Research Article

International Journal of Advanced Technology and Engineering Exploration, Vol 11(121)
ISSN (Print): 2394-5443 ISSN (Online): 2394-7454
http://dx.doi.org/10.19101/IJATEE.2024.111100967

Experimental testing and numerical modeling of fatigue and mechanical properties of composite materials for partial foot prostheses

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Received: 05-June-2024; Revised: 03-December-2024; Accepted: 10-Deecember-2024

Abstract

Loss of mobility and functional capacity is a common consequence of partial foot amputation (PFA). Several effective partial foot prostheses (PFP) have been developed to restore movements. The mechanical behaviours, safety features, von Mises stress, and total deformation of a prosthesis designed for Chopart amputation were examined in this research. The fatigue properties of various composite laminate materials were included in the analysis. Three different types of composite materials were analyzed for their fatigue and tensile properties in the fabrication of PFP. The process utilized a vacuum system and an 80:20 composite material matrix for lamination. The composite types included three different combinations: (1) 6 perlon layers; (2) (3perlon – 4glass fiber- 3perlon); (3) (3perlon - 4carbon fiber- 3perlon) layers. Mechanical testing was conducted for each group to determine their ultimate strength (σ_{ULT}), yield stress (σ_Y), and modulus of elasticity (E), providing valuable performance characteristics. The mechanical properties of the composite materials varied across the three groups. The second group demonstrated improved properties, compared to the first group, with E of 1.4 GPa, σ_Y of 75 MPa, and σ_{ULT} of 143 MPa. The third group showed the highest performance, with E of 1.9 GPa, σ_Y of 105 MPa, and σ_{ULT} of 197 MPa. In addition, critical data were obtained from pressure measurements conducted on the patient's stump and socket using the Matrix scan F-sensor. The maximum pressure values recorded were 235 and 273 kPa in the lateral and posterior regions, respectively. Fatigue testing of each group, performed using ANSYS 17.2 software, revealed that the composite material for the third group exhibited a safety factor of 2.43. This safety factor confirms that the material meets acceptable standards, making it suitable for use in the prosthetic design. This research indicates that the material meets acceptable safety requirements, making it suitable for design applications.

Keywords

Partial foot prosthesis, Composite materials, Mechanical properties, Fatigue testing, Chopart amputation, Safety factor.

1.Introduction

Prosthetics are devices designed to replace body parts lost by amputees due to illness or accidents. Typically consisting of a socket, shank, and foot, these devices are crafted to mimic the anatomy and function of human lower limbs [1, 2]. Partial foot amputation (PFA) prostheses aim to replicate a natural stride, protect the remaining extremities, and absorb shock during heel strike and toe-off. Their designs and functions vary depending on the level of amputation, ranging from the ankle to the lower leg [3]. The surgical technique of amputation is crucial for maintaining normal foot and ankle function by preserving residual foot length, which is essential during walking [4, 5].

PFA, also known as ankle disarticulation are surgical techniques used to replace a portion of the foot, with the precise position of the amputation being considered [6, 7]. Amputations may result from trauma, such as frostbite, or congenital abnormalities [8, 9]. In order to treat these diseases, Chopart primarily investigated level amputations via the joint [10, 11]. Chopart's amputation, previously considered limited due to the risk of equinovarus deformity, has gained popularity for its ability to preserve limb length. When combined with ankle fusion and the use of modified shoes, it facilitates propulsive ambulation and improves functional outcomes [6, 12].

Lee's 2020 search identified risk factors for PFA, including osteomyelitis, peripheral artery disease, chronic renal disease, ulcer size, and forefoot ulcer

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site [13]. Martins-Mendes D observed that diabetic foot ulcers had an independent effect on the probability of lower limb amputation and mortality [14]. Dillon et al. conducted a comprehensive analysis to compile epidemiological data on the incidence rate and prevalence of PFA. Their findings indicate that individuals with diabetes have a significantly higher likelihood of developing PFA, approximately 25 times greater than those without diabetes [1]. Kaminski identified prevalence rates of 5.9% for amputation and 14.4% for foot ulcers among individuals undergoing dialysis [15].

Among the numerous issues connected with PFA are medical consequences such as diabetic foot ulcers, peripheral artery disease, osteomyelitis, and chronic renal sickness, all of which increase the risk of amputation and death. PFA often causes functional and mobility impairments, making sophisticated prostheses necessary to regain movement. Designing durable and customised prostheses utilising materials like carbon fibres and simulations using programs like ANSYS for stress and fatigue analysis is still difficult, though. Effective management of PFA is further complicated by socioeconomic hurdles, such as high costs, limited access to care, and insufficient awareness of prevention and rehabilitation.

This study aimed to evaluate the mechanical behavior, safety characteristics, von Mises stress distribution, and overall deformation of various lamination samples composed of (3 perlon - 4 glass fibers - 3 perlon), (3 perlon - 4 carbon fibers - 3 perlon), and (6 perlon), all utilizing an 80:20 resin lamination matrix. The test results were applied in the socket production process to improve durability, elasticity, and load-bearing capacity for PFA prostheses.

The article is structured to provide a comprehensive overview of the work across several sections. Section 1 includes the background, challenges, inspiration, objectives, and contributions. Section 2 reviews relevant studies, followed by a detailed explanation of the proposed methodology in Section 3. The results are presented in detail in Section 4, followed by an in-depth discussion and the study's limitations in Section 5. The conclusions are provided in Section 6.

2.Literature review

This review summarizes recent and comprehensive studies about studying the mechanical properties of the composite materials used in partial foot prostheses (PFP).

A review conducted by Mondal and Ghosh [16] highlighted the impact of loads acting at different gait cycle subphases, such as heel strike and toe-off. Better design and long-term results from newer designs necessitate addressing numerous aspects that are still being defined, such as manufacturing components of prosthetic design, minimal bone resection, misalignment, and bone remodelling. Future research should aim to resolve these clinical challenges.

Finite element analysis (FEA) is widely used to test the material's strength. Researchers used von Mises stress distribution and deformation data to predict the mechanical behaviour.

In their 2021 study, Oleiwi and Hadi [17] examined the specification of materials and foot patterns. The research determined that the SACH feet exhibit superior durability compared to other varieties. The mechanical parameters of SACH are enhanced by its components and the sole rubber. The absorption criteria of SACH are superior to those of Jaipur and Seattle. The Jaipur prosthetic design closely resembles the human foot compared to SACH and Seattle designs. In recent decades, polymers like carbon fibres, fibreglass, and Kevlar fibres have been extensively used in the production of artificial elastic feet due to their high strength-to-weight ratio and excellent shock absorption properties. In their study, Abbas et al. 2018 [18] examined three different polymer sequences used in prosthetic feet to determine their fatigue and mechanical properties by experimental and computational methods. The sequence consisted of eight layers of perlon, followed by four layers of perlon, two levels of carbon, four layers of perlon, four layers of perlon, two layers of Nglass, and finally, four layers of perlon. The results indicate that there was an 8.93% disparity between the findings obtained from the experimental tests and the finite element method (FEM).

A dependable method for assessing a prosthetic socket below the knee against static failure and determining the socket thickness based on the factor-of-safety (FOS) for every amputee was presented by Karthik et al. [19]. The mechanical behaviour, von Mises stress distribution, and damage state of stacked AM sockets with different thicknesses were all simulated with FEA and Hashin's transversely isotropic mechanical damage model, which was created for composite materials. Experimental results indicated that the lobe corner of the 4 mm socket failed at 918.5 N. For this failure load, FEM analysis predicted 896.6 N, which was 2.45% lower than the experimental result. For socket thicknesses of 3, 5, and 6 mm, the FEM predicted failure loads to be 618.1, 1008.6, and 1105.2 N, respectively.

A recent study [20] examined two composite laminates for ankle-foot orthoses [20]. Sequence 1 (2perlon, 1carbon fibre, 2perlon, 1kevlar, 2perlon) and sequence 2 (2perlon, 3carbon fibre, 2perlon, 3kevlar, 2perlon) are used. The first and second layers exhibit ultimate tensile stresses(σ_{ULT}) of 67 MPa and 80 MPa, respectively. ANSYS reports that the safety factors for composite material (sequence 1), composite material (sequence 2), and polypropylene are 2.8, 3.6, and 1.9, respectively. The gap between the yield stresses of composite materials and the maximum stresses on orthoses highlights their ability to support patient weight and function effectively as material alternatives. Sequence 1 polypropylene and sequence 2 composite material peak stresses are 18.0 MPa and 17.5 MPa, respectively. The yield strengths are 50 MPa for sequence 1 and 63 MPa for sequence 2, compared to 24.3 MPa for polypropylene.

Wagle and Padhi [21] use 3-D printing to create a partial foot template mould and a glass fibre partial foot orthosis (PFO) model for diabetics, war veterans, and vascular deficiency patients. The test rig evaluated the PFO model under various stress conditions, simulating real-world forces acting on partial foot amputees. The design was modeled using Unigraphics, and ANSYS 16.0 was employed to analyze the mechanical performance of the PFO under simulated conditions. A load was applied to the partially amputated foot and the reconstructed heel during the analysis. In the test rig, a vernier caliper was used to measure the deformation of the heel and toe. The toe and heel deformed under stresses of 196.13 N and 392.26 N, respectively. The deformation results demonstrated safety and reliability, both analytically and empirically. The findings indicate that glass fiber PFO models can be produced and distributed affordably, making them a viable solution for underdeveloped nations with limited resources.

Carbon fiber has been a dominant material in prosthetic manufacturing for decades due to its superior strength-to-weight and stiffness-to-weight ratios. It is suitable for prosthetics due to their high strength-to-weight and stiffness-to-weight ratios. Previously, prosthetic devices were made of wood and metal. Maruo et al. [22] raised concerns about the susceptibility of traditional prosthetic materials, such as wood and metal, to moisture, corrosion, and environmental dampness. Research has emphasized the necessity of using robust and stable synthetic fibers for prosthetics to ensure amputees experience optimal comfort and control. The biocompatibility and superior strength-to-weight ratios of composite materials make them ideal for prosthetic applications [23]. Carbon fibres' energy storage and versatility have made them vital to prosthetic use, according to Rościszewska and Wekwejt [24]. Carbon-reinforced composites made from epoxy resins and woven carbon fibres may change their characteristics. Oleiwi and Hadi [25] stated that manipulating angles and matrix might provide lamination with unique tensile qualities and stiffness. Abbas and Kubba [26] established an experimental approach to build blended polymer frameworks for prosthetic limbs using lamination methods and examined a novel resin composition with the goal of improving socket safety factors. In order to give the patients both functioning and support, Zaier and Resan worked together on a number of lower limb prosthetics-related features [27, 28]. Furthermore, a study examined the effect of carbon fiber layers on the mechanical performance of through-knee prosthetic socket assemblies, highlighting significant improvements in strength and durability [29].

Orthotic and prosthetic devices are constructed using metal, plastic, and advanced laminations through modern rapid prototyping techniques. With hand layup without hoover bagging, Sehar et al. [30] examined the effects of increasing carbon fibre concentration in carbon-fibre composites. In sample preparation, more carbon-fibre layers increase the mechanical strength of carbon-fibre laminate. Various carbon-fibre laminates are tested for tensile strength. Laminating black 100% 3 K 200 gsm carbon fibre with a fabric thickness of 0.2 mm and tensile strength of 4380 Mpa using two parts epoxy resin Araldite® LY556 and Aradur hardener at a 100:30 ratio produced the sample specimen. Adding carbon-fibre layers to successive samples increased carbon-fibre laminate tensile strength. The testing procedure achieved a maximum tensile strength of 576.08 N/mm² with 10 carbon fiber layers. In tensile testing of built samples, increasing loading from 2 to 5 mm/min enhanced deformation. The tested carbon fiber composites, with modifications to experimental parameters, show potential for use in the construction of prosthetic feet. Building on the reviewed literature, researchers explored various laminations to improve the mechanical properties of prosthetic materials. Therefore, this study included tensile and fatigue testing on several samples of laminations composed of (3 perlon - 4 glass fibres - 3 perlon), (3 perlon - 4 carbon fibres - 3 perlon), and (6 perlon) with a lamination matrix of (80:20) resin. These test results guided the fabrication of a socket for partial foot amputees (PFA), aimed at achieving a material with superior mechanical properties. The fatigue safety factor of the fabricated socket was determined through simulations performed in the ANSYS workbench environment.

3.Materials and methods

3.1Materials

Selecting the materials needed to make the partial foot prosthesis was an important stage in the manufacturing of prosthetic sockets according to the patient's lifestyle. Acrylic resin (80:20), glass fibre, carbon fibre, perlon, and PVA were the materials selected for the partial foot socket lamination, as shown in *Table 1*.

Table 1 Mechanical properties of different materials [31]

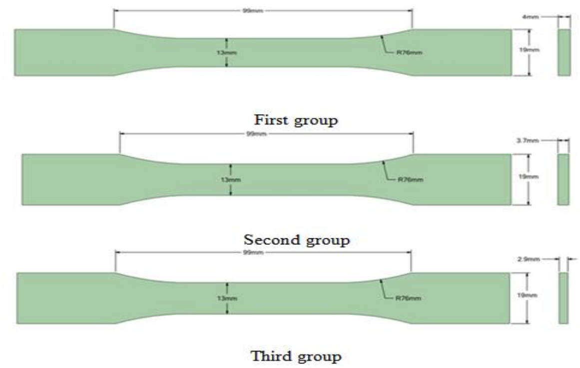
Materials	Density (g/cm^3)	Elastic modulus (GPa)	Poisson ratio
Perlon fibre	1.083	0.8-2	0.39
Carbon fibre	1.76	1.5-3	0.3
Glass fibre	1.44	1.2-4	0.4
lamination resin	1.04	1.24	0.35

3.2Procedure for manufacturing sample test

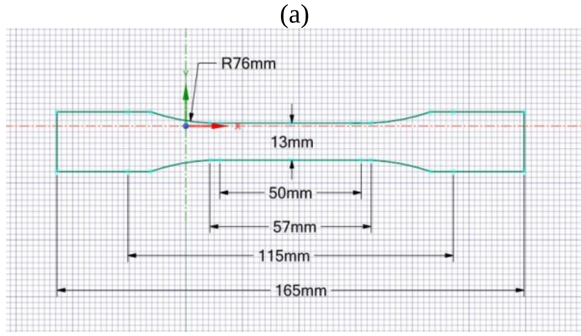
After selecting the materials, tests were conducted to verify their suitability for constructing a prosthetic foot for partial Chopart amputation. Following the procedures outlined below, the samples were prepared for testing:

- Laying the rectangular mould out on the vacuum pressure system's platform.
- Stockinets are utilized in an overlaying layup composed of perlon, carbon fiber, and a combination of perlon and glass fiber.
- The overlay resin, composed of an 80:20 polyurethane mixture, is combined with the hardener.
- After the laminations have cooled, maintain a constant vacuum pressure at ambient temperature within the range of 40 to 60 kPa. Once stabilized, cut the laminations into the required number of samples.

In accordance with the American society for testing and materials (ASTM) D638 type I standards [32], three samples for each lamination were created for the tensile test; the dimensions of the tensile sample are shown in *Figure 1*.



(a)



(b)

Figure 1 (a) Dimensions of tensile specimen for first, second, and third groups. All dimensions are in millimeters (mm) (b) Detailed drawing of tensile specimen with key dimensions in mm

For every material group, many fatigue samples were evaluated in order to evaluate the fatigue properties at room temperature. For the fatigue test, eight specimens per group were employed under repeated load (tension-compression). The load was applied to the fatigue samples based on the tensile test results for each group. The sample specifications, as shown in *Figure 2*, were as follows: a length of 100 mm, a breadth of 10 mm, and a thickness that varied depending on the type of laminate.

Testing was conducted to evaluate the mechanical and fatigue characteristics of the materials after the tensile and fatigue samples were prepared. The fatigue tools used for the material fatigue tests, as shown in *Figure 3*, were designed for flat specimens.

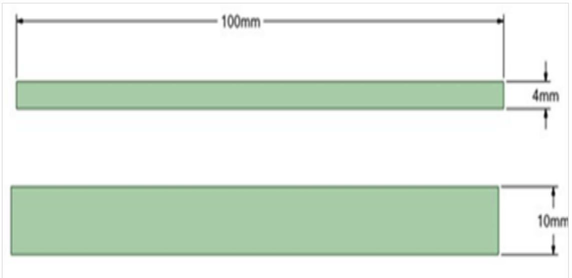


Figure 2 Specimen for fatigue test

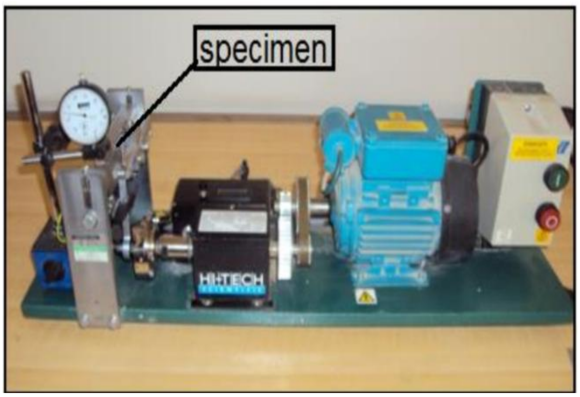


Figure 3 Fatigue test device

3.3Procedure for manufacturing partial foot prosthetic

The following crucial processes are involved in the production of the partial foot prosthesis:

- Measurement: Measure the patient's length, the length of their normal foot, the circumferences of the stump, the M-L diameter at the calcaneus and malleolus, and the heel height of their shoes.
- Hand casting: Form a cast around the patient's remaining appendage.
- Rectification and cast rectification: Get the cast ready for processing by making the necessary corrections.
- Fabrication of the soft socket: Build the prosthetic's soft socket.
- Application of lamination, trimming, and finishing: Finish the finishing touches, laminate the socket, and execute any necessary trimming.

3.4F-socket test

The F-socket is used to test the partial foot prosthesis following the manufacturing process. An interface pressure (IP) test is conducted on a patient with a left foot Chopart amputation caused by an explosion. For this dynamic load evaluation, a sensor system (Mat Scan) is utilized, as illustrated in *Figure 4.*

Figure 4 Left foot amputation for patient

3.5Participant

The research participant was a 48-year-old man who underwent a PFA of his left leg eight years ago due to an accident. Testing was conducted in compliance with ethical approval granted by Al-Nahrain University's College of Engineering (02/2020). *Table 2* presents the demographic characteristics of the amputee subject and details of the prosthetic limb.

Table 2 Demographic features of the participant

Variable	**Amputee Subject's Data**
Age (year)	48
Height (m)	1.80
Body weight (kg)	82
Body mass index (kg/m^2)	25.3
Duration of amputation (year)	8 years
Type of amputation	PFA
Amputated side	Left
Cause of amputation	Explosions

3.6Numerical analysis

The fatigue properties of the prosthetic component (safety factor, total deformation and stresses) were assessed in this study using the FEA. The prosthetic geometry is measured and simulated in ANSYS software as shown in *Figure 5*. The mechanical characteristics that were employed in this model were obtained from the three sets of composite material experiments that were employed to create the socket. The stresses caused by the weight and walking loads on the prosthesis were assessed using the mechanical characteristics of the three distinct groups. Ten-node tetrahedral (SOLID187 element type) with 9303 elements and 7804 nodes were adopted for the numerical simulation of the prosthesis.

3.6.1Prosthetic geometry

As shown in *Figure 5*, the geometry of the prosthesis was measured and input into the ANSYS program to enable analysis using the mechanical properties of the materials used in its construction. The stresses generated in the prosthesis under the influence of walking loads and body weight were evaluated based on the mechanical properties of three distinct material groups.

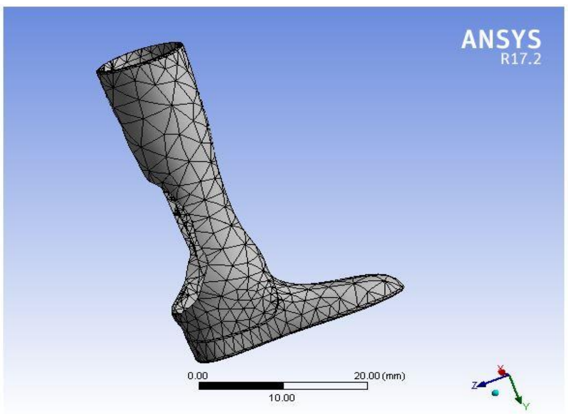


Figure 5 Prosthesis geometry

3.6.2Boundary conditions

Figure 6 illustrates the applied forces and boundary conditions for prosthesis displacement. Specifically, one of the key boundary conditions involves defining the interaction between the prosthetic foot and the ground to simulate real-world conditions.

In the first scenario, with stationary heel support, the constraints and forces applied are designed to replicate the support and movements associated with various activities. In the second scenario, with fixed metatarsal support, the focus shifts to simulating conditions related to the front portion of the foot. These boundary conditions are established to provide a comprehensive analysis of the prosthetic foot's behavior in different scenarios, contributing to the evaluation of its overall performance. The patient's weight, the experimental settings, and the applied forces were compared with the displacement boundary conditions. The program was modified to incorporate the fatigue mechanical properties of the materials used in this study to evaluate the safety factor of the prosthesis. For the numerical modeling of the prosthesis, as shown in *Figure 7*, a ten-node tetrahedral structure was employed. This structure is of the SOLID187 element type and consists of 9,303 elements and 7,804 nodes.

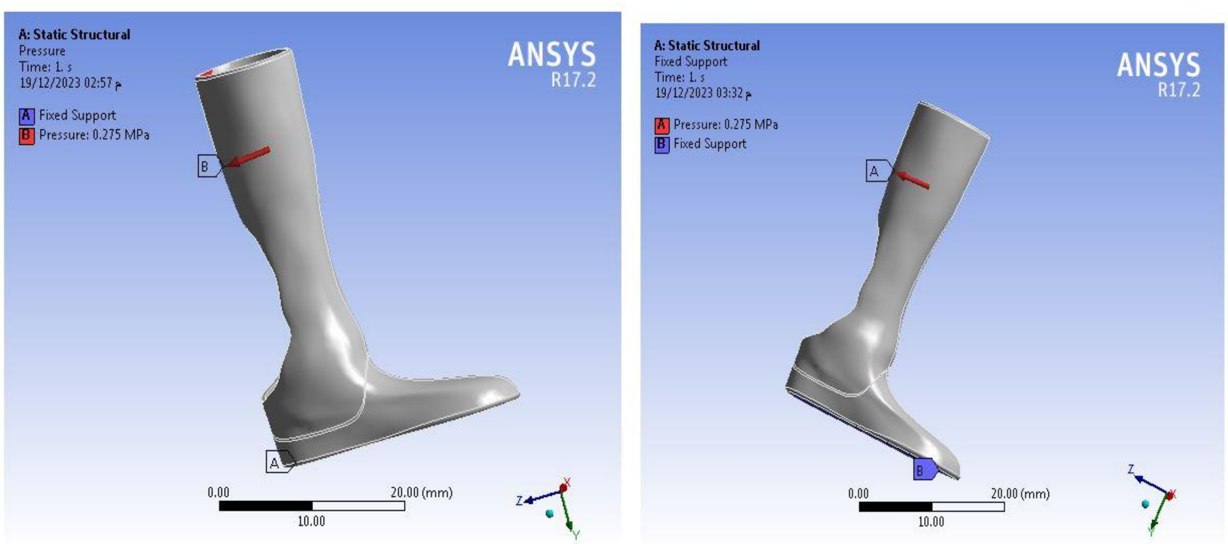


Figure 6 Prosthesis displacement conditions and applied forces: the right panel illustrates the heel strike, while the left panel depicts the toe-off phase

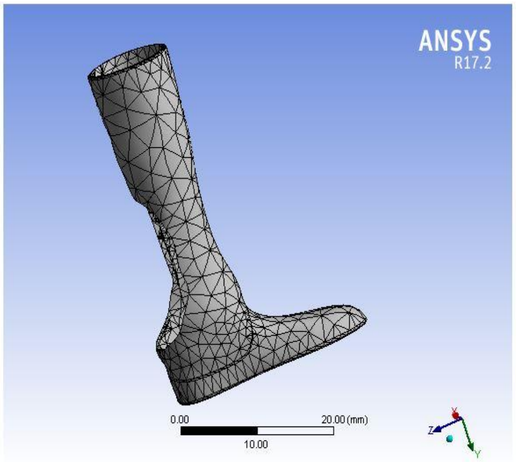


Figure 7 Prosthesis elements

4.Results

4.1Tensile test results

Mechanical properties for all groups are presented in *Table 3*. From the curves, the tensile of samples are determined and recorded in *Figure 8.*

4.2Fatigue properties results

When a flat specimen breaks under alternate stress, fatigue failure takes place. The number of cycles till specimen breakage was recorded by the fatigue tester equipment. *Figure 9* shows the S-N curves for fatigue. Notably, at a constant temperature, the number of cycles required to reach failure increases while the failure stresses drop. This data sheds light on the various laminations' fatigue behaviour and longevity under repeated stress scenarios.

Table 3 Mechanical properties of composite groups with different layer configurations

No.	**Total Layers**	**Thickness (mm)**	**Yield tensile stress (σ_Y) MPa**	**Ultimate tensile stress (σ_{ULT}) MPa**	**E GPa**
First group (6Perlon) layers	6	4	18	36	1.16
Second group (3perlon – 4glass fiber- 3perlon)	10	3.7	75	143	1.4
Third group (3perlon - 4carbon fiber- 3perlon)	10	2.9	105	197	1.9

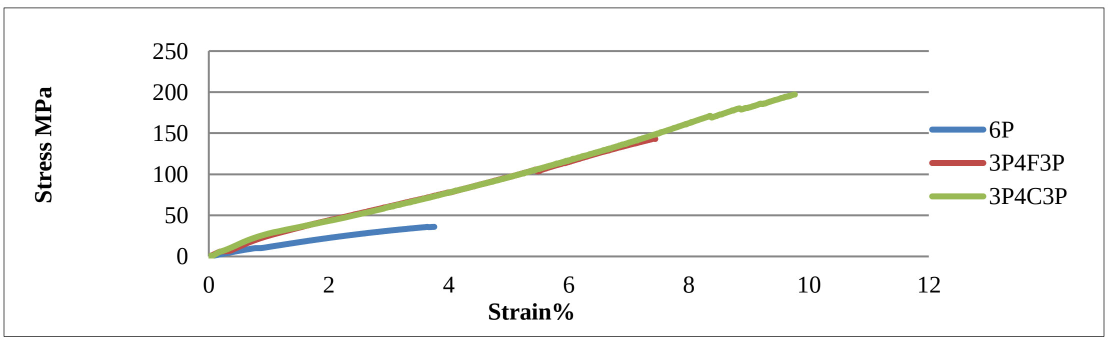


Figure 8 Tensile test curve for each group

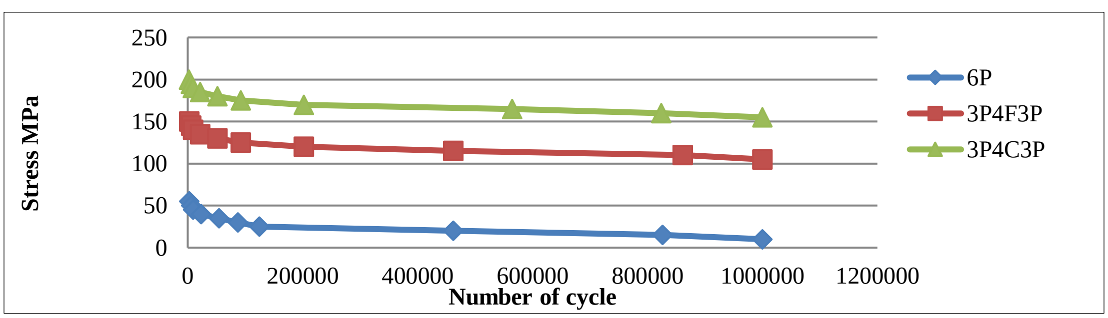


Figure 9 S-N curve for fatigue

4.3F- socket result

F Socket sensor measures the pressure at the contact between the patient's stump and socket. To determine the pressure by sensor, the sensor was linked to software (F-Socket). As seen in *Figures 10 to 13*, the sensor was placed on each side of the socket with the stump. *Table 4* provides specific values for the socket pressure and locations. There is an increase in pressure at the lateral (236 kPa) and posterior (275 kPa) regions. The patient's lateral and posterior muscles contract more during movement, preventing pressure from building up at the tibia bone.

Table 4 Interface pressure across different regions

Regions	**Anterior side**	**Lateral side**	**Posterior side**	**Medial side**
Interface Pressure (KPa)	144	236	275	174

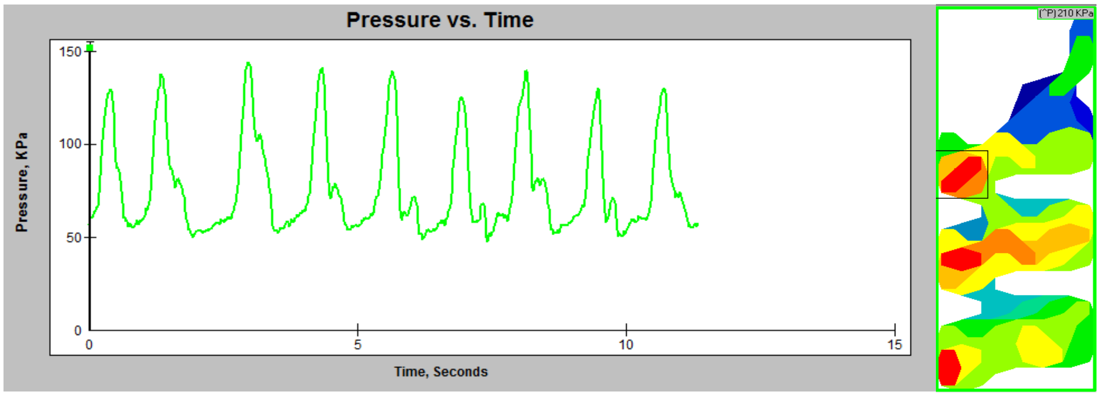


Figure 10 Anterior pressure – time

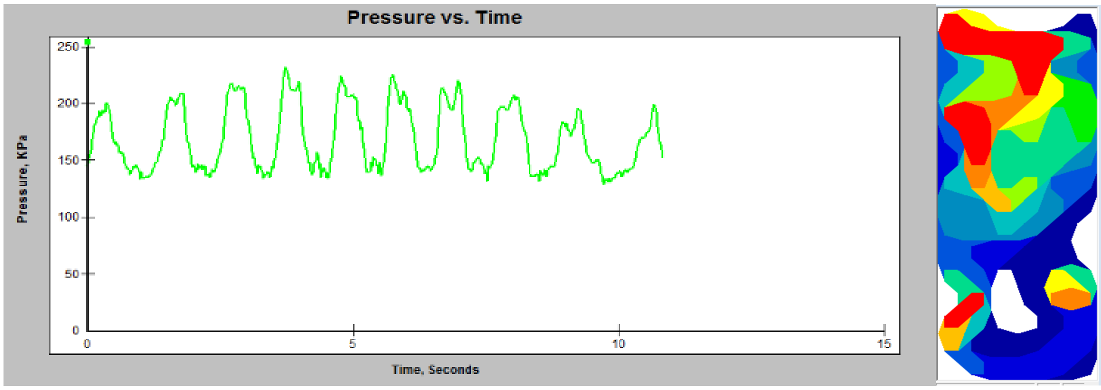


Figure 11 Lateral pressure – time

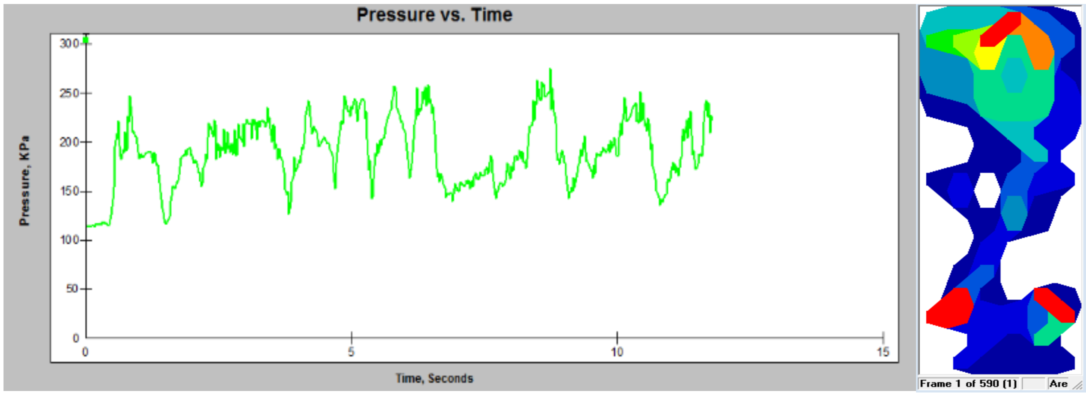


Figure 12 Posterior pressure – time

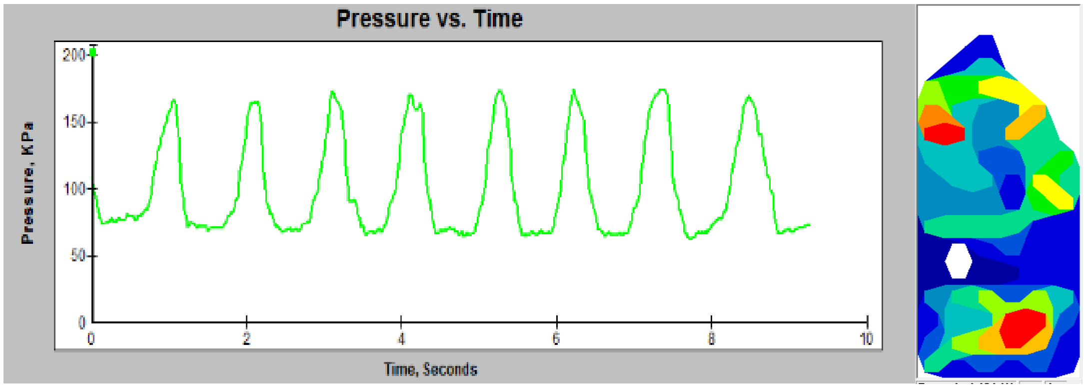


Figure 13 Medial pressure – time

4.4Numerical results

The equivalent (von Mises) stress, total deformation, and fatigue safety factor were calculated using FEM software on a partial prosthetic model for the patient. Von Mises theory states that the yield stress is evaluated using three criteria: $\sigma_E < \sigma_Y$, safe; $\sigma_E = \sigma_Y$, critical; and $\sigma_E > \sigma_Y$, failed. When the fatigue safety factor is equal to or more than 1.25, it is safe in design and acceptable.

Table 5 and *Figures 14 to 16* illustrate the safety considerations of different composite prosthesis socket models. The von Mises stress values are affected by changes in material qualities. *Table 6* presents the von Mises stress for three groups of composites, which is further illustrated in *Figures 17 to 19*. *Table 7* and *Figures 20 to 22* display the overall deformation findings for each kind of lamination composite.

Table 5 Safety factor comparison of fixed positions across different groups

Fixed Position	First group	Second group	Third group
When the heel is fixed	0.436	1.116	1.483
When the metatarsal is fixed	0.686	1.161	1.593

Table 6 Von Mises stress comparison of fixed positions across different groups

Fixed position	First group	Second group	Third group
When the heel is fixed	197.37	223.93	251.2
When the metatarsal is fixed	186.61	215.32	233.26

Table 7 Total deformation comparison of fixed positions across different groups

Fixed position	First group	Second group	Third group
When the heel is fixed	0.0639	0.1937	0.1727
When the metatarsal is fixed	0.0604	0.1799	0.1662

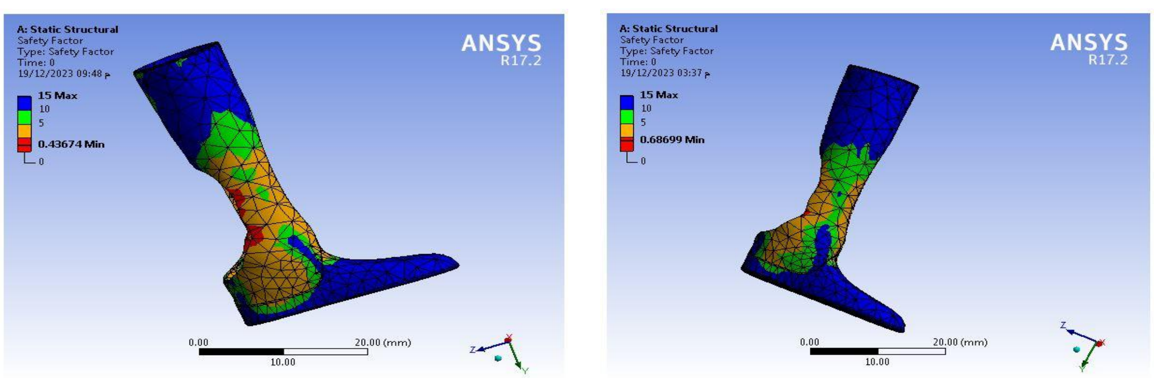


Figure 14 Factor of safety for the first group: the right panel illustrates the heel strike, and the left panel at the toe off

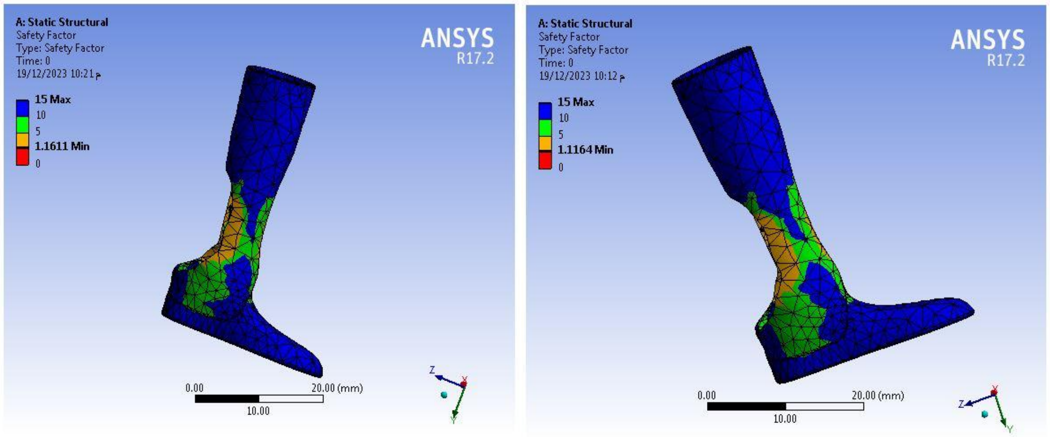


Figure 15 Factor of safety for the second group: the right panel illustrates the heel strike, and the left panel at the toe off

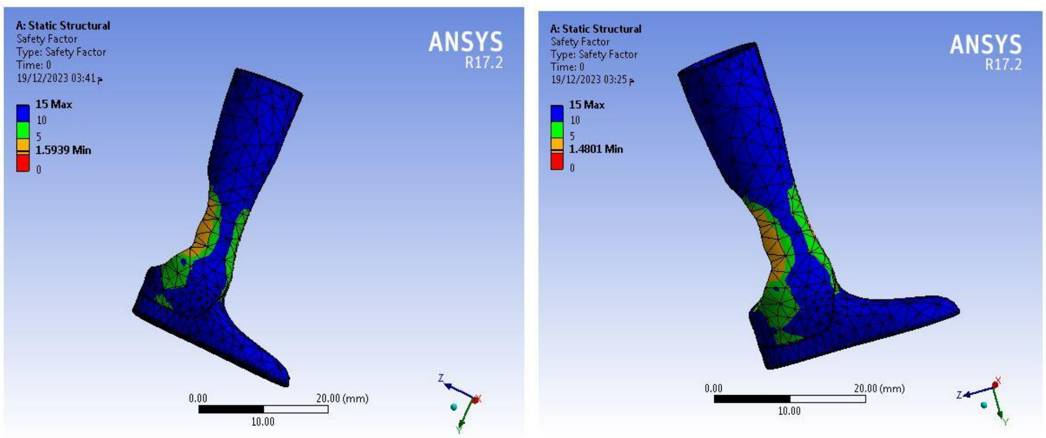


Figure 16 Factor of safety for the third group: the right panel illustrates the heel strike, and the left panel at the toe off

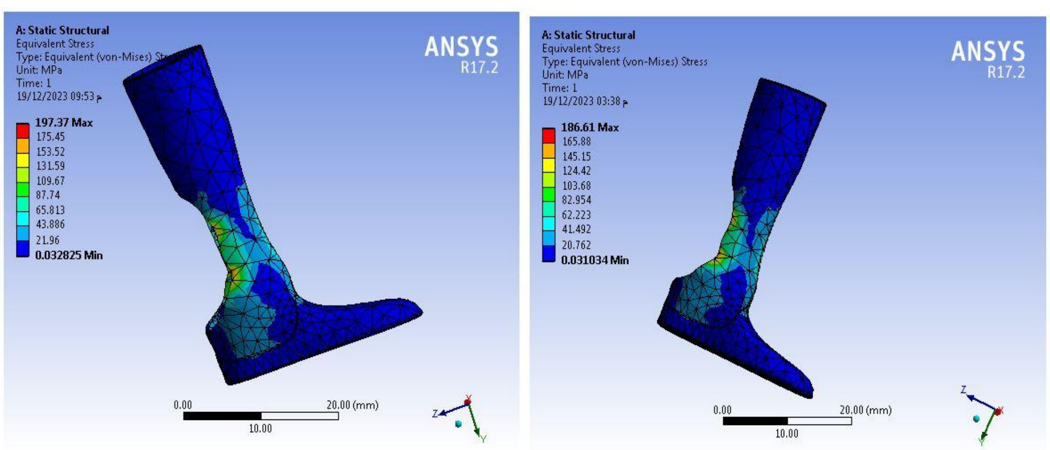


Figure 17 (von Mises) For the first group: the right panel illustrates the heel strike, and the left panel at the toe off

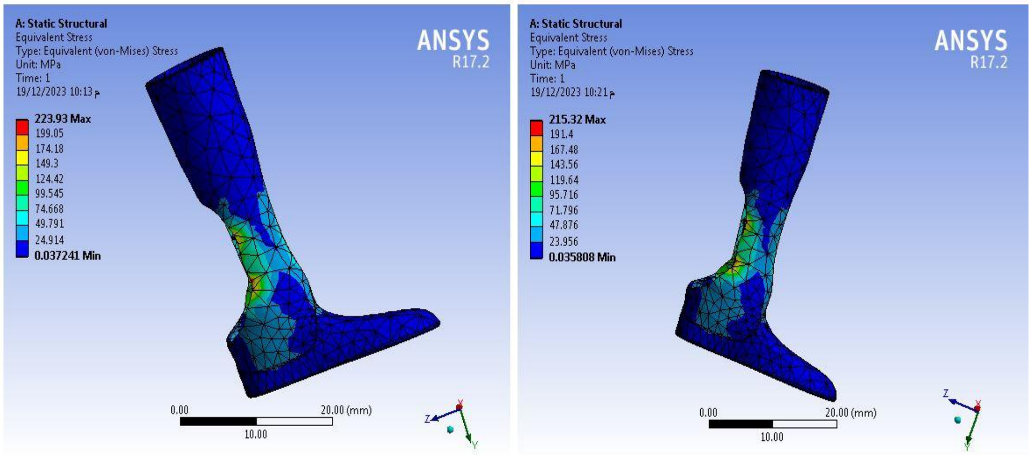


Figure 18 (von Mises) For the second group: the right panel illustrates the heel strike, and the left panel at the toe off

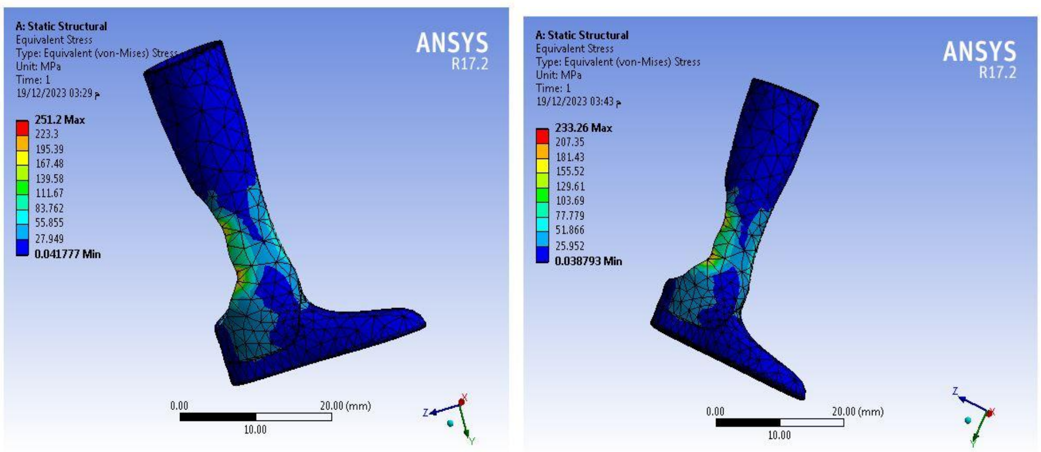


Figure 19 (von Mises) For the third group: the right panel illustrates the heel strike, and the left panel at the toe off

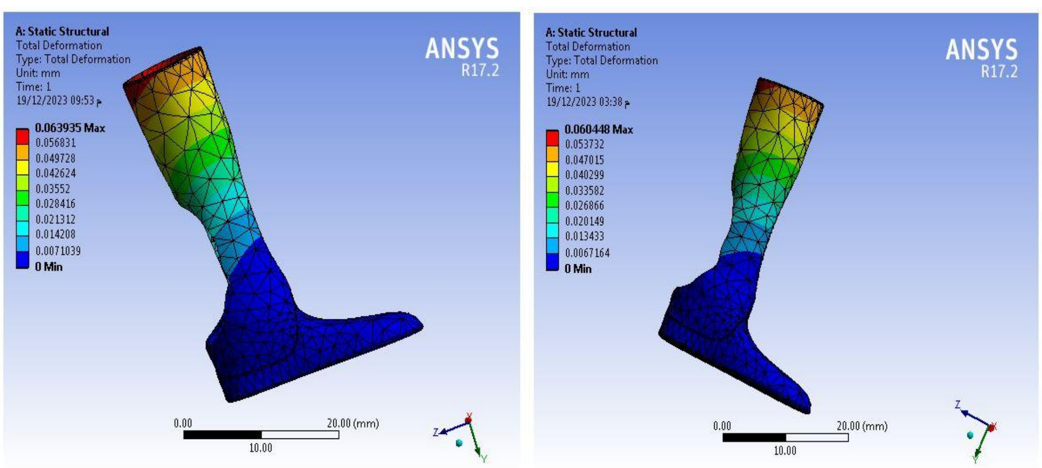


Figure 20 Total deformation for the first group: the right panel illustrates the heel strike, and the left panel at the toe off

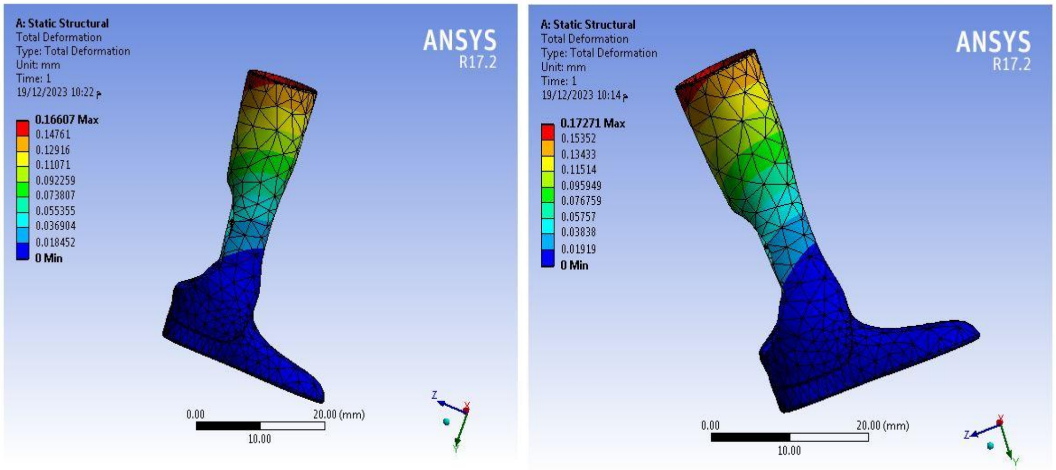


Figure 21 Total deformation for the second group: the right panel illustrates the heel strike, and the left panel at the toe off

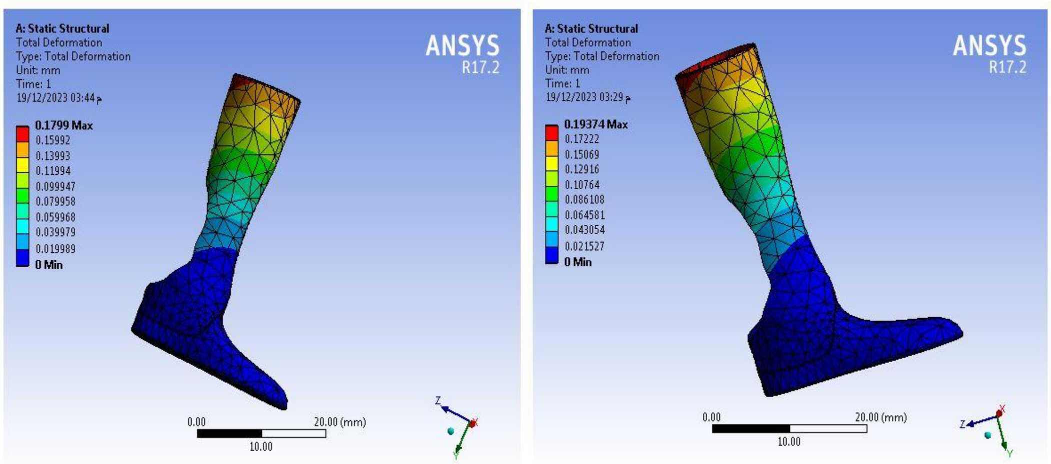


Figure 22 Total deformation for the third group: the right panel illustrates the heel strike, and the left panel at the toe off

5.Discussion

The main aspect of this study is that the four layers of carbon fibre in the midst of the perlon layers and their matrix design are largely responsible for the optimal assessed composite material's superior performance. How do the layers contribute to increasing σ_Y and σ_{ULT} in the study. The materials used to make the prosthesis in this study underwent a tensile test, and the results indicated that when compared to the first group, the second group showed a 76% increase in yield strength (σ_Y), an approximate 75% increase in ultimate tensile strength (σ_{ULT}), and a 17% increase in modulus (E).

In addition, four more layers of glass fibre, along with continuous perlon were added to this second group. In comparison to the first group, the third group's yield strength (σ_Y), ultimate tensile strength (σ_{ULT}), and modulus (E) rise noticeably with the addition of four more layers of carbon fibre with constant perlon, by around 80%, 83%, and 39%, respectively as compared with [33–36].

The superior mechanical properties of carbon and glass fibres over Perlon account for these improvements. This indicates that it has high mechanical strength in comparison to polypropylene's mechanical qualities (maximum of 43 MPa), where it is typically utilised in orthoses or socket prostheses. These differences in results, according to the adding number of carbon or glass fibre layers, lead to an increase in the mechanical properties to improve the socket prosthesis.

The recommended composite material (Group C) performs better mechanically than other studies. When comparing the ultimate stress from this study (197 MPa) to 55 MPa [37] and 121 MPa [38], it shows a significant increase. This improvement can be primarily attributed to the carbon layer structure of the matrix and its four layers.

Additionally, it was found that when the temperature remains constant, the failure stress rate of the samples decreases as the number of cycles increases and the samples approach their failure stress. The recommended composite material (Group C) improves stress endurance, as demonstrated by the fatigue testing. This enhancement is due to the superior mechanical properties of carbon fiber. The homogeneity of the prosthetic socket fit for prosthetic fabrication operations was verified using the F-socket. The test, as shown in *Table 3*, revealed that the lateral (236 kPa) and posterior (275 kPa) areas exhibited the highest-pressure values between the stump and the inner surface of the socket, which is consistent with findings in [39, 40]. When the patient walked, the pressure exerted on the prosthetic stump was clearly evident due to the muscular activation in certain regions of the stump.

An F-socket was used to efficiently capture residual limb pressure. In the posterior and lateral regions, the peak muscular pressure was much higher. The lateral and posterior sides of the stump have the highest IP values due to their tissue composition, while the anterior and medial sides have the lowest values due to their bony nature and inability to tolerate high pressure. These findings are derived from the F-Socket test. The patient is made comfortable in this manner, preventing pressure on the proximal and distal transtibial bones. Also, there was no redness on the bony parts when the prosthesis was put on. It should be noted that the third group is capable of withstanding larger stress values and overall deformation amounts while doing FEM analysis. This gives the prosthetic socket the strength, durability, and flexibility it needs before failure. This gives the patient sufficient comfort while walking and moving from the heel contact phase to the metatarsal contact phase with the ground with complete comfort.

The equivalent (von Mises) stress, total deformation, and fatigue safety factor were calculated using FEM software on a partial prosthetic model for the patient. Von Mises theory states that the yield stress is evaluated using three criteria: $\sigma e<\sigma_Y$, safe; $\sigma_E = \sigma_Y$, critical; and $\sigma_E>\sigma y$, failed. When the fatigue safety factor is equal to or more than 1.25, it is safe in design and acceptable [19, 41]. Composite material (3 perlon + 4 carbon fibre + 3 perlon) layers have safety factors of approximately (1.483) at heel fixed and (1.593) at metatarsal fixed, while (3 perlon + 4 glass fibre + 3 perlon) layers have safety factors of approximately (1.161) at metatarsal fixed and (1.163) at heel fixed. These layers are safe and acceptable in design applications, but the (6 Perlon) layers fail. Overall, while the proposed method holds promise process to create a socket material with improved mechanical qualities for PFA, follow-up research is essential to comprehensively evaluate its potential and limitations for the long term.

Limitations

The pressure measurements were only taken during lab walking sessions, which may have caused variations in socket pressures. It is important to consider that these sensors could provide prosthetic socket pressure values during real-world activities. An important issue is the lack of documented user feedback on the prosthesis system. The absence of user input may hinder the understanding of the prosthetic system's functionality and comfort.

A complete list of abbreviations is listed in *Appendix I*.

6.Conclusion and future work

It has been shown that the numerical approach, using ANSYS software, is an effective method for evaluating safety factors, von Mises stress, and total deformation for each group. The third group has an estimated 83% increase in σ_Y, 80% increase in σ_{ULT}, and 39% increase in E compared to the first group. When carbon fibres are added to the samples, the second group's σ_Y increases by around 76%, σ_{ULT} by 75%, and E by 17%. The lifecycle of a partial foot is contingent upon the type of composite material and the value of applied stress. The lamination of three perlon, four carbon fibre, and three perlon layers has a longer endurance limit stress (σ_E) than other laminations. This increases the lifespan of patients who have partial prosthetic limbs. The patient's posterior and lateral muscles are in active state during movement, which prevents pressure from accumulating in the anterior and medial regions of the tibia. This is due to the increased pressures applied to the lateral and posterior regions, which are 236 KPa and 275 KPa, respectively. The safety factors for the (3 Perlon - 4 Carbon Fiber - 3 Perlon) layers are approximately 1.483 when the heel is fixed and 1.593 when the metatarsal is fixed, in contrast to the six Perlon layers, which fail. Similarly, the safety factors for the (3 Perlon - 4 Glass Fiber - 3 Perlon) layers are approximately 1.161 at the metatarsal and 1.163 at the heel. In summary, while the suggested approach highlights the promise of the suggested approach as a viable option in lower limb prostheses, providing both customization and cost-effectiveness, more study is necessary to thoroughly assess its long-term viability and constraints.

Acknowledgment
None.

Conflicts of interest
The authors have no conflicts of interest to declare.

Data availability
None.

Author's contribution statement
Sarah A. Hamood: Contributed to the conceptualization and writing of the original draft. **Saif M. Abbas:** Contributed to conducting experiments and investigation, analysis of experimental results, and writing reviews. **Aseel Ghazwan:** Contributed to the verification and visualisation, analysis of experimental results, and editing.

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Appendix I

S. No.	Abbreviation	Description
1	ASTM	American Society for Testing and Materials
2	E	Modulus of Elasticity
3	FEA	Finite Element Analysis
4	FEM	Finite Element Method
5	FOS	Factor-of-Safety
6	IP	Interface Pressure
7	PFP	Partial Foot Prostheses
8	PFA	Partial Foot Amputation
9	PFO	Partial Foot Orthosis
10	σ_E	Effective Stress
11	σ_{ULT}	Ultimate Tensile Stress
12	σ_Y	Yield Tensile Stress