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Design and Analysis a New Prosthetic Foot Made of Discontinuous Glass Fiber

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ABSTRACT

The present study introduces a novel design of passive prosthetic foot based on the principle of energy storage and release to enhance gait performance. Also, due to noticeable limited applications of discontinuous glass fiber in prosthetic industry, the primary objective is to investigate the fiber mechanical performance as an alternative to the typical high-cost constituents such as s-glass and carbon fibers. To achieve this goal, the key biomechanical parameters were evaluated namely: weight, cost, forward energy and foot movement in different planes. To begin, the mechanical properties of short glass fiber combined with polyester resin were experimentally determined through tensile and fatigue testing. Finite element software, SolidWorks and Ansys, were employed to simulate the developed prototype for 80 kg individual. Numerically calculated mass of the new counterpart is 0.45 kg, fulfilling the essential demand for lightweight structure. Furthermore, the current design provides an adequate level of energy storage 0.19 J/kg, satisfying the commonly accepted range 0.10-0.40 J/kg for most prosthetic feet. The calculated plantar flexion of 10.2°is consistent with the established heel flexibility limits 5°-30°. In the frontal plane, the foot safely twisted to 26.5° reflecting a proper rolling compared to the previously recorded data up to 30°. Finally, the utilization of discontinuous glass fiber reduced the cost roughly 50% compared to s-glass and 70% compared to carbon fiber. Thus, the study findings demonstrated an optimal balance of low weight, high mechanical strength and economic feasibility of discontinuous glass fiber as a reliable candidate for advanced prosthetic solutions.

1. INTRODUCTION

A lower-leg amputation is a dramatic event that inevitably leads to mobility disabilities. Prosthesis is designed to recover at least some of the lost mobility. However, enabling their users walking with a roughly natural gait style in a range of speeds on level and uneven ground still challenging tasks for these individuals [1]. Relative to healthy people, lower limb losing persons have great differences in a multi of biomechanical functions such as locomotion speed, balance and stability, gait asymmetry and metabolic consumption. Thus, research is needed to enhance the rehabilitation and quality of life for the growing amputee population [2,3].

Researchers have proposed numerous layouts for the prosthetic feet. They can be classified into three major categories: conventional, energy storing and return (ESAR) and powered feet.

Conventional prosthetic feet, like SACH foot, offers basic support and cushioning but lack dynamic performance abilities. In comparison, developed energy storage and return (ESAR) feet utilize advanced materials such as carbon fiber to improve walking efficiency by storing energy during stance and returning it during push off. Recent innovations include microprocessor controlled and actuated prosthetic feet that actively adapt to terrain and walking speed, thus improving amputee movement and minimizing metabolic cost [4]. However, the introduction of actuators and external power sources to the active prostheses make them high mass and bulkiness, these devices are heavy compared to passive devices and need more maintenance. As the prosthetic weight increase, the user feels discomfort when wearing it for a long time so it is not often the preferable solution for a considerable rate of users [5-7].

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The commercial available prosthesis are (ESAR) feet and they are the most prescribed counterpart for people with third and fourth level of mobility which are always made of flexible elements [8]. They store energy during the stance phase and release it during the swing phase to minimize the metabolic energy during walking [9].
This study explores a detailed design approach of an innovative dynamic ESAR prosthetic foot. The research gap is to investigate the mechanical performance of discontinuous E- glass fiber as a lightweight, low cost and efficient alternative to the typical costly fibers used in fabrication of prostheses members. The research shows a limited studies in its niche application area, particularly in the detailed design, fabrication, and static and dynamic testing. It contributes to progressive development in the prosthetic technology to achieve low weight products and enhance prosthetic gait using advanced hybrid materials. The design is particularly appropriate for individuals who prefer a cost effective, lightweight and mechanical passive prosthetic counterpart.

2. LITERATURE REVIEW

Prior studies showed that the superior design of structures in flight and vehicle industries through the utilization of composites instead of conventional metals due to their excellent strength to weight ratio and high specific stiffness. These specifications make composites very attractive and convenient for the production of alternative prosthesis [10]. The weight of prostheses have always been a challenge for prosthetic researchers because the main complaints from amputees remains its perceived weight [11]. Active and passive ESAR elements are usually fabricated from layers of carbon fiber [12]. The authors in [13] designed and manufactured a passive foot utilizing woven carbon fiber, wherein a unidirectional carbon fiber was used for developing a prosthetic feet in the works of papers [14-17]. Glass fibers have evolved into many engineering fields due to their low cost and fairly robust structural strength compared with aramid and carbon [18]. Based on these aspects, multi types of composites including glass fibers have been utilized in the design of light weight athletic prosthetic feet [19-21].

Design enhancements of ESAR prosthetic feet focus on optimization gait efficiency, adaptability on uneven terrain, and energy return using advanced materials and geometry. The authors in [22] presented a novel design of new prosthetic foot through optimization of geometry, energy storage, and stress distribution using nonlinear FEM methodology. Also, the paper in [23] introduced a novel, accessible method in designing a prosthetic foot for production on standard 3D printers, by making it available to any people with the requisite resources. A cost-effective modelling process was performed to reduce material waste and improve testing through the employing of dimensional analysis and scaling approaches. Recently, group of researchers characterized user preference and advantages of a novel ESR foot compared with the individuals previously worn Ossur Pro-Flex foot. For numerous activities, e.g., walking on inclines, over uneven terrain and walking up slopes are easier in the designed foot vs. their everyday feet [24]. The research paper in [25] employed polyester matrix reinforced with both natural and synthetic fibers to design and fabricate athletic foot using a vacuum-assisted system.
From previously outlined researches, carbon fiber and S-glass fiber are the most widely used materials in prosthetic foot studies and industry due to their proven mechanical performance. However, despite their benefits, the fabricated counterparts from both materials are still expensive and less accessible. Therefore, further research is needed to explore alternative affordable materials that can maintain functional performance while improving expenditure and accessibility.

The originality of this research lies in its approach to design an innovative dynamic prosthetic foot. The aim of this study is to investigate whether it is possible to incorporate discontinuous E- glass fiber as a lightweight, affordable and efficient alternative to expensive typical fibers usually used in fabrication of prostheses components. Tailoring the design to lightweight device is an emerging technology. The research shows a low degree of originality within its niche application area, particularly in the detailed design, fabrication, and dynamic testing of the material. It contributes to ongoing efforts in the prosthetic technology to reduce weight and improve performance using advanced composite materials. The design is particularly appropriate for individuals who prefer a cost effective, lightweight and mechanical passive prosthetic system.

3. METHODOLOGY

3.1 Materials

This research focuses on utilization of discontinuous glass fiber reinforcement in the design of new prosthetic foot. Chopped fibers are generally produced by cutting continuous glass filaments fiber into short lengths ranging from few millimeters to several centimeters. In recent years, chopped fibers have been widely used to enhance the mechanical properties of composite due to their low cost, easy and safe application, and simple fabrication. Chopped fiber strand mat has short strand of fibers held together with a resin binder, randomly oriented and is often used where high thickness is required. The matrix binds the fibers together, holding them aligned in the important stressed directions which serves as the load transferring medium. It protects the reinforcing filaments from mechanical damage and from Environmental attack [26]. In this work discontinues glass fiber was used in the design of prosthetic foot. The fabric is composed of randomly oriented fibers with an approximate length of 50 mm and a diameter of 13 μm. Polyester and epoxy resins combined with glass fiber create strong, lightweight composite materials, used widely in boats, automotive, aerospace, and construction for their high strength, stiffness, and corrosion resistance. Polyester used in this study is favored for lower cost and ease of use with general fiberglass mats. The effect of chopped fiber utilization with the various ratios on the mechanical properties of composites has been investigated by many researchers. Although it is observed that the fiber fractions are used between 0–1 % in the studies, it is seen that one of the most preferred ratios is 0.5 % in many composite applications [26].

3.2 The suggested new foot profile

A passive flexible prosthesis functions like a mechanical spring that stores and returns elastic energy during the stance phase of walking. Such prostheses are often referred to as energy storage and return (ESAR) prostheses.

ESAR were introduced in the 1980s and continually improved to enhance the strain energy storage .This type of prosthesis uses a foot-modeled plate (usually carbon fiber made) that stores elastic potential energy and progressively releases it as kinetic energy [27,28]. Currently, ESAR foot prosthetics are assumed to minimize metabolic energy consumption during locomotion. Additionally, the energy storage and return structures are claimed to enhance walking performance. Thus, ESAR foot prosthesis can be a priority choice due to its high functionality [29]. Based on these aspects, the proposed prototype in the present study is Ottobock size 27 foot suitable for 80 kg of weight individuals. By concern on essential qualities in foot design specifically, strength, weight and energy absorption and recover, the suggested design for keel and heel profiles were created by Solidworks software as a powerful tool to generate the accurate shape as shown in Figure 1, The foot length is 250 mm, width is 85 mm, the foot thickness is a design factor. The effective ankle joint location, height and the sections of the device which are in contact with ground have determined according to ISO standard as indicated in Figure 2.

The prototype explained in Figure 1 is generally preferred over solid ankle cushioned heel feet by people with a lower limb amputation. While ESAR feet have been shown to have only limited effect on gait economy, other functional benefits should account for this preference. A simple biomechanical model is suggested to enhance gait stability and gait symmetry. Also to increase push-off power by control the center of mass velocity at push off and enhance intact step length and step length symmetry while preserving the margin of stability during locomotion in people with a transtibial and transfemoral prosthesis. Based on Ottobock size 27, the foot length is 250 mm and the width is 85 mm. The device height is standardized to 80 mm because it mimics the natural anatomical distance between the bottom of the heel and the ankle as explained in Figure 2. Through the thickness grooves are made along heel and keel to provide an adequate level of flexibility, minimizing weight, ensure adaption to uneven surfaces, and damping the ground reaction forces on the residual limb.

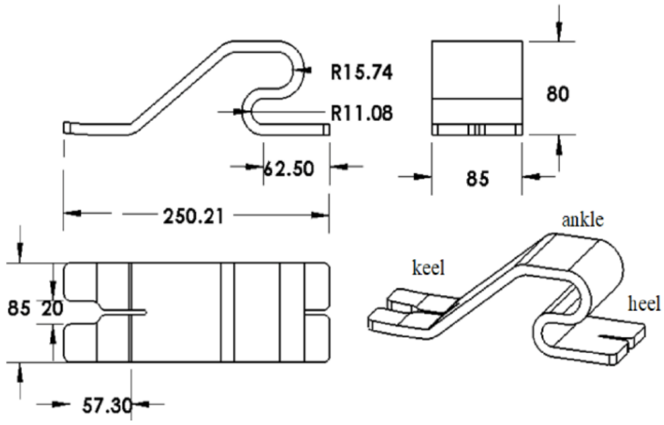


Figure 1. The proposed foot layout (dimensions in mm)

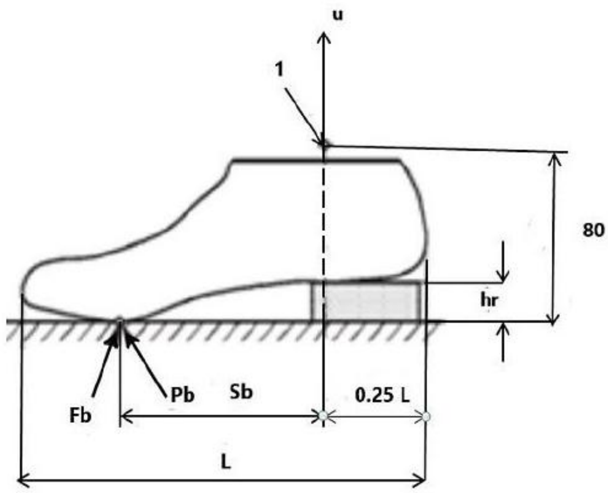


Figure 2. Effective joint center, dimensions in (mm) [30]

The foot height is 80 mm, the distance from heel tip to the ankle center is (0.25L= 62.5 mm), the distance from the ankle center to the keel point contact with ground is (0.5L =125mm).

3.3 Finite element analysis (FEA) of the gait cycle

Gait study is the examination of humans' movements patterns. In fact, walking is substantially a repetitive activity; thus, its analysis means the motion event repeated during walking. Two successive strikes on the ground of the same heel restrict one cycle [31,32. Two primary phases of the gait events are the stance phase (60% gait cycle) and the swing phase (40% gait cycle). The stance phase starts and ends with heel strike of the same leg in three modes: controlled plantar flexion, controlled dorsiflexion and push off [33,34]. Previously published studies on the utilization of FEA in prosthetic foot analysis has concerned on investigating structural properties of constituents [35]. The authors in [36] employed FEA to analysis and design a new three prototypes of ESAR foot to decide the best design. The implementation of FEA for direct design purposes was also presented in [37] where the mechanical properties and the distribution of loads and stresses on all components of the prosthetic limb were analyzed. In the current study, to evaluate the required foot thickness to bear the body weight, the CAD model was exported to ANSYS software to perform three necessary tests of gait cycle phases. The element types used in the analysis for modeling plantar flexion phase, mid stance and dorsiflexion phase is SOLID185. Elements number was refined iteratively during meshing process pending to results convergence at 19308 elements as shown in Figure 3. The mechanical properties used in this analysis were calculated experimentally. Loads and constrained degrees of freedom are explained in Figures 4. The load values were applied based on ISO 10328 dynamic test [8].

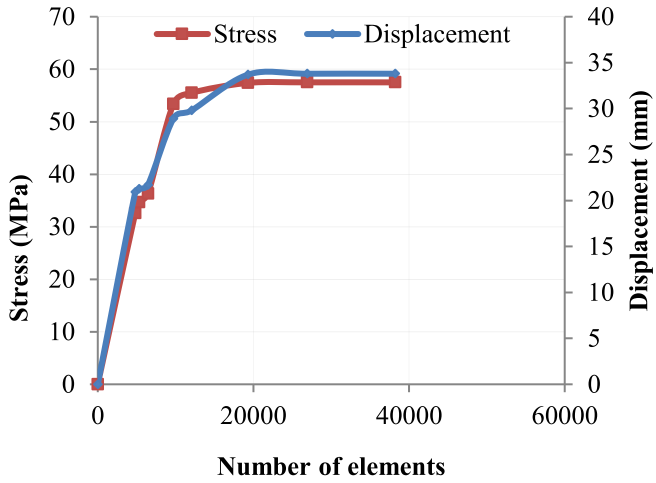


Figure 3 Results convergence at maximum load.

Thus, a load value 130 % of body weight (1020.24 N) was applied in heel strike phase, in mid stance phase the load value is 80% of body weight (627.84 N), while in fore foot phase the maximum load value is 108% of the body weight (847.584 N). The thickness was increased gradually via iterative solution of gait cycle analysis until achieving the safe foot thickness with permissible resulting stress and displacement.

3.4 Experimental work

3.4.1 Tensile Test

The effective mechanical properties of composite material were determined experimentally and then used for performing the numerical solution. The materials pictured in Figure 5 were employed to fabricate the samples of mechanical tests by means of hand lay-up technique. The materials are produced by Changzhou Tianma group in China. To gain a fine surface finish laminate and ensure perfect isolation condition, stiff nylon sheet was placed on a piece of ceramic base in a wood frame mold. A thin layer of polyester resin was spread uniformly by suitable brush with fraction of 0.5. Glass fiber ply was applied in the next stage to construct a layer of composite material. The materials fractions used in fabrication of composite plate are 50% fiber, 40% polyester resin and 10% hardener.

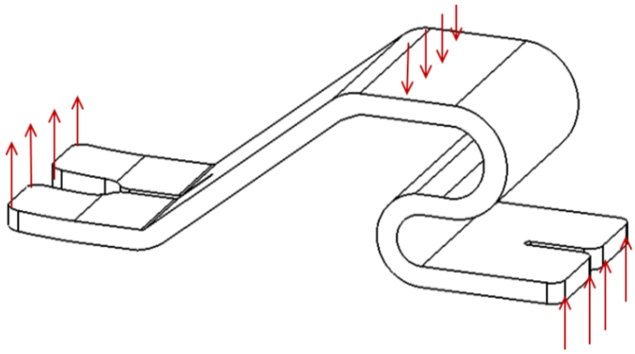


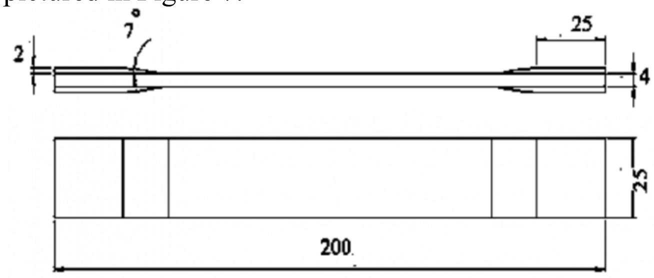
Figure 4. Distributed load was applied during three phases of gait cycle.

Figure 5. Materials used in lamination of composite plate, 1- short glass Fiber, 2- Hardener, 3-Polyster 4- Insulating wax, 5- Thinner 6- Brushes, vernier and permanent marker

To accomplish the specified tensile specimen thickness (4 mm) based on the ASTM D 3039 guideline as shown in Figure 6a [38], eight layers were laminated since the thickness of one layer of composite is approximately 0.5 mm. Again, nylon sheet and piece of ceramic was placed at the top of laminate. The composite was exposed to limited compression at room temperature for 72 hours to verify the solidification state. Figure 6b shows the fabricate samples by hand lay-up approach.
Then tensile specimens were machined by reciprocating cutter and grinding to the final shape as pictured in Figure 6b. The fabricated coupons were tested under tensile loading until reaching the failure condition and the average value were considered in determination of the mechanical properties.

3.4.2 Fatigue Test

Fatigue test is generally performed to characterize the fatigue strength limit of the material. However, in order to obtain the S-N curve, it is necessary to test at least five samples in response to each displacement or stress value that is below the static strength of the material [39]. Thus, eight fatigue samples were prepared in a suitable dimension (100 mm, 10 mm, 4mm) [23]as pictured in Figure 7.



(a)

(b)

Figure 6. Tensile samples (a) ASTM Figure D3039 standard dimensions [38], (b) Fabricated coupons.

The fabricated specimens are satisfying the specifications and requirements of the fatigue reciprocating bending machine. The specimen was fixed as a cantilever in the root and subjected to a fully reversed bending load at the free tip as indicated in Figure 7c.

The common fatigue mathematical modeling involving cyclic loading is [40]:

$$R = \frac{\sigma_{max}}{\sigma_{min}} \quad (1)$$

$$\sigma_a = \frac{\sigma_{max} - \sigma_{min}}{2} \quad (2)$$

The experimental stress was calculated as follows [41]:

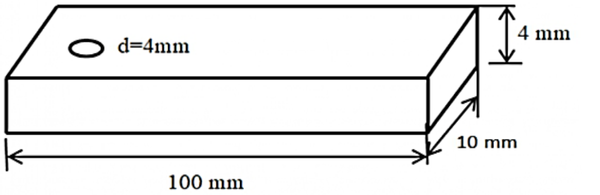
$$\sigma_a = \frac{1.5Et\delta}{L^2} \quad (3)$$

Where R is the stress ratio, σ_a is the amplitude stress in (MPa), E: Young modulus (MPa), t: is the specimen thickness(mm) and δ is the applied deflection(mm).

The test was performed using 8 specimens with R= -1 and frequency of 7.15 Hz. The experimental fatigue life can be predicted by fitting S-N curve through the Basquin's formula as follows [41]:

$$\sigma_a = aN_F{}^b \quad (4)$$

N_F is the number of cycles to failure, and a, b are material fitting parameters.



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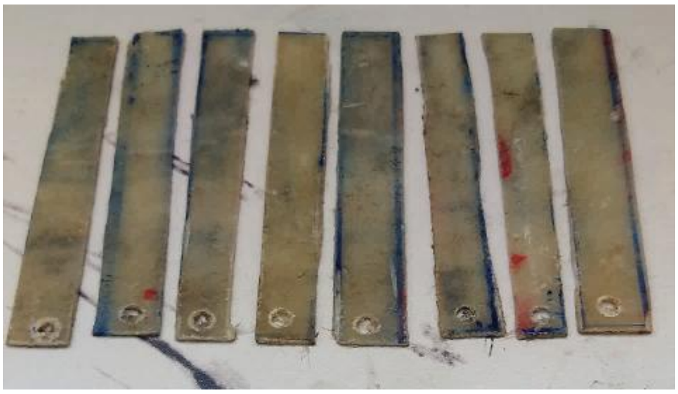


(b)

(c)

Figure 7. Fatigue specimens, (a) Standard dimensions [23], (b) Fabricated specimens, (c) Reciprocating bending fatigue machine

4. RESULTS AND DISCUSSION

4.1 Tensile test results

The material properties of composite materials were determined experimentally and used in FEM solutions. Tensile tests were carried out for three samples to determine their mechanical properties. This involved plotting a stress-strain curve for each specimen as shown in Figure 8. The test was conducted with a 5 mm/min feeding speed on an universal testing machine until the specimen fail under load. The stress-strain curve was used to determine the modulus of elasticity and the ultimate tensile strength, the results are: The Young modulus = 3878 Mpa, Density= 1523 kg/m3, ultimate tensile strength= 252.28 MPa.

4.2 Fatigue test results

Figure 9 shows the S-N curve for 8 samples of composite discontinues glass fiber and polyester resin have been tested under reversed bending load (R= -1), the load was applied through controlled displacement at the cantilever free tip.

Thus, the resulted Basquin, s formula is:

$$\sigma_a = 190.5\ N_F^{-0.136} \quad (5)$$

The Basquin, s equation is used to estimate the experimental service life of a designed component subjected to dynamic stress and used in this work for prediction of foot life.

4.3 Calculating the required foot thickness

Iterative solutions were implemented to estimate the safe foot thickness at which the induced stresses are below the ultimate strength of material. In addition, the corresponding deflection in heel part during initial strike and keel deflection during dorsiflexion phase must be sufficient for resulting a suitable plantar flexion and dorsiflexion angles to ensure a comfortable walking as summarized in Table 1. The gait phase angles were calculated based on foot configuration at maximum displacement as follows:

$$\emptyset = \tan^{-1}\frac{y_{\max}}{L_{\text{effective}}} \quad (6)$$

Where in heel loading, ymax represents the maximum heel displacement (mm), $L_{effective}$ represents the distance from heel tip to the to the effective center of ankle joint (mm). Similarly, in the case of keel loading, y_{max} represents the maximum keel displacement (mm), $L_{effective}$ represents the distance from keel tip to the effective center of ankle joint (mm).

Table 1 clearly shows that the generated maximum Stress is 115 MPa, this value is extremely below the ultimate tensile stress 252.28 MPa through the entire modes of locomotion for all values of foot thickness. However, other important factors in prosthetic foot design are the plantar flexion and dorsiflexion angles. These values are considered to be 10.3°and 10.2° for foot thickness t=9 mm in Table 1.

To enhance the reliability of the selected design thickness(t=9mm), the current design reached a factor of safety 2.9 at heel region and 1.6 at the keel region also the default maximum fatigue life 10^6 as clear in Figure 10. Actually, the generated cyclic stress on the foot is vary from zero to maximum. Therefore, by using equation (2) the amplitude stress will be half of the maximum stress listed in the last row of Table 1.

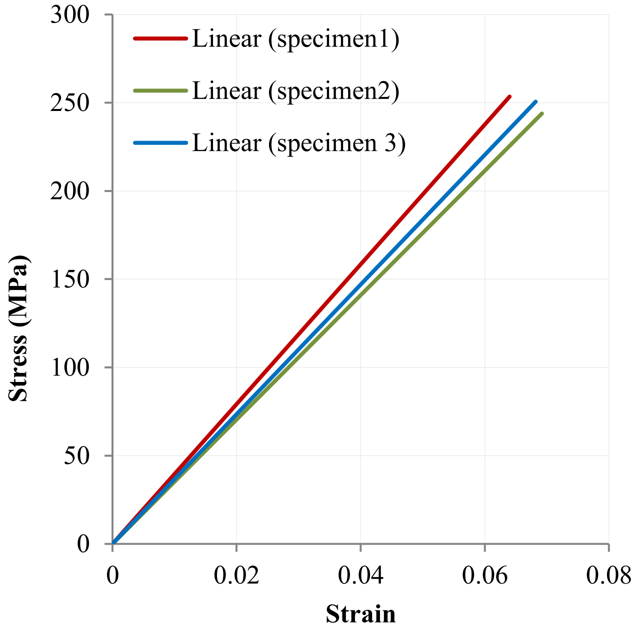


Figure 8. Tensile test of three samples

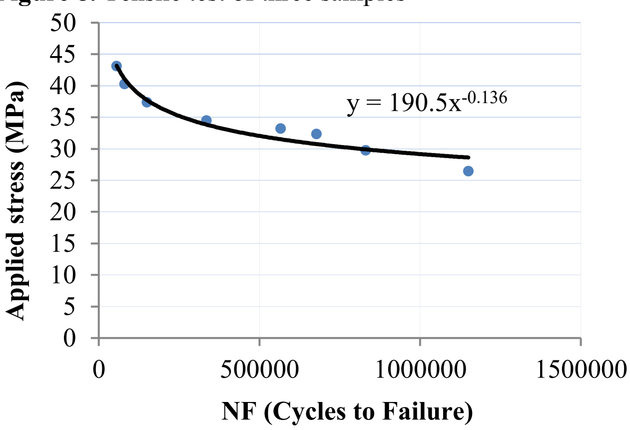


Figure 9. Experimental S-N fatigue curve

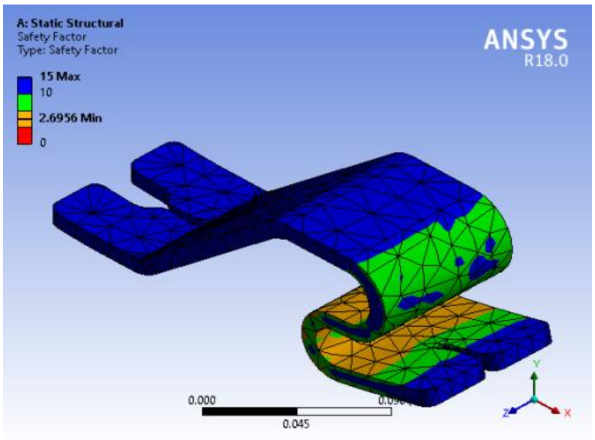
Table 1. The resulted stresses and displacements with respect to foot thickness

	Heel		Mid stance		Keel		Total mass (kg)	Plantar flexion angle (°)	Dorsi flexion angle (°)
Thick. (mm)	peak stress (MPa)	peak disp. (mm)	Peak stress (MPa)	peak disp. (mm)	peak stress (MPa)	peak disp. (mm)			
5	.838	39.6	11.2	1.8	115	103.86	0.26	23.3	28.9
6	61.2	23.9	10.9	1.19	80.7	66.06	0.30	20.9	19.04
7	43	16	7.8	0.82	71.5	51.61	0.35	14.3	15.3
8	40	13.6	6.4	0.72	63.9	42.12	0.40	12.2	12.6
9	34.1	11.4	5.7	0.52	57.5	33.8	0.45	10.3	10.2

Applying equation (5), the corresponding experimental fatigue life is (819238.27 cycle) in good agreement with the simulation result. A fatigue factor of safety greater than or equal to 1.25 indicates safe design [23], moreover, the authors in [42] designed a prosthetic foot with 1.5 factor of safety. Therefore, the foot thickness was selected as 9 mm with factor of safety 1.6 correspond to low resulting stresses and desirable foot bending in plantar and dorsiflexion periods.

4.4 Stress analysis results of gait cycle at designed foot thickness t=9mm

Figure 11a shows the stress distribution generated in the foot at initial strike and Figure 11b shows the corresponding contour plot for displacement at maximum load. Stress analysis indicated that the maximum stress (34.1 MPa) generated at the heel root as clear in Figure 11a, this value remains below the ultimate tensile strength of materials. Also, from Figure 11b, the maximum displacement at the heel tip is 11.4 mm, this value is sufficient to achieve a desirable plantar flexion 10.3° as mentioned in Table 1. The plantar flexion angle is relevant to a comfortable walking by minimizing the effect of ground reaction force on the residual limb and also minimize the fracture opportunity for the whole prosthetic leg components



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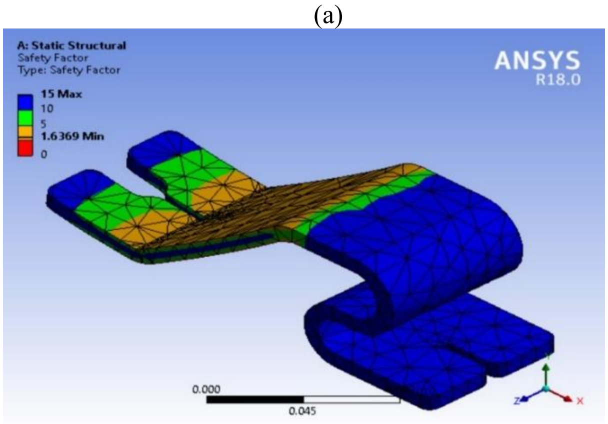

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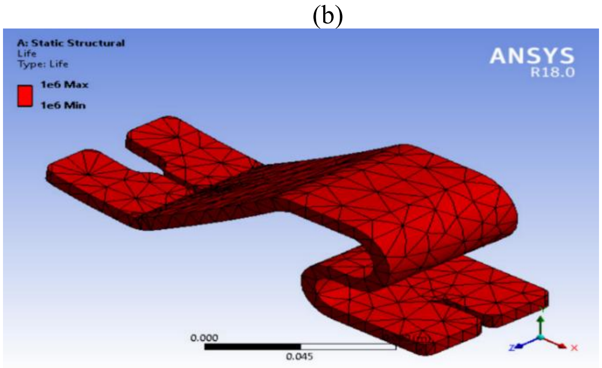
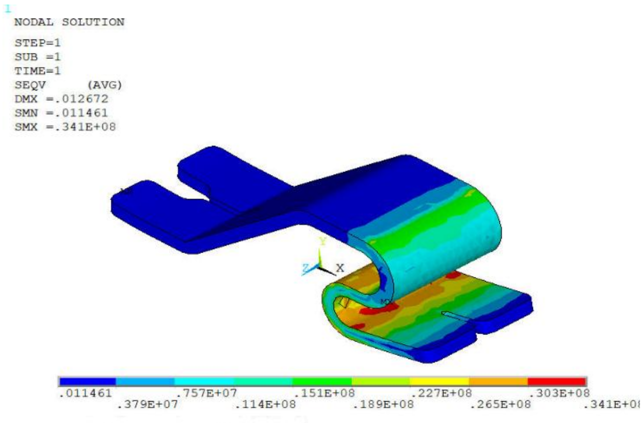


(c)

Figure 10. Fatigue results, (a) Factor of safety in heel strike phase, (b) Factor of safety of keel in dorsiflexion phase, (c) Fatigue life



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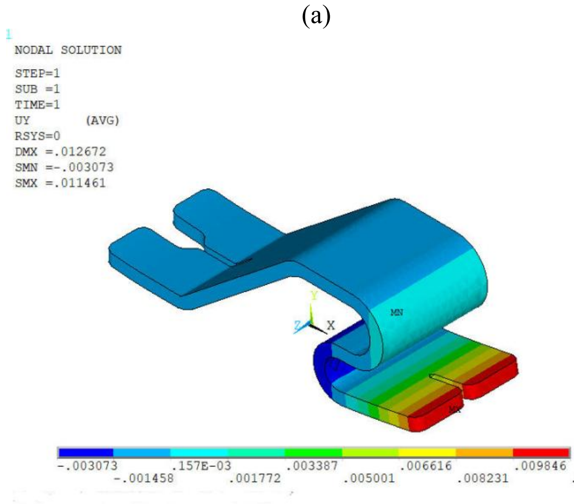


(b)

Figure 11. Stress analysis of heel at maximum load, (a) Stress distribution (Pa), (b) Displacement distribution (m)

The mid stance phase shows a minimum resulting stress and displacement in contrast to other modes. In this case, the foot loaded at ankle region and totally constrained at the contact surfaces with ground.

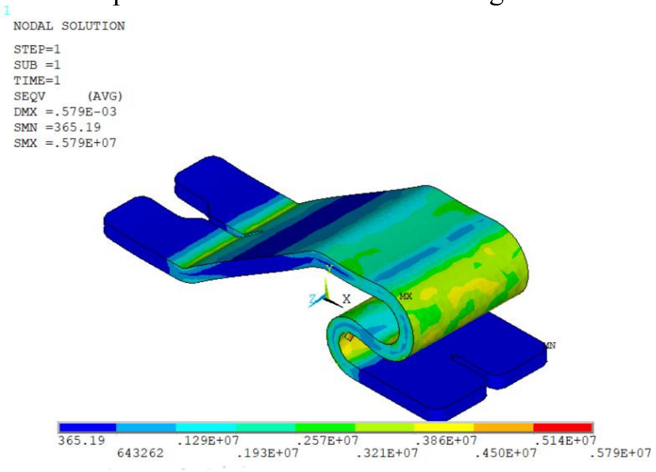
The maximum stress value is 5.7 MPa as obvious in Figure 12a, the corresponding maximum vertical displacement is 0.52 mm as explained by the contour plot in Figure 12b.

The highest values for the induced stresses and displacements were observed in dorsiflexion phase compared to the other two phases. In this mode analysis, the load was applied at the keel tip and the foot is completely fixed at ankle region. The value of peak stress was 57.5 MPa as explained in Figure 13a whereas Figure 13b indicates the maximum displacement during keel loading which is 33.8 mm.

The resulting displacement value is suitable that means the keel will be bent at 10.2° in dorsiflexion phase.

4.5 Estimation of eversion and inversion angles

Inversion and eversion primarily occur at the subtalar joint. Inversion is the inward turning of the sole of the foot and the eversion is outward turning in the frontal plane [43]. In the present design the eversion and inversion have been simulated in the standing phase considering half of the body weight was applied on the prosthetic leg. The load was subjected at inner side assuming right foot, the outer side and ankle regions were constrained for all degrees of freedom. Figure 14 shows the resulting displacement due to eversion load; at this displacement value the eversion angle is 2.5°.



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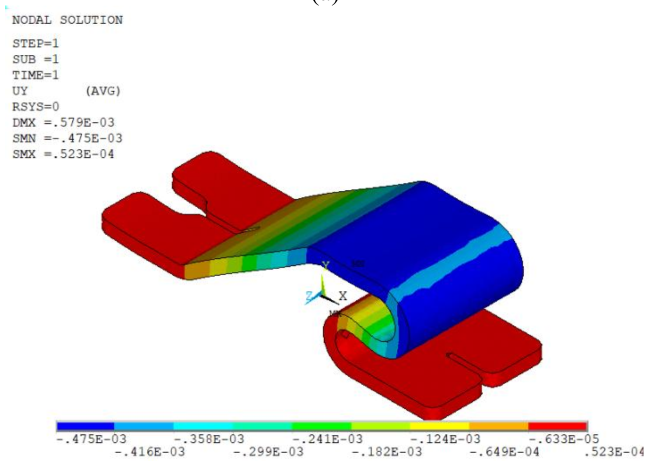
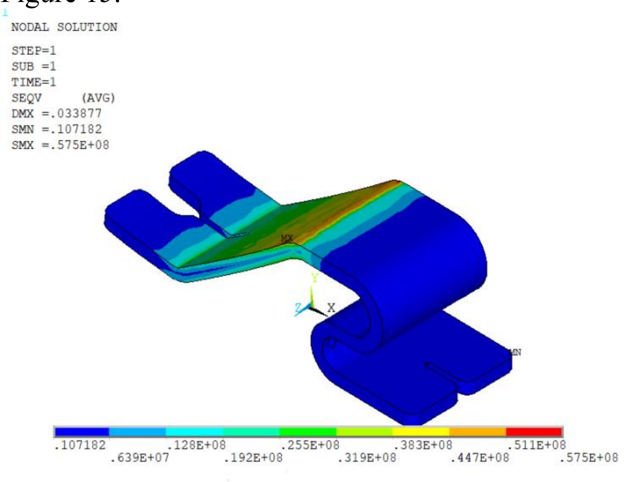


(b)

Figure 12. Mid stance analysis at maximum load**,** (a) Stress distribution (Pa), (b) Displacement distribution (m)

The inversion rolling is dependent of the distance between the two legs during standing, thus the inversion analysis was simulated for three arbitrary angles (θ= 10°, 20° and 30°) between the body centerline and the prosthetic leg centerline as explained schematically in Figure 15.



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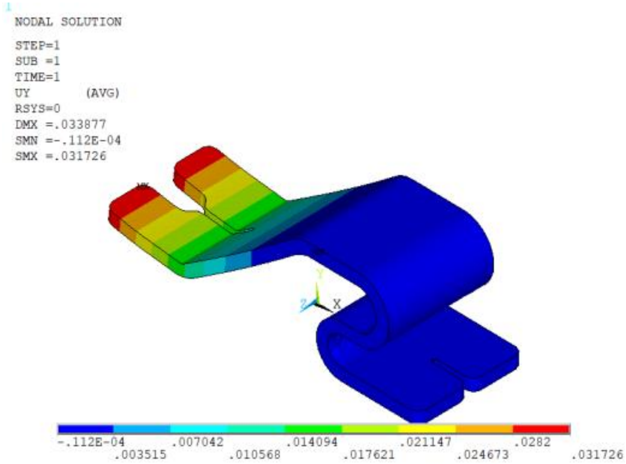

(b)

Figure 13. Keel analysis at maximum load**,** (a) Stress distribution (Pa), (b) Displacement distribution (m)

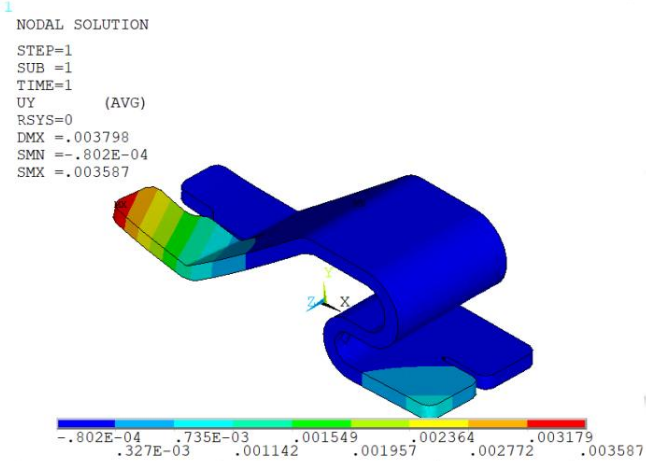

Figure 14. Eversion simulation; displacement contour (m)

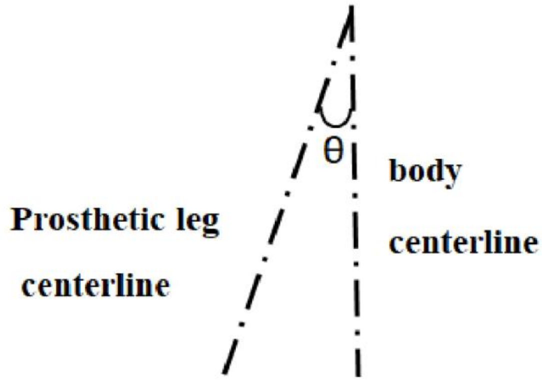

Figure 15. Representation of inversion analysis in standing case

In FEA analysis, CONTA 174 and TARGET 170 elements were utilized to simulate the surface-to-surface contact problem between the prosthetic foot and the ground. The load was applied at the ankle region and the nodes in the left side of the foot, that are in contact with ground, have fixed for all degrees of freedom. The analysis result is explained in Figure 16 for the maximum displacements and the whole results are summarized in Table 2. The specifications of fully designed model are illustrated in Table 3.

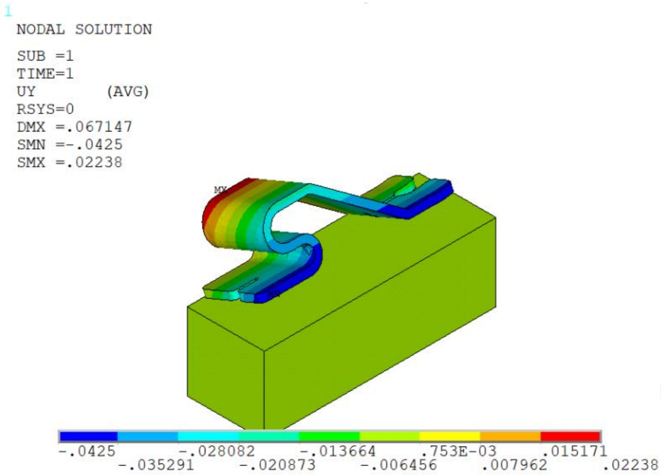

Figure 16. Inversion analysis for θ=30°, Inversion displacement(m)

Table 2. Inversion angle with respect to the angle (θ)

S. NO.	θ(°)	Inversion angle(°)
1	10	9.85
2	20	18.83
3	30	26.5

Table 3: Specifications of fully designed model

Length	250 mm
width	85 mm
Thickness	9 mm
Height	80 mm
mass	0.45 kg
Maximum plantar flexion angle	10.3°
Maximum dorsiflexion angle	10.2°
Eversion angle in standing case	2.5°
Inversion angle in standing case θ=30°	26.5°

4.6 Determination the amount of the energy storage

The benefit of the ESAR foot prosthesis is the capability to store energy. In the field of prosthetics, energy can be quantified using the formula of work (W). In physics, work is defined as the force applied to an object causing it to undergo displacement. The concept of work is inherently linked to the concept of energy. Mathematically, work is expressed as the product of force and displacement and formulated as [29].

$$W = \int f \, ds \tag{7}$$

The relation between the applied load and the resulting displacement in plantar flexion and dorsiflexion periods is shown in Figure 17. The amount of energy stored in the heel can be determined by considering the area under the load – displacement curve. Thus, the energy stored in the heel during plantar flexion mode is 5.53 J. Likewise, the energy stored in keel is 15.9 J. Finally, the whole results of the current study were compared with various prior studies as indicated in Table 4.

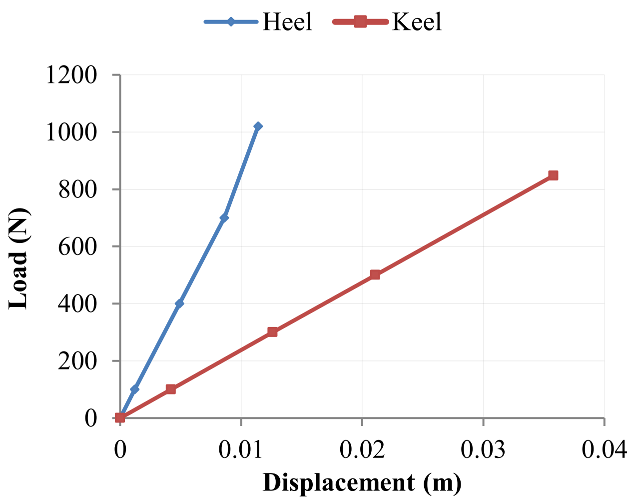

Figure17. Load-displacement curves for heel and keel loading

Table 4: Comparison of current results with literatures

Reference	Dorsiflexion angle (°)	Plantar flexion angle(°)	Inversion angle (°)	Eversion angle (°)	Mass(kg)	Energy stored in dorsiflexion phase(J/kg)	Energy stored in plantar flexion phase(J/kg)
[6] (passive)	7	-	-	-	-	-	-
[8] (passive)	15	6.4	-	-	-	-	-
[29] (passive)	-	-	-	-	-	0.17	0.15
[31] (passive)	-	-	-	-	-	-	0.044
[35] (passive)	-	-	-	-	-	0.27	0.071
[44] (active)	3.62	3.51	3.14	8.69	-	-	-
[45] (active)	15	7	-	-	1.32	-	-
[46] (passive)	14.54	6.7	5.8	4.5	-	0.16	-
present	10.2	10.3	26.5	2.5	0.45	0.19	0.069

5. CONCLUSIONS

This work explores the design and structural analysis of a novel prosthetic foot through the integration of discontinuous glass fiber with polyester resin. The main objective was to investigate the opportunity of involving a low cost random fiber in prosthetic technology via evaluating the biomechanical design parameters. The study presented step by step design methodology of a new prototype through combining numerical and experimental tests. The primary findings of this research comprise the following:

- Required foot thickness is 9 mm (18 layers) of discontinues fiber and polyester resin has standard length and width is sufficient for 80 kg amputee. The design is compact with reduced weight and size. The mass of the new alternative is 0.45 kg. Weight reduction lowers the metabolic energy required for ambulation and enhances gait efficiency.
- The stresses generated in the new design are relatively low in comparison to the excellent mechanical strength of composite materials listed in Tables 1 and 2. Thus, the main finding that the current model is capable of withstanding substantial loads during static and dynamic phases of gait.
- The new model exhibits superior fatigue behavior, enduring over one million cycles with 1.6 fatigue factor of safety. This outcome indicates the ability to absorb and redistribute dynamic load without rapid crack initiation or propagation leads to improved service life and reduced maintenance requirements.
- The results listed in Table 3 show the acceptable adaptability of the new component in terms of plantar and dorsiflexion in sagittal plane as well as inversion - eversion in frontal plane. The proposed device can provide stable and an efficient performance on uneven, sloped or irregular surfaces.
- The cost of the new designed foot composed of reinforcing discontinuous glass fiber with polyester resin is approximately 9000 ID exactly by hand layup technique. In contrast, the cost may be doubled if S -glass is used, four times higher if carbon fiber is used, and up to seven times higher if aramid fiber is used.
- The substantial feature of woven and unidirectional fibrous materials is the ability of tailoring the material to a specific fiber direction to improve stiffness and strength, such a feature is not common in randomly orientated fibers which provide approximately isotropic mechanical properties. This is a main drawback of discontinuous fibers compared to direction dependent fibers.

Finally, the new prosthetic foot design addresses several key performance criteria essential for amputees' satisfaction and functionality. The findings support further development of cost-effective, lightweight, compact sizing and high-performance composite based prosthetic systems for individual user needs. Recommendation for future work is to study the effect of combining hybrid composite materials such as discontinuous fiber and unidirectional fiber or natural fiber in the design of prosthetic foot.

REFERENCE

[1] M. Ernst, B. Altenburg, T. Schmalz, A. Kannenberg, M. Bellmann, "Benefits of a microprocessor controlled prosthetic foot for ascending and descending slopes ", Journal of Neuro Engineering and Rehabilitation, vol. 19, no.1, 2022. https://doi.org/10.1186/s12984-022-00983-y

[2] A. Berettoni, S. D. Giuseppe, G. Mariani, N. Boccardo, M. Laffranchi, M. Semprini,, "Exploration of biomechanics, mental workload and metabolism factors during prosthetic gait", Progress in Biomedical Engineering **7** (2025) 043002. https://doi.org/10.1088/2516-1091/ae0f11

[3] C. E. Bosman, C. K. van der Sluis, J. H. B. Geertzen, N. Kerver, A. H. Vrieling," User-relevant factors influencing the prosthesis use of persons with a transfemoral amputation or knee-disarticulation: A meta synthesis of qualitative literature and focus group results", PLOS One, vol. 18, no. 1, 2023. https://doi.org/10.1371/journal.pone.0276874

[4] D. R. Thakre, M. Behera, P. V. Patole, "Low-Cost Energy Storing and Return Prosthetic Foot With Toe Break", International Journal of Novel Research and Development, Vol. 10, Issue 5, 2025, https://ijnrd.org/viewpaperforall.php?paper=IJNRD2505483

[5] S. R. Ahamed, S.Saha, A. Bhushan, "Lower-limb prosthetic mechanisms: recent progress, control innovations and barriersto real-world adoption", BioMedical Engineering OnLine,(2026) 25:76, https://doi.org/10.1186/s12938-026-01558-x

[6] S. K. Mohammed, A. H. Mohammed, " Design and Manufacturing of a New Prosthetic Foot", Journal port Science Research, vol. 4, no. 2, pp.118-124, 2021. https://doi.org/10.36371/port.2021.2.8

[7] G. Marconi, A. A. Gopalai, S. Chauhan, " Effects of powered ankle–foot orthoses mass distribution on lower limb muscle forces—a simulation study", Medical & Biological Engineering & Computing, vol. 61, pp. 1167–1182, 2023. https://doi.org/10.1007/s11517-023-02778-2

[8] J. Tabucol ,T.M. Brugo, M. Povolo, M. Leopaldi, M. Oddsson, R. Carloni, A. Zucchelli, "Structural FEA-Based Design and Functionality Verification Methodology of Energy-Storing-and-Releasing Prosthetic Feet", Appl. Sci., vol. 12, no.1, 2022. https://doi.org/10.3390/app12010097

[9] H. Kamel, O. Harraz, K. Azab, T. Attia, "Developing an Optimized Low-Cost Transtibial Energy Storage and Release Prosthetic Foot Using Three-Dimensional Printing", ASME J of Medical Diagnostics, vol. 3, no. 2, 2020. https://doi.org/10.1115/1.4046319

[10] D. Hern´andez-Lara, R.G. Rodríguez-Ca˜nizo, E.A. Merch´an-Cruz, ´A.M. Santiago-Miguel, E.T. Ju´arez-Vel´azquez, C.A. Trejo-Villanueva, "Optimal design of a foot prosthesis insole with composite materials applying metaheuristic algorithms", Results in Engineering, vol. 13, 2022, 100322. https://doi.org/10.1016/j.rineng.2021.100322

[11] G. Preatoni, G. Valle, F. M. Petrini, S. Raspopovic, "Lightening the Perceived Prosthesis Weight with Neural Embodiment Promoted by Sensory Feedback", Current Biology, vol. 31, no. 5, pp. 1065-1071, 2021. https://doi.org/10.1016/j.cub.2020.11.069.

[12] A. N. Al Thahabi, L. M. Martulli, A. Sorrentino, M. Lavorgna, E. Gruppioni, A. Bernasconi, "Stiffness-driven design and optimization of a 3D-printed composite prosthetic foot: A beam finite Element-Based framework", Composite Structures, vol. 337, 2024. https://doi.org/10.1016/j.compstruct.2024.118053.

[13] M. I. Hameed, A. A. Ali, "Finite Element Design and Manufacturing of a Woven Carbon Fiber Prosthetic Foot", Association of Arab Universities Journal of Engineering Sciences, vol. 29, no. 2, pp. 09–18, 2022. https://doi.org/10.33261/jaaru.2022.29.2.002

[14] B. Sehar, A. Waris, S.O. Gilani, U. Ansari, S. Mushtaq, N.B. Khan, M. Jameel, M.I. Khan, O. T. Bafakeeh E. M. Tag-ElDin, "The Impact of Laminations on the Mechanical Strength of Carbon-Fiber Composites for Prosthetic Foot Fabrication", crystals, vol. 12, no. 10, 2022. https://doi.org/10.3390/cryst12101429.

[15] R. Elgamsy, M. I. Awad, N. Ramadan, A. Amer, Y. Osama, R. El-hilaly, A. Elsabbagh, "Localization of composite

prosthetic feet: manufacturing processes and production guidelines", Scientific Reports, vol. 13, no.1, 2023, 17421. https://doi.org/10.1038/s41598-023-44008-7

[16] M. Falbriard, G. Huot, M. Janier, R. Chandran, M. Rechsteiner, V. Michaud, J. Cugnoni, J. Botsis, K. Scho¨ nenberger, K. Aminian," A functional approach towards the design, development, and test of an affordable dynamic prosthetic foot", PLOS One, vol. 17, no. 5, 2022. https://doi.org/10.1371/journal.pone.0266656

[17] B. Li, G. Xu, Z. Teng, D. Luo, J. Pei1, R. Chen, S. Zhang,"Intelligent ankle–foot prosthesis based on human structure and motion bionics", Journal of NeuroEngineering and Rehabilitation, vol. 21, no. 1, 2024.http://dx.doi.org/10.1186/s12984-024-01414-w.

[18] H. Jariwala, P. Jain, and V. Maisuriya, " Experimental and statistical analysis of strength of glass fiber reinforced polymer composite for different fiber architecture", Polymer Composites, vol. 42, no. 2, pp. 1407-1419, 2020. http://dx.doi.org/10.1002/pc.25911.

[19] H. K. Talla, J. K. Oleiwi, A. F. Hassan, "Performance of Athletic Prosthetic Feet Made of Various Composite Materials with PMMA Matrix: Numerical And Theoretical Study", Revue des Composites et des Matériaux Avancés, vol. 31, no. 4, pp. 257-264, 2021. https://doi.org/10.18280/rcma.310410

[20] N. K. Jamagh, M.J. Jweeg, A. K. Hassan," Testing of suggested laminated composite materials for the manufacturing of an athletic prosthetic foot", AIP Conf. Proc., vol. 3292, Issue 1, 8May (2025). https://doi.org/10.1063/5.0267490

[21] Y. Akkaş and S. Alsancak," Lightweight design of running prosthetic feet using honeycomb sandwich composites: a comparative FEA study", Frontiers in Bioengineering and Biotechnology, vol. 14, 2026. https://doi.org/10.3389/fbioe.2026.1785315

[22] Th. M. Balaramakrishhan, S.Natarajan, S. Sujatha, "Energy storage and stress–strain characteristics of a prosthetic foot: a priori design and analysis with experiments" International Journal for Numerical Methods in Biomedical Engineering, Vol. 38, Issue 4, 2022. https://doi.org/10.1002/cnm.3579

[23] . A. M. Gómez-Amador, C. Pérez-Carrera , L. P. Fernández, H. Rubio-Alonso, 3D-printed prosthetic foot design: Mechanical similarity and testing, Materials & Design 253 (2025) 113918. https://doi.org/10.1016/j.matdes.2025.113918

[24] . M. Pawel et al., User preference and patient benefits of a novel energy storing and return foot: A randomized, cross-over clinical trial, Prosthetics and Orthotics International 49(6):p 645-653, December 2025. http://dx.doi.org/10.1097/PXR.0000000000000415.

[25] N. K. Faheed, R. A. Issa, Q. A. Hamad, M. J. Jaafar, M. S. Mahmood, "Design and evaluation of a novel bioinspired prosthetic foot for running applications in lower limb amputees", Biomaterials Translational, 2025 https://doi.org/10.12336/bmt.25.00016

[26] C. K. Anyiam and G. C. Onuegbu, "Mechanical Properties of Chopped Glass Fibre Reinforced Polyester", Journal of Materials Science Research and Reviews, Vol.9, Issue 2, PP 342-350, 2026. https://doi.org/10.9734/jmsrr/2026/v9i2480

[27] Z. A. Colvin, J. R. Montgomery, and A. M. Grabowski, "Effects of powered versus passive-elastic ankle foot prostheses on leg muscle activity during level, uphill and downhill walking", Royal Society Open Science, vol. 9, no. 12, 2022. https://doi.org/10.1098/rsos.220651

[28] J. Z, López, L. A. Z. Avilés, A. H. V. González, J. C. Á. Vilchis, "Foot/Ankle Prostheses Design Approach Based on Scientometric and Patentometric Analyses", applied sciences, vol. 11, no. 12, 2021. https://doi.org/10.3390/app11125591.

[29] A.F. Istiqomah, R. Ismail, S. Sulistyo, D. F. Fitriyana, D. Setyawan, H. Prawibowo, A.P. Fardinansyah, M.A. N. Safitri, J. Jayatin, R.B. Taqriban, "Manufacture of Carbon Fibre-Based Energy Storage and Return (ESAR) Foot Prosthesis Using Vacuum Bagging Curing Method", Proceedings of the 2nd Lawang Sewu Internasional Symposium on Engineering and Applied Sciences, LEWIS-EAS, 2023. https://doi.org/10.2991/978-94-6463-480-8_3

[30] International standard ISO 10328, "Prosthetics- Structural testing of lower –limb prostheses –Requirements and test methods", 2nd edit. 2016.

[31] M.H. Ahmed, A. Jamshid, U. Amjad, A. Azhar, M.Z. ul Hassan, M.I.Tiwana, W.Sh. Qureshi, E. Alanazi, "3D Printable Thermoplastic Polyurethane Energy Efficient Passive Foot", 3D printing and additive manufacturing, vol. 9, no. 6, pp. 557- 565, 2022. https://doi.org/10.1089/3dp.2021.0022

[32] R. D. Gregorio, L. Vocenas, "Identification of Gait-Cycle Phases for Prosthesis Control", Biomimetics, vol. 6, no. 2, 2021. https://doi.org/10.3390/biomimetics6020022

[33] X. Sun, F. Sugai, K. Okada, M. Inaba, "Design, Control and Preliminary Test of Robotic Ankle Prosthesis", International Conference on Intelligent Robots and Systems, Madrid, Spain, October 1-5, pp. 2787- 2793, 2018. https://doi.org/10.1109/IROS.2018.8594498.

[34] V. Raveendranathan, V. G. M. Kooiman, R. Carloni, "Musculoskeletal Model of an Osseointegrated Transfemoral Amputee in OpenSim", PLOS One, vol. 18, no. 9, 2023. https://doi.org/10.1371/journal.pone.0288864

[35] H. Tryggvason, F. Starker, C. Lecomte, F. Jonsdottir, "Use of Dynamic FEA for Design Modification and Energy Analysis of a Variable Stiffness Prosthetic Foot," applied sciences, vol. 10, no. 2, 2020. https://doi.org/10.3390/app10020650

[36] A.F. Istiqomah, R. Ismail, D. F. Fitriyana, Sulistyo, A. P. Fardinansyah, M. A. N. Safitri, J. P. Siregar, "Design and Analysis of The Energy Storage and Return (ESAR) Foot Prosthesis Using Finite Element Method," Journal of Biomedical Science and Bioengineering, vol. 1, no. 2, pp. 59-64, 2021. http://dx.doi.org/10.14710/jbiomes.2021.v1i2.59-64

[37] S. M. Ali, S. S. Mahmood, "Design and Manufacturing of a Low-Cost Prosthetic Foot", Ingeniería e Investigación, vol. 43, no. 3, 2023. http://doi.org/10.15446/ing.investig.98916

[38] K. Hamza, A.Y. Shash, M. Abdrabou, "Application of Short Fibers Reinforced Composites in Power Transmission Coupling", Latin American Journal of Solids and Structures, vol. 14, no. 10, pp.1789-1803, 2017 http://dx.doi.org/10.1590/1679-78253414

[39] H. E. Balcıoğlu, R. Sakin, H. GÜN, "Development of Fixed End Type Flexural Fatigue Test Machine and Static and Dynamic Behaviour of Glass/Epoxy Laminated Composite" , IV. International Ege Composite Materials Symposium KOMPEGE, 6-8 September 2018. https://doi.org/10.17515/resm2018.67me0917

[40] P. Alama, D. Mamalis, C. Robert, C. Floreani, C. M. O Bradaigh, The fatigue of carbon fiber reinforced plastics - a review, Composites Part B: Engineering, vol. 166, pp. 555-579, 2019. https://doi.org/10.1016/j.compositesb.2019.02.016

[41] S. Ghafel, N. S. M. Namer , A. H. Ali, "Study the Mechanical Properties and Fatigue Effect of Multilayer Woven E-Glass/Epoxy Composite under Constant and Variable Loading", Tikrit Journal of Engineering Sciences, vol. 30, no.10, pp. 21-30, 2023. http://doi.org/10.25130/tjes.30.2.3

[42] B. S.Vinary, T. Thinlay, S. K. Jayswal, S. Pradeep, M. Bais, K.D. Prasad, J. I. P.Singh, "Design and structural analysis of a passive ankle-foot prosthesis with manually adjustable

stiffness and having two degrees of freedom", Materials Today, vol. 65, no. 8, pp. 3496-3505, 2022. https://doi.org/10.1016/j.matpr.2022.06.086

[43] H. Masuma, S. Bhaumika, R. Rayb," Conceptual Design of a Powered Ankle-Foot Prosthesis for Walking with Inversion and Eversion", Procedia Technology, vol. 14, pp. 228 – 235, 2014. http://dx.doi.org/10.1016/j.protcy.2014.08.030

[44] E. M. Ficanha, G. A. Ribeiro, H. Dallali, M. Rastgaar, "Design and Preliminary evaluation of a Two DOFs cable-Driven ankle–Foot Prosthesis with active Dorsiflexion–Plantarflexion and inversion–eversion", Frontiers in Bioengineering and Biotechnology, vol. 4, 2016. https://doi.org/10.3389/fbioe.2016.00036.

[45] L. Gabert, S. Hood, M. Tran, M. Cempini, T. Lenzi, "A Compact, Lightweight Robotic Ankle-Foot Prosthesis : Featuring a Powered Polycentric Design", IEEE ROBOTICS & AUTOMATION MAGAZINE, vol. 27, no. 1, pp. 87-102, 2020https://doi.org/10.1109/MRA.2019.2955740.

[46] J. Tabucol, V. G. M. Kooiman, M. Leopaldi, T. M. Brugo, R.A. Leijendekkers, G.Tagliabue, V. Raveendranathan, E. Sotgiu, P. Benincasa , M. Oddsson, N. Verdonschot, R. Carloni, A. Zucchelli, "The Functionality Verification through Pilot Human Subject Testing of My Flex-d: An ESR Foot Prosthesis with Spherical Ankle Joint," applied sciences, vol. 12, no. 9, 2022. https://doi.org/10.3390/app12094575