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EMG-Based Control of Active Ankle-Foot Prosthesis

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Abstract

Most below-knee prostheses are manufactured in Iraq without considering the fast progress in smart prostheses, which can offer movements in the desired directions according to the type of control system designed for this purpose. The proposed design appears to have the advantages of simplicity, affordability, better load distribution, suitability for subjects with transtibial amputation, and viability in countries with people having low socio-economic status. The designed prosthetics consisted of foot, ball, and socket joints, two stepper motors, a linkage system, and an EMG shield. All these materials were available in the local markets in Iraq. The experimental results showed that the maximum range of motion to move the designed prosthetic in the sagittal and frontal planes reached 70% of the healthy foot range of motion relative to the signals of the gastrocnemius muscle of a healthy leg person. The angles that represented the range of motion achieved in various directions at the ankle joint were Dorsiflexion Angle (35˚), Plantar Flexion Angle (25˚), Inversion Angle (20˚), and Eversion Angle (15˚).

Keywords: Prosthetic, Dorsiflexion, Plantar flexion, Inversion, Eversion, Ankle joint.

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نمذجة والسيطرة بمتحسسات العضلات على طرف قدم- كامل فعال

رؤى مصطفى احمد[1،*]، محسن عبدالله عبد الحسين الشمري[2]

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الخلاصة

يتم تصنيع معظم الأطراف الاصطناعية تحت الركبة في العراق دون الأخذ بعين الاعتبار التقدم السريع في هذا المجال من البدلة الذكية التي يمكن أن تقدم الحركات في المطلوب الاتجاهات حسب نوع نظام التحكم المصمم لهذا الغرض غاية. أن التصميم المقترح يتمتع بميزة البساطة ، وبأسعار معقولة ، وتوزيع أحمال أفضل ، ومناسب للأشخاص مبتوري الاطراف وقابل للتطبيق في البلدان التي يكون فيها الأشخاص ذوي الوضع الاجتماعي والاقتصادي المنخفض .نظرًا لأن الجزء المهم الذي يضمن هدف تحريك الأطراف الاصطناعية في كلا المستويين ، فقد بدأ النهج التجريبي لهذه الدراسة من استخدام نظام مفصل كروي ومشغل. تتألف الأطراف الصناعية المصممة من القدم والكرة ، ومحركين ، ونظام ربط و EMG ، وجميع هذه المواد كانت متوفرة في الأسواق المحلية في العراق. أظهرت النتائج التجريبية أن أقصى مدى للحركة لتحريك الطرف الاصطناعي المصمم في المستويين السهمي والأمامي وصل إلى 70٪ من نطاق حركة القدم الصحية بالنسبة لإشارات عضلة الساق لدى شخص سليم. كانت هذه النتائج تعتمد على العديد من العوامل مثل دقة الإشارة التي جاءت من مخطط كهربية العضل ، والتأثيرات المحيطة على أقطاب مخطط كهربائية العضل والقدم المستخدمة في الطرف الاصطناعي. ان الزوايا التي تمثل الحد الأعلى لمستويات الحركة والتي تم الحصول عليها بأتجاهات مختلفة لمفصل الكاحل كانت الأنثناء للأعلى $(35)^{o}$ والأنثناء للأسفل $(25)^{o}$ والأقلاب للداخل $(20)^{o}$ والأقلاب للخارج $(15)^{o}$.

الكلمات المفتاحية: الأطراف الصناعية، انثناء للأعلى، انثناء للأسفل، اقلاب للداخل، اقلاب للخارج ، مفصل الكاحل

1. INTRODUCTION

Artificial limbs have passed through many developments in the last decades. The major targets of these developments are to provide patients with limbs lighter and stronger by using materials with these properties, such as carbon fiber and plastic. Moreover, improving the artificial limbs' shapes is very important to the patient for their comfort and familiarity with society. However, one of the most important studies was on the joints. In the beginning, the mechanical joints were in one simple axis and depended on the strength of the amputee. Hydraulic joints were adopted in the middle of the nineteenth century to justify the joint's oil level. Bionic joints are the modern types of joint, especially the knee joint, by connecting it with the computer, which introduces moves like the healthy knee joint. Electronics contributed widely to the development of joints by combining them with the brain to get nerve signals from the brain. The lower limb amputation **(Wilson, 1969)** types depend upon the extent level of the amputee as below:

1. Hemipelvectomy prosthetics: likewise, designated hindquarter amputation, which included the excision of a portion of the pelvic bone

2. Hip Disarticulation (HP) prosthetics: Which is the mutilation of the limb would be at the hip level
3. Transfemoral amputation (the mutilation would be Above the Knee).
4. Knee disarticulation (KD).
5. Transtibial amputation (the mutilation is below Knee BKA).
6. Ankle Disarticulation.
7. Amputation of the foot: Mutilation is fulfilled at the mid-foot, which is named a Chop-art; mutilation close to the metatarsal is designated as a Lisfranc mutilation through the metatarsal bones is designated as a Trans-metatarsal amputation and mutilation of toe (s) excessive part of metatarsal bone(s) is designated Ray amputation a platform for weight bearing Toe amputation.
8. Disarticulation of Toe(s).

Prosthetics are the best replacement for missing limbs, and their manufacturing shape and parts depend on the extended level of the amputation **(Baptiste and Dokeh, 1999)**. However, these parts should bear weight and movements function using appropriate material. The previous studies of the advance for below knee prosthesis designs are summarized below:

(Park et al., 2014) designed and manufactured a controlled wearable robotic device powered by an actuator system for synthetic muscle activity. The feature of the designed device was providing the amputee with active assistance without shackling by the restriction in the ankle joint's natural Degrees Of Freedom DOF. **(Azocar et al., 2018)** introduced a design for an open-source leg and described the controlled performance in the time domain and frequency. The scalable robot in the ankle-knee prosthesis adopted the investigations of the controlling strategies. **(Yu et al., 2013)** designed and tested an actuator system that controlled the mechanism of a portable robot for knees, ankles, and feet and analyzed its biomechanical system. The important features of this design were the ability to justify the force by using a control actuator, the designed robot's lightweight, which is made of composite materials to manage portability, and the soft consolidated linkage mechanism of the design. **(Thatte, 2019)** studied the effect of prosthesis control techniques in maintaining balance using a powered robotic trans-femoral prosthesis. The objective of **(Bellman et al., 2008)** of contemporary prosthetics is to duplicate the transplanted limb or member's assignment most competently and unobtrusively feasible. **(Zeng, 2013)** designed an Ankle prosthesis to store and return energy to the amputee, allowing them to propel their body forward during push-off. **(Eilenberg et al., 2010)** described an adaptive muscle-reflex controller that employs an ankle plantar flexor composed of a Hill-type muscle with affirmative force feedback reflexes based on simulation experiments. The model's parameters were adapted to match the torque-angle profile of the human ankle acquired from level-ground walking measurements of weight and height-matched fit subjects walking at 1 m/s. **(Alleva et al., 2020)** introduced and fabricated a biomechanical design of the prototype of a unique power-driven ankle-foot prosthetics with energy stored within. **(Geeroms, 2011)** presented the modeling, designing, and developing of a variable stiffness mobilized knee-ankle prosthetics actuator that transfers energy from the knee to the ankle. This technique provides a decent estimate of the joint torques. A direct current motor was employed in this device to compress and retain a carbon-fiber ankle joint. **(Cimolato et al., 2022)** provided a comprehensive analysis of research on using electromyography (EMG)-driven control in lower limb prosthesis. **(Liu et al., 2018)** described the development of a novel robotic ankle rehabilitation platform designed specifically for hemiplegic patients recovering from strokes. **(Rusu et al., 2015)** determined the temporal fluctuation of force and torque in the human ankle joint during a single walking stride. In this search, two devices have been employed to measure the ground response force and the angular movement of the ankle joint. **(Au and Herr, 2008)** introduced a power-driven prosthetic design with a unidirectional spring constructed parallel with the controllable force actuator with a combination elasticity. **(Kokz et al., 2021)** designed and simulated ankle joints to mimic human walking on sloped surfaces. **(Mitchell et al., 2013)** presented the modeling, planning, and developing of a variable stiffness mobilized knee-ankle prosthetics actuator that transfers energy from the knee to the ankle. This technique provides a decent estimate of the joint torques. **(Jweeg et al., 2010)** studied and reviewed the prosthesis designs that streaked lower leg foot prosthetics that overcome these basic issues. A unidirectional spring is built in tandem with a force-controlled actuator with configuration flexibility in the prosthesis. **(Dipietro et al., 2005)** discussed a novel approach to stroke rehabilitation using an interactive robotic treatment called EMG-triggered therapy. **(Sawicki et al., 2006)** This study explored the impact of powered ankle-foot orthoses (AFOs) on individuals with incomplete spinal cord injury (SCI) during walking. **(Dickinson et al., 2017)** reviewed the state of the art work in residual limb finite element analysis published since 2000.

(Brockett and Chapman, 2016) proposed a design of an ankle joint to help people with single-limb trans-tibial amputations adapt to their surroundings while retaining normal gait patterns and stability. **(Crenna et al., 2018)** studied a Randomized Withdrawal design to investigate a unique experiment design for arranged medical instrument studies. **(Dundass et al., 2003; Li and Bhuiyan, 2020)** designed and developed a rotator ankle joint, allowing Amputee to squat and kneel easily. **(Alam et al., 2014)** studied the literature on articulated ankle-foot orthoses developed for drop-foot therapy. **(Baker, 2019)** discussed the assignment of the designed novel for an orthotic of the ankle joint to handle drop-foot for the individual's post-stroke. **(Kapti and Yucenur, 2006)** analyzed the walking gait cycle pattern for walking, standing, sitting, stairs walking, and walking backward. The major part of this prosthesis actuator of knee joint functions of flexion and extension with a motor. **(Amiot et al., 2017)** designed and tested an actuator system that controlled the mechanism of a portable robot for knees, ankles, and feet and analyzed its biomechanical system. **(Dawe and Davis, 2011)** designed a computer-controlled system suitable for typical ankle-foot prosthetics to trap and postpone the release of absorbed energy from the ankle until certain intervals in the gait cycle. **(Lecomte et al., 2021)** discussed a method for comparing the biomechanics of walking (gait) in individuals with trans-tibial (below the knee) amputations and proposed a novel robotic ankle rehabilitation platform (RARP) to help patients do ankle exercises. **(Grabowski and D'Andrea, 2013)** presented a case study where an EEG-based brain-computer interface successfully controlled a lower-limb prosthesis. In the literature review, many studies of prosthetics with different designees with different techniques and mechanisms had been reviewed, and most of these studies focused on the effectiveness method for prosthetics. The previous review discussed the theoretical and experimental searches that designed and simulated a prosthetic. Controller's techniques were utilized and applied via time domain and frequency. Others discussed the various moment measurement methodologies in the knee or ankle joints, while others studied more than one joint. The ankle joint with a powered mechanism will be designed to simulate the movements in both the dorsiflexion and plantar flexion planes and the inversion and eversion planes. The methodologies employed in EMG-driven controllers to translate user intention and volition to high-level control parameters have been grouped into direct control, pattern recognition, and model-based. Each class adopts a different principle to interpret user volition and intention from EMG signals and translate it into a high-level control of the EMG-driven controllers classified types shown in **Fig. 1.** This study aimed to design a low-cost and lightweight adjustable prosthetic limb with 3 Degrees of Freedom DOF at the ankle joint, powered by EMG for a subject with transtibial amputation.

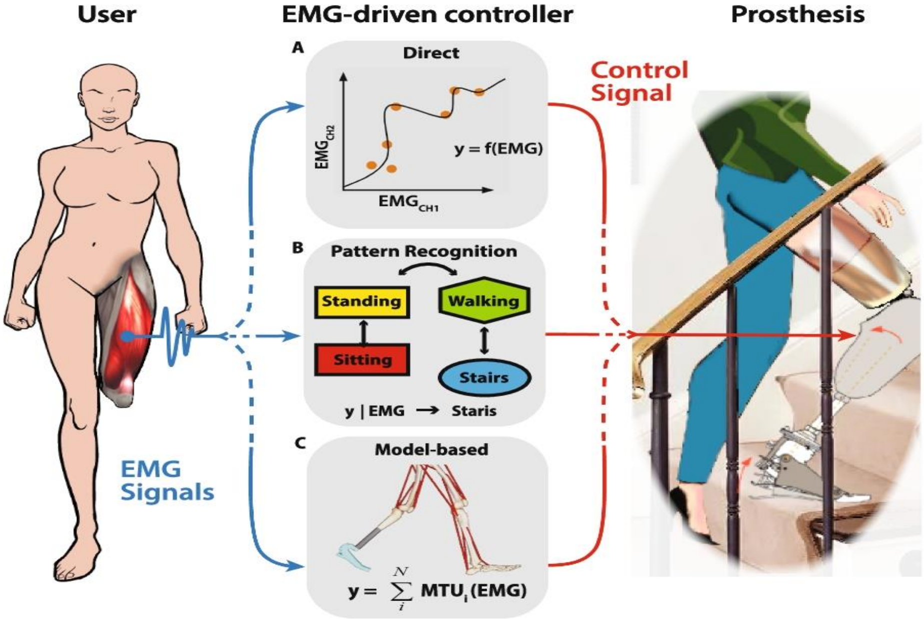

Figure 1. The classified types of EMG-driven controllers **(Cimolato et al., 2022)**.

2. EXPERIMENTAL WORK

The experimental work has been explained in detail to clarify the intelligent powered ankle-foot design, its limitations, and the basics that led to this design. The designed prosthetic consisted of three major parts: 1) prosthetic foot and mechanism. 2) Stepper motors. 3) sensors system. **Fig. 2** shows a CAD model of designed ankle-foot model components.

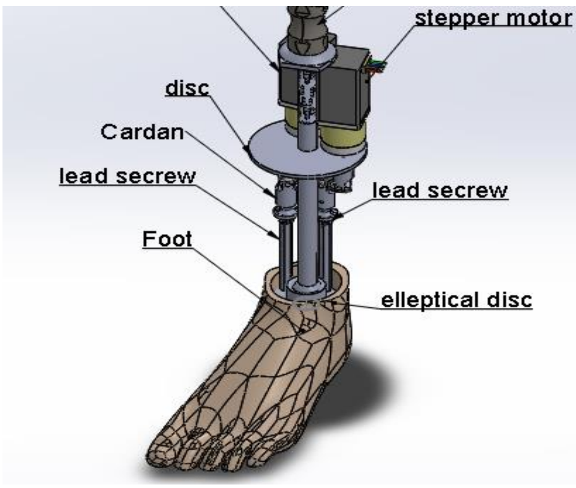

Figure 2. CAD model showing the components of the design ankle-foot model.

3. PROSTHETIC FOOT AND MECHANISM SYSTEM

Articulated Multi-axis foot was chosen in this prosthetic since it's available adjustable, and its price is low compared to both above foot manufacturing processes. The size of the used feet in the experiment was 26 cm, and the diameter of the screw was 10 mm ends with the adapter, whose basic function was to pin the foot with the pylon.

3.1. Ankle Joint Mechanism

The mechanism of the ankle joint depends on the ball and socket to get movements range of motions as possible, similar to the ankle joint **Fig. 3**. The mechanism of the ankle joint consists of the following parts:

Figure 3. Front view of moving in plantar flexion plane.

3.1.1 Elliptical Plate

The size of the foot made the plate elliptical in the radiuses respectively, and it has been extruded in the center with a circle radius of 15 mm and two circles with a 2.5 mm radius about 5 mm from the edge of the plate with an angle between these two circles 90º. This plate is fixed in the foot, and its major task is to stabilize the suspension ball in the central extruded circle and two screws in the small extruded circles.

3.1.2 Suspension Ball

The range of motion in the ankle joint was in 3 planes, dorsiflexion-plantar flexion plane, and its angles in these axes were 40º and 30º, respectively. In the eversion-inversion plane, the angles were 20º and 30º, respectively. In the adduction-abduction plane, the foot rotates from the center line about 20º and 30º, respectively. Despite the suspension ball's ability to rotate in the adduction-abduction plane, the motion in this plane is neglected since the hip joint can rotate in these axes, and the designed prosthetic target was the below-knee amputee. The major specification in the designed prosthetic was that the Articulated Multi-axis foot screw adapter size from the side of the socket and the screw through the Articulated Multi-axis foot were 14 mm which made the suspension ball must be the same size to connect it with the socket. The suspension ball used in this framework was made of steel and welded in the bottom of the suspension ball with a stud diameter of 14mm to set it in the foot. The part of the ball connects with another plate from the upper part welded with stud 14mm, and its length was 100 mm

3.2 Linkage Systems

The linkage systems were two; however, the basic part of the linkage in both systems was a nut with a 6 cm length fixed by a screw with 10mm to the elliptical plate and another screw with a 6 cm length. Two screws with 6 cm length were welded with incompressible spring with 5 cm length, and another two screws connected with three sections swivel joint connector. The swivel joint wasn't providing the ankle joint with the appropriate displacement for the motor to move the ankle with the range of motion, like the incompressible springs.

3.3 Circular and Squirt Plates

The upper part of the ball was fixed by a nut to the plate used to stabilize the servo motors, which provide the ankle-foot with the range of motion as explained below. These plates were extruded in the center 14mm circle and many other circles to fix the motors above the plates. The dimensions of the circular plate were 2 cm in radius, and the square plate was 4*4 cm in length. However, the circular plate has a 3mm thickness, and the square plate is 4mm.

3.4 Stepper Motor

DC motors that rotate in exact increments or "steps" are known as stepper motors. It is a good choice when the required position is needed in something precisely. Steppers have been used in 3D printers to position the print head accurately and CNC machines to position the cutting head accurately. A stepper motor will likely be used if the digital camera has autofocused or remote zoom. **Fig. 4** shows the stepper motor connected to the linkage system. Stepper motors, unlike DC motors, are controlled by applying DC electrical pulses to their internal coils. Each pulse moves the motor one step or a fraction of a step. The latter is referred to as "micro-stepping."

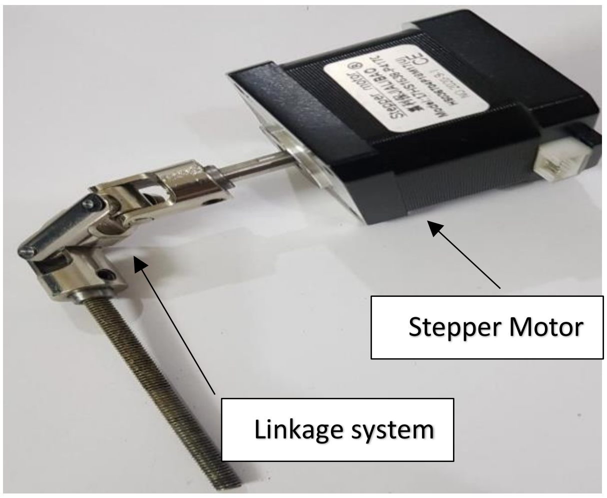

Figure 4. The Stepper Motor connected to the linkage system.

The stepper motors used in the experimental approach specifications were summarized in the below:

1. Brand: GKTOOLS
2. Model: 17HS1538-P4170
3. Motor length: 40mm

4. Diameter of axle: 5mm
5. The front shaft length: 20mm
6. Static torque: 2.2N.cm
7. Size: 42mm*42mm*40mm
8. Certificate: CE, ROHS
9. Field: Widely used with laser engraver machines, 3D printers, monitor devices, medical/textile/packaging machines, stage lighting, and ECT.

While the mechanization parameters were as summarized below:

1. Step Angle: 1.8°
2. Insulation Class: B
3. Phase: 2
4. Voltage: 3.12 V
5. Current: 1.7 A.
6. Resistance: 1.5 Ω.
7. Inductance/Phase: 2.8 MH.
8. Holding Torque: 40 N.cm.
9. Mass: 0.28 kg.

The design concept is to convert the rotary motion to linear from the stepper motor to the lead screw strategy. A lead screw is a screw turned while a nut attached to the screw is prevented from spinning. This technology allows quick and powerful movements while maintaining a low sound volume. Also, the technique enables for more exact placement than toothed belts. However, the number of screws thread tightened per cycle can be calculated by measuring the distance traveled by the thread. The motor is 1.8° per step. This is (360/1.8) or 200 steps/rev. The distance measured was 4 mm every 360° or 200 steps.

3.5 Electromyography (EMG)

Nerves use electrical signals called impulses to control the muscles in the body. These signals cause the muscles to respond in predictable ways. Muscles can react abnormally as a result of nerve and muscle disorders. The electrical activity of muscles at rest and during contraction is measured by an electromyogram (EMG). The technical specs used by the UNO Arduino in the experimental work are shown in **Table 1**. According to SENIAM recommendations for describing the electrode location and orientation and describing the starting posture and clinical test for recording the SEMG of that particular muscle, the electrodes fixed on the lateral gastrocnemius muscles. The recommendations for the individual muscles are organized according to the body parts in which the muscles are located in the Intelligent Bio-feedback, an EMG-employed sensor to measure a muscle's filtered and rectified electrical activity. The sensor's output is a proportional voltage proportionate to the amount of activity in the specified muscle. The EMG sensor used in the search experiment was the ECG/EMG biofeedback Shield **Fig. 5**. The ECG/EMG biofeedback Shield enables Arduino-compatible boards to capture information from electrocardiography (ECG) and electromyography (EMG). This shield allows bio-feedback signal experimentation. The heartbeat can be monitored, and motions can be recognized by monitoring and analyzing muscle activity.

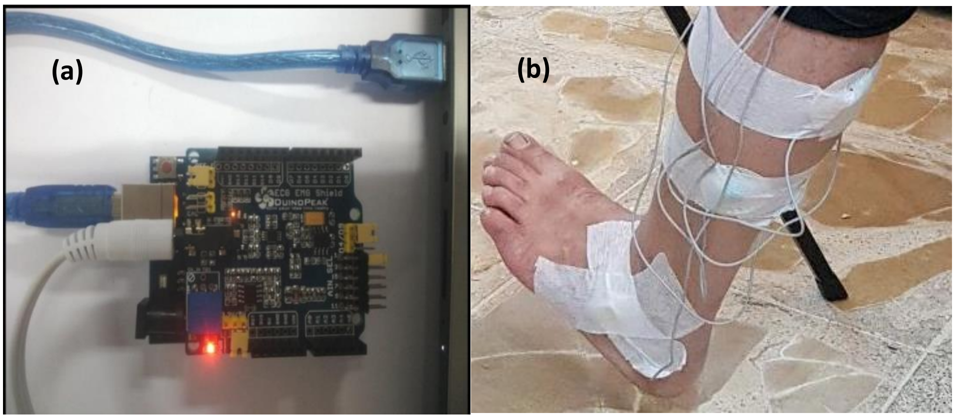

Figure 5. (a) Arduino UNO board connected to the PC, (b) ECG/EMG biofeedback Shield sensors

Table 1. Technical specs of the UNO Arduino.

Microcontroller	ATmega328P
Operation Voltage	5 V
Input Voltage (recommended)	7-12 V
Input Voltage (Limit)	6-20 V
Digital I/O Pins	14 (of which 6 provide PWM output)
PWM Digital I/O Pins	6
Analog Input Pins	6
DC Current per I/O pin	20 mA
DC Current for 3.3 V pin	50mA
Flash Memory	32kB (AT mega328P), of which 0.5 kB used by bootloader
SRAM	2 kB (ATmega328P)
EEPROM	1 kB (ATmega328P)
Clock speed	16 MHz
LED_Builtin	13
Length	68.6 mm
Width	53.4 mm
Weight	25 g

4. EXPERIMENTAL WORK PROCEDURE AND RESULTS

The experimental tests and the major results for the designed prosthetic have many parameters that affect the resulted movements, and those parameters are summarized in the following points:

1. The tests were examined by pinning the electrodes on the lateral gastrocnemius muscles of a male 50 years old, his weight was 88 kg, to investigate the ability of the moving mechanism in the designed prosthetic.
2. The major parts of the powered prosthetic are the activity muscles' signals, which have to be translated to the prosthetic to move the foot in different planes according to these signals. However, it should be clear that EMG electrodes have been pinned on the gastrocnemius muscles since it is the largest muscle in the leg, and the activity signals are

very clear in these muscles. Those signals represented about 70% of the ROM of the healthy foot muscle activity, which was clear by comparing the angle of the moved prosthetic with the healthy foot by moving the ankle in the maximum range that could be moved, according to **(Liu et al., 2018)**. The EMG electrodes were fixed on the lateral gastrocnemius muscles, and sensor placement procedures and signal processing methods for SEMG were fixed according to **(SENIAM). (SENIAM)** is an European concerted action in the Biomedical Health and Research Program (BIOMED II) of the European Union). It should be noted that these translated signals varied according to the person using the electrode and the effects on the signals of the muscle's activity sensor (**Fig. 6**).

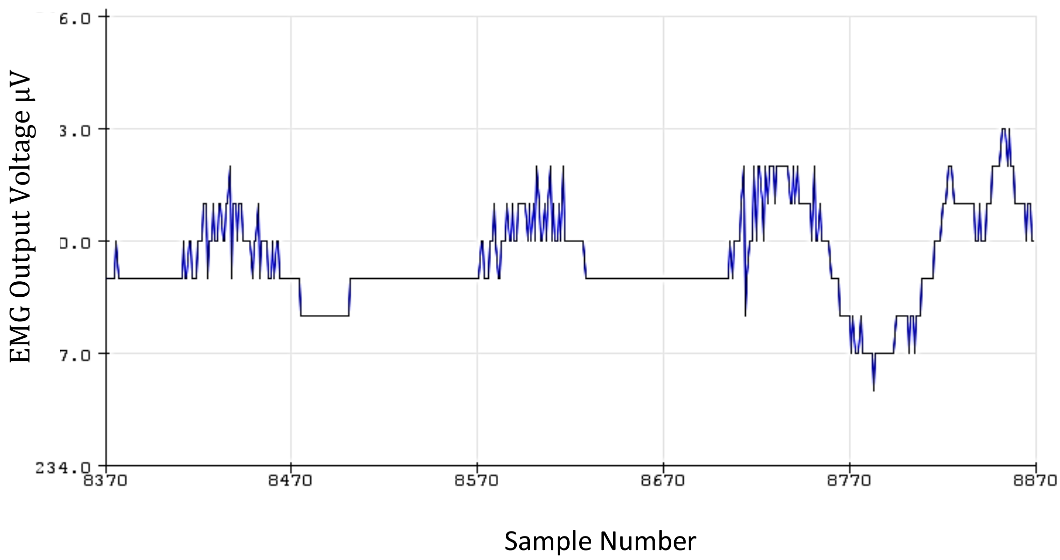

Figure 6. EMG signals.

3. The EMG sensor, connected with the prosthetic and its mechanism system, was verified with the Arduino that controls the motor's speed and its rotation direction that moves the ankle joint in the planes of movement. The system's mechanism consisted of the screws moving up and down to rotate the system in a different direction. It should be taken into consideration that the speed of the motors was constant, and to control the signals of the muscles' activity speed, different sensors should be used. Therefore, the tests were considered to be at constant speed. The Experimental work of the designed prosthetic flow chart is shown in **Fig. 7**.
4. The designed prosthetic moved in two planes by moving the screws connected with the motors; since it moved upward and downward relativity. These moves depended upon the reads of the muscle's activity, which means there was no such a constant move every time the reads of the EMG sensor gave an output move different from the previous movement because its input signals read from the healthy leg and couldn't be controlled.
5. The mechanism of the designed prosthetic itself was also restricted by the length of the screw and the foot used in the prosthetic since the foot used in the experimental tests was available in the markets made by OTTOBOCK Company, and its number was 2529 with 26 cm size. The flexibility of change in this foot was very difficult because it could affect the foot's strength. However, the results could be improved if the foot was made and designed to be compatible with the designed mechanism and sensors.

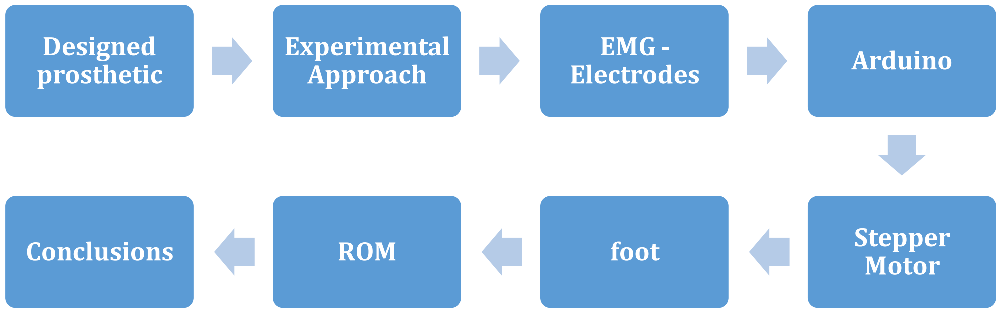

Figure 7. Experimental work of the designed prosthetic flow chart.

5. CONCLUSIONS

Using residual limb EMG signals to control the ankle position of an active ankle-foot prosthesis holds promise for improving the functionality and adaptability of prosthetic limbs. However, technical challenges must be addressed to ensure accurate and reliable control in real-world scenarios. From the experimental work, the following conclusions are summarized below:

1. The search experiment showed the ability to move the designed prosthetic in different planes depending on the nerve signal from the muscle center.
2. The effectiveness of the moves in the different plans depended upon the signal that came from the activity of the muscles sensed by the EMG sensor used in the tests, which was very sensitive and affected by external movements as a result of the moving angles of the designed prosthetic.
3. The moving in the different angles of the sagittal and frontal planes at the experimental part of the search was noticed that the resulting angles were 70% of the ROM of the healthy moving foot in the same planes.
4. Thus, the dorsiflexion angle in the experimental part reached about 35˚ maximally, the plantar flexion angle in the experimental part reached about 25˚ maximally, the inversion angle in the experimental part reached 20˚ maximally, and the eversion angle in the experimental part reached to 15˚ maximally.
5. The moving parts in the different planes were relatively restricted by the foot chosen in the search since it's heavy and inflexible. Besides, not all the muscles' activity could be translated from the electrode to the EMG shield according to the surrounding effects and the efficiency of translating the signals from the EMG shield to ARDUINO.
6. The stepper motor could also affect the resulting moves according to its speed, efficiency, and the mechanism connected with the motors.

Several recommendations could be introduced for future work, which can be summarized below:

1. Study the effect of using different feet or design a carbon steel foot that could be integrated with the ball joint in a design that gives more flexibility in moving the foot with different planes.
2. Study the effect of using different muscle activity reading sensors, which gives a more accurate reading of the muscle's activity.
3. Compatible prosthetic with the pylon and the socket to test with an amputee patient to study the relaxation and flexibility in movements supplied to the patient.

4. Programming the Arduino to study the effect of using the moves in two planes simultaneously, which causes different stresses on the ball joint.
5. Study the effect of using different materials of the same ball joint with a more lightweight material, giving the patient more comfort.

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