UDC 621
DOI: 10.15587/1729-4061.2022.265736

DESIGN AND KINEMATIC INVESTIGATION OF AN ACTUATED PROSTHETIC ANKLE DURING WALKING

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Due to the varied needs of persons who have lost a lower limb in their everyday lives, ankle-foot prosthetic technology is continually evolving. Numerous prosthetic ankles have been created in recent years to restore the ankle function of lower limb amputees. Most ankle foot prostheses, on the other hand, are passive, such as the solid ankle cushion heel and the energy storage and release foot (ESAR). The solid ankle foot can only provide steady vertical support during ambulation; however, the ESAR foot can store energy and gradually release it throughout human walking periods, hence increasing the walking pace of amputees. The aim of this work is to describe the design and manufacture of an actuated ankle-foot prosthesis. The main benefit of powered ankles is that they are capable of mimicking natural stride, particularly in steep or uneven terrain conditions. The primary objective is to establish two degrees of freedom of ankle rotation in two planes, plantar flexion and dorsiflexion in the sagittal plane, besides inversion and eversion in the frontal plane. As software can improve the gait stability, an automatic modifiable transmission arrangement was prepared for delivering the current design motions in the sagittal plane based on empirical collected biomechanical data related to passive prosthetic normal gait circumstances. However, the ankle rolling in the frontal plane was guided mechanically by means of mono leaf spring. The majority of the ankle mechanical components are made of 7075-T6 aluminum alloy and are integrated onto ESAR carbon fiber laminated foot. For a unilateral above-knee amputee, the ankle function at self-selected walking was assessed, achieving maximum results of 10° inversion, 10° eversion, 12° plantar flexion and 18° dorsiflexion ankle angles. Also, the patient gait experiment in a normal cadence showed an improvement in plantar flexion behavior for the powered ankle in contrast with the passive ankle

***Keywords:** prosthetic ankle, ankle kinematics, above knee amputee, prosthetic gait, energy storage and release foot*

Received date 02.08.2022
Accepted date 03.10.2022
Published date 30.10.2022

***How to Cite:** Hameed, M. I., Ali, A. A. H., Saleh, M. M. (2022). Design and kinematic investigation of an actuated prosthetic ankle during walking. Eastern-European Journal of Enterprise Technologies, 5 (8 (119)), 68–75. doi: https://doi.org/10.15587/1729-4061.2022.265736*

1. Introduction

Most passive prostheses cannot deliver positive ankle work because there are no devices that add output energy [1]. Traditional passive prosthetic feet are solid ankle-cushion heel feet (SACH feet), which have no moving parts and an internal keel that is too stiff to be bent. Currently, typical passive ankle foot prostheses are composed of one or more carbon leaf springs. These carbon leaf spring ankle feet are less stiff than SACH feet, so they could be flexed slightly once they are loaded by weight, and absorb shock at heel strike. However, since there is no movable ankle joint, plantar flexion and dorsiflexion cannot be realized, and they are still impeding the user's ambulation ability [2]. Robotic design can improve gait performance for lower-limb amputees during walking. These devices can mimic normal ankle and knee kinematics, decrease metabolic energy, and supply direct neural control of the limb. As robotic technologies improve, active prostheses are expected to enhance performance even further.

Ankle inversion-eversion is an essential feature of prosthesis function. Commercial prostheses commonly include a passive inversion-eversion degree of freedom, either using an explicit joint or a flexure. This mitigates nasty inversion moments produced by rough ground. The inversion moment has a strong effect on the side-to-side motions of the body during human walking, and its pattern is altered among individuals with amputation. Side-to-side motions seem to be less stable in bipedal locomotion, particularly for amputees. Uncontrolled inversion-eversion torque in the prosthetic ankle may partially explain reduced stability and increased fear of falling and fall rates among people with amputation [3].

Researchers have developed innovative actuators that integrate passive and active devices to imitate the natural stride. Despite that, the current typical actuation technologies have significant limitations. Thus, enhancement of prosthetic electrical and mechanical elements is still necessary to treat the unmet functions for the below knee amputated leg persons [4].

Most previous studies have been concerned with the improvement of a powered prosthesis for below knee missing limb individuals.

Therefore, research on the development of an actuated novel ankle prototype that is fit to persons with above knee leg amputation is relevant. Thus, collected virtual gait events data for a transfemoral patient served as the basis for the design of the current alternative prosthetic counterpart.

2. Literature review and problem statement

Most fundamental objects of studies in the prosthetic foot and ankle domain have been involved in various related fields to enhance performance, such as design, energy storing and release, control and dynamics. The qualitative patients' requirements and functional targets have led to the production of multi-prosthetic technologies. Recently, prosthetic feet have been standardized according to their functional performance as dynamic and multiaxial [5].

In [6], the position of the motor and gearbox was studied to improve the metabolic cost and prosthesis biomechanics by using a cable-driven system to control the ankle in the frontal and sagittal planes. The introduced mechanism achieved an 85.4 % transmission efficiency. The prosthetic device sufficiently emulates the natural ankle kinematics with adaptable mechanical strength in two DOFs. The low height profile allows the prosthesis to fit a wide range of residual leg individuals. However, one limitation of this design is the need to use two motors for ankle movements. The study in [7] focused on performing mechanical testing and numerical optimization on an ESAR foot prototype for a certain gait style to provide the designers a clear insight into the essential design parameters rather than depending on clinical feedback in the selection of a user's prosthetic device. The construction of a passive and powered ankle joint system, besides the combination system being both powered and passive with its benefits, was discussed in [8]. The required power for running was achieved in this design, which is not covered in the past robotic prosthesis. Also, the researcher presented a novel approach for the analysis and control of the combination system. The study outcomes showed that the energy method is a proper behavioral analysis of any robotic joint mechanism. Many researches worked on increasing prosthetic foot energy return during walking such as [9], where they examined the action of a Pro Flex foot and a traditional Vari Flex foot on five amputees who walked on different terrains, including fine and sloping. The tests indicated that the novel linkage foot was able to return more energy than the traditional prosthetic foot, and this additional energy was used to increase whole-body propulsion. Despite that, passive feet do not have the potential to generate positive work within the roll-over period. An unusual non-linear elastic actuator was utilized to drive a powered ankle in [10]. The developed controlled device made the transition between phases fine, and the controller could respond to various users' gait conditions. The trial results showed that the powered prosthesis can reduce the individual's metabolic consumption by 15 %, compared to the original passive prosthesis. The period of active propulsion is limited to battery power capacity. The paper [11] proposed an actuated plantar and dorsiflexion rotation with mechanical inversion-eversion rolling prosthetic ankle via a novel universal joint device that 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. The relatively bulky size, a height of 4.3 cm and a mass of 2.95 kg, makes this design not a perfect solution. The improvement of typical powered ankles, which are limited to walking at normal speed, by assembling the common elastic actuator with a lever arm, was presented in [12]. The work focused on the optimization of the lever arm length of the prosthetic device to be adaptable to walking at different speeds. The main drawback of a fully active prosthesis is the inability to keep amputee locomotion in the case of lack or absence of a power source. The authors in [13] introduced a hydraulic damper in the prosthetic foot design. The structure includes two one-direction flow valves to control the damping ratio in the plantar flexion and dorsiflexion independently. The design improved the movement and reduced the size and weight of the foot. However, fluid leakage is a big challenge of pneumatic and hydraulic systems. The researchers in [14] proposed a powered structural design with two degrees of freedom at the knee and ankle joints that could simulate the human walking motion. Moreover, the prosthesis was optimized to minimize the weight while maintaining the safety of the structure, and it can be used for stair walking. The outcomes of this work are helpful for amputees in their rehabilitation fashion. In [15], the study proposed a carbon fiber foot with a skin and core structure, as well as a powered ankle, which was developed to maintain a wide range of motion and adequate energy for a push-off step. The prosthesis is designed for walking only on a horizontal plane. In [16], a novel passive ankle design that is simple to fabricate whereas providing high performance was presented. The proposed design utilizes double springs, one to mimic the Achilles tendon, and the other to mimic the Tibias tendon. The kinetics of the ankle was analyzed via Working Model 2D finite element software. The experimental tests showed that this design was premium compared to the simple SACH foot. The statistics explained in [17] identified the state of the art for constructing a prosthetic foot. Considering patents and a bibliometric analysis over the last 6 years, the collected data showed a target of ESAR prostheses designs for patents of nearly 70 % and active prostheses designs for scientific documentation of 40 %. A novel prototype of a coupling that uses the discrete variation in viscosity happened in shear thickening fluids was proposed in [18]. The developed coupling was added to springs to achieve a speed-dependent stiffness system. This design can modify typically presented prosthetic feet. The design shows good performance in low-speed ambulation. Also, the compliant reaction in the standing style is unfavorable.

Regarding various design limitations mentioned previously, several questions are still open. All this allows us to assert that it is expedient to conduct a design and study on a novel mechanical structure with compact volume and light weight, which can emulate the biological counterpart.

3. The aim and objectives of the study

The aim of the study is to design and manufacture a novel powered prosthetic ankle, with actively controlled two degrees of freedom in the sagittal plane and passively controlled two degrees of freedom in the frontal plane, which can emulate the biological counterpart.

To achieve the aim, the following objectives are to be accomplished:

– to design the proposed prototype using CAD software and fabricate the model to fit a carbon fiber ESAR foot;

– to evaluate the maximum plantar flexion, dorsiflexion, inversion and eversion angles that the proposed prototype can experimentally reach during walking;

– to verify the kinematics of the developed actuated foot with a passive present one.

4. Materials and methods

4. 1. Object and hypothesis of the study

The hypothesis of this research is to develop a powered biaxial prosthetic ankle to provide outcome design parameters, 15°, 20°, 10° and 10° for plantar, dorsi, inversion and eversion flexure angles, respectively. The geometry dimensions were adopted to be appropriate to the user case study foot size 26 according to Ottobock. SolidWorks software was used as a helpful means for constructing the mechanical structure of the current prototype. Theoretical formulations of actuator components design are based on actual biomechanical information measured during amputee case study passive walking experiments. Active control in the sagittal plane and sensors placement were improved through experimental gait tests. Thus, the layout methodology is specified into three major items: mechanical structure design, actuator design and inversion-eversion mechanical setting. The current study proposes a novel mechanical structure with compact volume and light weight with biaxial rotation for mimicking human ankle movements in sagittal and frontal planes. The developed mechanism has been fabricated and kinematically investigated with an amputee.

4. 2. Mechanical structure design

4. 2. 1. Specifications of the Proposed Ankle Prototype

It is essential that an alternative prosthesis has the potential to mimic the function of the sound limb, as well as match the body size to maintain adequate activity. Thus, the following distinguishing characteristics are to be achieved in the artificial ankle design process:

1. Amputees should be able to put a lot of weight on their prostheses, but they should also be able to maintain their daily activities.
2. The mass of the alternative prosthetic limb must be close to the lost human limb.
3. The ankle must be capable of delivering the needed output power and torque during actuated plantar flexion.
4. In the absence of an active power source, the ankle should work passively to keep the user's gait.
5. For more comfortable walking, the ankle should perform flexible rolling in the frontal plane (inversion and eversion).
6. The prostheses must provide adequate shock damping to reduce the stress on the residual limb and to avoid any mechanical damage to the system.

4. 2. 2. CAD Modeling

The current study examined the foot capacity to rotate around two axes in order to conduct plantar flexion and dorsiflexion, as well as inversion and eversion motions. The ankle is composed of two primary components that rotate around a universal joint: the top portion with a pyramid connection for attachment to the prosthetic limb and the bottom portion with two bolts for attachment to the foot. Two compression springs were used to offer stability when standing and to reposition the foot during the stride. Additionally, the springs-ankle connection is not rigid, but rather utilizes a rotating plate joint to provide ankle movement during the plantar flexion phase. For a streamlined design process, a SolidWorks CAD model was initially developed by combining several elements as shown in Fig. 1.

Fig. 1 demonstrates a general assembly of the proposed ankle mechanism, which provides biaxial movement: plantar flexion-dorsiflexion rotation in the sagittal plane and also inversion-eversion rolling in the frontal plane through a universal joint link. The base part can be screwed to the foot blade by two bolts and connected to the universal joint. When the heel strikes the ground, the upper part rotates counterclockwise; consequently, the hinged stiffened plate releases to avoid the bending load on the compression springs. After the plantar flexion mode, the two tension springs relocate the plate joint to the original position. Then the upper part rotates clockwise in the dorsiflexion phase. At this time, the springs are compressed via two sliding bars and then gradually return the foot to the original configuration during the swing phase of the gait period.

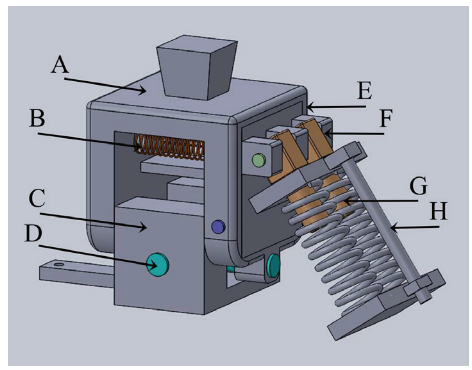


Fig. 1. Proposed Ankle Joint: A – upper part with a pyramid fixture; B – two tension springs; C – base fixed part; D – universal joint; E – hinged stiffened plate; F – two bars, G – two compression springs; H – sliding guide

4. 2. 3. Spring Stiffness Estimation

The required spring stiffness was calculated with respect to the ankle configuration in the dorsiflexion phase. Based on the AOPA project [19], the assumed design plantar flexion and dorsiflexion angles are 15° and 20°, respectively. Thus, the spring force and stiffness are:

$$F_s = \frac{w}{\sin(\theta_1 - \theta_2)}, \quad (1)$$

$$K = \frac{F_s}{\delta}, \quad (2)$$

where F_s – spring force (N);

W – load (N);

θ_1 – dorsiflexion angle (°);

θ_2 – spring inclination with respect to the foot (°);

K – spring stiffness (N/mm);

δ – spring deflection in dorsiflexion phase (mm).

According to the standard ISO load range (P_3–P_5), the value of w in (1) is P_3 (900 N) for 60 *kg* of an individual's weight [7].

4. 3. Actuator Design

4. 3. 1. Subject Walking Considerations

For a clear insight into the selection of suitable actuator characteristics and before adding electronic and sensory devices, the novel designed ankle mechanism was screwed to a carbon fiber laminated foot. A preliminary test was performed for the new prototype by an above knee left amputated leg individual, a prosthesis user since 2015, male, aged 42 years, his mass is 62 *kg*, for two reasons: first, to check the durability of the mechanical devices and also to estimate the maximum angles that the new ankle design can experimentally flex rather than depending on assumed design values of plantar flexion, dorsiflexion, inversion and eversion angles. The second important reason is that the biomechanical parameters of the gait cycle were calculated for a multi-trial walking at normal speed and are collected in Table 1.

These biomechanical data (Table 1) served as the basis for the design of a linear actuator.

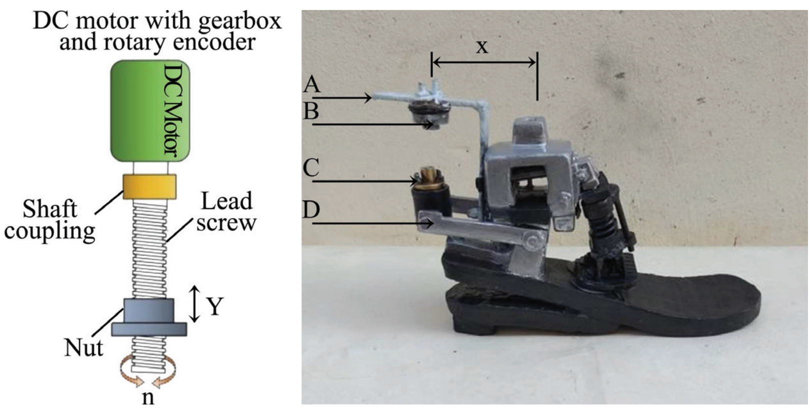
Table 1

Subject walking data

Optional walked distance	10.5 m
Total time of walking	15 s
Step distance	1 m
Speed	0.7 m/s
Time of one step	1.42 s
Time of stance period	0.852 s
Time of plantar flexion	0.426 s

4. 3. 2. DC Motor and Power Screw Design

An efficient means to convert a rotary motion of a DC motor to a linear motion is using a power screw and toothed collar system. The motor and screw assembly can drive the powered ankle via two torque transmitted arms. Fig. 2 shows the mechanical scheme of a lead screw work principal and also the position of the two ankle torque supplying arms.



a *b*

Fig. 2. Actuator Transmission System and its Position: *a* — power screw and nut; *b* — ankle and foot assembly; A — motor holder; B — ball bearing; C — nut; D — left and right torque transmitted arms

Based on the estimated data (Table 1), the required speed of the power screw was calculated as follows [20]:

$$y = x\tan\theta, \tag{3}$$

$$n = \frac{y}{L}, \tag{4}$$

$$N_{screw} = \frac{n \times 60}{t}, \tag{5}$$

$$T_{screw} = \frac{FL}{2\pi\eta_{screw}}, \tag{6}$$

$$N_{motor} = \frac{N_{screw}}{G}, \tag{7}$$

$$T_{motor} = \frac{T_{screw} \times G}{\eta_{gearbox}}, \tag{8}$$

where y – linear motion of the nut (mm);
x – length of the torque transmitted arms (mm);
θ – plantar flexion angle (°);
n – number of revolutions of the power screw;
N – angular velocity (RPM);
(t) – time of dorsiflexion (s);
F – axial force (N);
L – lead (mm);
G – gear ratio;
η – efficiency.

(3)–(8) were used for lead screw design and appropriate motor characteristics selection as described in Tables 2, 3.

Table 2

Lead screw design

Spring stiffness	111.98 N/mm
Spring deflection	10.6 mm
Spring force	1186.988 N
Lead	4 mm
Screw axial load	1067.97 N
η_{screw}	75 %
T_{screw}	0.906 Nm
Nscrew (plantar flexion)	598 rpm
Nscrew (dorsiflexion)	1,196 rpm

Table 3

Motor specifications

Power	30 W, 24 V
Motor torque	0.207 N·m
Motor speed	6,000 rpm
G	5.02:1
$\eta_{gearbox}$	87 %

The designed actuator was added to the passive ankle at distance (x) from the ankle joint as shown in Fig. 2, *b*.

4. 3. 3. Control System

A DC motor with a gearbox and rotary encoder was used to drive the ankle by means of converting the rotation speed of the motor to the linear motion of the power screw nut. The up/down movement of the nut will generate a CCW/CW torque, which is transmitted to the ankle through two arms. Arduino Uno R3 was used to regulate the motor speed and direction together with the L293D motor driver and two force-sensing resistors (FSR). The FSRs are located at the zones of contact of the heel and the keel with the ground to detect the gait phase of walking as shown in Fig. 3. The flowchart of the software to control both plantar flexion and dorsiflexion movements is shown in Fig. 4.

Generally, Fig. 4 illustrates the controlled plantar flexion, dorsiflexion and swing phases. At the beginning of the gait cycle, the heel always bears the entire body weight when it touches the ground in plantar flexion mode, thus the FSR signal attached to the heel (p_1) is very high compared to the (p_2) signal of the FSR at the keel region, which is assigned to a very small value (R_1). Consequently, the motor will rotate but not exceed (n_1) CW turns to perform the plantar flexion angle. In the end of the previous foot flexion phase, the body weight will be distributed on the whole foot and not only on the heel as the initial stride, so the (p_1) value will decline ($p_1>p_{1i}$) whereas the (p_2) value will rise ($p_2<p_{2i}$). This feedback will be provided to the controller so as to reverse the motor rotation direction less than or equal to (n_2) CCW turns in the dorsiflexion phase. Finally, the controller can sense the swing phase based on the FSRs response in the toe-off period of the gait cycle when the keel provides the body a forward energy of locomotion, thus the FSR reading in the keel is larger than the FSR reading in the heel $R_3<R_2$. In this case, the motor rotation direction will reverse (n_3) turns to relocate the foot to the flat configuration.

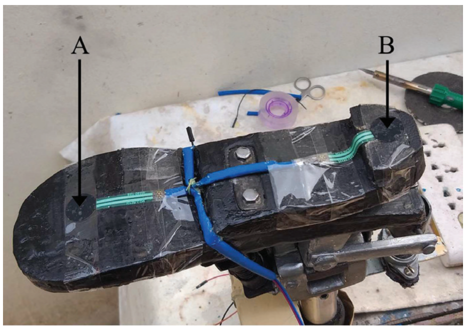


Fig. 3. Force Sensing Resistors Location: A – keel sensor; B – heel sensor

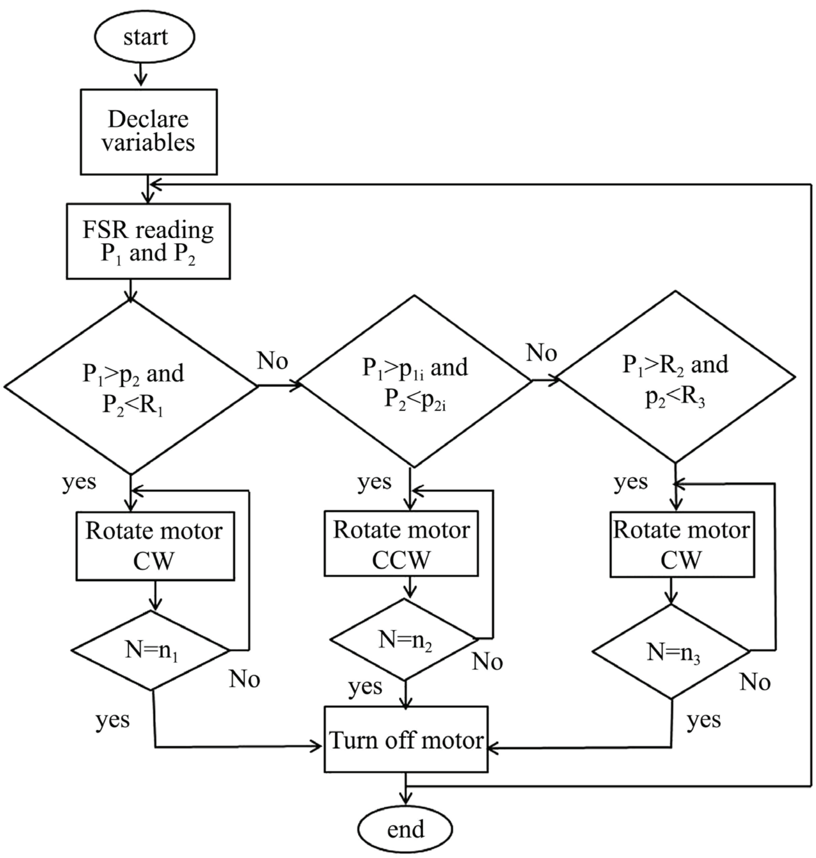


Fig. 4. Flowchart of the control system

4. 4. Inversion-Eversion Mechanical Setting

For comfortable walking and adaptability on rough terrain, the ability of the foot to turn inward or outward was included in this design. During stance events, a twisting phenomenon in the frontal plane always occurs when a foot strikes a rough surface. The prosthetic foot turning about the frontal axis assists amputees' normal gait and enhances the ground contact. It is substantial that the movements allowance be in a range close to that of the human ankle, 20–30° inversion and 5–15° eversion angles [21]. The foot free rolling was adjusted via a carbon fiber laminated beam and rubber fixed in the upper ankle part in a cantilever manner as described schematically in Fig. 5.

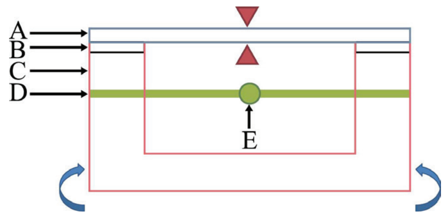


Fig. 5. Rear View of the Base Fixed Part of the Ankle on the Foot: A – laminated beam; B – rubber; C – base ankle part; D – universal joint; E – axis of rolling

When the amputee walks on a sloped surface or rough terrain, the beam will deflect under the effect of the induced ground reaction torque. The elastic strain energy stored in the beam will relocate the foot to the neutral position in the swing phase of the locomotion period.

5. Results of the proposed actuated ankle prototype

5. 1. Fabricated Model

The proposed ankle structure was made of 7075-T6 aluminum alloy and added to a carbon and epoxy composite foot. The totally fabricated model is shown in Fig. 6.

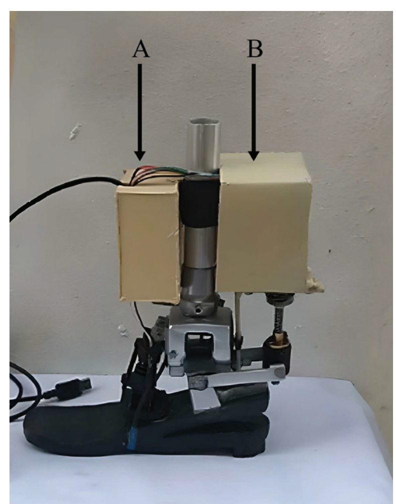


Fig. 6. Fabricated Powered Ankle Foot: A – control devices; B – motor, gearbox and battery

A leg of 165 mm in length was screwed to the ankle through a connected adapter and carrying a fabricated carbon fiber electronic board holder as shown in Fig. 6. The device becomes ready to fit the entire above knee prosthesis for performing experiments.

5. 2. Subject ambulation trial and angles measurement

The fabricated system was assembled to the entire prosthesis. Fig. 7 shows a left above knee amputated leg person wearing the fabricated design to perform experiments for the purpose of performance verification. A digital protractor shown in Fig. 8 was utilized to measure the ankle angles with camera video during the walking trial. The protractor was fixed on the leg during the plantar flexion and dorsiflexion test while it was fixed on the foot in the inversion-eversion angles measurement. The higher values achieved are summarized in Table 4.

Fig. 7. Subject wearing the fabricated foot

Fig. 8. Digital protractor

Table 4

Assumed and achieved peak ankle angles

Phase	Assumed angle (°)	Achieved angle (°)
plantar flexion	15	12
dorsiflexion	20	18
inversion	10	10
eversion	10	10

Table 4 indicates a little difference between the supposed and the accomplished actively controlled plantar flexion and dorsiflexion angles. This is due to the limited space allowance for the actuator movement. However, perfect results are accomplished for passively controlled inversion and eversion angles.

5. 3. Giat ankle kinematics

Fig. 9 describes the behavior of the designed powered ankle and a passive ankle for walking at a normal cadence on a fine surface. Overall, the recorded data showed that the maximum plantar flexion angle is higher than the dorsiflexion angle for both feet. In contrast, the novel ankle flexed a peak of 9.4° in the plantar phase versus 7.2° for the passive ankle. The peak dorsiflexion angle is nearly the same, which is about –4° for both feet. Furthermore, Table 5 illustrates a comparison of the present work with various powered ankle features.

Table 5

Characteristics of powered ankle feet [4]

Foot	Height (mm)	Weight (kg)	Plantar flexion angle (°)	Dorsiflexion 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
Reference [4]	120	1.32	27	28
Present design	135	2.2	12	18

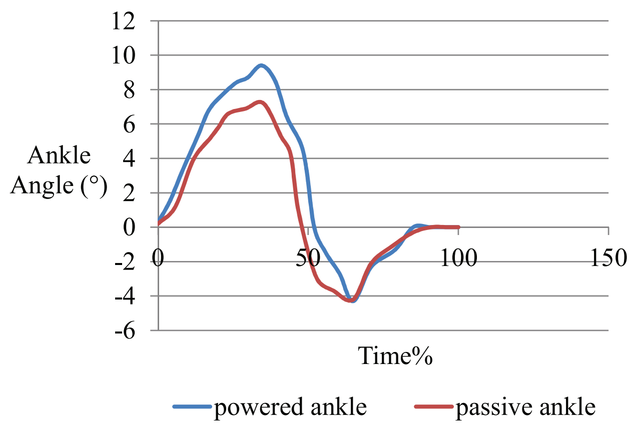


Fig. 9. Ankles kinematics

As illustrated in Table 5, the outcomes of the current device match the literature powered ankles in the essential specifications, such as height, weight and walking kinematics.

6. Discussion of the results of the proposed actuated ankle prototype

The present prototype was designed and fabricated taking into account the biomechanics of the human ankle and the essential design specifications for the alternative counterparts previously mentioned in the initial design stage. Numerical software (CAD), theoretical formulation and experimental collected information about the amputee case study, all have been utilized for achieving the goal of this study.

Fig. 1 explains the whole mechanical components of the novel proposed prototype simulated in SolidWorks software. The required spring stiffness was estimated by theoretical approach in Eq. (1), (2), and the result is 111.98 N/mm as listed in Table 2. Fig. 2, *b* shows the fabricated passive model, which is simulated previously in Fig. 1. The design of the assistive actuator was based on virtual biomechanical information illustrated in Table 1. The time of plantar flexion was 0.426 s. To perform the plantar flexion phase just at this period, Eq. (5) was applied for calculating the appropriate lead screw speed to be 598 RPM as mentioned in Table 2. In a similar manner, the required speed of the power screw in the dorsiflexion phase was calculated to be 1196 RPM. It is obvious that the speed in dorsiflexion is doubled due to the time spent on flexing the foot after reposition to the neutral configuration. Accordingly, the needed delivering motor speed was selected with respect to the outcome of Eq. (7), which is 6000 RPM (Table 3). The main goal of using (5.02:1) gears reduction speed (Table 3) is to generate high power screw torque of 0.906 Nm (Table 2) determined from Eq. (6).

In terms of control strategy, a controlled speed motor (Table 3) was used for driving the ankle movements in the sagittal plane based on the feedback of two FSRs used for gait phase detection professionally. Nevertheless, the inversion and eversion motions in the frontal plane were modified manually to avoid extra prosthesis weight. Fig. 6 shows the fully actuated fabricated device.

A volunteer amputee, ambulated at normal speed, demonstrated the kinematics of the current novel design as shown in Fig. 9. Compared to the passive ankle, the actively controlled prototype enhances the plantar flexion behavior 9.4° versus 7.2°, while the dorsiflexion amplitude is roughly the same –4° when walking on a typical surface. However, more distinguished results may appear on terrain locomotion. Table 4 shows that the inversion and eversion rolling are accomplished perfectly where the supposed and the achieved values are the same 10°. A little convergence can be shown for planar flexion results 15° versus 12° and dorsiflexion 20° versus 18° for the designed and achieved angles, respectively. However, the entire outcomes of the current study are comparable as explored with several previous powered ankles in Table 5.

The study explains sufficient effectiveness of a novel ankle design. It also provides virtual above knee amputee biomechanical information, which can be used for design and comparison purposes. The height and size of the fabricated ankle are relatively compact, which is another advantage that can be added to this work. The verification of the prototype behavior was carried out by a transfemoral individual, it is important that a transtibial amputee individual try the current device.

Large weight is still an explicit limitation of actuated prosthetic devices. Although the weight of the current prosthesis is 2.2 kg, it is reasonable compared to previous designs in Table 5. In addition to weight, another limitation of the present prototype is the adaptability to various locomotion speeds and slopes. Thus, it is necessary to update the control arrangement to be responsive to an increase or decrease in speed to maximize or minimize the required power also for walking circumstances uphill or downhill. However, extra studies are important to completely prove the effect of the proposed ankle design on the wider extremity lost population.

7. Conclusions

1. The achieved specifications of the entire fabricated prosthesis are: height 135 mm, length 240 mm and weight 2.2 kg.

2. The experiments showed that the ultimate flexion of the proposed actuated ankle is: 12° plantar flexion, 18° dorsiflexion, 10° inversion and 10° eversion angles.

3. Assessment of the kinematic behavior of the current device, by above knee amputee individual walking on level ground at ease throughout the gait events, indicated an improvement in plantar flexion response where the recorded outcomes are 9.4° and 7.2° for active and passive prosthesis, respectively. This is due to the modulation ability that the actuated systems possess.

Conflict of interest

The authors declare that they have no conflict of interest in relation to this research, whether financial, personal, authorship or otherwise, that could affect the research and its results presented in this paper.

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