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RESEARCH ARTICLE

Development of Ankle-Foot Orthosis Using Servo Motor with On-Off Control System for Drop-Foot

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ABSTRACT - This research addresses the imperative need for a lightweight, compact, and efficient design of an Active Ankle-Foot Orthosis (AFO), with the potential for everyday use and enhancement of gait cycles in individuals with drop-foot. The proposed AFO employs a servo motor to generate ankle moments based on gait phase detection, ensuring low power consumption and a structurally lightweight configuration. The study encompasses a kinematic analysis of the gait mechanism utilizing a graphical method, facilitating the determination of the requisite force for the motor actuator. Additionally, a finite element approach is employed to assess the strength of linkages. The connecting mechanism utilizes Poly (methyl methacrylate) or PMMA board due to its lightweight nature and ease of fabrication. Based on the simulation result, the minimum safety factor of link bars was 1.8. While, based on the kinematic analysis, the minimum required torque that should be provided by the servo motor is 3.838 Nm. The servo motor serves as the active element within the AFO, and an on-off control system, driven by a rotary encoder detecting gait phases, governs the AFO's movement. Ankle joint angle readings obtained through AFO implementation exhibit values within the range of normal individuals. The entire control system operates with remarkably low power consumption, registering at 0.06 W over one running cycle. Based on these analyses, a prototype has been developed for further evaluation in patient trials, underscoring the potential efficacy and practicality of the proposed active AFO design.

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Gait phase,

1.0 INTRODUCTION

The act of walking constitutes a fundamental facet of human mobility, facilitating interaction with the surrounding environment. Regrettably, a considerable segment of the population grapples with gait disorders stemming from muscular and neural irregularities [1]. One such condition, known as drop-foot, manifests as a pronounced weakness in dorsiflexion and the toes [2]. This weakness can lead to challenging situations, such as individuals stumbling on their toes or needing to lift their knees higher than usual during each step. The presence of drop-foot often signals a more significant underlying issue, and the nature of its persistence or resolution depends on the root cause, making it either a temporary or permanent condition [3]. Drop-foot is a condition characterized by the inability to lift the front part of the foot, leading to dragging the toes or foot while walking. It can result from various causes, and one common cause is damage to the peroneal nerve.

Peroneal nerve damage caused by pressure on the fibular head is the most common nerve damage in the lower extremities. According to a 2019 study, the reported incidence is 19 per 100000. Drop-foot is the most visible symptom. All age groups have the same chance of getting drop-foot, but it is more common in men (male-to-female ratio, 2.8:1). About 90% are unilateral or occur only on one side and can occur on the right or left side with equal frequency [2], [4]. Drop-foot is a condition where the sole of the foot can no longer be actively lifted against gravity due to inadequate dorsiflexion, an abnormal gait pattern appears that affects the standing and swing phases [5], [6]. This results in the dragging of the toes, decreased walking speed, shortened stride distance, increased metabolism while walking, and a higher risk of tripping [6]. Drop-foot can be caused by a variety of diseases and injuries. In principle, injury to the motor pathways of the muscles involved in foot dorsiflexion can occur at all levels, from central to peripheral, or simultaneously at several levels [5]. There are several treatment options available for patients with ankle disabilities (such as surgery, therapy, or orthotics [9]) with a full understanding of the etiology of the drop-foot experience [5]. Orthotic treatment is the most commonly performed. In addition, the plantarflexion muscles are not often affected so most Ankle-Foot Orthosis (AFO) devices are designed for drop-foot prevention [6]. Drop-foot at least by some patients is seen as a disorder with a relevant impact on daily activities [5]. Based on a study in 2005 it was found that patients with drop-foot experienced a significant decrease in quality of life, physically and psychosocially. The results of the study showed that 69% of the patients studied needed mobility aids [7].

Ankle-Foot Orthosis (AFO) is an assistive device designed to improve the daily mobility of patients with incurable drop-foot [5]. The AFO supports the ankle and foot by directing dorsiflexion and plantarflexion of the ankle [8]. AFO is the most commonly used device (about 26%) among all existing assistive devices [9]. The AFO has an important role in gait improvement by helping to maintain plantarflexion of the ankle and thus acting as a stabilizer for gait balance in

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drop-foot patients [10]. The use of AFO significantly increased the angle of ankle dorsiflexion in the standing phase and reduced the maximum angle of knee extension in the standing phase. Despite having a positive impact on the standing phase, one study reported that AFO did not have a significant impact on knee flexion during the swing phase [11].

In general, there are three types of AFO devices, passive, semi-active, and active devices. Passive AFO does not use any electrical or electronic elements or power source, but allows the use of mechanical elements such as dampers or springs to control movement at the ankle. The semi-active AFO can vary the flexibility of the ankle joint using computer controls. An active AFO has a built-in power source, control system, sensors and actuators. Based on a study conducted by Alam, et al. [6] regarding the mechanism analysis and AFO design for drop-foot patients, it was concluded that a lightweight, compact, and efficient AFO design is needed. Active AFO has the potential for everyday use as well as improving gait cycles in drop-foot patients. In this study, an active AFO was designed with a servo motor to provide a moment on the ankle based on the detection of the gait phase. The formulation of the problem is how to design an active AFO that has low power consumption and a lightweight structure.

AFO research with various types has been done by previous researchers. Wong, et al. [12] conducted a study of AFO with a passive brace that has the advantages of being compact and lightweight, but has limited movement with resistance only in plantarflexion direction throughout the walking cycle. AFO design using MR brake carried out by Furusho et al. [13] has the advantage of large resistive torque, but the device has a heavy weight. Blaya and Herr [14] designed an active AFO with a Series Elastic Actuator (SEA) which has the advantage of an ankle impedance that can be adjusted according to the walking phase, provides both plantar and dorsiflexion resistance, but it is too heavy and requires improvement in the power consumption used on the AFO system. Other studies related to AFO have also been carried out as shown in Table 1 below.

Table 1. Previous research on AFO

Research	Input	Control System	Actuator	Performance
Adiputra, et al., 2016 [15]	EMG	-	Magnetorheological Brake	Plantar and dorsiflexion successfully helped; Accuracy not identified; Energy consumption more than 20 W.
Van der Wilk, et al., 2018 [16]	Motion	On-off	Solenoid	Plantar and dorsiflexion helped, but uncomfortable; No accuracy required; Energy consumption about 184 W
Kikuchi, et al., 2013 [17]	Motion	PID	Magnetorheological Brake	Plantar and dorsiflexion successfully helped; Adaptive torque detection reference; Energy consumption more than 15 W
Kikuchi, et al., 2010 [18]	Motion	PID	Magnetorheological Brake	Plantar and successful dorsiflexion are helped by focusing on the foot-slap; Constant torque detection reference, not suitable for all patients; Energy consumption more than 15 W
Shorter, et al., 2013 [19]	Motion	PID	Solenoid	Plantar and dorsiflexion effects were not tested; Position detection is good, but response is slow; Energy consumption not identified

Many researches on the development of AFO were done. However, the study of the AFO device which focused on the kinematic analysis in parallel with strength evaluation of the linkages was not found in the previous reports. These two aspects are very important for AFO designers since they have to decide the appropriate specification of the actuator for the AFO. Moreover, the existing products do not provide external linkages which ease the motor to do its job for foot positioning. The failure or drawbacks of the existing product are more on the low success rate of the drop-foot patient during their recovery. This is due to the limitation of the conventional AVO which is not flexible. Consequently, the training for post-stroke patients needs a longer duration. Based on the background and description above, the contribution of this work is to design an active AFO that has low power consumption and a lightweight structure through conventional kinematic and strength analysis. Also, the identification of power consumption is found in the study by Shibata, et al. [20] of 57.36 mW. There has been no AFO research using rotary encoder input, on-off control system, and servo motor actuator.

2.0 DROP-FOOT WALKING PHASE

During walking, the drop-foot patient elevates the foot higher than normal to prevent the dragging of the toe [21]. The gait cycle is divided into two phases, the standing phase and the swing phase. From the standing phase, the maximum angle of ankle dorsiflexion is obtained, and then from the swinging phase, the maximum plantarflexion angle of the ankle is obtained. One gait cycle starts from the heel touching the ground until it hits the ground again [22]. Based on a study by Romkes, et al. [8], data on gait of patients with drop-foot were compared with normal values.

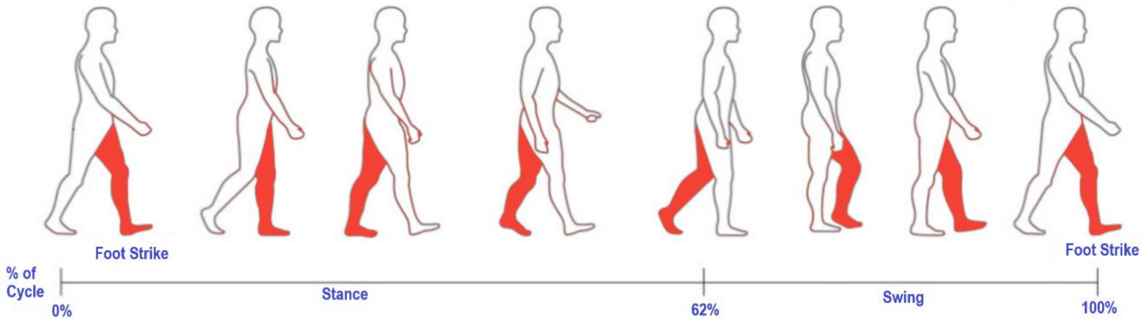


Figure 1. One Gait Cycle [22]

2.1 Ankle Joint Angle

The ankle joint angle is a crucial parameter in the analysis of human gait, which refers to the manner in which individuals walk. The ankle joint plays a significant role in the overall gait cycle, influencing the efficiency and stability of walking. The ankle joint angle is typically measured during different phases of the gait cycle to understand how the foot and ankle move. The ankle joint angle is usually measured in degrees and provides information about the position of the foot relative to the leg. In normal gait, the ankle joint angle undergoes changes throughout the gait cycle. Table 2 states the range of angle for the drop-foot patient and normal human body.

Table 2. Ankle joint angle data [8]

Variable	Drop-Foot	Normal
Initial contact Dorsiflexion	-18.5 ± 10.3°	2.3 ± 3.3°
Stance phase Peak Dorsiflexion	4.9 ± 8.8°	12.9 ± 4.1°
Toe off Plantarflexion	17.1 ± 11.3°	10.9 ± 8.4°
Swing phase Peak Plantarflexion	26.7 ± 12.6°	14.9 ± 7.6°

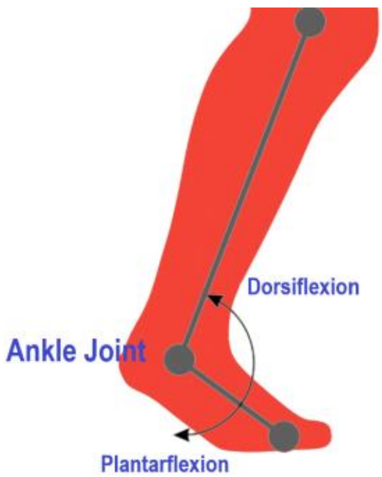


Figure 2. Definition of Ankle Joint Angle [22]

2.2 Ankle Moment

The plantarflexion moment at the ankle joint refers to the rotational force or torque that causes the foot to point downward, away from the leg. This moment is crucial in various activities such as walking, running, jumping, and maintaining balance. The peak plantarflexion moment of the ankle was obtained in two phases, the first 30% of the gait cycle and 30-65% of the gait cycle as detailed in Table 2.

Table 2. Kinetic data of peak plantarflexion moment [8]

Plantarflexion Moment (Nm/kg)	*Drop-Foot*	Normal
0-30% gait cycle	1.12 (±0.29)	No data
30-65% gait cycle	0.82 (±0.18)	1.37 (±0.14)

3.0 MATERIALS AND METHODS

This study draws upon advancements in ankle-foot orthosis (AFO) technology, encompassing a range of mechanical and control systems employed in previous research. Lessons learned from prior improvements serve as a foundation for the design methodology adopted in the current investigation. The design process initiates with the creation of a preliminary hand sketch, outlining the envisaged linkages mechanism and componentry. This sketch serves as a conceptual guide for formulating an initial design that aligns with specified requirements. Subsequently, a kinematic analysis of the proposed mechanism is conducted to ascertain the generated moment's adequacy in meeting predetermined criteria. Upon successful verification, the design progresses to the detailed stage; otherwise, iterative refinement of the initial concept ensues until momentary requirements are fulfilled. The resulting design of linkages and machine elements is translated into a comprehensive three-dimensional representation through the utilization of Fusion 360 software. Subsequent to the creation of the three-dimensional design, a simulation of the mechanism is executed using the same software. Detailed fabrication drawings of the linkages mechanism components are generated. The integration of the control system into prefabricated components follows suit. A comprehensive movement test is then conducted on the AFO after the assembly of all mechanism and control system components. Analysis of the test results facilitates drawing conclusions, allowing for a comparative evaluation with findings from previous studies.

3.1 Mechanical Design

The development of the AFO design goes through several stages of the process. The first process is to convert an initial sketch into a four-bar linkages mechanism for calculating moments. In the initial concept design process, acrylic is chosen as the component material for the connecting mechanism because of its light weight and easy to fabricate. The Servo motor is used as an actuator, and acts as the active element in the AFO. The control system used to control the movement of the AFO is an on-off control system based on a rotary encoder input that detects the phase of the gait. The results of the initial concept design then enter the design process using Fusion 360 software. The design of the connecting mechanism and the initial concept design of the AFO can be seen in Fig. 3. and Fig. 4.

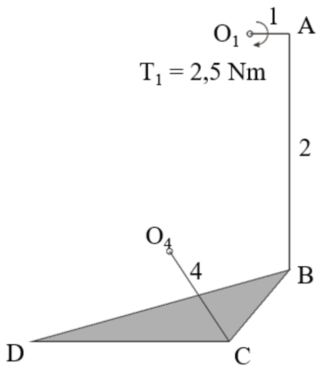


Figure 3. Connecting Mechanism Design

The calculation of the kinematics of the four-bar linkages uses a graphical method according to the Dynamics of Machinery book [23] to determine the value of the moment that occurs in the ankle joint generated by the torque of the servo motor rotation.

$$T = F(h) \tag{1}$$

where T is moment or torque (Nm), F is Force (N), and h is the perpendicular distance between forces (m). The basic moment equation above is the last stage of the calculation using the graphical method which is the result of multiplying the force vector acting on the link with the perpendicular distance between the forces.

Description:
1. Four bar linkages
2. Servo motor
3. Rotary encoder
4. Footwear mount
5. Servo mount
6. Strap

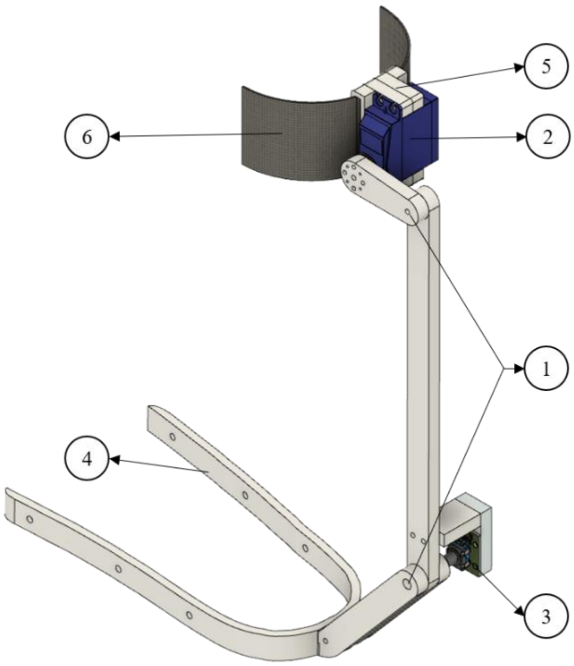


Figure 4. AFO concept design

3.2 Simulation Management

Design simulations were carried out using Fusion 360 software, the first was a stress test and the second a displacement test. Where the simulation is given a load of 250 N according to the capacity of the motor. This value is the maximum force that can be created by the servo motor. Therefore, it is used to determine the strength of the linkages with the base material used for the linkages being clear acrylic. The value of 250 N is equivalent to a 25 kg load accelerated in 1g. This also mimics load distribution during walking in which not all patient weight is supported by the AFO mechanism. The total number of linkage simulated in strength simulation is 6 bodies as shown in Fig. 5.

Table 3. Clear acrylic material properties

Acrylic, Clear	
Yield Strength	40 MPa
Tensile Strength	60 MPa
Poisson's Ratio	0,395

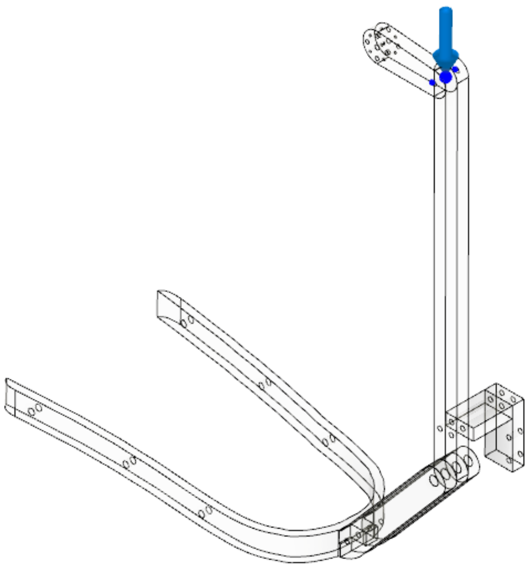


Figure 5. Simulation Model

3.3 Control System

The control system used to control the movement of the AFO is an on-off control system based on a rotary encoder sensor input (Rotary Encoder type LPD3806-100BM-G5-24C, China) that reads the angle of foot movement according to the walking phase [24, 25]. Arduino Uno as a microcontroller provides output to the servo motor (Tower Pro MG995, Capacity 10 kg) in the form of an angle rotation to provide a moment on the ankle to improve the maximum angle of dorsiflexion and plantarflexion of drop-foot patients. The control system is supplied by a 9 V battery alkaline, ABC, Indonesia.

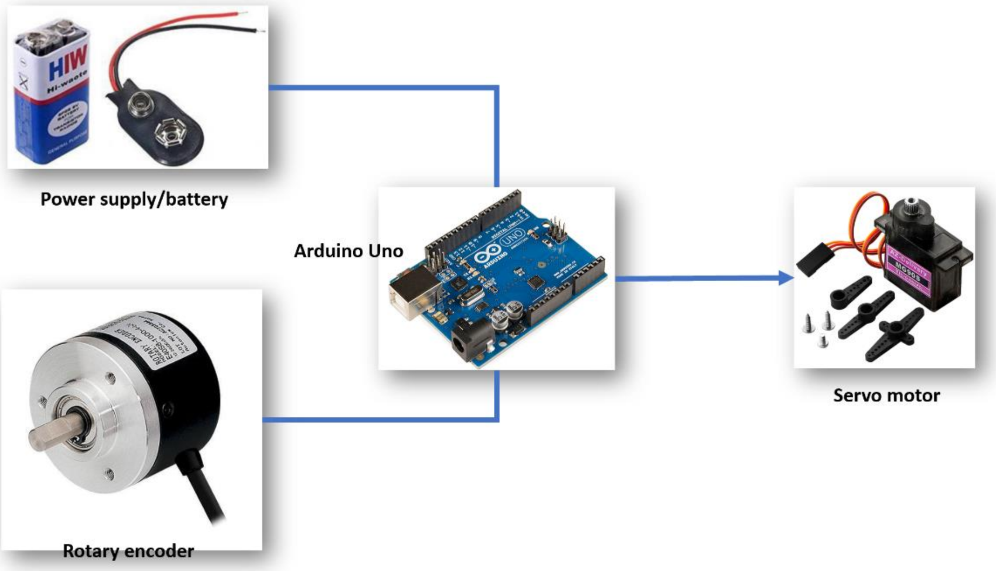


Figure 6. Control system scheme

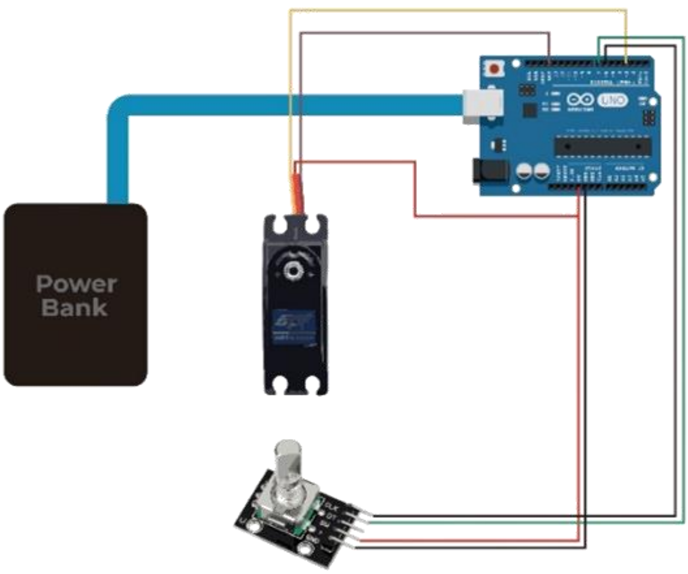


Figure 7. Control system circuit

3.4 AFO Movement Test

AFO movement testing is carried out after all the fabrication and assembly of the AFO control system has been completed. Using Fusion 360 software, the need for rotation of the motor to improve the angle of dorsiflexion and plantarflexion can be determined. Then the prototype was tested on the foot directly and recorded to determine the ankle angle result. The position of the rotary encoder can be known through the serial monitor on the Arduino software. Energy consumption testing is obtained by using a digital multimeter to read the voltage and current that works on the motor for one gait cycle. The calculation of power and energy consumption was carried out to determine the efficiency of the control system used.

$$P = V.I \tag{2}$$

$$W = V.I.t = P.t \tag{3}$$

where V is voltage (V), I is Current (A), t is Time interval (second), P daya Power (W), and W is Energy (Joule).

4.0 EXPERIMENTAL RESULTS

The design process was carried out by calculating the moments produced by the linkages mechanism on the footwear. If it meets the needs of the drop-foot patient moment, a 3D design is made using Fusion 360 software and a stress test and displacement test design simulation is carried out to determine the strength of the linkages. AFO movement testing is done by taking a sample image from the recorded AFO usage and looking at the ankle angle obtained to be compared with the ankle angle data in drop-foot patients and normal people. Energy consumption that works on the motor during one gait cycle is obtained from the voltage and current readings and then compared with previous studies according to the control system used.

4.1 Kinematics Analysis

Kinematic calculation from the design of the linkages mechanism determines whether the moment result is sufficient for the peak plantarflexion moment of the ankle or not. Link 1 is analyzed first, and there is T_1 given by the servo motor rotation of 2.5 Nm, with the moment equation applied to obtain F_{21}, as shown in Fig. 7(a). F_{21} was obtained at 62.5 N. Link 3 cannot be fully analyzed because the value and direction of F_{03} and F_{43} are unknown. F_{43} is broken down into its components, F_{43}^{T4} and F_{43}^{N4} as shown in Fig. 7(c). By taking the moment at point D, we can obtain F_{43}^{T4} by equation $(F_{23})(DM) = (F_{43}^{T4})(DN)$. The similar triangles shown in Fig. 7(d). will give the answer to the equation for the value of F_{43}^{T4}. Its direction is that it opposes moment F_{23} to point D.

The value and direction of F_{03} and F_{43}^{N4} are unknown, link 3 can be reduced to a three-force system by combining F_{23} and F_{43}^{T4}. F_{03} must work through the intersection of the resultant obtained and F_{43}^{N4} by applying a moment equation. The force polygon for link 3 is shown in Fig. 7(h). The moment applied to the foot part of the AFO connecting mechanism can be obtained with the value and direction F_{03} and the foot length of 200 mm as shown in Fig. 7(i). Graphical kinematics calculation gets the moment that acts on the footwear is 3.838 Nm. Where this figure meets the needs of the ankle moment of drop-foot patients based on a study by Romkes, et al.

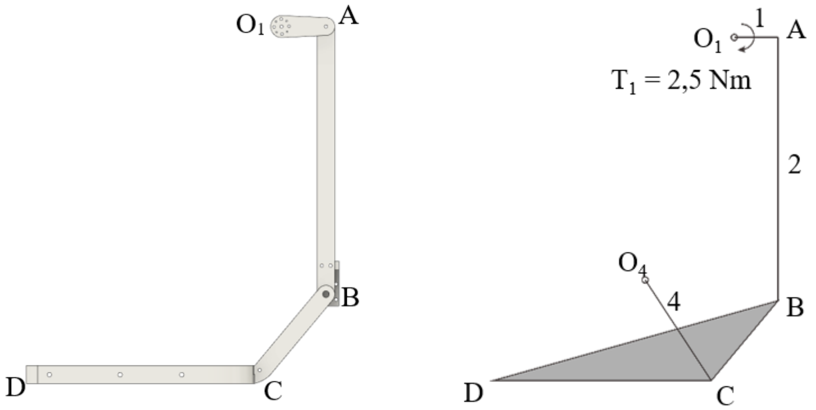


Figure 8. AFO linkages mechanism

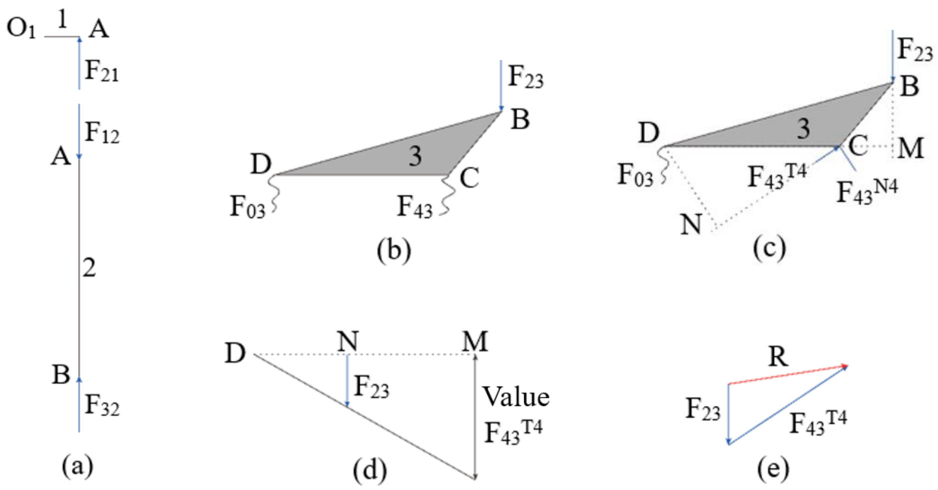


Figure 9. AFO linkages analysis

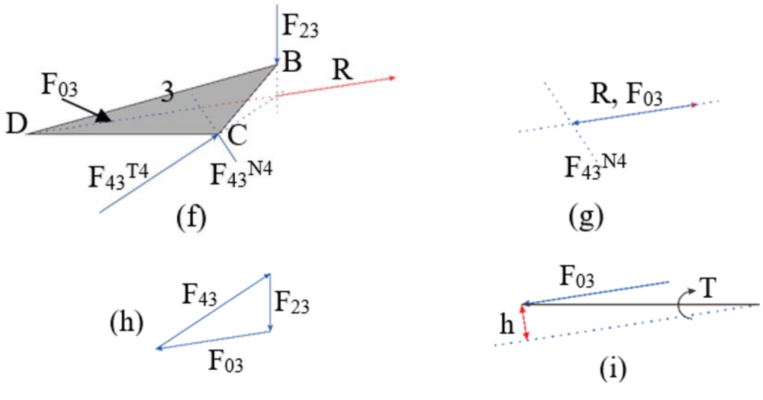


Figure 9. (cont.)

4.2 Linkages Strength Simulation

In accordance with the initial goal of designing a lightweight non-metallic AFO structure, clear acrylic material was chosen for the connecting mechanism which has a yield strength of 40 MPa, tensile strength of 69 MPa, and a Poisson's ratio of 0.395. Fusion 360 software was used to design the AFO prototype and run stress test and displacement test simulations before being implemented directly. Stress test using von Mises Stress showed a minimum yield strength of 2.885E-21 MPa and a maximum yield strength of 21.96 MPa. The minimum displacement of 0 mm and the maximum displacement of 1,108 mm are in the position of the rotary encoder holder.

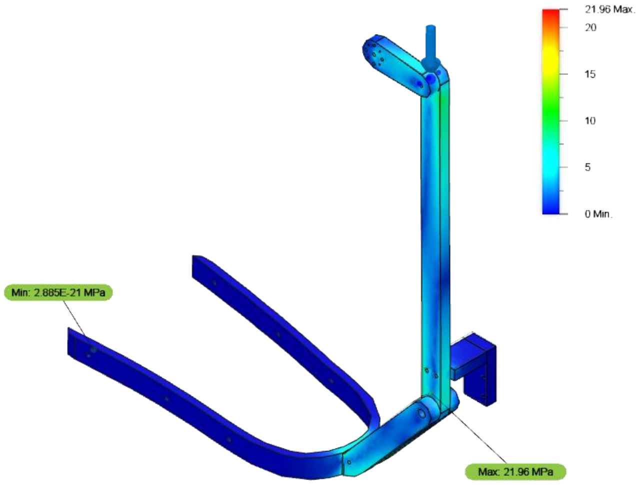


Figure 10. Location of maximum and minimum yield strength after applying stress test

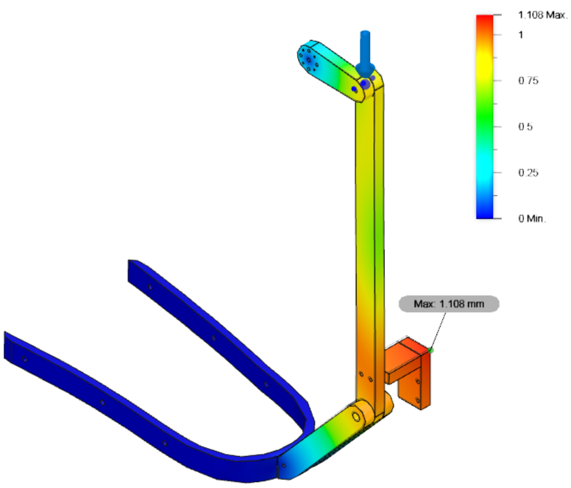


Figure 11. Maximum displacement and its location after applying the displacement test

The minimum safety factor shows a result of 1.821 which is a small value, which may be sufficient to meet the design requirements but with external factors, it can cause bending or breaking. The largest displacement position in the rotary encoder holder can be caused by the condition of the rotary encoder being locked with the linkages so that when the linkages move downwards due to the load, the rotary encoder holder supports the load.

4.3 Ankle Movement

The movement test on the AFO prototype was carried out to determine the performance of the linkages mechanism and control system used. In the movement test, angle input to the motor is given for 4 phases in the gait cycle, initial contact, standing phase, toe off, and swing phase. Referring to the ankle joint data in normal people, using the Fusion 360 software, it can be seen the angle of rotation of the servo motor is needed, so that the ankle angle reaches a normal condition.

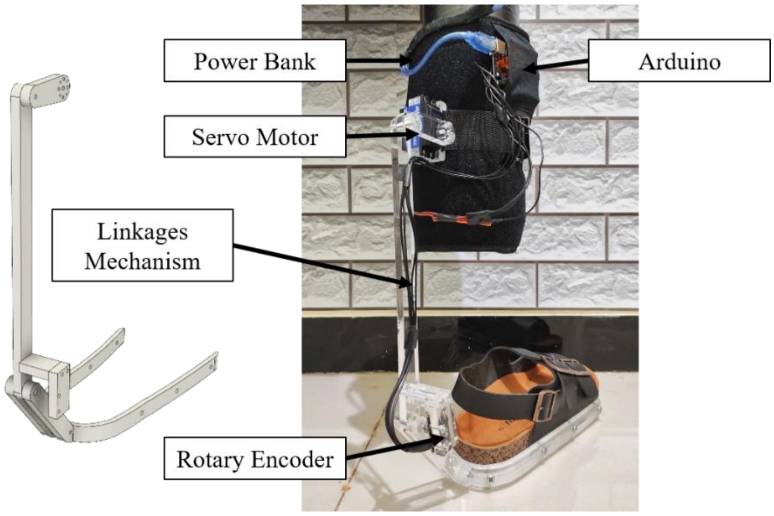
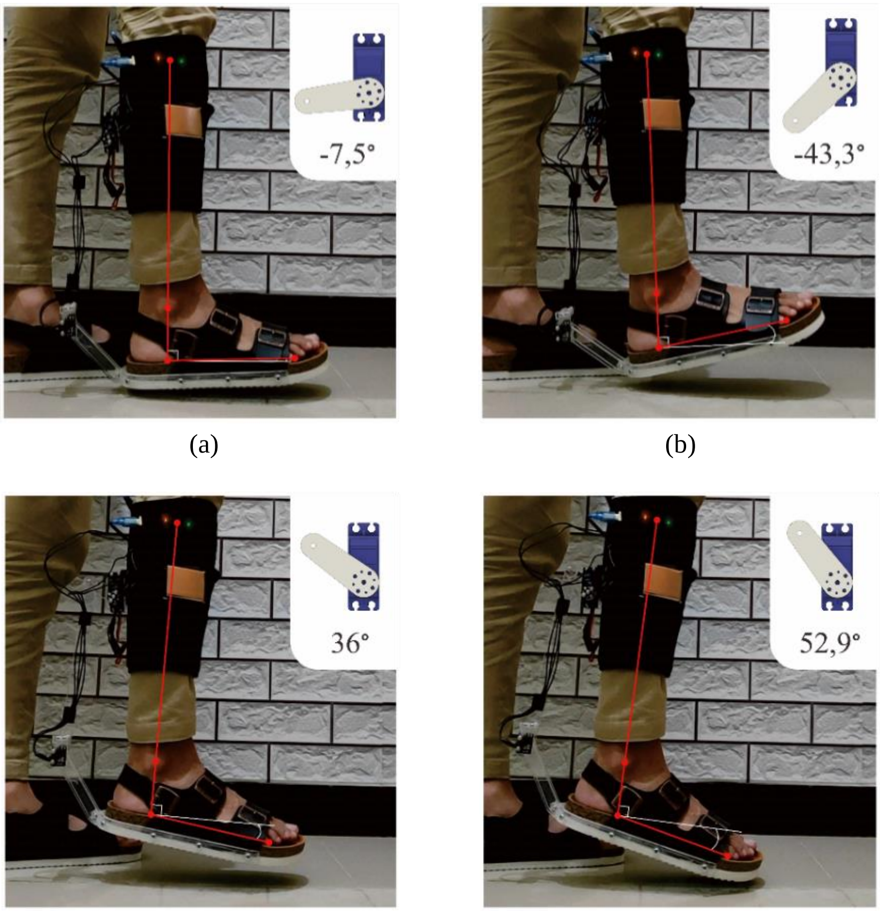


Figure 12. Implemented AFO model



(a) (b) (c) (d)

Figure 13. Ankle joint angular motion testing (a) initial contact; (b) standing phase; (c) toe-off; (d) swing phase

To correct the patient's ankle angle, the angle required at initial contact is 7.5° counterclockwise and used as a zero value from the rotary encoder reading. Then 43.3° counterclockwise in the standing phase which is the peak dorsiflexion, with the rotary encoder reading at position 3. 36° clockwise for the toe-off phase and 52.9° clockwise in the swing phase which is the peak plantarflexion with the rotary encoder position reads the same at position -1.

Table 4. Result of ankle joint angle using AFO

Variable	Drop-Foot	AFO	Normal
Initial contact Dorsiflexion	-18.5 ± 10.3°	1.582 °	2.3 ± 3.3°
Stance phase Peak Dorsiflexion	4.9 ± 8.8°	9.763 °	12.9 ± 4.1°
Toe off Plantarflexion	17.1 ± 11.3°	8.120 °	10.9 ± 8.4°
Swing phase Peak Plantarflexion	26.7 ± 12.6°	12.652 °	14.9 ± 7.6°

The ankle joint angle is obtained by taking a right angle to the foot as a reference and looking at the dorsiflexion or plantarflexion angles of the ankle joint. The ankle joint angle obtained with the use of AFO shows a value that meets the deviation of values in normal people so it can be said that the AFO movement has succeeded in improving the ankle angle in drop-foot patients.

4.4 Energy Consumption

To determine the efficiency of the control system used, a power test was carried out by reading the voltage and current using a digital multimeter that worked on the servo motor for one gait cycle.

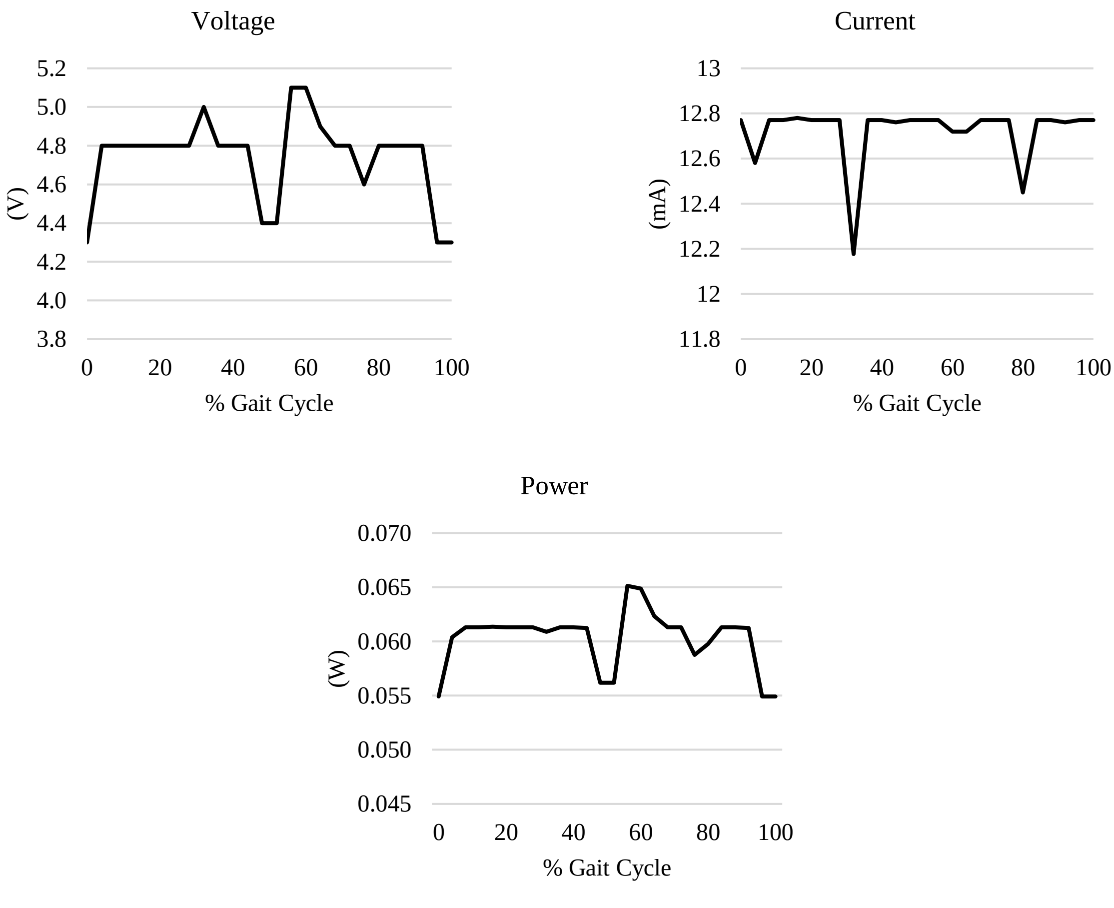


Figure 14. Power test result for one gait cycle

The voltage has an average value of 4.7 V and a current of 12.7 mA. The average power obtained is 59.69 mW. With a drop-foot patient's gait velocity of 1.15 m/s and a stride length of 1.13 m based on research by Romkes et al., the energy consumption for one gait cycle was 0.06134 Joule. To determine the efficiency of the control system used, it is compared with the other control system used in previous studies by Shibata, et al. which uses two 5 V batteries, one as a motor power source and one connected to a step-up chopper circuit to charge and has a power consumption of 0.05736 W. So the required energy consumption is 0.05837 Joule for one gait cycle, so that it can be said that the application of the AFO control system has low energy consumption. Several active or semi-active AFOs have been proposed by Adiputra et al., [15], Kikuchi et al., [17,18], and Van der Wilk, et al., [16]. Based on these four works, their power consumption was more than 15 W. Compared to the existing device, the power consumption of the existing product is much less than the previous work. This means that this device can be considered as an efficient device and therefore it will potentially get more attention from the industry.

5.0 CONCLUSIONS

Kinematic analysis was carried out to determine the minimum required torque of the linkage which was 3.838 Nm. The design of the AFO mechanism meets the needs of the moment on the footwear. The simulation results of the structure show that improvements are still needed on the design side, especially on the encoder bracket. So, it is expected to increase the safety factor value of the AFO mechanism structure by at least 2.5 for the general requirement of dynamic devices. The AFO movement provides an improvement in the angle of dorsiflexion and plantarflexion of the ankle joint from drop-foot patients to near-normal movements. The whole control system running during one gait cycle has low power consumption compared to the previous studies. The future works will be focused on the real patient test which was not covered in this manuscript.

6.0 ACKNOWLEDGEMENT

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