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At-Home Computer-Aided Myoelectric Training System for Wrist Prosthesis

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Abstract. Development of tools for rehabilitation and restoration of the movement after amputation can benefit from the real time interactive virtual animation model of the human hand. Here, we report a computer-aided training/learning system for wrist disarticulated amputees, using the open source integrated development environment called "Processing". This work also presents the development of a low-cost surface Electro-MyoGraphic (sEMG) interface, which is an ideal tool for training and rehabilitation applications. The processed sEMG signals are encoded after digitization to control the animated hand. Experimental results demonstrate the effectiveness of the sEMG control system in acquiring sEMG signals for real-time control. Users have also the ability to connect their prostheses with the training system and observe its operation for a more explicit demonstration of movements.

Keywords: Training system · Computer-aided · sEMG · Control prosthesis

1 Introduction

Hands are important parts of human body without which it is difficult to carry out most of the daily tasks. The loss of hands not only poses a huge barrier in daily life of an amputee, but also impacts them both emotionally and socially. The causes for upper extremity amputations include cancer, diseases, traumas, or congenital complications. The negative impact of amputation on the life of an amputee is partly overcome through upper limb myoelectric prosthesis, which is an artificial device controlled by the myoelectric signals from muscles in the residual limb. The use of myoelectric prosthetic limbs is not new, as the first myoelectric prosthetic device was demonstrated at the Exportmesse, in Hannover in 1948. However, the field progressed at a slower pace, especially in the initial years due to the post-war impact on industry in Europe. This is evident from the fact that it took 16 years (i.e. 1964) to commercialize the first myoelectric-controlled prosthetic limb [1]. Since then, there have been several studies focusing on the control techniques, for example, using toe gesture sensors [2], targeted muscle reinnervation (TMR) [3], and fully implanted myoelectric sensors [4]. The majority of these works involve commercial robotic/prosthetic hands. The most commonly used control system is the surface EMG (sEMG) based control, where a number

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of sEMG electrodes are placed on one or two target muscle groups to acquire myoelectric control signals.

Evidences suggest that despite promising improvements the satisfaction of users and the rate of use have not increased [5, 6]. This is partly because the training/learning systems, which is an important element, has received little attention over the past years. An intuitive training system is essential during rehabilitation process allowing patients to get trained in order to control their muscle signals. In this regard, a number of training approaches have been explored, but often they are found to be insufficient. For example, the *tracking training system* [7, 8], which uses a 1-D position-tracking task to allow the user to control the position of a mark on a graphical environment with the sEMG signals, is less intuitive as there is no visual representation of the movement of the prosthesis. Therefore, the user has less control over the learning experience. Another technique is the *prosthesis-guided training system* [9, 10], which uses the motions of the prosthesis to automatically recalibrate it. However, it does not allow the user to estimate the grip force. Further, users cannot be trained without having their prosthesis, which is usually expensive and takes a significant amount of time to be fabricated. A different training technique is the *bilateral training system* [11, 12], where the continuously evolving EMG signals can be associated with different parameters, such as the speed and direction of the movement. However, this system can be used only by unilateral amputees excluding a large part of patients. A fourth technique that has been proposed, which is also used in this work, is the *computer-based training system* [13–16]. Commonly, these applications require complicated control algorithms relying on the requirements of the configuration of the personal computer, and on the availability of software tools that will allow the design of sophisticated animations.

We have developed a 3-D interactive animation using the open source integrated development environment called “Processing” for training wrist amputees, which provides visual feedback by eliciting the desired response of the virtual hand. As there are no specific requirements for the configuration of the personal computer, users can be trained at home using their computers. A simple control algorithm has been developed to allow users to choose between *four* pre-determined gestures, and to estimate the applied grip force. To test and evaluate the proposed training/learning system we have developed a low-cost sEMG control system based on the Arduino microcontroller, and integrated the two systems.

This work is organised as follows: The challenges and the proposed solutions on the design of myoelectric-controlled prostheses is presented in Sect. 2. The architecture and implementation of the proposed system is presented in Sect. 3. The integration of the sEMG control system with the computer-aided myoelectric training system, and the experimental results of the integrated system are presented in Sect. 4. Finally, the concluding Sect. 5 summarises the results and presents directions for future work.

2 Myoelectric Control System

The real challenge in designing a myoelectric-controlled upper limb prosthesis is to provide an intuitive myoelectric control system with high functionality and dexterity. Figure 1 shows a typical control system for upper limb myoelectric-controlled

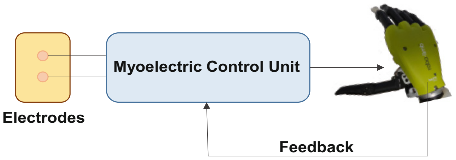

Fig. 1. Myoelectric control system.

prostheses, which incorporates the differential surface electrodes, the myoelectric control unit including the analogue and digital circuitry and the control algorithm, and finally the feedback loop which provides to the user information such as the applied grip force.

One of the major problems affecting the operation of the prosthesis is the acquisition of a reliable and useful signal. This problem may be caused by the dislocation of the differential surface electrodes with respect to the target muscle, or by the introduction of biological activities of other muscles within the residual limb when the target muscle is contracted; cross talk. For that reason, the electrodes are fixed in the prosthetic socket housing. Solutions that have been explored include prosthetic sockets incorporating methods of suspension [17], incompressible fluid [18], and struts arranged longitudinally with respect to the residual limb [19], so as to allow small movements of the prosthetic socket without altering the position of the electrodes. Studies towards the improvement of signal recognition [20], have also demonstrated a substantial improvement in the acquisition of the signal. However, there is a significant increase in the number of surface electrodes, which increases the discomfort of the users, and the training time in which they learn how to operate their prostheses. Towards this direction, there is an increased research activity on thin, flexible epidermal electronics bonded to the skin that can capture muscle activity, and without being perceived by the user in the course of the day [21, 22]. Furthermore, a gap in the market still exists concerning the feedback that users should have from their prosthesis. This refers especially on the applied grip force, so as an object to be neither firmly, nor loosely grasped. Solutions that have been explored towards this direction focus on tactile sensing chips that mimic the properties of the human skin [23–26].

3 Architecture and Implementation

The proposed apparatus consists of five wet surface electrodes; two electrodes located on Flexor Capri Ulnaris muscle, two located on Extensor Capri Radialis muscle, and the last electrode was located on the elbow as the reference electrode. The electrodes were connected in a bipolar configuration in order to eliminate the common noise between the two electrodes, and lower cross talk. All the bipolar electrodes are connected to a Printed Circuit Board (PCB) designed as an Arduino-microcontroller shield. On PCB, the surface myoelectric signals taken from the surface of the forearm are detected, filtered, and rectified. Subsequently, the processed analogue EMG signals are sent to the microcontroller where they are digitised and encoded, so as to send the

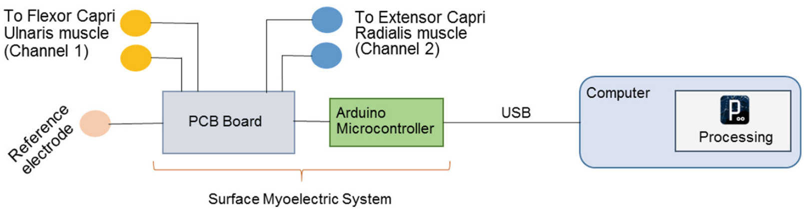

Fig. 2. System architecture.

appropriate commands via the serial port to the virtual animation model, which is designed using the integrated development environment called “Processing”. The block diagram of the implemented system is illustrated in Fig. 2.

3.1 Myoelectric System Design

The myoelectric system is powered by a bipolar analogue power supply, created by a 9 V battery connected to a decoupling capacitor and a TLE2426 rail splitter from Texas Instruments®. The signal conditioning system consists of two channels as illustrated in Fig. 3. Each channel comprises of two elements; the pre-amplification stage, and the signal conditioning circuit. The pre-amplification stage consists of a dual INA2126 instrumentation amplifier from Burr-Brown®. Each one of the two signal conditioning circuits (one for each channel) contains a quad op-amp TLC2274ACN integrated circuit from Texas Instruments®, and consists of a second order band-pass filter made up of a high-pass filter section in series with a low-pass filter section in a second order Sallen-Key topology. By the use of the spectrum analyser on a bit scope, the usable range of EMG frequencies was found to be approximately 10–600 Hz. However, the frequency content of power line interference, which is 50 Hz in Europe, can cause major problems. To reduce the electromagnetic interference from power line and other electromagnetic sources, the filters suppress the unwanted noises [27] and pass the low frequencies from 50 Hz to 600 Hz to a full wave rectifier which is subsequently

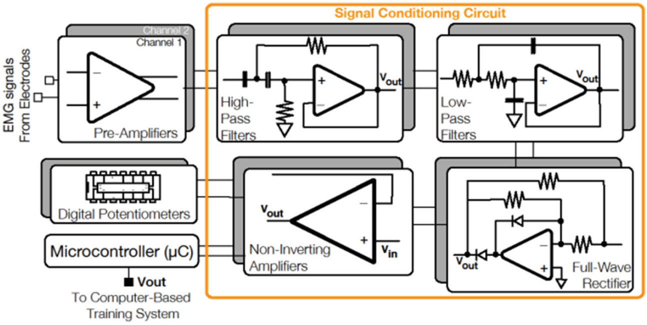

Fig. 3. Signal conditioning system schematic on the PCB.

Table 1. Myoelectric upper-limb control system performance summary

Specification	Value
Supply Voltage [V]	±4.5
Power Consumption [mW]	130
On PCB	
Pre-Amplifier Gain [dB]	1000
Pre-Amplifier Offset Voltage [μV]	±150
Pre-Amplifier CMR [dB]	90
High Pass Filter Cut-Off Frequency [Hz]	50
Low Pass Filter Cut-Off Frequency [Hz]	600
Non-Inverting Amplifier Gain [dB]	256/n, 0 ≤ n≤256
Off PCB	
Microcontroller	Arduino Leonardo

connected into a non-inverting amplifier. The gain of the non-inverting amplifier can be adjusted between 1 and infinity in theory, by a digital potentiometer MCP4151-503 from Microchip®, which is controlled by Arduino microcontroller through the serial peripheral interface (SPI) protocol, bypassing a byte with a decimal value n to the digital potentiometer. The output of each channel is routed to one of the eight available ADCs on Arduino microcontroller, where their digital values are processed. Finally, these processed values are sent serially to the program "Processing". Table 1 summarized the performance specifications of the implemented system.

3.2 Arduino Firmware Design

There are two array variables in the program storing 30 EMG samples, where 15 samples correspond to Channel 1, and the other 15 to Channel 2. The samples are summed and their Mean Absolute Value (MAV) is calculated. Subsequently, the MAVs are stored into two arrays. There are two threshold values (upper and lower) that can be changed easily at any time from the software, and against which the MAVs will be subsequently compared. A subroutine searches for a pattern of *three* values. Each value corresponds to one MAV stored in one of the two arrays, and only after the EMG signals go below the lower threshold, a new MAV can be registered as a valid one. Essentially, users have to flex their Flexor Capri Ulnaris muscle (Channel 1) or the Extensor Capri Radialis muscle (Channel 2) three times, in order to access the control of the four available states. When the three detected values are above the upper threshold value, the hand will move at high speed, when the three detected values are in between the lower and upper threshold value, the hand will move at low speed, and when they are below the lower threshold the hand will be instructed to stop. When the three values are detected in *Channel 1*, which corresponds to the Flexor Capri Ulnaris muscle, there will be a shift to the next mode of operation, and when the three values are detected in C*hannel 2*, which corresponds to the Extensor Capri Radialis muscle, the state machine will move to the previous mode of operation. For every different state and speed, the program sends the appropriate commands, via the serial port, to the program “Processing” where the commands are decoded in order to control the computer-animated hand. An optimal communication is established by the *Serial.begin* command at a 9600 baud rate, since there was not a significant time delay in the visualisation of the movement, and at the same time the transferred data are less likely to be corrupted at such a slow rate of information transfer.

3.3 User Training System Design

“Processing” is an open source integrated development environment with a huge supporting community. Since it can be interfaced with Arduino projects, it offers a powerful tool that runs in every computer or tablet without any demanding requirements, providing solutions for the interaction and visualisation of dynamic systems in a 2-D or 3-D environment.

Each finger was designed using three cones and four spheres. Cones play the role of the phalanges (distal, middle, and proximal), and spheres play the role of finger joints and fingertips. By changing the spatial coordinates of the cones and spheres the index, middle, ring, and pinkie fingers were created. Thumb was created in the same way as the other fingers, but it consists of two cones instead of three (proximal and distal phalange), and three spheres instead of four. Furthermore, palm was designed by creating two identical polygons separated by a distance and by connecting these polygons together. Thumb was attached with the palm by creating a thumb housing. The thumb housing consists of two identical polygons separated by a pre-defined distance and connected together. Finally, a sphere was designed playing the role of the thumb joint. The animation model is programmed in that way that it can perform all the postures of a human hand without making any abnormal moves, by setting the minimum and maximum permissible angle of the movement between 0 and 90^{o} for each finger joint. The hand animation is also customizable, in the sense of permitting the user to choose the colour of the skin, the colour of the background, or the viewing orientation.

The proposed code in “Processing” offers a very important characteristic, as each finger is independently instructed to move at a certain direction and speed. Users can easily change, or create the gestures that they want to include only by simply changing the values of five variables; *enable[0], enable[1], enable[2], enable[3],* and *enable[4]*. These five variables correspond to the direction of the thumb, index, middle, ring, and pinkie, respectively. When a variable equals to the value *1.0* the corresponding finger is instructed to open, when it equals to *2.0* the finger is instructed to close, and when it equals to *0.0* the finger is instructed to stop. The user can adjust the direction of movement for each finger by changing these values and without changing anything else in the code. In addition, users can adjust the parameter of speed simply by changing one variable; *stepSize*. In our proposed training system the value of 2.0 for that variable gives a nice representation of high speed movement, and the value of *0.*7 a nice representation of low speed movement. After the completion of training/learning process, the prosthesis can be programmed to offer the same modes of operation that users decided that are well fitted to their needs.

Another aspect of the proposed training system is that it can be interfaced with the robotic prosthesis through a second serial port. When the user is pleased with the selection of gestures, the operation of the robotic prosthesis can be adjusted appropriately so as to perform the same gestures that the user selected. Users can also train themselves by observing the hand animation and the prosthesis operating at the same time. This offers the advantage of modifying accordingly the parameters, such as the speed of the movement, or the desired position of the fingers during the performance of the gesture by the prosthesis.

One of the most important characteristics of the natural human limb is the ability to quickly interchange between the relatively small number of the main hand grips used in daily activities [28, 29], making accurate predictions. The number of implemented modes of operation in our proposed system is four. A state machine is implemented to allow users to choose between these four available modes of operation. Figure 4 shows the interactive hand animation performing four different gestures.

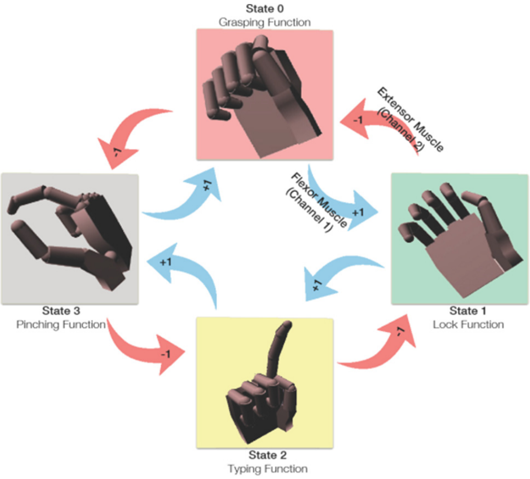

Fig. 4. The training system in the four modes of operation.

4 Experimental Results

The designed hand animation model is instructed to perform the corresponding gesture at the corresponding speed depending on the commands received from the Arduino microcontroller, by initialising the serial port in “Processing” using the *port = new Serial(this, “COM7”, 9600);* command. Each finger is instructed to move in the given direction (open, close) and speed (high, low, stop), depending only on the level of the *MAV* and the *Channel* that the signals are detected. Figure 5(a) shows the measurement setup, and the fabricated PCB with soldered components has been depicted in Fig. 5(b).

During training process users can adjust these thresholds to their particular case, so as to minimise the fatigue of the muscles. Users can also adjust equally easily from the

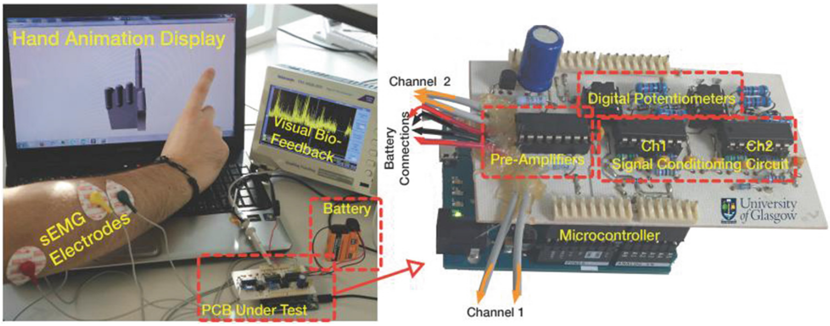

Fig. 5. (a) Measurement setup and (b) fabricated PCB with highlighted crucial components.

software the gain of the amplifiers by controlling the digital potentiometer, in order to increase or decrease the amplification of their muscle signals. Initially, the calibration of the system can be done with the help of the prosthetist using an oscilloscope, and later by the users, who will decide if the level of thresholds and the levels of amplification of their signals suit their needs. Figure 6(a) and (b) show the three different regions of the useful EMG signals captured in Channel 1 (Flexor Capri Ulnaris muscle) and Channel 2 (Extensor Capri Radialis muscle) highlighting the thresholds against which the MAVs of these signals will be subsequently compared.

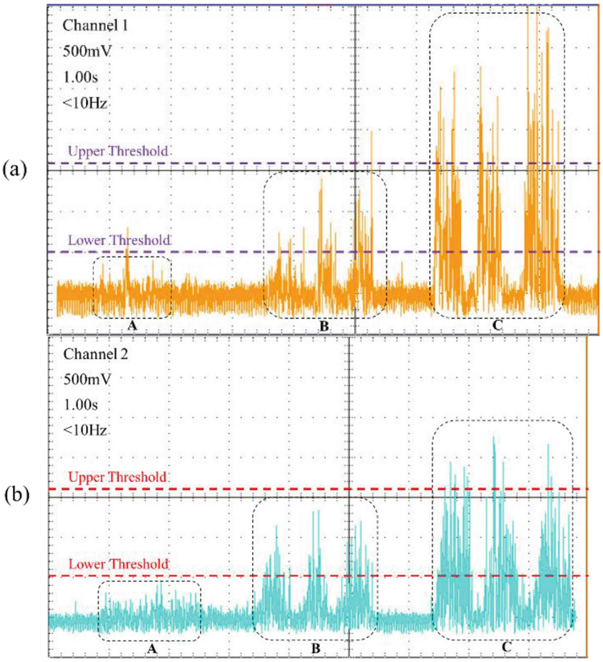

Fig. 6. Measured sEMG signals from (a) Channel 1, and (b) Channel 2. A: STOP, B: LOW speed, C: HIGH speed.

Users can train themselves through “simple moves”, or specific “tasks”. The “simple moves” refer to opening, closing, and the transitions between the states, so as users to learn which the available modes of operation are and how they can achieve the desired gesture at the desired speed. The “tasks” refer to assignments like answering the phone, opening the door, or taking a pen from the table that users can assign to themselves in order to have an effective operation of their prosthesis through the daily activities.

5 Conclusion and Future Work

This work presents a computer-aided training/learning system for myoelectric-controlled prostheses. The training system was simulated in “Processing” by creating a re-programmable code to allow users to add or change the parameters as they like, initially with a trained clinician and later by themselves. The proposed training system can be used for at-home training. Bio-feedback can be provided from a portable oscilloscope to either adjust the thresholds or the gain of the amplifiers to required levels.

Future work will focus on the implementation of a control system which will accommodate motions of the wrist, such as flexion/extension, or wrist rotation to allow the user to train on a fully functional below wrist model. The addition of the forearm and upper arm will allow a more complete training experience for higher level amputees.

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