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A Novel Design of the Articulated Lower Limb Prosthetic Foot Using Fiber-Reinforced Polymer

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Abstract

The prosthetic foot (PF) is a device used to make the lower limb for the amputee patients perform approximately like the normal leg, while the articulated Ankle-Foot orthosis (AFO) is a brace for patients with severe drop foot. The articulated prosthetic lower limb prefers to add several functions to patients rather than a non-articulated prosthetic like a higher degree of freedom, comfort, and easier maintenance. A novel Articulated Ankle-Foot Prosthetic (AAFP) is designed, manufactured, computerized, and numerically analyzed by "LINKAGE PROGRAM" that is connected to the second program "KINOVEA PROGRAM" in order to investigate all the main characteristics items of the prosthetic foot. The AAFP is based on mixing between prosthetic foot (PF) and the articulated ankle-foot orthoses (AAFO), it consists of several links and one slider, all made from Carbon Fiber-Reinforced Polymer CFRP. The novel prosthetic foot is experimentally studied in terms of dorsiflexion angle, weight, sawing time, and stance time. The AAFP shows an excellent dorsiflexion angle and lighter weight in comparison with commercial feet. The dorsiflexion angle is 8o with a good increment reached to 2 % and 20 % in comparison to non-articulated foot and SACH foot respectively. A self-mechanically return mechanism foot, lighter weight, and a simpler design are obtained. It has a lighter weight of 39 % and 266 % of the total prosthetic foot weight in the comparison with nonarticulated and SACH feet respectively. Finally, the AAFP design is closely approached to the normal gait of a healthy person, by an approximately identical swing and stance phases time with a difference of no more than 2 % and 4.66 % respectively. The numerical analysis is based on the finite element method using AUTODESK INVENTOR PROGRAM. The numerical results showed that the induced stresses and strains in the prosthetic foot give a similar factor of safety as steel but with a lighter weight when the present novel design is made from Carbon Fiber-Reinforced Polymer (CFRP). Higher deformation is obtained from using CFRP (PP: 5, CF: 1 by weight) in comparison with steel, so a more elastic and damping response is offered against ground reaction forces.

Keywords: Ankle- Foot prosthetic, articulated foot, CFRP, dorsiflexion angle, swing time, Autodesk Inventor.

Nomenculture

A : Cross-Sectional area of the link (m2).
CF : Carbon Fiber.
CFRP : Carbon Fiber-Reinforced Polymer.
E : Modulus of Elasticity of Link Material (N/m2).
HDPE : High density polyethelene.
K : Spring Constant (N/m).
KE : Kinetic Energy (Joule).
L : Link Length (m).
m : Mass (kg).
PE : Potential Energy (Joule).
PP: Polypropylene.
U : Total Energy (Joule).
v : Speed of the foot mechanism (m/sec).
W : Frictional force (N).

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1. Introduction

The great support provided by both biomechanics and biomaterials for the disabled in the lower and upper extremities was done by orthoses and prostheses (O&P) assistants.

The artificial devices which are substituted for missing body parts or supported (due to accident or diseases) are aimed to support, stabilize, and return motion. Orthoses (braces) hold up and adjust the structural and utility characteristics of disabled neuromuscular and musculoskeletal systems. For disabled with impairments that subscribe to practical activity limitations, orthoses are used to direct forces on the limb for biomechanical requirements[1].

The prosthesis is an artificial substitution for a missing limb, the remaining part of the amputee is devoted as the residual limb. The mechanical loads are passed from the residual limb to the prosthesis by the socket. The socket is a cup-like structure that dresses appropriately for the residual limb.

The design and the cost of these prosthetic and orthotic devices represent the important elements in the development of the biomechanical industry, guiding an excess in the quality of life of amputee patients.

In fact, the choice of artificial or as called adaptive foot for amputee patients is yet complicated. To guarantee effective rehabilitation, the clinical examination of the amputee patients is necessary by the prosthetist. The prosthetist recommendation for the appropriate prosthesis is chosen for each case depending on the objective items like stability in the gait cycle and accurate fit of the prosthetic leg in lower limb amputee case [2].

The absence of one or both legs sure makes an imbalance in the amputee patient's gait cycle. The more important factors that affect the function of the lower limb prosthetic is the dorsiflexion angle of the foot, impact resistance, the foot life against loadings, and the material that the Solid Ankle Cushion Heel (SACH) foot is manufactured from. The SACH foot represents the traditional foot especially used by old man amputee patients who have a short stride in the gait cycle [3].

A significant improvement is obtained from using multi-axes in SACH foot [4]. These axes provide several degrees of freedom for the non-articulated foot which is reflected in a good enhancement observed in the overall gait cycle of amputee patients, but still, they have several disadvantages such as having a bulky volume, and poor dorsiflexion angle. To overcome the mentioned challenges, the composite material is a good replacement for traditional polyethylene in manufacturing the SACH foot. Other attempts to improve the mechanical properties are carried out by using date palm wood. The maximum dorsiflexion angle and fatigue life achieved are 7.5o and 1029135 cycles respectively [5].

The composite material is presented in recent research as an optimum solution for higher mechanical strength and lighter weight [6]. The carbon-reinforced polymer material demonstrates excellent mechanical properties when a polymer (as a matrix) is mixed with different compositions of carbon fibers. The great enhancement in modulus of elasticity, ultimate tensile strength, and impact resistance is obtained as carbon fiber increases, with a significant decrease in elongation [7].The fiber-reinforced polymer in nature forms like straw or rice husk may work to improve the mechanical properties of the material to a great extent [8].The finite element method with experimental design together works to obtain the required optimum contributions [9], [10].

Figure 1 shows the different designs of prosthetic feet AMP, energy-saving, and Adjustable AFO respectively. The mechatronics foot is demonstrated as an excellent development in the manufacturing of the prosthetic foot for amputee patients. A small motor, 60 watts, works to generate a maximum torque of 120 N.m in both clockwise and anticlockwise directions, assisting to make the required movement of the Ankle Mimicking Prosthetic (AMP) foot. Figure 1-a shows the schematic diagram of the AMP foot after being converted to simple springs and damper [11]. The partitioned foot supplied its power and motion from the ankle, acquired stability and tendency to the normal foot. The heavy-weight mechatronics foot consists of two motors with several springs and links [12].

Some of the novel energy storage prosthetic feet types depend on the storing of the kinetic energy generated during the stance phase and releasing some amount of it in the next gait phases [13], [14]. The "energy storage and return foot" ESAR enhanced the stride length for the patient to a good extent with symmetrical steps in gait cycle [15].

The ankle-foot prosthetic is a good replacement device for the traditional solid prosthetic foot in both; cost in the manufacturing material and the walking speed, depending mainly on its mechanism on the energy-saving or as called "energy return" by using springs elements to return back of the foot to its original position after mid-stance phase for the patient as shown in Figure 1-b [16].

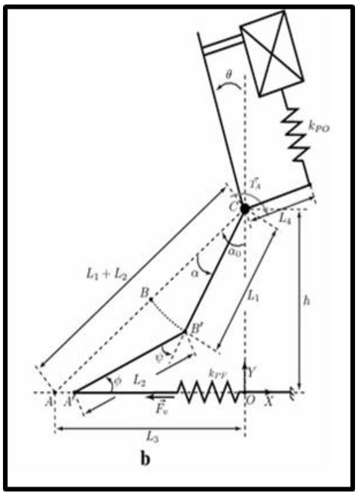
The return of the energy as a mechanism is done (in a dynamic foot) by saving the potential energy in the stance phase before releasing some of its value in the next gait cycle phase [17].

The articulated ankle-foot orthosis is a suitable device for patients suffering from severe from post-stroke. The articulated AFO provides a solution for resistance to dorsiflexion and plantar flexion by applying the mechanical forces required to gain the normal reflection for the ankle and knee joints [18]. An articulated AFO has a spring to resist plantar flexion of the foot at a beginning of the mid-stance phase which effected the height of the center of mass of the patient, in addition to knee and ankle reflection [19].

The advantage of AFO has adjustability in the degree of freedom constructed under several ground cases of gait including ascending and descending stairs. The design of AFO has a damper that controlled ankle joint movement as shown in Figure 1-c. The AFO device is manufactured from composite material while the bearing weight of the patient is supported by two steel links distributed on two sides of the leg. The adjustable device provides 8 and 26 degrees of freedom in plantar flexion and dorsiflexion respectively [20], [21].

The hydraulic damper is a perfect choice, especially wherever a higher range of motion and patient walk down in inclination is required. A computerized design may give a understanding of motion and the requirements of a prosthetic foot. The programs MATLAB and SIM HYDRAULIC are used to simulate the activity functions of the prosthetic feet with lower volume and weight [22].

This work aimed to design " Articulated Ankle-Foot Prosthetic " (AAFP) based on mixing between traditional prosthetic foot and articulated ankle-foot orthoses. It depends on the links and single slider only in design, without any springs and/or dampers. This novel design is done by LINKAGE PROGRAM that enables editing of the precise sketching before obtaining the final design that comes as a later step to the KINOVEA PROGRAM as a video clip of foot movement. The KINOVEA PROGRAM analyzes the video clip and converts it to required items. The numerical investigations are done using AUTODESK INVENTOR PROGRAM, all stresses, deformations, and factors of safety are determined by using present material (CFRP) for manufacturing of new prosthetics, in addition to steel and high-density polyethylene HDPE for comparison purposes.



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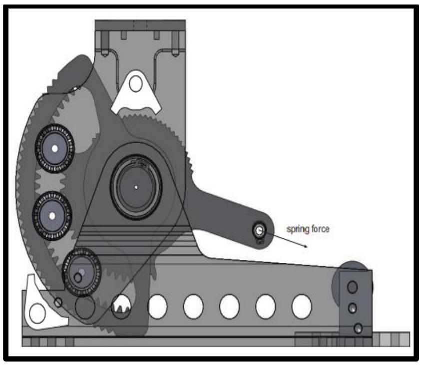

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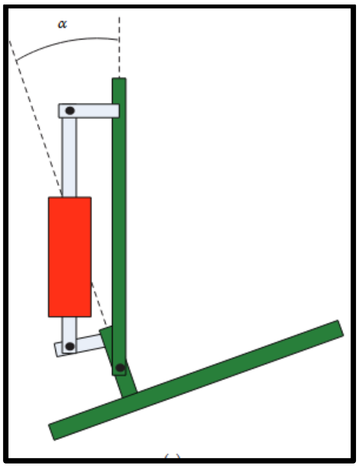


(c)

Figure 1. Different prosthetic foot.

(a) AMP Foot [. 8.], (b) Energy return Foot [.12.], (c) Ajustable AFO [.16.].

2. Methodology

The mechanism of the foot is designed according to what is called the " LINKAGE PROGRAM ". This program is similar to Computer-aided design "CAD" but in an easier form. This program has the ability to change the dimensions of the links and their quantity quickly with animation for each try. The program is exporting its file as a video clip which later represents an input to KINOVEA PROGRAM. The latest program has treated the videos by converting them to a separate sequence of images, in addition to measuring angles, distances, and times. The accurate results determined from the KINOVEA PROGRAM are the dorsiflexion angle response with time in the swing phase of the gait cycle. The results are bolstered by the evaluation of the horizontal movement of the slider, in addition to a determination of the cartesian displacements of the cross-links, in addition to measuring the time for the swing and stance phases. The present work procedures are demonstrated as block diagrams including design, experimental tests, and numerical results which are illustrated as shown clearly in Figure 2.

Regenerator at a flow rate of u_2 and an initial temperature of T_0. After heating in the regenerator area and the porousmedium combustion area, it flows out of the middle cross-

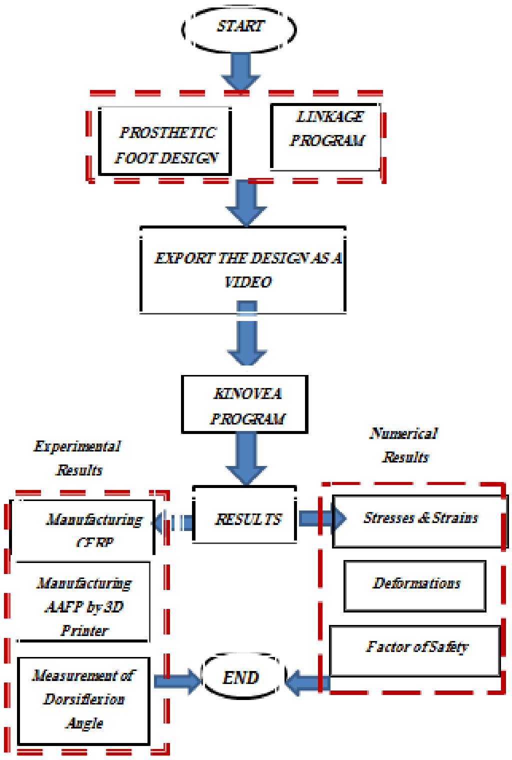


Figure 2. Flow chart of the current work.

3. Theory

One of the principal strategies in the formulation of the dynamic equations that are required to describe the overall motion of the prosthetic foot mechanism is the converting of the bearing load, the weight of the patient, to electrical power [23].The electrical power is generated is approximately 0.8 w which required in order to lift the foot upward.

In the present work, the formulation of equations is dependent mainly on the energy conservation law. The patient weight, which causes the potential energy in the foot, is converted to kinetic energy needed to make the dorsiflexion of the foot under the principle of return foot.

The total energy developed in the foot is divided into kinetic and potential energies as follows [24]:

$$U = KE + PE \tag{1}$$

The potential energy in the proposal AAFP is represented by the vertical extension of cross-links in the **y-axis as follows:**

$$PE = \frac{1}{2} K y^2 \quad (2)$$

The spring constant K is substituted by an equivalent link constant as follows:

$$K = K_{eq} = \frac{E.A}{L} \quad (3)$$

The sum of all equivalent link constants is represented the total link constant as follows:

$$K_{total} = \sum_{i=1}^{n} K_i = \sum_{i=1}^{n} \frac{E_i A_i}{L_i} \quad (4)$$

Where i = 1, 2, 3, ……, n

The kinetic energy in the slider can be formulated as follows:

$$KE = \frac{1}{2} m v^2 \quad (5)$$

The derivatives of Equation 1 items gives:

$$\Delta KE + \Delta PE = W \quad (6)$$

If the frictional forces (µ.N) in the slider are not to be ignored and other additional frictional forces in pins of links, otherwise Equation 6 is equal to zero.

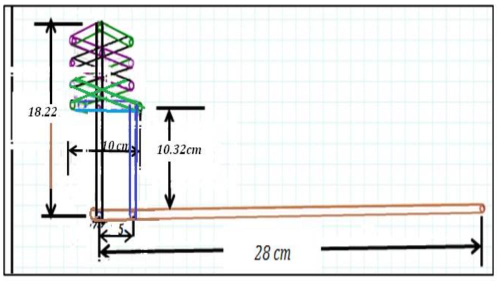
4. Proposed Design

The present proposed design is carried out using the Linkage program by connecting several links (15 links) to form a final structure. The dimensions of the foot are (2.0 * 2.5 * 28) cm of 244 g weight. The re-motion of these links several times is required to achieve the best dorsiflexion angle obtained from the program. All dimensions, front and side views of the final design, are shown in Figure 3. The present novel mechanism design is working closely with the normal foot, which makes a dorsiflexion angle reaching 8 degrees in the swing phase, and returns to the null position (horizontal position with zero angles) in the stance phase immediately with no needing for the extra equipment like springs or dampers.

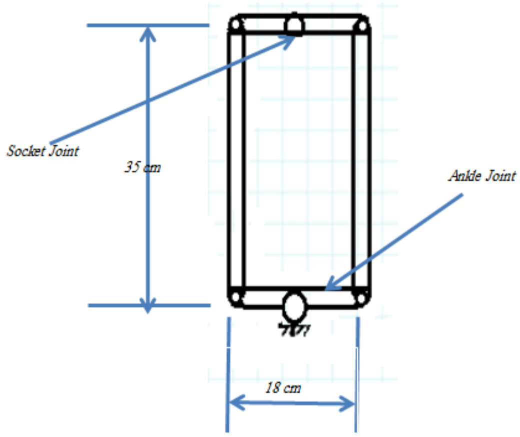
The main idea of the present prosthetic foot is constructed and presented as clearly shown in Figure 4-a, which illustrates the sketch of the mechanism animation of the foot. The white arrows represent the swing phase motion, while the red arrows refer to the stance phase. Figure 4-b shows the new prosthetic foot (AAPF) in the swing phase and its connection to the socket for amputee patients in the lower limb. In the stance phase, the red arrows represent the mechanism of motion that occurred, where the upper disc (up of the prosthetic foot) is attached to the lower disc and after that, the required supporting of patient weight by prosthetic foot is obtained. In conjunction with supporting the loading, the inclined foot is returned back to the null position (horizontally laid to the ground) by the kinematics of the cross-links.

Finally, the mechanism of the present foot can be summarized as follows:

1. Loading phase is working to generate what is called "stance Phase" by making the inclination of the foot zero.
2. The swing phase occurs when lifting the prosthetic foot from the ground in an unloading case.
3. The foot lift (at this instant the dorsiflexed angle in the foot is generated) starts and develops during the swing phase until reached maximum lift value before returning back to zero dorsiflexion angles at end of the stance phase.

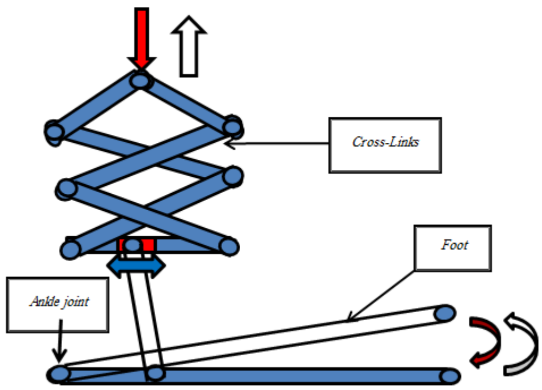


(a)

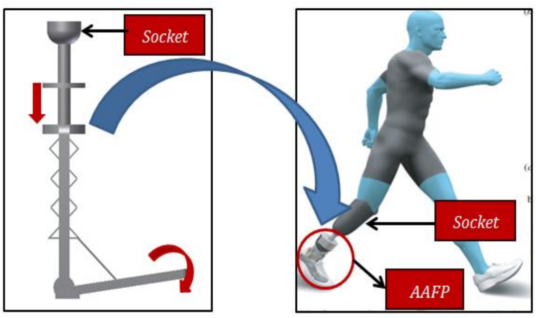


(b)

Figure 3. Dimensions of articulated prosethetic lower limb.
(a) Front view, (b) Side view



(a)



(b)

Figure 4. Ankle-Foot Prosthetic.
(a) Mechanism of Prosthetic animation (b) Connection AAFP to socket.

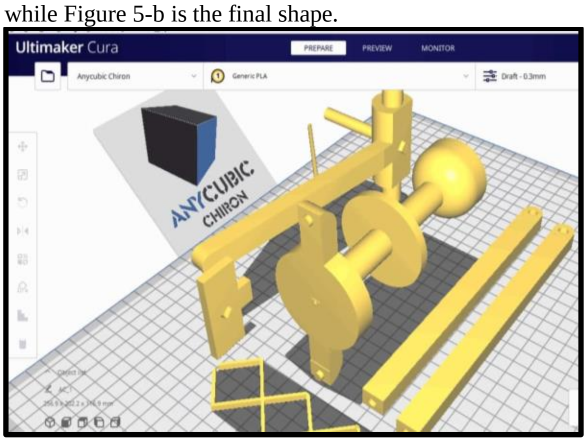
5. Results

5.1. Experimental results

5.1.1. Manufacturing the foot

The 3D printer represents the most accurate manufacturing machine in recent years for fabricating several mechanical parts, especially in prosthetics and orthotics fields. The prosthetic model manufactured from CFRP is shown in Figure 5 after being painted red color.

Figure 5-a represents details of the new prosthetic foot pre-manufacturing before executing the printing order, while Figure 5-b is the final shape.



(a)

(b)

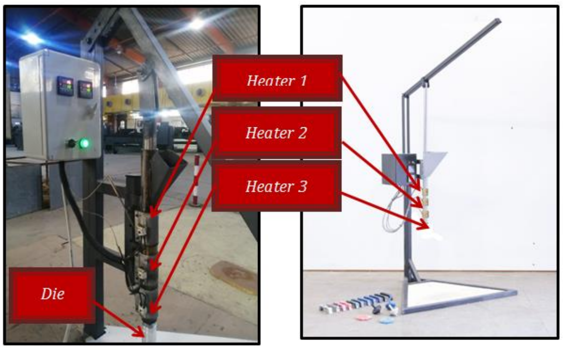
Figure 5. Manufacturing the prosthetic.

(a) Details, (b) Assembly

5.1.2. Manufacturing the CFRP material

The CFRP is produced using a rig as shown in Figure 6. The new rig consists of three heaters working to molten and mix the polypropylene particles with carbon fiber pieces. All heaters are adjustable regarding temperatures according to the material used. The last heater (at bottom of the rig) is adjusted to maintain a temperature more than the first two heaters by 20^o C to make the molten mixture like a fluid to easily occupy the shape of a die at end of the rig. The final shape of CFRP is demonstrated in Figure 7. The different compositions (PP / CF) by a percentage of weight are taken into account. The composite material (PP: 5 to CF: 1) represents the best-mixed materials for better modulus of elasticity and ultimate strength.

The investigated mechanical properties of the new material are shown in Table 1 by using the compression test of the black specimen shown in Figure 7.



(a) (b)

Figure 6. CFRP Production Rig.

(a) Real Image, (b) Schematic drawing

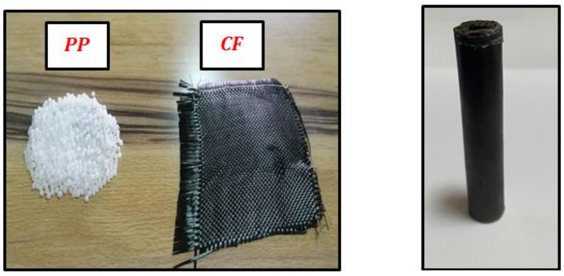


Figure 7. CFRP Specimen.

Composition, (b) Final product

Table 1. Mechanical poperties of materials under study.

Materials	Modulus of Elasticity (GPa)	Ultimate Tensile Strength (N / mm^2)	Poisson's ratio
CFRP	21.55	831	0.33
Steel[22]	200	450	0.30
HDPE[22]	1016	16.14	0.34

5.1.3. Dorsiflexion angle and swing time

The main goal of any prosthetic foot is the dorsiflexionangle which lifts the toes up during the swing phase and then becomes flat in the mid-stance phase.

The Kinova program converts the video clip to images that can be measured in foot inclination angle sequence with time as shown in Figure 8. All results are collected in one chart and compared with articulated ankle-foot orthoses [21] as clearly shown in Figure 9. The present foot presents a similar dorsiflexion response as the nonarticulated one. The time is taken logarithm because of the different time intervals in

seconds for the present work while in milliseconds in reference. The dorsiflexion angle of the proposed prosthetic foot converges its temporal response closely to the response of a drop foot patient that wears an AFO.

The proposed AAFP exhibits a maximum dorsiflexion angle with less weight reaches to 8^o and 244 gm respectively as compared with a nonarticulated prosthetic foot and SACH foot, where an 86.13 kg.f (846 N) of the weight of the amputee patient is applied at the heel in the stance phase at the gait circle. From Table 2, the suggested design acquired the foot a good increment in dorsiflexion angle reached 2.5 % and 20 % in contrast with non-articulated foot and SACH foot respectively. The new design is just a collection of 15 steel links with different dimensions. The links are made from steel with a cross-sectional area of (2.0 x 2.5 mm), the total mass of about 244 g indicated a saving in weight in contrast with a heavyweight of the non-articulated prosthetic and SACH foot by 39 % and 266 % respectively.

Table 3 presents another advantage of the new AAFP which has a swing time very close to the natural foot of a healthy person running with wide strides. The time in Figure 6-d is half of a swing time. Hence, the good convergence of the foot swing time with the runner's foot reached only 2 % as a difference which makes excellent insurance of the safety of the present kinematic foot that is approximately equal to the natural one. In the stance phase, the duration time for present AAFP is greatere than for healthy persons only by 4.66 % as shown in Table 3 which lists the gait phases time.

Table 2. Comparison of different items with Ref. [27]

Items	Present Foot	Non-articulated Foot	SACH Foot
Dorsiflexion Angle (Degree)	8.0	7.8	6.4
Wight (g)	185	400	650

Table 3. Comparison of the selected gait phases time with Ref.[28]

Items	Present Foot	Natural Foot
Swing Time (seconds)	8.72	8.90
Stance Time (seconds)	9.81	10.29

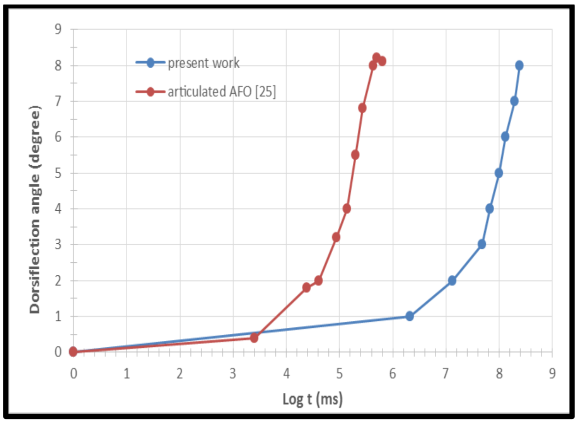
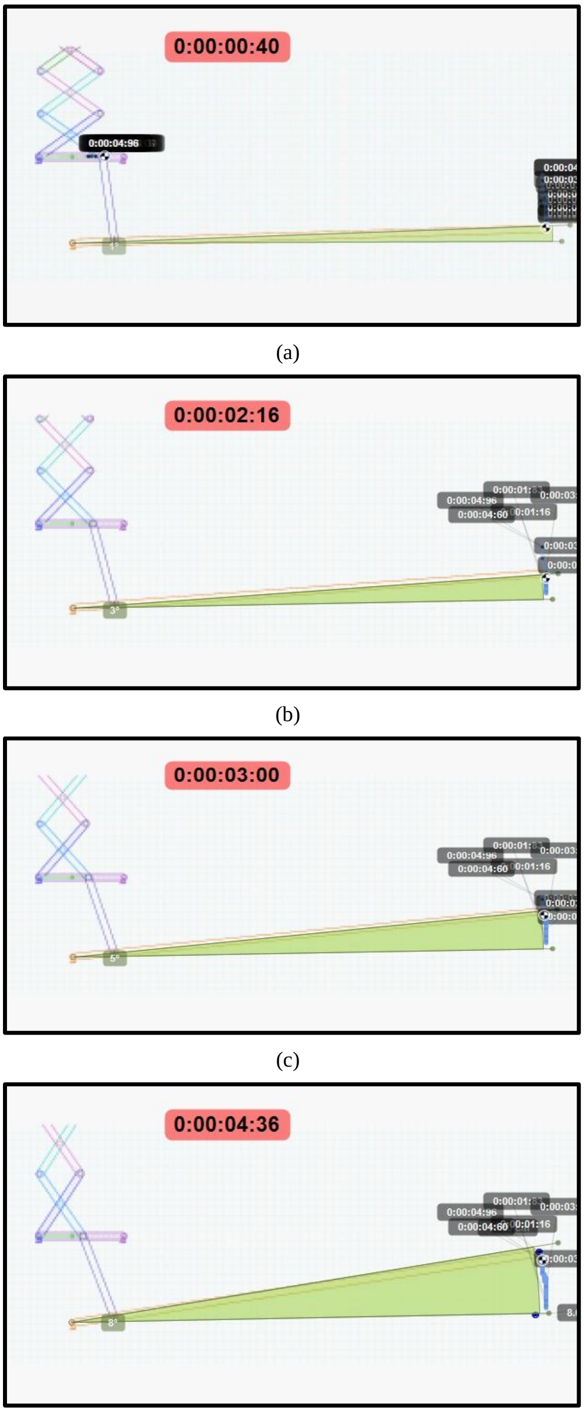


Figure 9. Dorsiflexion angle versus time.



(a)

(b)

(c)

(d)

Figure 8. Dorsiflexion angle in swing phase.
(a) 1 degree, (b) 3 degrees, (c) 5 degrees, (d) 8 degrees.

5.1.4. Cross-Links lift and slider movement

To assess and evaluate the maximum elongation in a vertical direction or as called vertical lift of the whole prosthetic foot is evaluated according to the measure of the track path of the point A indicated in Figure 10. The criteria for the determination and assessment of the vertical lift and upward and downward movements are to achieve the "optimum" dorsiflexion and also plantarflexion of the foot that can be reached from cross-link movement. The dorsiflexion angle is obtained from the upward movement of cross-links while plantarflexion is obtained from downward movement.

The vertical displacement versus the time of point A in Cartesian coordinate when it moves upward in the swing phase is illustrated in Figure 11. The red and blue curves represent the y and x-axes movement. This means each pin that connected the cross-links move upward and downward by 1.0 cm, therefore the three pins in our design move a total distance of 3.0 cm. The time consumed in turning from stance phase to swing phase is 4.6 seconds only. Therefore, a 0.64 cm/sec constant linear vertical speed of the cross-links has represented the foot speed.

The cross-links are a good replacement for ordinary spring and damper with an excellent saving in weight and volume obtained, with reasonable kinematic items like distance and speed.

The vertical movement of the cross-links produced a horizontal movement for the small slider as shown in Figure 12 for one stroke. The horizontal reciprocated distance of the slider works to produce the required inclination of the laid link (foot). Figure 13 shows the linear response of the horizontal motion with time, by reaching a maximum slider movement of 2.2 cm.

The combined animation of cross-links and slider exhibits a linear distance of 3.0 and 2.2 cm respectively which works, as a result, making up an 8º dorsiflexion angle of the foot.

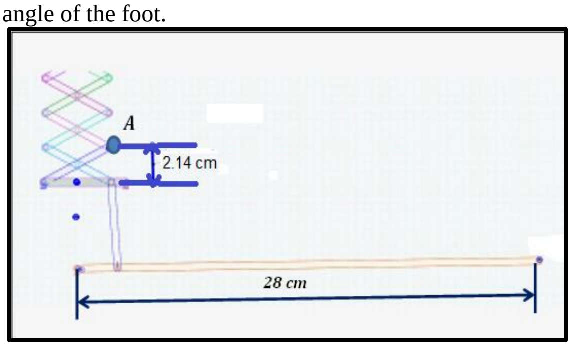


Figure 10. Null position of a prosthetic foot.

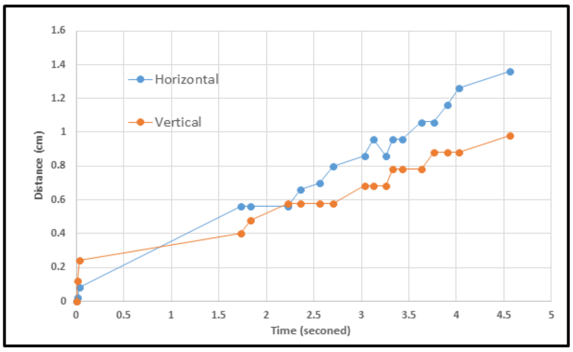


Figure 11. Movement of pin A in cartesian coordinate.

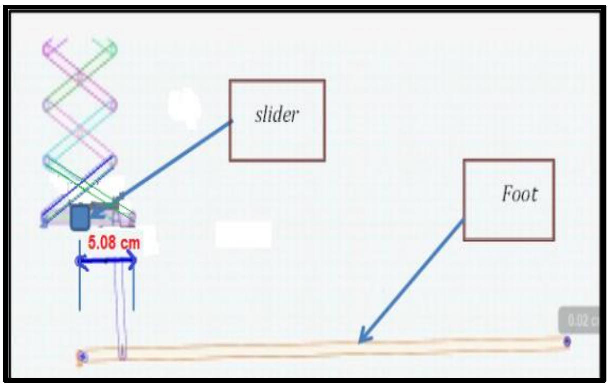


Figure 12. Movement of the slider in a horizontal coordinate.

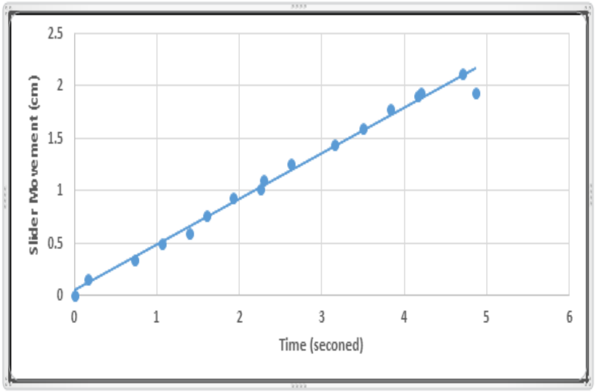


Figure 13. Horizontal movement of the slider versus time.

5.2. Numerical Results

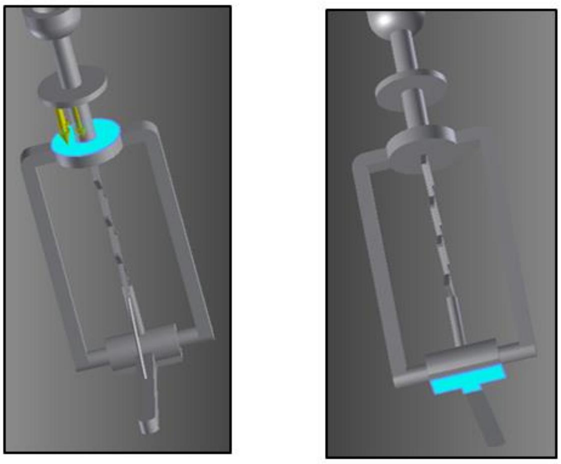
The mechanical properties of the thermoset CFRP which represented the prosthetic foot material are shown in Table 1 in addition to comparison materials (steel and HDPE). The orientation of carbon fiber is selected to be 90º (which exhibits a higher strength in comparison to other angles) and all mechanical properties are achieved by tensile test as listed from reference.

The loading applied to the prosthetic foot is 846 N which represents the patient weight. The concentrated force is converted to a bearing pressure (107.77 KPa) subjected to a lower disc that has a 10 cm diameter as shown in Figure 14-a, while the fixed support is located at the bottom face of the prosthetic heel made from silicon (for damping purposes) as shown in Figure 14-b. The total number of nodes generated in the mesh model is 5786 nodes.

The values of Von-Mises stress, deformation, shear stress, equivalent strain, and factor of safety that are presented as colored fringes across the main body of the present prosthetic foot made from CFRP are seen in Figures 15, 16, 17, 18, and 19 respectively. The maximum values of different items mentioned later are collected in Table 4 for prosthetic feet made from CFRP, steel, and HDPE materials**.**

The numerical results present higher Von-Mises stress of present CFRP prosthetic foot in comparison with the same foot made from steel, but less than prosthetic foot made from HDPE. These stresses developed in the new foot for different materials which were taken for comparison purposes. A safe design for both CFRP and steel only (reached to more than 15) in comparison to failed foot that made from HDPE.

The new material offers higher deformation and strain reaching maximum values of 0.03137 mm and 94.6 * 10^{-5} respectively so that a more elastic foot during gait phases. Other materials present lower deformations, then as a result more brittle feet are gained against shook pulses from the ground. The CFRP foot exhibits intermediate shear stress between both steel and HDPE.



(a) (b)

Figure 14. Loading and Fixation of the prosthetic foot.

(a) Bearing pressure of 107.77 KPa, (b) Fixed support

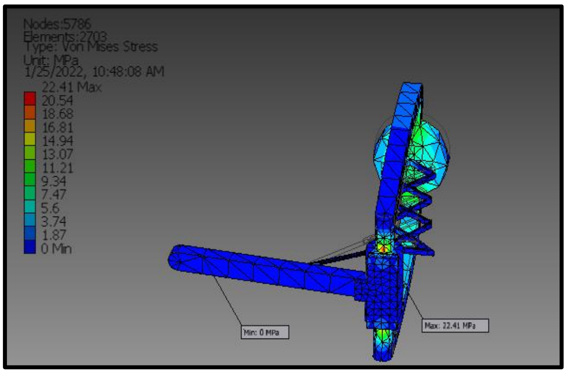


Figure 15. Von-Mises stress in prosthetic foot made from CFRP in MPa.

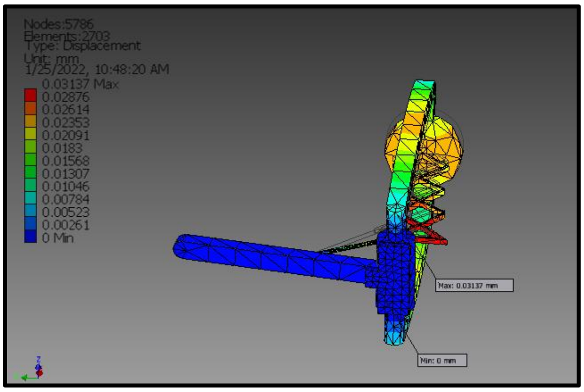


Figure 16. Deformation in prosthetic foot made from CFRP in mm.

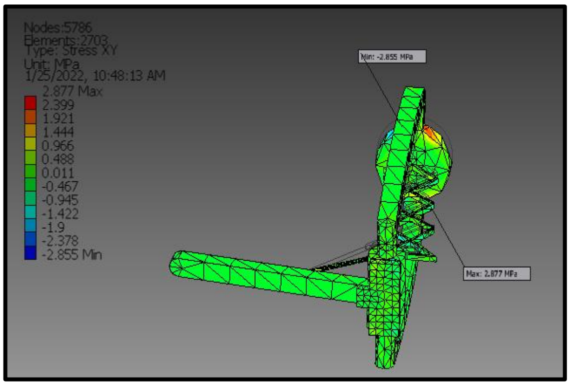


Figure 17. Shear stress in prosthetic foot made from CFRP in MPa.

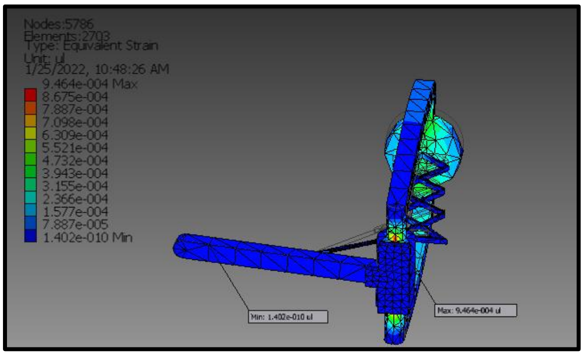


Figure 18. Equivalent strain in prosthetic foot made from CFRP.

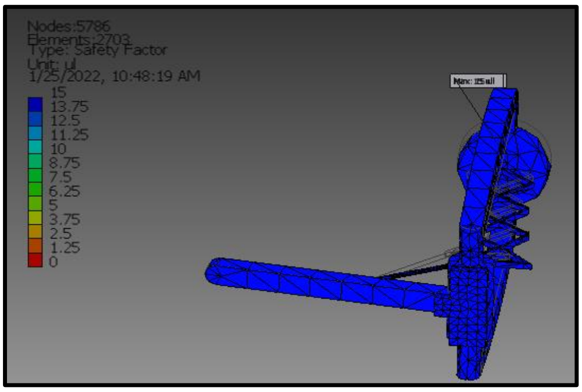


Figure 19. Factor of safety in prosthetic foot made from CFRP.

Table 4. Numerical values of items under study.

Items	CFRP	Steel	HDPE
Maximum Von-Mises stress (MPa)	22.41	10.88	31.19
Maximum Deformation (mm)	0.03137	0.001609	0.001014
Maximum shear stress (MPa)	2.877	1.455	4.272
Maximum equivalent strain	94.6 * 10^{-5}	4.65 * 10^{-5}	2.802 * 10^{-5}
Minimum Factor of Safety	15	15	0.39

6. Conclusions

The main conclusions obtained from the present AAFP can be summarized as follows:

1. High dorsiflexion angle is obtained which reached 8º, showing a good increment of about 2 % that for the nonarticulated foot, and a superior increment reached about 20 % versus SACH foot.
2. Lightweight in comparison to nonarticulated and SACH feet of 39 % and 266 % respectively, in addition to higher resistance to that patient's weight forces and impact resistance are gained due to being manufactured from CFRP.
3. Simplicity in design with mechanically foot return accomplished by no need for any complementary heavy pieces of equipment like springs or dampers.
4. Swing and stance phases time is approached closely to the normal foot, which the differences are 2 % and 4.66 % respectively.
5. Safe rigid design (like steel) but lighter weight and more elastic accomplished with using CFRP material in manufacturing the foot.

References

[1] V. Rajťúková, M. Michalíková, L. Bednarčíková, A. Balogová, and J. Živčák, "Biomechanics of lower limb prostheses," *Procedia engineering,* vol. 96, pp. 382-391, 2014.

[2] T. H. Weerakkody, T. D. Lalitharatne, and R. Gopura, "Adaptive foot in lower-limb prostheses," *Journal of Robotics,* vol. 2017, 2017.

[3] K. K. Resan, H. Ali, A. Hilli, and M. M. Ali, "Design and analysis of a new prosthetic foot for people of special needs," *the Iraqi Journal for Mechanical and Material Engineering,* vol. 11, 2011.

[4] F. Paradisi, A. S. Delussu, S. Brunelli, M. Iosa, R. Pellegrini, D. Zenardi, *et al.*, "The conventional non-articulated SACH or a multiaxial prosthetic foot for hypomobile transtibial amputees? A clinical comparison on mobility, balance, and quality of life," *The Scientific World Journal,* vol. 2015, 2015.

[5] H. S. Mohammed and J. M. Salman, "Design and Modeling the Prosthetic Foot from Suitable Composite Materials," *American Journal of Engineering and Applied Science,* vol. 13, pp. 516-522, 2020.

[6] F. Alfaqs, "Dynamic Behavior of Thin Graphite/Epoxy FRP Simply Supported Beam Under Thermal Load Using 3-D Finite Element Modeling," Jordan Journal of Mechanical and Industrial Engineering, vol. 15, No.3, PP. 301-308, Aug., 2021.

[7] A. Hadi and J. K. Oleiwi, "Improving tensile strength of polymer blends as prosthetic foot material reinforcement by carbon fiber," *Journal of Material Science & Engineering,* vol. 4, pp. 2169-0022.1000158, 2015.

[8] S. G. Ghalme, "Improving Mechanical Properties of Rice Husk and Straw Fiber Reinforced Polymer Composite through Reinforcement Optimization," Jordan Journal of Mechanical and Industrial Engineering, vol. 15, No.5, PP. 411-417, December, 2021.

[9] E. Al-Momani and I. Rawabdeh, "An application of finite element method and design of experiments in the optimization of sheet metal blanking process," Jordan Journal of Mechanical and Industrial Engineering, vol. 2, No. 1, pp. 53-63, March, 2008.

[10] A. Hadoush, "Finite Element Formulation of Internally Balanced Blatz Ko Material Model" , Jordan Journal of Mechanical and Industrial Engineering, Vol. 14, No.2, PP. 215-221, June, 2020.

[11] P. Cherelle, V. Grosu, M. Cestari, B. Vanderborght, and D. Lefeber, "The AMP-Foot 3, new generation propulsive prosthetic feet with explosive motion characteristics: design and validation," *Biomedical engineering online,* vol. 15, pp. 21-36, 2016.

[12] A. J. Young, A. M. Simon, and L. J. Hargrove, "A training method for locomotion mode prediction using powered lower limb prostheses," *IEEE Transactions on Neural Systems and Rehabilitation Engineering,* vol. 22, pp. 671-677, 2013.

[13] W. L. Childers and K. Z. Takahashi, "Increasing prosthetic foot energy return affects whole-body mechanics during walking on level ground and slopes," *Scientific reports,* vol. 8, pp. 1-12, 2018.

[14] X. Bonnet, H. Pillet, P. Fode, F. Lavaste, and W. Skalli, "Finite element modelling of an energy–storing prosthetic foot during the stance phase of transtibial amputee gait," *Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine,* vol. 226, pp. 70-75, 2012.

[15] H. Houdijk, D. Wezenberg, L. Hak, and A. G. Cutti, "Energy storing and return prosthetic feet improve step length symmetry while preserving margins of stability in persons with transtibial amputation," *Journal of NeuroEngineering and Rehabilitation,* vol. 15, pp. 1-8, 2018.

[16] B. Brackx, M. Van Damme, A. Matthys, B. Vanderborght, and D. Lefeber, "Passive ankle-foot prosthesis prototype with extended push-off," *International journal of advanced robotic systems,* vol. 10, p. 101, 2013.

[17] N. P. Fey, G. K. Klute, and R. R. Neptune, "The influence of energy storage and return foot stiffness on walking mechanics and muscle activity in below-knee amputees," *Clinical Biomechanics,* vol. 26, pp. 1025-1032, 2011.

[18] T. Kobayashi, M. S. Orendurff, G. Hunt, L. S. Lincoln, F. Gao, N. LeCursi, *et al.*, "An articulated ankle–foot orthosis with adjustable plantarflexion resistance, dorsiflexion resistance and alignment: a pilot study on mechanical properties and effects on stroke hemiparetic gait," *Medical engineering & physics,* vol. 44, pp. 94-101, 2017.

[19] A. Daryabor, M. Arazpour, G. Aminian, M. Baniasad, and S. Yamamoto, "Design and evaluation of an articulated ankle foot orthosis with plantarflexion resistance on the gait: A case series of 2 patients with hemiplegia," *Journal of biomedical physics & engineering,* vol. 10, p. 119, 2020.

[20] M. Alam, I. A. Choudhury, and A. B. Mamat, "Mechanism and design analysis of articulated ankle foot orthoses for drop-foot," *The Scientific World Journal,* vol. 2014, 2014.

[21] A. K. Lapre, "Semi-active damping for an intelligent adaptive ankle prosthesis," 2012.

[22] A. Naseri, M. Mohammadi Moghaddam, M. Gharini, and M. Ahmad Sharbafi, "A Novel Adjustable Damper Design for a Hybrid Passive Ankle Prosthesis," in *Actuators,* 2020, p. 74.

[23] C. Pylatiuk, F. Metzger, R. Wiegand, and G. Bretthauer, "Kinetic energy scavenging in a prosthetic foot using a fluidic system," *Biomedizinische Technik/Biomedical Engineering,* vol. 58, pp. 353-358, 2013.

[24] L. Bowen, J. Vinolas, and J. L. Olazagoitia, "Design and potential power recovery of two types of energy harvesting shock absorbers," *Energies,* vol. 12, p. 4710, 2019.

[25] M. J. Jweeg and S. H. Ameen, "Experimental and Theoretical Investigations of Dorsiflexion Angle and Life of an Ankle-Foot-Orthosis Made From (Perlon/Carbon fiber/Acrylic) and Polypropylene Materials," *Journal of the Japanese Society for Experimental Mechanics,* vol. 11, pp. s305-s310, 2011.

[26] S. H. Ameen, D. M. N. Mahmood, and S. A. Hussein, "Towards an Efficient Electric Pole's Material for Iraqi Electric Network," *Al-Nahrain Journal for Engineering Sciences,* vol. 20, pp. 1198-1206, 2017.

[27] M. M. Ali, "Design and Analysis of a Non-articulated Prosthetic Foot for People of Special Needs," M. Sc. Thesis in Mechanical Engineering, AL-Nahrain University, 2010.

[28] L. C. Smith and B. Hanley, "Comparisons between swing phase characteristics of race walkers and distance runners," *International Journal of Exercise Science,* vol. 6, pp. 269-277, 2013.