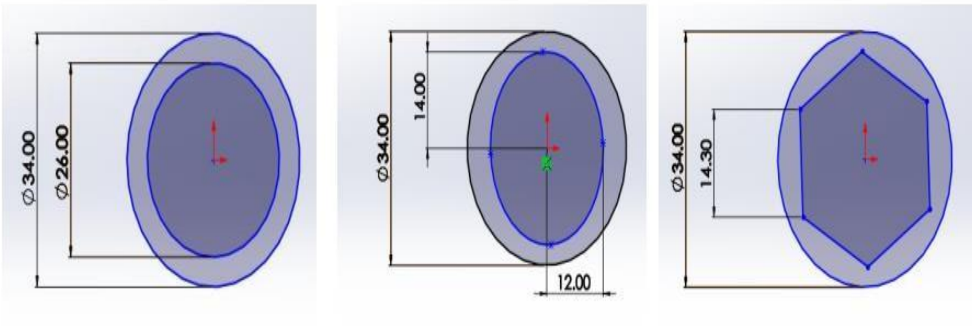


Fig (3.3) the dimension of three geometry

Static Loading

Permissible or design stress = $\frac{Yiels\ stress}{FOS}$ (For Ductile material)

(Or) FOS= $\frac{Yieldstress(\sigma_y)}{permissiblestress(or)Designstress}$

$$permissiblestress(or)Designstress = \frac{Ultimate\ stress}{FOS}\ (for\ Brittle\ material$$

FOS= $\frac{\sigma_u}{Desigm\ stress}$

In General (The factor of safety)

N		Ductile	Brittle	Brittle
1	Static Load	1.5-2	3 to4	4.5-6
A	Repeated	3	6	9
B	Revered	4	8	12
2	Heavy shock	5	10	15

order to predict the failure under combined loads, failure theories are used

.

3.3PRINCIPAL STRESSES: -

Machine components experience various external loads of diverse characteristics. Consequently, it is essential to determine the equivalent single stress by utilizing principal stresses. At any location within a strained material, three mutually perpendicular planes exist where only direct stresses are present, with no shear stresses acting. These planes are referred to as principal planes, and the direct stresses acting on them are known as principal stresses or normal stresses. Out of three Principal stresses $\sigma 1, \sigma 2, and\ \sigma 3$ one is maximum, one is minimum and the other one is intermediate .

Two-dimensional

Max. principal stress

$$= \sigma 1 = \frac{\sigma_x + \sigma_y}{2} + \sqrt{\left(\frac{\sigma_x - \sigma_y}{2}\right)^2 + t_x^2 y} = \frac{\sigma_x + \sigma_y}{2} + \frac{1}{2}\sqrt{\left(\sigma_x - \sigma_y\right)^2 + 4t_x^2 y}$$

MIN. principal stress

$$= \sigma 2 = \frac{\sigma_x + \sigma_y}{2} - \sqrt{\left(\frac{\sigma_x - \sigma_y}{2}\right)^2 + t_x^2 y} = \frac{\sigma_x + \sigma_y}{2} - \frac{1}{2}\sqrt{\left(\sigma_x - \sigma_y\right)^2 + 4t_x^2 y}$$

Max .shear stress = $t^{max} = \frac{\sigma_1 - \sigma_2}{2}$

Chapter 4 Experimental Work

Chapter 4 Experimental Work

4.1 Introduction: -

This chapter addresses the instruments employed in the study, along with the evaluations performed on the materials.

4.2. 3D Printing Work: -

Additive manufacturing is a production technique characterized by the addition of material in successive layers to form a component, rather than subtracting material to achieve the final shape. This method has a wide range of applications in 3D printing, including prototyping, the creation of specialized components for sectors such as aerospace, military, and biomedical engineering, as well as dental applications. Additionally, it is utilized in hobbies and home projects, with potential future applications in the medical field (such as the production of body parts), as well as in the construction of buildings and vehicles.

3D printing employs software that divides the three-dimensional model into layers, typically measuring 0.01 mm or less in thickness. The printer subsequently traces each layer onto the build plate. Once the pattern for a layer is completed, the build plate is lowered, allowing for the addition of the next layer atop the preceding one.

Conventional manufacturing methods are referred to as 'Subtractive Manufacturing' due to their reliance on the removal of material from a pre-existing block. Techniques such as milling and cutting fall under this category. This approach often results in significant waste, as the material that is removed is usually rendered unusable and discarded as scrap. In contrast, 3D printing significantly reduces waste by depositing material only where it is necessary, leaving the surrounding areas as empty space. In this work have been used 3D printing for design samples for mechanical testing as shown in figure 4.1.

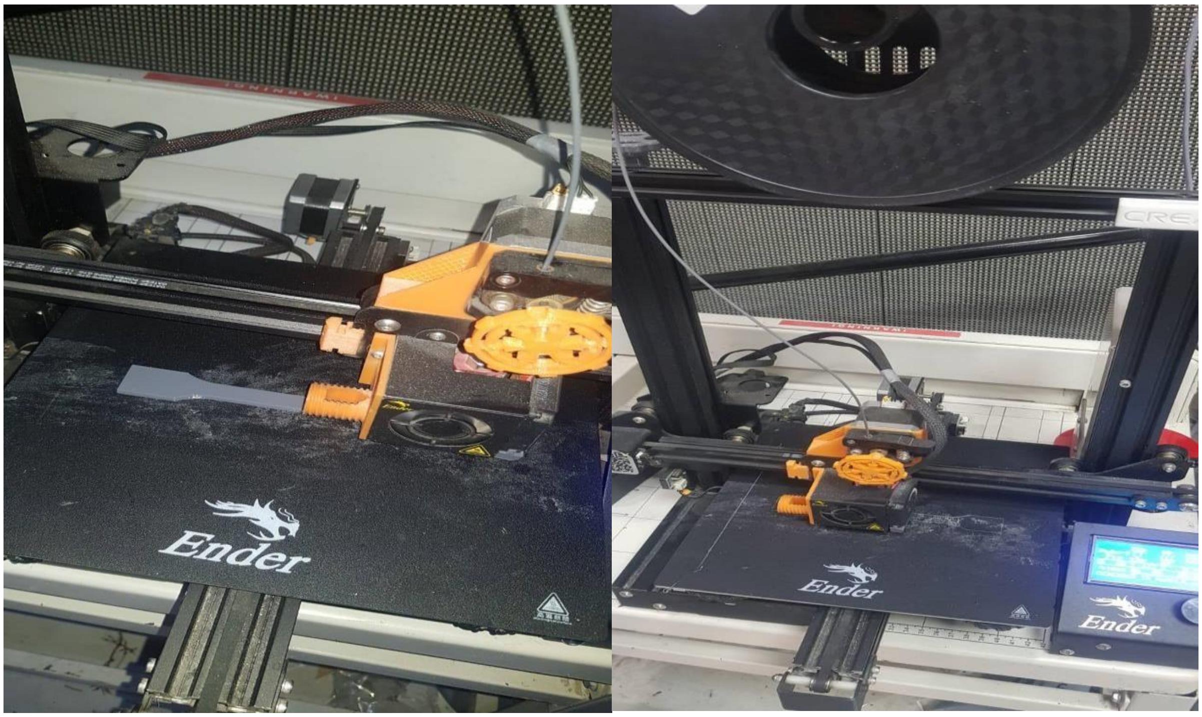
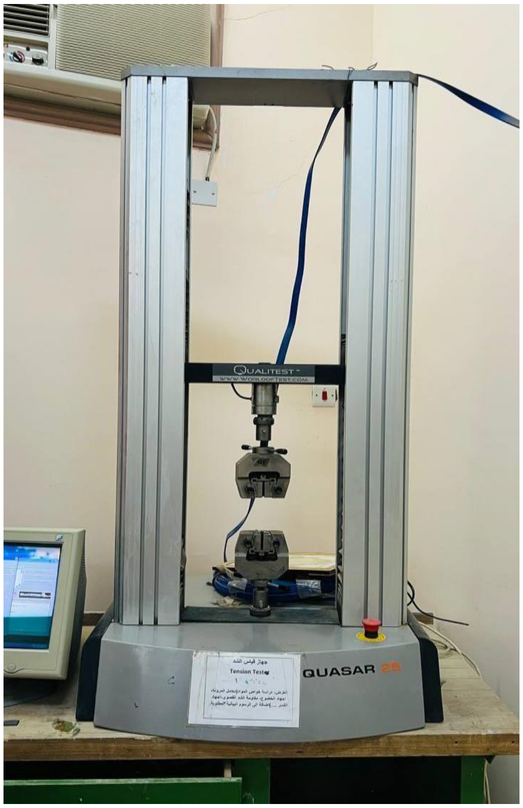


Fig (4.1) shows 3D printing Work 4.3Tensile test.

Tensile testing, often referred to as tension testing, is a crucial procedure in materials science and engineering that involves applying a controlled tensile force to a sample until it fails. The primary properties measured during a tensile test include ultimate tensile strength, breaking strength, maximum elongation, and reduction in area. From these measurements, additional properties such as Young's modulus, Poisson's ratio, yield strength, and strain-hardening characteristics can also be derived. Uniaxial tensile testing is predominantly employed to assess the mechanical properties of isotropic materials, while some materials may undergo biaxial tensile testing. The key distinction between these testing methods lies in the manner in which the load is applie to the materials.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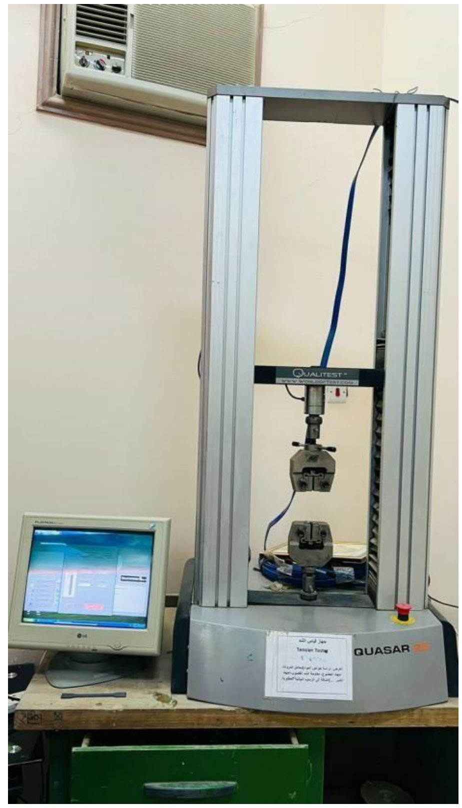
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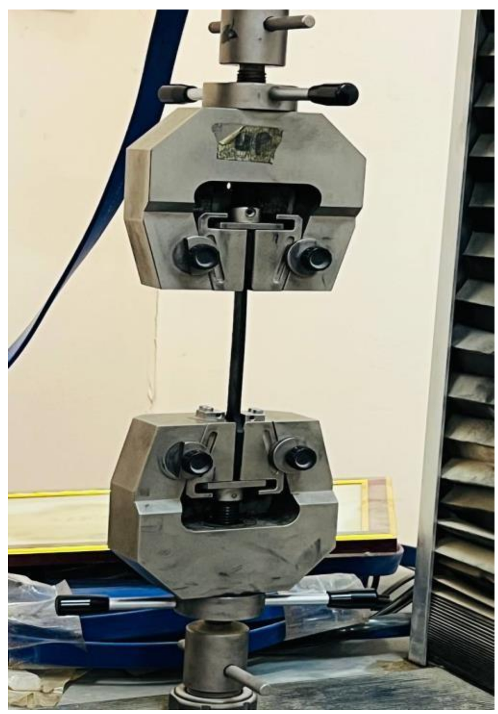

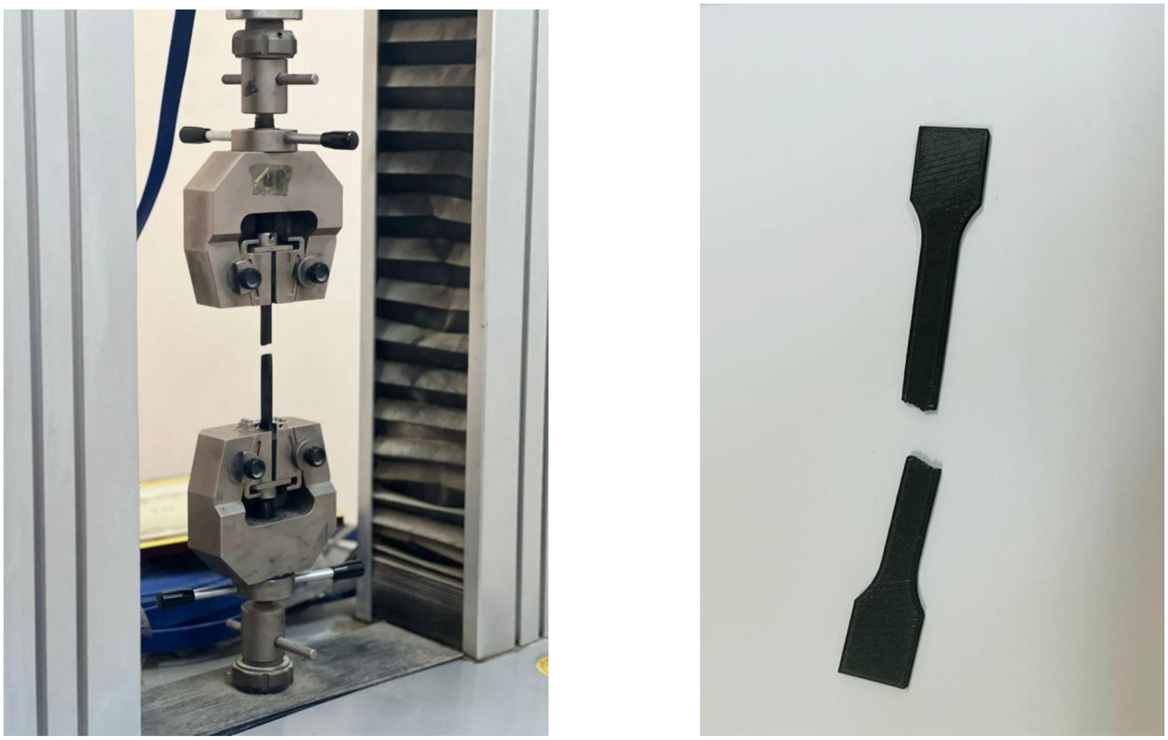

Fig (4.2) Shows Tensile Test.

4.3 Rehabilitation: -

Rehabilitation is a form of care designed to assist individuals in recovering, maintaining, or enhancing the skills necessary for daily living. These skills may encompass physical, mental, and cognitive functions, which could have been diminished due to illness, injury, or as a consequence of medical treatments. Engaging in rehabilitation can significantly enhance your daily life and overall functioning.

This process is particularly beneficial for individuals who have experienced a loss of essential daily living skills. Common causes of such losses include:

- Injuries and trauma, such as burns, fractures, traumatic brain injuries, and spinal cord injuries
- Stroke
- Severe infections
- Major surgical procedures
- Side effects from medical treatments, including those related to cancer therapies
- Certain congenital anomalies and genetic disorders
- Developmental disabilities
- Chronic pain conditions, including back and neck pain

The primary objective of rehabilitation is to facilitate the recovery of lost abilities and promote independence. However, the specific objectives vary for each individual, influenced by the underlying cause of the impairment, whether it is persistent or temporary, the specific abilities that have been affected, and the severity of the condition.

Fig (4.3) Shows Rehabilitation processing in Nasiriya city

Chapter 5 Results &Discussion

Chapter 5 Results &Discussion

5.1Introduction: -

In this chapter, the findings from both the experimental and theoretical investigations concerning the prosthetic material will be presented, focusing on the tensile and fatigue tests, as well as the numerical analysis.

Figure (5.1) Shows the result from tensile test in the experimental work, which noted that the magnitude of tensile yield stress is 60 Mpa .while ultimate stress is 70 Mpa .

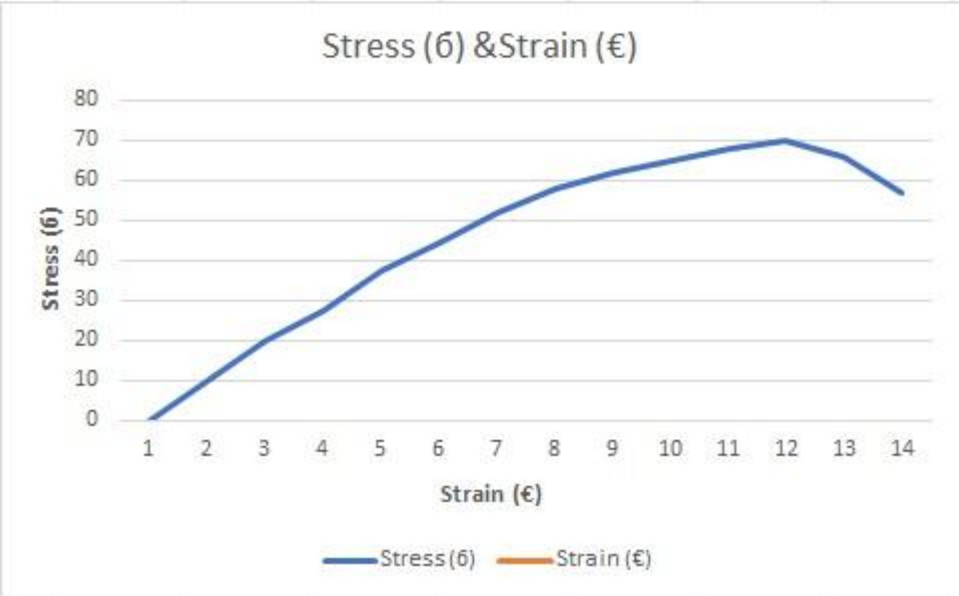

Fig (5.1) Shows Stress & Strain For fibber carbon Material.

5.2The Numerical Analysis: -

The finite element method (FEM) has become extensively utilized across various domains of engineering and science. This widespread adoption is facilitated by the rapid advancements in digital computing, characterized by substantial memory capacity and swift processing speeds. FEM is esteemed as one of the most robust numerical techniques due to its proficiency in handling intricate geometric boundaries and non-linear material characteristics. In this study, FEM, supported by ANSYS Workbench 18.2 software, serves as a numerical instrument to demonstrate the impact of fatigue performance on a structural element. It is employed to assess the behavior of maximum stress, total deformation, fatigue life, and safety factor.

Figure (5.2) describe the modeling process with three cross section area of shank by using carbon fiber material

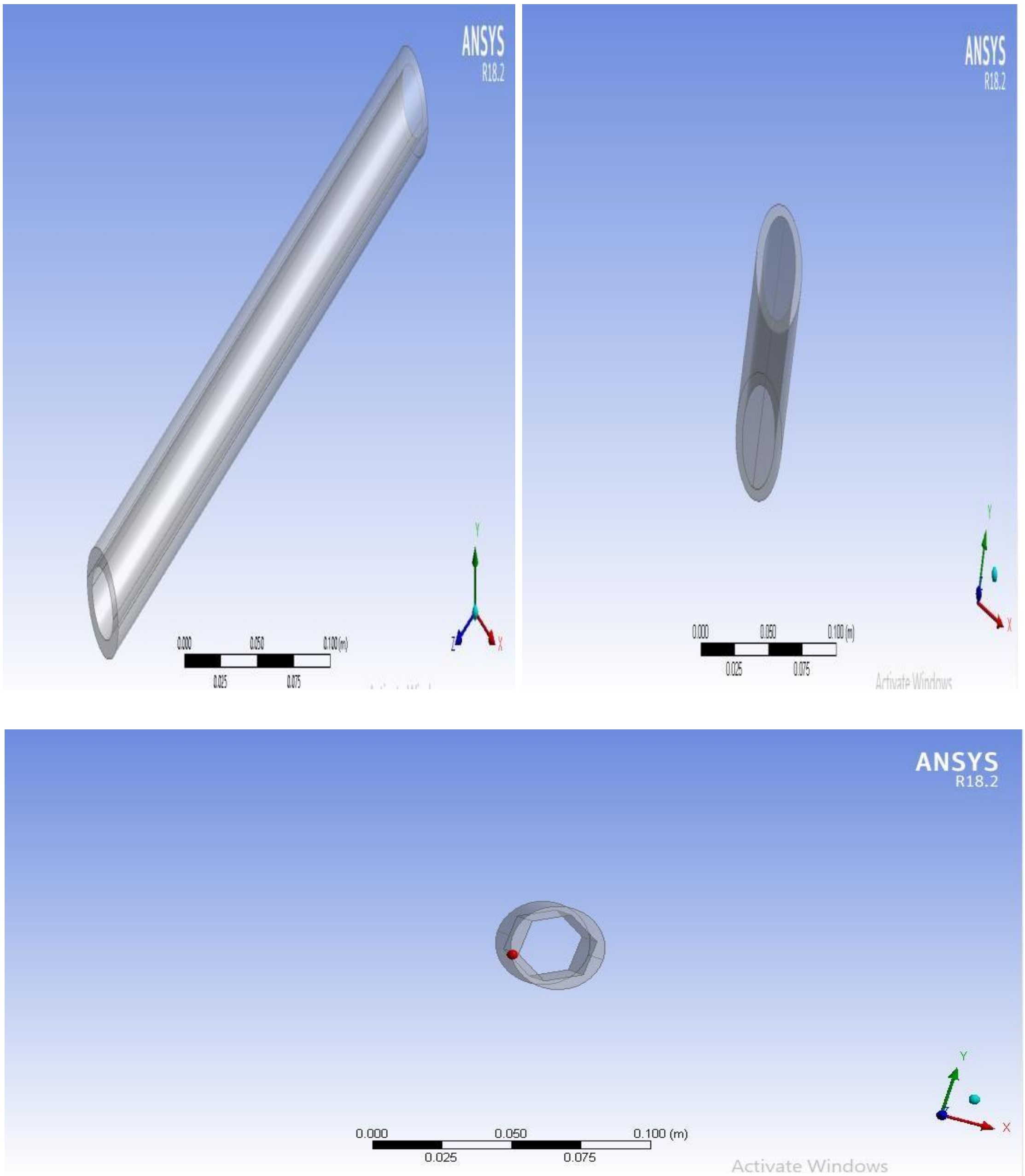

Fig (5.2) modeling process with three cross section are.

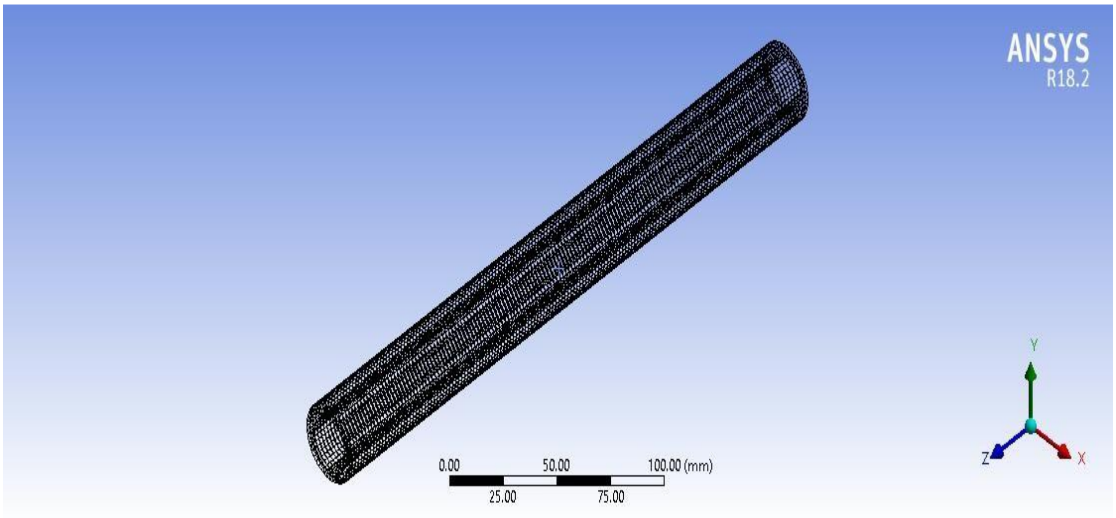

Fig (5.3) show meshing process.

5.3 Case one of analysis with human wights is 75: -

Figure (5.4) shows the maximum deformation through applied Wight was magnitude is 75kg with circle cross section area , which noted the maximum deformation on the shank is 0.69 mm.

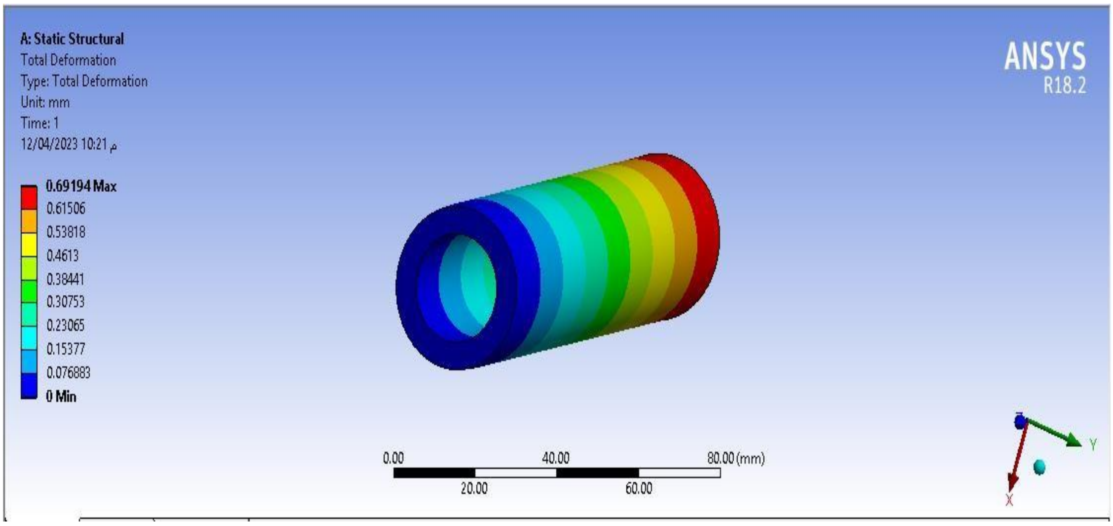

Fig (5.4) show maximum Deformation of circle cross with 75 kg.

Figure (5.5) shows the maximum elastic strain of circle cross through applied Wight was magnitude is 75kg , which noted the maximum on the shank is 0.0045 .

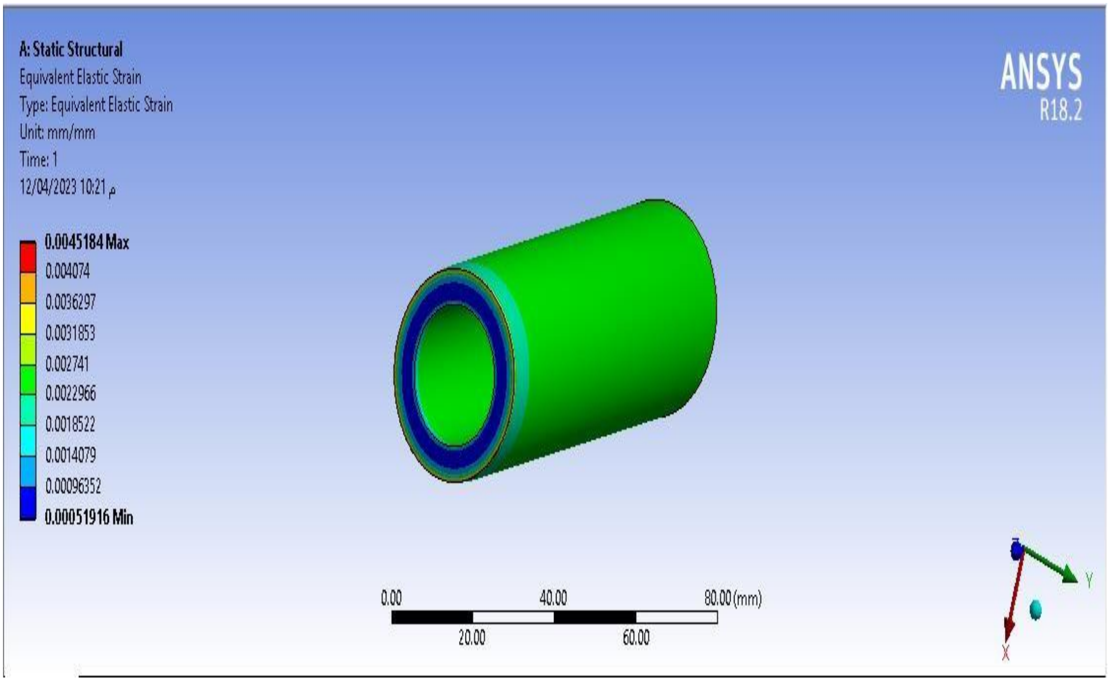

Fig (5.5) shows the maximum elastic strain of circle cross with 75 kg .

Figure (5.6) shows the maximum stress of circle cross through applied Wight was magnitude is 1kg , which noted the magnitude of this Wight is 4.9 Mpa

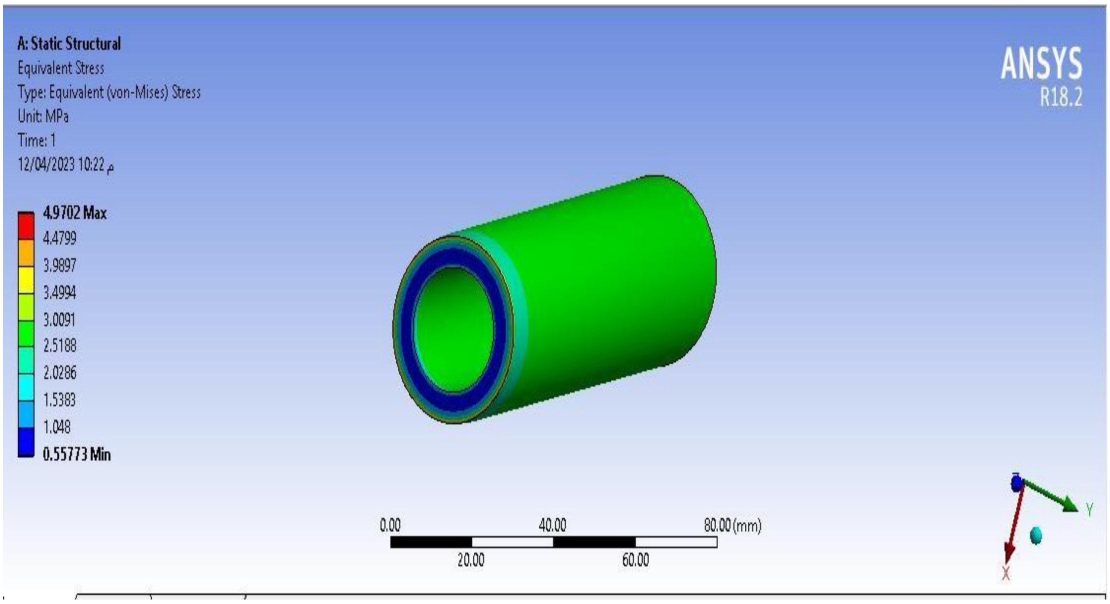

Fig (5.6) shows the maximum stress of circle cross with 75 kg

Figure (5.7 shows the safety factor through applied Wight was magnitude is 75 kg , which noted the minimum is 5

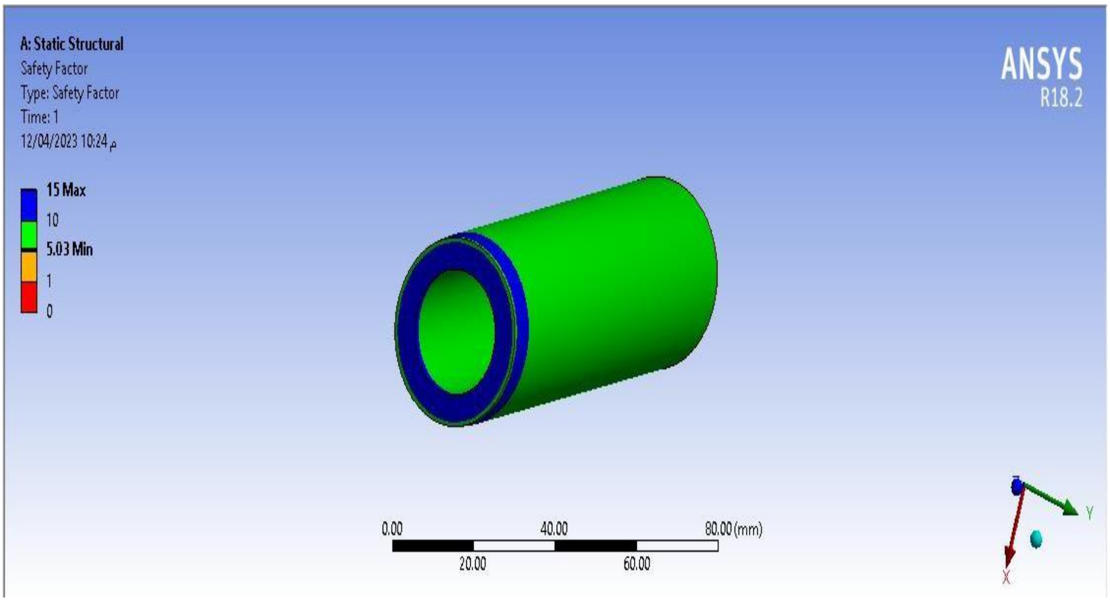

Fig (5.7) shows the safety factor of circle cross with 75 kg

Figure (5.8) shows the maximum deformation of ellipse cross section through applied human Wight's was magnitude is 75 kg , which noted the maximum deformation on the shank is 0.71 mm

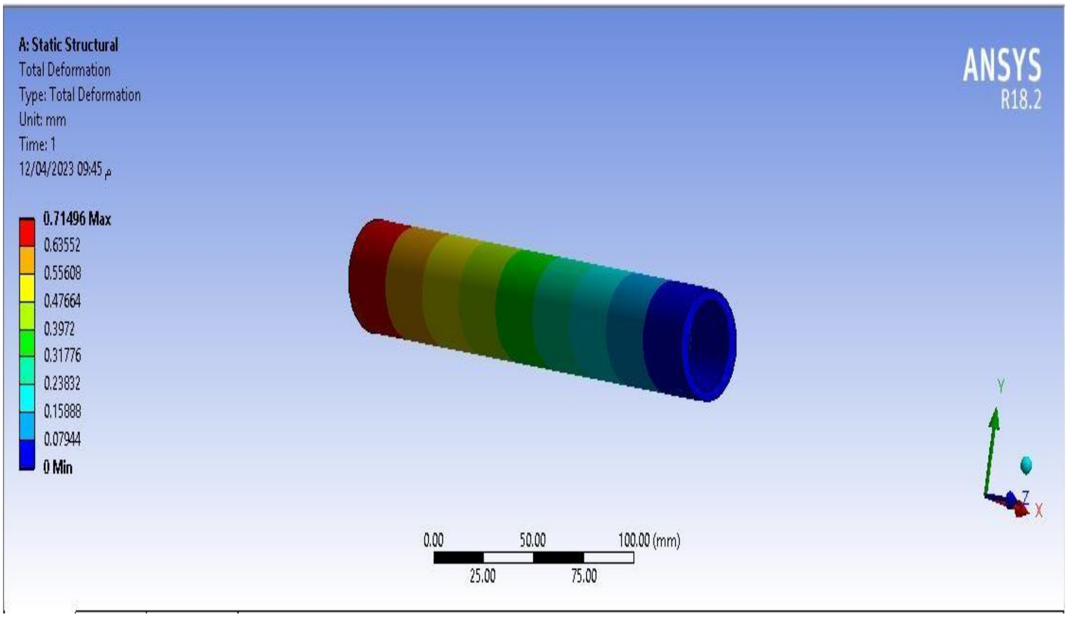

Fig (5.8) shows the maximum deformation of ellipse with 75 kg

Figure (5.9) shows the maximum stress through applied Wight was magnitude is
75kg , which noted the maximum on the shank is 5.4 Mpa

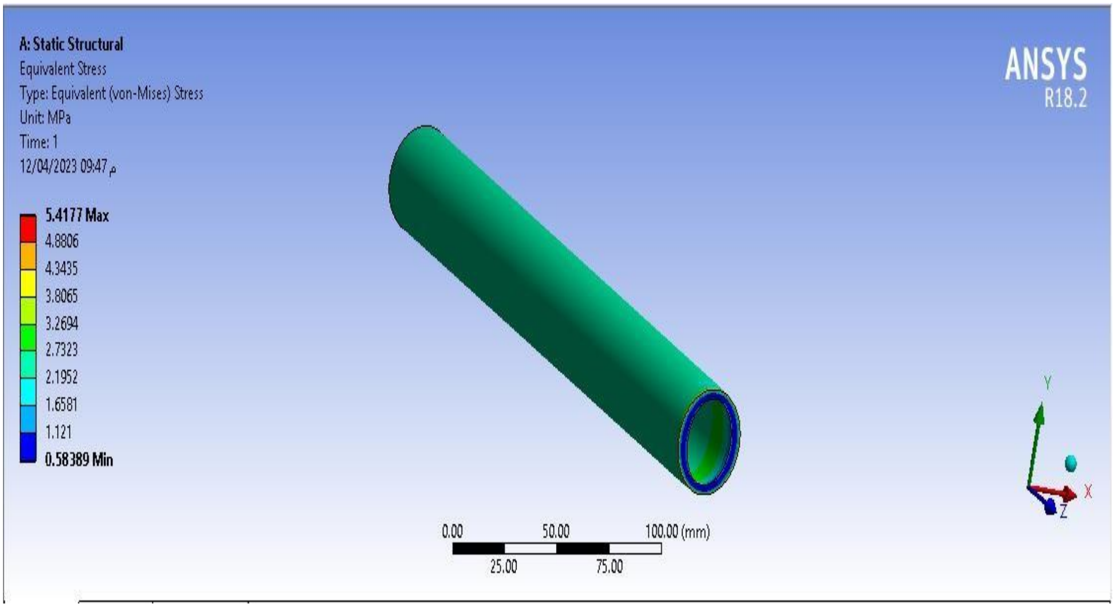

Fig (5.9) shows the maximum stress of ellipse with 75 kg

Figure (5.10) shows the maximum strain through applied human Wight's was magnitude is 75 kg , which noted the maximum stress on the shank is 0.00496

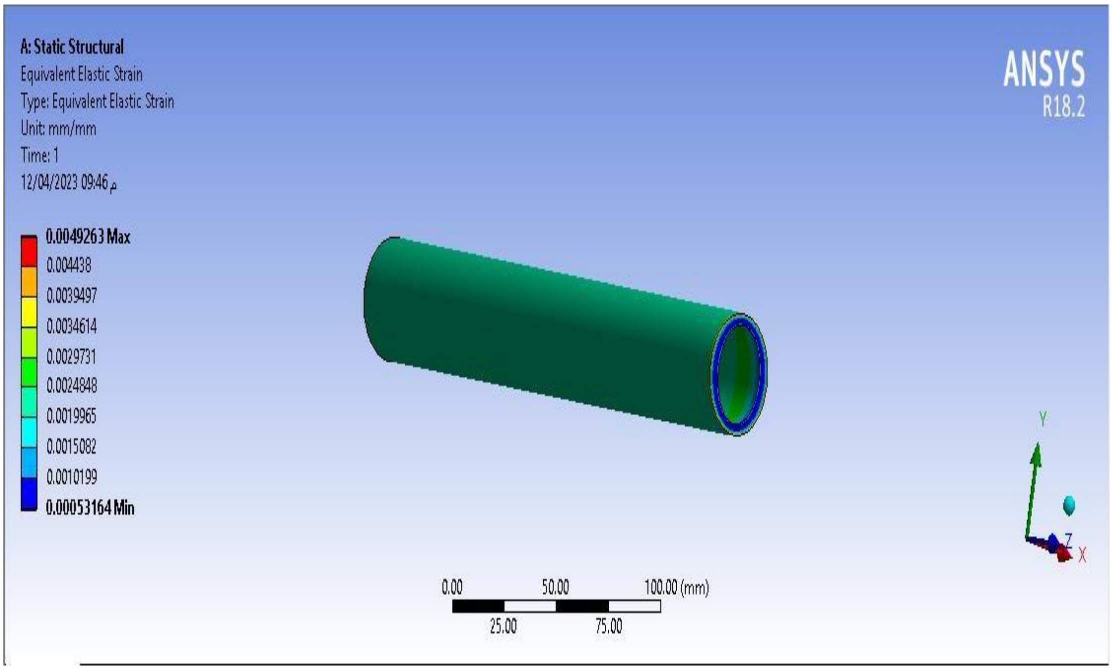

Fig (5.10) shows the maximum strain of ellipse with 75 kg

Figure (5.11 shows the range of safety factor through applied human Wight's was magnitude is 75 kg , which noted it on the shank minimum is 4.6

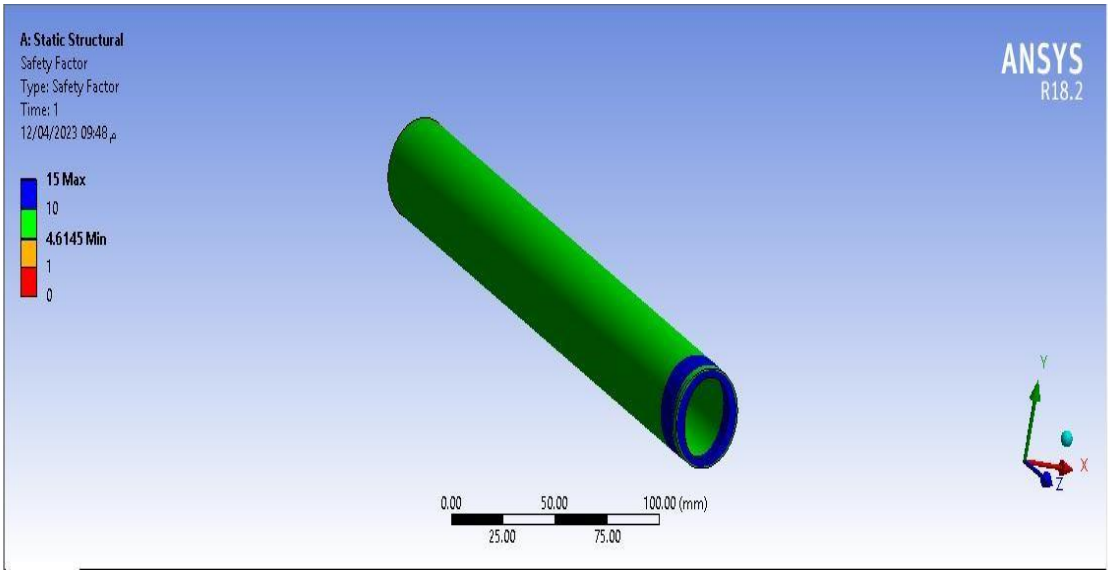

Fig (5.11) shows the range of safety factor of ellipse with 75 kg

Figure (5.12) shows the maximum deformation of hexagonal cross through applied human Wight's was magnitude is 75 kg , which noted the maximum deformation stress on the shank I is 7 mm

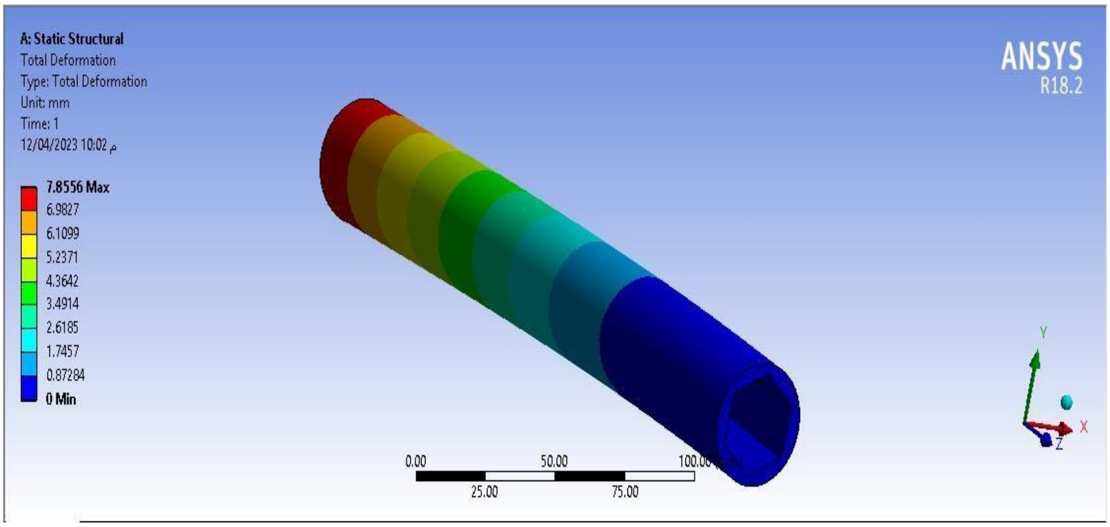

Fig (5.12) shows the maximum deformation of hexagonal with 75 kg

Figure (5.13) shows the maximum stress through applied wight was magnitude is 75kg , which noted the magnitude of this wight is 12.4 Mpa

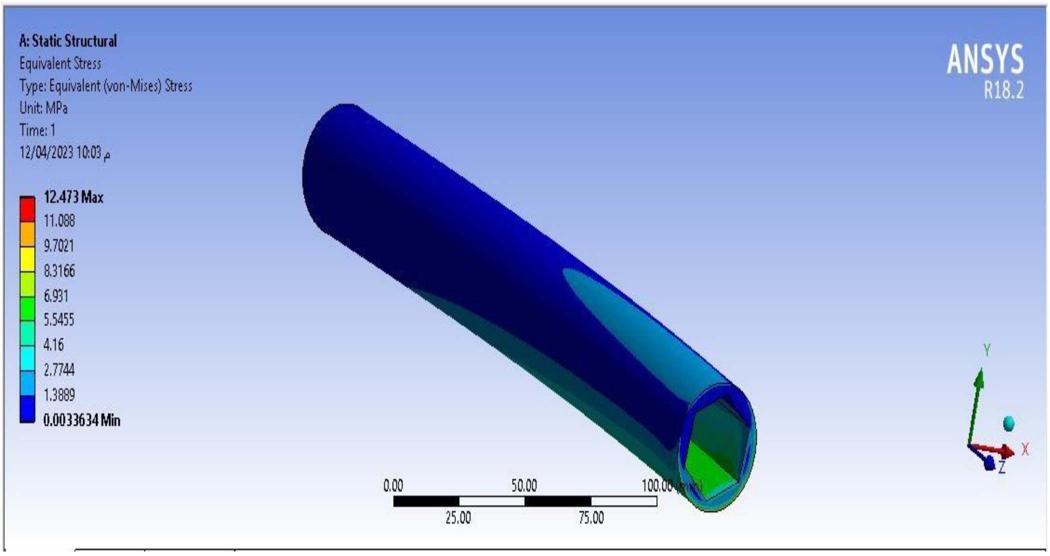

Fig (5.13) shows the maximum stress of hexagonal with 75 kg

Figure (5.14) shows the maximum strain through applied human Wight's was magnitude is 75 kg , which noted the maximum strain on the shank is 0.0113

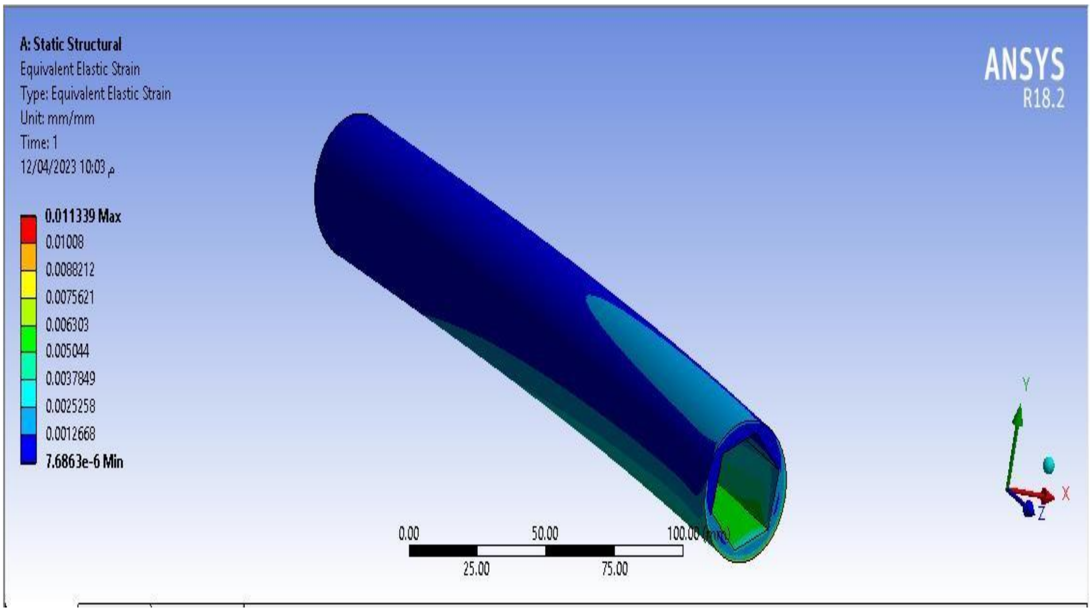

Fig (5.14) shows the maximum strain of hexagonal with 75 kg

Figure (5.15 shows the range of safety factor through applied human Wight's was magnitude is 75 kg , which noted it on the shank minimum is 2

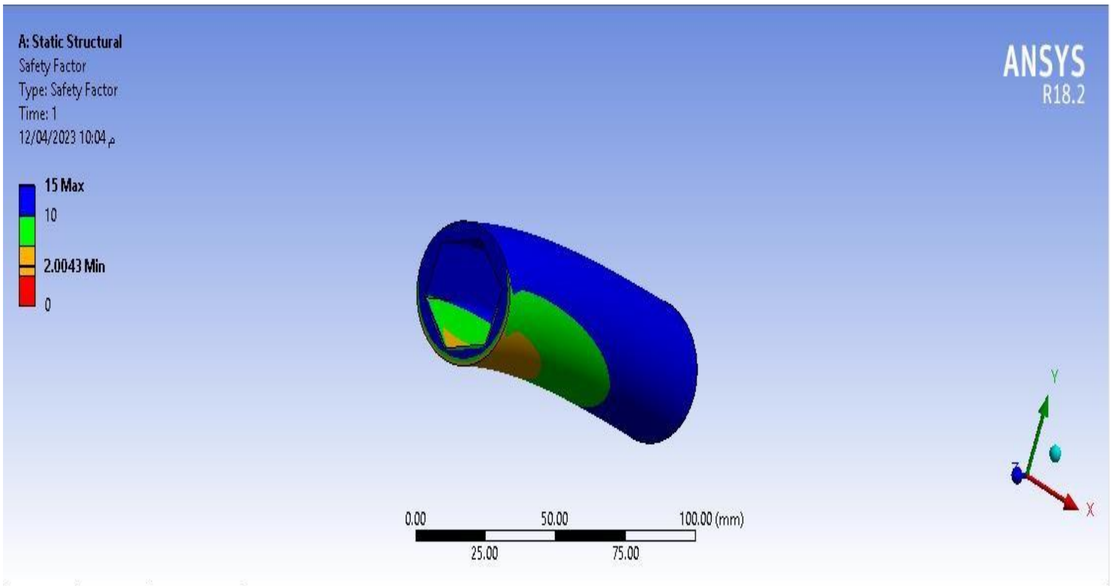

Fig (5.15) shows the range of safety factor of hexagonal with 75 kg

Cross section	Max deformation	Max stress	Max strain	Safety factor
Circle	**0.69**	**4.9**	**0.0045**	**5**
Ellipse	**0.722**	**5.4**	**0.0046**	**4.6**
hexagonal	**7**	**12.4**	**0.0113**	**2**

Table 5.1 show summary of maximum effect on shank with wight 75 kg

5.4 Case Two of Analysis With Human Wights is 100 Kg : -

Figure (5.16) shows the maximum deformation through applied Wight was magnitude is 100kg with circle cross section area , which noted the maximum deformation on the shank is 0.86 mm

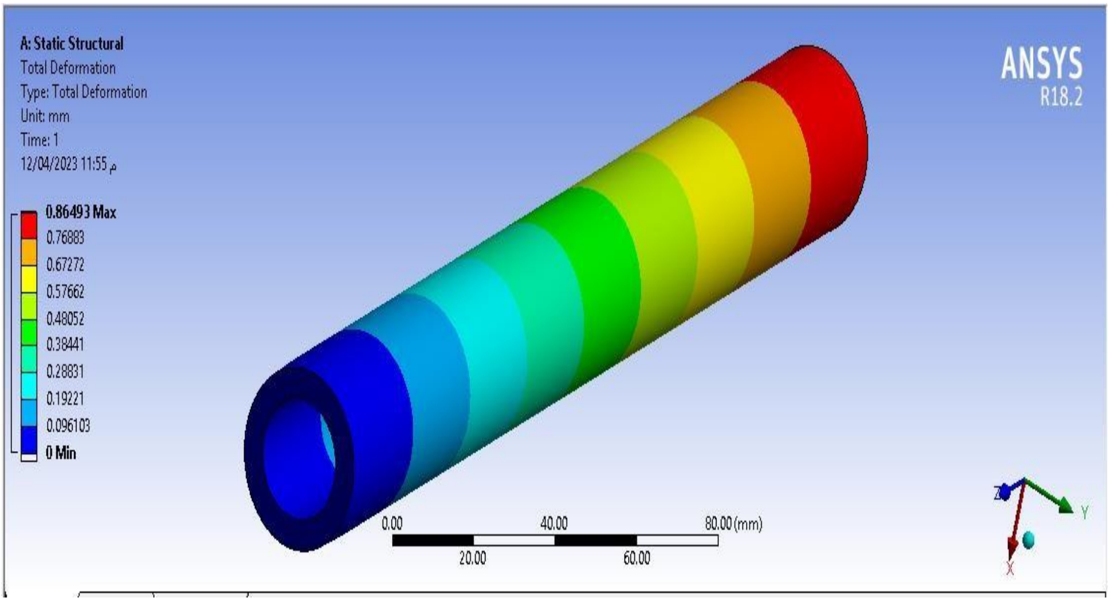

Fig (5.16) shows the maximum deformation with circle with 100 kg

Figure (5.17) shows the maximum stress of circle cross through applied Wight was magnitude is 100 kg , which noted the maximum on the shank is 6.2 Mpa .

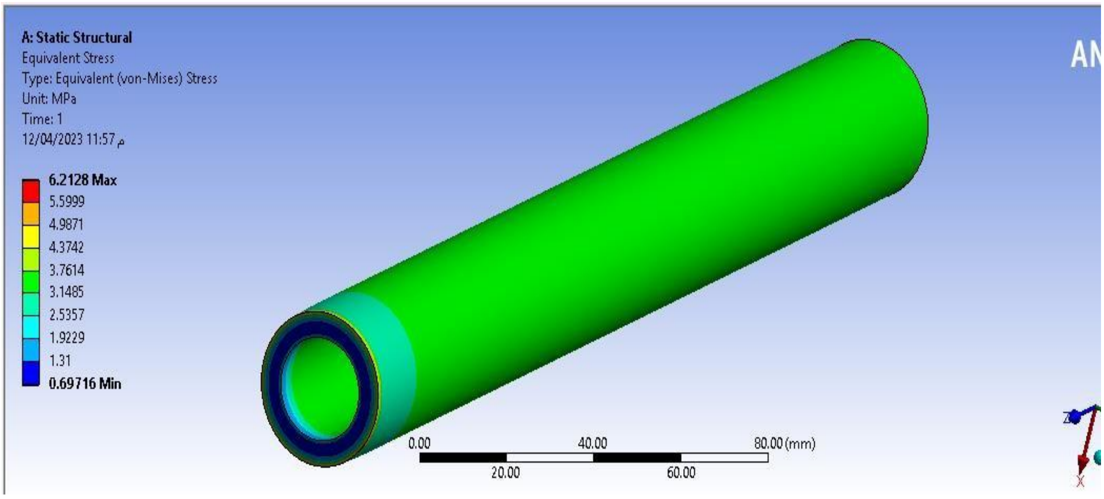

Fig (5.17) shows the maximum stress of circle cross with 100 kg

Figure (5.18) shows the maximum elastic strain of circle cross through applied Wight was magnitude is 100kg , which noted the maximum on the shank is 0.0055 .

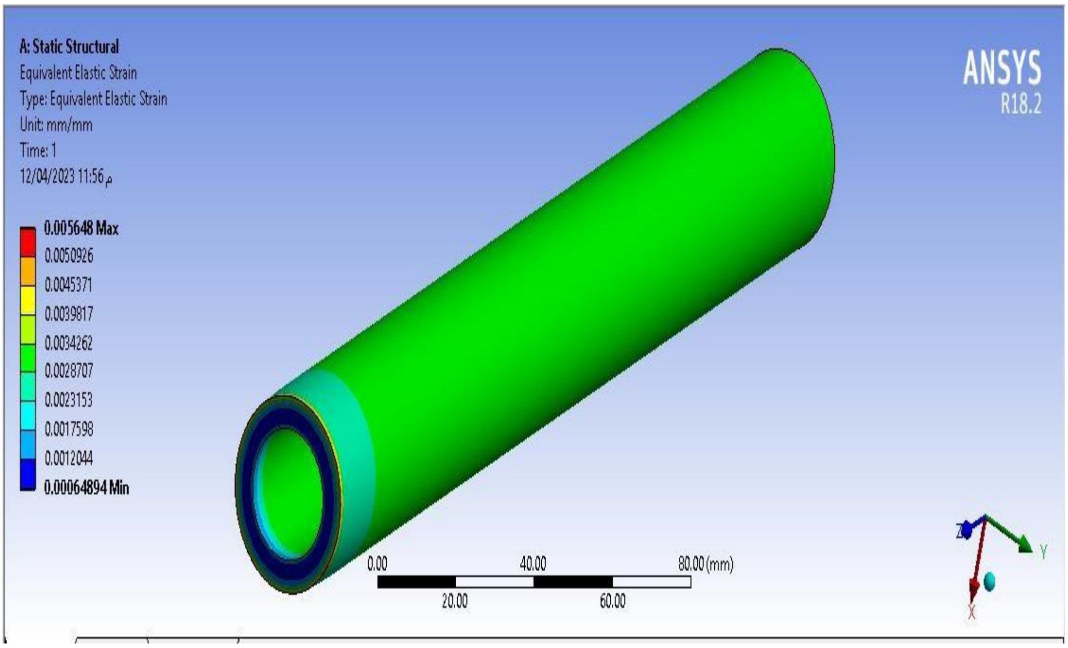

Fig (5.18) shows the maximum elastic strain of circle cross with 100 kg

Figure (5.19) shows the safety factor through applied Wight was magnitude is 100 kg , which noted the minimum is 4.02

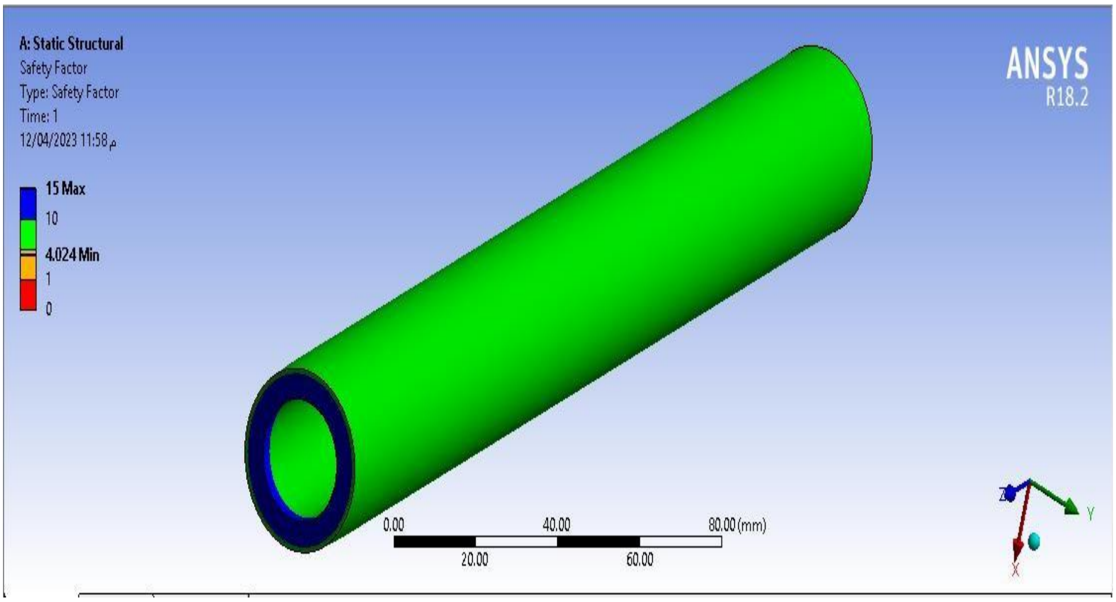

Fig (5.19) shows the safety factor of circle with 100 kg

Ellipse

Figure (5.20) shows the maximum deformation of ellipse cross section through applied human Wight's was magnitude is 100 kg , which noted the maximum deformation on the shank is 0.89 mm

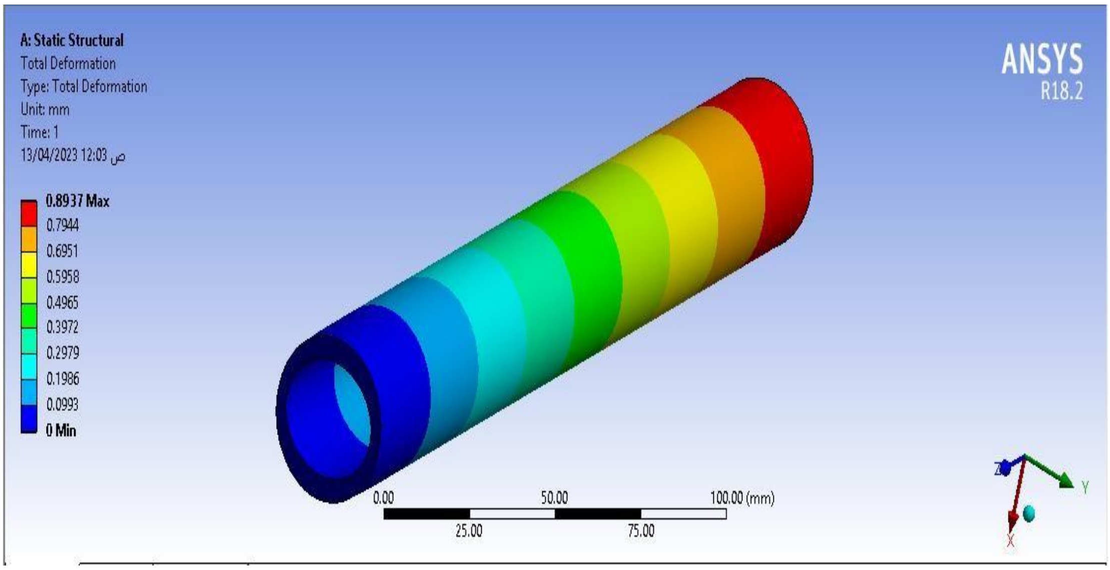

Fig (5.20) shows the maximum deformation of ellipse with 100 kg

Figure (5.21) shows the maximum stress through applied Wight was magnitude is 100 kg , which noted the maximum on the shank is 6.7 Mpa

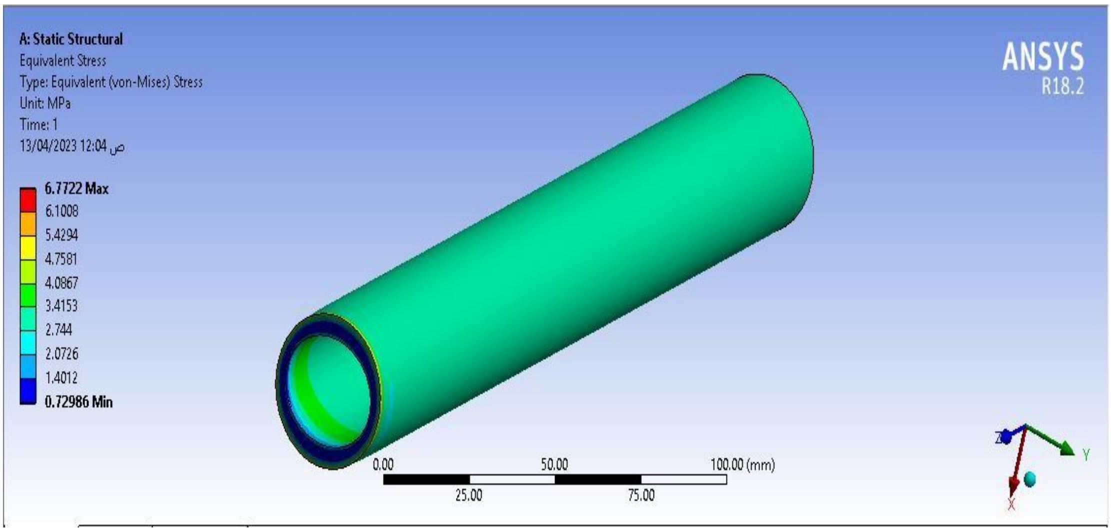

Fig (5.21) shows the maximum stress of ellipse with 100 kg

Figure (5.22) shows the maximum strain through applied human Wight's was magnitude is 100 kg , which noted the maximum stress on the shank is 0.006

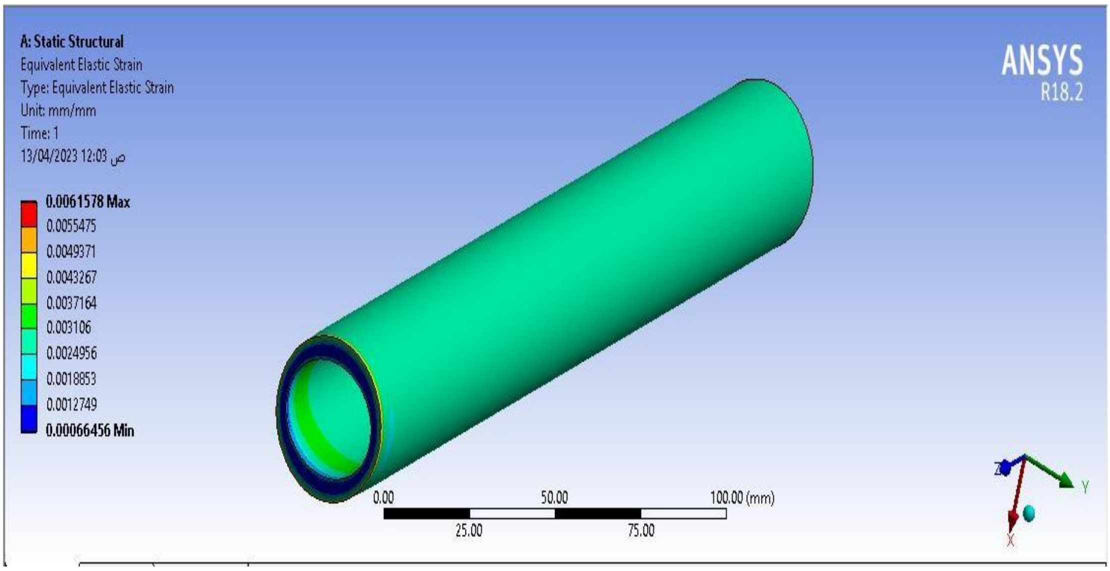

Fig (5.22) shows the maximum strain of ellipse with 100 k

Figure (5.23) shows the range of safety factor through applied human Wight's was magnitude is 100 kg , which noted it on the shank minimum is 3.6

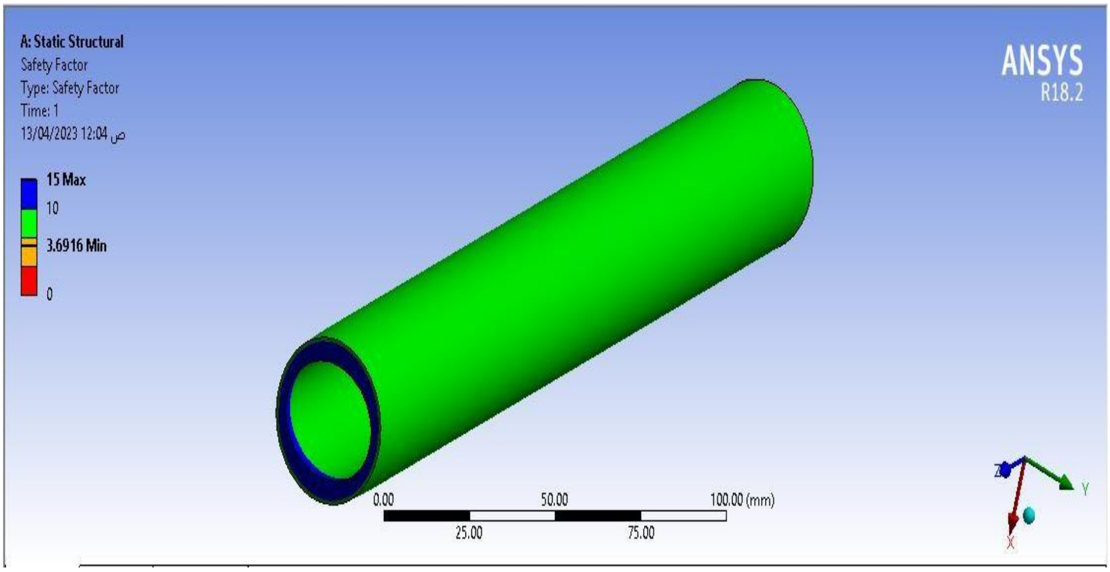

Fig (5.23) shows the range of safety factor of ellipse with 100 kg

Hexagonal

Figure (5.24) shows the maximum deformation of hexagonal cross through applied human Wight's was magnitude is 100 kg , which noted the maximum deformation stress on the shank I is 9.8 mm

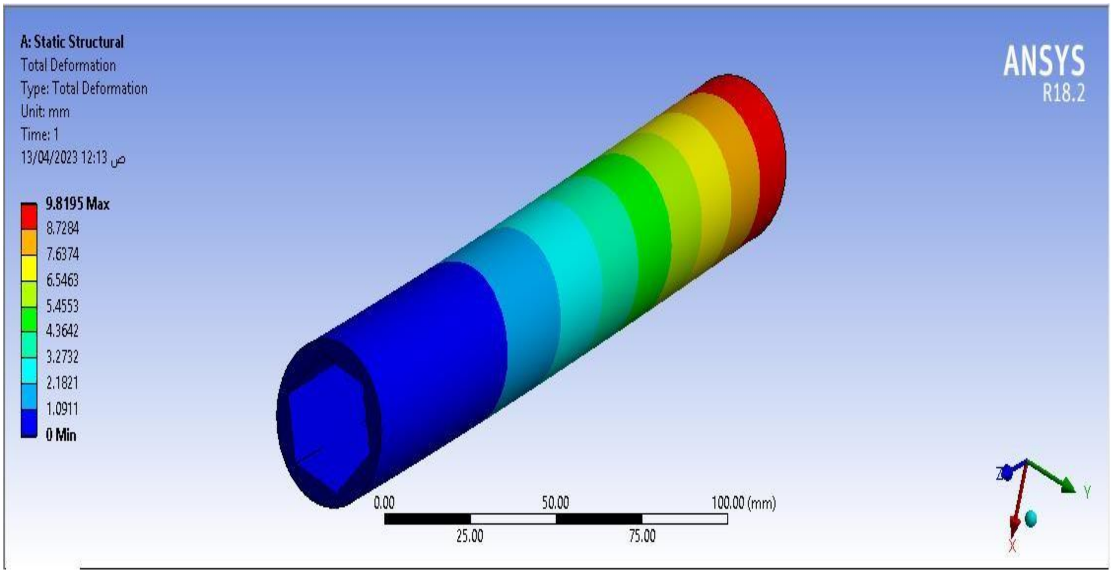

Fig (5.24) shows the maximum deformation of hexagonal with 100 kg

Figure (5.25) shows the maximum stress through applied Wight was magnitude is 100 kg , which noted the magnitude of this Wight is 15.5 Mpa

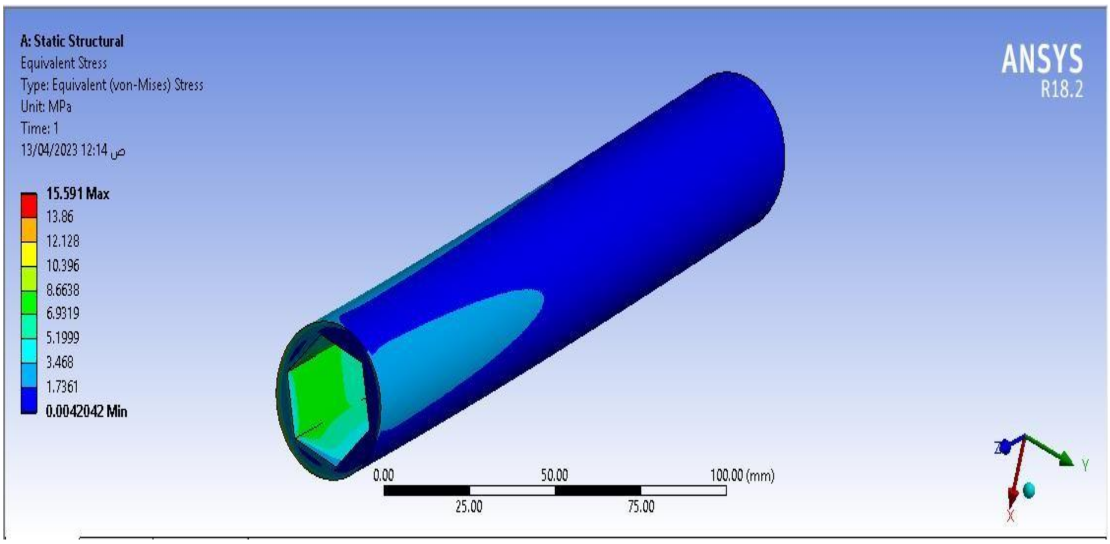

Fig (5.25) shows the maximum stress of hexagonal with 100 kg

Figure (5.26) shows the maximum strain through applied human Wight's was magnitude is 100 kg , which noted the maximum strain on the shank is 0.0141

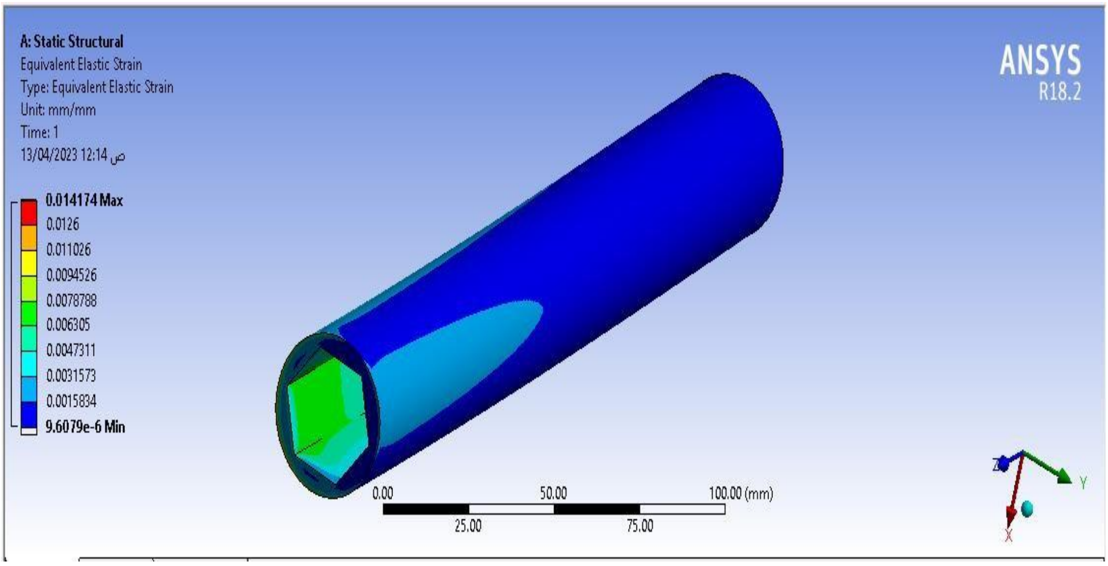

Fig. (5.26) shows the maximum strain of hexagonal with 100 kg

Figure (5.27) shows the range of safety factor through applied human Wight's was magnitude is 100 kg , which noted it on the shank minimum is 1.6

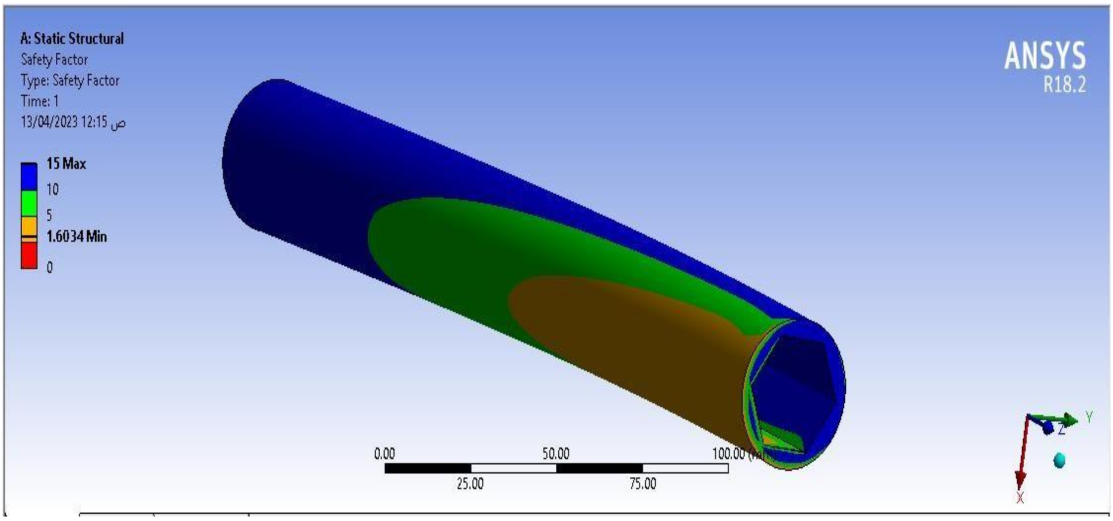

Fig. (5.27) shows the range of safety factor of hexagonal with 100 kg

Cross section	Max deformation	Max stress	Max strain	Safety factor
Circle	**0.86**	**6.2**	**0.0056**	**4.02**
Ellipse	**0.89**	**6.7**	**0.006**	**3.6**
hexagonal	**9.8**	**15.5**	**0.0141**	**1.6**

Table (2.5) shows the maximum result of weight is 100 kg

5.5 Conclusion of Results

1.The result from tensile test have been noted that the maximum tensile stress is 60 Mpa , while Young's modules stress is 4.2 Gap. This results have compared with other work [23] , which noted that yield stress for carbon fiber material is large than in other material is show in table (5.1).

2. in this work we have been observed that the result from ellipse and circle were very closing of all cases

3. the design of stress is less than of yield stress of all geometry and cases, it referees this material is high safety by comparison with the different design stress of geometry with yield stress

4. safety factor for circle & elipse were large than from hexagonal cross section for two cases.

5. There are large different of deformation of hexagonal with circle& ellipse, which observed that it reached to 7 mm &9 mm, while in circle &ellipse the magnitude is less than from 1 mm

6 . The case two with 100 kg no significant difference was observed when increasing the Wight with circle & ellipse, while in hexagonal the safety factor become very less is 1.06

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