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Design, Development and Evaluation of Novel Force Myography Based 2-Degree of Freedom Transradial Prosthesis

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ABSTRACT Limb loss is a traumatic event as it has physical and psychological effects on an amputee. Recent advancements in mechatronics and biomedical engineering have resulted in development of dexterous myoelectric prostheses for rehabilitation of amputees. In addition, evolution in manufacturing and sensing technology presents ample room for improvement in mechanical design and control system of prostheses to enhance amputee experience while using prosthetic devices. The present study is focused on design of a novel and cost-effective externally powered two-degree-of freedom prosthesis for assisting amputees to switch from body-powered devices to externally powered prosthesis. The control system of the developed prosthesis is based on the muscles signals acquired through force myography (FMG) technique. For precise integration of force-sensitive resistor (FSR) inside the socket to measure muscle activity, a stand-alone housing for FSR was designed with the feature of mechanical adjustment to control sensitivity of FSR and auto-calibrate its threshold to meet the requirements of individual amputees. The housing was designed to handle the fabrication inconsistencies during socket shaping process and thus ensure that sensor is in-firm contact with the muscle to sense volumetric changes. The developed mechanical design and FMG based muscle acquisition technique was successfully tested on a transradial amputee and extensive experimentation was performed for characterization of the prosthesis. FMG signal for various gestures was successfully extracted from muscles of the amputee to control the prosthesis according to the developed control technique. The results suggested that integration of FSR in the socket has significantly reduced the effect of sweat and volumetric changes on the performance of the sensor. Due to its novel design, embedded features, and cost-effectiveness the developed prototype holds the promise to be successfully commercialized to assist transradial amputees in becoming active citizens for contributing towards socio-economic growth of their country.

INDEX TERMS Electromyography, force myography, transradial, prosthesis, active actuation, muscle pressure, force-sensitive-resistor.

I. INTRODUCTION

Limb amputation for example hand loss is a disturbing event not only for amputees but also for their families, as it limits the patient's ability to perform Activities of Daily Living (ADL). According to World Health Organization (WHO), 15% of the world population is suffering from some sort of disability [1], with limb amputation as one of the major contributor to this figure in making it a global issue. Major causes of limb amputation include accidents, peripheral vascular diseases, tumors, infections, diabetes and congenital conditions [2], [3]. Public Health England (PHE)

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reports 27,465 amputations between 2015 and 2018, compared with 24,181 cases between 2012 and 2015 - a rise of 14% [4], whereas, National Health Service revealed, 17,845 upper limb amputees in Scotland between 1981 and 2013 [5]. Similarly, Ziegler-Graham *et al.* [6] estimated 41,000 upper limb amputees in the United States and expected this number to be doubled by 2050. Among limb amputees, 60% have transradial amputation [7], therefore, the robotics and bio-mechatronics community is contributing significantly to rehabilitate and create a barrier-free environment for them by developing technologically advanced prosthetic devices.

Transradial prostheses can be categorized as passive, body-powered and externally powered devices [8]. Passive prostheses are used to only restore the cosmetic appearance of the amputee, whereas, body-powered grippers presents a cost-effective solution for rehabilitation of amputees and greatly assist them in performing activities of daily living (ADLs) including grasping abstract objects, pouring liquid in a glass and turning the door handle, however, discomfort caused by the harness, unappealing cosmetic appearance, fatigue and excessive motion or power required to operate body-powered prostheses keep them far from being an ideal solution and results in more than 50% rejection rate of these devices [9]–[11]. To address the limitations of the body-powered solutions, externally powered prostheses have actuators powered through batteries contained inside the system for assisting amputees in performing ADLs. However as reported in the literature, externally powered transradial prosthesis provide less feedback to the amputee and lack intuitive and dexterous control. Furthermore, a single-degree-of-freedom externally powered gripper costs more than $6,000 making these devices beyond the affordability of the amputees [12].

This high initial and maintenance cost severely affects the interest of amputees in externally powered devices. WHO reported that affordability of health services, lack of technical facilities, expertise and skills for providing sufficient training and lifelong care to amputees is a significant barrier in acquiring externally powered prostheses [1], [13], [14]. In developed countries, various government, non-government, and health insurance organizations assist amputees in acquiring these devices, whereas, developing and third world countries lack financial resources and therefore, amputees may not receive financial and technical support from the government. Therefore, the National Academies of Science, Engineering and Medicine reported that only 27.6% of the upper limb amputees are using externally powered devices [15], whereas, remaining amputees are either not availing prostheses or using body-powered solutions due to technical and financial constraints [16]. Hence, externally powered grippers are highly uncommon in developing countries. However, due to rapid increase in the rate of amputation because of the aging population and security conditions, the need for active prosthetic limbs having less initial and maintenance cost along with an easy-to-use control interface is growing to eliminate barriers in the acquisition of externally powered devices [17].

The state-of-the-art control system of externally powered prosthesis utilizes Surface Electromyography (sEMG) for measuring neuromuscular activity and extracting amputee's intent for controlling prosthesis [18]–[20]. However, the signal extracted from these electrodes is unstable with amplitude ranging from hundreds of micro-volts to few milli-volts only [21]. Furthermore, sEMG electrodes are highly sensitive and require excellent skin contact to minimize calibration issues that may arise due to slight limb movement and hairy/sweaty skin [22]. The variable nature of these electrodes due to electrode shifts, motion artifacts and crosstalk among deep adjacent muscles have also been reported in the literature [23]–[25]. Therefore, signal filtration including various pre-processing operations such as noise removal and averaging of EMG signal is necessary to extract meaningful signals which in-turn may result in the following two issues: a) Pre-processing procedures accumulate time and causes a delay between onset of muscle activity and motion actuation b) pre-processing procedures require advanced filtering hardware and software techniques that adds up to the cost of EMG electrodes.

Therefore, there is a need to explore novel sensing mechanisms for controlling upper limb transradial prosthesis [26], [27]. Ultrasonic imaging (US) is one of the techniques that can detect muscle activities [28]–[31], however, it has lower wearability and is sensitive to arm/hand displacement [32]. Other researchers have explored Mechanomyography (MMG) that uses oscillations of frequencies ranging from 5 to 100 Hz to detect deep muscle activities [33]. However, like US imaging, MMG also lacks essential features for integration in commercial products.

Forcemyography (FMG) is one of the other potential candidates to replace sEMG in transradial prostheses. FMG consists of Force Sensitive Resistors (FSRs) which produce changes in resistance upon application of force. FSR can be used to sense the activity (contraction/relaxation) of muscles by measuring the pressure difference between residual limb and the socket. Since, FSR has the potential to be integrated in the socket, therefore, several studies have explored the possibility of controlling transradial prostheses through FSR.

The application of FMG for controlling prosthesis was first introduced by Yungher and Craelius [34], as they successfully generated topographic maps of the pressure exerted by the muscle against the socket. After their findings, many researchers have explored applications of FMG in transradial prostheses. Ferigo *et al.* [35] evaluated FMG for controlling prosthesis in both static and dynamic scenarios. They developed an array of 80 FMG sensors placed inside a conventional socket fitted to a transradial amputee and collected the data for various static and dynamic gestures. Although, they successfully achieved an accuracy of 81% and 75% for static and dynamic classification of six grips, however, the study has potential limitations including complicated electronics and control system along with high computational time for classification of gestures which is not considered by the authors. Furthermore, the FSRs are directly mounted inside the socket which may result in reduced accuracy due to bending of the sensors, hairy and sweaty skin [36].

The authors in [37] identified that sEMG electrodes have relatively low accuracy to control advanced prosthetic devices. Therefore, they performed a feasibility study to investigate the potential of FMG to acquire muscle signals for classifying various grips of Bebionic-3. Four transradial amputees participated in this study and accuracy of 70% was achieved even though only static gestures were recorded for classification purposes.

To compare the performance of EMG and FMG for controlling prosthetic devices, the authors in [38] collected data of various grips from intact subjects. Their findings highlighted that FMG is more stable than EMG and has a better accuracy in grip classification. They proposed that fusion of data from FMG and EMG may be necessary to obtain the best results. However, since the developed control schemes were tested on intact subjects only, therefore, the study has little significance. Other studies [39]–[42] also explored the potential application of FMG in controlling advanced prostheses. However, the majority of these studies overlooked the integration of FMG in the socket and placed FSRs directly in contact with the skin.

Based on the discussion of state-of-the-art, the research problems along with the contribution of this study are summarized below:

- Direct mounting of multiple FSRs in the socket is successfully demonstrated in previous studies including [35], [41], however, this approach requires precise integration of socket with FSRs to ensure that during the functioning of prosthesis sensor is in firm contact with the targeted muscle and the socket to measure volumetric changes. Any inconsistency during the shaping of sockets may result in a gap between the socket and the skin resulting in degradation of FSRs performance. It is worth mentioning here that the shaping of sockets is already a persistent challenge in developing countries [43]. Therefore, this research work presents a novel mechanism for integrating FSR in the socket to handle fabrication inconsistencies during socket manufacturing process and minimize the impact of socket loading.
- Apart from the shaping of the socket, direct placement of FSR in the socket also causes bending of the sensor film and further exposes it to sweaty and hairy skin. These factors result in degradation of the performance of FSR and affects its accuracy [36]. Therefore, the authors have developed a stand-alone sensor housing to minimize effect of sweaty/hairy skin and avoid bending of FSR while the amputee is performing activities in different orientations of prosthesis.
- The affordability of prosthetic devices is also a significant barrier to the rehabilitation of amputees. Therefore, an externally powered prosthetic gripper with a detachable wrist unit is designed and manufactured for assisting body-powered amputees to switch to externally powered solutions.

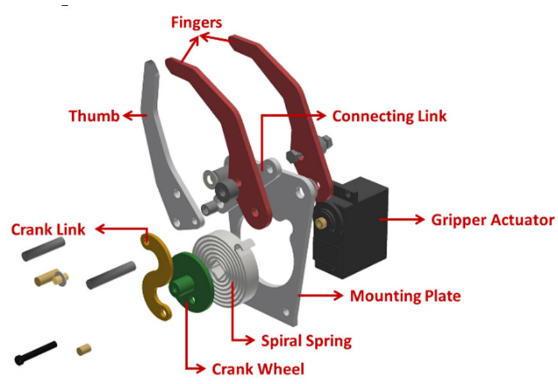


FIGURE 1. Exploded view of gripper.

II. MECHANICAL DESIGN OF AUTOMATED PROSTHESIS

The developed prosthesis is a two-degree-of-freedom device comprising of a gripper and a wrist unit. The design of the gripper is inspired by existing body-powered solutions that are modified to replace the function of the Bowden cable for opening and closing of the gripper with an actuator. The gripper consists of an index finger, middle finger, thumb, and actuation mechanism as illustrated in fig 1. The index finger and the middle finger acts as a single unit to form finger subassembly that is connected to the thumb via a connecting link. The thumb in turn is coupled to a crank link which acts as a power transmission device between the crank wheel and thumb. The crank wheel is directly mounted on the shaft of a rotary actuator. The activation of the actuator rotates the crank wheel resulting in a push on the crank link which moves finger subassembly and thumb towards and away from one another (opening/closing of the gripper). The crank wheel is also pre-loaded with a spiral spring to assist in providing constant force while gripping the object. Savox SB 2230 SG servo motors was preferred for the actuation of gripper because of its low cost, high torque and precise position control.

Although, several servo-controlled grippers are available commercially including [44]–[46], however, these devices are suitable for robotic applications such as pick and place and does not incorporate the complexity of human hand anatomy. Furthermore, servo actuated 3D-printed prosthetic grippers are also reported in various studies including [47], yet they are not successfully commercialized as the application of these grippers in rehabilitation of upper limb amputees is limited since they lack cosmetics and their mechanical characteristics such as maximum pinch force, maximum opening, etc. do not fulfill the criteria of transradial prosthesis. In contrary to these prostheses, the cosmetic appearance of the proposed gripper matches closely with body-powered solutions and the mechanical design parameters such as maximum opening, gripping force etc. are according to the recommendation of Belter *et al.* [48], hence, compared to existing servo-controlled grippers it will have higher acceptance rate.

The gripper sub-assembly is connected to a one-DOF detachable wrist unit for pronation and supination

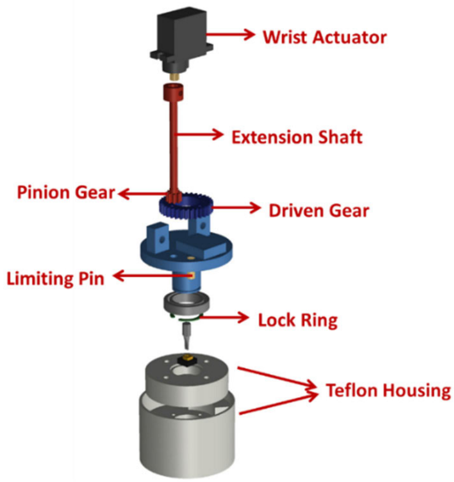
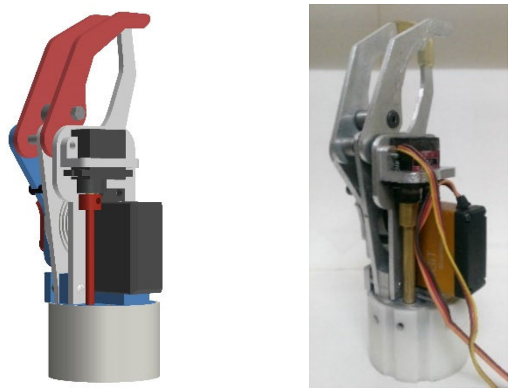


FIGURE 2. Exploded view of wrist unit.



(a). 3D CAD Model **(b): Manufactured and Assembled Prosthesis**

FIGURE 3. (a). 3D CAD model. Fig 3(b): Manufactured and assembled prosthesis.

movements. The exploded view of the developed wrist unit is presented in fig 2. The wrist actuator is incorporated inside the gripper's body and connected to a pinion gear through an extension shaft. The pinion gear meshes with a stationary-driven gear. When the actuator is powered the orbiting of pinion around the driven gear results in the rotation of the gripper subassembly. Although the wrist unit has a 360° range of motion, however, for constraining rotation within a safe limit, mechanical limiting pins were embedded within the wrist unit. Power Pro MG90S servo motor was used for the actuation of the wrist unit.

A detailed review of state-of-the-art in artificial wrists is presented by Bajaj *et al.* [49]. From this study it is evident that artificial wrists for rehabilitation applications are incorporated at the distal end of the terminal device or gripper, however, this may result in increasing the overall dimensions of the prosthesis. Considering the restrictions due to long residual limb lengths and significant variation in level of amputation, designing a prosthesis that occupies much of the forearm to house and drive the wrist unit is impractical.

In response, the proposed design of the wrist unit is unique as the wrist actuator is incorporated inside the gripper's body to reduce length of wrist subassembly for facilitating amputees with long residual limb lengths. A comparison of the developed wrist unit with the existing artificial wrists is presented in table-1. The length and diameter of active 1-DOF wrist actuators ranges between 57 to 60 mm and 40 to 47 mm respectively, whereas, the wrist unit developed in this study has a length of 35 mm and diameter of 45 mm respectively. The reduction in the length of the wrist unit facilitates amputees with long residual limb to assist them in restoring functionalities of their wrist joint.

The 3D-model and assembled prosthesis is presented in figs 3 (a) and 3 (b) respectively. For the wide acceptance of the prosthesis, it must be lightweight having enough strength to withstand various forces while performing ADLs [50]. Therefore, after considering various materials, the fingers and main body of the prosthesis were manufactured from Aluminium-7075 due to its high strength-to-weight ratio, toughness and resistance to fatigue. The housing of the wrist unit was manufactured from synthetic polymer as it only acts as an enclosure for pinion and stationary driven gear.

TABLE 1. Comparison of the developed wrist unit with the state-of-the-art commercial and research based one-DOF artificial wrists.

S. #	Name	Length (mm)	Diam. (mm)	Weight (g)	Range of Motion	Torque (Nm)
1.	MC wrist rotator [49]	70	47	143	360	1.13
2.	Manus Hand [51]	----	----	---	170	2
3.	Zinck *et. al* [52]	65	40	87	360	0.06
4.	TB Supro Wrist [49]	57	----	154	---	----
5.	**Proposed Design**	**35**	**45**	**100**	**360**	**0.88**

The next section discusses FMG based technique for acquiring muscle signals of an amputee to control the prosthesis.

III. MUSCLE SIGNAL ACQUISITION

For acquiring muscle signals of the amputee to control the two-DOF prosthetic limb, the authors opted Force Myography (FMG) technique. The option of direct placement of FSR on the residual limb is not effective as it may cause displacement and bending of the sensor during wearing of the socket and operation of the prosthesis, which may result in the unpredictable measurement of muscle pressure. Therefore, the authors have developed a novel mechanism for precise integration of FSR capable of handling the fabrication inconsistencies during the shaping of the socket.

The exploded view of the developed housing is shown in fig 4. FSR-408 from Interlink Electronics having a length of 609.40 mm (24 inches) and force sensitivity from 0.2 N to 20 N [53] is used for signal acquisition. For the best fit in the

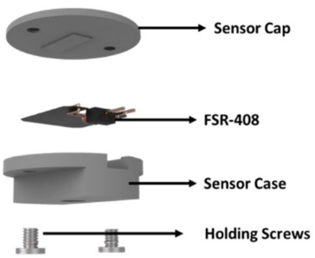


FIGURE 4. Exploded view of force sensitive resistor housing.

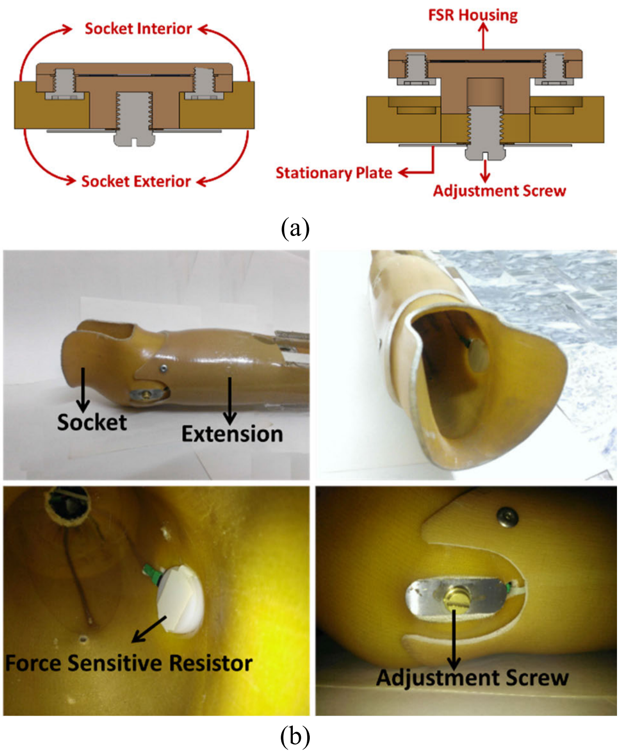


FIGURE 5. Sensor casing along with adjustment screw mechanism integrated in the socket.

casing and hence the socket FSR sensor was reshaped into a rectangular strip having an area of 5.08 mm^2. Inside the sensor case, FSR film was placed on a flat surface through double-sided laminating adhesives, whereas, in the sensor cap there is a rectangular step on top of FSR which ensures that force/pressure is applied uniformly on the sensor film. The sensor cap and its case were fixed through holding screws to form a single body and were manufactured from high strength polymer such as Teflon as it is light-weight and easy to machine. Since, FSR is placed inside the sensor case, therefore, during wearing and operation of the prosthesis, the bending of sensor film can be easily avoided and impact of hairy/sweaty skin on the performance of FSR can be minimized.

The sensor housing was attached to the socket using an adjustment screw mechanism as shown in figs 5 (a) and (b). Since, the screw is stationary with reference to the socket,

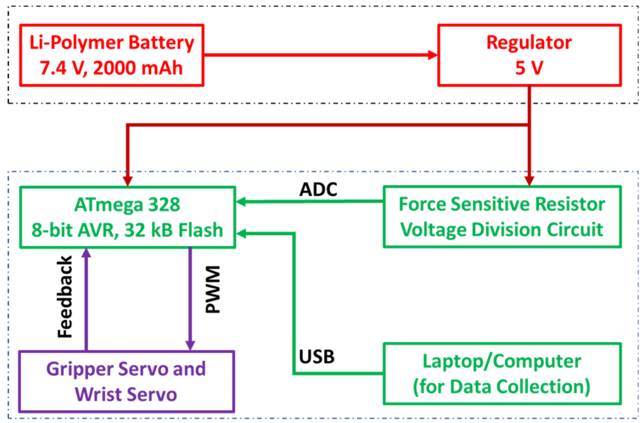


FIGURE 6. Block diagram of embedded system of the developed prosthesis.

therefore, screwing/unscrewing operation results in linear motion of the sensor housing. The linear adjustment mechanism greatly reduces the complexity in the shaping of the sockets during manufacturing to integrate FSR casing in it.

As an instance, if there is a gap between the sensor housing embedded in the socket and targeted muscle due to various fabrication inconsistencies during the socket manufacturing process, then unscrewing operation will result in moving the sensor case towards the forearm to compensate for the gap. The linear adjustment mechanism also regulates the sensitivity and threshold of the FSR to better suit the muscle strength of amputees. For example, if it requires large effort from the muscles of the amputee to cross the set threshold of FSR, then the position of the sensor can be adjusted by providing a single screw to move it closer to the stump resulting in a reduced effort to operate the prosthesis.

The next section discusses embedded system architecture for the proposed two DOF prosthesis.

IV. EMBEDDED SYSTEM ARCHITECTURE

An overview of the embedded system architecture for the developed prosthesis is shown in fig 6. The power from two cell 7.4 V 2000 mAh custom lithium-ion polymer (LiPo) battery was supplied to a DC-DC step-down LM-2596 voltage regulator. The power regulator ensures the supply of a constant voltage level of 5 V at which servo motors and micro-controller unit can operate efficiently. Arduino Pro Mini was used as a Microcontroller Unit (MCU) due to its small footprints and low power consumption. The MCU performs two-way communication as it detects amputee intent when the force exceeds a pre-defined threshold and then actuate and monitor servo motors. A current feedback sensor reported in [47] was used to monitor the current drawn by the gripper. When the gripper grasps an object, it draws more current and accordingly the motor was stopped.

Since Arduino pro mini requires a separate programmer for updating firmware, debugging code, and calibrating sensors, therefore, PL2303 USB to RS232 TTL converter was interfaced with it to establish communication between the

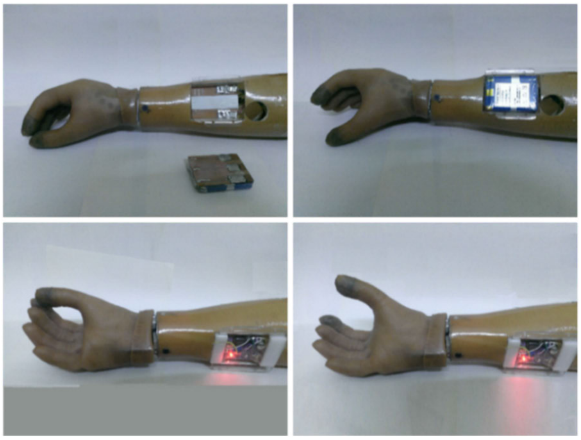


FIGURE 7. Prosthesis fitted in the socket along with battery and embedded system.

host device such as a computer and the microcontroller. This two-way communication apart from transferring the program to MCU also allow for the data logging to analyze useful information about the change in resistance of FSR in terms of voltage, finger/thumb position, etc.

For force to voltage conversion voltage division circuit recommended in the datasheet of FSR-408 was used. The value of the measuring resistor (R_M) was chosen to be 10 kΩ because in this configuration the sensitivity of FSR is relatively good in detecting changes in the applied force. Op-Amp (LM-358) was used to reduce the error due to the source impedance of the voltage divider. The output voltage of FSR can be predicted using equation 1. When the amputee contracts his residual muscles, it results in the application of pressure on the FSR due to which resistance of the sensor decreases and ultimately voltage drop across FSR increases which can be interpreted as the user/amputee intent to control the prosthesis.

$$V_{out} = \frac{R_M * V_S}{R_M + R_{FSR}} \tag{1}$$

For developing the control algorithm for the prosthesis, several discussions were carried out with amputees, physical rehabilitation training coordinators and prosthetists. Since the objective of the study is to develop a cost-effective prosthesis for developing countries having low literacy rates, therefore, the control algorithm is kept as simple as possible such that a maximum number of amputees can benefit from the device. In addition, there was a possibility to use two separate FSRs and gestures for controlling the gripper and wrist unit, however, to avoid complications only one FSR was used for actuation of both degree of freedoms.

V. RESULTS AND DISCUSSION

The manufactured prosthesis covered with an inner shell and cosmetic glove along with the battery and embedded system fitted in the socket is shown in fig 7. The component-wise details of the developed prosthesis are presented in table 2. The total cost of the prosthesis including socket, inner shell and cosmetic glove is nearly 250 USD which is comparable to cosmetic and body-powered solutions available in developing countries. Other studies [47], [54] have also explored the development of cost-effective prosthetic devices, however, the torque of the actuator and thus the gripping force is on the lower side. Furthermore, the mechanical structure of the hand is mostly 3D printed from low strength materials such as PLA which may result in breaking/cracking of the components in the extensive use of the device. Contrary to these studies, the developed prosthesis utilizes a high torque servo with the structure manufactured from Aluminum which is not only lightweight and easy to machine but also has superior strength as compared to most of the 3D printable materials. The total weight of the entire system excluding the socket and cosmetic glove was recorded as 461 grams and thus meet the criteria discussed by Belter [55].

A. CONTROL SYSTEM FOR THE DEVELOPED PROSTHESIS

After successful integration of mechanical structure, electronic devices and control systems, various tests were performed for the characterization of the prosthesis. For this purpose, a right-handed male transradial amputee voluntarily participated in this study. The amputee was healthy having 27 years of age and does not have any skin-related disorders. Before testing the prosthesis, the amputee was briefed about the experimentation protocol and accordingly informed written consent was obtained from him. This study was duly approved by the ethical committee of the University of Engineering & Technology, Peshawar, Pakistan.

Initially, the data from the developed sensor was collected in a free hanging position from wrist extensor muscles. The experiment consisted of performing the following hand and wrist movements: a) Relax, b) Power Grip, c) Flexion of wrist d) Extension of the wrist, e) and pronation of the forearm. For the convenience of the amputee visual instructions for every movement appeared on the screen of the host device. These instructions consisted of the name and the images of the gesture along with an audible sound for switching from one gesture to another. The amputee was provided initial training for 30 minutes to familiarize him with the experimental procedure. Every movement then appeared on the screen for 5 sec and the amputee was asked to mimic these gestures with repeatable and moderate force. In between two movements, the amputee was provided a rest of 3 seconds to avoid fatigue. A sampling rate of 10 Hz based on literature [37] was adapted for the collection of data through PLX-DAQ [56].

Fig 8 presents the FMG signals obtained while the amputee was performing the instructed gestures. The objective of this experiment was to evaluate the stability over time of the acquired FSR signal, therefore, the experiment was repeated ten times. Fig. 9 presents the standard deviation of various gestures across the trials. From fig. 9 it is evident that the standard deviation of power grip across the trials is fairly constant with less variations as compared to other grips.

TABLE 2. Bill of material along with the cost breakdown of developed prosthesis.

Sr. #	Component	Type	Amount (USD)
1	Gripper Servo Motor	SB-2230 SG	150.00
2	Wrist Servo Motor	MG90S	3.00
3	Structural components including finger, thumb, crank link, spring, gears in the wrist unit	-----	60.00
4	Microcontroller Unit	Arduino Pro Mini	3.15
5	USB to TTL Converter	PL203	2.83
6	Voltage Regulator	LM-2596	0.60
7	Operational Amplifier	LM358	0.19
8	Battery	7.4 V Lithium Polymer	6.50
9	Battery Casing	Poly (methyl methacrylate)	3.15
10	Controller Casing	Transparent heat shrinks tubing with sides fitted with connectors	3.50
11	Force Sensitive Resistor	FSR 408	4.00
12	Sensor Casing	Machine from high-strength polymer	4.41
Total			**241.33**

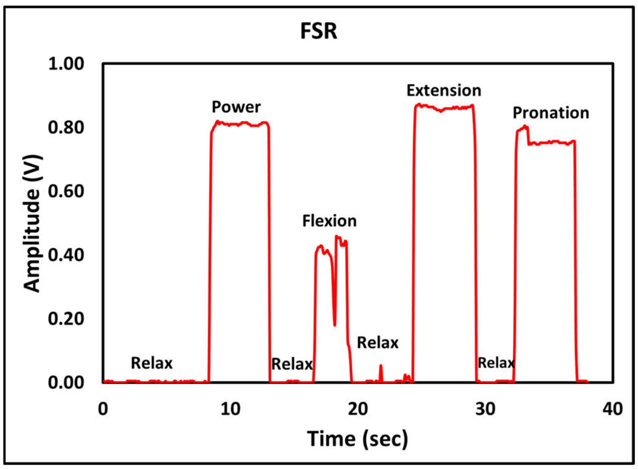


FIGURE 8. **FMG data acquired for various gestures.**

The fact is also endorsed by computing average standard deviation (trial-11) of the four gestures across first 10 trials. Since, the power grip has the least average standard deviation as compared to other three grips, therefore, for controlling the developed prosthesis in real time the algorithm was developed on the basis of FSR signal acquired from power grip.

The assembled prosthesis was then fitted to the amputee and controlled through the scheme presented in fig. 10. When

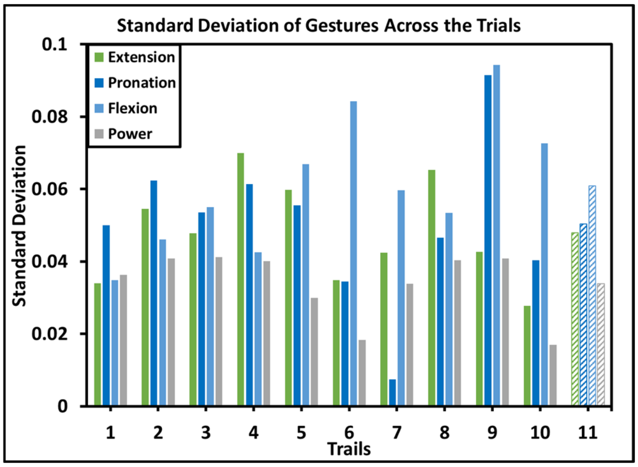


FIGURE 9. **Standard deviation of the four gestures across the trials with average standard deviation of each grip plotted in trial-11.**

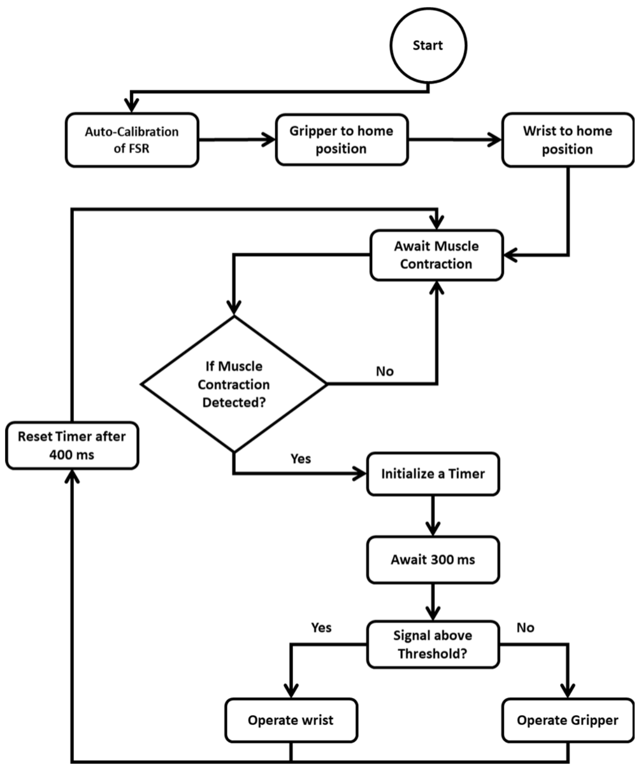


FIGURE 10. **Flowchart for actuation of developed prosthesis.**

the controller is powered on, the process of auto-calibration was initiated. As evident from fig. 11 for the initial three seconds the amputee was directed to not perform any muscle contraction while the MCU collected and averaged data for the unloaded FSR. The threshold of FSR was then computed as 1.25 times of the unloaded averaged FSR value. As an instance, the average of the first 30 samples in fig 11 was computed as 1.3 V, whereas, the threshold for hand and wrist actuation was set as 1.62 volts. This process greatly assists in the auto-calibration of the sensor and minimizes the impact of

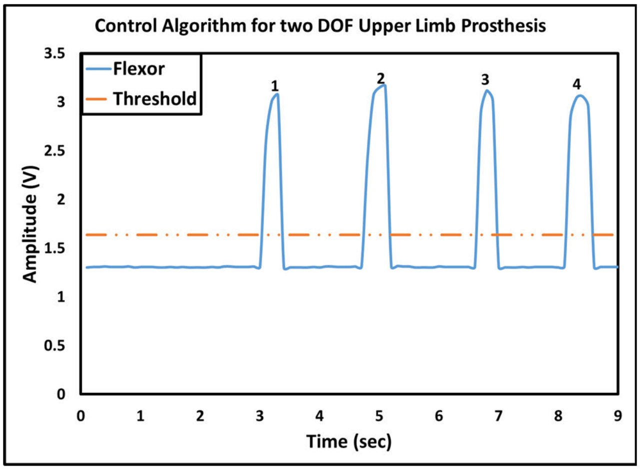


FIGURE 11. Four muscle contractions sensed by FSR for actuating gripper and wrist unit according to the pre-defined threshold.

TABLE 3. Four muscle contractions performed by amputee to activate gripper and wrist actuators.

Muscle Contraction	Signal Duration	Activated Actuator	Previous State	New State
1.	<= 300 ms	Gripper	Close	Open
2.	>300 ms	Wrist	ACW	CW
3.	<=300 ms	Gripper	Open	Close
4.	>300 ms	Wrist	CW	ACW

pressure/force variance due to slight movement of the sensor while wearing of socket or operation of prosthesis.

Once the auto-calibration is achieved, gripper and wrist units are moved to their home positions. For the gripper, the home position was defined as fully closed position whereas for the wrist unit the home position was in-between pronation and supination movements. The controller then waits for muscle intent for the desired actuation of the hand/wrist unit. Muscle intent can be defined as FSR signal that crosses the pre-set threshold. When the FSR signal exceeds the threshold value the MCU records and computes signal duration. Signal duration can be defined as ''the time over which FSR is active'' or ''the time over which the value of FSR is greater than pre-defined threshold''.

If the signal duration is less than or equal to 300 ms, the gripper is operated (based on its previous state), whereas for signal duration greater than 300 ms wrist unit is operated (based on its previous state) followed by resetting of signal duration counter to zero after 400 ms. Hence, the response time of the proposed algorithm for actuating either the gripper or the wrist unit is 300 ms which is in-line with the recommendations of Englehart and Hudgins [57].

For complete implementation of the proposed control algorithm, the amputee sustained four muscle contractions at different time stamps as shown in fig. 11 in the order presented in table-3. Since, as already discussed when the MCU is powered on, the gripper and wrist unit is moved to their respective home positions. By default, the MCU stores the previously actuated movement of the gripper as closed, whereas, for the wrist unit it assumes that previously supination of the wrist was performed. When the amputee sustains first muscle contraction, the MCU computes signal duration which is less than or equal to 300 ms, therefore, the gripper actuator is activated. The previous state of the gripper recorded by MCU was ''close'', therefore, the servo actuator is activated to open-up the gripper and the previous state is updated as ''open'' for the gripper.

The signal duration of the second FSR signal computed by MCU is greater than 300 ms, therefore, the wrist unit is activated. Since, previously the wrist unit performed supination, therefore, the signal to the wrist actuator by the MCU is to perform pronation movement and update the previous state to ''pronation'' as well. Similarly for third and fourth muscle contractions the gripper and wrist are activated respectively based on their previous states. Furthermore, during the execution of any gripper/wrist movement, if the MCU detects a new active muscle contraction then it is interpreted as amputee intent to stop the gripper/wrist at that specific position and halt that movement immediately.

B. EFFECT OF SWEAT ON FSR

Developing a standalone casing for FSR was not only to precisely integrate the sensor inside the socket but also minimize the impact of sweat on the performance of the sensor itself. For this purpose, the authors performed a test to evaluate the impact of sweat on FSR with and without the developed casing. Sweat was artificially introduced on the residual limb of the amputee and the obtained results are plotted in fig. 12. From the figure, it is quite evident that without casing the response of the sensor is highly unstable over time and there is a significant impact on its amplitude which may affect the future development of proportional control. Contrary to direct placement, when the FSR is protected in the casing and accordingly embedded in the socket, significant reduction in the effect of the sweat on the performance of the sensor is observed. This signifies that FSR possesses a huge potential to cost-effectively control active prostheses with advanced techniques such as proportional control and adaptive gripping.

C. EFFECT OF VOLUMETRIC CHANGES ON FSR

The volumetric changes in the residual limb are very common when the amputee moves the prosthesis in different orientations and hence may invalidate the calibration of FSR. Therefore, effect of volumetric changes in various poses was evaluated when the FSR is enclosed in the housing developed by the authors. For this purpose, a six-axis Inertial Measurement Unit (IMU) GY-85 was integrated with the already developed embedded system. The free-hanging position of the arm was calibrated as 0° and the data of FSR for various arm positions presented in fig. 13 was collected and the results are plotted in fig. 14.

A total of ten trials were performed for evaluating the effect of volumetric changes on FSR and the obtained results

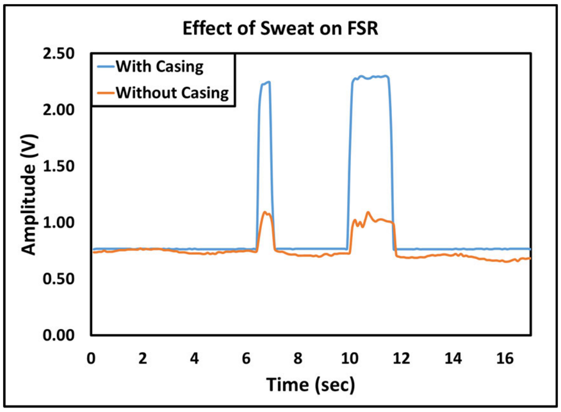


FIGURE 12. Effect of sweat on the performance of FMG with and without casing.

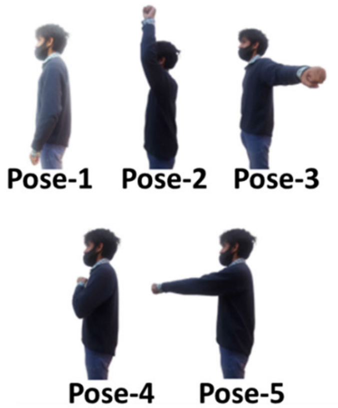


FIGURE 13. Dynamic protocol for evaluating effect of volumetric changes on FSR.

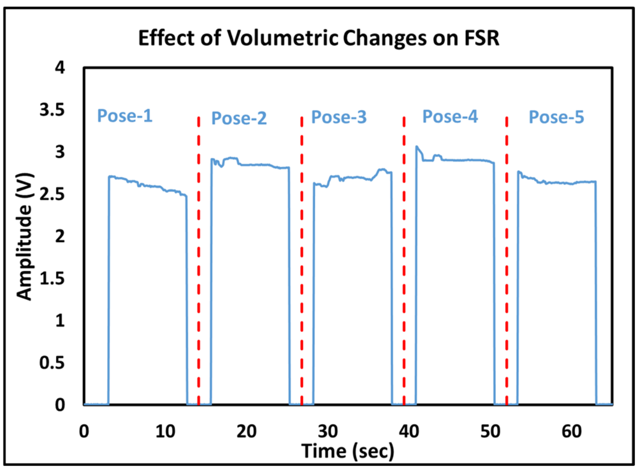


FIGURE 14. Evaluation of effect of volumetric changes on FSR.

are plotted in fig. 15. The average standard deviation of the poses across ten trials is also calculated and plotted as trial-11 in fig. 15. It was observed that average standard deviation of power in different poses ranges between 0.02538 and 0.037662 which indicates that the sensor housing assists in stabilizing the response of FSR, making it robust against the volumetric changes.

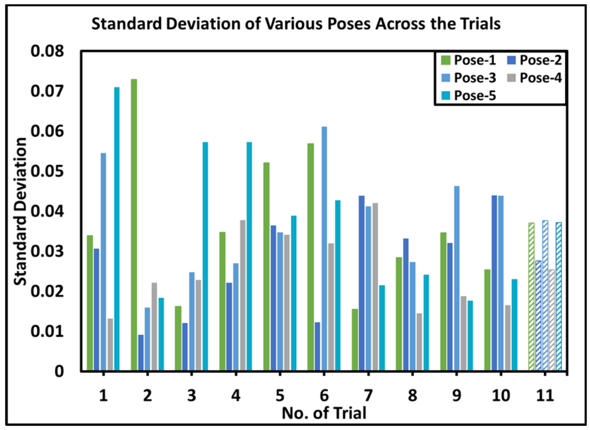


FIGURE 15. Standard deviation of the five different poses across Ren trials with average standard deviation of each pose plotted in trial-11.

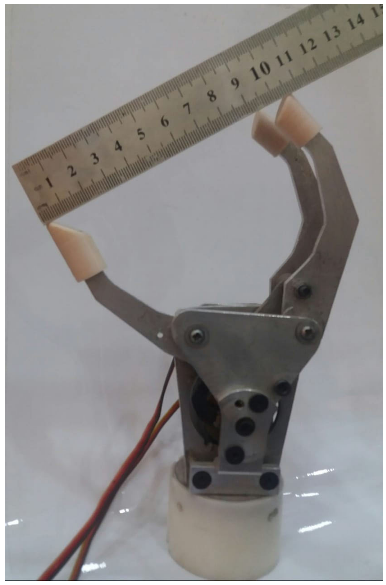


FIGURE 16. Maximum opening of gripper.

D. MAXIMUM OPENING OF GRIPPER

For characterizing the mechanical behavior of the developed prosthesis maximum opening of the hand was measured as it is an important feature to ensure that prosthesis can grasp and hold a variety of large and small objects. During trials, it was observed that the opening of the gripper was resistance-free. The maximum opening of the gripper was measured from the tip of the thumb to the tip of the finger as shown in fig. 16. A total of 10 trials were performed and it was observed that maximum opening is within a narrow range of 100 mm to 104 mm due to the closed-loop control architecture of the servo actuator incorporated in the gripper.

E. PINCH FORCE TEST

The gripper is designed to provide a pinch/tip grasp for the gripping of large as well as small objects, including a pen, paper, a glass of water etc. Therefore, a pinch force test was performed to estimate the maximum gripping force. For this purpose, JHBM-H3, HBM load cell was used. Initially,

FIGURE 17. Experimental setup to evaluate pinch force of developed prosthesis.

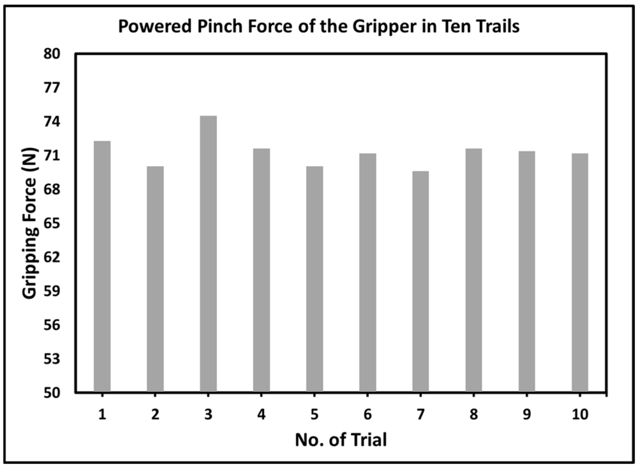


FIGURE 18. Powered pinch force of the gripper across ten trials measured using JHBM-H3 load cell.

it was interfaced with Arduino Nano for calibration, followed, by computation of pinch force as shown in fig. 17. Ten trials were performed and obtained pinch force across each trial is plotted in fig. 18. It was observed that pinch force was within the range of 65N to 75N.

F. DEVICE PERFORMANCE

Table-4 lists some features of the developed prosthesis in comparison to two commercially available prosthesis along with reference parameters for mechanical design of externally powered prosthesis listed in [55]. A similar comparison of the designed wrist unit with the state-of-the-art artificial wrists is already discussed in table-1. From table-4 it is evident that mechanical design parameters of the developed prosthesis match closely with the state-of-the-art devices. Although, the maximum pinch force of the prosthesis is less than Ottobock sensor hand, yet is comparable to bebionic hand and within the reference limits.

TABLE 4. Comparison of mechanical performance parameters of the developed prosthesis with commercially available transradial prosthetic devices.

Parameters	Developed Prosthesis	Ottobock Sensor Hand	Bebionic Hand	Reco m.Val ues
Max. Powered Pinch Force (N)	75	100	75	65
Max Unpowered Pinch Force (N)	24	----	-----	------
Fingers Close and Open Speed	1 sec 100 mm/sec	300mm/sec	1.7 sec	115 – 230 deg/se c
Weight of Prosthesis (grams)	461	350 - 500	495 - 539	<500 g
Overall Size (*l* x *w* x *h*)	150 mm long 70 mm wide 60 mm thick	160 mm long 70 mm wide 65 mm thick	198 mm long 90 mm wide 50 mm thick	
Actuator type	Brushless Servo	Brushed DC	Brushed DC	Brush less

Similarly, the weight and overall dimensions of the prosthesis is comparable to state-of-the-art devices. The actuators for the gripper and wrist unit in the developed device are different from both Ottobock and bebionic as in these devices brushed DC motors are used, whereas, brushless servo motors are used in the developed prosthesis because of the closed loop feedback, long life expectancy and improved performance versus weight ratio.

VI. CONCLUSION

This study was focused on the design, development and evaluation of Forcemyography based novel prosthesis for upper limb amputees. The contributions of this research are summarized below:

- Development of a transradial prosthesis comprising of a gripper and a detachable wrist unit with cost comparable to body powered prosthesis and mechanical performance parameters similar to state-of-art devices. The detachable wrist unit has length less than existing artificial wrist and thus facilitates amputees with long residual limbs in restoring the functionality of their wrist joint. Furthermore, the gripper has enhanced cosmetic appearance and mechanical performance parameters including pinch force and maximum opening in comparison to existing low cost-grippers, thus will have a higher acceptance ratio.
- Improve the process for integration of FSRs in the socket by developing a specialized housing for the sensor that is capable of overcoming fabrication inconsistencies during socket manufacturing process through built-in linear adjustment and auto-calibration mechanism.
- Enhance the performance of FSR by enclosing it in the custom housing that minimizes the effect of

sweaty/hairy skin on the sensor in addition to avoiding bending of FSR when the amputee is performing various tasks.

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