ABSTRACT

Prosthetic foot is considered one of the most important parts of prosthetic limbs because it is in direct contact with the ground and supporting the body of amputee. Developing designs and materials of prosthetic feet are very important field so as to improve their characteristics to produce better satisfaction for amputees. In this work, previous designs are studied, and new design foot was modeled depending on Niagara foot design. Three groups of new composite materials are prepared. First group is manufactured from polyethylene as a matrix and chopped carbon fibers. Second group is prepared from blending two different grades of polypropylene (polypropylene grade 575, polypropylene grade 513), Third group is prepared from mixing calcium carbonate with polypropylene. The mechanical tests such as tensile, fatigue and impact tests are achieved for these new suggested materials to study the mechanical properties.

For the first group, the results of the tests [illegible] fiber with 20% volume fraction produced highest ultimate tensile strength and modulus of elasticity. The optimum results of toughness obtained from adding 15% volume fraction of carbon fiber, 28 .57KJ/m^2 for blend of 15% carbon fiber and 85% high density polyethylene, 30.76 KJ/m^2 blend of 15% carbon fiber and 85% linear low density polyethylene.

For the second group, the results showed that the highest toyghness of 55.74 KJ/m^2 is obtained for the blend of 50% polypropylene grade 575- plus 50% polypropylene grade 513.

The results for the third group showed that, when adding calcium carbonate to polypropylene, the material toughness is decreased while the modulus of elasticity is increased. Adding 0.5% calcium carbonate to

polypropylene, the material toughness is decreased by 40% and when adding 2% decrease the toughness by 50%.

The Niagara foot and the new design foot are analyzed by finite element method by using ANSYS15 software to compute the equivalent stresses, total deformation and factor of safety. The suitable range of amputees weight was determined and it was in the range of (40-75) Kg for Niagara foot, while for the new suggested foot design, it is in the range of (45-75) Kg.

The total deformation is calculated for toe off phase for both feet design to obtain the dorsiflexion. The results showed that, both feet give acceptable values of dorsiflexion.

For the calculations of safety factor, the results showed higher safety factor for Niagara foot than the new des[illegible] properties of Niagara foots material. T[illegible] suggested foot ranged from 1.4 to 2.4 while the results for Niagara foot ranged from 2.2 to 3.87 which means that the obtained results are acceptable according to design principles

CHAPTER ONE
Introduction

1.1General

Thousands of people every year lose a limb because of diseases landmines, cancer, diabetes, violence and automobile collisions .Global conflict is something that has been prevalent for years. Victims are still being counted for wars that have taken place years ago. Landmine have killed or injured over one million people since1975.Currently, there are over one hundred million active landmines around the world.

Landmine victims lose their place in society along with losing their limbs .Many countries see amputees as a burden since they lack mobility and the skills to work .Many organizations have gathered to combat this issue by having countries sign treaties against using landmines .The most famous treaty, which bans the use of landmines, was presented at the Ottawa Convention in 1997. [1]

1.2 Prosthetic limb components of Bel [illegible]

[illegible] of the lower limb prosthes [illegible] the shank transmits the bodyweight from the limb to the foot and then wrapped with a cosmetic soft foam cover. The parts of the prosthetic shown in figure (1-1) which include

1-Socket.

2-Pylon (shank).

3- Foot.

4-Adaptor.

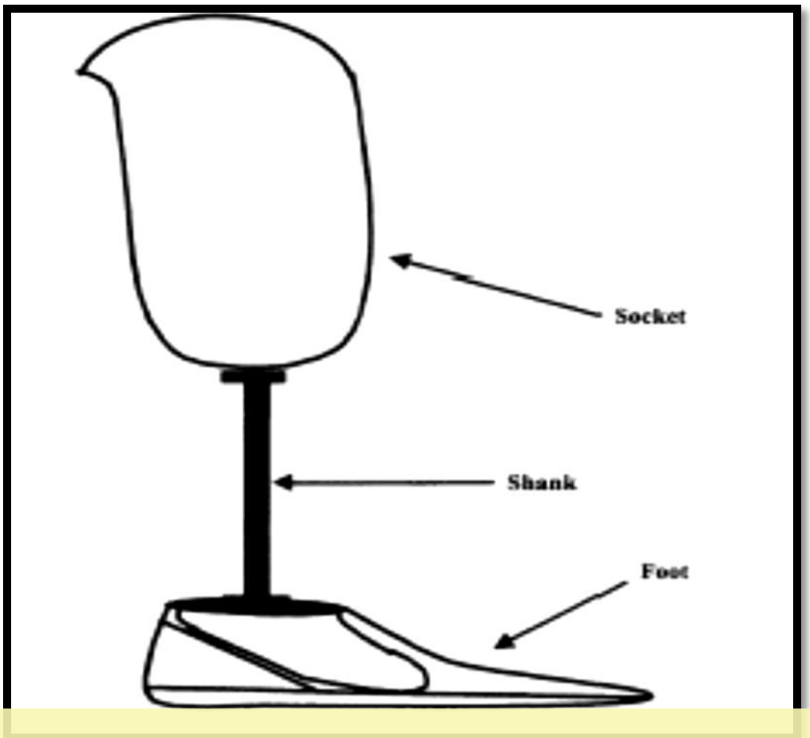


figure 1-1 Prosthetic limb components [2]

1.2.1 The socket

The socket is the interface between the patient and the prosthetic, it is important [illegible] responsible [illegible] [illegible] ed by discomfort [3].

1.2.2 The pylon (shank)

The pylon corresponds to the lower leg which used to connect the socket to the assembly of ankle-foot. In an endoskeletal pylon, a central shank, which is a narrow vertical bolster, rests in cosmetic foam cover **[3].**

1.2.3 The Foot.

Foot is considered from the complicated structures of a body, and it is an important part via which the body contacts with the ground. It offers high degree of activity and stability, and it's not just bears the weight of the body, but has multi-functions such as shock absorption, compression, extension, bouncing, tortuosity, and friction **[4].**

Current prosthetic designs provide a wide range of choices for amputees. The appropriate choice of prosthesis can significantly improve the comfort and performance of the patient. Most of the currently available prosthetic ankles, however, do not provide enough energy to propel the body forward **[5]**.

1.3 Prosthetic Foot Characteristics

The key factor in a new prosthesis designing is the analysis of amputee's response. This opinion is very important thing because if the prosthesis does not provide practical, cosmetically or functional acceptable characteristics, the amputee will not feel restful with it. The characteristics considered important by amputees to achieve normal gait include:

- Impact absorption.
- Eversion.
- dorsiflexion.
- Ankle torsion.
- Energy return.

1.3.1 FOOT FLEXION

The flexion of a foot measured about many planes include

- Sagittal: is the ability of the foot to rotate.
- Plantar: is the ability of the foot to bend down.
- Dorsi: is the ability of the foot to bend up.

1.3.2 Dorsiflexion

The dorsiflexion angle of the prosthetic feet considered one of the most important characteristics in designing prosthetic foot since it provide relief for amputee in toe off phase. The value of dorsiflexion depend on the material properties of the prosthetic foot ,weight of amputee and walking speed, type of foot, Table 1-1 shows values of dorsiflexion angles with different walking speeds.

Table 1-1 Dorsiflexion of a normal foot at different walking speeds.

Walking Speed (km/h)	**Dorsiflexion** (Degrees)
1 - 3 km/h	3 - 4°
3 - 5 km/h	5°
7 - 8 km/h	7 - 10°

*The acceptable range of walking (3-5° dorsiflexion) **[6]**

1.3.3 EVERSION

The ability of a foot to roll from side to side, it called inversion and eversion, it is important while walking on uneven surfaces. Foot must make compensations so as to balancing the person, as shown in Figure (1-2).

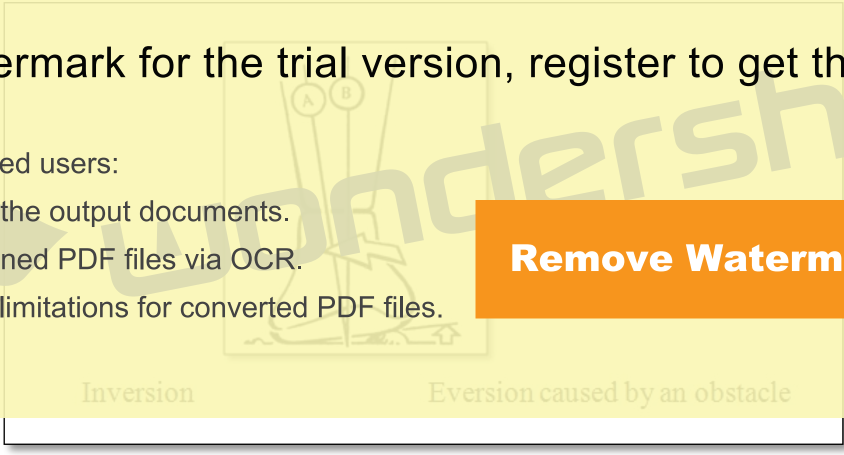


figure 1-2 Eversion and Inversion [6]

If a prosthesis is to reproduce motion of a normal feet then it should be show some eversion/ inversion characteristics. If an amputee was to walk on a stone, or any other obstruction, it would cause a displacement of the leg and result in a fall. Instead, due to the inversion / eversion characteristics of the prosthesis, the leg remains vertical and would cause the wearer not to lose their balance.

1.3.4 ENERGY RETURN

The capacity of a prosthetic foot to store the energy is very substantial to replicate the movement of a sound foot. Through the process of a sound foot, the energy is stored through the stance phase of walk and is progressive on transferal of weight.

In figure (1-3), the cycle of a sound foot is shown, the heel contact called HC, FF is mean a flat foot, heel off is HO and TO is toe off. At the time that the heel strikes the ground, the ankle will be in the neutral position. When the foot becomes flat on the ground, ankle must plantar - flex 12 to 15 degree. The foot becomes flat at 9% and at 63% of the cycle for the other foot. When the heel begins in leaving the ground, the foot will store the energy. The foot store energy so as to provide momentum for the prosthesis so as to rolling over the foot. Sound limb stores energy about of 14.18 Joules and releases about of 15.74 Joules, therefore its efficiency about 119.6%.

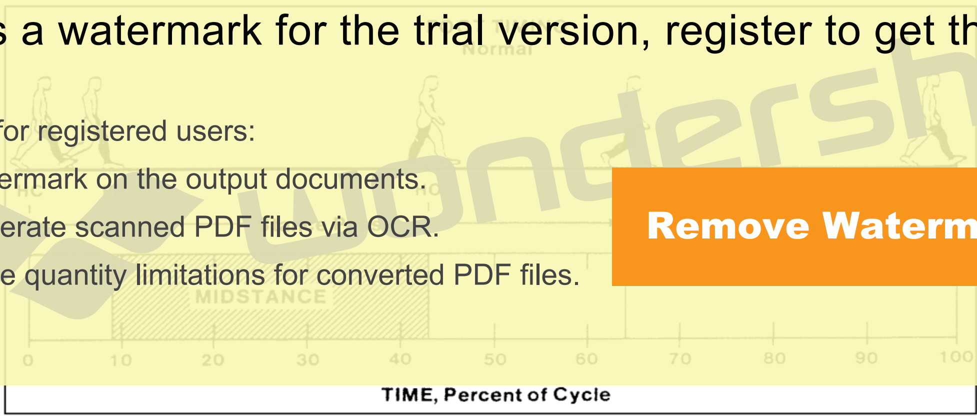


figure 1-3 The cycle of a human foot

1.4 Gait Cycle

Gait refers to how people propel themselves using their lower limbs. Gait cycle occurs in the time when the foot impacts the ground. [7].This process divided into two phases, stance phase and swing phase, as shown in Figure (1-4).

In the stance phase, the foot has been in contact with the ground surface. This typically makes up 60% of the cycle. Swing phase makes up the 40% remaining of the gait; it happens when there is no contact between the feet with the ground surface [8]. Twice during each gait cycle, the feet contact with the supporting surface; this is indicated to as double support.

In a normally walking speed, every period of double support takes up 10% of the gait cycle; so 80% of the time of gait cycle, a person's body is supported by just one limb [9]. The stance can be further being divided into three parts: firstly heel strike then mid stance and finally push off.

At heel strike phase, the heel firstly be in contact with the supporting surface. In a typical subject, the horizontal velocity decreases to 0.4 m/sec while the vertical velocity decreases to 0.05 m/sec [10].

In abnormal gait, the heel may not be the first part of the foot that contacts the ground; it could be the toes or the whole foot [7].

Both the midstance and push off phases can be further broken down into subsections. During the midstance phase, foot flat and the midstance point occur. Foot flat indicates to the first time at which fo[illegible] after heel strike phase at nearly 7% of the cycl[illegible]

The midstance point when a bodyweight of a person is over the supported limb, 30% of the way out of the cycle.

The push off phases include heel off phase and toe off phase. Through heel off phase, the heel leaves the ground, at 40% of the cycle. Then, the toe leaves the ground (toe off) at nearly 60% of the gait cycle.

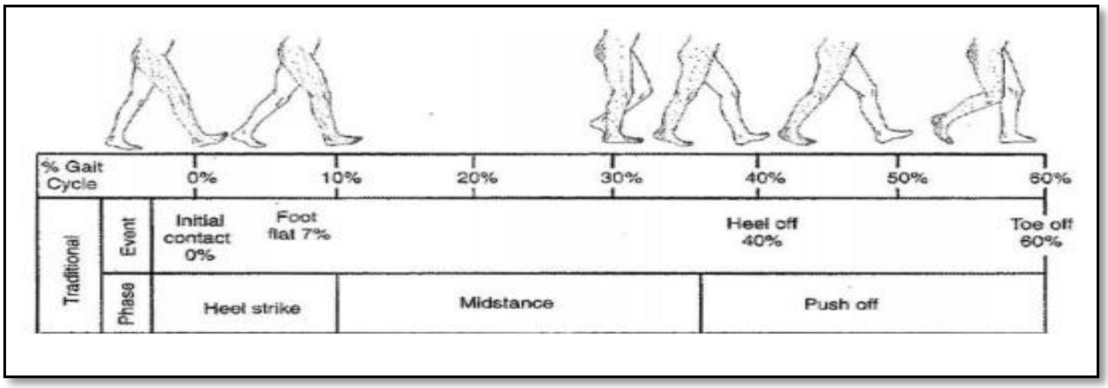


figure 1-4 The stance phase of the gait cycle for the right lower limb progressing. [7].

Swing can also be broken down into three phases: acceleration, mid swing and deceleration.

Once the toe leaves the ground'8, the leg starts to increase the angular speed; the acceleration continues till mid swing phase. Mid swing phase begins when the leg has been directly beneath the body and then continues until deceleration. During deceleration at late swing, the leg beings to decrease its angular speed to preparing for the heel strike phase [7].

1.5 Thesis Objectives

The main aim of this work is to suggest a new type of composite material which be available locally, cheap, has a good mechanical properties, can sustained the fatigue loading and offers manufacturing suitable prosthetic foot reliable to be used in the center of rehabilitation and manufacturing prostheses.

In order to achieve the above objective, the following steps are followed.

1-Prepare blends of composite materials have good characteristics such as light weight and high mechanical characteristics used for manufacturing prosthetic feet.

2- Study the mechanical properties of the new composite materials by using different mechanical tests such as tensile test and impact test.

3-Design a new prosthetic foot which has good characteristics such as accepted dorsiflexion and light weight.

4- Compare the new suggested prosthetic foot with Niagara foot by analyzing them theoretically.

1.6 Thesis Outline

The general theme of the thesis is covered by six chapters.

Chapter one includes a general introduction of amputation and its causes, foot in general, prosthetic foot, gait analysis, prosthetic foot characteristics. It also introduces the main objective to be achieved.

Chapter two deals with a literature survey of the optimal design of composite prosthetic foot generally done by others.

Chapter three includes the theoretical study, the numerical solution which has been done by the using of computer software ANSYS Workbench 2015 to calculate the von MISES equivalent stress, t[illegible] factor of safety.

Chapter four includes the experimental work which involves the tensile test, fatigue test, and impact test to determine the mechanical properties of the composite materials.

Chapter five contains the numerical, theoretical, and experimental results achieved by comprehensive analysis. In addition, the results are discussed.

At last, **chapter six** provides the main general conclusions that can be drawn from this work and hence the suggested possible recommendations for the future work in corresponding fields.

CHAPTER TWO

Literature Review

2.1 Introduction

The prosthesis of earliest cultures began as uncomplicated crutches or leather and wooden cups. Prosthetic foot was designed with the aim of restoring simple occupational tasks and basic walking. Athletic or active amputees need more than ambulation from their prostheses. These amputees have additional aims of being able to jump, run and take part in sports. The requirements for prostheses eligible of higher performance, more suitable shaped the design and easy to manufacturing **[11]**.

2.2 Literature Review on Prosthetic Foot and Composite Materials

Arya et al. (1995) have studied comparing biomechanically between Seattle, Jaipur and SACH fee[illegible] [illegible] as they represent the most use[illegible] generally. The study was to estimation the performance characteristics for Jaipur foot by compare its shock absorption ability by using the ground reaction forces and influence on gait style in compared with SACH and Seattle foot. The conclusions of this study can be stated as follows:

1. The performance of Jaipur foot was nearer and more natural to the normal foot as comparing with Seattle and SACH feet.
2. The shock absorption capacity for SACH foot was better than the Jaipur and Seattle feet.

There were no other considerable differences in of gait produced by the Jaipur, SACH or Seattle feet **[12].**

Francis J. Trost (2000), investigated various materials that have ability to store energy when compressed by the weight of the body through early stance phase. The analysis includes measurement of the consumption of oxygen and the determinant of gait. Fifty two amputees were studied, energy storing feet types were including Sten foot , Seattle foot, Carbon copy foot, and Flex-foot.

In evaluating particular activities,about most patients responded that climbing stairs , jumping , running, were easier with energy storing feet**[13]**.

Bryant and J. T Bryant (2002) Compared Niagara foot with SACH foot, and concluded no significant difference between two types of feet in walking speed, the purpose of this observation to determine the standard activities for sample amputee and to comparing these to the testing activities and the design for the foot. The two observed amputees were male, the first was a barber and the [illegible] were working in a standing position on [illegible] periods, there was a small amount of load bearing. The farmer mostly was using a tractor for most activities; the dynamic load exceeded that suggested in ISO 10328 for that cyclic testing **[14]**.

Glenn K. Klute, et al (2004), studied the properties of heel region of shoes and prosthetic feet. To compute and model the heel in impact response, a pendulum used to simulate mechanically the conditions following immediately at initial ground reaction at heel contact through the walking. The mass of pendulum was 6 kg used to duplicate the effective mass of the stance limb at the moment of ground reaction at heel contact.

The velocity directly earlier to impact, by using two fiber optic photoelectric sensors and the energy dissipation capability of different prosthetic feet was calculated using a diagram of force deformation**[15]**.

SAM et.al.(2004) ,investigated a mechanical dynamic properties of eleven prosthetic feet, including resonant frequencies and damping ratios characteristics. The data of walking illustrated that, the feet could be modeled by using their characteristics, since damping ratios and walking frequencies were small comparing with the natural frequencies. The feet deformed under loads of walking, and this computed by using a technique of loading and a spatial transformation of the ground reaction force's center of pressure. For most prosthetic feet, the roll-over shapes studied were similar to that of the SACH prosthetic foot **[16]**.

Andrew H. Hansen et al. (2004) achieved the foot effective length ratio for different feet such as Niagara fo[illegible] foot length ratio(EFLR) multiplied by 10[illegible] of a prosthetic foot.

Effective foot length is measured by computing the distance from the center of pressure to the heel for each prosthetic foot. The Effective foot length ratios of the prosthetic feet were between 0.63 and 0.81**[17]**.

Al-Kinani K.K. (2007) investigated a lot off commercially available feet. All prosthetic feet attempt to return some of the lost gait functions, but may use different mechanical principles to do so, such as **SACH** foot (commonly in Iraq), therefore, the researcher designed prosthetic foot.

This prosthetic foot is designed and manufactured from polyethylene and a comparative study with SACH foot was used to determine if there were differences in the gait pattern while wearing the new foot and whether these differences would be problematic. It has good characteristics when compared with the SACH foot, such as good dorsiflexion (4.2^{o} and 1.9^{o}), stored energy return (58.9 and 13.14), force transmitted at impact heel (154N and 205N), the effective length ratio (0.76 and 0.64) and life of foot (1233417 and 896213) cycles **[18]**.

Anne Schmitz in 2007 used a Niagara foot model with finite element methods (FEM) to analyze mechanical properties. The stiffness responses of toe off and the heel were obtained using ISO 10328 by applying displacements at a rate of 5mm/minute by a [illegible] of load [illegible] d15° on the toe and heel respectively. The maximum applied force was 1600 N. The applied [illegible] were recorded using a data acquisition syst[illegible]

Nolan (2008) reviewed improvement carbon fiber prostheses. The aim of improving the prostheses was to try increase the performance by reducing the functional disadvantage of the prosthetic foot compared to the human foot. The study would be taken by used carbon fiber prostheses and their effect on the running technique of transtibial amputees because of effect of carbon fiber prostheses on amputee running is limited by the number of studies. The chosen speeds of running and current prostheses of running do not equivalent the normal foot in terms of efficiency of energy due to having to decrease loading on the residual limb **[20].**

Carpenter et.al. (2008) designed a low cost prosthetic foot by Center International Rehabilitation (CIR). It was made by from 90% PP and 10% PE, CIR has developed a lower limb prosthetic (mono-limb) for transtibial amputees and studied the effect of coupling their mono-limb with two existing prosthetic foot designs SACH foot and North western's Shape and Roll (SR) prosthetic foot. The results were then compared to the physical testing of the prosthetics under various static loading conditions seen during walking. The results showed that the SR foot strains of greater magnitude than the SACH and Von Misses stresses do not exceed the tensile yield strength of the PP with either assembly. This led to the conclusion that based on the testing conducted that both feet were acceptable during static loading conditions in spite of their higher strains [21].

Ali. A. A. (2010) investigated a non-articulated prosthetic foot and all its related characteristics. These characteristics involved in normal walking include: [illegible] impact abs[illegible] characteristics displayed in the manufactured non-articulated foot compared with those of SACH foot. A non-articulated prosthetic foot was designed and manufactured from polyethylene and a comparative study with SACH foot is used to determine if there are differences in the gait pattern while wearing the non-articulated foot, and whether these differences would be problematic.

The basis of the prosthetic foot design combines the current prosthetic foot design elements, such as prosthetic components and materials.

The analytical part presents the results of the static and fatigue analysis by methods, such as numerical methods (Finite Element Method, FEM) and experimental methods. Thus, the non-articulated foot was designed, and the number of cycles, dorsiflexion and impact were measured. **[22]**

Scholz (2011) investigated the using of composite materials in a prosthetic devices and orthopedic medicine. The major advantages in using composites reinforced by fibers for orthopedic purposes are related with their biocompatibility and exceptional specific strength characteristics. parts of research are continues in the development of porous materials which are better suited for applications of biomedical field, as well as material processing techniques and advanced manufacturing process **[23]**.

Shao-Yun Fu & Bernd Lauke, The investigation researched an analytical method to studying the [illegible] distributions and fiber length of polym[illegible] (SFRP) to predict the tensile strength (TS). Two density functions were used to modeling the distributions of fiber orientation and fiber length. The strength of SFRP calculated as a function of fiber orientation distribution and fiber length taking into account.

The dependences of the critical fiber length and ultimate fiber strength on the angle of inclination and the effect of the inclination angle on bridging stress of the oblique fibers. Then the effect of the mean fiber length, most probable length called (mode length), the mean fiber orientation, the critical length of fiber, the probable fiber orientation and coefficient of fiber orientation on the strength of tensile of SFRP have been studied **[24]**.

K. S. Santhosh Kumarand et al., studied the modulus and mechanical strength of polybenzoxazine composites reinforced by chopped carbon fiber (CF), the investigation achieved by changing the fibers length. Flexural properties, compressive, and tensile were studied. The contents of voids were higher for the composites with short fibers.

With increasing fiber length, due to increasing in strength of tensile and optimized at around 17mm length of fiber while strength of compressive exhibited a continuous diminution. Strength of flexural increased with length of fiber and the optimization results found at around 17mm length of fiber. The increasing in the composites strength with length of fiber is attributed to the increasing in effective area of contact of the matrix with fibers.

The results of experimental investigation showed that increase in strength of tensile [illegible] 470% and the flexur[illegible] [illegible] comparing with neat polybenzoxazine, while the properties of compressive were affected adversely. The composites showed a maximum increase in flexural modulus of about 800% and 200% in modulus of tensile. Enhancing the length of fiber, due to fiber entanglement in the composites, which resulted in increasing in plastic deformation at higher value of strain **[25]**.

J. Michael Starbuck et al. investigated the effect of fiber length, tow size, volume fraction, profile radius on mechanical properties of chopped carbon fibers in composite material and found that higher strength of tensile and higher modulus of tensile resulted with smaller contents of chopped carbon fiber. Lower strengths of tensile and stiffness's were obtained with higher volume fraction and shorter length of chopped fiber were used. From the tests of compression, smaller areal density panels results higher strengths of compressive and strains of failure comparing with larger areal density panels. The effects of fiber volume fraction and fiber length on maximum strain, stiffness and strength compressive was inconclusive. Consistent with the data of compression and tension tests, the flexure data shows that the higher stiffness and strengths resulted when testing a smaller density panels. The effect of volume fraction of fiber on the pure tension is opposite that of flexural response, where the higher volume fraction of fiber tests resulted higher stiffness and strengths of flexural. The effect of length of fiber was higher stiffness of flexure and lower strength of flexure when shorter lengths were used. The property data showed that a randomness orientation of the chopped carbon fiber and a non-homogeneous material system and **[20]**.

Adalarasuet.al. (2011) studied the comparison on SACH, Jaipur, and Madras foot. This investigation encompassed many different parameters for prosthesis selection. Analysis and Comparison were done by obtaining the values of property of the material tests for the common materials used in different feet and simultaneously verifying these results through subjective feedback. By performing study theoretically as well as analysis of material of the three available commonly prostheses, properties of material such as shrinkage, relative density, flex crack growth and hardness etc. were quantified. The results compared with feedback subjective received from amputees to obtain the best configuration for different conditions. Jaipur foot was served more than one million amputees in the abroad and nation providing durable comfortable and light weight. Moreover, Jaipur foot is time tested and has been proved on the grounds of properties of material like tensile strength, hardness, relative density ,abrasion, resilience and hence outweighs SACH foot and Madras foot in performance and composition as well, be it comfort while walking, occupation retrieval rate, aesthetics, lesser exhaustion rate, and be it socio-cultural acceptance **[27]**.

Muslim M. Ali. **(2011)** designed and manufactured a non-articulated prosthetic foot from polyethylene and a comparative study with SACH foot is used to determine if there are differences in the gait pattern while wearing the non-articulated foot and whether these differences would be problematic. The non-articulated foot is compared with SACH foot in cost and weight, and it is shown the cost of new foot is lower than that of the other kinds of by about (80%), also it was found that the new weight is lighter than that of the other of by about (25%) [28].

Anderson et.al. (2012) designed a new prosthetic foot by The Centro de Miembros Artificiales (CMA) from (PE+PP) copolymer .These models were used to refine the dimensioning of the designs and to run static FEA stress analysis on the models. Analyses showed where stress concentrations occurred on each design and allowed for the stress levels to be compared between the designs. The simulations were done using loads that mimicked what was seen in a natural foot. The final design was determined based on the performance of each of the designs under the static models as well as its historical performance based on prosthetics research or the experience of the CMA. The final design incorporated elements of three different initial designs used on the Shape and Roll foot at the CMA and was performed well in the field **[29]**.

Hansen A., et al (2012), investigated a standard height of heel shoe to fit with College Park foot shells. Prosthetic feet S&R conforms to the suitable rocker effective shape for walking through closure of saw cuts in the forefoot. The “flat region” was adjusted by blocking 0, 2 and 4 cuts.

They were having amputees with bilateral and unilateral transtibial amputations. Tests of balance performed by using a long force plate, Neurocom Smart Equitest and Clinical Research System. Motor control test, limits of stability, sensory organization test, tests of mobility (speed of Walking). Participants asked to provide a ranking preference of the conditions of foot **[30]**.

Hassan S. Mohammed (2015): concerned developing two types of composite materials, which can be used in manufacturing the prosthetic foot. It was found the Young’s modulus of 40% high density polyethylene (HDPE) filled with 60% [illegible] significantly increased to 90% compared [illegible] HDPE. Moreover, the yield and ultimate stresses were improved, which was approximately 2 times higher than that observed for neat HDPE,. On the other hand, it has been noticed that adding 10% of linear low density polyethylene (LLDPE) to 90% HDPE increasing from elongation at break to 27% compared with those values of pure HDPE while the ultimate , yield stress and impact energy have been decreased.

The new two models of prosthetic foot of (HDPE and LLDPE) and (HDPE and DPW) have good characteristics, such as good dorsiflexion and life of foot cycles respectively ($9^{o}$and 2,193,228 -7.5^{o} and 1,049,135) respectively when compared with the SACH foot of dorsiflexion (6.4^{o} and 896213).

The results of two models were demonstrated that added to get better ability of design prosthetic foot to match the mechanical properties of prosthetic foot to amputee specific parameters, including, abilities, needs and biomechanical characteristics **[31]**.

2.3 Concluding Remarks

From the literature review, it is clear that many studies has been carried out on the prosthetic foot as comparison between different types of foot ,developing designs and composite materials for manufacturing prosthetics feet, studied foot length ratio for different types, the failure of SACH foot at toe region ,and investigating the effect of chopped carbon [illegible] previous study concentrated comparison between types of feet but in this work study focused on developing different new composite materials and [illegible] prosthetic foot [illegible] with high characteristics with low cost, this design is derived from Niagara foot design which considered a good alternative for developing countries because of its simple design and weather resistance characteristics.

CHAPTER THREE

Theoretical Analysis

3.1 Introduction

Many prosthetic feet designs are available. These designs of prosthetics feet serve main functions such as: absorb shock through heel contact , supportting the body through walking and standing and mimic metatar sophangeal function during the stance phase of gait, preventing the fatigue failure, principle of energy storing as the stance limb accepts the weight of the body and returns the energy when the foot lift off from the ground and good lifts off the ground and good eversion and dorsiflexion .

Theoretical model and Analysis for the mechanical behavior of Niagara and proposed [illegible]

3.2 Prosthetic Foot

The stiff part structural of a prosthetic feet is called the keel. In single unit designs, the keel is incorporated in the rest of the prosthetic foot. In some designs of prosthetic foot, the keel is inserted into a separate cover which allows replacing the cover when needed, with using the same keel [32].

Prosthetic feet can generally be placed into one of two categories: Non Articulated and Articulated Feet

3.2.1 Non-Articulated Feet:

Two common non-articulated feet are the SACH and the SAFE. The solid ankle-cushion heel foot referred to as the SACH foot has a rigid keel. The keel, which can be seen, supports the frame of the foot. In walking, a normal ankle swings about a hinge, but in the SACH foot the ankle is rigid and cannot swing. Therefore the soft rubber heel provides the ankle action by compressing during the early phase of walking. The rubber heel wedges are available in soft, medium, and hard densities. The different densities allow for different gaits [33].

The SAFE foot has the same characteristics as the SACH foot with added the ability for the sole to conform on uneven surfaces. The flexibility of the sole allows for an easier and more comfortable style of walking [33].

Generally Non Articulated feet are basic designs that have no moving components. Such as Solid Ankle Cushion Heel feet which known as SACH foot, this type of rigid feet are without ankle articulation where the forefoot simulates the dorsal flexion of the feet and the heel damps the shock.

SACH foot has a rigid plastic or a wooden keel that extends until the toe section. Dense foam makes the remainder heel and of the prosthetic foot is rubberized foam. The Belting is attached to the keel end and extends into the forefoot region [34] [illegible]

figure 3-1 SACH Foot

3.2.2Articulated Feet

A foot could be with a certain range of liberty in only sagittal plane to simulate movements of planter- and dorsi-flexion, or in sagittal and frontal planes to simulate planter- and dorsi-flexion as well as pronation and supination. There are feet with different degrees of amortization of all those movements.

Advantage of imitating the normal movements for the ankle is not just for cosmetic reason but also for a functional one: after heel contact, the instantaneous forefoot contact happens, promoting the ground reaction forces that keep the extension of the prosthetic knee which can be used for uniform prosthetic treatment of the plurality of patients, in particular for TF patents.

3.2.2.1 Single-Axis Foot:

As the name suggests, a single-axis foot has a hinge or other mechanism that allows the foot to plantar flex and dorsiflex, as shown in Figure (3-2).

Single-axis feet were the first prostheses to provide ankle articulations. They typically consist of a keel, with an ankle joint and a molded foot shell. The keel has a plantar flexion bumper located in the heel behind the ankle joint. Some feet have a second dorsiflexion bumper anterior to the ankle. Feet that do not have this second bumper have a dorsiflexion stop. Similar to the SACH foot, the keel is attached on the keel end and extends to the forefoot

figure 3-2 single axis foot

3.2.2.2 Multi-Axis Foot:

A typical multi-axis design shown in Figure (3-3), has a multi-directional hinge that allows for eversion and inversion as well as plantar and dorsiflexion. The Endolite Multiflex, a newer style of multi-axis foot shown in Figure 8d, has a rubber ball inside the stem of the ankle assembly with an O-ring sitting just below it. This system allows for some rotation of the foot, in addition to eversion, inversion, plantar flexion and dorsiflexion.

The multi-axis foot is slightly heavier than the single-axis foot, and also requires more maintenance [35].

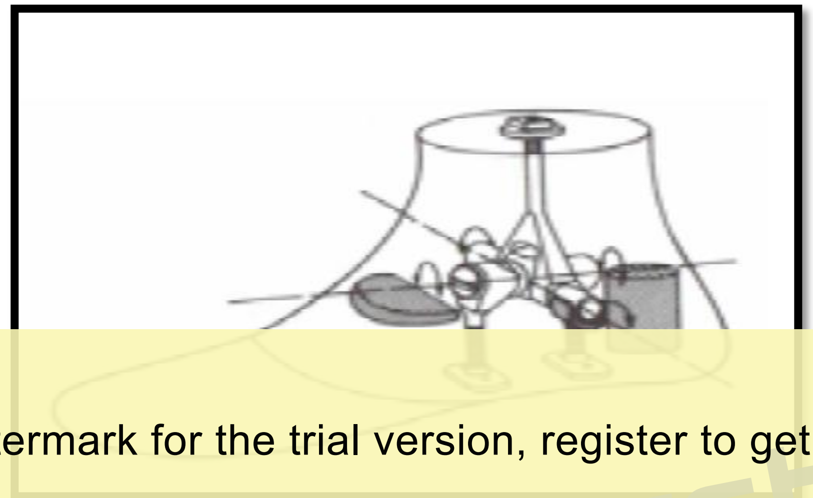


figure 3-3 Multi a

3.2 Principle of Design Prosthetic Feet

Prosthetic feet are designed to compensate for the bone musculature structure lost because of amputation and to ease the different actions that might happen during a normal gait [**36**]. During gait, the human ankle and foot are able to adjust to uneven surfaces and to adapt the length of the lower limb while providing the stabilization of the knee and shock absorption. Perfectly, prosthetic feet should be able to provide these functions; but, this still a challenge [37].

Prosthetic feet are designed to damping the impact when the heel hitting the supporting surface during the heel strike phase to decrease the forces transferred to the amputees residual limb. The contact point of the feet with the supporting surface is backward to the ankle through the heel strike which causes planter flex on the foot. Prosthetic foot control the plantar flexion rate, which controls the time that it takes to reach a flat foot, a position of stability. In transtibial patents, that helps them to control the rate of tibia advances. Then the tibia next progresses from posterior of the ankle to the anterior, through the midstance, causing the foot to dorsiflex. With an unaffected limb, the speed at which this happens controlled by musculature which helping to maintain the stability. The stability provided by the keel of the prosthetic foot. The tibia holds on to advance through the heel off phase. In normal gait, the contact point of transfers to the forefoot as the foot rolls over the metatarsophalangeal joints. This joint is sometimes changed with a mechanical element. A dorsiflexion of the toes simulated by a toe break. Toe off phase begins and the weight of the body is carried to the other leg. Prosthesis should provide supporting to help in balance, which allowing for smooth transition. At the push off end, rapid knee flexion occurs which allowing begin of the swing phase [36].

3.4 Material analysis

Prosthetic feet can be hamper proper out[illegible] can be a high effective cornerstone of treatment. The [illegible] choosing [illegible] prosthetic material and [illegible] device. A thor[illegible] prosthetic will enable designer to mix them to determine the best advantage for amputees [36].

3.4.1 Polymers

Polymers can be classified as either thermoplastics or Thermosets. Thermoplastics always be soften when heated and when removing the heat, become hard again. The term implies that the material becomes “plastic” when heat is applied. While when heating the thermosets, it will not soften as in thermoplastic, but it will char and decompose. Thus thermoplastic materials can be heated and shaped to form required shapes, Thermosets cannot. Thermoplastic materials are generally flexible and relatively soft. Polyethylene is an example of a thermoplastic, being widely used as films or sheets for such as cable insulation, bags “squeeze” bottles, and wires. The term elastomer is used for polymers which because of their structure has considerable extensions that are reversible. Polymers have a low thermal conductivity and low electrical conductivity, hence their use for thermal and electrical insulation. Compared with metal, they expand more when there is a rise in a temperature and they have lower densities, have a lower stiffness and generally more corrosion resistant. **[38]**.

Polyethylene: Polyethylene is a thermoplastic. There are two types **[39]**:

1. The high density polyethylene.
2. The low density polyethylene.

The LDPE is a partially crystalline solid melted at 115˚C. LDPE has a low specific gravity, flexibility without the use of plasticizers, good resilience, high tear strength, and good moisture and chemical resistance. High density polyethylene, having higher crystallinty, is stiffer than low density polyethylene, with greater brittleness and higher strength

Polyethylene has excellent resistance to [illegible] although oxidizing acids will cause deterioration. The advantages are good impact resistance over a wide temperature range, outstanding heat resistance, excellent vibration resistance and low cost. Therefore it is used in the proposed foot design. **[10]**.

3.5 The Ground Reaction Force

Ground reaction force (GRF) develops through the gate cycle as a result of the applied force to the ground while the foot contacts it. Ground Reaction Force is opposite and equal to the force that applied by the foot on the supporting surface. Since the Ground Reaction Force is an external force acts on the human body through the locomotion, it is of great interest in gait cycle analysis.

The plot of the ground reaction force during one gait cycle is called the M curve since it likes the shape of M that as shown in Figure (3-4).The value of F_zReaches a maximum value of about 120% body weight through the phase of double stance and drops to 80% body weight through the single stance.

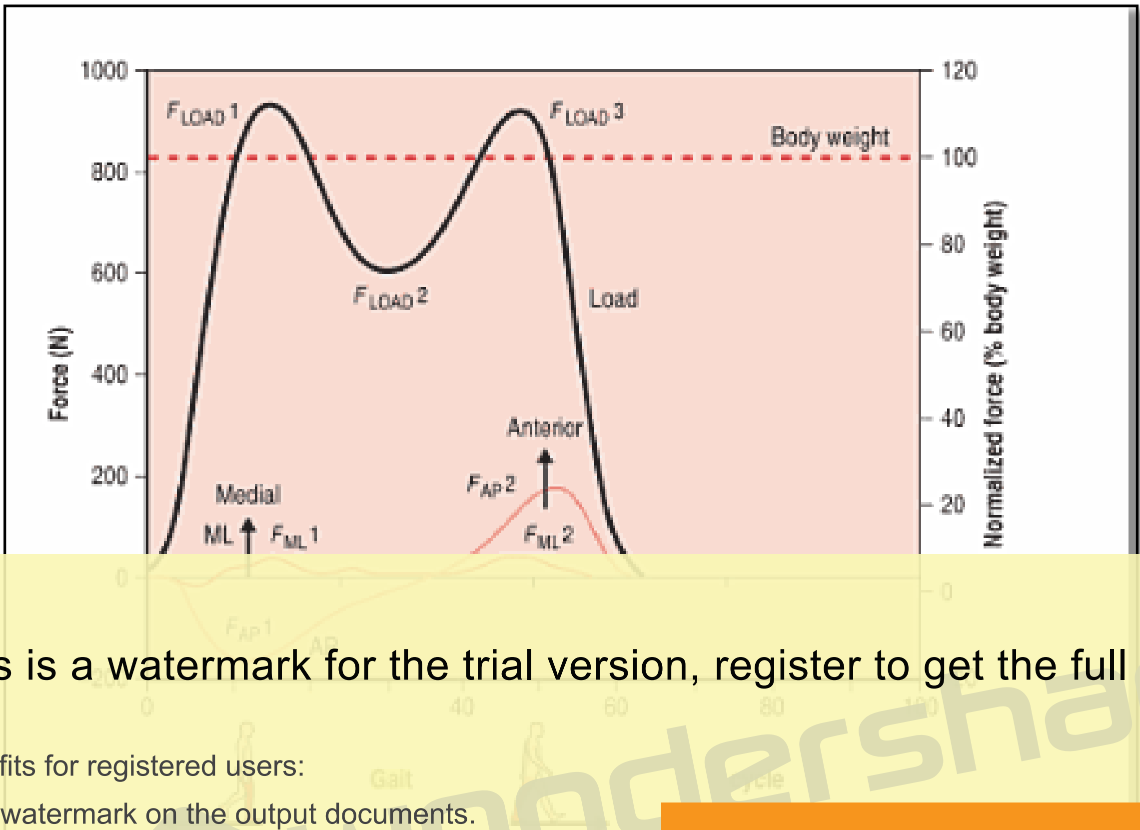


Figure 3-4 the three components of GRF during normal gait. Fz, the vertical component of GRF, is here referred to as F_{LOAD}. F_{AP} represents the anterior/posterior force component of GRF, and F_{ML} its medial **[40]**.

3.6 Niagara Foot:

The Niagara foot is an energy return low cost prosthetic foot designed to provide good performance in lower limb patients. It's developed as a part of the landmine victim's relief program of The Canadian Centre for Mine Action Technology (CCMAT) by Queen's University [41]

The Niagara Foot uses Delrin Dupont with 70 MPa yield strength and 2.6 Gpa modulus of elasticity, which is a special polyacetyl plastic that has a unique combination of toughness and elasticity. The Niagara Foot heel design, inverted from

S-shaped, which gives a spring action that provides a push to impulse the wearer forward while the foot be in contacts with the support surface.

Figure (3-5) shows the design of Niagara foot. It is designed to be used with any system of prosthetic which attaches the pylon to the foot by using a single bolt. In the upper portion, the hole is placed and provided with open base for tightening tool [42].

Niagara Foot fully tested, in international field trials, mechanically, in local and surpasses ISO 10328 [43].

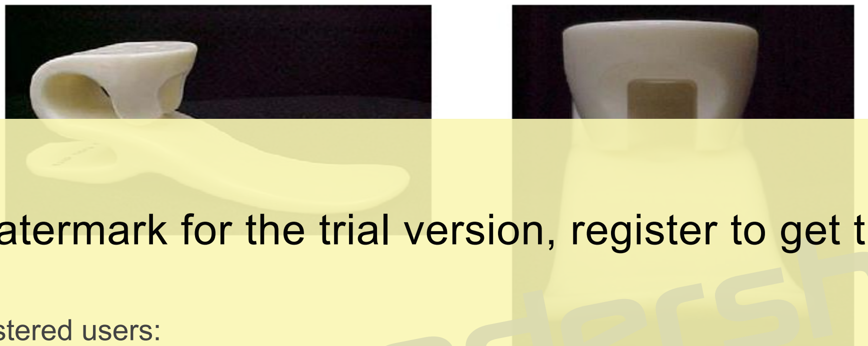


figure 3-5 Niagara foot (a) side view and (b) front view

3.7 Analysis of Niagara and a New Design Foot:

The aim of mechanical analysis is to find out equivalent stress, safety factor and dorsiflexion angle. Therefore, the mechanical properties of the materials should be known to used in a mechanical analysis.

- First,Niagara foot is modeled by using solid work 2015 software as a preperation for finint element analyses by using ANSYS Workbench 2015 as shown in figures (3-6) to (3-8)when the mechanical properties of the material that used in Niagara foot manufacturing re known,the model is analysed in to charecterize the ability of this type of foot to withstand amputees weight and the allowable dorsiflexion and factor of safety as well.

- Second, After testing many blends of materials and knowing the mechanical properties of these new materials, it was decided that , the new materials are not suitable to be used with Niagara foot design. The optimum result of mechanical properties got from the blend of (50%PP575+50%PP513), and based on that ,the new design foot was modeled by using solid work 2015 software as shown in figure (3-9), (3-10) and (3-11) and analysed by using ANSYS Workbench 2015.

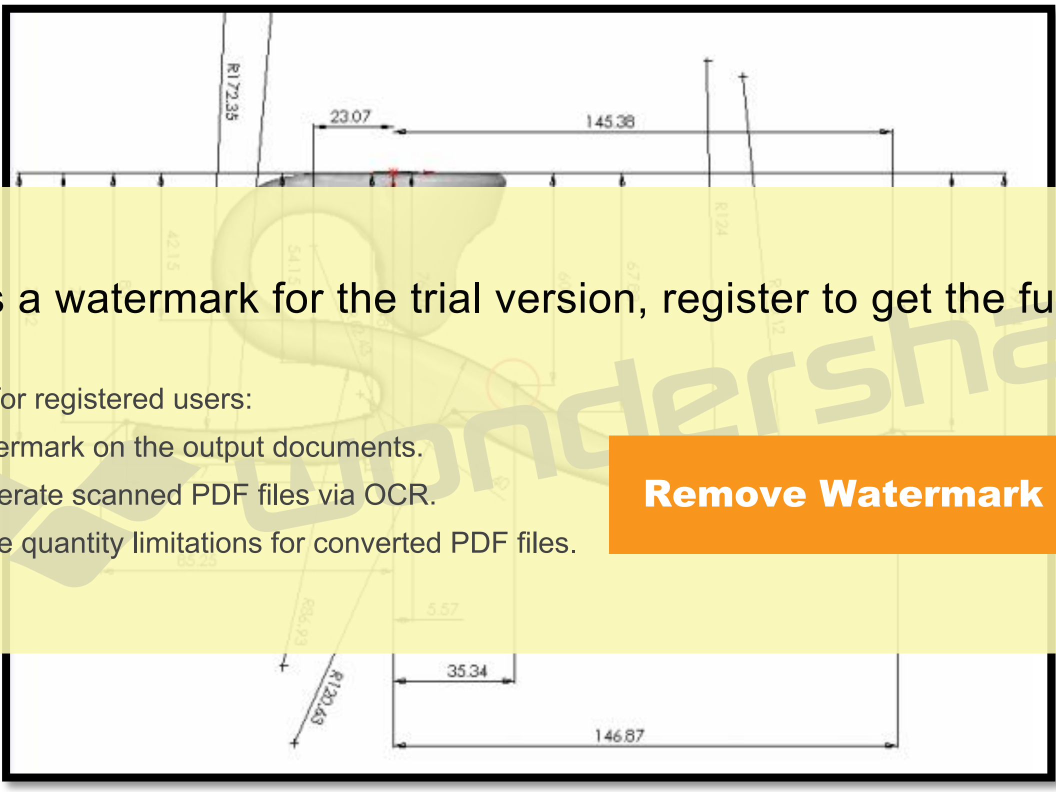
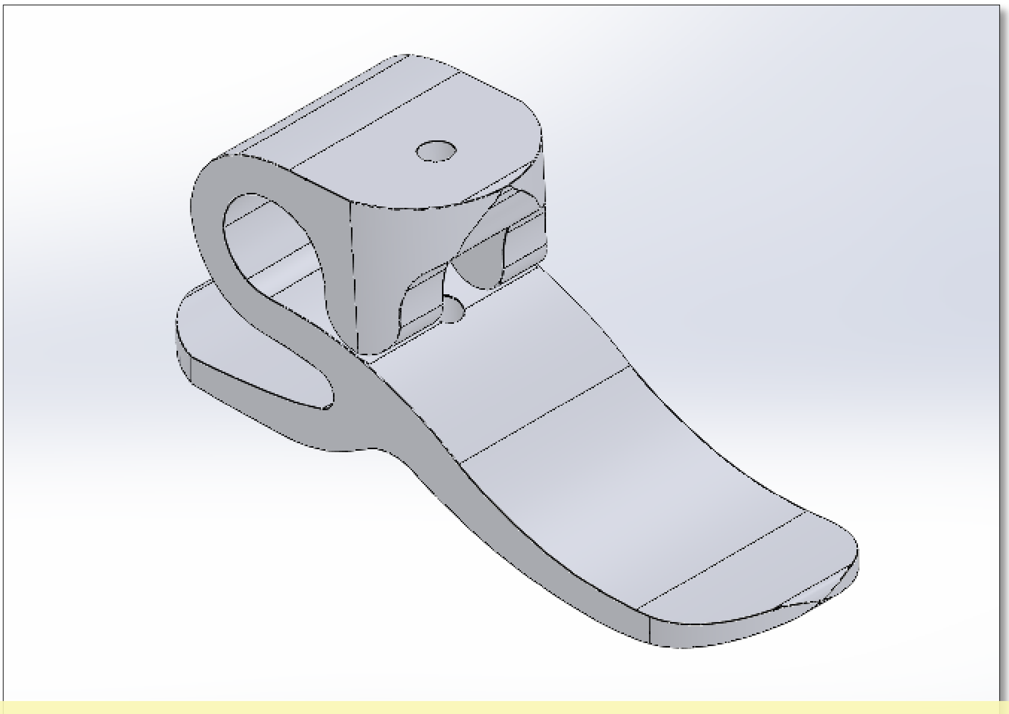


figure 3-6 The dimensions of Niagara foot



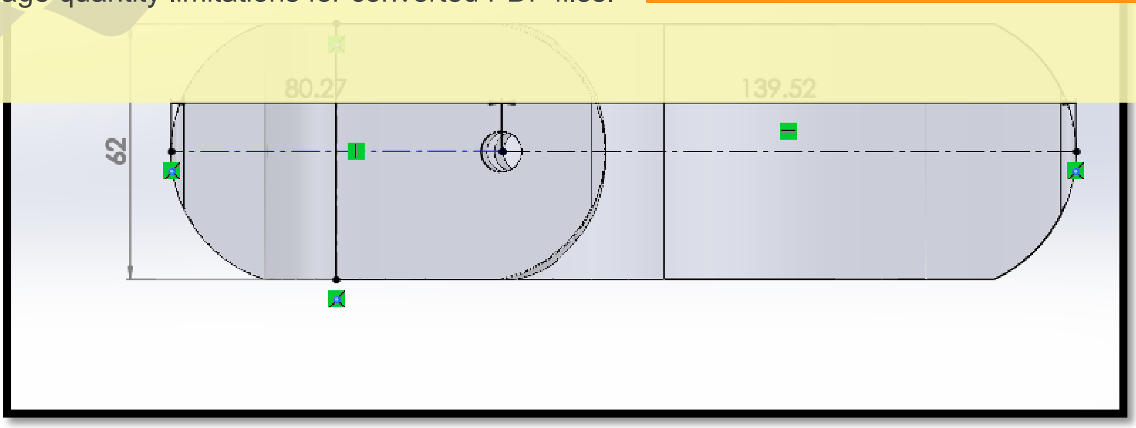


figure 3-8 Top view with dimensions of Niagara foot

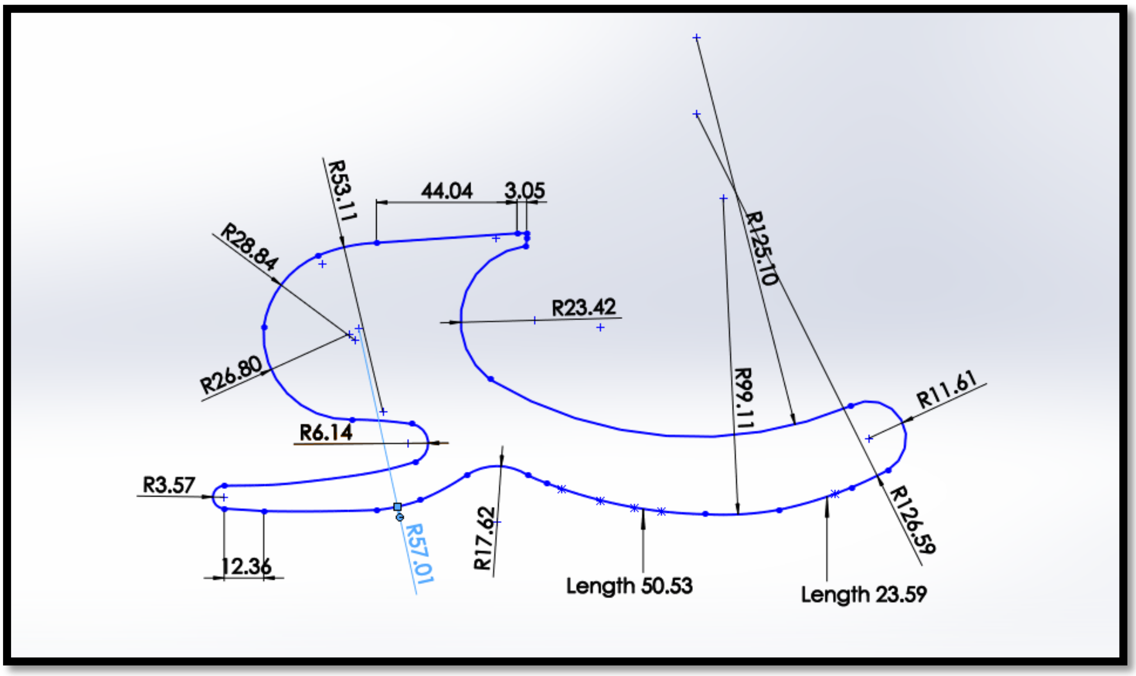


figure 3-9 The dimensions of the suggested new design foot

figure 3-10 Iso. Drawing of the suggested new design foot

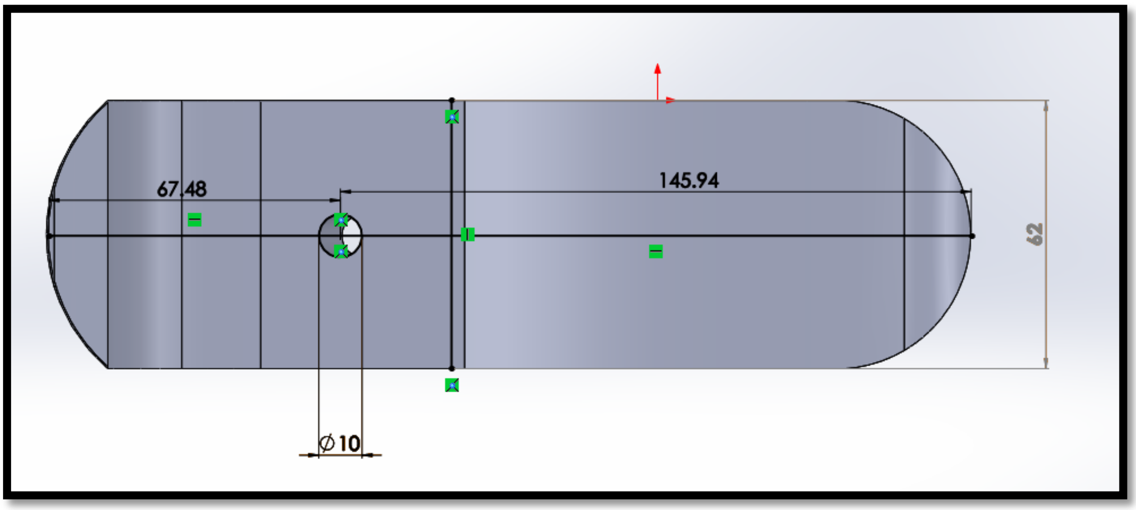


figure 3-11Top view with dimensions of the suggested new design foot

3.8: Finite Element Analysis

3.8.1 Meshing of Niagara and New Prosthetic Foot

[illegible] modeling of non-articulate foot .The type of element was chosen according to the shape of the model, the standard tetrahedral elements (SOLID-185) were used as shown in Fig.[illegible] and Fig.[illegible]. The number of tetrahedral elements was (1696) and (1907) elements with total number of nodes of (3481) and (3681) nodes for Niagara foot and new design foot respectively.

figure 3-13 Meshing of the new suggested design foot

3.8.2 Loading and boundary conditions:

The maximum applied force by a person walking on the ground is about 1.2 times body weight [9]. So, to know the range of amputee's weight that would be suitable with Niagara and new design foot, the feet are loaded with different loads which equal to amputee's weight ranging from 40kg to 75kg, for a 65 Kg amputee, the resulting applied force on the foot is about 765N which was maximum force applied on Niagara foot. For a 75Kg amputee, the resulting applied force on the foot is about 883N which was maximum force applied on new design foot, the loading and supporting cases are shown below.

1-Heel contact : During heel strike, the forces of 765 and 824 N were applied on the back of the heel for both Niagara and new design foot respectively , 15° anterior to the tibial axis according to loading conditions in the standards in ISO 10328 as shown in Fig (3-14) and Fig(3-15).

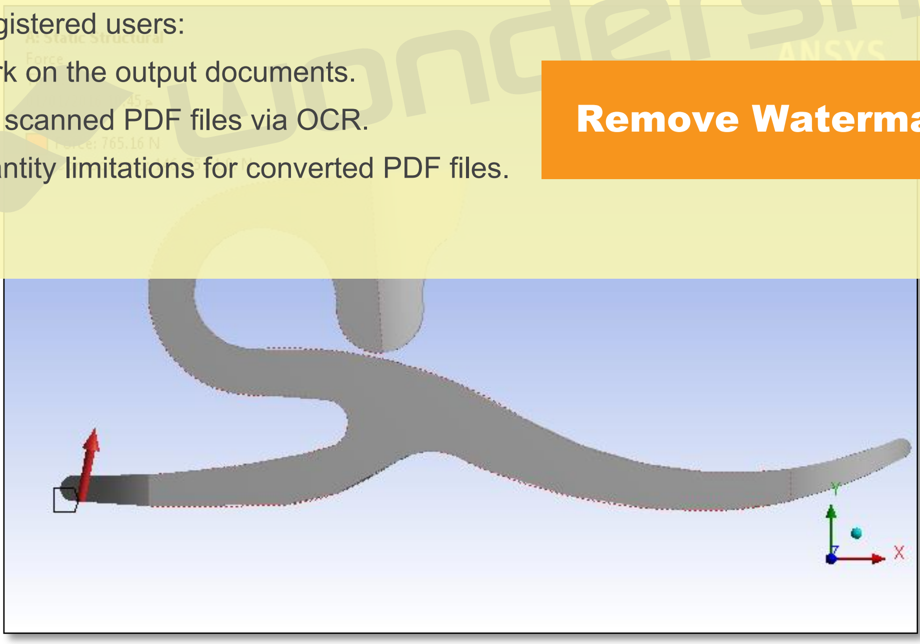


figure 3-14 Heel contact loading of Niagara foot

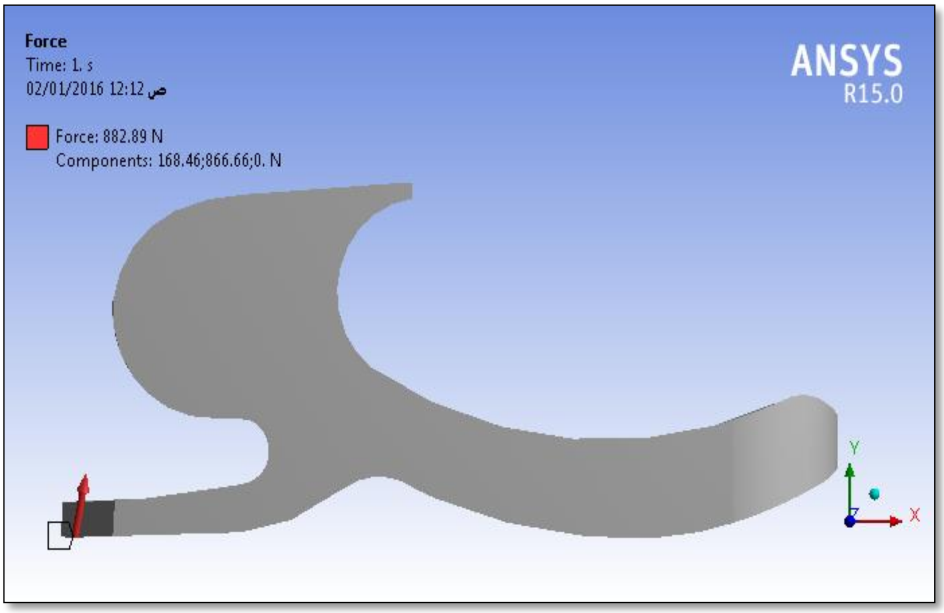


figure 3-15 Heel contact loading of the new design foot

2-Toe off phase: During toe off, the forces of 765 and 824 N were applied on the toe for both Niagara and new design foot respectively at 20° angle posterior to the tibia axis according to loading conditions in the ... figure 3-16 and figure 3-...

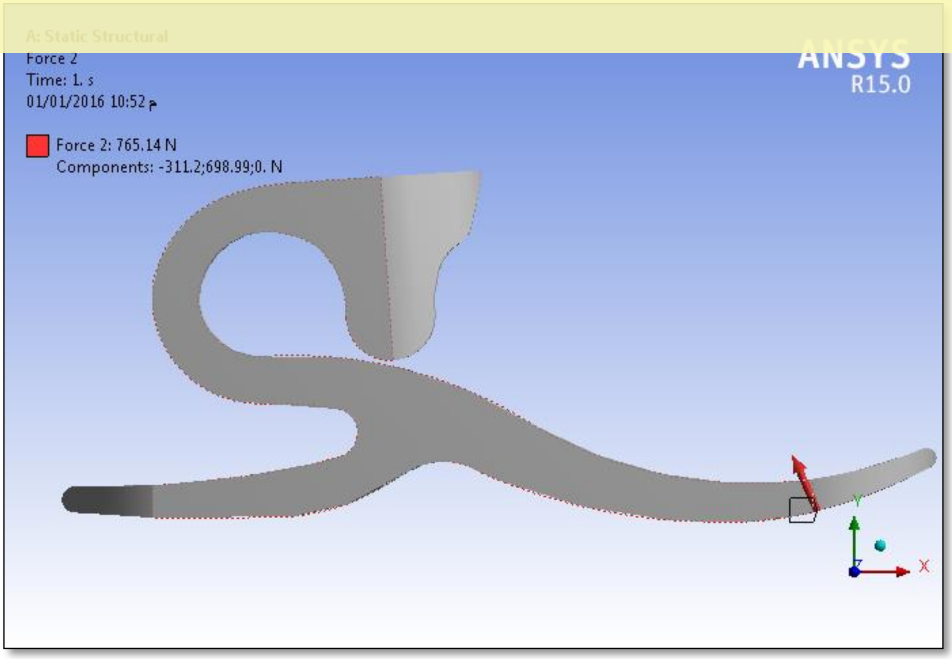


figure 3-16 Toe off load of Niagara foot

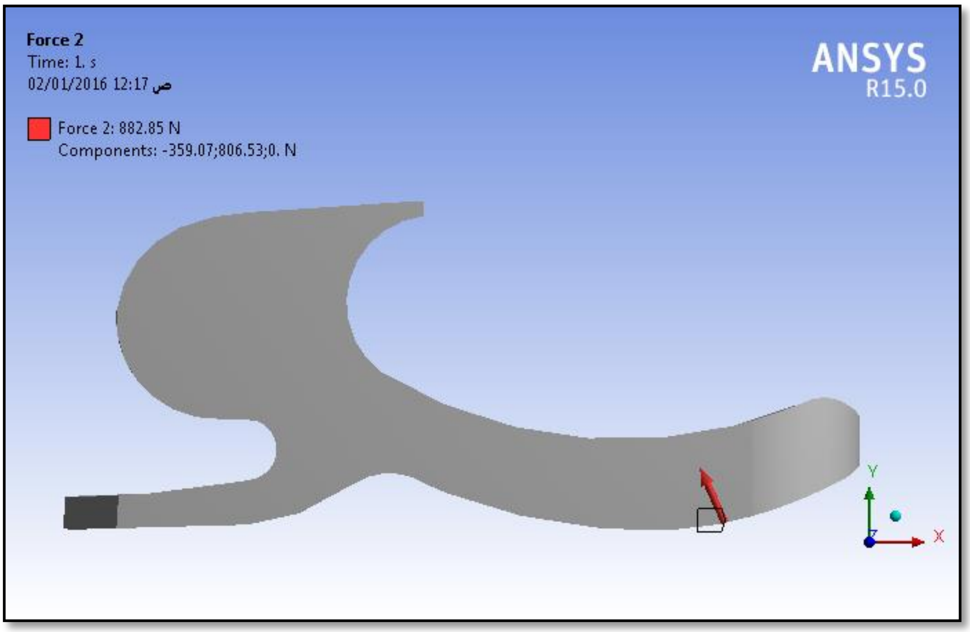


figure 3-17 Toe off load of the new design foot

3-suppotring: The upper portion of the feet supported by a fixed support as shown in figure 3-18 and figure 3-19.

figure 3-18 Fixed support of Niagara foot

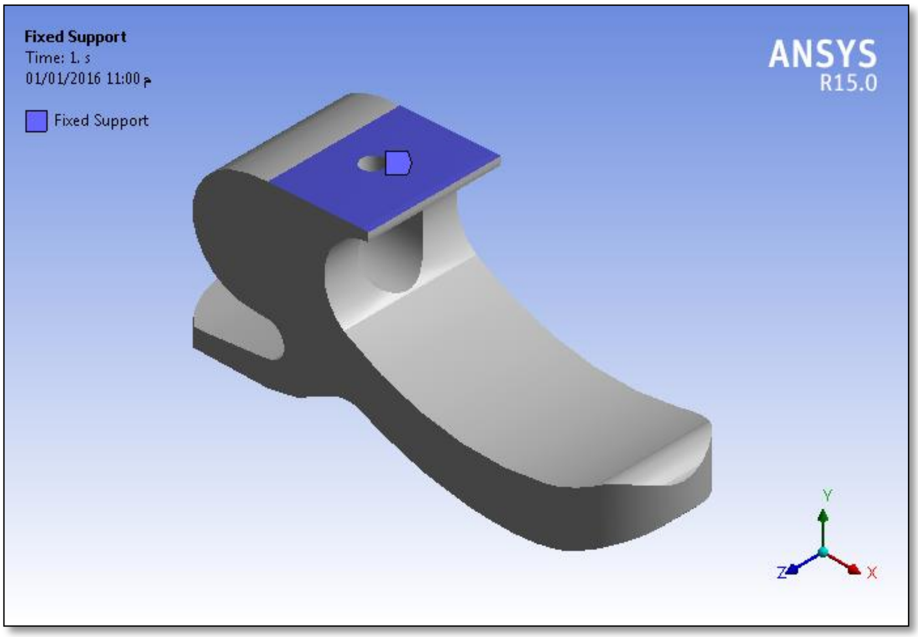


figure 3-19 Fixed support of the new design foot

3.8.3 The Solution

After input the properties of materials and boundary conditions, Ansys would be able to analyze the model and computing the equivalent stress, safety factor and total deformation.

1-the tool of maximum equivalent stress tool based on the theory of maximum equivalent stress failure for ductile material, which is called the Von Mises theory.

2-factor of safety is calculated by using equation 3-1

$$S.F = \frac{\text{Yeild Strength}}{\text{Mmaximum Equivalent Stress}} \quad \ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots 3\text{-}1$$

3-Dorsiflexion angle is calc’2ulated form the total deformation in toe off phase loading

$$\theta = \tan^{-1}\frac{Y}{X} \quad \ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots 3\text{-}2$$

Y: is a total deformation.

X: is distance from the tibial axis to the fore foot.

CHAPTER FOUR

Experimental Work

4.1 The Materials Used in the Research

1-High Density Polyethylene (HDPE).

2-Linear Low Density Polyethylene (LLDPE).

3-Carbon Fiber (CF).

4-Polypropylene grade P513MN and PP575P.

5- Calcium Carbonate ($CaCo_3$).

This chapter includes the experimental work which includes the following research points:

1-Preparation composite material compound of HDPE and LLDPE with carbon fiber.

2-Preparation of blends of polymers.

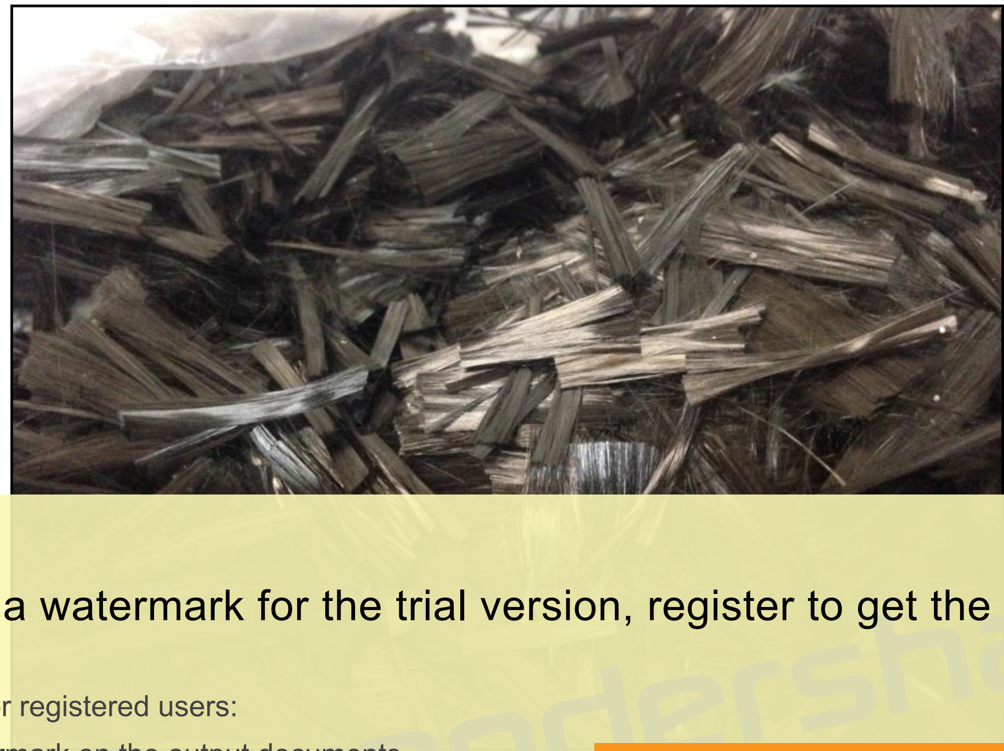
3-Preparation composite material compounds of [illegible] carbonate $CaCo_3$.

4-Testing the Mechanical properties of the composite materials.

4.2 Preparation of Composite Material Compound of HDPE and LLDPE with Carbon Fiber.

Carbon fiber is most notably used to reinforce composite materials, particularly the class of materials known as carbon fiber or graphite reinforced polymers.

Firstly, suitable grade of high density polyethylene and linear low density polyethylene was selected as a composite matrix has been supplied by(SABIC company) with melt flow index (MFI) of 0.45g and 1g /10min at 5kg/190°C for high density polyethylene and linear low density polyethylene respectively. Carbon fibers with diameter of 5 micron and 17 mm length. Figure (4-1) shows carbon fiber that was used in this work.



[illegible] 4-1 Chopped Ca[illegible]

Fig.(4-2) illustrates the internal mixer of type Brabender plasti-corder model PL2200 by which the HDPE/CF and LLDPE/CF composites were acquired. The mixer operated with temperature of 185°C and a rotor speed of 60 rpm, Table 4-1shows the six groups that were prepared. The groups have compounded of different volume fractions of carbon fiber and polyethylene as well.

Figure 4-2 Internal mixer

Table 4-1 Group of composite material

GroupName	Composition(volum fraction)
Group A	90%HDPE+10%CF
Group B	85%HDPE+15%CF
Group C	80%HDPE+20%CF
Group D	90%LLDPE+10%CF
Group E	85%LLDPE+15%CF
Group F	80%LLDPE+20%CF

4.3 specimen preparation:

The specimens for tensile, impact and fatigue test were prepared by compression molded by hydraulic press (Amir Kabir university/Tehran-Iran) as shown in Figure **(4-3)**.

The molding procedure involved heating the composite material at 185C for five minutes without applying pressure. After that for another five minutes under pressure of 50 bar, the mold was cooled by water circulation under pressure to 40 C^o .

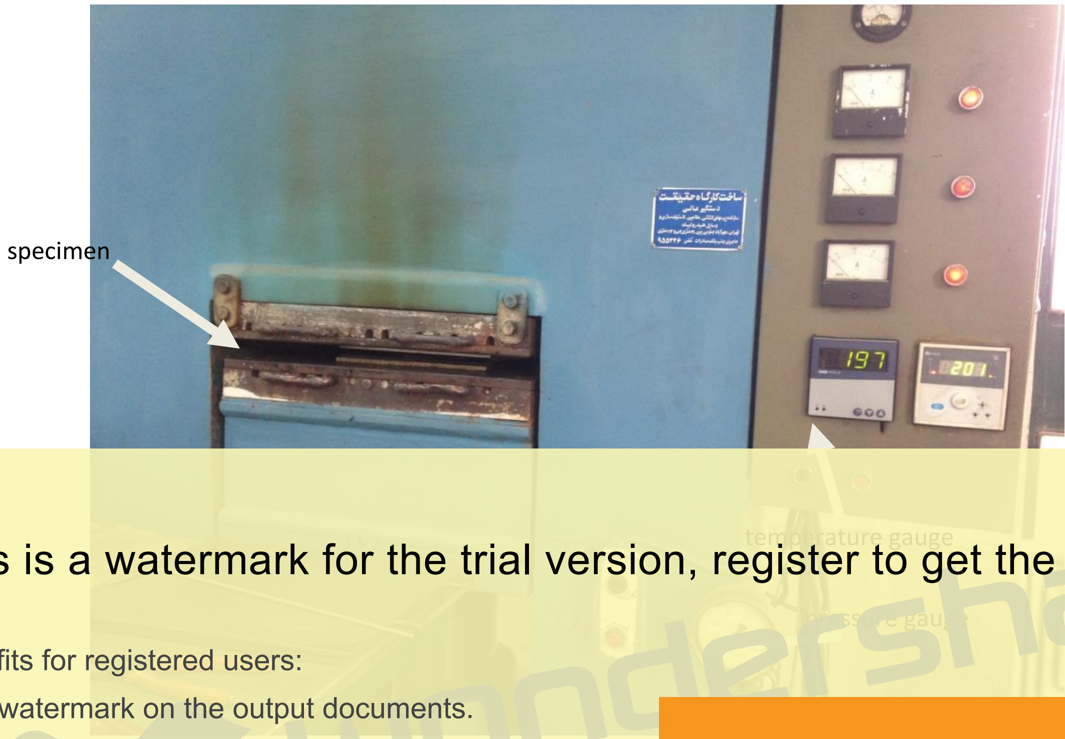


figure 4-3 Hot p

4.4 Preparation Polypropylene Groups:

Polypropylene (PP) is considered a thermoplastic material, produced by polymerizing the molecules of propylene, into long polymer chains or molecules.

Polypropylene has desirable and excellent mechanical, physical and thermal properties when it is used in applications at room-temperature. It is relatively has a low density, and good impact resistance [1].

Copolymer

Is a type of polymer when there are two types of joining monomers to produce a polymer, this is also known as heteropolymer. Two monomers can join in any fashion to make a polymer which called alternating or random polymers depending on the arrangement of monomers [2].

Polypropylene grade (PP575p) and polypropylene grade (PP513MN40) were used. The specific gravity of the PP575P and PP513 is 0.905 g/cm3, with melt flow index of 11.0 and 70 g/10 min respectively. Pure PP575P, pure PP513 and their blends were processed in a machine of injection-molding with different PP575P/PP513 contents of weight as shown in Table (4-2),

Other composite material was prepared consist of calcium carbonate and polypropylene with two different $CaCo_3$/PP weight ratio, the Groups consist of 5/1000 and 25/1000 weight ratio Named Group 6 and Group 7 respectively.

All of the above blends were processed by injection molding technique. The materials mixed and heated by using injection machine which shown in Figure (4-4). The machine contains four heat zones, first heater is (80˚C) and fourth heater is (200˚C), this temperature increases gradually, then the molded material injected with injection temperature 200˚ into the mold with pressure of 5 bar.

Table 4-2 Polypropylene Groups Weight Ratio (w/w)

Group	[illegible]	2	[illegible]	[illegible]	[illegible]
PP575P %	100	75	50	25	0
PP513 %	0	25	50	75	100

.

figure 4-4 Injection Machine

4.5 Metal mold:

Metal mold contain from two parts figure (4-5) one of them include the shape that inj[illegible] another rep[illegible] dimensions of the product sheets was 200mm*175mm *5mm, as shows in figure (4-6), then these sheets cut into specimens for mechanical tests by using cutting machine .

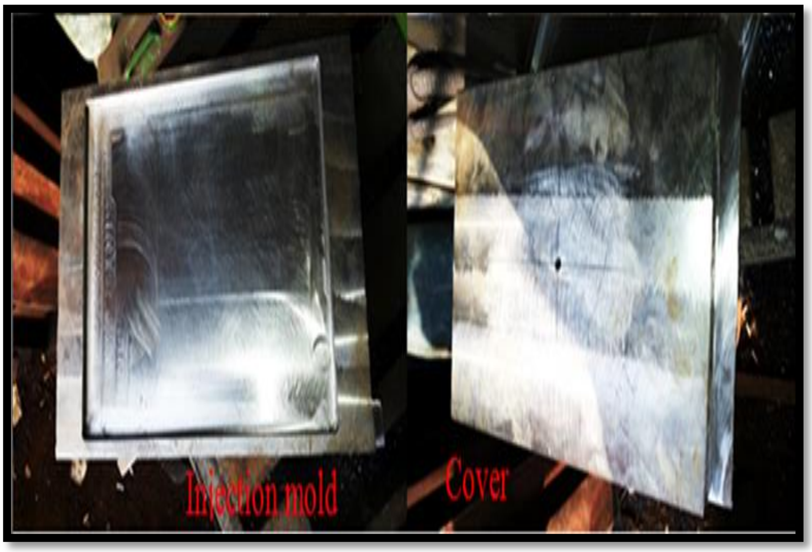


figure 4-5 Metal Mold

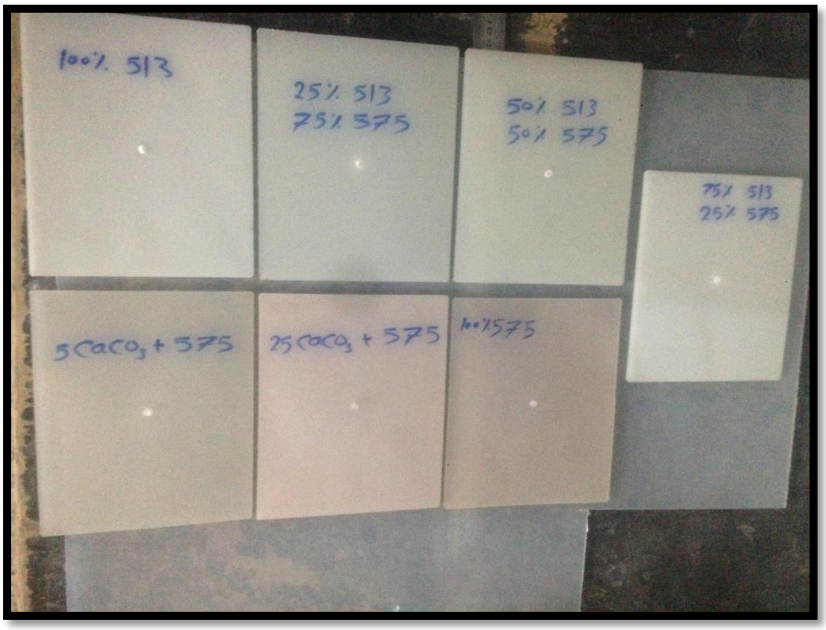
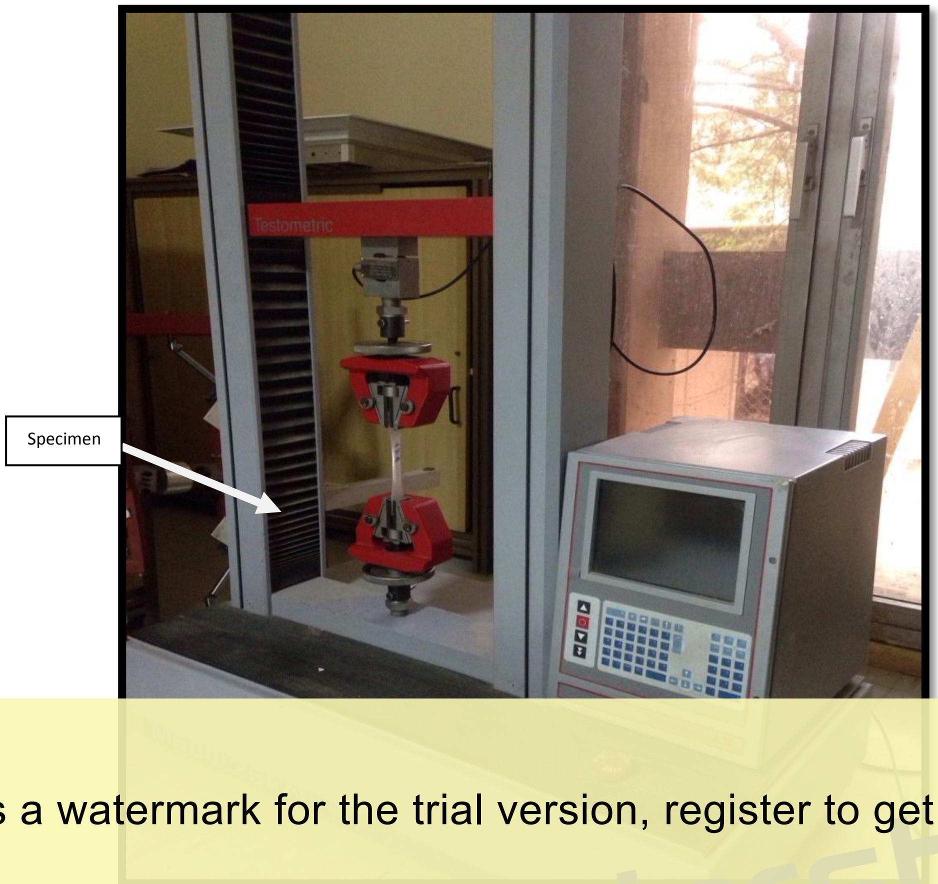


figure 4-6 Product Sheets

4.6 Mechanical Tests:

4.6.1 Tensile Test:

The mechanical properties for the materials mechanical tests such as tensile test for standard specimens. Tensile testing of the specimens performed according to ASTM D638 by using the Testometric machine (Al-Nahrain University / mechanical Engineering). Specimen dimensions were as shown in figure (4-7).The samples tested at about 25°C. At least, three specimens of every composition were tested. The cross head speed is 5mm/min , figure (4-8) shows the tensile specimen dimension.



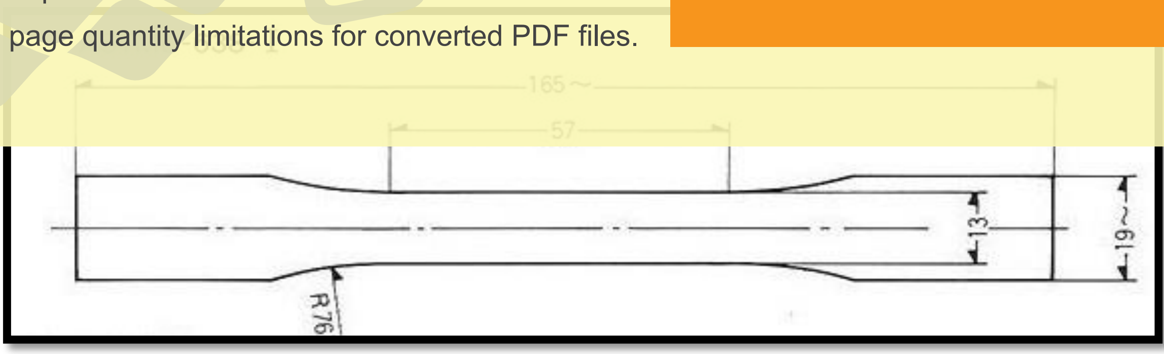


figure 4-8 Tensile specimen's geometry and dimensions

a

b

figure 4-9 Standard Specimens of Composite Material for Tensile Test

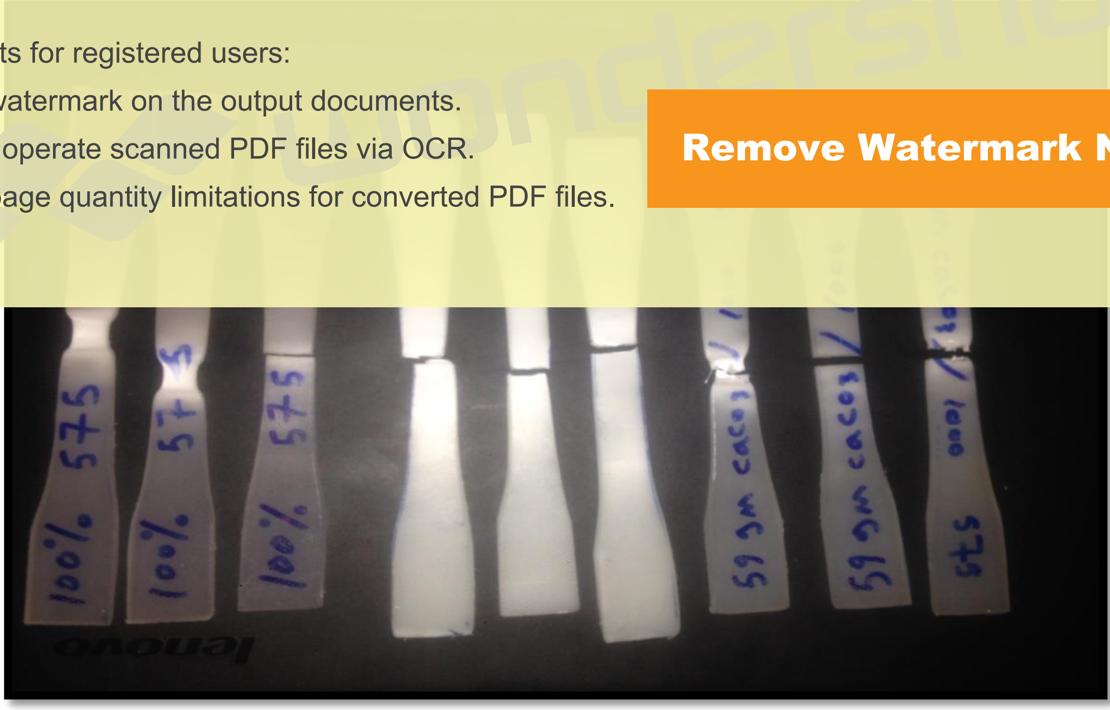


figure 4-10 Specimens of Composite Material after Tensile Test

4.6.2 Impact Test:

For impact testing, IZOD impact test was used The test specimens was carried out on an Impact Tester using instrumented impact pendulum with an overall impact energy of about 5 J striker in accordance with ASTM D256,as shown in figures (4-11),(4-12).In addition ,the hammer was selected according to type and shape of composite materials. Specimen dimensions were (65mm * 13mm * 5mm), where test worked on testing impact machine as shown in figure (4-13). A sharp blow on the specimen breaks the test piece and the impact was recorded from the read out. This device is available in University of Technology/Materials Engineering Department.

figure 4-11 Impact specimens of composite material

figure 4-12 Impact specimens of polypropylene blends

figure 4-13 impact tester.

4.6.3 Fatigue Test:

The [illegible] of fatigue testing figure ([illegible]) [illegible] with constant amplitude. The specimens were subjected to deflection which is normal to the specimen’s axis one the specimens’ side and another side is fixed to develop the stresses of bending.

So, the specimens’ surfaces are under compression and tension stresses when the machine rotates. Dimensions of samples are as follows: the length of 100mm, the width of 10mm and the thickness of 4mm according to the fatigue device test. Figure (4-15) shows the shape and dimensions of fatigue samples.

A dial gauge was used to measure the deflection; their values were used to determine the maximum alternating bending stress.

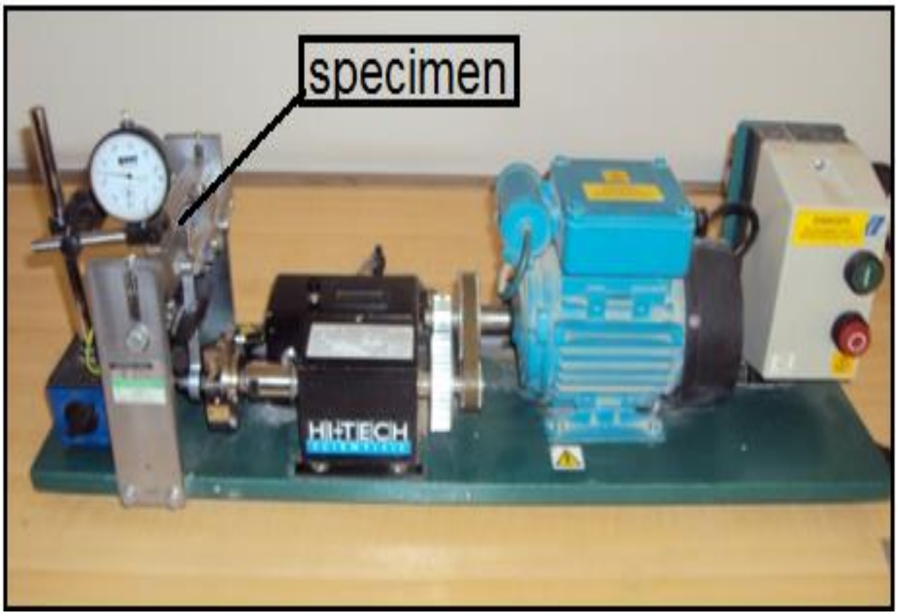


figure 4-14 Fatigue Test Device.

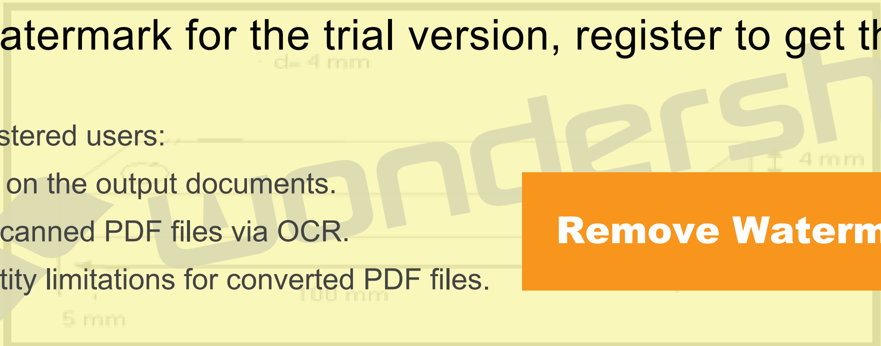


figure 4-15 geometry and dimensions of fatigue specimen

CHAPTER FIVE

Results and Discussion

5.1 Introduction

This chapter presents the results of the experimental work for the selected composite material for the tensile and impact test. Also, the result of the analysis for the prosthetic foot models will is be illustrated here to know the results of the analysis of (Von-Mises) stress, safety factor, total deformation and dorsiflexion of the feet.

5.2 Mechanical Properties Results

The results of the mechanical properties (tensile and impact) for the prosthetic foot's materials are presented separately:

5.2.1Tensile tests

Table (5-1) provides the mechanical [illegible]ertis of [illegible] composite materials blends A-F. Figures [illegible] and [illegible] show the stress-str[illegible]). Figure (5-4), (5-5) and (5-6) show stress-strain curve for blends of (CF, HDPE). From these curves the mechanical properties σ_y, σ_{ult}, and E and for each sample are determined. The results explain the effects of increasing carbon fiber volume fraction with polyethylene in mechanical properties which lead to increasing in the ultimate tensile strength.

Table 5.2 presents the results of groups 1-7. Figures (5-7) to (5-13) show the stress-strain curves for group 1-7.

Table 5-1 The results of tensile test for the groups of A-F

Group Name	Composition	σ_Y(Mpa)	σ_{ult}(Mpa)	E (GPa)
Group A	10%CF+90%LLDPE	12.169	18.6	2.299
Group B	15%CF+85%LLDPE	10.17	21.93	2.36
Group C	20%CF+80%LLDPE	10.984	32.4	2.7
Group D	10%CF+90%HDPE	11.067	33.5	2.212
Group E	15%CF+85%HDPE	11.117	35.7	4.205
Group F	20%CF+80%HDPE	9.48	39	3.8

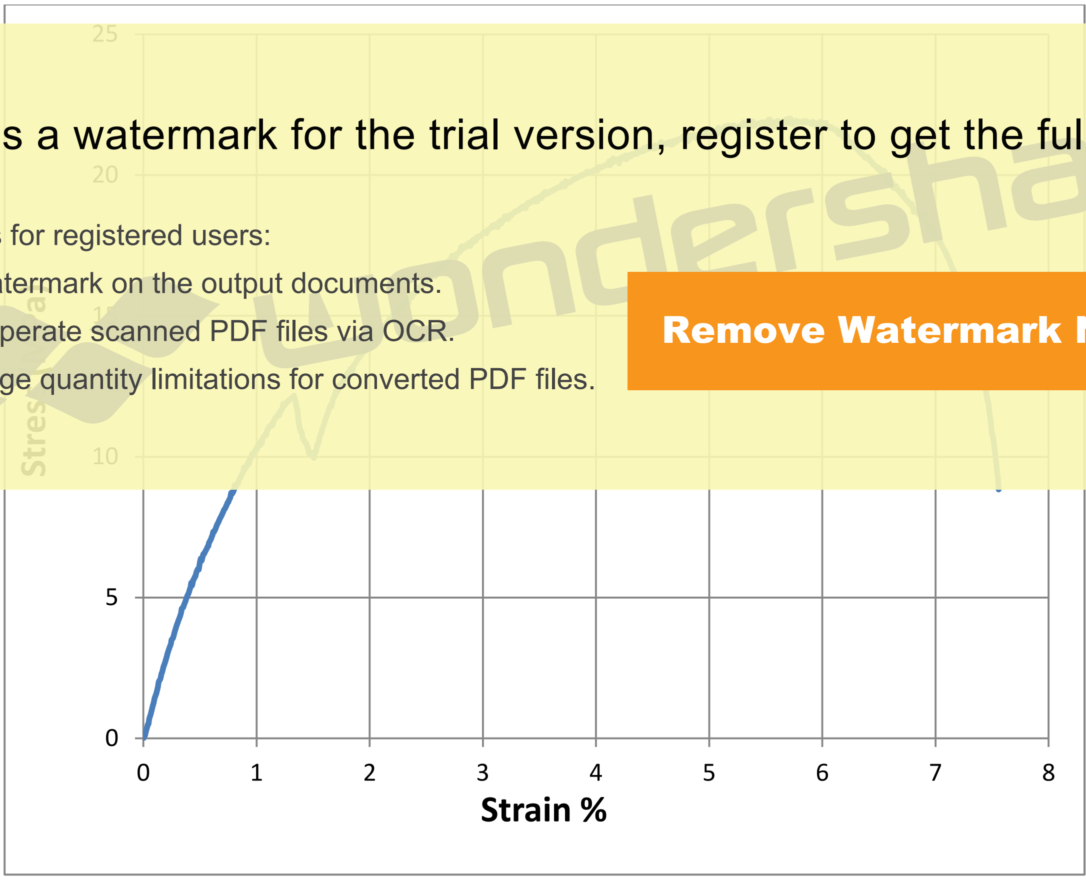


figure 5-1 Stress-strain curve for the group of (10%CF,90%LLDPE)

figure 5-2 Stress-strain curve for the group of (15%CF,85%LLDPE)

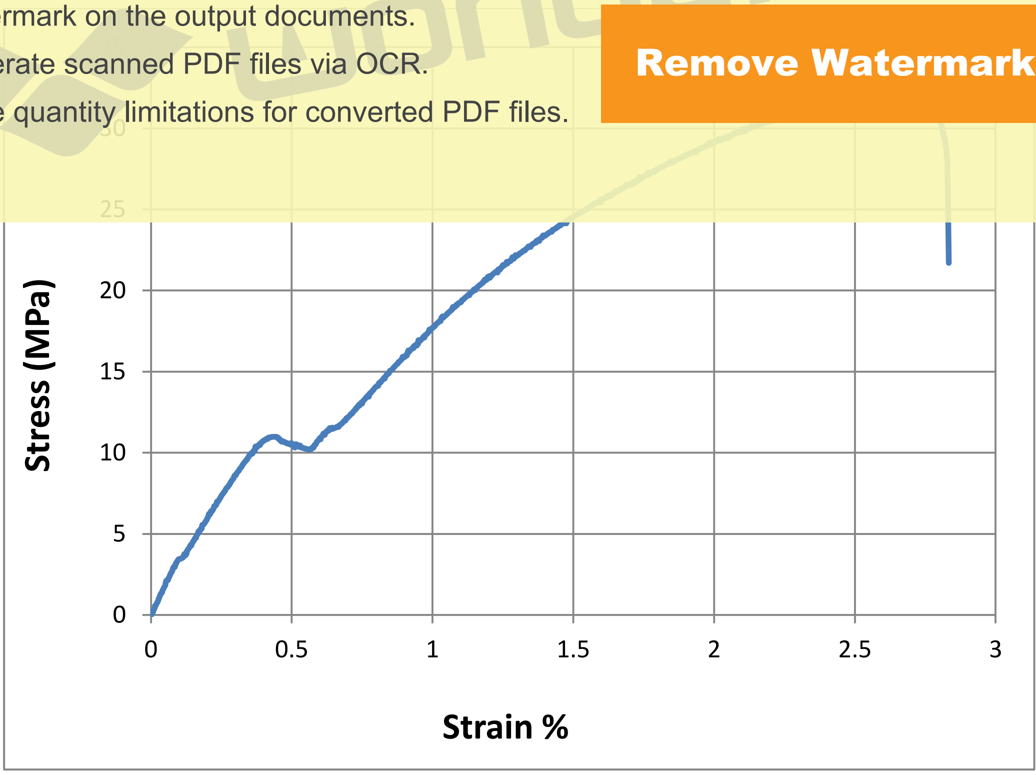


figure 5-3 Stress-strain curve for the group of (20%CF, 80%LLDPE)

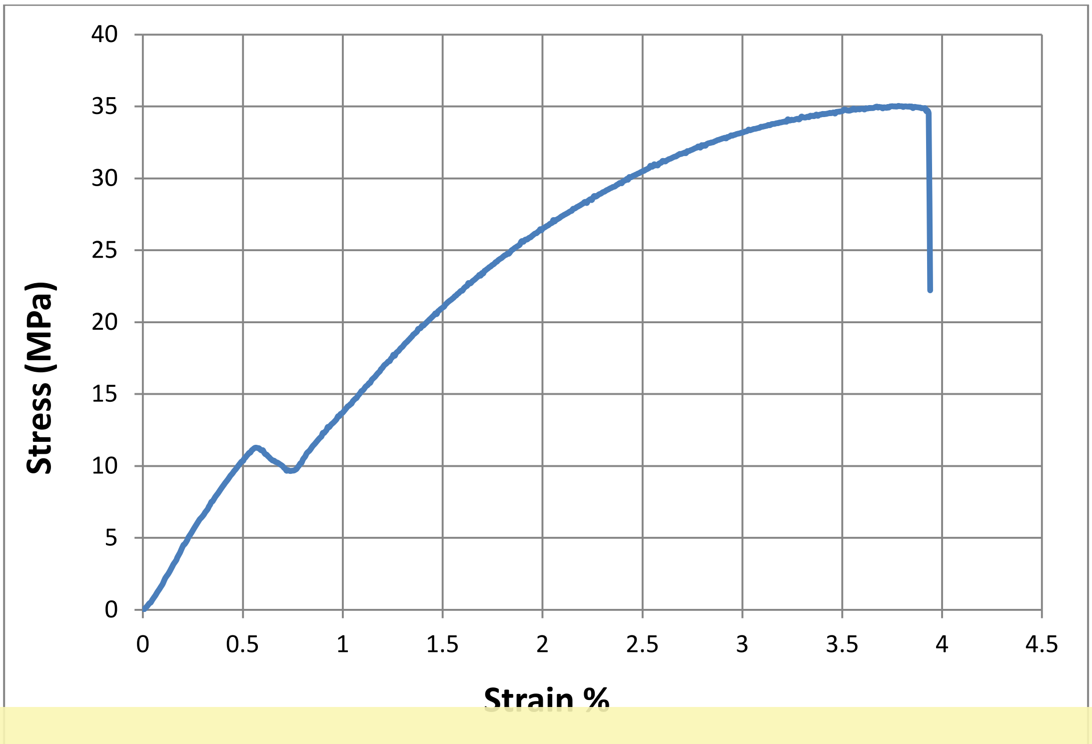


figure 5-4 Stress-strain curve for the group of (10%CF, 90%HDPE)

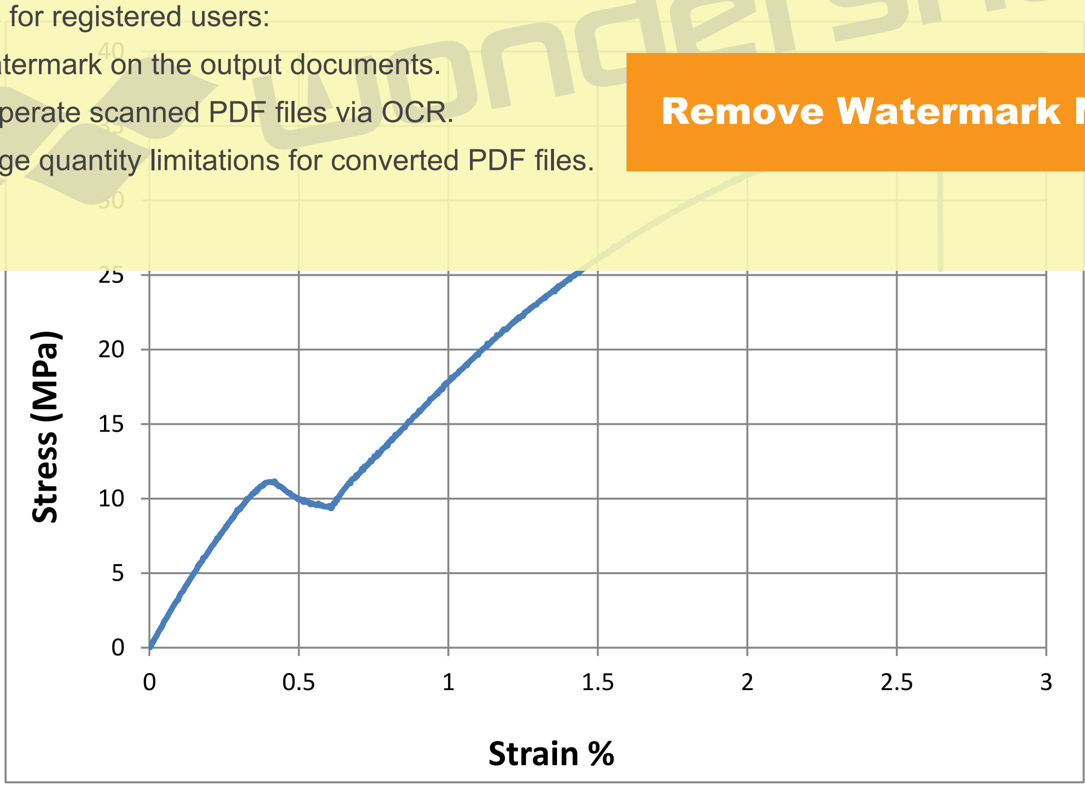


figure 5-5 Stress-strain curve for the group of (15%CF, 85%HDPE)

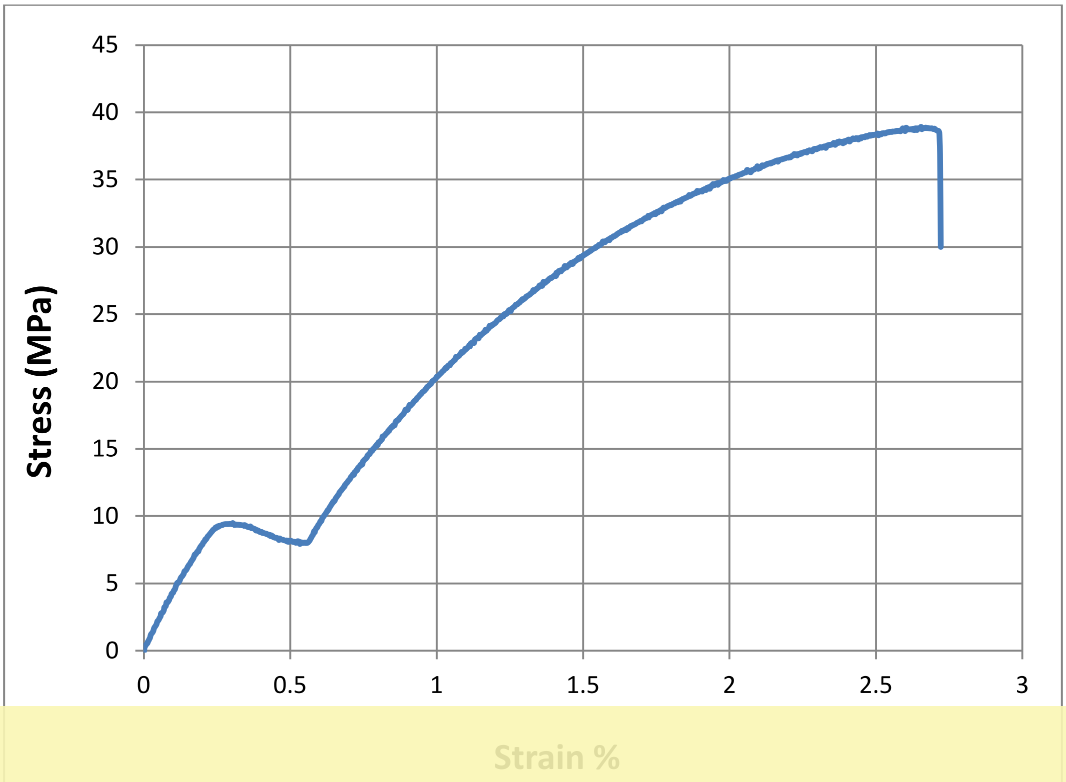


figure 5-6 Stress-strain curve for the group of (20%CF+80%HDPE)

Table 5-2 The results of tensile

	Composition			
Group 1	100%PP575	41.049	18	0.788
Group 2	75%PP575+25%PP513	38.66	23	0.841
Group 3	50%PP575+50%PP513	37.437	20	0.806
Group 4	25%PP575+75%PP513	29.525	17	0.751
Group 5	100% PP513	26.785	16	0.707
Group 6	5/1000,$CaCO_3$/PP575	40.32	19	0.879
Group 7	25/1000,PP575+$CaCO_3$	40.887	17	0.830

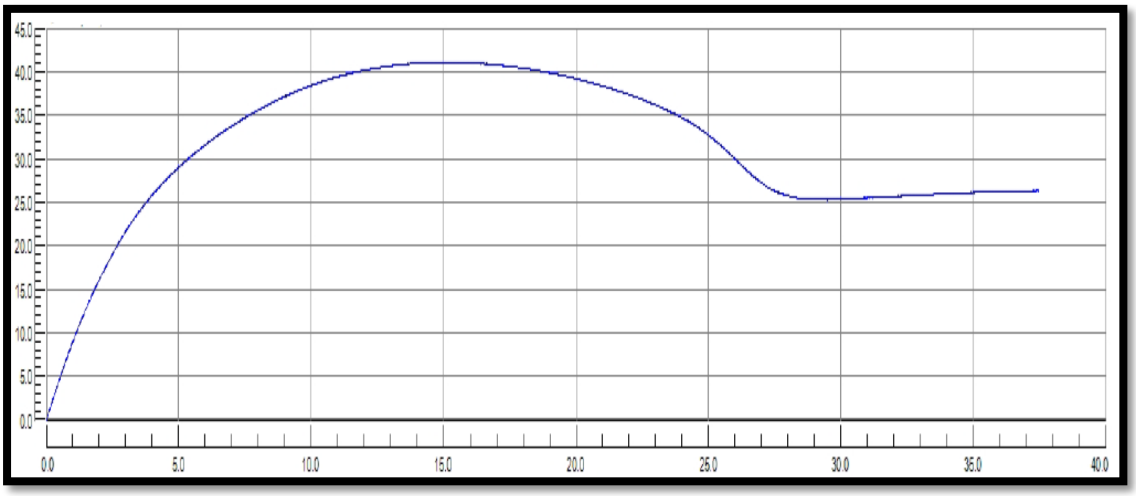


figure 5-7 Stress-strain curve for the group of 100%PP575

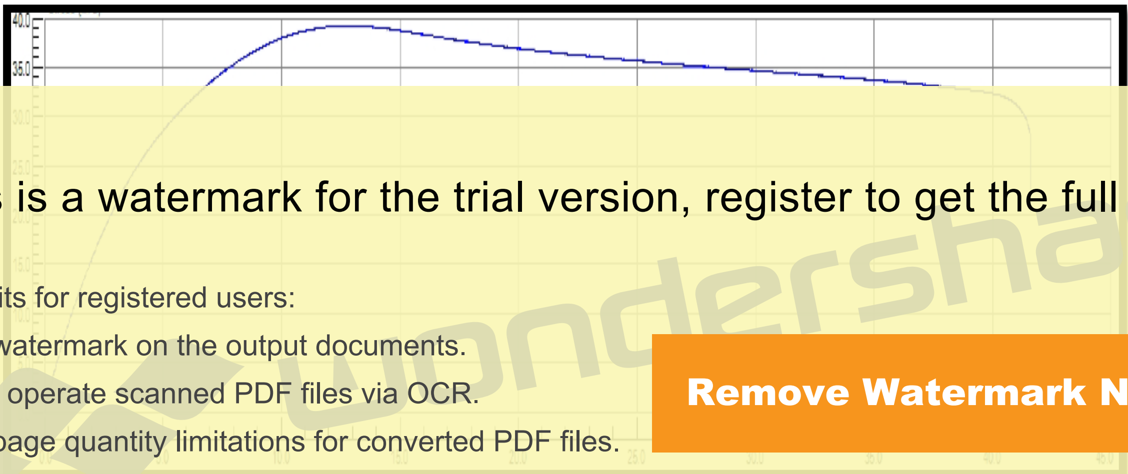


figure 5-8 Stress-strain curve for the group of (75%PP575+25%PP513)

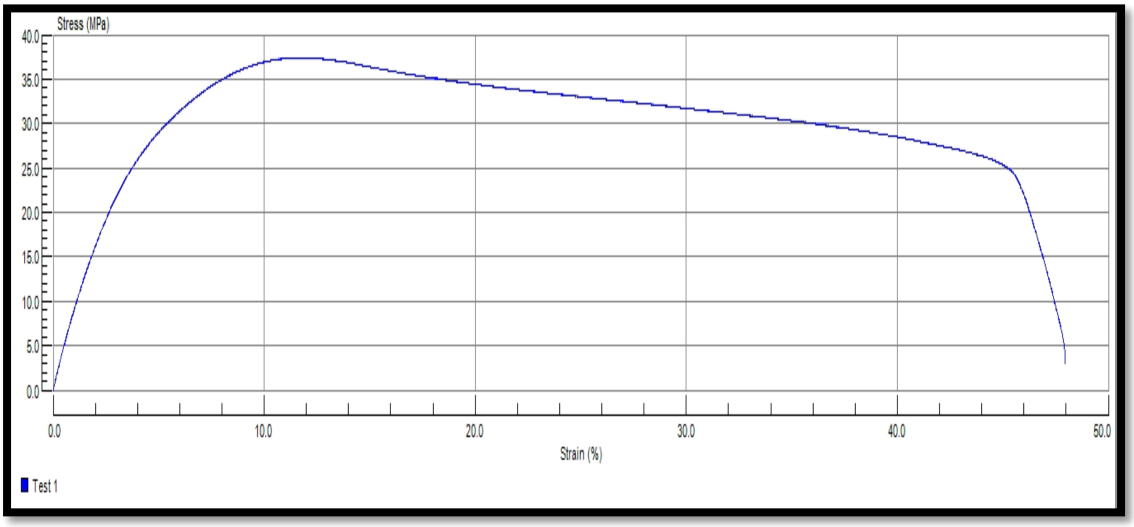


figure 5-9 Stress-strain curve for the group of (50%PP575+50%PP513)

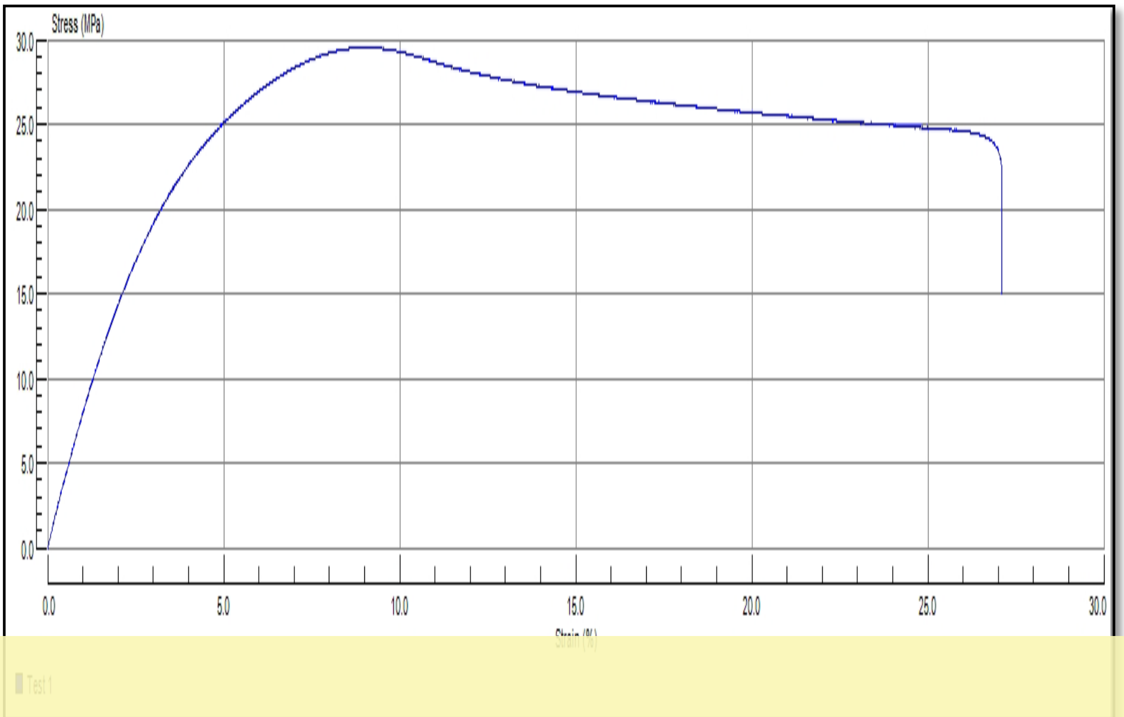


figure 5-10 Stress-strain curve for the group of (25%PP575+75%PP513)

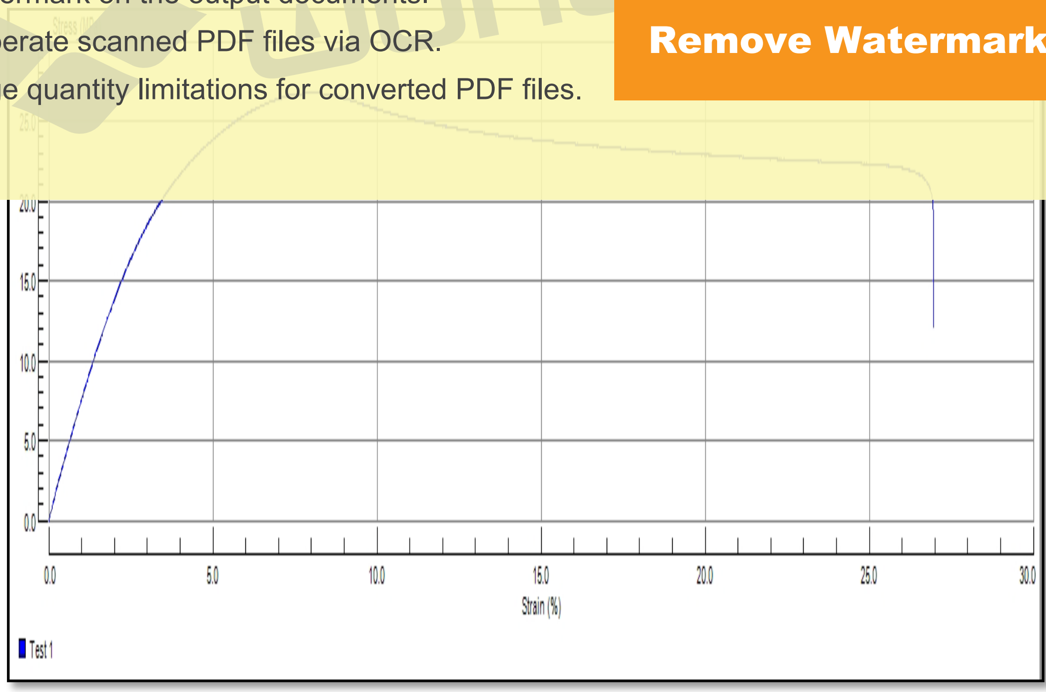


figure 5-11 Stress-strain curve for the group of 100%PP513

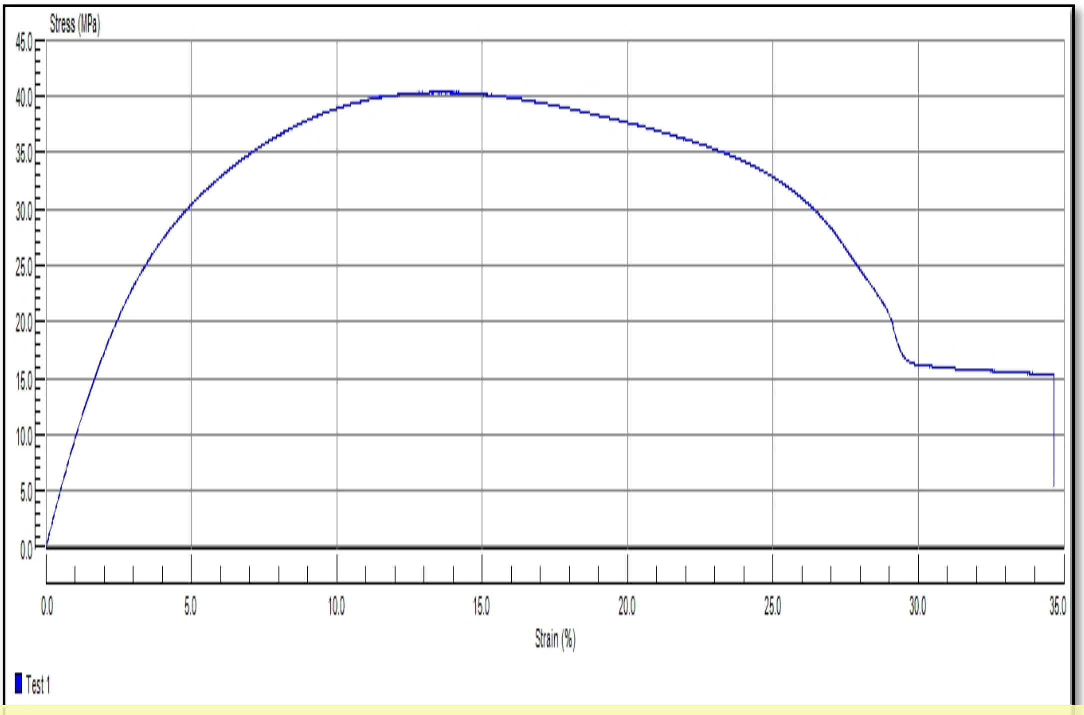


figure 5-12 Stress-strain curve for the group of (5 g $CaCo_3$, PP575)

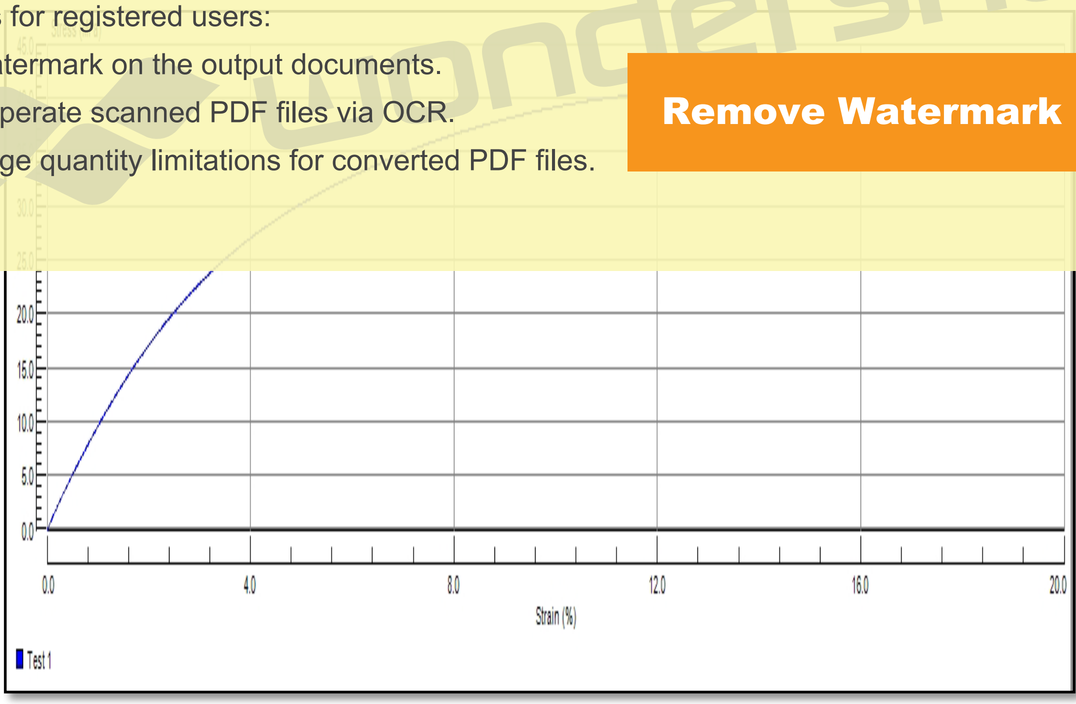


figure 5-13 Stress-strain curve for the group of (25g $CaCo_3$, PP575)

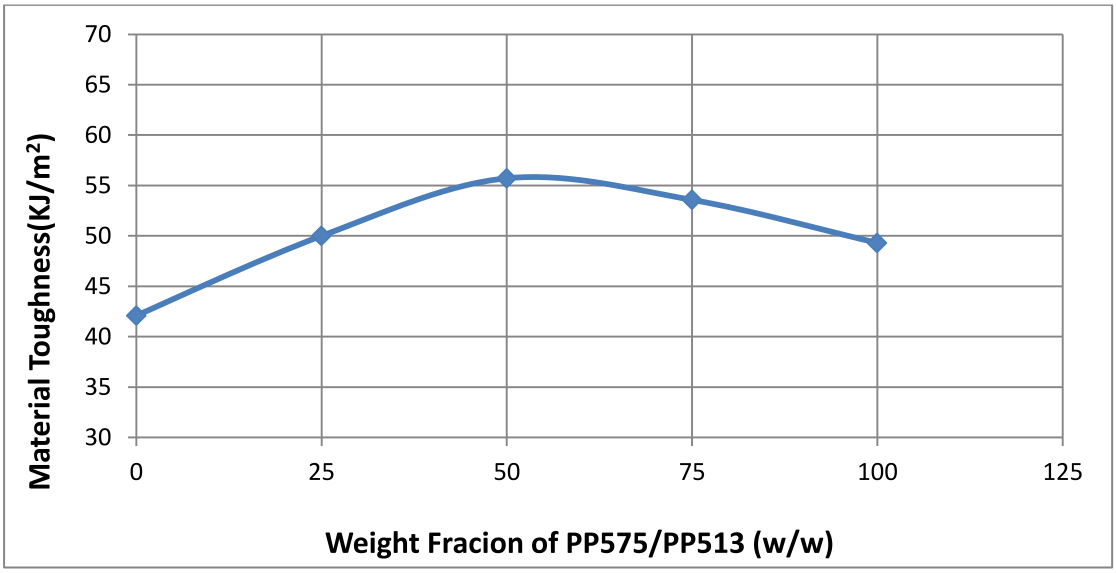


figure 5-14 effect of adding PP513 on toughness

5.2.2. Fatigue tests

After tensile tests for the blends of HDPE and LLDPE with carbon fiber, the results showed that the optimum result [illegible] (20%CF, 8[illegible] [illegible]or fatigue test. Eight samples tested with different stresse[illegible]

Table 5-3 The results of fatigue test of group F

Stress (MPa)	Number of cycles
7.3	11258
6.5	20647
5.2	43528
4.15	132476
3.6	257472
3.1	442782
2.7	573247
2.3	786324

5.1.3 Impact Test

Impact testing is widely used to characterize the material toughness of materials because it attempts to simulate the most severe load conditions to which a material can be subjected.

For the composite materials group A,B,C,D,E and F, the results of impact test are shown in Table(5-4).It can be noticed that the values of material toughness significant decrease when decrease or increase the volume fraction of carbon fiber from 15%. The maximum values of absorbed impact energy were with blends which contain 15% CF.

For the copolymer and composite materials Group 1-7, the result of impact test are shown in table (5-5).It significant clearly that, when adding of PP513 to PP575 leads to increasing in absorbed impact energy of the copolymer.

The maximum values of material toughness were with group [illegible]

Table 5-4 Impact test results of groups A-F

Blend No.	Composition	Absorbed Energy [illegible]	[illegible]
Group A	10%CF,90%HDPE	0.7	15.38
Group B	15%CF,85%HDPE	1.3	28.57
Group C	20%CF,80%HDPE	0.25	5.49
Group D	10%CF,90%LLDPE	0.9	19.78
Group E	15%CF,85%LLDPE	1.4	30.76
Group F	20%CF,80%LLDPE	0.5	10.98

Table 5-5 Impact test results of the groups 1-7

Group Name	Composition	Absorbed Energy (J)	Material Toughness(KJ/m^2)
Group 1	100%PP575	3.5	50.714
Group 2	75%PP575+25%PP513	3.7	52.857
Group 3	50%PP575+50%PP513	3.9	55.714
Group 4	25%PP575+75%PP513	3.5	50
Group 5	100% PP513	2.95	42.094
Group 6	5/1000,$CaCO_3$/PP575	2.1	30
Group 7	25/1000, $CaCO_3$/PP575	1.8	24.48

5.6 The Result of the Numerical Analysis and Discussion

The results of static analysis of Niagara and the suggested new design foot are illustrated. These results include Von Mises stresses, safety factor, total deformation and dorsiflexion for both feet.

5.6.1 Von Mises Stresses.

Calculating Von Mises stress is very important for the design and the analysis of the foot because it is considered an indicator of failure. It is an index that combines the three principal stresses of the material. Although the material may not indicate failure based on the principal stresses alone, the von Mises stress can show how these principal stresses combined potentially cause a failure in the material Von Mises stresses were also extracted from the FEA models, and the resulting contours can be found in Figures (5-16), (5-17) for Niagara foot with load of 65Kg as a maximum amputees weight can be used. Figures (5-18), (5-19) show the results contour for suggested new design foot with load of 75 Kg as a maximum amputees weight can be used for both phases of heel contact and toe off phase respectively. The results of Von Mises stresses for heel contact are illustrated in figure (5-20) and table (5-6). The results for toe off phase are illustrated in figure (5-21) and table (5-7) for Niagara and the new design foot respectively. The results show that, the maximum Von Mises stress in heel contact region were 29.344MPa and13.513MPa, while in toe off phase were 26.085MPa and 17.23MPa for Niagara and the suggested new design foot respectively.

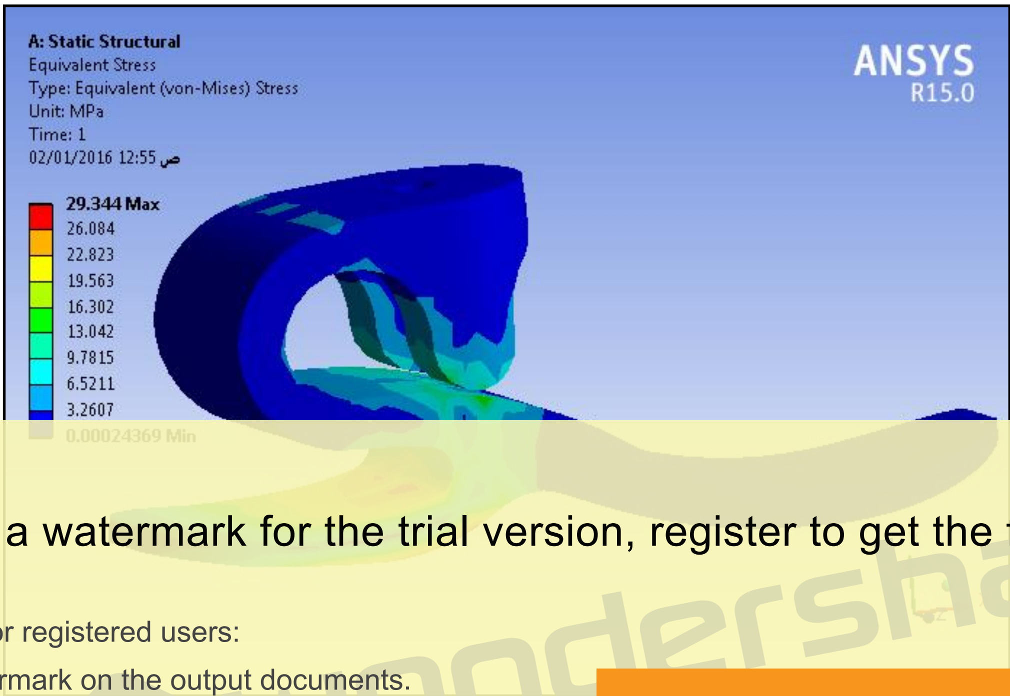


figure 5-16 Equivalent Stress of Niagar

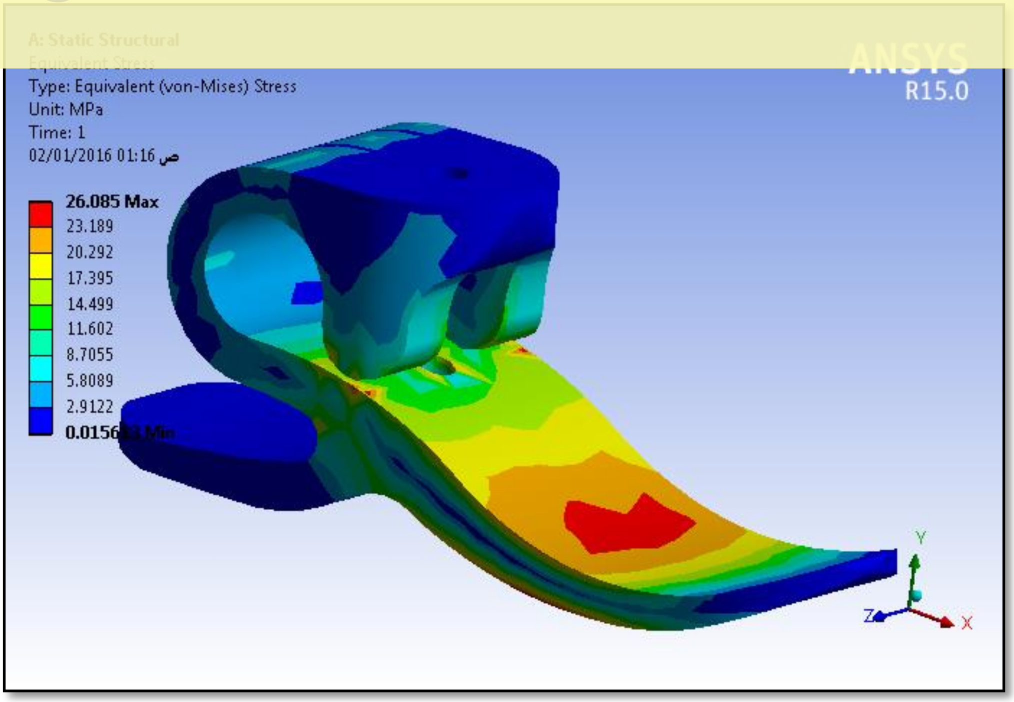


figure 5-17 Equivalent Stress of Niagara Foot (Toe-Off phase)

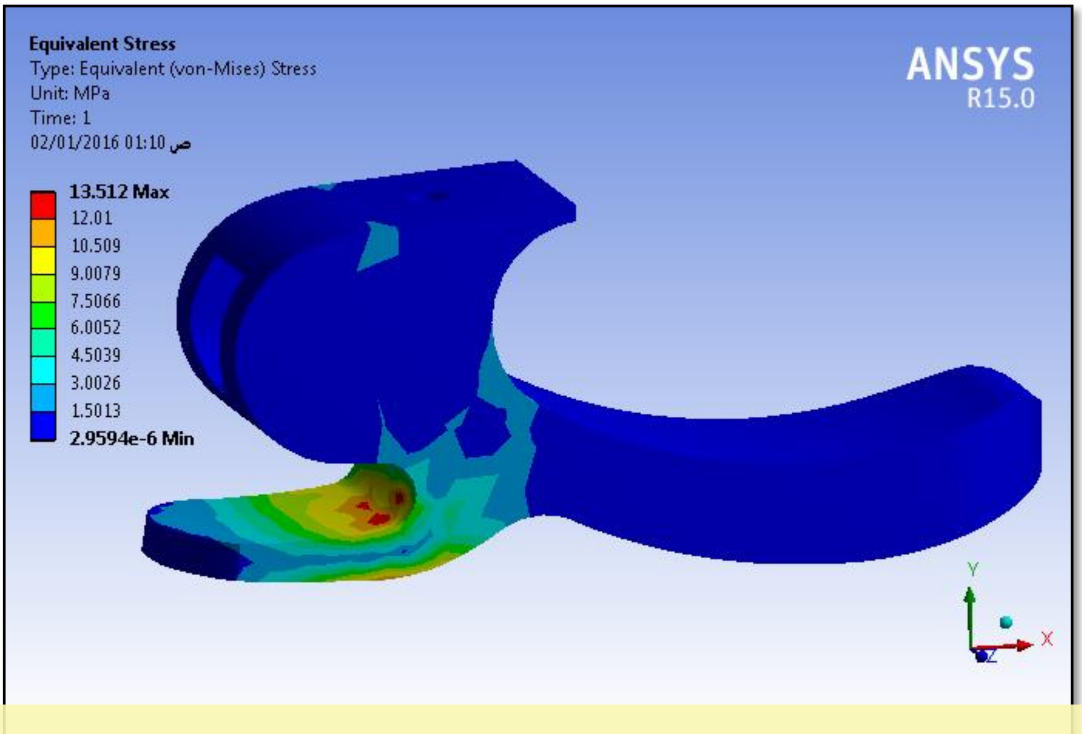


figure 5-18 Equivalent Stress of the Suggested New Design Foot (Heel Contact phase)

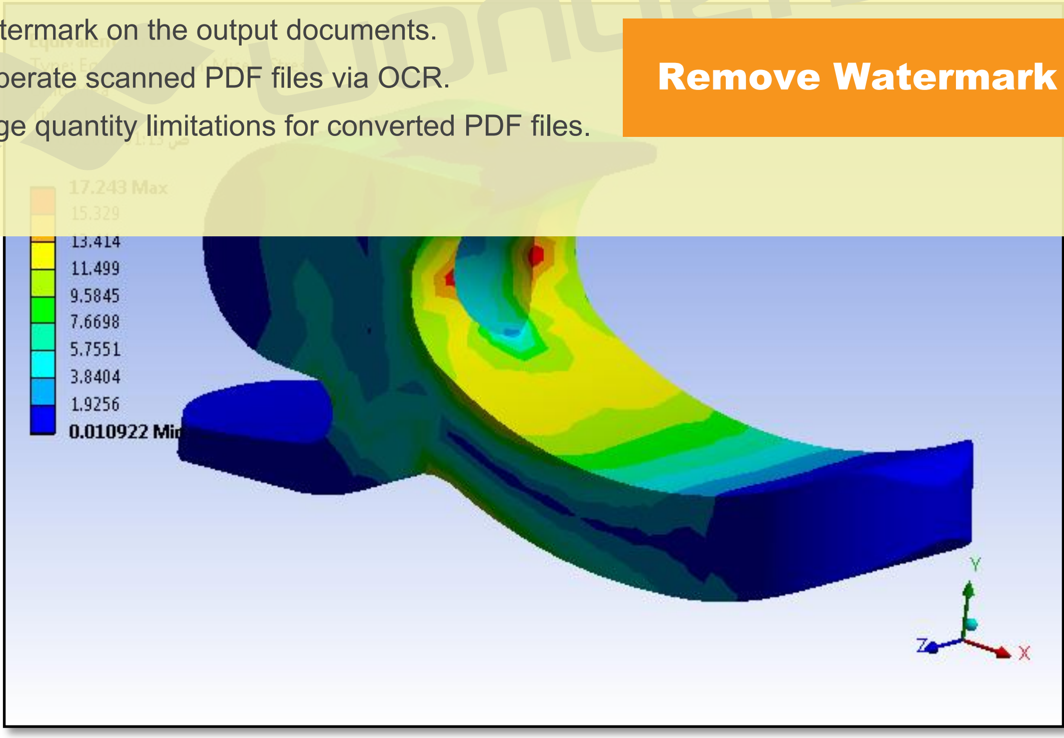


figure 5-19 Equivalent Stress of the Suggested New Design Foot (Toe-Off phase)

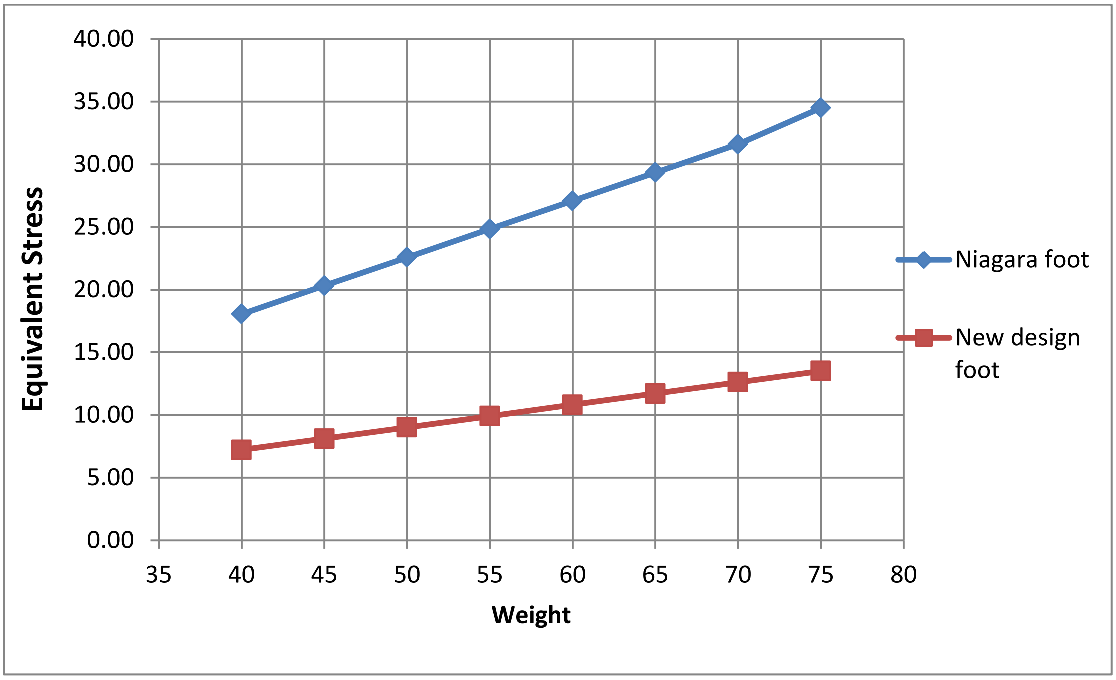


figure 5-20 Equivalent Stress of Heel Contact Phase

Table 5-6 Results of Equivalent Stress of Heel Contact Phase for the prosthetic foots

Weight (kg)	Von-Mises Niagara foot (MPa)	Von-Mises new design foot (MPa)
40	18.058	[illegible]
45	20.315	8.107
50	22.572	9.007
55	24.830	9.908
60	27.087	10.809
65	29.344	11.710
70	31.601	12.611
75	34.510	13.512

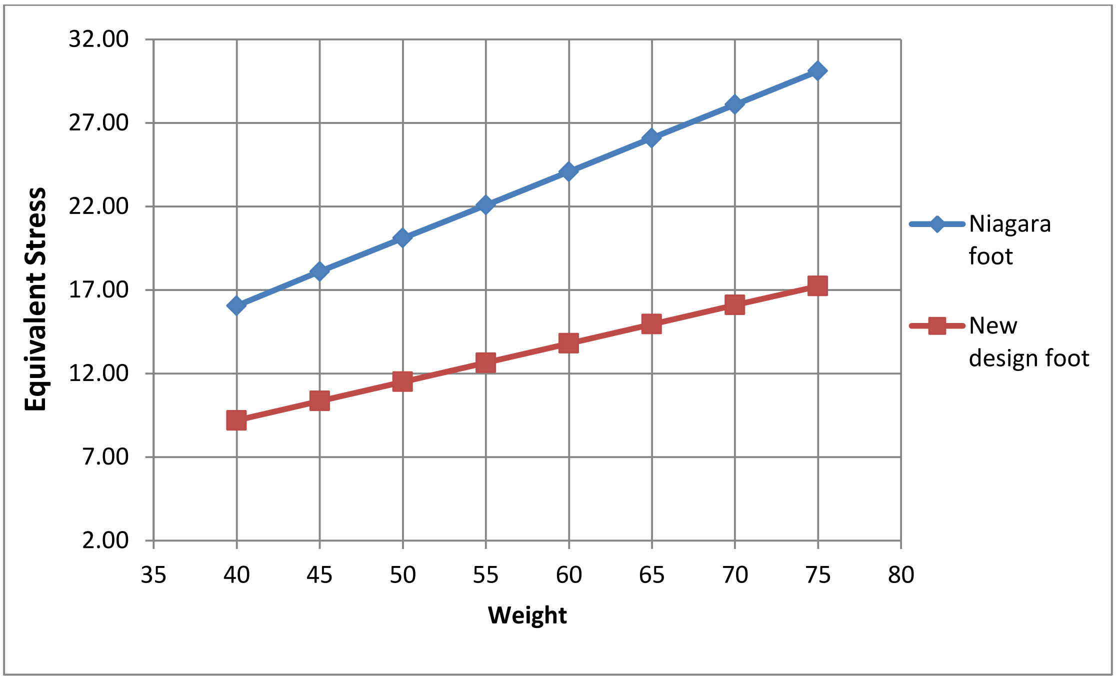


figure 5-21 Equivalent Stress of Toe-off Phase

Table 5-7 Results of Equivalent Stress of Toe-off Phase for the prosthetic feet

Weight	Von Mises Niagara (MPa)	[illegible]
40	16.052	9.19
45	18.095	10.346
50	20.099	11.496
55	22.072	12.64
60	24.079	13.795
65	26.085	14.944
70	28.092	16.094
75	30.098	17.23

5.6.2 Factor of Safety:

The Von Mises stress is compared to the tensile yield strength of the material in order to compute safety factor .The analysis results for Niagara foot shown in figures (5-22) and (5-23), for the new suggested design foot, the results are shown in figures (5-24) and (5-25) for both phases heel contact and toe off phase respectively. In heel contact phase, the minimum factors of safety were 2.38 and 1.7 while in toe off phase were 2.68 and 1.33 for Niagara and the suggested new design foot respectively. The results for different amputee's weight are illustrated in tables (5-8) and (5-9). Figures (5-26) and (5-27) show the comparison for the results of safety factors between Niagara foot and the suggested new design foot for heel contact and toe off phase respectively.

figure 5-22 Safety Factor of Niagara foot (Heel Contact)

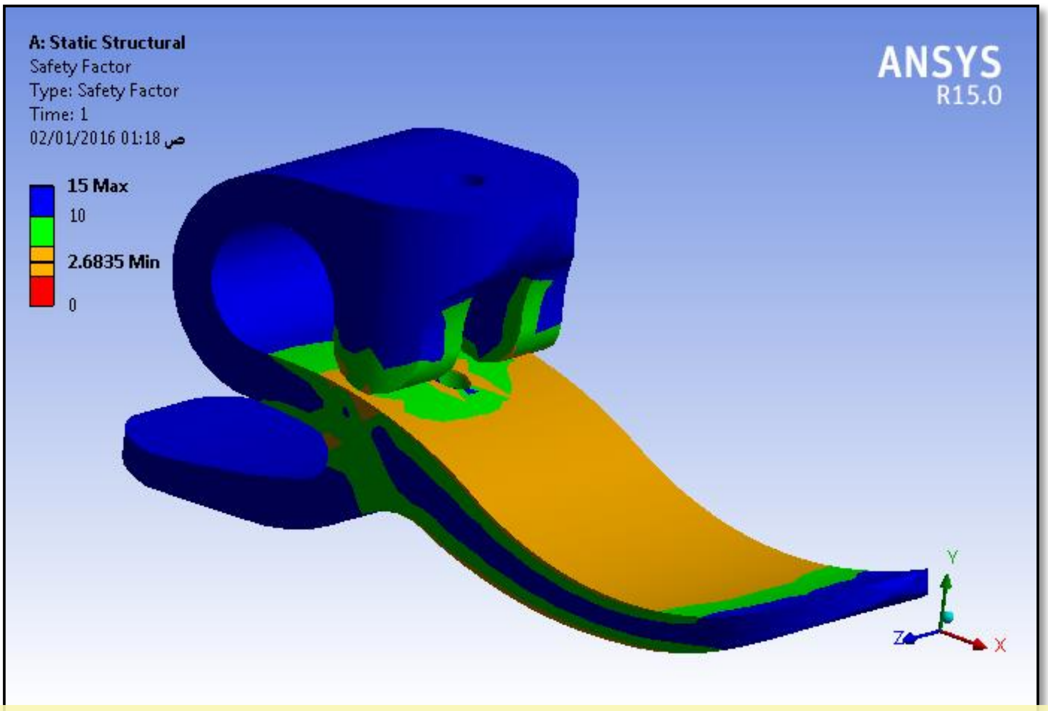


figure 5-23 Safety Factor of Niagara foot (Toe-off)

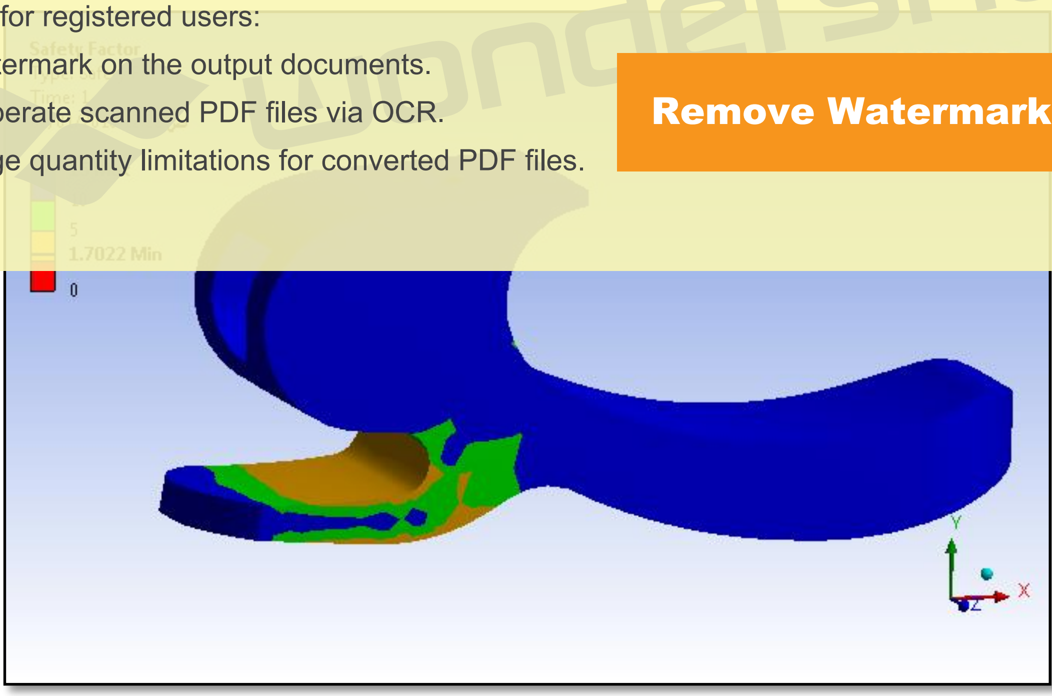


figure 5-24 Safety Factor of the suggested New Design Foot (Heel Contact phase)

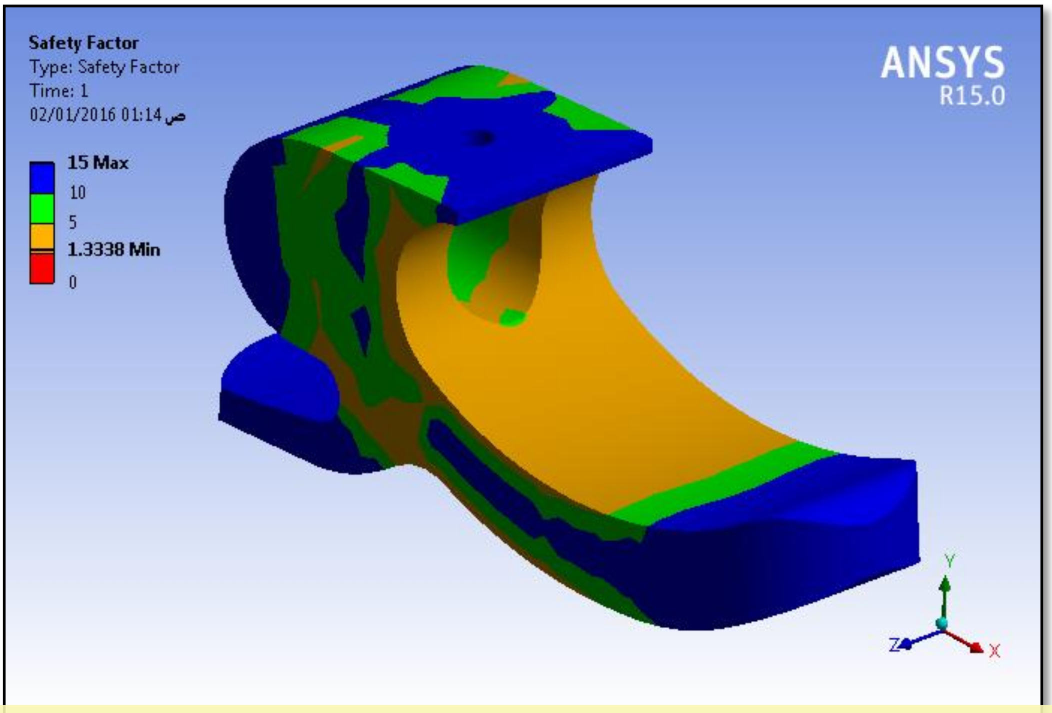


figure 5-25 Safety Factor of the suggested New Design Foot (Toe-off)

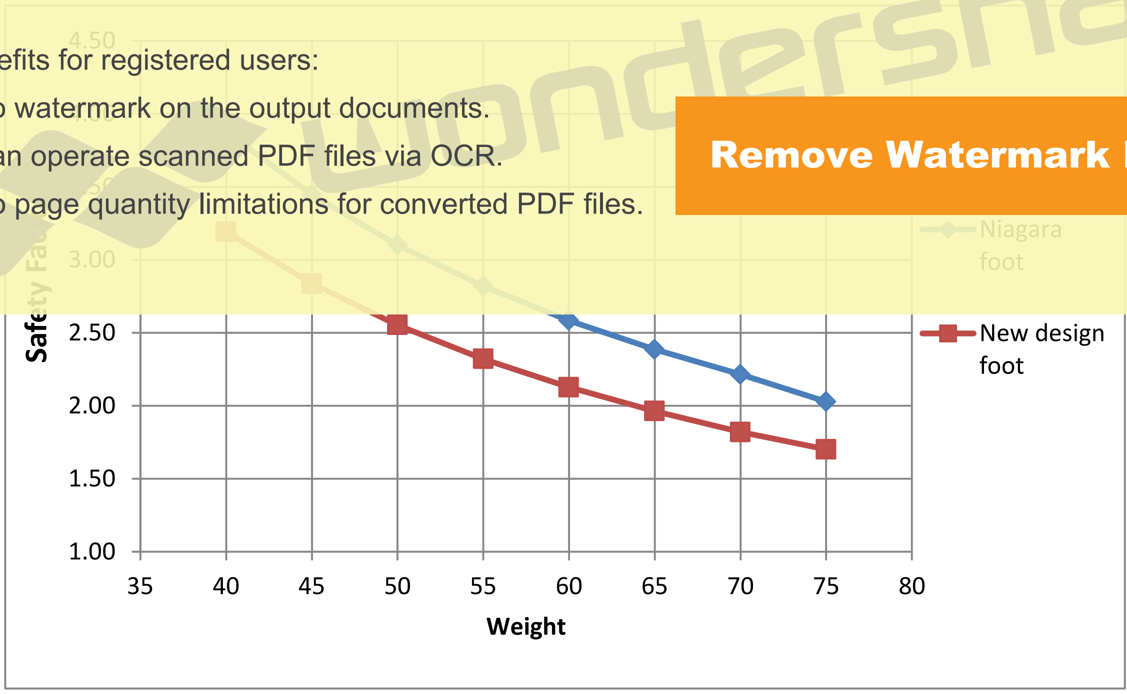


figure 5-26 Safety Factor of Heel Contact Phase

Table 5-8 Safety Factor of Heel Contact Phase for the Prosthetic Feet

Weight (kg)	S.F. of Niagara foot	S.F. of new design foot
40	3.876	3.191
45	3.445	2.837
50	3.101	2.553
55	2.819	2.321
60	2.584	2.127
65	2.385	1.964
70	2.215	1.820
75	[illegible]	[illegible]

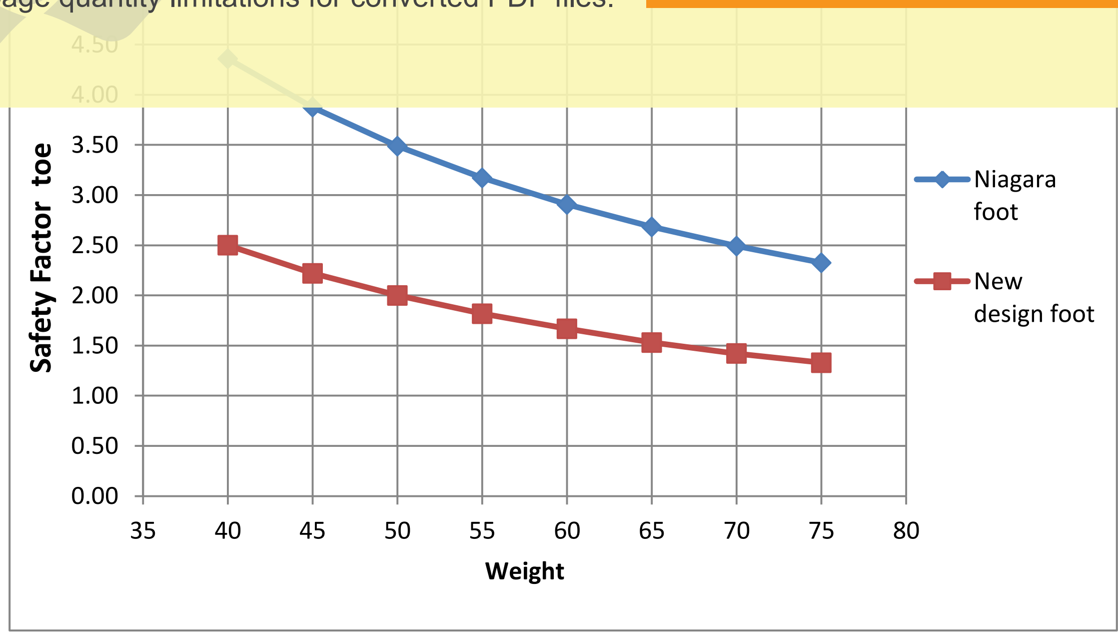


figure 5-27 Safety Factor of Toe-off phase

Table 5-9 Safety Factor of Toe-off Phase for the Prosthetic Feet

Weight (kg)	S.F. of Niagara foot	S.F. of new design foot
40	4.36	2.5
45	3.876	2.22
50	3.488	2
55	3.171	1.818
60	2.907	1.667
65	2.683	1.53
70	2.491	1.42
75	2.325	[illegible]

5.6.3 Total deformation in the Dorsiflexion:

The dorsiflexion angle is considered one of the [illegible] e prosthetic feet, because its effect on the amputee's relief, the dorsiflexion angle calculated from the total deformation in toe off phase as shown in equation (3-2).The total deformation results of analysis are illustrated in Table (5-11) and figure (5-27) for both feet. For Niagara foot, the shape of deformation is shown in figures (5-28) and (5-29). For the new suggested design foot, the shape of deformation is shown in figures (5-30) and (5-31) for heel contact and toe off phase respectively.

Table 5-10 Dorsiflexion Angles for the Prosthetic Feet

weight	dorsiflexion Niagara	dorsiflexion new
40	3.160	2.734
45	3.560	3.075
50	3.955	3.410
55	4.349	3.757
60	4.743	4.098
65	5.136	4.435
70	5.529	4.77
75	5.921	[illegible]

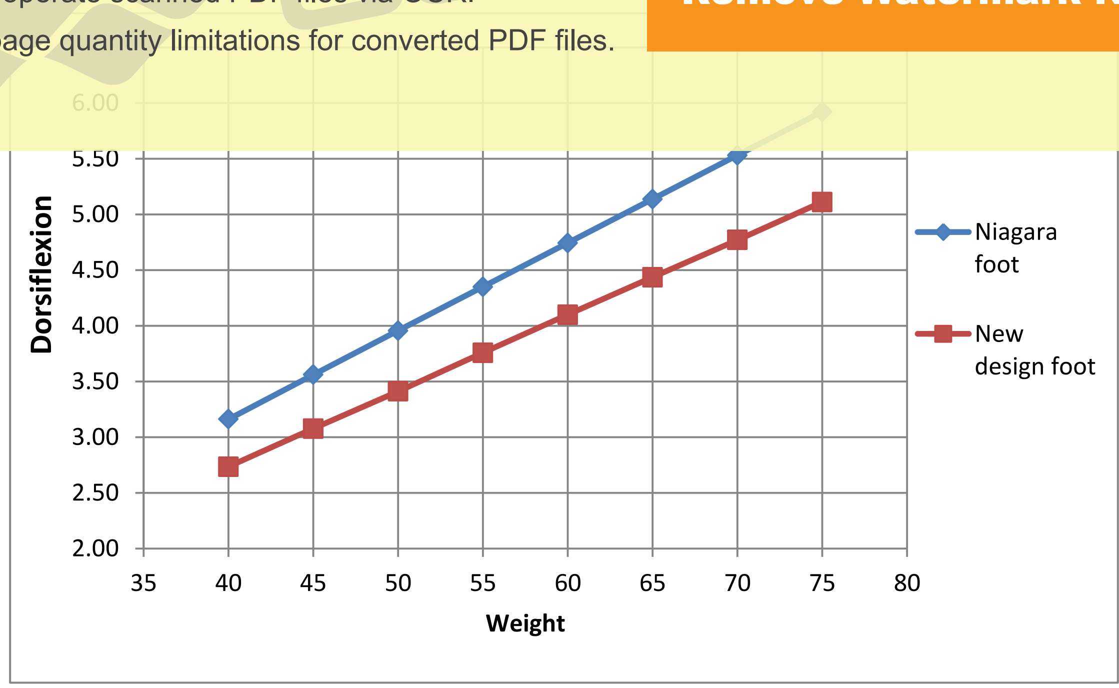


figure 5-28 Dorsiflexion Angles for the Prosthetic Feet

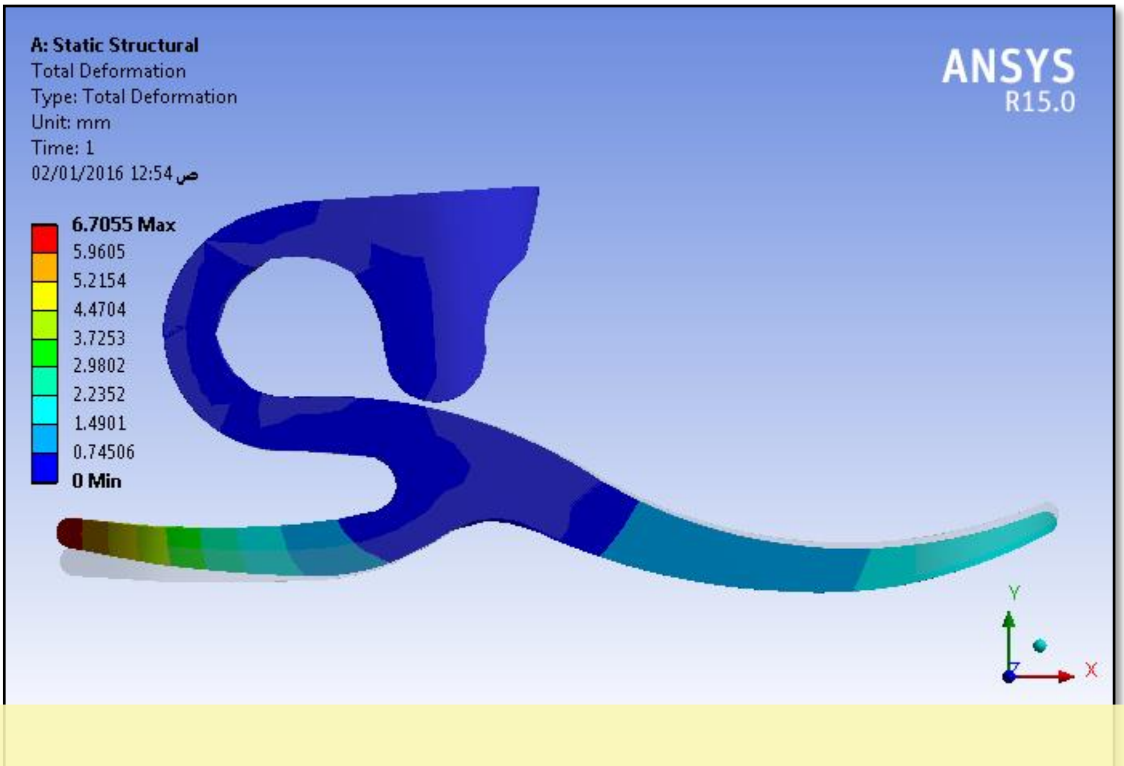


figure 5-29 Total Deformation of Niagara foot (Heel Contact)

figure 5-30 Total Deformation of Niagara foot (Toe-off)

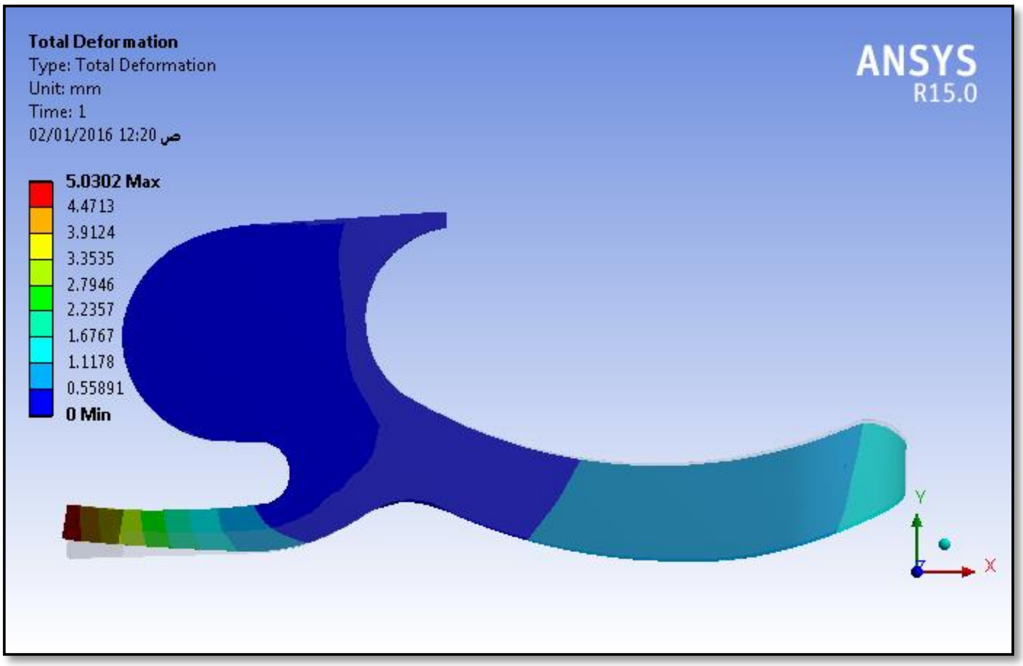


figure 5-31 Total Deformation of the suggested New Design Foot (Heel Contact)

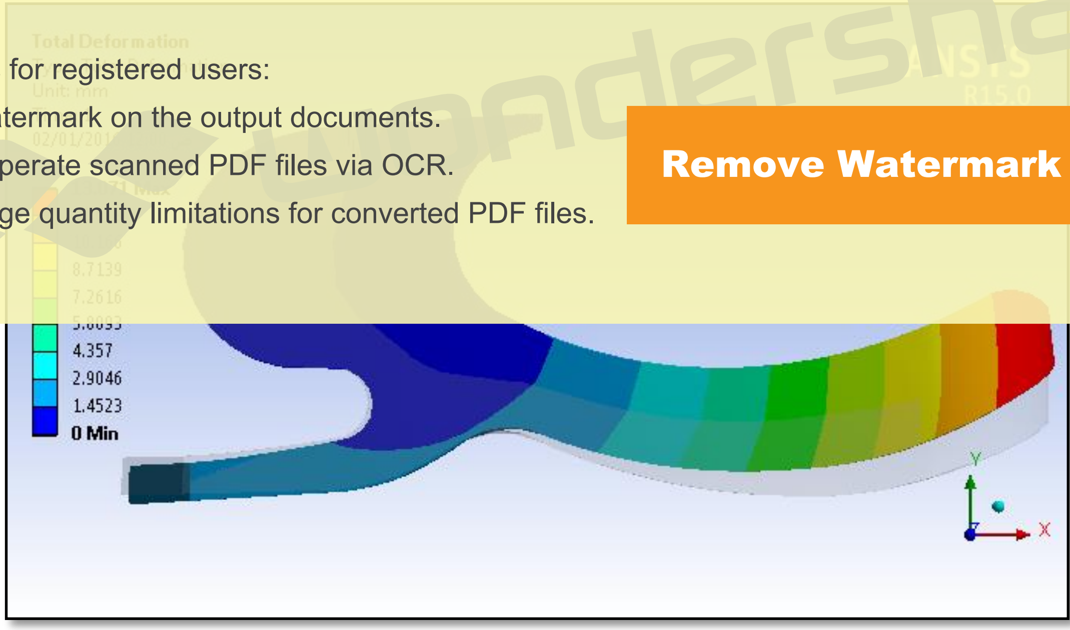


figure 5-32 Total Deformation of the suggested New Design Foot (Toe-Off)

CHAPTER SIX

Conclusion and Recommendation

6.1 Conclusions

This study concentrated in manufacturing new composite materials that can be used with non-articulated developed design foot in this thesis. From the results obtained both numerically and experimentally, the following conclusions can be drawn:

1-The effect of adding chopped carbon fiber to polyethylene increases the ultimate tensile strength and modulus of elasticity.

2-Adding 15% carbon fiber leads to highest toughness comparing to the other ratios of adding chopped carbon fibers to polyethylene.

3-The optimal results achieved from the blends of different grades of polypropylene was with the blend of (50% PP575+50%PP513) which results in high[illegible] toughness and modulus of elast[illegible]

4-Adding $CaCo_3$ to polypropylene leads to increase young modulus and ultimate tensile strength, and that cause a decrease in toughness.

5. The dorsiflexion angle for the Niagara foot is greater than that of

the new design foot by 9% which considered small percentage and it may could be acceptable limit.

6-The range of amputees weight for New design foot up to 75Kg comparing with Niagara foot which has range up to 65Kg.

7. The new design foot is compared with Niagara foot in cost and weight, so that the cost of the new design foot is lower than that of the other by about (85%). Also, the new foot weight is almost equal to that of Niagara foot comparing with using materials with lower properties and design greater in volume.

6.2 Recommendation

Several recommendations for further work can be summarized in the following points:

1. Study the effect of temperature on the mechanical properties of the new composite materials by using creep relaxation device.

2. Test the fatigue resistance for the polypropylene blends.

3.Manufacturing the new design foot and test it experimentally, and compare the experimental and theoretical results.

4. Try to develop new composite material to be used for manufacturing the new design foot which lead to improve the performance of the foot.

Committee Certificate

We the Examining Committee, after reading this thesis **"Design and Analysis of a Non-Articulated Prosthetic Foot Using Composite Materials"** and examined the student **"Mustafa Abdulsalam Kodi"** in its content , find it is adequate as a thesis for the degree of **Master of Science in Mechanical Engineering.**

Asst.Prof.Dr. Abdulrahman N. Abood

(Chairman)

/ /2016

Asst.Prof.Dr. Kadhim K. Resan

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Dr. Ghanim Sh. Sadiq

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Approved by College of Engineering / Al-Nahrain University

Prof.Dr. Jassim Abbood Abbas Aldabbagh

Dean, College of Engineering / Al-Nahrain University

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ABSTRACT

Prosthetic foot is considered one of the most important parts of prosthetic limbs because it is in direct contact with the ground and supporting the body of amputee. Developing designs and materials of prosthetic feet are very important field so as to improve their characteristics to produce better satisfaction for amputees. In this work, previous designs are studied, and new design foot was modeled depending on Niagara foot design. Three groups of new composite materials are prepared. First group is manufactured from polyethylene as a matrix and chopped carbon fibers. Second group is prepared from blending two different grades of polypropylene (polypropylene grade 575, polypropylene grade 513), Third group is prepared from mixing calcium carbonate with polypropylene. The mechanical tests such as tensile, fatigue and impact tests are achieved for these new suggested materials to study the mechanical properties.

For the first group, the results of the tests sh[illegible] with 20% volume fraction produced hig[illegible] modulus of elasticity. The optimum results of toughness obtained from adding 15% volume fraction of carbon fiber, 28 .57KJ/m^2 for groub of 15% carbon fiber and 85% high density polyethylene, 30.76 KJ/m^2 groub of 15% carbon fiber and 85% linear low density polyethylene.

For the second group, the results showed that the highest toughness of 55.74 KJ/m^2 is obtained for the groub of 50% polypropylene grade 575- plus 50% polypropylene grade 513.

The results for the third group showed that, when adding calcium carbonate to polypropylene, the material toughness is decreased while the modulus of elasticity is increased. Adding 0.5% calcium carbonate to polypropylene, the

material toughness is decreased by 40% and when adding 2% decrease the toughness by 50%.

The Niagara foot and the new design foot are analyzed by finite element method by using ANSYS15 software to compute the equivalent stresses, total deformation and factor of safety. The suitable range of amputee’s weight was determined and it was in the range of (40-65) Kg for Niagara foot, while for the new suggested foot design, it is in the range of (45-75) Kg.

The total deformation is calculated for toe off phase for both feet design to obtain the dorsiflexion. The results showed that, both feet give acceptable values of dorsiflexion.

For the calculations of safety factor, the results showed higher safety factor for Niagara foot than the new design foot because of the higher properties of Niagara feet material. The factor of safety [illegible] from [illegible] to 2.4 while the results for Niag[illegible] which means that the obtained results are acceptable according to design principles

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ABBREVIATIONS

HDPE	High Density Polyethylene
LLDPE	Linear Low Density Polyethylene
LDPE	Low Density Polyethylene
PP	Polypropylene
CF	Carbon Fiber
SACH Foot	Solid Ankle Cushioned Heel
TF	Through Foot
TO	Toe off
HO	Heel off
EFLR	effective Foot Length Ratio
FEM	Finite Element Method
CIR	Center International Rehabilitation
GRF	Ground Reaction Force

LIST OF TABLES

LIST OF FIGURES

Supervisor Certificate

I certify that this thesis entitled **"Design and Analysis of a Non Articulated Prosthetic Foot Using Composite Materials"** was prepared under my supervision at College of Engineering /AL-Nahrain University in partial fulfillment of the requirements for the degree of **Master of Science in Mechanical Engineering.**

Supervisor

In view of the available recommendation, I forward this thesis for debate by the Examining Committee.

Asst. Prof. Dr. Ali H. Mohammed

Head, Department of Mechanical Engineering

/ /2016

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43. http://www.niagarafoot.com/niagarafoot/index.html)

الخلاصة

تعتبر القدم الاصطناعية احد الاجزاء المهمة في مجال الأطراف الاصطناعية وذلك لانها في تلامس مباشر مع الارض وتقوم باسناد الجسم. ان تطوير التصاميم والمواد المستخدمة في تصنيع الاطراف يعتبر مجالا مهما في تحسين الخصائص لانتاج اطراف صناعية ذات كفاءة افضل. في هذا البحث تم دراسة التصاميم السابقة واقتراح تصميم جديد لقدم اصطناعية وتمت النمذجة اعتمادا على تصميم قدم نياجارا (Niagara foot.8م)

تم تهيئة ثلاث مجموعات من المواد المركبة حيث تم تصنيع المجموعة الاولى من البولي اثيلين كمادة اساس مدعم بالياف الكاربون المقطعة. المجموعة الثانية تم تحضيرها من خلط صنفين مختلفيين من البولي بروبيلين (575,513) والمجموعة الثالثة تم تحضيرها من كاربونات الكالسيوم والبولي بروبيلين. انجزت الاختبارات الميكانيكية [illegible] لهذه المواد.

[illegible] %20 [illegible] نتائج المثلى(28.52) [illegible] الكاربون الى 85% من البولي اثيلين عالي الكثافة, باضافة 15% من الياف الكاربون الى [illegible] اثيلين الخطي الواطيء الكثافة كانت المتانة $30.76KJ/m^2$

بينت نتائج المجموعة الثانية ان خلط 50% من البولي بروبيلين صنف (575) مع 50% صنف (513) يعطى اعلى متانة مقدارها $55.714KJ/m^2$

نتائج المجموعة الثالثة بينت ان اضافة كاربونات الكالسيوم الى البولي بروبيلين تقلل من متانة المادة وتزيد من معامل المرونة حيث ان اضافة 0.5% من كاربونات الكالسيوم الى البولي بروبيلين تقلل من متانة المادة بنسبة 40% بينما اضافة 0.25% تقلل المتانة بمقدار 50%.

تم تحليل قدم نياجارا وتصميم القدم المقترح باستخدام طريقة العناصر المحددة باستخدام برنامج ANSYS Workbench15 لحساب الاجهادات المكافئة ,التشوه الكلي ,معامل الامان والانحناء الضهري للقدم.

من خلال النتائج تبين ان القدم نياجارا ملائم لاشخاص مبتوري القدم باوزان تتراوح بمدىKg(40-65) بينما التصميم المقرح يكون الوزن بمدى Kg (45-75).

تم حساب التشوه الكلي في طور (Toe-off) لكل من القدمين والذي يمكن من خلاله حساب الانحناء الظهري للقدم وقد بينت النتائج ان قيم الانحناء الضهري تعد مقبولة لكلا القدمين للاوزان المحددة.

حسابات عامل الامان, حيث بينت النتائج ان قيم معامل الامان كانت لقدم نياجارا اعلى من ما هو في التصميم المقترح للقدم وذلك بسبب استخدام مواد ذات مواصفات عالية في تصنيع قدم نياجارا مقارنة بالمواد المحلية المقترحة.قيم عامل الامان تراوحت بين 1.4 الى 2.4 للتصميم المقترح للقدم بينما كانت القيم لقدم نياجارا 2.2 الى 3.87 وهذا يعني ان النتائج التي تم الحصول عليها في استخدام المواد المحلية تعد مقبولة من ناحية التصميم والاستخدام.

DESIGN AND ANALYSIS OF A NON-ARTICULATED PROSTHETIC FOOT USING DEVELOPED COMPOSITE MATERIALS

A Thesis

Submitted to the College of Engineering of Al-Nahrain University in Partial Fulfillment of the Requirements for the Degree of Master of Science

in

Mechanical Engineering

by

MUSTAFA ABDULSALAM KODI

(B.Sc.2012)

Rajab 1437

April 2016

تصميم وتحليل قدم اصطناعية غير مفصلية من مواد مركبة

رسالة

مقدمة ألى كلية ألهندسة في جامعة ألنهرين

وهي جزء من متطلبات نيل درجة الماجستير

في

ألهندسة ألميكانيكية

من قبل

(بكلوريوس 2012 م)

رجب ١٤٣٧هـ

نيسان 2016 م