Design and development of an ultra-low-cost electro - resistive band based myo activated prosthetic upper limb

by

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Statement of Authentication

I certify that the work in this thesis is to my best knowledge and belief is my own work except as acknowledged in this thesis. I hereby confirm that the thesis has not been submitted anywhere else, either in whole or in part, for any degree at any institution.

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Abbreviations

ASEMG/SEMG	Analogous Surface Electromyography
CNS	Central Nervous System
DSO	Digital Signal Oscilloscope
ERB	Electro-Resistive Bands
NMJ	Neuro Muscular Junction
PWM	Pulse Width Modulation
UEA	Upper Limb Amputation
ULP	Upper Limb Prosthesis

Abstract

In developing countries, many amputees have no access to the prosthesis. This is due to the challenges of the environment they are living in and to the prohibitive costs of available prostheses. To reduce this gap, a new concept design for an extremely low cost but highly functional upper limb prosthesis is presented. This goal is attained using a low-cost embedded platform (Arduino) and a wearable stretch-sensor adapted from Electro resistive bands (ERBs).

In the proposed design, a sensor based on ERB is used to detect residual muscle contraction which detects the volumetric shifts of contraction instead of electromyography signals. The signals received via this sensor is then processed via an Arduino micro-controller to drive a single DC servo motor. The DC servo motor is directly geared onto a claw-style two-fingered prosthesis which is printed in-house from PLA plastic using a standard 3-D printer. The amount of closure of the prosthesis is fed-back to the user via a second ERB sensor directly connected to the claw in the form of haptic feedback. To make the design easier to maintain, the gears and mechanical parts are made so simple that can be crafted even from recovered materials.

The entire design of prosthesis is presented in this thesis. The overall cost for the proposed prosthesis is estimated to be AUD 29. The proposed design can be easily scaled up to accommodate more complex designs such as having multiple individual fingers or wrist rotation.

Keywords: SEMG, Upper Limb prosthesis, ERB, less privileged amputees

Publications as a result of the thesis

1. G. D. F. Ulloa, N. Sreenivasan, P. Bifulco, M. Cesarelli, G. Gargiulo and U. Gunawardana, "Cost effective electro — Resistive band based myo activated prosthetic upper limb for amputees in the developing world," *2017 IEEE Life Sciences Conference (LSC)*, Sydney, NSW, 2017, pp. 250-253. doi: 10.1109/LSC.2017.8268190
2. N. Sreenivasan, D. Felipe. U. G, G. Gargiulo, and U. Gunawardana, "Towards utltra low cost myo-activated prostheses," Biosensors, 2018 (under review)'.

INTRODUCTION

1.1 General Background

The human body is a well-organized machine consisting of various connected organs and parts. Almost all of them are vital, functioning in specific and distinctive manner. So, when all the parts are in the perfect coordinated state, it can function exactly as a well maintained-lubricated system. Among them, the human hand is an important part having unique capability from other animals. For instance, has fingers and an opposable thumb that enables execution of more complicated movements and postured prerogative of the human being. Therefore, hands are one the most important component of our being. they are important for self-presentation, after the face. In addition, it has a tremendous part in creating a sense of wholeness during the interaction with the social and psychological setting. Furthermore, hands can carry out heterogeneous tasks, from the intricate and elaborated to the precision and power. From day to day activities of picking up things, holding, use of tools, communication, defence and entertainment, hands do myriad functions.

The loss of whole or the section of the upper limb (or hand) also known as upper extremity amputation (UEA), occur for various reasons. It may happen due to wars, accidents, infections, burns, and trauma or by the congenital factors. Amputation in one way or other always creates great challenges for the normal living of people [1]. The loss of one or more limb of a person can significantly affect the autonomy level of participation in life roles thereby evading their natural blend with society. It demands additional pressure and stress for performing their routine activities [2].

The common causes of amputation in many developed countries are advanced vascular failures which can cause a build-up of plaque in the artery walls and eventually leads to the blockage of free blood flow to a limb or an extremity. Whereas in several war-prone countries of the world which hardly struggling to survive with their limited public resources consists of more than 75% of amputees from repeated landmines. Every year the land mines alone are responsible for creating more than 25,000 amputations in these countries and around 300,000 amputees worldwide [3]. And, in some other underdeveloped countries regular hygiene and nutrition deficiency prompted the reasons for amputations. Further to this workplace injuries, natural calamities, violence, and the lack of elementary level public hygiene which also lead to diabetes and infections comprises the causes of amputation. World Health Organization(WHO) has estimated that around 650 million people worldwide have various disabilities, and the clear majority of them are surviving in low-income countries [4].

As stated earlier, due to birth defects, diseases, or accidents, a person might lose the ability of one or more extremity ranging from reduced function to total absence. In all such cases, an integration of an additional implant known as 'prosthesis' or 'artificial limb' could be able to make the life relatively normal [5]. The innovation and succession of progressed artificial limbs in the recent years have created a new world of opportunities in front of limb disabled people. Those extra construct can continually provide a life-support by upgrading the lost functioning ability to carry out normal routine. In turn, it helps the person to deliver the social commitments in a more confident and positive way [6].

The several artificial limbs available in the market, now a day have the most difficult parts to replicate the limb movements, which are very well-worked and exactly like the motion of our natural hands. The result is a very complex hand with huge cost of making, fitting, conditioning and servicing of parts. These ongoing charges cannot always be reasonable with the budget of low-income amputees surviving in underdeveloped or developing countries. According to some report by World health organization(WHO) on rehabilitation services provided worldwide, only less than 5% of disabled persons who are living in low income countries are using any kind of artificial rehabilitation facilities [6]. Interestingly for most of the everyday tasks, one can perform with the help of wo fingers, though it is not able to satisfy in full. The simple two finger arms to multiple joint arms are available in the market. Even though that is the case, the choice and use of these arms among amputees are limited mainly by high cost.

Considering the constraints in developing prosthetic world, the thesis outlines the proposal of a new prosthetic device with 3D printed arm that might enable a better life for the individuals having upper limb disability, especially in low-income areas. And, by the inclusion of latest 3D printed technology, the overall cost of production of artificial arm and assembly will further be reduced. But sadly, this is not the mainstream process yet in the artificial assistance sector. The practical benefits and worthiness of this activity in artificial healthcare options are still in its infancy in many underdeveloped countries in developing and underdeveloped countries [7]. Through this work I am trying to evoke some changes in the utilization factor of upper prosthesis.

1.1.1 Upper Limb prosthesis (ULP)

The use of a prosthetic limb is the best-recommended measure for the rehabilitation of amputees at various levels and variations of upper limb loss. And this assistive com adaptive equipment can serve them to continue a healthy fulfilling life even after an amputation. Typically, there are three main parts for all upper limb prosthetics as in Figure 1. A socket with the suspension system is the first part, which is the interfacing part used as the connection between the amputees' residual limb and the terminal device. It also adds more physical support to the rest of hand. Second, a piece of lengthening either a hollow or a flat part with interposed joints that can take the place of the lost limb length. Lastly, a terminal moving unit such an artificial hand with controlled grippers or fingers.

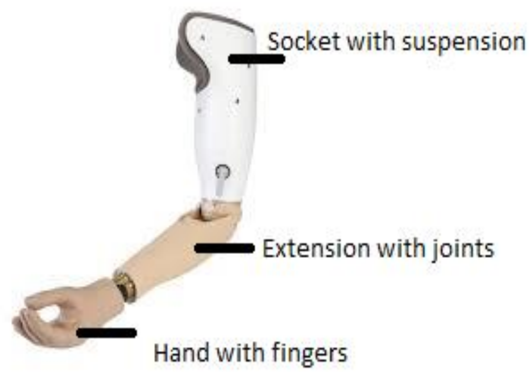


Figure 1. Typical upper limb prosthesis [8]

The socket-suspension system is the meeting place which acts as the conveyance path of myoelectric forces between the remnant hand and the artificial device. Because the continuous use of additional fitting is bound to be damaged on the underside skin of the residue arm, the socket suspension always ensures to have the prosthesis in the right position. The extension is the important structural backbone of the prosthetic arm rendering required mechanical support. Traditionally, various metals and woods have been used for making extensions and in the modern times, plastic and silicone-fibers replaced it with lightweight applications. This part usually comes with a textured cover related to the remaining hand which give a more instinctive and appealing appearance.

There are various extends for upper extremity amputations, so for different types of amputation problems the rehabilitation procedure requirement varies and so on with the prosthesis. The selection of a suitable prosthesis considers the amputation level, the contour of the residual limb, expected function and vocation of amputee, financial capability and appearance. The common upper extremity amputations [2] include the following levels as in Figure 2.

- Digits or partial arm (trans carpal)
- On the wrist (wrist disarticulation)
- Under the elbow (trans radial)
- On the elbow (elbow disarticulation)
- Over the elbow (trans humeral)
- On the shoulder (shoulder disarticulation)
- Over the shoulder (fore quarter)

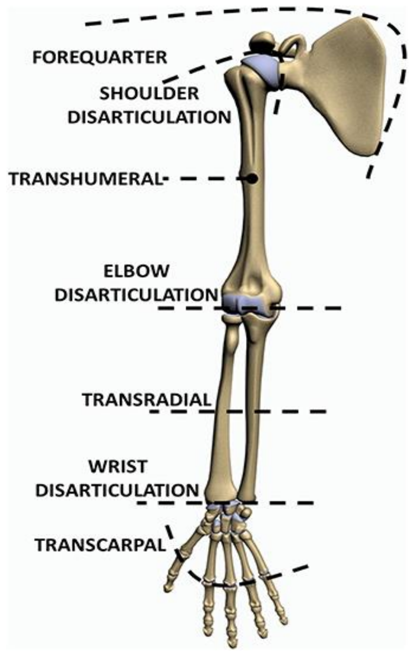


Figure 2. Different levels of UEA [2]

Most common of the above all in developing countries is at the trans-radial level of amputation, or just below the elbow. That means a clear majority of the prosthesis design approach in poor income countries demands only a wrist with a functioning arm.

There is always a growing demand for prosthetic limbs by the several replacements and repairs occur over a lifetime of an amputee. For nearly two decades, although there have been so many researchers working on upper limb prosthetic technologies, most outcomes shows the varied limitations of prostheses [9]. Unlike lower limb prosthesis, an upper-limb prosthesis is more complicated with multiple joints and motions. And unfortunately, so far none of the highly sophisticated and versatile devices can perform the complex functions done by our natural hands.

1.1.2 Mechanism of the normal hand and a prosthetic hand

All the actions of human body take place through muscles with the intricate interaction mechanism of the nervous system. The movements we need are made effectively and precisely because of these mutual interventions happens by some specialized brain cells in a timely manner. This engagement is what makes the light motor movements called as locomotion coordination. Such nerve cells in interaction which provide initial direction of excitement from body's Central Nervous System (CNS) are called motor neurons. It enables the initial introduction and distribution of the neurotransmitters through some non-specific tunnels called neuron-muscular junctions(NMJ) to the need for delivering energy in the form of synapses. This is the boundary of the excitation commencement, NMJs, is said to be the place where the motor neurons deploy a few substantial synapses obtained after the first stimulation to the neurotransmitters [10]. Those transmitters merged to arrive at the acceptor regions of the muscle fibers and then follow the trigger instructions and make the focused muscle move. The release of neurotransmitters thus controls the movements of the entire musculoskeletal system.

In human beings, the neurotransmitters have a stimulating action for developing muscular contractions. Similarly, just as the movement stimulates by these neuron groups, they need to relax the muscles even as the stimulation moves. The state of release is obtained through the interruption of neurons by the central nervous system. With, a continuous string of shrinking tensions and expansions, the necessary movements have put through as guided by the brain. The neurons can always energize multiple muscle fibers at a time. For any motor task, the brain undertakes the available sensory information such as sight, depth, height, weight and so on to ascertain the trajectory of operation it must possess. This information is certainly proprioceptive in nature having the capacity to identify the location, posture, orientation and alignment of the body and its parts. The brain then integrates all the information's to provide an array of motor commands to the specific action site [11].

For instance, during the shaking of hand, a series of activities is produced when CNS initiate a self-awareness, i.e. data regarding with the current position, information of the required elevation and angle, data of fine motor skills, estimation of the quantity of physical strength to achieve the duty and so on. Apart from this, it decides the sequence of minute steps in the whole process, integrating the visual data such as height to be raised, the angle to be at the various joints (such as angle at the wrist, elbow and finger nodes) and the amplitude of the shaking for the action of the musculoskeletal system. When the mode of operation is selected after feedback from visual data, a set of instructions will be given by CNS through the spine to generate motor movements. The instructions are conveyed as a succession of stimulated neuron signals, which in turn results in the production of special transmitters from neurons creating an action potential [12]. The result is depolarization of neurons and change in the membrane potential, which then causes the triggering of muscle cells resulting in the shrinkage of the corresponding muscle fibers. On contrary, the similar mechanism of the polarization of neurons itself directs to the relaxation of the muscle. The connected sequences of such contractions and relaxations of muscle fiber develops with the desired movement of the hand. The process is a highly complicated one involving transfer of chemical ions, which may occur within a fraction of a second thus making complex the job of replicating through an artificial means.

In an artificial well-fitting prosthesis, the device can be secured to the residual arm with a socket and suspension system. Identical to the natural limbs which are controlled by muscles stimulated by brain, the basic model of movable and workable prostheses is made up of long cables or chords enabling them moving back and forth, similar in doing the job of the muscle fibers. The typical example for such prosthetic hand is a body-powered hand which open or close the fingers and joints by pulling different cables attached to the moving pattern to the other hand or some convenient part. The recent myoelectric prosthesis employs the muscular impulses (EMG) via electrodes in the residual arm. The basic operation of hand i.e., open and close is then performed by the help of a motor controller. With the controller, fine controls such as soft pinch, hard grip, etc. are tailor made with the need of the disabled person. As muscles in the hand flexes, myoelectric potential is transferred through the electrodes into the logical control unit. Switches in the control unit operates a battery-powered motor to open and close the hand mechanism. Not as complicated as the normal hand, the artificial hand is also working on complex connected activities.

1.2 Objective

The active prosthetic devices with power supply from an external source are chosen by most of the users around the world. Surface electromyography (SEMG) signal is the most commonly used signal for many externally powered prostheses. The desire is because of their ease in functional capability with the living circumstances and by their pleasing compact look. These myoelectric prosthetic devices (uses EMG) are controlled by muscle contractions of residual limb. By the purposeful control, myoelectric arm can open and close fingers with a motor. But, in several developing areas of the world, many amputees have no access to any prosthetic hands for a normal living [13]. The complexity and huge cost of currently available prosthetic hand may limit the spread and its wide part custom-made applications to the underserved people in developing countries. So, to avoid the gap between the cost factor and prosthetic design an efficient design of artificial upper limb is proposing in this work.

The main aim of this project involves the design of a low-cost myoelectric ULP for disabled living in developing countries. This can be achieved by using a wearable band sensor (known as Electro resistive bands (ERB) instead of conventional electrodes) [14] to detect muscle contractions, with an inexpensive control platform and simple 3-D printed mechanical hand. Besides, the projected design of low-cost myo-activated prosthetic is aiming only on the basic functionality of the hand; one degree of freedom i.e. to open or close of hand. This design model at any rate can provide a suitable solution for the economic barrier existing in the developing world prosthetics. The aim of the project can be summarized as follows:

- The main objective is to design and develop a low cost artificial limb for people with disabilities living in developing countries.
- To move the mechanical hand in response to changes in SEMG with an ERB transducer sensor and a microcontroller assembly.
- Finally, to incorporate a closed loop and sensorial feedback control for better performance.

1.3 Scope of thesis

The proposed thesis typically aims to design and develop a prototype of an ultra-low cost upper limb prosthetic hand. The design is mainly meant for the rehabilitation support of upper limb amputee life in the developing and under developed countries. The model simply includes the necessary parts for basic hand movements, few materials and fast acting, cheap modern microcontroller technology. Even though the function of proposed hand is minimized to single degree of freedom i.e. open or close of arm, the design can provide successful changes in the survives of underprivileged class of amputees. The maximum cost of the production of the proposed design (including sensory and current feedback together) will not be higher than AUD 32 in any form, which is a great demand in the existing societal set up. The sole purpose of this work is to make sure the rehabilitation services using new upper limb prosthetic design can be provided easily to all persons with upper limb disabilities, regardless of their living backgrounds, ethnicity, whether they live in a city or in the countryside or whether they are rich or poor. This design also aims to understand the efficiency of wearable electro resistive bands over traditional electrodes with Arduino based assistance in the control of EMG signals for the voluntary opening and closing prosthetic hand.

1.4 Outline of the thesis

This project can be broadly divided into six chapters. This ensures a detailed analysis of the project work.

Chapter 2 Gives a brief description of the various papers and works so far related with my topic – Literature survey.

Chapter 3	It includes the detailed design of myoelectric sensor with Electro resistive bands for the detection of SEMG signals.
Chapter 4	Closed loop feedback methods are discussed in detail.
Chapter 5	Results and discussion.
Chapter 6	Future scope and conclusion

2 Literature Review

Artificial implants or prosthesis are important tools meant to substitute the role of a lost body part. The key intention behind all the prosthesis is to providing natural like support in the life of an amputee individual. As a result, self-confidence will improve to equip them for better career openings and emotional balance with the world. As mentioned before artificial upper limbs benefit the limb disabled individuals to recover the lost control on their survival skills to prepare things for themselves. They can get belongings, wear their own attire without any other personal aid. A prosthesis thus can progress the social, emotional and personal life of an amputee. Recent prosthesis makes use of innovative integrated circuit technology and they remain to advance by additional functionalities. The fusion of such innovative constituents has made them supportable to develop tougher, low weight devices with natural compliance to the residual limb. Within the last decade, the research and development on upper limb prosthesis have been focused more on the myo sensor-based design optimization and control strategies [15]. The design optimization on the device for the suitability of underdeveloped countries gains more attention in the recent years [16]. The major pullbacks occur in current available artificial arms are their inadequacy to accommodate simple control with fewer parts and less maintenance. These inefficient features are considered as the major reasons behind the abandonment of many advanced prostheses [17]. The design of an inexpensive prosthesis that can operate with any environmental conditions must necessarily follow a modular approach, with accessible materials. Moreover, the design should incorporate quick remedial strategies with minimal technical faults and interferences, affecting the device operation. Past research papers certainly can throw some light into different stages of development of ULPs till date with their adoption level into the amputee community[3], [18], [13]. The chapter also discusses the idea of the use of electro resistive bands (as the proposed SEMG sensor) which has got the inspiration from earlier biomedical researchers in various other fields [19], [20], [21].

2.1 History of the prosthesis

The use of artificial limbs is not a recent invention, and it is assumed to have appeared from the starting stages of human progression. Earlier prostheses were established only for bodily support function, beautifying presence, and to deliver an emotional feeling of fullness. The first recorded instance of the use of an artificial limb was mentioned in Rig Veda [22], an old Sanskrit textbook written in the period between 3500 and 1800 B.C. in India. As a poetic verse, it mentioned about the details of an iron leg which was used by the Warrior Queen Vishapala after her amputation from the battlefield. The earliest evidence of deformity restoration which commenced as the modest props made of timber hooks and leather clutches clearly indicates that unlike these days the facility for moving function stood more like an inevitable preference than a choice. Artificial limbs made of various natural materials have been found in some of the coverings of Egyptian mummified body. This shows even at that time only the rich could then afford to have prostheses [3]. During the period of 15th century, there were metal prostheses designed as an extension of the armor for soldiers. They were designed to use in the battlefield but not for the normal daily functions. At that time, prostheses were more beauty enhancing than usefulness; and were destined to conceal the mere disability [23]. Due to a large number of amputees from civil wars, a demand aroused later for better medical developments with advanced functional prosthetics [24]. After that, during the 16th century, Ambroise Paré, a great military surgeon from France, designed somewhat functional upper and lower limbs with rods and gears [25]. He was succeeded in illustrating the natural movements of artificial limbs through the adjustable harness and several lock mechanisms. Through his many other inventions, the amputation surgery existed in those times was completely modified. By early 20th century, the sudden increase had witnessed in the advanced artificial limbs. This era led to the invention of myo-controlled prosthesis in mid of the century.

Later on, during the period of 1960, the very first purposeful artificial arm was made as a functioning tool for the handicapped [26]. After that during 1980-2000 period demonstrated the emergence of microprocessor-based control of prosthetic limbs named as 'Intelligent prosthesis' and 'Adaptive prosthetics' having hydraulic-pneumatic controls [27], [28] served as a new stepping stone to the prosthetic upgrade. Further, with the refinements of technology, stronger and lighter prosthetics were developed offering a greater level of control and comfort [10]. In 1987, a novel method [29] was projected to distinguish the way for planned gestures between several kinds of body extremity actions and selectively choose the unique action that needed to deliver. This has opened the use of multichannel EMG signals processed by suitable levelling filters. The placing of an electrode on to the body had made versatile by the cross-matching of data between the numerous EMG signals. The change in amplitude and frequency data of the SEMG signals are provided by different stages of filters. The technique could discriminate the very delicate fluctuations within the EMG patterns by means of the connected neural systems. In 1996, another work was published proposing an innovative category of the myo-potential prosthetic arm [30], called as Biomimetic EMG-Prosthesis hand. This model could demonstrate the elementary connected behavior of the neuron and muscles in the human hand. The literature specified the need of necessary parts for the working of an artificial hand. The vital parts consist of EMG signal processing unit, a replication scheme of neuromuscular arrangement with changing aspects, and a servo system associated to the terminal device. The edge device is the motorized hand having a single degree of freedom. The EMG processing unit analyses the isometric forces of the muscle fibers. The angle is designed from the measured gripping power P and the amount of torque requirement is also estimated by the EMG dealing unit. The servo system confirms that the position of the digits of the extreme device is in perfect match with the estimated position. This format presented in the literature showed inspiring outcomes in the soft grasping of an object.

Later, in 1999 another paper proposed the hardware embedded chip control of prosthetic hand [31]. The chip presented the feature of shorter learning time, unlike the neural network models. Besides that, improved efficiency of the chip during the operation proved the design is better among the myoelectric controlled prosthesis at that time. Then during the 21st century, the thought of low-cost artificial hands commenced on track.

In the recent years, the sudden boom of CAD-based 3 D printing and designing have laid a new era in the manufacturing industry thereby triggering a cost-efficient stage in the growth of prosthesis. Modern prosthesis is equipped with multiple degrees of freedom allowing amputees to stand on their own feet. A quick overview of the historical development of prosthesis is given in Table 1.

Table 1 Historical overview of the prosthesis

Year	Details
3500 B.C	The ancient Indian poetic book, Rig Veda, the first recorded document about prosthesis.
Ist B.C	Oldest usable bronze peg prosthetic was discovered by archaeologists.
1508	German Knight Gotz Von Berlichingen used two prosthetic iron arms for the missed right arm during the battle of Landshut.
1529	Ambroise Pare introduced amputation recover to the medical community. The person is considered as the ‘Father of prosthetics’.
1696	Pieter A. Verduyn, a prosthetic surgeon from Dutch made first non-locking prosthesis below the knee.
1843	Sir James Syme described ankle amputation.
1861-65	Start of American Prosthetic field with North Carolina as the center.
1914-18(World War I)	European boom in prosthetic research.
1939-45(World War II)	Advanced research in American prosthetic necessities with military group tie-ups.
1944-48	The very first myo-potential prosthesis was formed in this period by Reinhold Reiter, a science student from Munich University. But he did not gain clinical or commercial acceptance.

1957-69	Researchers in many countries invented various myoelectric control. Groups led by Bottomley in England; Herberts in Sweden; Kato in Japan; Kobrinski in Moscow; Reswick, Lyman and Childress in the USA were some among the pioneers.
1970	Commercial development of myoelectric prosthesis.
1987	Multi-channel prosthesis control.
1980-2000	Intelligent and adaptive prosthesis.

The current expansions in limb prosthesis area are concentrating on the practice of acquiring one's physical motions such as EMG, EEG, voice, vision, etc. of the user to make the anticipated gesture in the device [32]. The know-how of transformation of muscle activities into corresponding electrical variations permitted the dawn of myoelectric class of extremities. The idea is highly appreciated due to the sensitive capability of obtaining feeble muscle motions and can alter them into equivalent movement, thus permitting a reduced amount of annoying movements [33]. The indicators that remained till now are EMG signals, EEG signals and neural signals. Additionally, the conceivable measure of physique signal can be pH value between the neuron and muscle junction, which depends upon the bodily chemical ion exchange and movements across the muscle tissue [10]. The prosthesis used by sports people is of different nature than a common man using in the everyday task. So, the mode of selection of body signal sensors is based on the specific demand of the user such as climatic conditions, work nature, affordability and the frequency of use [34].

2.2 Types of ULPs

The evolution of Upper limb prostheses (ULP) has generally classified into various categories based on the functionality, SEMG acquisition methods, control parameters and generation of implementation.

2.2.1 Classification based on functionality

The main categories based on the functionality are passive prostheses and the active prostheses. They can be subdivided as follows in Figure 3 chart. [35], [36].

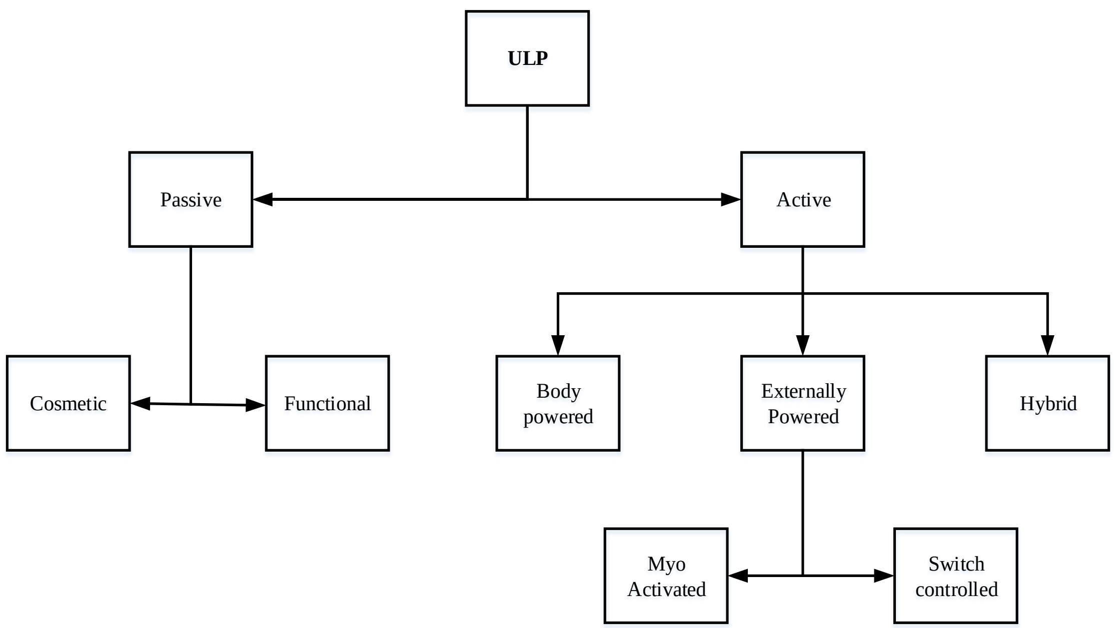


Figure 3. Types of Upper Limb Prosthesis [37]

A passive prosthesis as in Figure 4 is lightweight devices with no moving parts. It can assist with the function of an intact hand with the fullness of cosmetic appearance. Whereas in the active prosthesis, motors and mechanical systems make it more functional.

Figure 4. Passive arms [37]

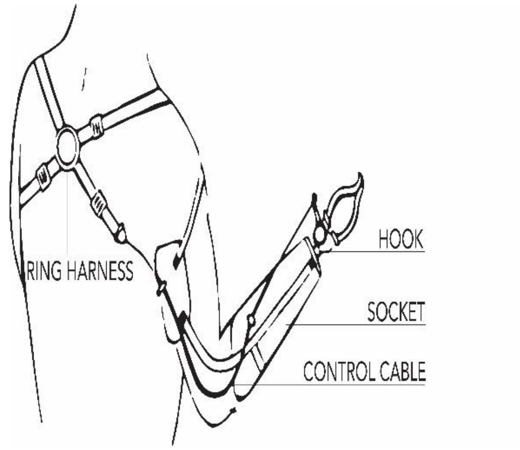


Figure 5. Body powered [37]

Among them, the body-powered ones are moved by remaining body parts of the amputees (see the Figure 5). A harness and belt are attached to the artificial hand and with the body; by the cable and brake arrangement, the hand can be made to open /close. The body-powered prosthesis is relatively durable and lightweight than other active ones. But, the user must pull the cable arrangement with enough strength to make finger movements through the attached harness. This often leads to additional pressure and stress on the use of prosthesis [38].

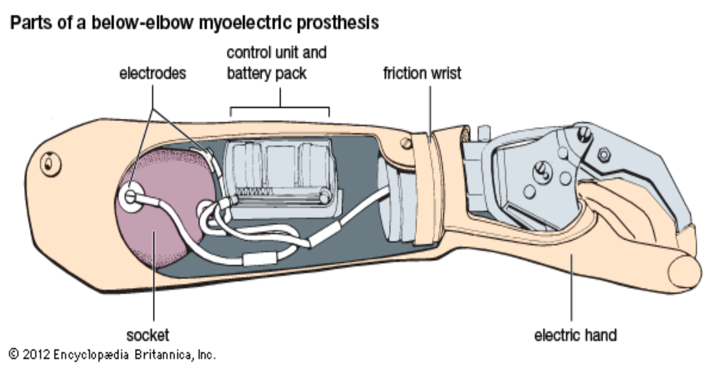


Figure 6. Myoelectric Arm [39]

The problem was rectified with the externally powered prosthesis [40]. They are powered by batteries contained in the system. The input to these devices can vary from electromyographic signals, force sensor signal, pull/push switches. There is no connecting cables and harnessing belts on the device, so more attractive in physical look (see the Figure 6) but, as demerits it demands more attentive and careful operation, frequent battery recharge and extra maintenance. Although these category prostheses have weight issues, the efficiency is better than earlier ones. The benefits of both body-powered and external powered devices are incorporated in other category known as hybrid prosthesis [35].

2.2.2 Categorization based on SEMG acquistion

Recent trends in prosthetic expansion emphasis on the practice of using motions that the human body develops to initiate rotation of servo motor. As mentioned earlier, the common signals that are used in for actuation of the artificial hand are SEMG, brain signals and neural signals. Among them, SEMG is the most demanding signals to control the modern world prostheses. EMG signal itself is an intricate biomedical signal that produce as electrical pulsations in muscles during its tightening and easing by neuromuscular actions of the brain. The amplitude and frequency of these signals are normally dependent on the anatomical and physiological properties of muscles.

Because of this anatomical dependence, the method of SEMG sensor control of the artificial limb involves two techniques: invasive and non-invasive techniques [41]. Invasive techniques consist of electrodes and sensors directly connected to the patient's nerves while non-invasive involves the interfaces to be placed on the patient's skin. Both categories make use of muscle signals so they are called as myo-activated or controlled prosthesis.

On the other hand, the control device used in myo-activation in turn makes another subdivision for upper limb prosthetic solutions. All prosthesis can thus classify into two categories: Non-microprocessor prosthesis (Mechanical) and Microprocessor/controller based prosthesis [27]. The Non- microprocessor or mechanical prosthesis are old fashioned one with harness, mechanical suspension systems, hatches and release mechanisms. Whereas the microprocessor based ones are current solutions with compact integrated units. Microprocessors are logical units used to control various functions of artificial devices. The emergence of microprocessor based artificial limbs control for the people with upper limb disabilities has significantly expanded the spectrum of rehabilitation and treatment options. It allows prosthetic management with smooth control options [28]. Additionally, feedback can be adjusted according to desired myoelectric control. The important advantage of microprocessors is that it can accept a wide variety of input devices and parameters enhancing the prosthetic function. It also has

- More functionality and stability in hand movements.
- Less time for operation and release

2.2.3 Classification based on different generation of ULPs

The evolution of microprocessor/controller based prosthetic control systems can be classified into three separate generations. The development of first generation was based on digital systems available at that time. While the second generation was a bit modified with low power components. And, the third generation which are the existing ones, based on microprocessors and digital signal processors. First generation microprocessor used simple logical operation of on- off control method using voluntary closure and voluntary open of prosthetic hands using electronic drives, wrist rotators and elbows. It can be operated only by a single speed setting. The number of input devices attached to the unit is limited and the entire system was attached to body harness for support. Lack of correct suspension was the major issue regarding with this generation of controllers [32].

The second generation involved better proportional control using specified reliable electronic packages. By lowering the number of muscle thresholds used for the device operation it maintained and incorporated low microvolts for the power requirements. The second-generation control includes single site, dual site interfaces with separate electronic packages. The main issue with this generation of controllers were the lack of inter changeability and the management. Dual site control was difficult from user perspective so if it was tried to replace with the single site control, the entire package must be changed. This produces more cost and time involvement [32], [9].

The problems with two generations were rectified by the versatile programmable controllers called as third generation. The compact size and multiple controls allow them to provide more realistic nature to the functionality.

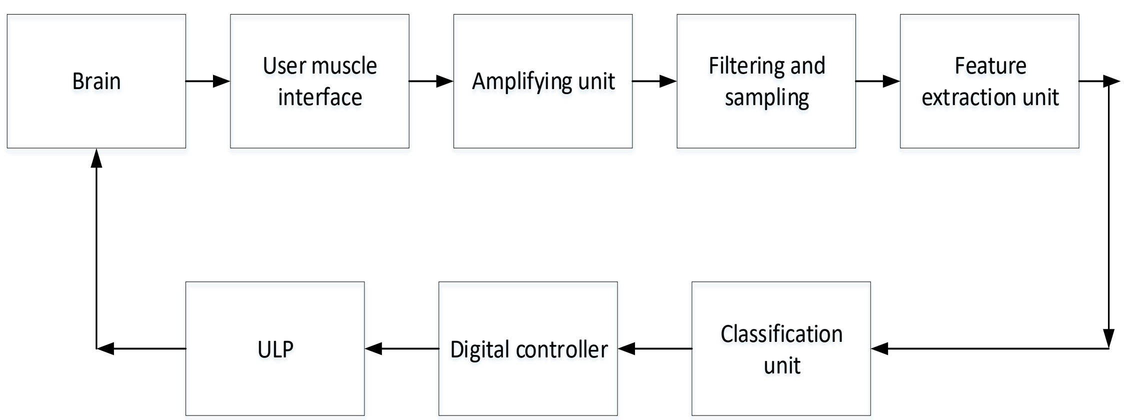


Figure 7. Layout of typical microprocessor based prosthesis

Third generation technologies are making artificial limbs more functional than ever before. The use of microprocessor controllers along with compact materials has made the prosthesis easier to use (see the Figure 7). All microprocessor based artificial hands developed by researchers include sensors, encoders, software systems, noise filters, resistance system, drive controls and battery. Most of the new age prosthetic devices are developed with EMG/EEG signal measurements consisting of a filter and control network to provide the actual torque signal for drive motion. The EMG signals are measured using electrodes and the signal is then given to conditioning unit of the controller. With microprocessor technology, body interfacing sensors are placed over the socket of the prosthetic limb and the person with an amputation is trained when and how to their flex certain muscles. The controller then decides the actuation for the prosthesis. The sensors send a signal to the drive motor to do a specific motion programmed by the controller like open or close. A visual feedback obtained from the ULP is used by the brain signals provide a further level of gripping force.

2.2.4 Classification based on control parameters

The functional quality and quantity, component and control, cost and risks are the aspects which decide the control parameters of ULPs. The available variations in upper solutions are

1. Electrodes – surface, percutaneous, implanted
2. Channel – single channel, Multi-channel
3. Current flow – unipolar, bipolar
4. Triggering – EMG triggering, EEG triggering, Button, cyclical
5. General factors – frequency, pulse width, temperature, Intensity, duration, dosage, ramp, On/Off cycle, waveform shape and so on.

As detailed before, whatever classification is the socket is the most important part of the upper limb prosthesis. Socket creates a direct connection between the residual limb and the prosthetic device, so it is important to design it properly to fit with individual's shape and size of the limb. The use of vacuum technology into suction method has allowed disabled to continue in physically challenging jobs and proactive in sports events. The existing market systems include normal suction sockets, flexible sockets, vacuum sockets, custom liners, and lock-and-pin systems [1].

2.3 Recent developments in ULPs

The recent advancement in intelligent and controlled prosthesis is the use of multi-articulating prosthetic hands [33]. The kind of dedicated and customized operations are also possible with such microcontrollers based prosthesis. The low cost, easy programmable Arduino controllers are the current trend in the prosthetic market. The multi articulation hands are controlled by multiple motors to drive different fingers and to obtain multiple hand positions. They have several pre-programmed hand gestures that a patient can select from their connectivity device. The gestures such as finger point, pinch, grip, and so on with wrist rotation and elbow extension. When the gesture of hand is selected, using acquired myoelectric signals, the user can control the voluntary opening and closing of the fingers with selected speed. The current developments also are in the direction which involves the advanced control of actuation using a single motor instead of many motors. It makes use of state feedback control given to the prosthesis to develop the correct gripping force. The detailed layout is presented in Figure 8. So, the actual grip can be modified according to the size, shape and type of the object to grab. With highly sensitive torque and pressure sensors, the controller can detect changes in pressure as they are applied to the fingers. These are converted into electrical signals and fed back into the user's brain.

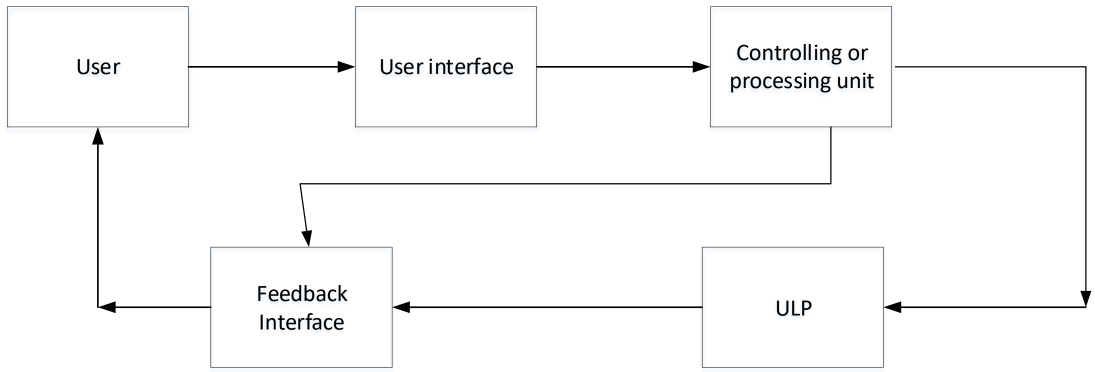


Figure 8. State feedback model with user interface

Other recent emergences in electronic prosthesis are brain controlled prosthesis. It uses the brain computer interface technique for the acquisition of signals directly from the brain cells. Instead of collecting EMG data as in the traditional controllers, mind controlled prosthesis (Figure 9) uses EEG data to process the signals for actuation. The signals generated from the brain can be easily identified using wearable electrodes rather than clinical measures. The main advantage of this category prosthesis is the exact replication of delicate hand movements and precision grips. A feeling of an embodiment can be obtained from the sensory feedback which in turn keeps the user to act accordingly.

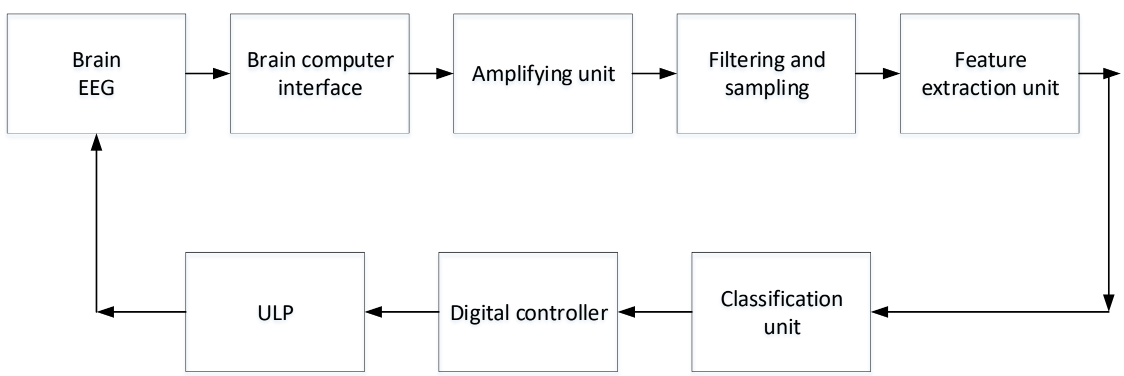


Figure 9. Mind controlled prosthetic design

To get a feel of real world, the prosthesis is changing their way from functional cosmetic appearance to direct skeleton integration level. The union between patient's body and the prosthetic machine is done through mind controlled mechatronics. For long-term stable mechanical systems, the direct attachment of artificial arm to the residual limb skeleton with bidirectional interface proves better. The bidirectional interface makes the prosthetic arm to send signals in the opposite direction i.e.to the brain. This technology is an important step towards the more natural control of artificial limbs with intuitive, reliable communication and sensory feedback between the prosthesis and the patient's body.

As the signal acquisition is very critical with traditional methods, an alternative, with the use of electro resistive bands (ERBs) - a new easy to wear contactless sensor that changes its electrical resistance when it is stretched serves better the purpose. The new sensor does not require any preparation of patient's skin and can be comfortably worn as an elastic band [14]. The use of ERB bands in biomedical application were discussed in detail in [20], [19], [21]. A control scheme similar to the proposed design has detailed in the paper [14]. But, the cost associated with the literature model is relatively high compared to the proposal. Moreover, the literature design makes use of instrumentation amplifiers, advanced processing parts creating a much more complicated arm than the proposed design.

2.4 Issues of current ULPs

Though the advancement of multifunctional prosthetics in recent years have drastically changed the normal way of living for amputees; unfortunately, the costs associated with their manufacturing and maintenance is sometimes simply prohibitive.

Moreover, throughout the lifetime many limbs are crucial in backing the elementary requirements of an amputee [42]. Some of the current prosthetic limb designs available in the market having exact replication of the human hand gestures and features make the design extremely complex requiring constant fine-tuning, complex assembly, and continuous maintenance of parts. For example, the commercial hands with multiple grips and superior functions are expensive with costs up to USD 50,000 [43]. The devices that offer the highest support in the current market do not get a favorable approach from the user. Other than huge cost, the major facts of device rejection are probably due to precise factors relating to uneasiness, matters related to strength and other failures.

When an individual miss both the arms, he or she generally wears an artificial upper limb. This may nearly permanent, because of their body requirement. However, one who has lost one arm or a part of it can survive better than those having disability in both hands. They can learn to organize most things by themselves single-handed by means of the lasting parts of the residue arm, and even without an artificial arm. The need for an additional device may sometimes create unwanted insecurity and discomfort rather than confidence and easiness. For the individual who grows into adept with more single-handed skills, diminish the need for trying a prosthesis. Whereas in lower limb amputation the reverse always happens. The individual with a lower-limb amputation mostly selects to walk with prosthesis nearly full-time regardless of their single or full leg amputation. So, the choice of continuity and use of such device is purely depending on the comfort level factor and expertise in the use of the person in need.

A lot of upper limb amputees sometimes use the prosthesis only for a portion of daytime either for specific everyday jobs, or not at all. There is some difference in mentality with above elbow and below elbow amputees. A person having amputation over the elbow suffer more while carrying things with their residual limb. They make use of the core portion of their body to support such functions. This is always tedious, painful and demanding a high amount of energy. In contrary, a person with an amputation below the elbow can perform more easily than other levels of amputation. They can sometimes support the holding objects even with their pointed tip of the residue arm. The amputee can use both the arms (amputation arm and unaffected arm) for firmness and stable while sitting or in leaning position. Apart from how natural is the artificial arm, the frequency and skill of the users depend on the level of amputation. It acts as a significant vector in determining the abandon rate of the prosthesis. For example, about 80 percent of people with below-elbow amputations use their prosthesis daily against 20 percent of above-elbow amputations. There are several reasons for this dramatic difference [4]. Only 7 in 10 upper-limb amputees in developing countries were satisfied with their prosthesis, compared with more than 9 in 10 lower-limb amputees.

It is evident that people with disabilities have more healthcare needs than others. However, amputees and people with disabilities in developing countries are further disadvantaged from their economic limitations that in turn, make them unsuccessful in getting proper care when needed. The lack of experienced and professional caretakers in these countries set the hindrance for providing insistent prosthetic facilities. For the high expertise and deep knowledge in various shaping, moulding, fitting and adjusting of artificial limbs, training programs conducted by federal officials in these regions are very uncommon. The low per capita income of these countries sustained the deficit in supplying qualified technicians for the purpose. Some of the studies done by the World Health Organization (WHO) clearly mentioned that there is a large shortage of trained prosthetic technicians all over the world, especially in the underdeveloped countries. Around 40,000 people are currently short in this profession and more than 50 years is needed to train at least half the required technical personnel [4].

Another issue is regarding the reach and accessibility of the prosthesis components. Major industrialized countries are the primary source of supply of the main parts, which are of high cost and not always aligned with the natural environment, climate and geographical existence of the user [6]. Moreover, many sophisticated parts are sensitive to the tropical climatic conditions making the lifespan of the same into a very small period of 2 or 3 years. The compensation of import cost is not a practically possible thing in such speedy wear and tear conditions. Over the lifetime, this frequent replacement of arm is not always affordable for the average or below wages people. In many low-income countries, a clear majority of amputee population are refugees, beggars and labor workers. Those who are burdened with earnings for survival will look for a vital option, a cheap compact arm without compromising the functional capability. Though the rate of artificial limbs may vary with regions, the production cost associated with those limbs can cut to the minimum amount with the wise selection of components and compatible design. The comfort and effectiveness of a prosthesis are largely governed by how well it fits onto the remaining part of the patient's own limb [44].

Access to proper healthcare is one of the major challenge faced by amputees in all developing countries [45]. Due to lack of public health facilities and funds, the expenses of high-quality artificial limbs are not affordable by amputees in the rural world. Additionally, most of the artificial limbs are typically designed for developed lifestyle making them unsuitable for the rural environment. Regarding the major provision of prosthetics services in developing countries, there are several factors determine the choice of an artificial limb. Along with design viability, replacement and maintenance, level of comfort, overall cost, cultural and religious backgrounds are factors which decide the wide acceptance and use of artificial assistive technology. These choices are not always fulfilled due to lack of funds and trained professional assistance [46]. By making artificial healthcare facilities affordable and accessible for all persons with disabilities these barriers can eliminate. The increasing rate of amputations and growing demand for prosthetic limbs demands the need for a cost-effective alternative to current technology. The solution is to design a hand with commonly available resources and simple control technology offering basic hand functionality having reduced cost well below than most affordable myoelectric hands available in the market.

Though there are lots of advancements in the prosthesis, the disabled people are still living as an underserved population. The abandonment of most of the advanced assistive solutions can be summed up as the following reasons.

- Upper limb prostheses are more complicated and expensive than lower limb prostheses.
- Maintenance of health condition of the residual limb.
- The user's activities may get hindered using a prosthesis. Moreover, some prosthetic components create problems for the user.
- The high cost of manufacturing, assembly, and maintenance of parts, frequent replacement of parts and large weight.
- A good prosthesis is still beyond the reach of most people living in developing countries. The manufacturing materials are in the developed countries and accessibility of such materials is a hard thing for common people.
- Not always suitable for physically demanding occupation. Adaptation with attitudes and routines of the user is important.

- The overall cost involved.
- Difficulty in exposure to adverse weather conditions, temperature range and chemical corrosion.

Despite many products and techniques available, the advancement benefits the society in an extended way. However, the real-world contribution of artificial rehabilitation solutions is getting down and the rejection rate of design models are still increasing. The main influencing factors of selecting prosthesis are always cost efficiency and comfort rather than their technological advancements. Moreover, the easiness in maintaining and replacing also plays an important role in the selection of prosthesis.

The basic goal of any prosthesis is to improve or restore the function of the physically handicapped person. With the proposed ERB based low-cost assistive technique, amputee people can develop self-confidence which keeps restrain them from public interferences. Use of prosthetic limbs helps them gain a better outlook on their life. They feel less discomfort with their conditions and can develop an ability to blend in an effortless way with the society. This model can be essential to offer better service to physical disability consumer in the rural world on their activities of daily living, pain levels, psychological well-being, social participation and subsequently improved quality of life [47].

Today, the developing world prosthetics is facing issues on their available resources and technologies for physical rehabilitation. Funding is also an issue for amputees in these places, as they lack the necessary income to purchase the devices [48]. To contribute a better understanding of current challenges and solutions of assistive technologies in resource-limited environments of developing countries, it is important to carry out an investigation to classify the real problems in rehabilitation solutions. In many countries, access to latest assistive technology within health and welfare schemes in the public sector is very poor or doesn’t exist. Currently, the assistive products industry is only serving high-income markets. Also, lack of adequate funding delays the nationwide service delivery systems. Challenges to distribute prostheses in low-resource settings at an affordable price force the patients to take donations from Western countries. But, it may not always be suitable for the user and the mismatch leads to develop additional difficulties. Large sector poor people are depending on these charity services who are delivering pre-used products. These are often not appropriate for the user or the context, and lack mechanisms for repair and follow up [6]. With second rate equipment amputees not only have directly impacted on their physical wellbeing; but also lead to mental health issues such as disappointment and depression. Not much focus in health care setting is placed on the prevention of mental health issues of amputees. With proposed model, people will benefit from increased health and wellbeing.

In order to justify the accepted measures in rehabilitation by comparing the prosthetic sources, their providers and costs, the provision of assistive technologies needs to make affordable [17]. The main reason for not possessing such provision of assistive technologies in the rural world is due to the lack of affordable solutions. To ensure equitable services delivery at the lower level of society, the use of local resources, their collaboration, and coordination with the consideration of cultural factors [49] is important. Along with other amputee rehabilitation services, governmental policies should be made available as a support with documentation to aid in the patient’s ongoing health provision and care. So, it is relevant to have access to appropriate prosthetic services which should begin in the acute phase and continue if required as part of lifelong management.

The new proposed design with ERB sensor not only acts as better artificial limb solution but also it can benefit governments from decreased health care costs. For the research community, this model can act as a motivational tool to increase their knowledge of ERBs in the design of upper limb prosthetic solutions. In future, more developments are possible with added features like ergonomic design, multiple degrees of function and advanced control.

3 Design of ERB sensor prosthesis – Method and materials

As discussed in the previous chapter the current prosthetic solutions have been witnessed with increased rejection rate because of the huge cost. This report will discuss about removing this economic barrier to an extent with the development of a new design for ULP prosthesis. Here, this entire chapter explains about the requirements of the proposed system, the design stages and operations in detail. The proposed design model can be outlined as in the block diagram given below in Figure 10. The design and implementation includes a hardware section and a controlling software part. On the hardware side, two parts can be distinguished:

1. The electronic system that acquires the myoelectric signals and generates the control signals.
2. The gripping hand that is the terminal device of the prosthesis. It is a simple two-finger hand able to perform basic grips that serve as an aid in the user's daily life.

Arduino Nano board and the Arduino IDE form the myoelectric control system. It is responsible for processing the digitized myoelectric signals. Depending on detected sensor signal amplitudes, Arduino microcontroller generates the control signals to operate the servomotor of the artificial hand. Also, for the user comfort, the controller coordinates feedback signals from the of moving hand.

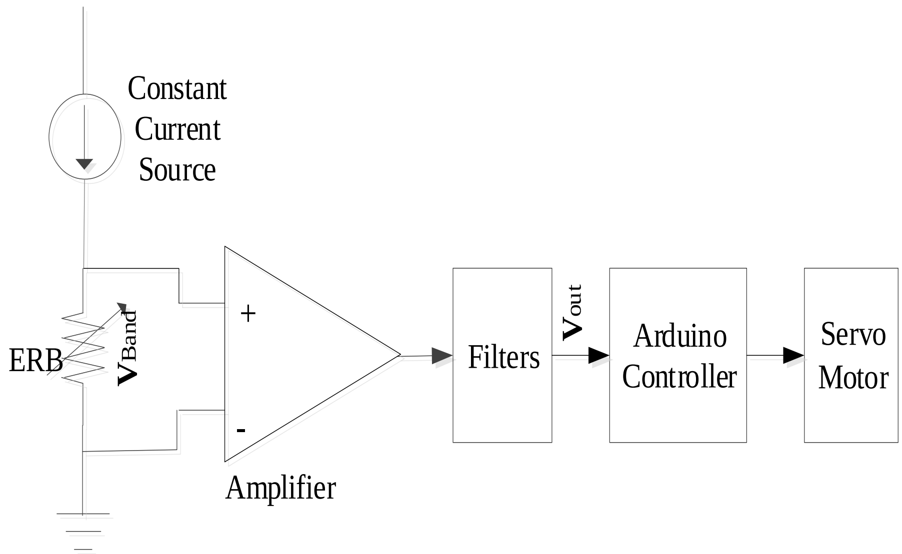


Figure 10. General layout of the prosthesis using ERB sensor.

Since the proposed design is a myoelectric activated hand, acquiring the myosignals is the important stage in the whole design. The potentials from muscles are the function of a group of selected muscle fibers that involved in flexion and locomotion. The larger the strength of electrical signals from fiber more is the tension generated from the muscles. Therefore, a generic correlation exists between the myopotentials and the muscle tension. SEMG signal acquisition systems are traditionally composed of pre amplifiers, instrumentation amplifiers, analog filters and multiple gain stages for the individual channel [50], [41]. To meet the goal of cost reduction using minimum components, a newer approach must be explored. As previously mentioned, to quantify the muscle tensions exerted by a group of muscle fibers, an ERB transducer can successfully employ onto the surface of the muscle [14].

In the proposed design, the ERB sensor signals are processed using an Arduino controller in the subsequent processing stages. Using the built-in analog to digital converter, the signal is digitized to produce PWM signals required for actuation of motor.

Along with the general structural, a light weight 3D printed artificial gripping hand finishes off the design to give a full sophisticated arm look. Initially, the design was planned with a haptic feedback to the user, and later a second phase was added with an external closed loop control of motor. To get a clearer understanding of the proposed design, all details are discussed in the following sections.

3.1 Components used in the project

3.1.1 Electro resistive band sensors

The electro resistive bands (ERBs) are a type of transducer composed of a cylindrical conductive rubber band (see Figure 11). The band is made of carbon impregnated rubber; with a diameter of about 2mm. Its resistivity is about 140- 160 Ω/cm [19]. ERBs functions on the principle that whenever the length of wire increases, its resistance also increases. The band changes its electrical resistance when it is stretched or vibrated. In a non-stretched mode, the variations in resistance is about 300-400 ohms per inch [51].

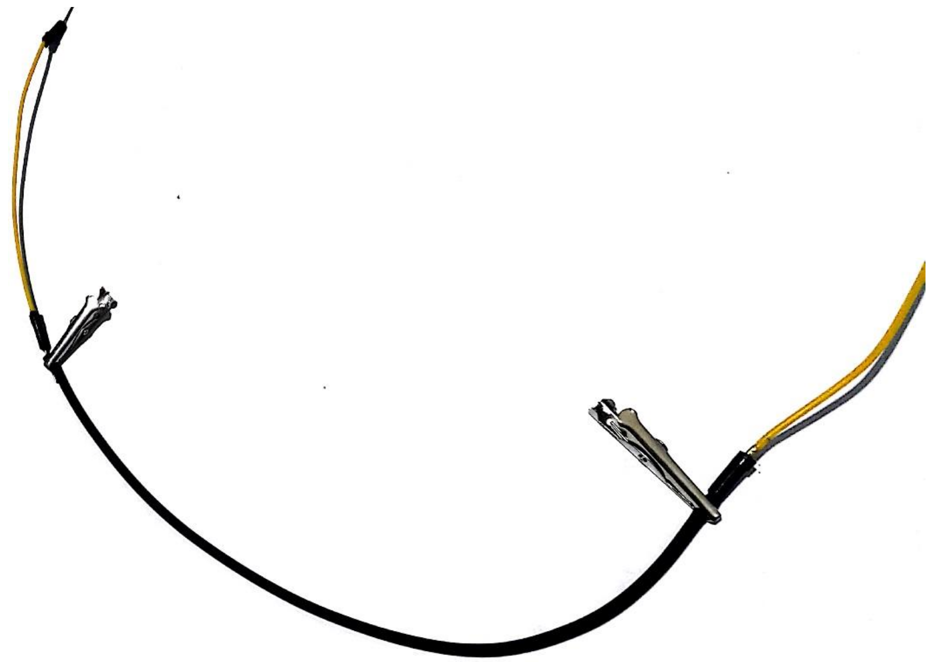


Figure 11. ERB stretch sensor

This property is made use in the current design of SEMG sensor based prosthesis. The amount of change in band resistance is somewhat proportional to the amount of muscle tension produced by flexion. Although a non-perfect linear change of resistance with stretch has been demonstrated [20] for this sensor; nevertheless, it can be employed to acquire Analogous of Surface Electromyography (ASEMG) signals. The impregnated rubber provides enough extension while flexion and can be stretched up to 25% longer than its normal length. Therefore, it can be easily adapted to any size of the residual arm. Above 25% of normal extension, the band will reduce its property of maintaining proportional impedance variations. It will no more be used as SEMG sensor, as the resistance varies from batch to batch. This will significantly affect the actuation control. But, without much variations the conductive rubber band can be wrapped around a stretchable fabric piece that can easily wear without any direct electrical contact between the sensor and the body [21]. This is a great advantage over the use of other EMG electrodes as it creates a cumbersome amount of difficulty to the normal use [52].

3.1.2 High-gain, low-power transducer amplifier (LM324)

The LM324 amplifier includes four individual, large-gain electronic voltage operational amplifiers which are designed to operate at high stability. This IC can work for wide range of voltages from a single power source. The action of multiple, dual power and low-power supply are also possible with LMx24 series. The frequency of op amps are internally compensated and continuous draining of current is not depending on the magnitude of the supply voltage [53]. The dual in line pin diagram of LM324 is as shown in Figure 12 below.

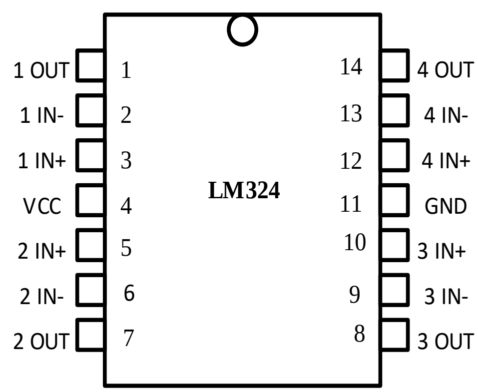


Figure 12. Pin configuration

The general features of LM324 comparator can be summarized as follows:

1. Unity gain frequency compensated
2. Large DC voltage gain – 100 dB
3. Supply voltage range – 3V to 32V
4. Bandwidth – 1MHz

The main application areas of this 14 pin IC include transducer amplifiers, signal oscillators, DC gain blocks and all other common operational amplifier circuits that can implemented with a single supply [53].

3.1.3 Arduino Nano

Arduino is an open-source development board of ATmega Co. electronics. Using the programmable IC in the development board various circuits can be developed to read different inputs such as sensor data or on/off status of a button. It also can be programmed to provide an output as activation of a motor, turning on/off devices and remote level of control. Arduino Nano is one among the Arduino boards which are compact, complete, and breadboard-friendly platform based on the ATmega328P (Arduino Nano 3.x) [54].

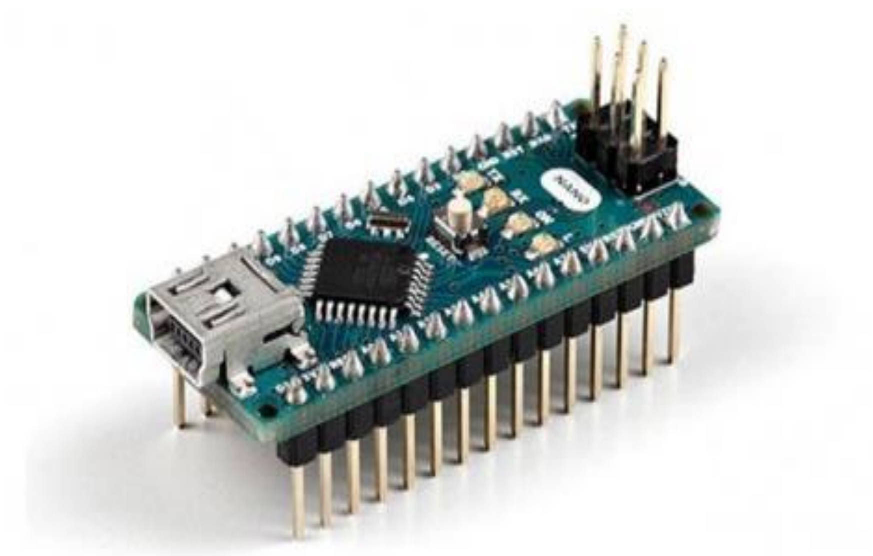


Figure 13. Arduino Nano [54]

Arduino Nano is a small size portable microcontroller working with a small-B type universal serial bus (USB) cable in the place of a typical serial connecting cable.

The Nano suits well in compact projects as the board is only the size of a normal match box. The controller uses easy programming language which is very much related to embedded C or C++ language. Apart from that this is a moderately inexpensive cross platform controller easy to use for the beginners. Clear programming environment makes it more flexible even for advanced users with the extensible hardware and software environment. Moreover, the significant features of this microcontroller (see Table 2) make it suitable for the current project design. The Arduino Nano acts as the CPU for the proposed system and utilizes information from the input processed SEMG signals to control the hand action. The different types and number of pins on the controller satisfies the requirements for the rest of the circuit components.

Table 2. Arduino Nano features

Microcontroller Board	**Nano**
Controller	ATmega328
Clock Speed	16MHz
I/O’s	22
Analog Inputs	8
PWM’s	6
Functioning Voltage	5V
Supply range of voltage	7-12V
Flashing memory	32 KB
SRAM	2 KB
EEPROM	1 KB

3.1.4 3D printed mechanical hand:

A gripper is a basic aid which can be adapted to the purpose of an object to be grasped and controlled in a skillful manner. The basic shape of a gripper is similar to the forefinger and thumb of the human hand aiding perfect grasp. Just like our hand, the gripper allows all the basic functionalities such as holding, tightening, handling and releasing of an object. Hence, the representation of the hand prosthesis in this thesis work is done in the form of two finger gripper enabling one degrees of freedom.

For the low-cost design of prosthesis, 3D printed hands are the best solution. The important advantage of the 3D printed design is that it can be improved with customized hand designs [55]. So, an open source project design available online from 'Think verse' group was appropriately modified to make suit with the current purpose. And, all pieces of the prosthesis were in-house printed using Maker Bot 3D printer with the PLA material.

In the prototype model, it is constructed in such way that, the hand is free to create 120 degrees. The connection between two fingers is through the spur type gears providing simple open and close movement. Since I used a gripper on the hand, it must be robust enough to track the entire weight of the interface part with the residue arm. Figure 14 shows the printed gripper part of the artificial hand. The ultimate mass of the printed hand is 48g and is very easy to assemble onto the selected servomotor.

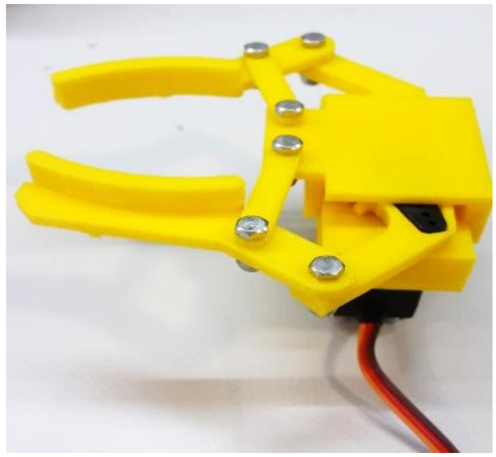


Figure 14. Model of 3D printed hand

3.1.5 DC Servo motor – Tower Pro MG995

Servos are the special type of DC actuators with precise control of position and velocity. It consists of a built-in gearing and position feedback loop for the closed control mechanism. The input to the control is digital or analog position commands for the output shaft. Every time the output position is compared with the command position, the input to the inbuilt controller. If they differ, an error signal is produced to make the motor rotate to bring down the error as possible. When the command position is reached, the error becomes zero and motor stops. These motors are mainly developed for making hobby circuits, radio controlled models and automation, that they are not meant for large industrial applications. Most of the servos can rotate about 90 to 180 degrees and with slight internal modifications, some can even complete full rotation, 360 degrees or more.

In this project, I used a servo from Tower pro MG995 (see Figure 15) having 180-degree rotation. The specifications are summarized in Table 3. It has three wires, one for supply, other for ground and PWM signal.

Figure 15. DC servo MG995 [56]

The connection of servo to the controller is very simple; connect the brown wire to ground potential, the red wire to a voltage source (typically 4.8-7.2V), and the yellow wire to a PWM signal source (such as from the microcontroller). The position and velocity control of servo is done by varying the pulse width of square wave from 1 to 2ms.

Table 3. Tower Pro MG 995 details

Stall torque	10 Kg/cm
Speed of operation	0.20 s/60 degree
Working voltage	4.8 – 7.2 V
Operating current	500mA-900mA
Stall current	2.5A
Dead bandwidth	1 µs
Allowable variations in temperature	0 ºC – 55 ºC

Total weight	55gm
Dimensions	L -1.6in, W-0.8in, H-1.69in
Motor type	3 poles
Gear type	Metal
Rotation	Dual bearings
Modulation	Digital

3.2 Working Principle

The working principle of the proposed prosthetic model is a very simple on (fingers close) and off (fingers open) control. With sensor on residue hand, if the user makes a sudden muscle flexion, the myoelectric signals are detected via the ERB band. The signals are then processed in different stages and filtered to get a smooth PWM signal from the Arduino controller, this is going to activate the DC servo motor in prosthetic hand. The working of the prosthesis without any feedback can be simply represented as the flow chart shown in Figure 16. The feedback inclusion and effects their detailed discussions are in following chapter.

A set value of flexion threshold is determined from the user beforehand and is used as the level trigger for the actuation of motor. In the first stage of development the raw transducer signals are processed as a near equal of RMS signal with 50 samples every time is generated to compare with the onset threshold limit. Once the limit is exceeded, the trigger is on to generate the PWM signal. So, for any of the careless and accidental motions of residue arm, ERB bands on muscles detect sensitive variations. To avoid such unavoidable circumstances in real world life the threshold limit for triggering is set so high as it won't affect the power drain of battery by continuous actuation of motor.

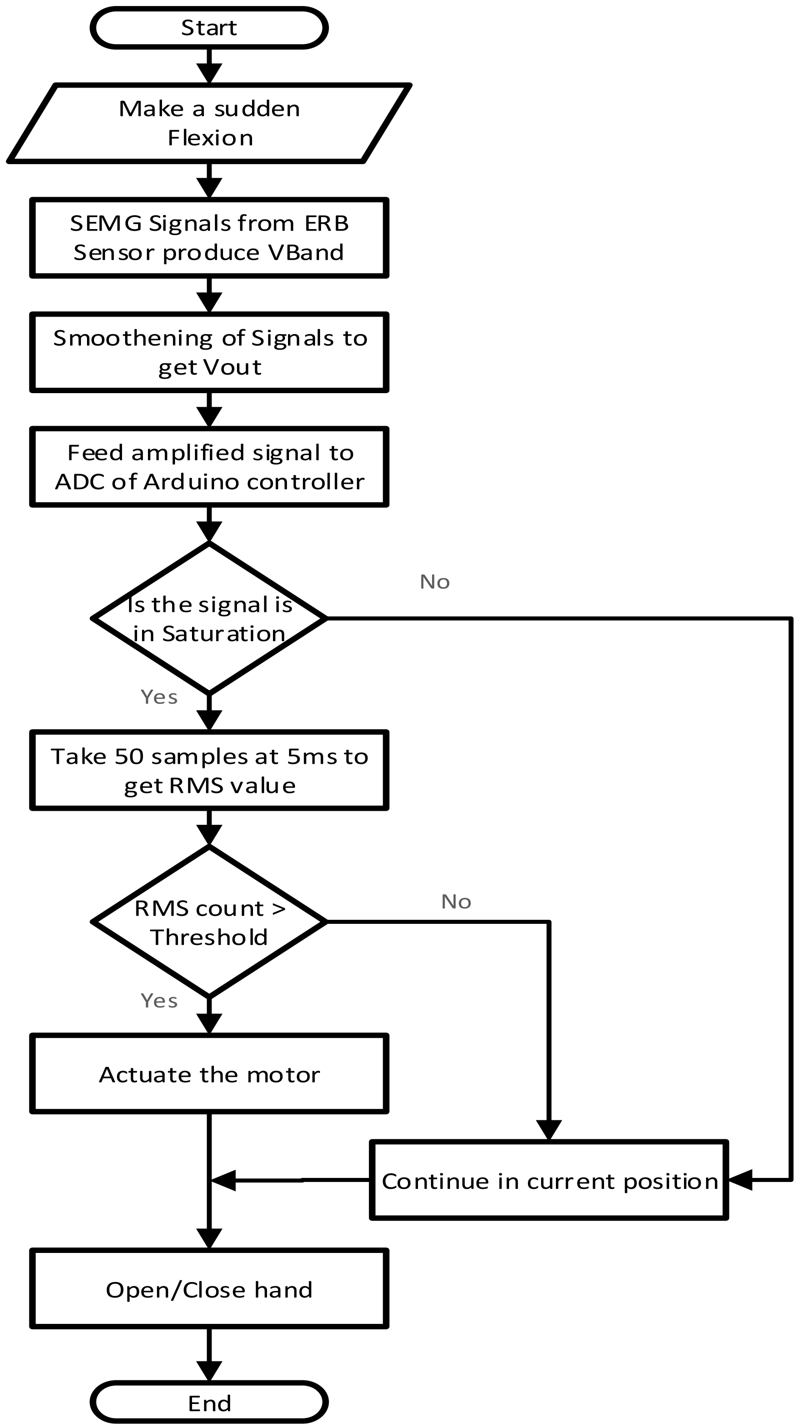


Figure 16. Flowchart of ERB hand

With the movement of the DC servo motor shaft, the user can get the sensory-haptic feedback along with the closed loop control of motor position. The complete model of prosthesis is developed in two stages: one with simple sensory feedback and stage two is with an additional closed loop feedback. The overall structure of model design of prosthesis is made up of the following five parts:

1. ASEMG detection Circuit
2. Microcontroller unit
3. Mechanical claw with the actuator
4. User feedback circuit
5. Closed loop feedback

The ASEMG detection circuit consists of contactless ERB sensors for sensing volumetric muscle activities and a Quad op amp circuitry for the signal processing. The mechanical claw which is followed by the detection unit contains a 3D printed hand and a DC servo motor. The mechanical hand works according to the signals from ASEMG unit. And, the feedback circuit is provided for the real time sensory (haptic) feedback to the user. The feedback is given in the form of vibrations from a buzzer and position control. The circuitry for haptic feedback is like ASEMG detection unit with extra sections for additional gain of control. The model is provided with support up to three ERB sensors, of which one is to be used to close the sensory feedback loop with the user. However, in the concept design, only a single ERB is used as a volumetric sensor. Additionally, the circuitry for second ERB could be used to scale up the design with changes in preferences.

In the design, for the practical realization of prosthesis, I did use only readily available components. An Arduino Nano board was selected as the controller for ease of prototyping and use. Moreover, the detection circuit and the feedback circuit was fully implemented on two separate printed circuit boards (PCBs) for much simplicity of control. The PCBs are developed using standard “thru-hole” packaged components that can be also assembled onto a universal prototype board. The following sections will give more details of design together with the full assembly and testing. As mentioned above, the details of the closed loop feedback section have included in the upcoming chapter.

3.3 ASEMG Detection Circuit

The main part of this unit is the ERB sensor for detecting muscular electric signals. As mentioned, ERB sensors are contactless wearable sensors; hence, skin preparation is not at all required. It can be worn on the top of the user’s clothes or on a nonconductive material. In the prototype, it was simply stitched onto a standard elasticated elbow support for the simplicity of use. However, this can be attached using pins to existing clothes or bandages. Contrary to electrodes, ERBs are not as sensitive to any external electromagnetic disturbances; therefore, the system can work in almost any environment [57], [52]. ERBs’ resistance changes with the band stretch, thus, to use this sensor to detect ASEMG. Normally, the SEMG signals have low range of amplitude values about 10 µV to 4 mV. The amplitude is highly unstable and is continuously varying depending on many factors (such as the proportions of the muscle in context, sensor stretch and sensor distance) [57]. This need to first transform the changes in resistance to elicited voltage changes. That means the resistance variations from ERB sensors should be converted to voltage signals for further stages of processing. The fine myoelectric voltage signals from the band obtained as variations in resistance should convert into potential variations by the continuous polarization of the band using a fixed current source [14]. This was achieved by using a small DC polarization current. For the purpose of continuous polarization, a constant current of 1mA was injected into the band sensor all the time to obtain voltage drop across its terminals. The output of variations in resistance are not enough to produce PWM signals to actuate the servo drive. These feeble voltage changes are then amplified and fed to the ADC using a single operational amplifier. In the design, a polarization DC current of ~1 mA is implemented using a simple biasing network composed by the resistors R1 and R2 and two small signal diodes. The full circuit for ERB sensors to detect SEMG is depicted in Figure 17.

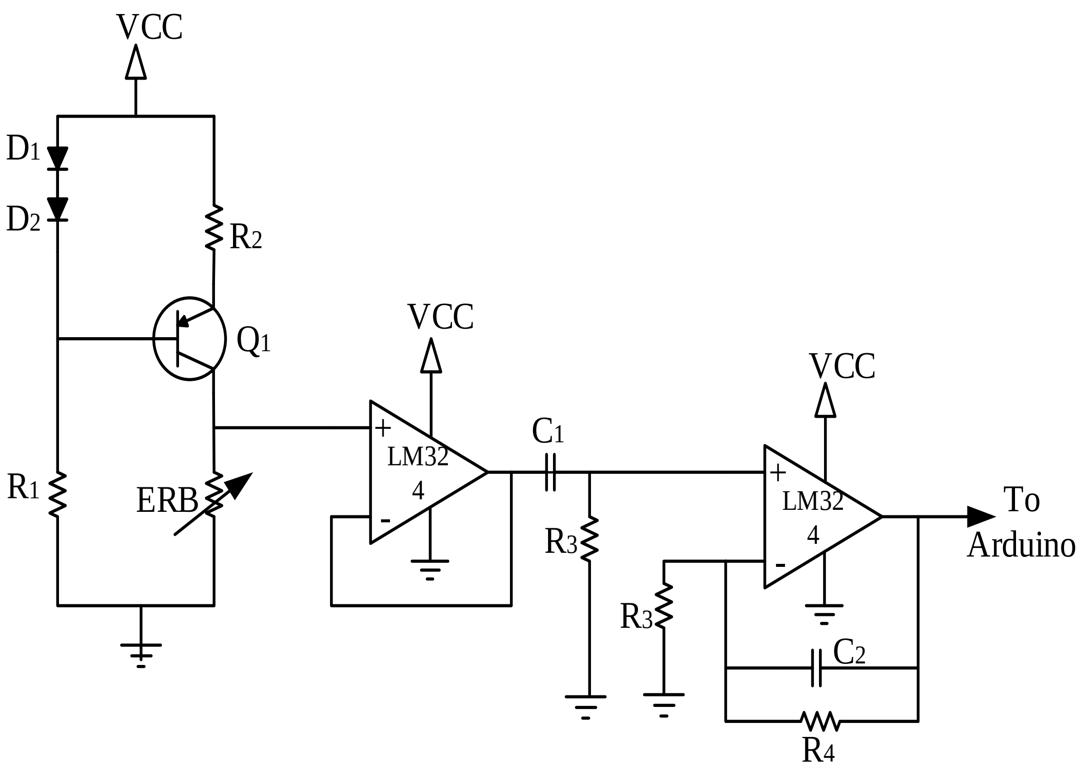


Figure 17. Circuit diagram of ERB sensors for ASEMG detection, limited to only one ERB

where R_1 = 5K, R_2 =470Ω, R_3 =100K, R_4 =10M, C_1 = 7.9µF, C_2 =68pF

Although the control has been provided for two ERBs; in the prototype type setup, I employed only one band as this is adequate to prove that the ultra-low-cost ULP is easily attainable. The DC polarization current is designed around a single BJT transistor (BC557B) which is not so critical and can be replaced by any of its equivalents.

Table 4. ERB band data from ASEMG unit

Band Voltage	**Relaxed**	**Stretched**
$\mathbf{V_{Band}}$	1.344	2.131
$\mathbf{V_{Out}}$	2.670	3.716

The current source is designed for approximately 1mA. With changes in supply voltage, fluctuations may arise in the current flowing through the circuit. This is achieved by replacing a normal resistor bias section with a voltage reference or signal diode. To find, a voltage loop equation is formed around D_1, D_2, Q_1 and R_2 loop. The KVL equation for the loop is given in equation 1. V_{D_1} and V_{D_2} are voltage drop across the diodes, V_{be} is the base emitter drop in transistor. Voltage drop in resistors R_1 and R_2 are V_{R_1} and V_{R_2} respectively.

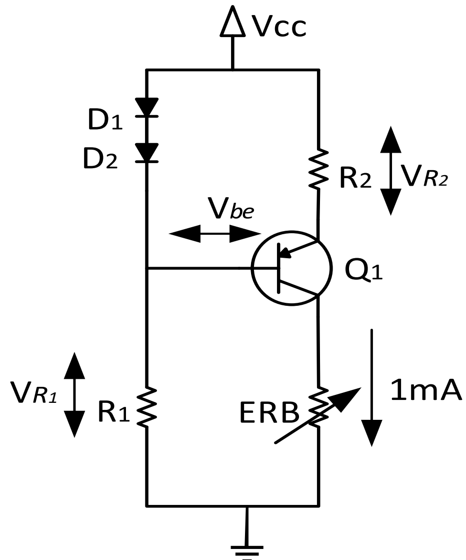


Figure 18. Constant current bias circuit

$$V_{D_1} + V_{D_2} - V_{be} - V_{R_2} = 0 \quad (1)$$

$$V_{R_2} \approx I_o * R_2 \quad (2)$$

$$I_o \approx \frac{(V_{D_1} + V_{D_2} - V_{be})}{R2} \quad (3)$$

Assuming, the diode voltage drop is below 0.5V and it nearly equals the transistor base-emitter voltage drop.

V_{D_1} =0.45V

Assume, $V_{D_2} \approx V_{be}$

$R_2 = \frac{V_{D_1}}{Io}$ / = 0.45V/1mA = 450 Ω.

R_1 = 5K Ω.

The polarization current induces a voltage drop across the ERB, which is the function of the band's base resistance (DC mean) and from the stretching induced by the muscle contraction. This signal is conditioned (filtered and amplified using two operational amplifiers from the four available inside the LM324.

The LM324 is a general-purpose quadruple operational amplifier and this component is being selected due to its widespread availability and low-cost. Because its characteristics are not critical, it can be easily replaced with any of the available general purposes operational amplifiers. One single LM324 is then enough for the entire ASEMG circuitry. The first amplifier is just used as voltage follower. The voltage follower helps to isolate output stages from any external disturbances and it reduces the high input impedance. Followed by this stage the output is fed to a high-pass passive cell and then this to an active low-pass cell with a measured gain of ~10 V/V.

$$A_v = 1 + \frac{R4}{R3} \tag{4}$$

The calculations for resistance and capacitance are done from the setting of corner frequency (cut off frequency). The formula for corner frequency is

$$f_c = \frac{1}{2\pi RC} \tag{5}$$

where f_c = the cut off frequency;

R = Resistor value, C = Capacitor value

So,

$$f_{c1} = \frac{1}{2\pi R_3 C_1}$$

$$f_{c2} = \frac{1}{2\pi R_4 C_2}$$

The measured corner frequency thus obtained for high pass filter is ~ 0.2 Hz and for the low pass filter is ~ 235 Hz. The gain and cut-off frequencies of the signal in the conditioning circuits are not very critical; so, in the prototype setup a standard high-tolerance is selected with low-cost passive components. So, whenever the user makes a sudden flexion, the detected ASEMG signals are converted via Arduino inbuilt ADC to drive the prosthesis servo.

3.4 Microcontroller unit

The process controller of the design is based on an Arduino Nano microcontroller. Due to very low power consumption (Nano watt technology) and inexpensive microcontroller category, Nano suited better for this design. The input signals to microcontroller are band-limited to 500 Hz hence it is suitable up to a sampling rate of 1kHz. The ASEMG detection circuit give signals to the mechanical hand after the detailed processing within the controller. In Arduino, the voltage outputs from ERBs are digitalized using the embedded 10-bit ADC. Therefore, the output for muscle flexion is obtained as continuous digital counts. In which a threshold value is selected and set to cut off the weaker signals from spurious and unintentional muscle activities. The threshold level varies from person to person according the capability of muscle flexion.

The threshold to actuate the servomotor is drawn in a calibration phase where the user is asked to perform one voluntary reliable contractions; the value of 50% is then used as threshold. If the flexion count value is more than the set threshold, the controller calculates the time based sampled average measured during the muscle contraction. This calculation produces a result like an RMS calculation; however, it is computationally lighter than a true RMS calculation across 50 samples. When the signal is above the set threshold, a proportional PWM signal is generated for the actuation of servo motor. If the signal is lower than the threshold value, the hand remains in the default position.

The analog pins in the controller reads the analog signal and then do the digital conversion using the command ‘analogRead ( ). In the Nano board, the pins are represented with a letter ‘A’ in front of their number label (from A0 to A7). The numbering specify that these pins can read analog voltages connected to corresponding pins. The analog to digital converter available in the Arduino Nano is a 10-bit resolution ADC; meaning it could detect 1,024 (2^{10}) discrete analog levels. i.e. For 0 -5 V analog voltage value the digital resolution can be obtained from 0 -1023 values.

For the servo control, servo setup is attached to the digital pin of Arduino using the command servo.attach (9). In Arduino Nano the Servo library supports only two digital pins: 9 and 10. The command to represent servo is available in void loop as the variable pos (position). This variable name which is used to evaluate the state of servo communicates with servo to remain in or leave the position changing loop. The variable decides the incrementing as well as decrementing of the counter include. More importantly, this variable represents the servo position value for the servo.write(pos) statement. As the loop runs, the pos value begin at 0 is incremented, and thus the position of the servo is changed five degree at a time by using the servo.write(pos) statement. The servo rotation is based on the code by using the time delay function delay( ) after each individual or group of write( ) position movements.

```
for (pos = 0; pos < 180; pos += 5) { // increment in steps of five
 servo.write(pos);
 delay(15);
}
```

For prototyping purposes, the design is controlled as two stages. One with haptic feedback and a push button is provided for the user to change the direction of the claw movement. Later, the second stage of modifications are made in the control part by introducing closed loop position control. However, it would be easy to target a secondary muscle using the second provided ERB polarization/sensing circuitry to select the servomotor direction.

3.5 Mechanical Claw with actuator

The processed signals from ASEMG unit through Arduino controller are used to drive the mechanical hand. The mechanical hand is a 3D printed claw shaped structure with two fingers and a servo motor as shown in Figure 11. A small DC servomotor MG995 is used for actuating the artificial hand. This motor has been selected for its low- cost, high-torque and wide range of motion, of about 180 degrees (90 in each direction). In the design, despite the motor can rotate approximately 0 to 180 degrees due to the restrictions of hand model printed the maximum span obtained from the motor is limited to 70 degrees. As mentioned earlier, the servo model is not specific and it is easy to replace this motor with any equivalent one. The total weight of the mechanical hand is approximately 103g (including the DC servomotor).

To detect an eventual motor stall and/or any over-current that could damage the prosthesis, the current drawn by the servomotor is monitored using a sampling resistor and an amplifier (see Figure 14). Although this circuit is implemented on the PCB, for this specific stage 1 of design, I excluded this signal because the sensory feedback control implemented is working well and never experience the motor stalling issue during the trial experiments.

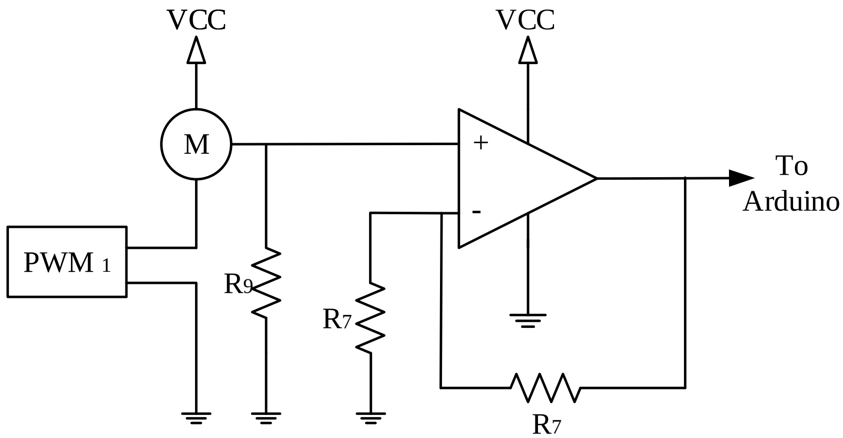


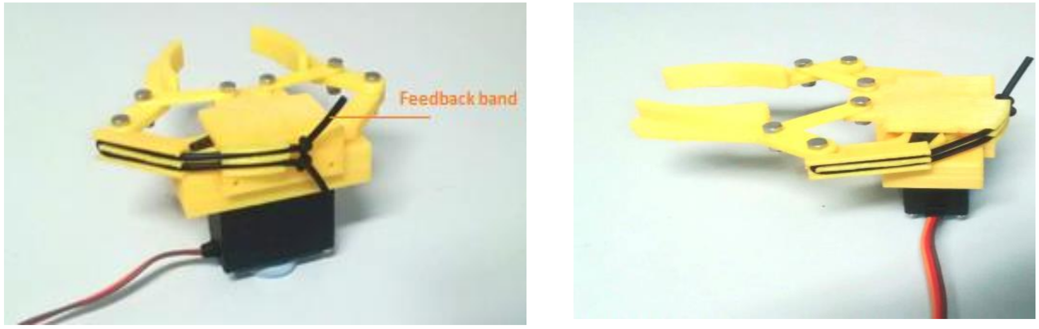
Figure 19. Circuit diagram of Servo motor current sensing

R_7 = 1K, R_9 =10Ω

3.6 User Feedback Circuit

It was necessary to design a method for gathering force data on the hand for the microcontroller to make decisions on actuation. Without such feedback, the motors driving the fingers would simply actuate as much as possible and stall when the fingers cannot move any longer. This does not allow for fine motor control, which would not be beneficial if the user wanted to, for example, shake someone's hand.

To avoid unintentional injuries or damages to the prosthesis while gripping an object, we provided a real-time sensory feedback of the prosthesis aperture to the user. The sensory feedback to the user allows the control of the pressure applied when grasping an object. The degree of the prosthesis aperture is fed back to the controller using a third ERB. Contrary to the ERBs used for ASEMG, this circuit requires an additional signal conditioning stage (see Figure 18).



(a) (b)

Figure 20. 3D printed hand: a) Back view; b) Side view; in both views the ERB is visible (see text) and the connection to the circuitry have been removed to avoid cluttering in the figure.

As it is possible to observe in Figure 18, the ERB is placed across the two arms of the claw in such a way that it stretches when the prosthesis closes, the ERB, it is anchored in position by simple knots that are also used to make the electrical connection to the ERB. The signal from this third ERB is used to activate a little buzzer placed on the user's body and its vibration amount gives information about the current aperture of the claw. To remove any ambiguity, a little amount of intermittent vibration is given when the claw is fully open. Full vibration is instead used to flag that the claw is fully closed.

To customize the feedback according to the user preferences, additional controls over the zero-span circuitry as in Figures 19 and 20 are used. Zero span circuitry allows the user to decide the level of full vibration and low vibration by means of gain adjustment. Adjusting the value of the potentiometer labeled Rp1; the user can change the vibration intensity level indicating the fully closed position to a comfortable value. The little vibration buzzer is directly driven by the Arduino using a simple BJT circuit as in Figure 20. The potentiometer Rp2 can be used to vary the gain of the signal conditioning circuit to adapt to different configurations, namely size/shape of the prosthesis. All circuit is made up of low-cost active components and materials and is powered directly from the 5V DC supply output of the Arduino.

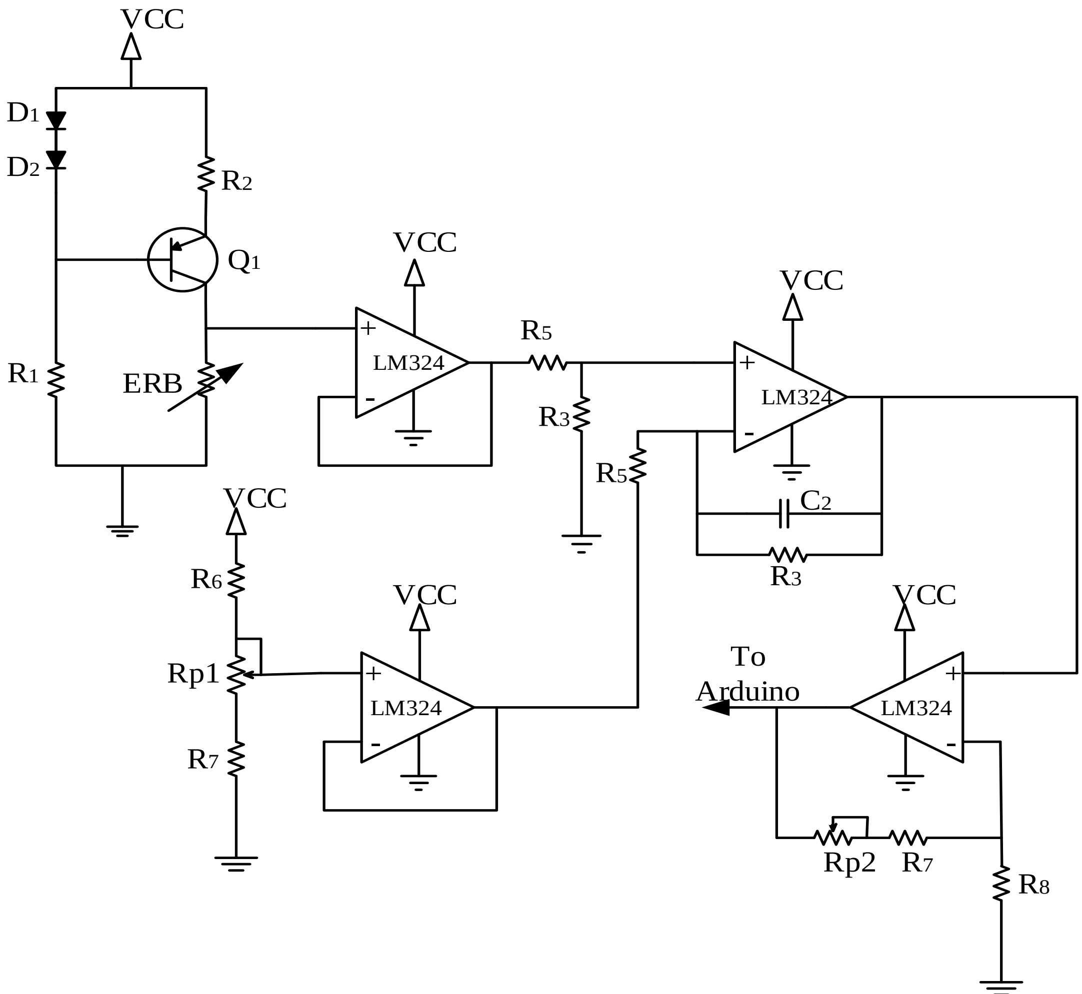


Figure 21. Circuit diagram of User feedback with span control

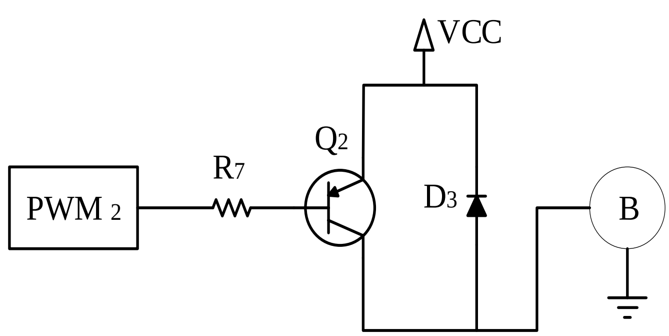


Figure 22. Circuit diagram of vibration buzzer in feedback control

where R_1 = 5K, R_2 =470Ω, R_3 =100K, R_5 =10K, R_6 = 39K, R_7 =1K, R_8 =56K,

R_{p1} =10K, R_{p2} = 100K, C_1 = 7.9µF, C_2 =68pF

The final design of this project is a one-DOF arms adapted for performing basic hand functions of hold/ grasp of an object from desired location. The prototype laboratory set up is given in Figure 21.

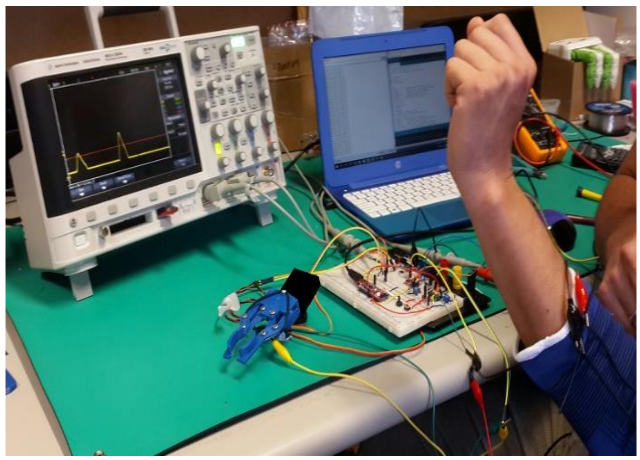


Figure 23. Prototype set up

4 Closed loop feedback control

The system without any feedback is known as open loop system. For the servo motor, without feedback there is no means to authenticate the position command has attained or not. But in a closed loop feedback system, the response data is used to alter the speed and path of the motion. To accomplish the desired outcome, an error check with available data is mandatory. In the same way to incorporate more versatile means control in the proposed model a closed loop feedback is essential.

In the first phase of the project, the prototype design was decided to finish at sensorial control with respect to the simple target customer needs and engineering requirements. Then a haptic feedback was realized successfully in stage 1. Through this feedback, the user would have a real time feel of adjusting the pressure of finger with change in its position. But, the physical reality of arm demands more than just a scratch level sensorial feedback. So, I decided to move on with an additional control strategy which will be bounded inside the scope of project. Thus, the idea of implementation of a closed loop feedback motor control came into the work design. This can be done by creating either a position feedback from the shaft of motor or by taking current/torque feedback from it.

One of the goals of this project is to design and develop a system that would automatically regulate the force the prosthetic hand applies to an object when it is gripping. To accomplish this, both position feedback of motor shaft and current feedback are to be identified, modified, characterized, and implemented on the control of hand. The feasibility of both methods was tested and demonstrated in this chapter in detail. As a second phase of design a closed loop feedback control has been introduced into the earlier design. While processing the sensor signal, rather than taking the nearly true RMS of the raw sensor data, the second stage has been developing by including the mean value of sensor signals. Both methods will work fine in terms of identifying the threshold level. For the simplicity and reduction in coding time, I chose for mean value in second phase.

4.1 Method I. Position feedback

In an ordinary DC servo motor, the position or angle feedback is obtained as the output data from the wiper of potentiometer (pot) linked directly with the shaft of the motor. The output obtained from the pot is so related to the location (angle intended) of the servo shaft. By the built-in model of servo details, the position of shaft is known to the internal circuitry control but it is always unknown outside.

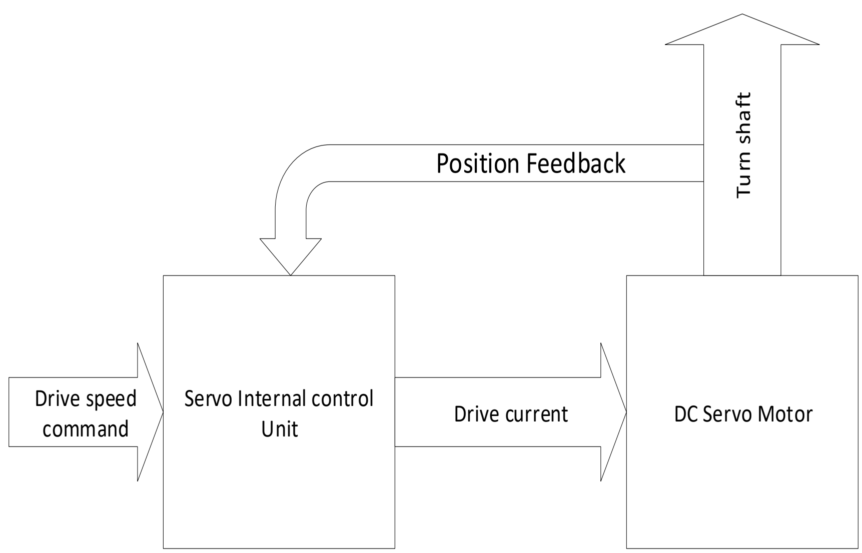


Figure 24. Closed loop (internal) servo control

In servo motors, the situation of the shaft is identified and given back to the internal control circuitry. When there is any mismatch with the data obtained and commanded the control unit regulates the flow of current to the motor to maintain required position. Similarly, every time it calculates the shaft speed and fed back the data into the control unit to maintain the demanding accelerator speed. The main issue with this type of control is that it is constantly communicate as closed loop inside the servo motor casing, but not with the main microcontroller [29]. Without any data to confirm what happens inside the servo control, it is impossible to use the servo in highly sophisticated and fine tune connections. Only solution to this is to get an external communication path from servos. This outer loop permits the microcontroller to act immediately with the deviation from desired output.

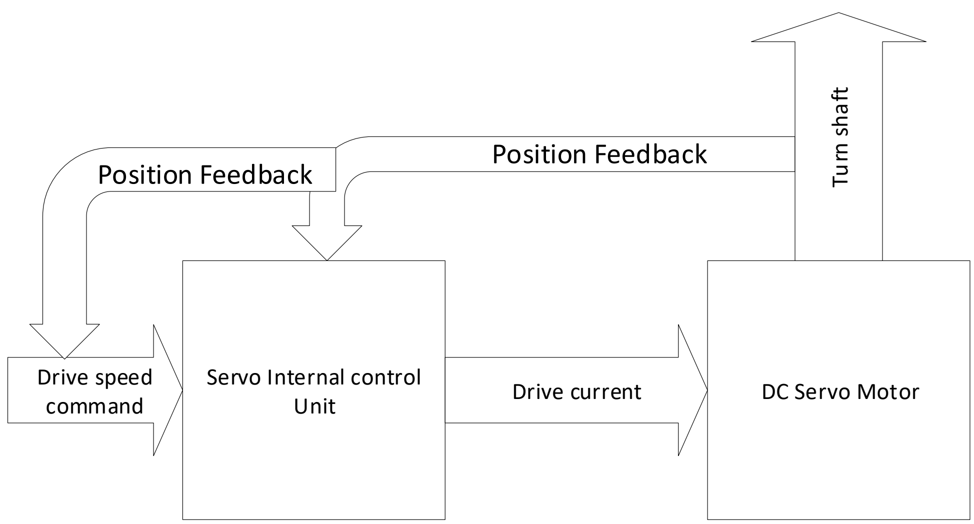


Figure 25. Closed loop (External) control of servo

4.1.1 What is the use of external position feedback in servo?

DC servos generally act with accordance to the command provided by the user, but if there are circumstances in which it ended up without the required position. This can be caused by various reasons including:

- Not sufficient in motor size
- Inadequate power supply
- Physical / External disturbances
- Electro-magnetic interferences
- improper assembly

In all the above state, if there is a communication between the main controller and servo is happening, then the issue will be attentive immediately by the user. And in some cases, though the motor is adequately sized and working flawlessly, the response time of motor may become longer than usual. With most of the situations, it is imperative to recognize whether the motor shaft has attained its position on specific command. Without feedback, in every servo programming the length of response is merely made on some assumptions. By giving certain amount of time delays in the code make the servo works satisfactorily for the simple applications. Aside from this the motor performs very slowly with hunting disturbances when it is trying to synchronize numerous motions or connections between the various fragments and additional sensors or even actuators. So, if the time delay provided are insufficient to track the changes happening, the servos will find it difficult to attain the looked-for state or position. The result is imperfect operation with jittering noises and even cause ruin to the running project. The improper delay can deny the normal working of servos mainly in battery powered projects. As the battery runs out of power, the motor commences to run slower than usual mode.

To avoid all these issues an external feedback is necessary. With position data as feedback the correct state of servo arms are available in any moment of time. The data of exact state of servo arm thus let accurate regulation of motor parameters. The feedback also can help the user to manually fix a random position and seize that same location for forthcoming reproductions These are accomplished by adding a new connection wire with the internal potentiometer of the RC servo motor. While the servo motor is working, the shaft changes the resistance of wiper in potentiometer. The value will rise or fall depending on the direction of motion of the motor shaft. By making use of this feature, the range of output voltage from motor can be determined. Using the range of shaft rotation, the output voltage is also matched up. Thu the voltage resolution or step change with change in angular position is calculated. To recognize the relative changes in voltage signal and angular position value, some calculations was carried out by pot wiper. The Arduino then turns to a data acquisition unit to accept the voltage feedback from the servo motor.

With extreme care the servo disassembled and internal board placed inside the casing has taken out. The board was only held with a small compression, so it was easy to lift. The wiper with the potentiometer was soldered with a new wire to get the position data. Before that, the maximum and minimum range of the voltage was calculated manually turning the potentiometer wiper. For the MG995 servo, it was between 2.49 and 0.54 volts.

A high resolution 10-bit Analog to Digital Converter(ADC), which is available on Arduino Nano controller, displays value between 0 - 1023 steps for corresponding analog voltage signal. From the 5V reference source ADC will yield ~ 5mV per step change of position. For the servo I have taken has a measured voltage difference of 1.95V (2.49V - 0.54V). The difference is obtained through servo feedback wire, so the ADC range would be around ~400 steps. The range is very impressive and enough for the current design method. However, the range can be enlarged by constructing a voltage divider on the controller A-Ref pin.

4.1.2 Calibrating the feedback

The servo arm is always attached with a small potentiometer. Whenever the position of the potentiometer changes, a signal from internal circuitry will try to match up with the commanded position. This will continue until they both match. For interpreting the feedback signal, the state of shaft is taken through resistance of pot in rotation. The white colour feedback wire is used to connect with analog read pins and it is available with the command code analogRead( ).

```
int fdbk = analogRead(fdbkpin);
```

The reaction signal is simply voltage signals from wiper of the potentiometer. Servo couldn't understand raw voltages, for a meaningful conversion as position calibration of servo is essential. By reading the feedback values at two known positions, we can interpolate the expected feedback values for every position in between. If a call "calibrate" is made in setup function, it will perform the calibration on the two points you specify. These servos operate over a range of about 0 to 180 degrees. For maximum accuracy, the minimumposition and maximumposition calibration points were chosen based on the range of motion required in the project.

```
void calibrate (servo servo, int A5, int minimumposition, int maximumposition {
  // change to the minimum position and noted the feedback value
  servo.write(minimumposition);
  minimumangle = minimumposition;
  delay(1000); // ensure enough time to reach position and settle
  minimumfdbk = analogRead(A5);// Change to the maximum position and noted the feedback value
  servo.write(maximumposition);
```

```
maximumangle = maximumpos; delay(1000); // ensure enough time to reach position and settle
maximumfdbk = analogRead(A5);}
```

After the calibration step, the position feedback is then again connected to Analog pin A5 in the Arduino for testing the nature of model prosthesis control.

The flowchart for the working of closed loop feedback was designed as shown below.

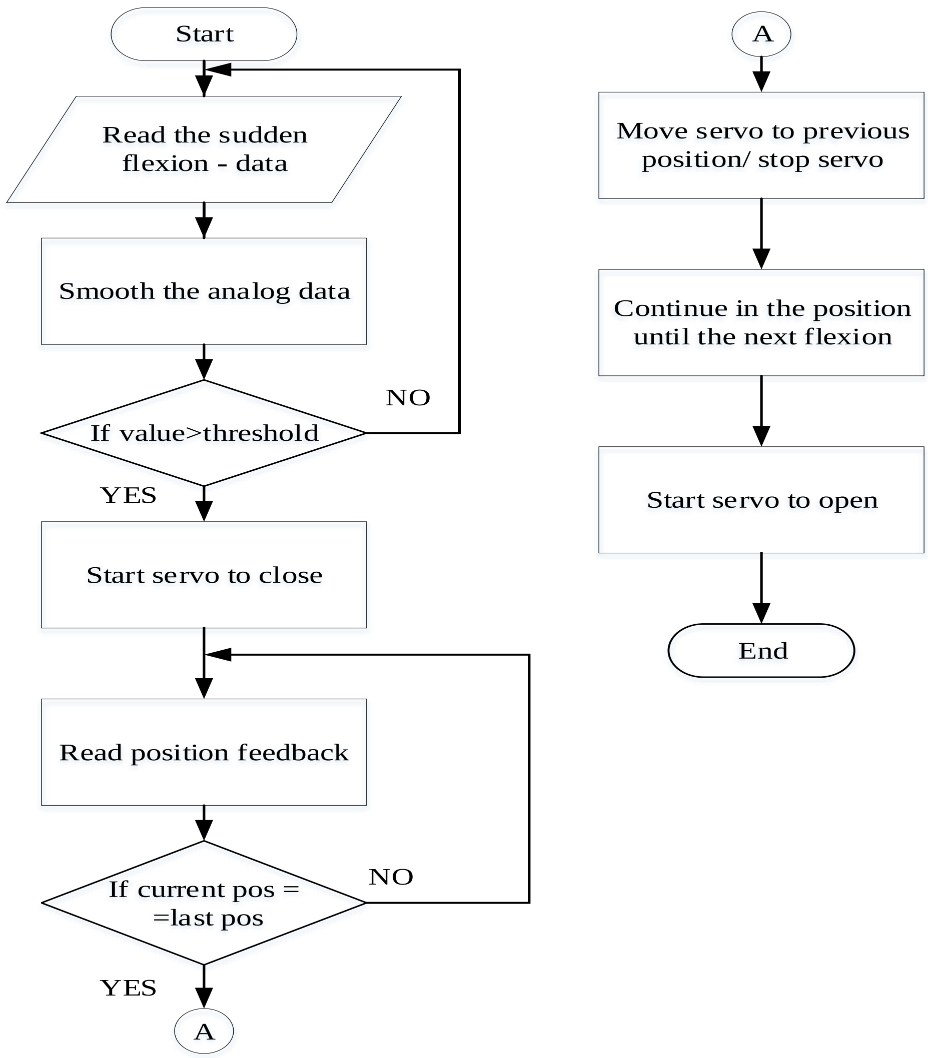


Figure 26. Closed loop position control flow chart

Surprisingly, the result was not as expected. Although the output position can be tracked from the servo, it is not linear in nature with the command position given. The data obtained from Arduino plotter is given for reference in Figure 27. The graph shows a clear nonlinear behavior of position feedback with respect to the change in rotor angle.

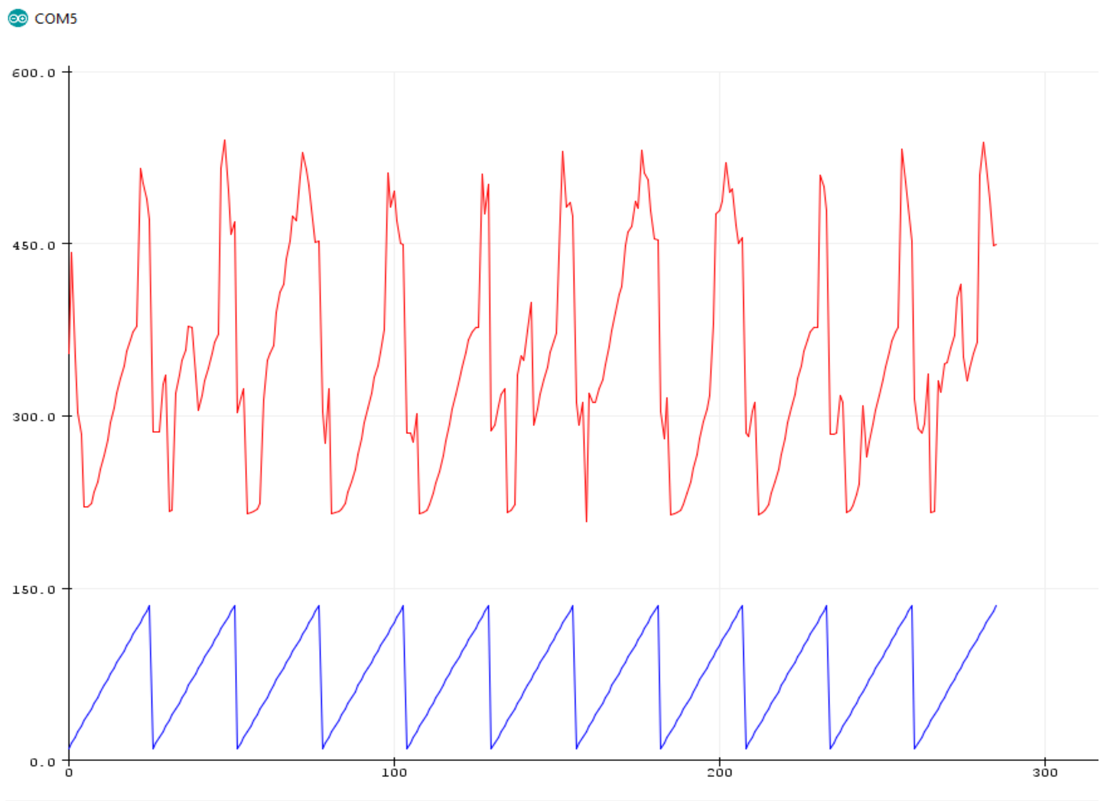


Figure 27. Servo motor sweep (Red – shaft position feedback, Blue- rotor angle)

This non-linearity makes the soft grip control process a difficult one. When the shaft motion is interrupted with an external object, the force put in to cease the motion is higher and rather takes long time to respond back with the controller. Because of the position data is not in line with the command, the position feedback is not a good choice of control for the system. The details of the testing results obtained from Arduino serial plotter was given next chapter.

4.2 Method II. Current Feedback

Due to the failure of Method I, the feedback current control was available as the next option. Compare to position feedback, the feedback in the form of current or torque is more convenient for the soft control of hand. And for all the servo motors electric current is required to induce the rotation. The electromagnetism produced by the combination of current carrying coils and permanent magnets interacts each other to produce the twisting force to change the motor shaft in desired direction. For a proper control of torque in servo, a well-controlled circuit current is mandatory.

Normally, inside the servo, a closed loop current control is running along with the position control by the built-in motor drive. The drive of the digital servo use power transistors to deliver enough energy. The power transistors inside the motor produce only voltage to actuate the motor. So, a feedback current loop is requisite to achieve precise control of current. The current loop compares the actual flow of current with the feedback current from load. Then, the controller fine-tunes the torque required to drive the motor to reduce the error between them. For much more dynamic control of hand the external closed loop current feedback is required. Instead of taking position of shaft, the load current is taken back to the Arduino Nano controller. The raw data is a highly varying voltage signal, because of the internal disturbances, so to obtain a current signal it is given to the resistive current sense circuit.

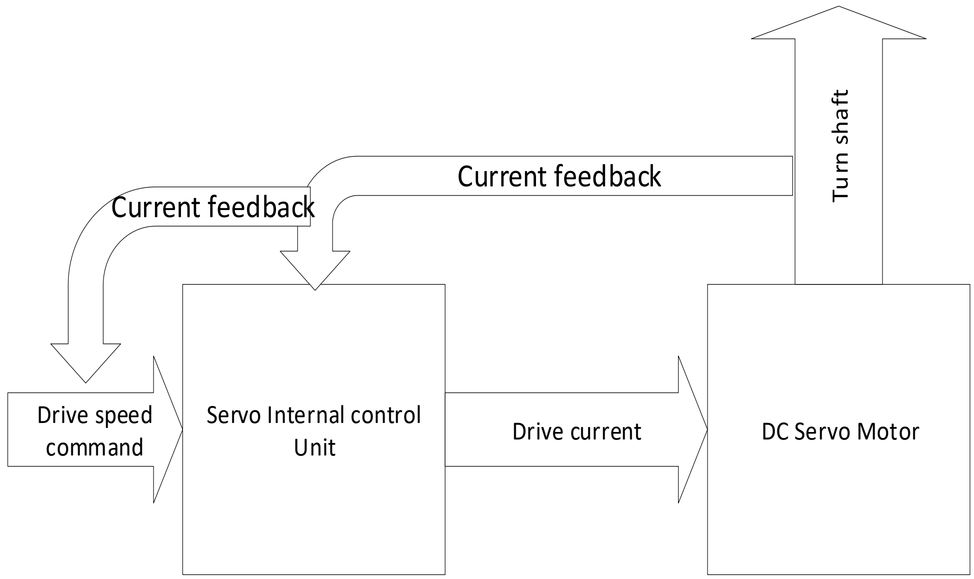


Figure 28. Closed loop (External) current control of servo

Using the current sense circuit given in Figure 17 thus the analog current data is collected to the analog pin A5 in the board. Same as in method I, band signals are analyzed as mean values for continuous 10 samples and for any of the level exceeding above the threshold a trigger is activated for the servo movement. The following code shows the triggering and advancing of servo position.

```
for (int rI1 = 0; rI1 < numvalue; rI1++) {
        sum = sum - value[rI1];
        value[rI1] = analogRead(inputpin)
        sum = sum + value[rI1];
        if (rI1 >= numvalue) {
          rI1 = 0; }
avg = sum / numvalue;
if((avg > threshold1)){//pulse detected
ang=ang+(DeltaAng*dir)
```

The advancing step of servo position is set as one degree and given as DeltaAng in the code. A direction 'dir' is prefixed as -1 for closing operation and 1 as open condition.

The default position of hand is setup as 170 degrees fully open. When it is achieved the controller give back the feedback and displayed it as ‘OK’ in the monitor. Once the trigger threshold is achieved it starts to move towards close of fingers or until it identifies and grips an object. Then the monitor will display ‘OBJECT’. All the time, the current is continuously reading through the analog pin A5 as shown below in Figure 29. The raw data is highly varying so smoothening function is applied to get some neat variations in current level.

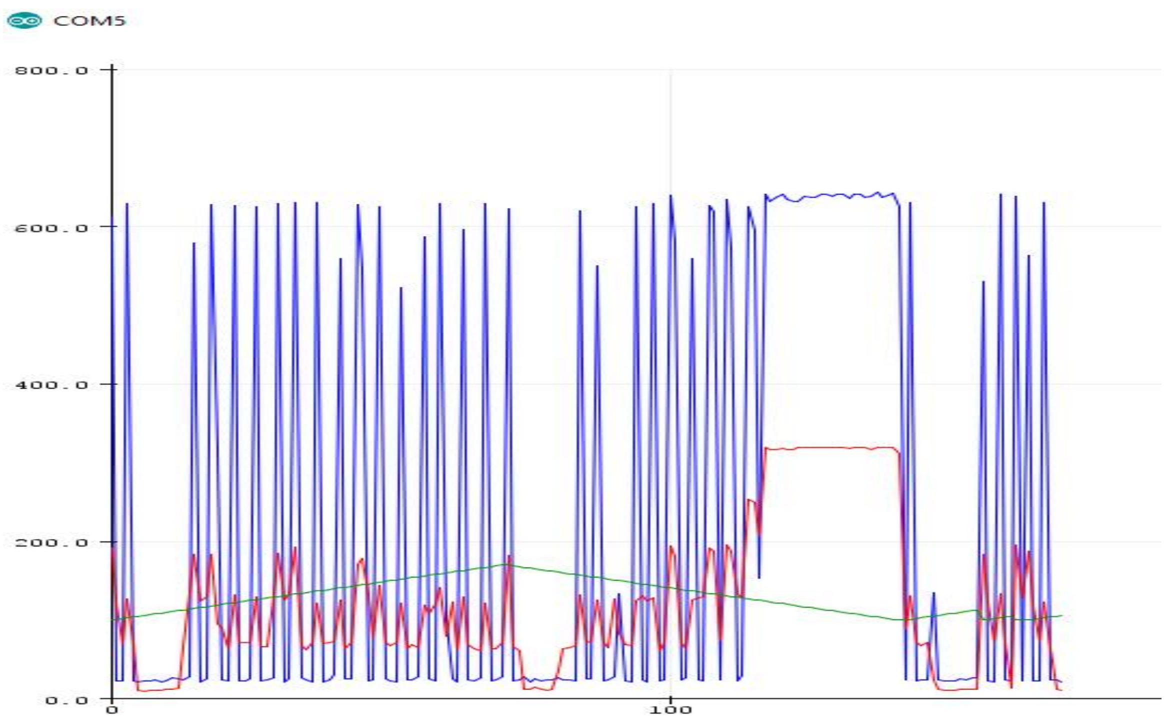


Figure 29. Current feedback data (Blue – current feedback, Red- smoothened signal)

Whenever the object is identified across the fingers, the current level jumps to a constant value. The constant value is continued if the object is staying in between the fingers. To protect the motor from overcurrent, as soon as the jump is identified the servo should stop its movement. This can be done by stepping back the servo to a small degree from the current position.

From the figure it is very clear that the level changes in current can be used as an effective communication tool between the controller and servo to make a smooth transition from forced close to soft grip. After so many trials with different sized and shaped objects a threshold current is set for this current jump. If that threshold exceeds, the position of the servo takes back to one step from the current angle.

```
if (average>threshold2){   //overcurrent detection
    ang=ang-DeltaAng*dir;
```

This way we can ensure that the grip is soft enough to hold the object and will not squeeze it more. The state achieved will continue until the next flexion of the band. Once the flexion reaches the threshold level which is different from the first threshold, the gripper starts to go back i.e. it loosens up the gripping of object and settles in the default position of fully open and wait for the next flexion of the muscles. The current feedback is tested and from the above result itself it is evident that the effectiveness of the motion is better with current control than the sensorial control.

5 Results

5.1 Fabrication of prototype

For the working analysis of model, a test setup was made with 5V DC supply. As a proof of the low-cost ULP feasibility model, all parts of the circuits were assembled on two small PCBs. The ASEMG detection PCB assembled is depicted in Figure 30, The PCB measures 33.5 × 47.5mm (ASEMG) and 44 × 65mm (User Feedback unit). In Figure 31 is depicted the user feedback PCB fully assembled. Contrary to the ASEMG PCB, this circuit uses three copper bridges that for convenience it was etched on the otherwise empty top copper layer. For convenience of prototyping, for both PCBs used the top copper layer for the components' designators. This greatly allowed to use a milling machine and not a chemical etching process. The results were captured from Digital signal oscilloscope. And for the efficacy of analysis some of the control results were taken from the serial monitor of Arduino IDE.

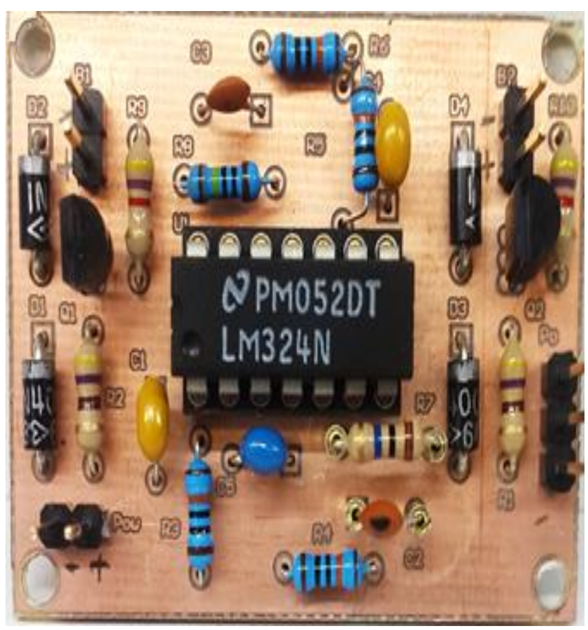


Figure 30. PCB assembly of ASEMG detection unit

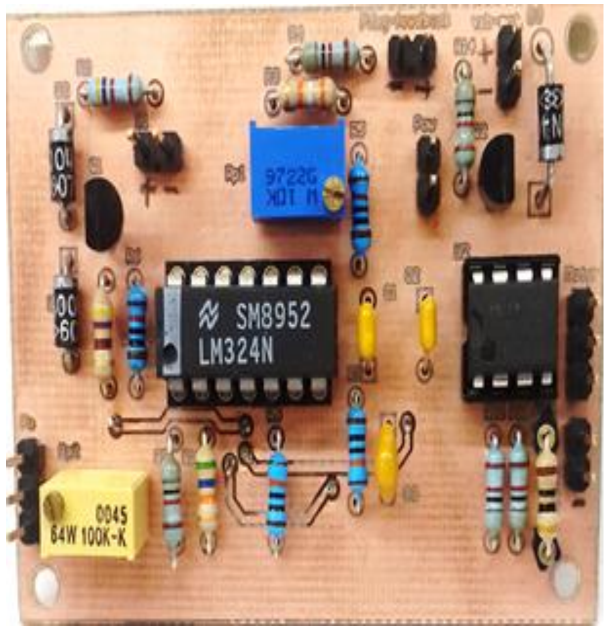


Figure 31. PCB assembly of User feedback unit

The circuit was completely realized by low cost readily available materials. The full Bill of Materials, as mentioned, is reported in Table 5. The total cost is calculated with the current market value of components available in Australia. None of the components are critical so that each of them can be replaced by their similar characteristic items available in developing countries.

Table 5. Full bill of Materials (Prices are in Australian Dollars and correct to December 2017)

Components	**Type**	**Quantity**	**Amount**	**Total**
Diodes	D1,D2-IN4148	4	0.05	0.2
	D3 - IN4001	1	0.08	0.08
Push button	MCDTS6	1	0.2	0.2

Resistors (Fixed)	R1 - 5K	3	0.08	0.24
	R2 - 470 Ω	3	0.05	0.15
	R3 - 100K	6	0.08	0.48
	R4 - 10M	2	0.2	0.4
	R5 - 10K	2	0.04	0.08
	R6 - 39K	1	0.08	0.08
	R7 - 1K	5	0.025	0.125
	R8 - 56K	1	0.024	0.024
	R9 - 10 Ω	1	0.05	0.05
Resistors (Variable)	Rp1 - 10K	1	1.19	1.19
	Rp2 - 100K	1	1.37	1.37
Capacitors	C1 - 7.9µF	2	0.06	0.12
	C2 - 68pF	2	0.03	0.06
	C3 - 0.33µF	1	0.08	0.08
BJT	Q1 - BC557B	2	0.29	0.58
	Q2 - BC548	1	0.05	0.05
Op Amps	LM324N	2	0.6	1.2
	LM358AP	1	0.4	0.4
Servo motor	MG996R	1	5	5
Vibrator (buzzer)	Coin type	1	2	2
ERB bands	10cm	3	0.13/cm	3.9
3 D printed hand	PLA make	48gm	0.08/gm	3.84

Arduino Nano	Atmega	1	6.95	6.95
Grand Total = 28.85				

5.2 Response of microcontroller to ERB change

From Figure 32 to Figure 36 represents some of the signals recorded during the experiments with the ASEMG circuit and user feedback circuit. The results have taken from Digital signal oscilloscope(DSO). The results are produced by placing the ERB band on one of the author’s normal hand during the voluntary flexion of upper wrist muscles to produce claws movements. In Figure 32 I showed the raw volumetric shifts (top trace, yellow color) and its filtered counterpart during continues muscle activations. In Figure 33 to Figure 34, shows the same signals recorded during sudden strong flexion and fingers movements. As it is possible to observe from the figures, the sensor is quite sensitive hence able to pick up even the smallest of the muscular volumetric shifts i.e. residual muscles on the stump.

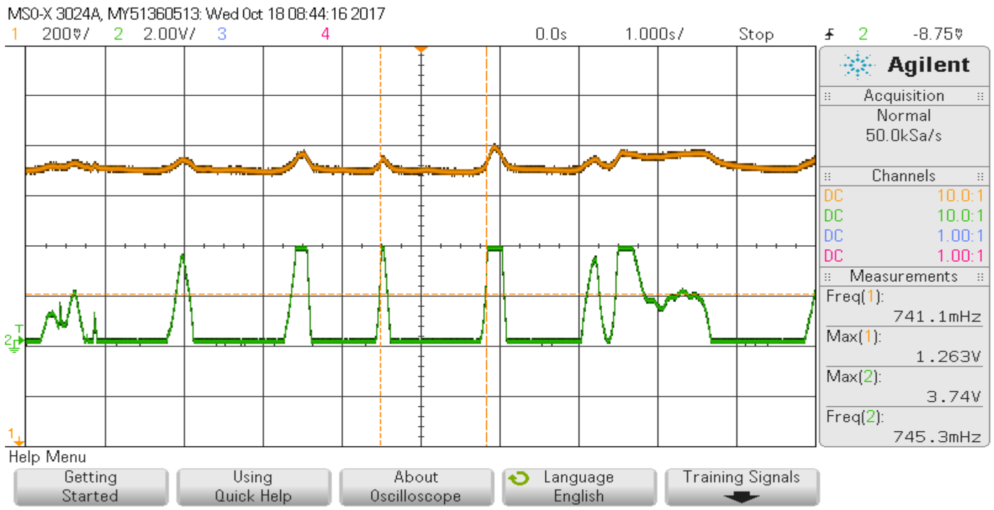


Figure 32. Raw signals of ERB sensor (Yellow) and filtered and amplified signal (Green) on continuous muscular variations.

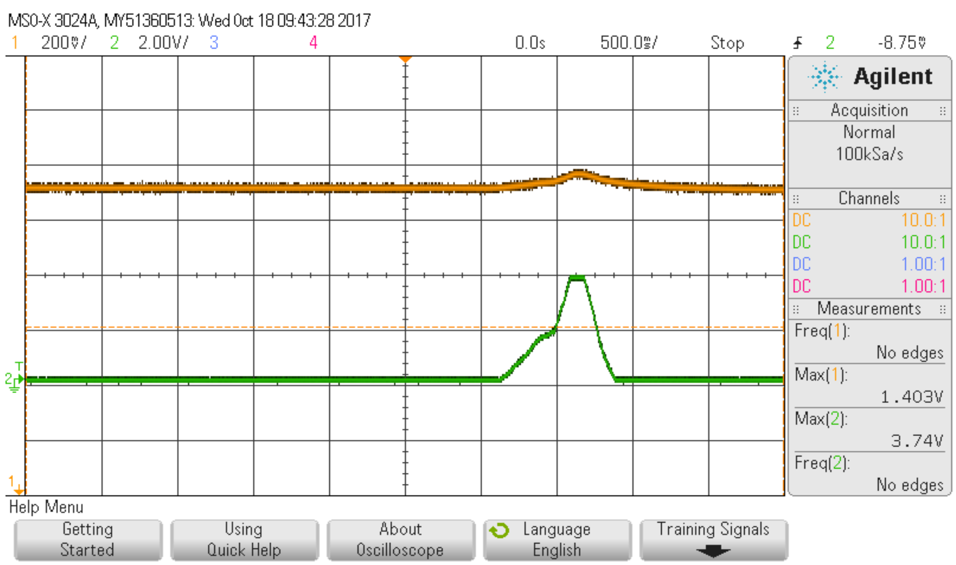


Figure 33. Oscilloscope results for signals of ERB sensor (Yellow) and filtered and amplified signal(Green): Signal after sudden flexion

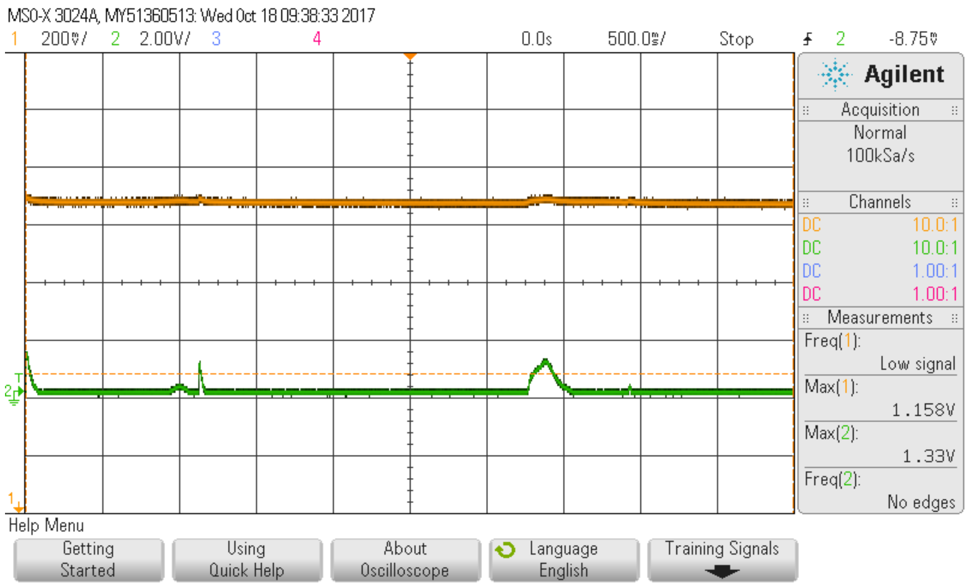


Figure 34. Oscilloscope results for signals of ERB sensor (Yellow) and filtered and amplified signal(Green): Signals when fingers are moved.

In Figure 35 to Figure 36 shows some example of translation of detected signals to pulse width modulated signals (PWM).

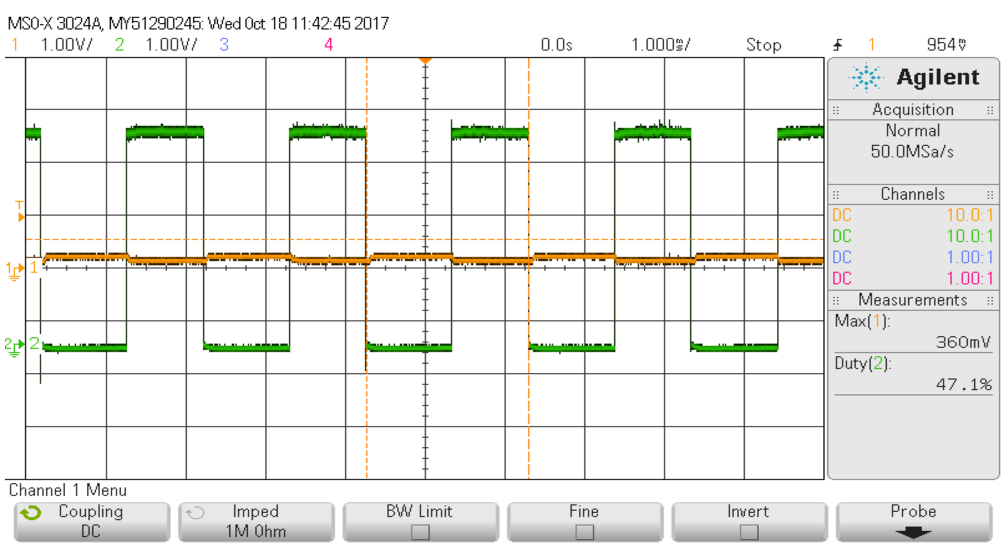


Figure 35. Raw signals of ERB feedback sensor(Yellow) and PWM2 duty signals(Green): fingers full closed

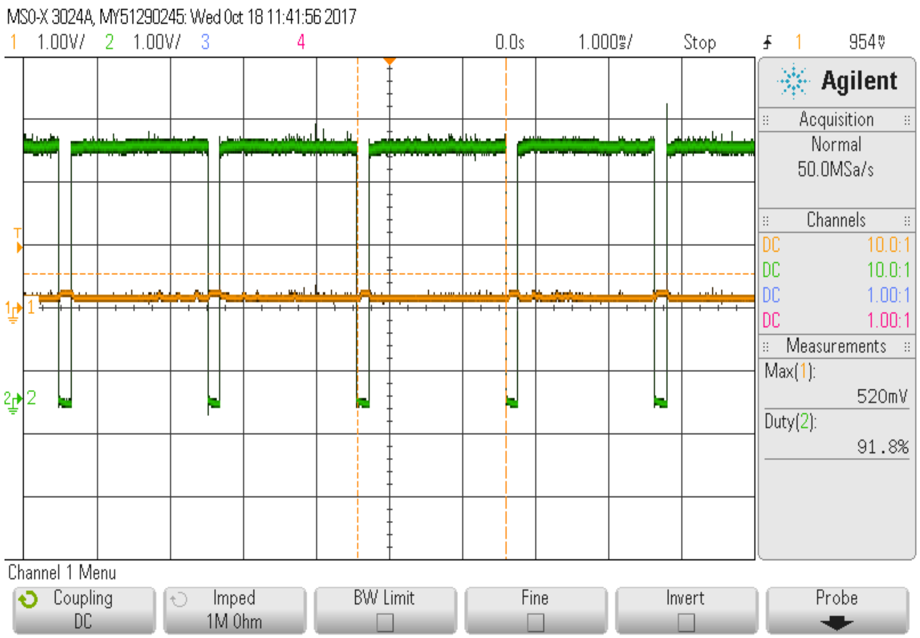


Figure 36. Raw signals of ERB feedback sensor(Yellow) and PWM2 duty signals(Green): (**a**)fingers full open

The total current consumption of the prosthesis varies (see Table 6) and it is 76 mA when powered by the 5V outlet on the Arduino during the opening of the claw. The largest current consumption during opening is due to the mechanical drags inside the servomotor gears and the additional drag associated with the claw itself. Although this value seems quite high, these are measured during operation the ULP consumes less power in average since that there will idle times.

Table 6. Current consumption of the circuit

Status	**Current consumption**
Opening	76 mA
Closing	62 mA

5.3 Response from Arduino Nano IDE

From Arduino software part, the programming results are taken only for the stage two analysis, through the inbuilt serial monitor plotter. The processed signal from analog read input pin is plotted with the smooth average of the signal. In Figure 30, the blue colour signal represents analog input and red colour represent smoothed average. The band is very sensitive, so for small changes it shows the corresponding variations in the controller output.

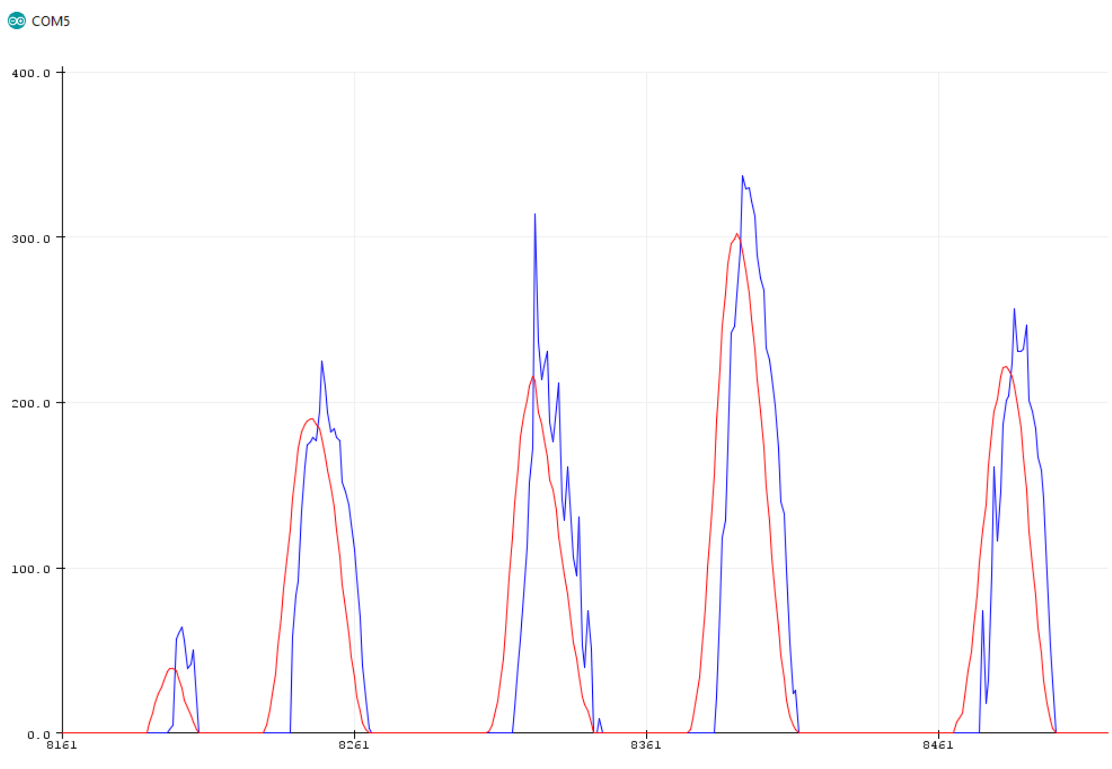


Figure 37. Arduino raw analog output (blue) and smooth output (red)

The performance of the servo motor response was also captured in Figure 33 and 34. A desired angle was set (straight blue line) and the actual angular position in terms of voltage was monitored (intermittent red line). It is clearly evident that the actual servo position is not linear with the rotor angle which limits its possible application for the current design of advanced soft control of prosthesis.

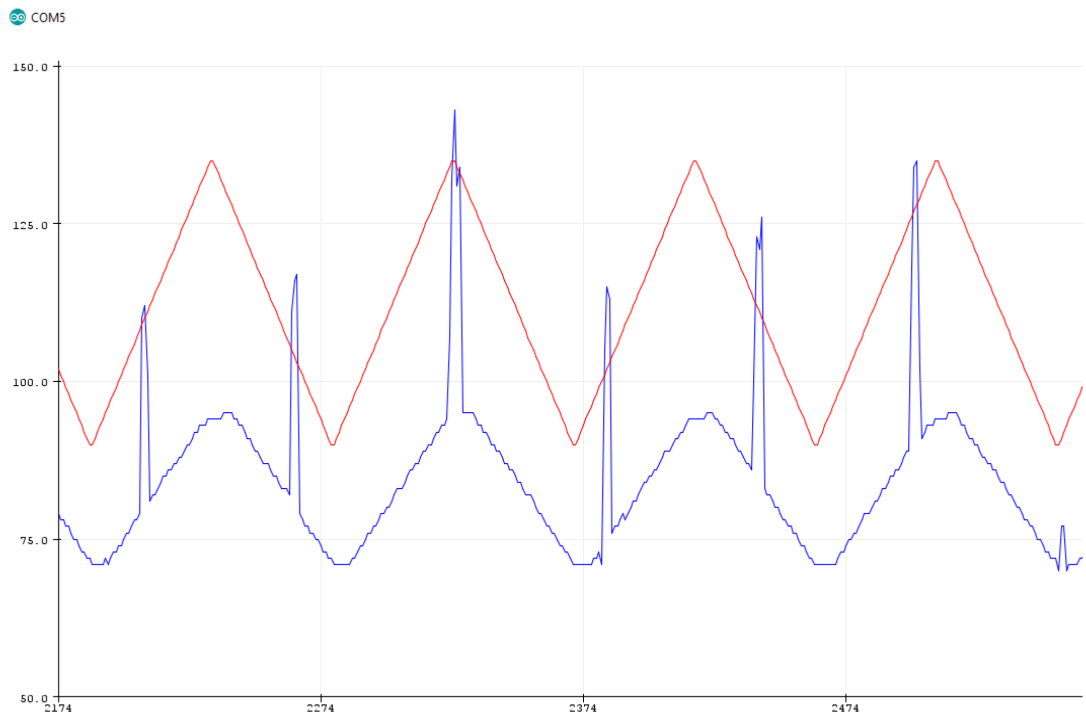


Figure 38. Servo motor data (Blue – shaft position feedback, Red- rotor angle)

But with the current feedback, control is more efficient and nearly smooth. In Figure 29 we can see the changes in current level when the motion of shaft is interrupted. That means, whenever the shaft hits with an object the current level changes and become constant until it is removed from the grip. This opened the possibility of fine tuning of control with prosthesis.

The option of sensorial feedback has not included in the stage 2 section of the feedback control. Because of the longer delay in getting response and difficulty in controlling prototype, I purposefully avoided sensorial feedback within stage 2. So, for the stage 2 design implementation, the component requirements of hand prosthesis are further reduced and the cost can be cut down to AUD 26. If both stages of control (current feedback and sensorial feedback) are incorporated together, in any way, it will not be higher than AUD 32.

Thus, the special EMG transducer sensor - ERB band sensor based prosthesis with no direct electrical connections on the body has developed, together with a hand mechanism actuated by a simple DC servo-motor with a battery supply of 5V. From all the results obtained above, this design can be taken as a guideline for the very promising feature of the cost reduction of prosthetic control. For the prosthetic research and design community, I believe the use of electro resistive band based circuit will create a new channel of possibilities. Ideally, the prototype behaves in the way which I expected, so with slight modifications the present prototype model can be realized as the real-world prosthesis and it is only in one hand distance away.

6 Conclusion

6.1 Discussion

Amputees in developing countries who have undergone a wrist disarticulation for one or many reasons often avoid the use of an artificial hand to regain part of the lost function. The restrictions for choosing a proper solution for amputation is mainly the cost, weight and reach of the artificial device. The main objective of our proposed model to provide a solution to the above problems by designing a low cost and simple prosthesis. And, as proposed, the idea of low cost design is completely successful; that the design is fully realized with a total cost of AUD 29 (see Table 5) with a minimum number of components and compact assembly.

One of the main advantage of this design is the usage of a non-invasive sensor, ERB band. Instead of the use of traditional EMG electrodes to acquire and measure muscle contractions, this very cheap sensor is a big boon to the present market of prosthetic control. Moreover, the sensor does not require any preparation of skin surface and are not much sensitive to external electromagnetic disturbances. The other striking feature of the proposed design includes, none of the components like BJTs and Op amps are not strictly explicit to the design, so they can be replaced with any equivalent components.

The placing of ERB bands on the claw like fingers of the artificial hand, provide a good level of vibrotactile sensory feedback to allow the user for soft gripping. Clearly, it is found that inclusion of stage 2 adds more value to the design. Among the two methods used for feedback, current control is more achievable than position control. i.e. The method II feedback control is the most appropriate means for the application stated above.

While the performance of the two stages of designs are practically identical, phase 2 is found to be the well-organized, simple and speedy technique. The major advantage of this is the possibility for user to adjust with the gripping pressure of hand with the shape of an object.

The design presented is suitable for all ages because of its contained weight and structure. Since the control is simple users will not require the assistance of expert technicians. Despite the use of convenient 3D printing technology, the physical parts are so simple that they could be easily replaced with recovered materials. With this design, the assistance of an expert in rehabilitation is not always required for children or old age people. The use of a 3D printed hand is the other added feature which reduces the cost. Mass customization and inexpensive production parts make it accessible to achieve the precise and detailed user requirements, with the dimensions of the limb, and its intricacy. The proportions of each limb could be matched with the size of the other limb. The compact structure of the limb project will help the user to upgrade distinct parts with their choices as they grow old and their capacities or likings deviate from the existing one.

Although the design can make a vast impact on the lives of deprived society, the functions of the planned proposal are restricted to a single degree of freedom. The purpose of design is simply to hold/grasp an object so that the hand model is limited and closed in a claw type of shape. With all design compromises and trade-offs, the new model can provide a suitable solution to an extent, towards the affordable prosthesis in developing countries. The proposed project of ERB based ultra-low cost SEMG controlled prosthetic hand with closed loop operation is completely enough and suitable for regaining lost function of the hand.

To summarize, the present concept design elevates the myo-activated prosthesis towards a more affordable and easy to use reality. The only disadvantage of the design is the two claws like gripper design. A more realistic finger type approach bounded with cost constraints is needed for the further expansion of this work. In appendix A, I have included the Arduino scripts for the Sensorial feedback control. And, in appendix B, the Arduino scripts for the closed loop feedback control is included. As well as the PCB Gerber files and the STL for the claw stile hand can be downloaded from https://github.com/neethurugma/Low-cost-prosthesis. These files together with the full bill of materials are everything that is required to reproduce the design.

6.2 Future recommendations

The current design system could be able to achieve the aim of a prosthetic hand to a reasonable extent. Yet, the explained prototype is more of an illustrative design than a real applicable one. Aimed at the sake of simple explanation, the manageable constraints in this thesis set aside to the minimum level. The actual artificial arm can be made as a scaled version of the model device. In the future, the proposed work can be proceeded further to remove the shortcomings of the discussed prototype and develop into a saleable practical design. In that way, this simple model is just a stepping-stone for the low-cost solution of ULPs in developing countries.

To be able to simulate the complex and often subtle, but at times powerful and broad movements more efficiently, this value needs to increase. This can be achieved with a better microcontroller and more effervescent programming algorithm. Also, a programming language more suited to the purpose can be used to increase the sensitivity and selectivity of the device. Another improvement that can be made is increasing the number of degrees of freedom. A great part of it will involve tweaking the mechanical part of our system to better resemble the mechanical aspects of a real hand. Additionally, the complexity of the finger can be increased, again by improvements over the quite rudimentary design that have incorporated into the system for simplicity. The use of a stepper-motor instead of the basic DC motor used in our system is also recommended to exert better control over the prosthetic movement.

Lastly, I would like to underline that the use of ERB bands as sensors in prosthesis could be discussed in a broader context where conventional electrodes play a major role and have been the reason for frustration and malfunction. This inexpensive sensor being contactless could be worn on the top of clothes and multiple sensors can be used to increase the prosthesis with much more sophisticated control and degree of freedoms.

References

[1] S. A. Ovadia and M. Askari, “Upper extremity amputations and prosthetics.,” *Semin. Plast. Surg.*, vol. 29, no. 1, pp. 55–61, Feb. 2015.

[2] F. Cordella *et al.*, “Literature Review on Needs of Upper Limb Prosthesis Users,” *Front. Neurosci.*, vol. 10, p. 209, May 2016.

[3] E. Strait, “Prosthetics in Developing Countries,” *White Paper*, 2006. [Online]. Available: http://www.glb.nist.gov/tip/wp/pswp/upload/239_limb_prosthetics_services_devices.pdf%5Cnhttps://notendur.hi.is/ats9/Prosthetics in DevelopingCountries.pdf.

[4] World Health Organisation and International Society for Prosthetics and Orthotics, “Guidelines for training Personnel in Developing countries for Prosthetics and Orthotics Services,” 2005.

[5] N. E. Walsh and W. S. Walsh, “Rehabilitation of landmine victims - The ultimate challenge,” *Bull. World Health Organ.*, vol. 81, no. 9, pp. 665–670, 2003.

[6] “WHO | Assistive technology,” *WHO*, 2016.

[7] B. ; Phillips, S. ; Ritter, Mehta, and Khanjan, “Printed Prostheses: Not yet ready for developing countries,” *Mech. Eng.*, vol. 138, no. 4, 2016.

[8] “DynamicArm transhumeral prosthesis | Ottobock AU,” *www.ottobock.com.au*. [Online]. Available: https://www.ottobock.com.au/prosthetics/upper-limb/solution-overview/above-elbow-system-with-dynamicarm/. [Accessed: 08-Apr-2018].

[9] E. A. Biddiss and T. T. Chau, “Upper limb prosthesis use and abandonment: A survey of the last 25 years.”

[10] A. Kumar Padhi and B. Prasad Nayak, "Development of a limb prosthesis by reverse mechano transduction," National Institute of Technology, Rourkela, 2013.

[11] J. N. CUMINGS, "How Does the Brain Function?," *Nature*, vol. 200, no. 4905, pp. 395–395, 1963.

[12] C. J. Forehand, "The Action Potential, Synaptic Transmission, and Maintenance of Nerve Function- The maintenanceof nerve cell funcion," .

[13] M. Marino *et al.*, "Access to prosthetic devices in developing countries: Pathways and challenges," *Proc. 5th IEEE Glob. Humanit. Technol. Conf. GHTC 2015*, no. December, pp. 45–51, 2015.

[14] G. T. Popa and A. Name, "A stretchable, conductive rubber sensor to detect muscle contraction for prosthetic hand control," in *The 6th IEEE International Conference on E-Health and Bioengineering - EHB 2017*, 2017.

[15] M. Asghari Oskoei and H. Hu, "Myoelectric control systems—A survey," *Biomed. Signal Process. Control*, vol. 2, no. 4, pp. 275–294, Oct. 2007.

[16] J. Kate, G. Smit, P. Breedveld, J. Kate, and G. Smit, "Disability and Rehabilitation : Assistive Technology 3D-printed upper limb prostheses : a review," vol. 3107, no. September, 2017.

[17] S. G. Kumar, G. Roy, and S. S. Kar, "Disability and rehabilitation services in India: issues and challenges.," *J. Fam. Med. Prim. care*, vol. 1, no. 1, pp. 69–73, Jan. 2012.

[18] D. Cummings, "Prosthetics in the developing world: a review of the literature," *Prosthet. Orthot. Int.*, vol. 20, pp. 51–60, 1996.

[19] G. D. Gargiulo, A. O'Loughlin, and P. P. Breen, "Electro-resistive bands for non-invasive cardiac and respiration monitoring, a feasibility study.," *Physiol. Meas.*,

vol. 36, no. 2, pp. N35-49, 2015.

[20] E. S. Varaki, P. Breen, and G. Gargiulo, “Quantification of a Low-Cost Stretchable Conductive Sensor Using an Expansion/Contraction Simulator Machine: A Step towards Validation of a Noninvasive Cardiac and Respiration Monitoring Prototype,” *Machines*, vol. 5, no. 4, p. 22, 2017.

[21] G. D. Gargiulo, U. Gunawardana, A. O’Loughlin, M. Sadozai, E. S. Varaki, and P. P. Breen, “A Wearable Contactless Sensor Suitable for Continuous Simultaneous Monitoring of Respiration and Cardiac Activity,” *J. Sensors*, vol. 2015, 2015.

[22] E. E. Vanderwerker Jr., “A Brief Review of the History of Amputations and Prostheses,” *Assoc. Child. Prosthetic-Orthotic Clin. Inter-Clinic Inf. Bull.*, vol. 15, no. 5, pp. 15–16, 1976.

[23] T. L. Wellerson, “Historical Development of Upper Extremity Prosthetics,” *Orthop. Prosthet. Appl. J.*, vol. 11, no. 3, pp. 73–77, 1957.

[24] R. H. Meier, “History of Arm Amputation, Prosthetic Restoration, and Arm Amputation Rehabilitation,” *Funct. Restor. Adults Child. with Up. Extrem. Amputation*, pp. 1–7, 2004.

[25] “Robotics, simulation, and telemedicine in plastic surgery | Plastic Surgery Key.” [Online]. Available: https://plasticsurgerykey.com/robotics-simulation-and-telemedicine-in-plastic-surgery/. [Accessed: 13-Mar-2018].

[26] “Myoelectric control of prostheses: A brief history,” in *MyoElectric Controls/Powered Prosthetics Symposium Fredericton*, 1992.

[27] P. J. Kyberd and J. L. Pons, “A comparison of the Oxford and Manus intelligent hand prostheses,” *Proc. IEEE Int. Conf. Robot. Autom.*, vol. 3, pp. 3231–3236, 2003.

[28] C. M. Light, P. H. Chappell, B. Hudgins, and K. Engelhart, “Intelligent multifunction myoelectric control of hand prostheses,” *Journal of Medical Engineering and Technology*, vol. 26, no. 4. pp. 139–146, 2002.

[29] H. Yu, “A high performance microchip control system design for DC motor,” *2011 Int. Conf. Electron. Commun. Control. ICECC 2011 - Proc.*, pp. 1210–1213, 2011.

[30] K. Akazawa, R. Okuno, and M. Yoshida, “Biomimetic EMG-prosthesis-hand,” in *Proceedings of 18th Annual International Conference of the IEEE Engineering in Medicine and Biology Society*, vol. 2, pp. 535–536.

[31] I. Kajitani *et al.*, “An evolvable hardware chip for prosthetic hand controller,” in *Proceedings of the 7th International Conference on Microelectronics for Neural, Fuzzy and Bio-Inspired Systems, MicroNeuro 1999*, 1999, pp. 179–186.

[32] Á. Villoslada Peciña, “Design and Implementation of a Myoelectric Control System for a Printable Robotic Hand,” p. 115, 2012.

[33] M. Tavakoli, C. Benussi, and J. L. Lourenco, “Single channel surface EMG control of advanced prosthetic hands: A simple, low cost and efficient approach,” *Expert Syst. Appl.*, vol. 79, pp. 322–332, 2017.

[34] L. Resnik, “Development and testing of new upper-limb prosthetic devices: Research designs for usability testing,” *JRRD*, vol. 48, no. 6, 2011.

[35] “Types of Upper Extremity Prostheses | P and O Care.” [Online]. Available: http://www.pandocare.com/upper-extremity-types/#1471966938595-7757743e-c7b3. [Accessed: 13-Mar-2018].

[36] K. J. Zuo and J. L. Olson, “The evolution of functional hand replacement: From iron prostheses to hand transplantation.,” *Plast. Surg. (Oakville, Ont.)*, vol. 22, no. 1, pp. 44–51, 2014.

[37] "Types of Upper Extremity Prostheses | P and O Care," *www.pandocare.com*. [Online]. Available: http://www.pandocare.com/upper-extremity-types/#1471968815523-57104117-7e02. [Accessed: 08-Apr-2018].

[38] S. L. Carey, D. J. Lura, M. J. Highsmith, CP, and FAAOP, "Differences in myoelectric and body-powered upper-limb prostheses: Systematic literature review," *J. Rehabil. Res. Dev.*, vol. 52, no. 3, pp. 247–262, 2015.

[39] A. Saikia, "Below Elbow Prosthetic: A Path to Independent Era," *Int. J. Adv. Inf. Sci. Technol.*, vol. 34, no. February, 2015.

[40] S. G. Millstein, H. Heger, and G. A. Hunter, "Prosthetic use in adult upper limb amputees: a comparison of the body powered and electrically powered prostheses.," *Prosthet. Orthot. Int.*, vol. 10, no. 1, pp. 27–34, Apr. 1986.

[41] M. B. I. Raez, M. S. Hussain, and F. Mohd-Yasin, "Techniques of EMG signal analysis: detection, processing, classification and applications.," *Biol. Proced. Online*, vol. 8, pp. 11–35, 2006.

[42] F. Cordella *et al.*, "Literature Review on Needs of Upper Limb Prosthesis Users.," *Front. Neurosci.*, vol. 10, p. 209, 2016.

[43] D. F. U. G, N. Sreenivasan, G. Gargiulo, and U. Gunawardana, "Cost effective electro - resistive band based myo activated prosthetic upper limb for amputees in the developing world," 2017.

[44] L. Resnik *et al.*, "Advanced Upper Limb Prosthetic Devices: Implications for Upper Limb Prosthetic Rehabilitation," *YAPMR*, vol. 93, pp. 710–717, 2012.

[45] P. Kumar Lenka, P. K. Lenka, A. R. Chowdhury, and B. Ratnesh Kumar, "Design & Development of Lower Extremity Paediatric Prosthesis, a Requirement in Developing Countries," vol. 19, no. 81, pp. 8–12, 2008.

[46] E. A. Biddiss and T. T. Chau, "Upper limb prosthesis use and abandonment: A survey of the last 25 years," *Prosthet. Orthot. Int.*, vol. 31, no. 3, pp. 236–257, Sep. 2007.

[47] Z. G. Xiao and C. Menon, "Towards the development of a wearable feedback system for monitoring the activities of the upper-extremities.," *J. Neuroeng. Rehabil.*, vol. 11, no. 1, p. 2, Jan. 2014.

[48] WHO, "World Report on Disability - Summary," *World Rep. Disabil. 2011*, no. WHO/NMH/VIP/11.01, pp. 1–23, 2011.

[49] J. Borg and P.-O. Östergren, "Users' perspectives on the provision of assistive technologies in Bangladesh: awareness, providers, costs and barriers," *Disabil. Rehabil. Assist. Technol.*, vol. 10, no. 4, pp. 301–308, Jul. 2015.

[50] E. Scheme and K. Englehart, "Electromyogram pattern recognition for control of powered upper-limb prostheses: State of the art and challenges for clinical use," *J. Rehabil. Res. Dev.*, vol. 48, no. 6, p. 643, 2011.

[51] "Conductive Rubber Cord Stretch Sensor + extras! ID: 519 - $9.95 : Adafruit Industries, Unique & fun DIY electronics and kits." [Online]. Available: https://www.adafruit.com/product/519. [Accessed: 04-Mar-2018].

[52] G. D. Gargiulo, P. Bifulco, M. Cesarelli, A. Fratini, and M. Romano, "Problems in Assessment of Novel Biopotential Front - End with Dry Electrode : A Brief Review," *Machines*, vol. 2, pp. 87–98, 2014.

[53] G. Description, U. Characteristics, and C. Diagram, "Low Power Quad Operational Amplifiers Low Power Quad Operational Amplifiers," no. December 1994, pp. 1–12, 1995.

[54] Arduino LLC, "Arduino Nano," vol. 9210, no. 662. p. 1, 2015.

[55] R. Elmansy, “Designing the 3D-printed prosthetic hand,” *Des. Manag. Rev.*, vol. 26, no. 1, pp. 24–31, 2015.

[56] “Tower pro,” *www.towerpro.com.tw*. [Online]. Available: http://www.towerpro.com.tw/product/mg995/. [Accessed: 08-Apr-2018].

[57] B. S. Day, “Important Factors in Surface EMG Measurement,” *Measurement*, pp. 1–17, 2002.

Appendix A

This program reads analogue inputs from the ASEMG circuits as well as from the fedback circuit and produces the proportional Servo/buzzer responses. The analogue values are also transmitted via Serial port for debug purposes.

```
#include <Servo.h>
Servo MySerVo;
//Variable Declaration PINs
 int AInput7 = 7;      //Analog Input pin A7 Band Position Control
 int AInput5 = 5;         //Analog Input pin A5 Sesorial position
Feedback
 int PWMOutput = 11;    //PWM Output pin 11 Sesorial feedback
 int ServOut = 10;    //Servo signal connected to pin D10
 int DirPIN = 3;      //Digital Pin to change direction
 int CalPIN = 4;      //Digital Pin to start the calibration protocol
 int ledPIN = 13;     //Digital Pin with a led
 //Global Variables
 /**Position Control**/
 int AValueCtrl = 0;
 float RMS = 0;
 int ServAng = 0;
 int DeltaAng=0; //add/dim servo's angle
 int counter=0; //use to calculate Signal's RMS value
 /**Sensorial Feedback**/
 int PWMDuty = 0; //Duty to control the vibrating motor
 int AValueFbk = 0;   //analogue value read from the analogue channel
 int NSample=1;    //counts to calculate the average
 float AnaAverg=0; //to calculate the average of the last 10 samples
 volatile int dir=-1; //to choose direction negative: close;
positive: open;
 /**Code Timing**/
 int TimeIni = 0;
 int TimeEnd = 0;
 int DeltaT = 0;
 bool Active = false;
void setup() {
  // put your setup code here, to run once:
  //Activate serial transmision
  Serial.begin(115200);
  //set analog referece for the ADC
  analogReference(DEFAULT);
  //Pin Setup
  pinMode(A5, INPUT); //Used for the control signal
  pinMode(A7, INPUT);//Used for the feedback signal
  pinMode(ServOut, OUTPUT);//Claw psoition control
  //Servo Setup
  MyServo.attach(ServOut);
  MyServo.write(40); //Initial position, Widely Open
  //Interrupts Setup
  attachInterrupt(digitalPinToInterrupt(DirPIN),ButtonRed,FALLING);
```

```
attachInterrupt(digitalPinToInterrupt(CalPIN),ButtonBlack,FALLING);
      }//End Setup
    void loop() {
    /******Position Control******/
       //Read a Value from the sensor:
      AValueCtrl = analogRead(AInput7);
      //Serial.println(AValueCtrl);
      RMS = 0;
      if (AValueCtrl>=550){
          TimeIni = millis();
          for (counter=0; counter<=100; counter++)
          { RMS = RMS + AValueCtrl;
         // Serial.print(RMS);Serial.print(", ");
          Serial.println(AValueCtrl);
         // Serial.println(counter);
          delay(5);
          AValueCtrl = analogRead(AInput7);
          if (AValueCtrl<550) {
              break;
              };
            };
       TimeEnd = millis();
       DeltaT= TimeEnd-TimeIni;
      // Serial.print("Delta T =  ");Serial.println(DeltaT);
       RMS = RMS*(counter+1)*(0.001);
       Serial.print("RMS                                              =
");Serial.print(RMS);Serial.print("/");Serial.println(counter);
       DeltaAng = 0.01*RMS;
       Serial.print("DeltaAng= ");Serial.println(DeltaAng);
       ServAng = ServAng + (DeltaAng*dir);
         if(ServAng>40){
          ServAng = 40;
          }; //Can't open wider!
          if(ServAng<0){
          ServAng = 0;
          }; //Can't close narrower!
       MyServo.write(ServAng);
      Serial.print("ServAng= ");Serial.println(ServAng);
      }
      /******End Position Control******/
     /******Sensory Feedback Control******/
     //Digital Low-pass Filter:
     //Reads 10 values from the AInput, and take the average:
     //Sampling rate, 5ms
     AnaAverg = 0;
     for(int NSample=1; NSample<=20; NSample++){
      //Read a Value from the sensor:
      AValueFbk = analogRead(AInput5);
      AnaAverg = AnaAverg + AValueFbk;
```

```
      delay(5);
      };
      AnaAverg = AnaAverg/20;
     /**End Digital Low-pass Filter  ********/
      // Escale value:
      PWMDuty = 0.25*AnaAverg + 25;
      if (PWMDuty<60) PWMDuty=70; //Min value that can be felt
      if (PWMDuty>255) PWMDuty=255;//Max value in the motor
     //Send the PMW value to the motor:
      analogWrite(PWMOutput, PWMDuty);
      // Print value via serial
      Serial.print("Position=  ");Serial.print(AnaAverg);Serial.print(",
");
      Serial.print("Duty= ");Serial.println(PWMDuty);
      /******End Sensory Feedback Control******/
      }//End Loop
      //Interruption Routine Service - Red Button
      //When the button is pressed, the direction of the claw changes
from opening to closing
      //and viceverse
      void ButtonRed(){
      dir*=-1;
        if(dir<0){
          Serial.println("closing");
          }
          else{
           Serial.println("opening");
            };
        };//End Button Red
        //Interruption Routine Service - Black Button
       //When the button is pressed, the configuration mode for the
feedback starts
       void ButtonBlack(){

      int Value = 0;
          //Blink three times, to start calibration
          digitalWrite(ledPIN, LOW);
          delay(500);
          digitalWrite(ledPIN, HIGH); //1
          delay(300);
          digitalWrite(ledPIN, LOW);
          delay(300);
          digitalWrite(ledPIN, HIGH);  //2
          delay(300);
          digitalWrite(ledPIN, LOW);
          delay(300);
          digitalWrite(ledPIN, HIGH);  //3
          delay(300);
          digitalWrite(ledPIN, LOW);
          /***************************************/
```

```
            do{
            //Read the analogue port and change resistance in Zero Pot
until it reaches 200 counts
                 Value = analogRead(AInput5);
                  delay(100);
                  }while(Value!=200);

                  //Blink once to continue calibration
                  digitalWrite(ledPIN, HIGH); //1
                  delay(300);
                  digitalWrite(ledPIN, LOW);
                  delay(300);
                  /**********************************/
                  do{
            //Read the analogue port and change resistance in Span Pot
            until it reaches 950 counts
                  Value = analogRead(AInput5);
                  delay(100);
                  }while(Value!=950);
                  //Blink Twice to finish calibration
                  digitalWrite(ledPIN, HIGH); //1
                  delay(300);
                  digitalWrite(ledPIN, LOW);
                  delay(300);
                  digitalWrite(ledPIN, HIGH); //2
                  delay(300);
                  digitalWrite(ledPIN, LOW);
                  delay(300);
                  /**********************************/
           loop();//Starts main program again
        };//End Black Red
```

Appendix B

```
#include <Servo.h>
Servo servo;
const int numvalue = 10;
int rI1 = 0;
int sum = 0;
int avg = 0;
int rI2 = 0;
int total = 0;
int average = 0;
const int threshold1 = 410;
const int DeltaAng=1;
const int threshold2 = 380;
int ang=0;
int inputpin = A0;  // to measure pulses
int input[10]={0,0,0,0,0,0,0,0,0,0};
int fdbkpin = A5 ; //to measure current
int fdbk[10]={0,0,0,0,0,0,0,0,0,0};
int dir=-1;
boolean PulseFLG=false;//true if a pulse is detected
boolean ObjectFLG=false;//true if avercurrent detect an object when
closing the claw
boolean OCurrentFLG=false;//true if overcurrent is detected
/*********Functions*********/
boolean ReadPulse(){
  for (int rI1 = 0; rI1 < numvalue; rI1++) {
        sum = sum - input[rI1];
        input[rI1] = analogRead(inputpin);
        Serial.print("band ");Serial.print(rI1);
        Serial.print(" ");Serial.println(input[rI1]);
        sum = sum + input[rI1];
        delay(50);
              if (rI1 >= numvalue) {rI1 = 0; };
        }//end for
  avg = sum / numvalue;
  Serial.print(" band avg ");Serial.println(avg);
        if((avg > threshold1)){ return true;}
        else {return false;};//end if else
        };//End Read Pulse
/********************/
boolean OverCurrent(){
  total=0;
  average=0;
  for(rI2=0; rI2<numvalue; rI2++){
        fdbk[rI2]=analogRead(fdbkpin);
        Serial.print(" current ");Serial.println(fdbk[rI2]);
        total = total+fdbk[rI2];
      }
      average=total/10;
       Serial.print(" average ");Serial.println(average);
      if (average>threshold2){ return true;}
      else {return false;};//end if else
};//End Overcurrent
void setup() {
  pinMode(A0, INPUT);
```

```
  pinMode(A5, INPUT);
  Serial.begin(9600);
  servo.attach(9);
  ang=170;
  servo.write(ang);delay(1000);
  Serial.println("OK");
}
void loop() {
/*****Reset Flags****/
PulseFLG=false;
ObjectFLG=false;
OCurrentFLG=false;
/*******************/
while(!PulseFLG){//loop to detect pulse
      PulseFLG=ReadPulse();
      };//end While
do{//close claw until obect is detected
  ang=ang+(DeltaAng*dir);
        if(ang==99){
                    ang=170;
                    PulseFLG=true;//to jump to reset
                    break;
                    };//limit condition, no object found
  Serial.print(" new ang ");Serial.println(ang);
  servo.write(ang);
  OCurrentFLG=OverCurrent();//check for overcurrent
        while (OCurrentFLG){//if Overcurrent is detected
          ObjectFLG=true;
          ang=ang-(DeltaAng*dir);
          servo.write(ang);
          };//end while no overcurrent
  }while(!ObjectFLG);//End DO-While
Serial.println("OBJECT");
delay(5000);
while(!PulseFLG){//loop to detect second pulse
    PulseFLG=ReadPulse();
    };//end While
/*******Reset Claw Position*******/
ang=170;
servo.write(ang);
delay(2000);// wait 2sec
Serial.println("reseted");
/*****************************/
    }
```