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Prosthetic Ankle Design and Performance Evaluation by Experimental Gait Comparison

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ABSTRACT

There are millions of persons in this world have been suffered from land mines or other accidental events which have caused amputations. The human body feet provide stability and balance when standing and moving. Amputation of a foot highly decrease the amputee's ability to practice common activities such as walking. The main target of a prostheses of any kind is to improve or return function to a physically disabled person. Although, the rapid developing of alternative prostheses technology, unfortunately, it is still a far from inquiring quite functional prosthetic limb replacement. The present study was focused on design and manufacturing of a two degree of freedom ankle rotation, plantar flexion- dorsiflexion in sagittal plane and inversion - eversion in frontal plain so as to mimic the normal human gait and also to reduce the pain and stress in the residual limb. Most ankle parts were formed from aluminum alloy and assembled to a carbon fiber foot laminated foot. The gait analysis was performed by the amputee user case study for both his prosthetic foot and the designed foot at the same optional ground surface conditions. The user foot angles responses were: eversion 2.60, inversion 2.60, plantar 8.70, dorsi 5.30, in contrast, the designed foot angles were: eversion 9.50, inversion 9.80, plantar 10.20, dorsi 10.40. The achieved designed maximum rolling in frontal plane was 100 inversion- eversion angle and the maximum rolling in sagittal plane were 120, 180 for plantar and dorsi flexion angle respectively.

1. Introduction

The main text format consists of a two Lower extremity amputee has enormous effects on the nature of life and autonomy. The Amputation patients exhibits considerable changes in many biomechanical and physical gait performance such as locomotion speed, muscles function, gait balance, and energy requirement. Prosthetic legs are so designed to return common bio mechanic function while minimize diversity influence such as gait stability [1].

Artificial foot designs have developed, obviously in recent years, via an iterative design procedure based on simple observations of how a change of users implement while using various prosthetic feet. Developing a next generation devices design have proved the possibility to get better walking performance. Despite that, challenging continuous what parameters are key to improve prosthesis efficiency [2].

Tommaso Lenzi et al.: implemented a new non back drivable cam depending transmission. Based on this modern application, they produced a consolidated and light weight artificial ankle foot. Gait and preliminary experiments with a healthy person illustrate that the developed structure efficiently relocate the foot in swing as important to rise foot allowance,

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while adapting the ankle location to the surface tendency in stance [3].

Matthew Tomkin et al.: compared the gait factors for the two Össur proflex and proflex xc prosthetic feet. A proflex foot is a three-blade carbon fiber designed to supply more ankle range of mobility compared to common energy store and release foot. The obtained data from this study provide clinical professionals with significant decision-making information about two innovative dynamic foot.[4]

Hamzah et al.: considered a novel design and manufacturing of a carbon fibre-epoxy composite material ankle-foot prosthetic. Numerical modeling and analysis were performed by means of AutoCad version 15 and ANSYS Workbench version 16.1 respectively. The proposed design presented a smooth roll-over shape and adequate energy response characteristic. The thickness of non-prismatic cantilever beam heel and keel was optimized based on utilized material considered [5].

Alexander Agboola et al.: developed two degree of freedom ankle system to permit both actuated rotation in the sagittal plane and mechanical rolling in the frontal plane. A series spring with linear actuator imitates the mechanical action of the gastrocnemius and Achilles tendon movements whereas a novel cross joint device permits two axes of rotation. Under both smooth and sloped level ground circumstances, the design was efficient to provide actuated sagittal plane rotation and torque close to a human ankle. These experiments established that the presented device contributes revolutionary development in the field of ankle design [6].

Huy-Tuan Pham et al.: described an optimization design approach for a carbon fiber prosthetic foot. Finite element analysis is utilized to simulate the mechanical behaviors during the optimization procedure. Numerical and experimental trails show that the fabricated device is capable of bending and compression to store strain energy and return it to help in forward propulsion of the amputee. These features make it more versatile and minimize the impact reaction on residual limbs [7].

Lukas Gabert et al.: proposed a novel design method performed to a new actuated polycentric ankle prosthetic. The implemented four bar ankle mechanism minimized the foot height and mass, whereas enhanced torque created and electric power. The foot performance was checked initially by healthy person using empirical orthosis.[8]

Amirreza Naseri et al.: introduced a hydraulic damper design for using in the transtibial prostheses to support amputee’s locomotion on slopes specially downhill. The hydraulic device was manufactured with two control [9]

Michael et al.: investigated the mechanical adaptability to uneven and sloped surfaces for six various designed energy storage and return prosthetic feet. The study showed that the split toe foot was better performance to adapt slops than the foot with continuous fore foot, also joints improve this by permitting eversion – inversion rolling [10].

Stefano Alleva: proposed a novel design of a powered ankle-foot prosthetic to provide a wide range of mobility and a sufficient power for forward step. An iteration cycle has performed to modify the ankle torque and angle until these values during the gait are as near as possible to the physiological quantities [11].

Michael Ernst et al.: performed experimental study for various designed feet to evaluate the performance on sloped surfaces. The results of the study showed that the split toe property in a typically present prosthetic feet enables and enhance adaptability compared to identically stiff feet with a continuous carbon forefoot [12].

Taehoon Lee et al. designed ankle exoskeleton robot with feature of rotation about two axes via utilizing a four-bar mechanism to obtain the anatomical locomotion of a one degree of freedom device generally used for ankle. Bidirectional tendon-driven actuators were introduced by taking into account the length change of both cables. Moreover, adaptive controller was used to regulate the system performance valves allow one direction flow for independent tuning of the damping ratio in the plantar flexion and dorsiflexion. To improve intact mobility, a carbon foot blade was utilized in series to a damper and spring [13].

The proposed ankle in this work is a two degree of freedom movements, plantar flexion - dorsiflexion in sagittal plane and inversion – eversion in frontal plane. This design tried to mimic normal ankle gait performance. The fixed ankle feet are lacked the inversion- eversion feature such as the one that has been used by the amputee case study and used in the present comparison study with the manufactured dynamic ankle foot which provide more natural gait and also reduce the pain and stress in the residual limb.

2. Human gait events and ankle joint kinematic

The gait cycle can be described as a time interval between any two typically identical proceedings in the ambulation path. These two commonly similar events correspond to the time when one foot touches the floor and finishes when the same foot touches the floor again [14]. Generally, the gait duration is divided into two major phases as scheme in figure (1), stance phase, covers 60-62 % of the entire cycle, begins at first heel strike and ends at toe-off. Swing phase covers 40-38 % of the cycle when the foot is not in contact with ground [15].

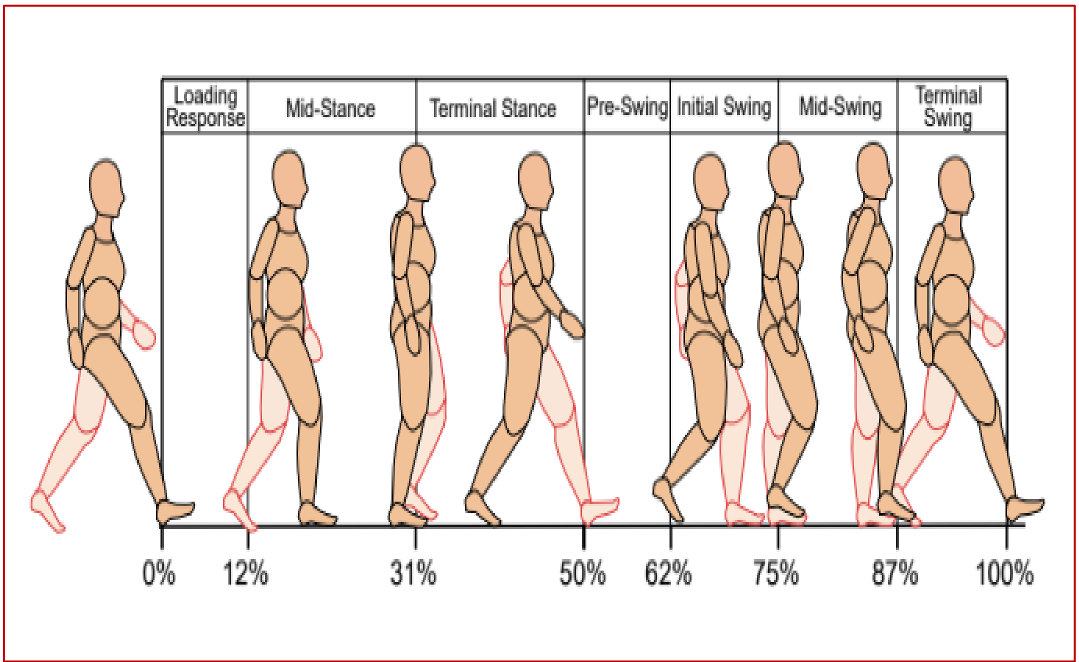
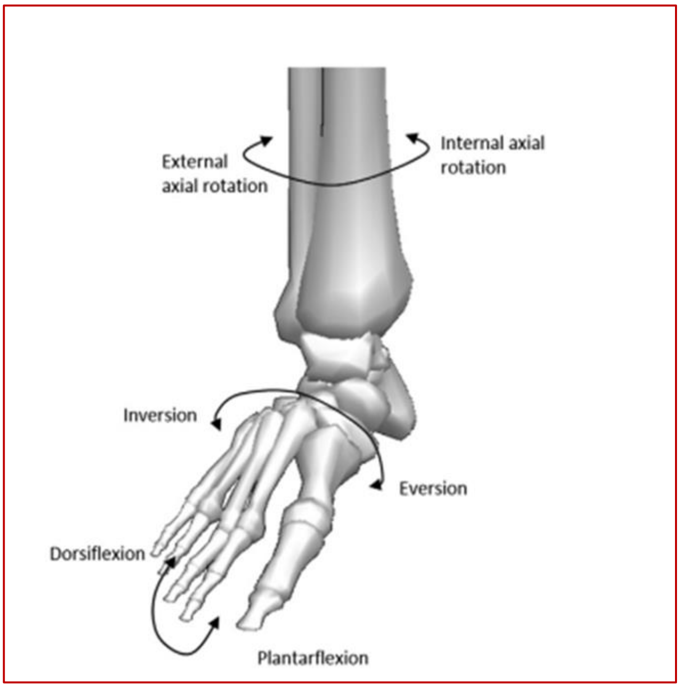


Figure 1. Gait cycle percent [16]

The ankle is a complex joint contained of the lower limb and the foot and composes the active linkage chain making the lower leg to react with the ground, a basis matter for propulsion and other activities of daily life. Although it bears high compression and shearing forces during walking, the ankle bones and ligaments structure enables it to perform with a high degree of stabilization [17].

The ankle structure allowing it to rotate in multi normal axis as shown in figure (2). The ankle movements in sagittal plane are plantar flexion and dorsi flexion, while in frontal plane it rolls from side to side, this movement known as inversion-eversion. Understanding of sound ankle kinetics and kinematics are considered the foundation towards design of a two degree of freedom artificial ankle in this work.



3. Proposed ankle design

The present study aimed to design, fabrication and performance evaluation of a two degree of freedom prosthetic ankle that allows planter flexion - dorsiflexion as well as inversion- eversion rotations. Figure (3) shows the assemble of the total components of the system designed via Solid works software version 15.0. The ankle is composed from two major parts, the lower fixed part was screwed to the carbon fiber foot and the upper part moving in two perpendicular planes by means of crossed joint and screwed to prosthetic leg by pyramid fixture. Two compression springs was utilized to provide stability during standing and also return the ankle to the original configuration beyond dorsi flexion phase. The required springs stiffness was calculated analytically and the springs selection based on experimental test.

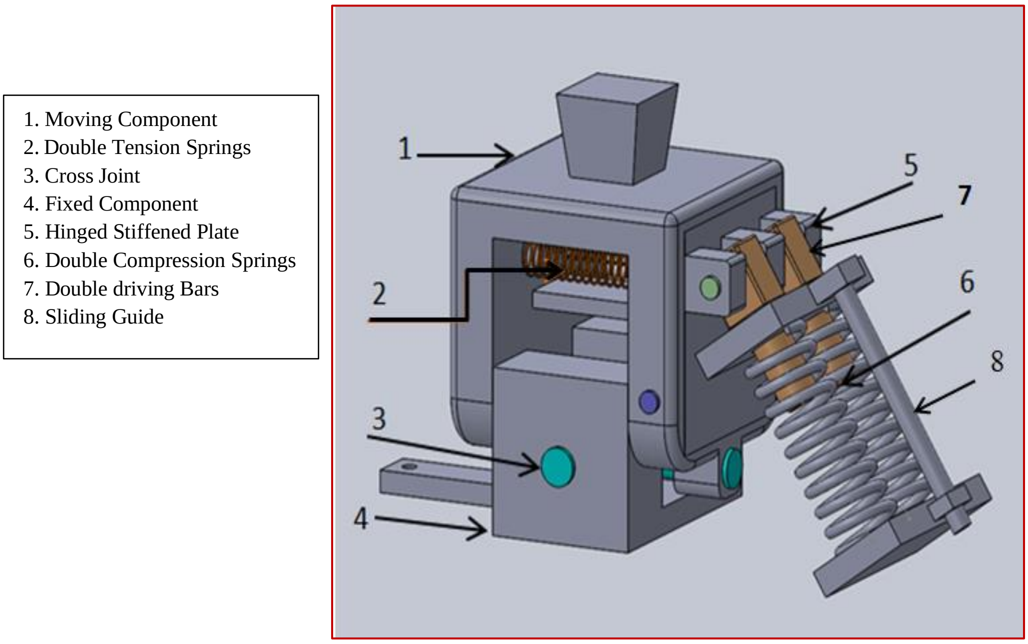


Figure 3. Designed ankle joint

Referring to Figure 3, part 1 can rotate in sagittal and frontal planes, via a universal joint part 3, to perform plantar flexion- dorsiflexion and inversion-eversion rolling. Part 4 can screw to the carbon foot by two bolts and connect to the cross joint. During the initial plantar flexion phase, the upper moving device rotates counter clock wise and in this time the hinged plate (part 5) joint releases to avoid the rotation of the two compression springs (part 6). The function of the two tension springs (part 2) is to return the plate joint to the normal position beyond the plantar flexion phase. Two compression springs (part 6) are driven by two bars (part 7) along a sliding guide (part 8) in dorsi flexion phase and the purpose of the springs are to relocate the moving upper part after clock wise rotation of the ankle.

4. Analytical analysis

The minimum required thickness of the ankle major components was calculated from fixed-fixed beam formula as below: -
Goodman fatigue equation [18]

$$\frac{1}{F.S.} = \frac{\sigma_m}{\sigma_u} + \frac{\sigma_v k_f}{\sigma_e k_b k_{su} k_{sz}} \quad (1)$$

$$\sigma_m = \frac{\sigma_{max} + \sigma_{min}}{2} \quad (2)$$

$$\sigma_v = \frac{\sigma_{max} - \sigma_{min}}{2}$$

And for Aluminum:

$$\sigma_e = 0.3\, \sigma_u \quad (3)$$

where
F.S. factor of safety
σ_m = Mean stress (Mpa)
σ_u = Ultimate stress (Mpa)
σ_v = Variable stress (Mpa)
σ_e = Endurance limit (Mpa)
K_f = Fatigue stress concentration factor
K_b = Load factor
K_{su} = Surface finish factor
K_{sz} = Size factor.
And for stainless steel universal joint:

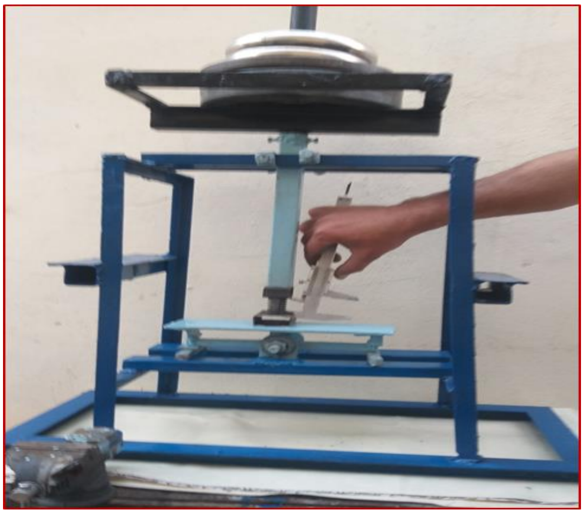
$$\sigma_e = 0.5\, \sigma_u \quad (4)$$

The required stiffness of spring was calculated from the following equation:

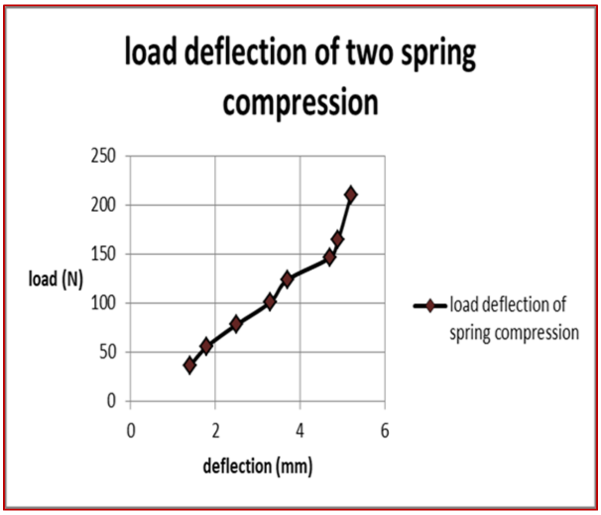
$$F_s = \frac{W}{\sin\theta_1 \cos\theta_2} \quad (5)$$

where
W: body weight (N).
Fs: spring force (N).
θ_1 dorsiflexion angle (degree).
θ_2 spring inclination with respect to the foot (degree).

The experimental stiffness value was calculated by applying weights and the resulting deflections was recorded as shown in Figure (4), the stiffness value is the slop of the load deflection curve.



(a)



(b)

Figure 4. Calculation of Spring Stiffness a) Load Appling and Deflection Measuring b) The Resulting Load Deflection Curve

5. Experimental work

The proposed ankle shown in figure (5) was cut and formed by grinding process from 10 cm shaft, 7075 aluminum alloy, which is preferable in wide application for many reasons such as ductility, strength, toughness, resistance to fatigue low density and corrosion resistance. The mechanical properties are: Young Modulus 71.7 Gpa, ultimate tensile strength 572 Mpa, yield strength 503 Mpa and poisson, s ratio 0.33. The main specifications of the entire foot are listed in Table (1).

To check the performance of the novel designed ankle, a left unilateral amputee subject was performed the gait comparison between his prosthetic foot and the fabricated foot. The prosthesis user foot is size 26 Ottobock with fixed ankle type and totally depend on a comparison spring on the heel to damp the impact force with ground and to provide the forwarded roll over energy during walking. Important information about the volunteer patient is listed in Table (2).

Table 1: Specifications of the fabricated foot

Foot materials	Carbon fiber and epoxy
Total length	24 cm
Maximum width	8 cm at keel contact region
Minimum width	5 cm at ankle region
Foot thickness	1.4 cm
Ankle materials	7075 aluminum alloy
Total foot height	13.5 cm
Total foot weight	1.3 kg

Table 2: Amputee case study information

Amputation level	weight	Gender and Age	Height	Prosthesis Specifications	Prosthetic foot weight
Above knee	62 kg	Male 43	168 cm	Glass fiber socket, 3R78 four bar pneumatic knee joint and spring type heel foot Ottobock size 26	0.65 kg

Figure (5) shows the patient, s foot and the fabricated foot, where a leg was screwed to the fabricated foot so the entire prostheses length is equal to the sound leg.

1. Socket and Knee
2. Fabricated Foot
3. The foot has been used by the amputee case study

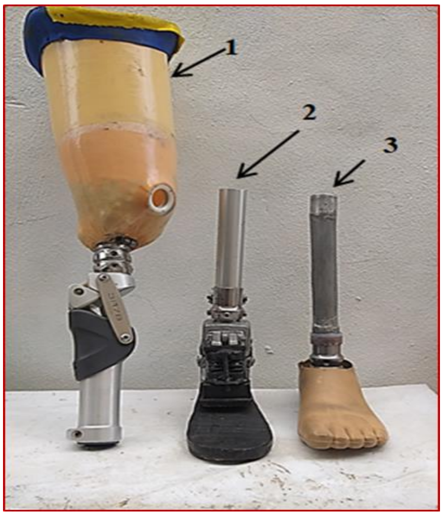


Figure 5. The two feet comparable study

6. Results and discussion

The performance of the feet has recorded during gait cycle on the fine ground surface by using the digital protractor and camera video as shown in figure (6). the plantar and dorsi flexion angle have extracted from the recorded video as illustrated in Figure (7).

(a)

(b)

Figure 6. The patient walking on fine surface, a) The user foot, b) Fabricated foot

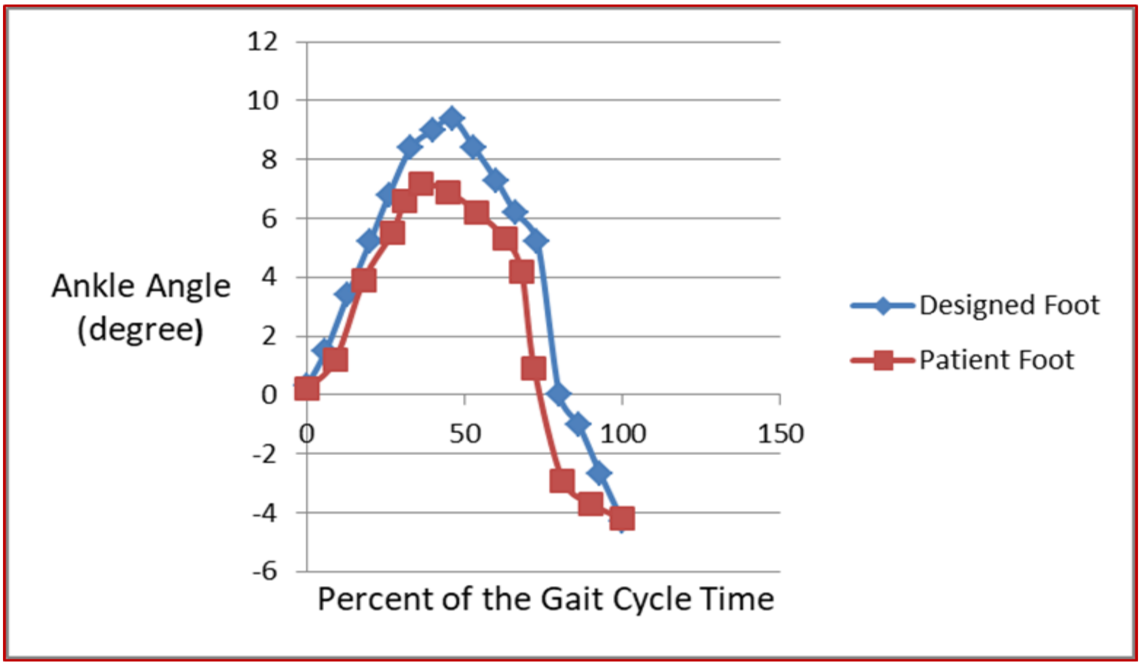


Figure 7. The feet performance on the flat fine surface

From the two curves, it is obvious that the plantar flexion angle is high for the two feet, the designed foot was deflected to 9.40, whereas the patient, s foot was deflected to 7.20. The dorsi flexion angles were closed, -4.30, -4.20 for the designed and patient, s foot respectively.

Figure (8) shows the first trail of the patient moving on rough road wearing the fabricated foot. Then reference optional locations have selected to compare the response of the two feet on the identical ground conditions.

Figure 8. The patient walking on rough surface by the fabricated foot

The ability of the foot to roll in sagittal plane was tested through the patient standing on 12 mm obstacle height (optional) as picted in figure 9. Furthermore, the plantar and dorsi flexion angles were tested on optional sloped surface and the data collected in Table (3).

(a) (b)

Figure 9. Eversion angle test a) The user foot b) Fabricated foot

Table 3: Comparison of the patient foot and the designed foot in the same ground conditions

Foot Type	Eversion angle (degree)	Inversion angle (degree)	Plantar flexion angle (degree)	Dorsi flexion angle (degree)
User Foot	2.6	2.6	8.7	5.3
Designed foot	9.5	9.8	10.2	10.4
Designed maximum angles	10	10	12	18

The testing circumstances representing by the walking surface cases were optional and they are necessary for checking the performance of the designed ankle from the point of view of the patient case study.

Overall, the conducted results, listed in Table (3), explains that the designed foot is more responsive to the walking surface conditions. In contrast, it was achieved higher ankle angles in the two cases of rolling in frontal and sagittal planes.

A clear improvement in the eversion and inversion movements and this result will reduce the stress and pain resulting from the contact of the prosthesis socket with the residual limb. The conducted results were compared with other researches as clear in Table (4).

Table 4: Features comparison of various designed feet [8]

Foot type	Height (mm)	Weight (kg)	Plantar flexion angle (°)	Dorsi flexion angle (°)
Empower and Ottobock (previously BIOM)	190	2.2	24	10
VU leg - generation 3	210	2.3	45	30
AMP	200	2.5	30	15
Walk Run	300	1.9	30	10
UMass ankle	184	1.9	10	0
Powered polycentric ankle	120	1.32	27	28
Present work	135	1.3	12	18

7. Conclusion

The present work aimed to design and fabrication of a prosthetic ankle to improve the gait performance especially in sloped and rough roads. The ankle was intended for a case study above knee prostheses user weighing 62 kg. The conducted gait results showed a clear improvement in ankle response compared to the foot that has been using by the patient. The allowance of the ankle inversion-eversion feature was added in this design so as to less the pain and stress in the residual limb when the user walking on hard roads conditions. Also, improvement of dorsi and plantar angles will enhance the ankle performance on sloped surfaces so as to mimic the normal human locomotion. Moreover, the light weight, small ankle size and low height are additional preferable features of the present design.

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