Refunctionalization of passive electromyographic electrodes to upper-limb prosthesis applications. ⋆

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Abstract: The rehabilitation potential of myoelectric prosthesis for upper-limb amputee led to its wide investigation in the last two centuries. However, its everyday use still faces several challenges, such as high cost of available commercial models and unsatisfactory functionality. Thus, the present work proposes the development of an upper-limb prosthesis, focusing on the myoelectric signal measurement system. The electromyography system was integrated to an adjustable armband with three measurement channels made with traditional disposable Ag/AgCl passive electrodes readapted as active dry electrodes. For purposes of performance evaluation and database creation, myoelectric signals were collected from 12 healthy volunteers performing daily activities using a bottle. It was verified the high influence at measurements of volunteer physical characteristics, especially those relating to physical fitness, amount of forearm skin or adipose tissue, and sweating. Furthermore, the electrodes also showed high sensitivity to positioning and movement artifacts at the electrode-skin interface. A database of myoelectric signals was created for future pattern characterization. The electromyographic armband performance was demonstrated by driving a claw-type prosthesis using direct control.

Keywords: Upper-limb Prosthesis, Myoelectric Signal, Electromyography.

1. INTRODUCTION

The hands are associated with creating, working, touching, communicating, manipulation and in some way they participates of almost every single action performed by a person. The movement synchrony and complementarity between each hand associated with other biological systems allow a wide manipulative capacity. Therefore, the amputation impacts must be addressed in all its extension considering personal, functional, social and psychological levels (DeLuca, 2006).

The prosthesis is an engineering simplification of the human hand complexity, reproducing at some level the aesthetic appearance and returning some key functionalities. The main differential of the myoelectric prostheses is the control method by the muscular contraction of the residual limb, through which it is intended to achieve greater naturalness and intuitiveness.

Wars have always driven the development of prostheses for the rehabilitation of knights, pirates and soldiers maimed during battle. It was also 2 years after the end of World War II, in 1948, that University of Munich physics student Reinhold Reiter created the first myoelectric prosthesis. The first clinically significant model was developed in 1960 by Russian scientist Alexander Kobrinski. (Zuo and Olson, 2014)

In recent years, the focus in the development of prostheses has focused on achieving greater functionality and anthropomorphology, thus seeking to improve the patient/prosthesis interface through myoelectric control and tactile feedback. Thus, the current state of the art of prostheses are 5-finger models with multiple DOFs (degrees of freedom), whose positioning is combined in 6 to 14 different configurations of the bionic hand. The drive is done by DC motors and the feedback mechanisms use pressure, slip, temperature and skin distension devices. A more natural appearance is achieved through cosmetic gloves that cover and protect the electromechanical system of the prosthesis. (Saudabayev and Varol, 2015)

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The works by Belter et al. (2013) and VanDerRiet et al. (2013) compare the main models of commercial and academically developed myoelectric prostheses. Among the commercial models are *iLimb*, *Bebionic* and *Michelangelo* developed by *Touch Bionics*, *RSL Steeper* and *OttoBock*, respectively. Meanwhile, *Vanderbilt University Hand*, *Southamption Hand* and especially *SmartHand* are models of great relevance from the various academic research developed in the world in this area.

The analysis presented allows us to identify the main limitations of the developed myoelectric prostheses, such as the weight greater than that of the human hand (average of 400g), which is an important factor in the comfort of use. Another point is the lack of adequate control of the force of the gripper to grip objects in relation to precision and power. Commercial models supply at least the power requirement required for everyday activities (minimum of 68N), but even the most intense force grips, have 4 times less force compared to a healthy hand. In addition, all commercial models have a high cost of more than 35 thousand dollars. Belter et al. (2013)VanDerRiet et al. (2013)

Electromyographic electrodes characterize myoelectric prostheses and are an essential part of the user-prosthesis control interface. They measure the myoelectric signal generated during the muscle contraction by which the user operates the prosthesis. Surface electromyography is already an established technique in clinical diagnosis. However, the application in myoelectric prostheses adds specific challenges, highlighting fundamental differences in the acquisition conditions and in the design requirements of the electromyographic device.

There are electrodes of the most varied types, varying the electrochemical characteristics according to the metal used, geometry, presence of electrolyte and incorporated electronics. Specifically for application in prostheses, the most traditionally used electrodes are made of silver coated with a thin layer of silver chloride (Ag/AgCl), with a non-invasive surface contact, with a dry interface (*dry*) without electrolyte and bipolar assets with signal pre-processing circuit.

Although several companies sell different models of prostheses, they are still not widely used, especially in Brazil. Still, the high cost and unsatisfactory adaptability compromise the accessibility of this rehabilitation technology.

Thus, this project proposes the development of a myoelectric prosthesis for transradial amputations of the upper limb with 1 degree of freedom (DOF) with an affordable price and easy adaptation. A new, cheaper measurement system is presented by recycling and retrofitting passive electrodes commonly discarded as dry active electrodes. In addition, the simplification of the system by limiting the functionality to 1 DOF, facilitates adaptation and also contributes to cost reduction. The specific objectives are guided:

- Development of a myoelectric signal measurement system observing the following specifications: non-invasive, without electrolytic conductive gel, low cost, reduced size and weight;
- Development of a protocol for collecting myoelectric signals from healthy volunteers.
- Formation of a database of myoelectric signals for future extraction of characteristics and parameters.
- Implementation of a control system to drive the gripper.

2. METHODOLOGY

2.1 Measurement

The electromyographic bracelet (Fig. 2-D) with three reading electrodes plus a reference electrode was developed for measuring myoelectric signals. The electrodes are housed in specially prototyped 3D printer housings, designed to slide through the elastic bracelet, allowing the adjustment of the position of the electrodes on the arm, while performing a gentle vertical compression ensuring local fixation.

3M passive surface electrodes, model 2223BRQ, were adapted and integrated into a pre-processing circuit forming bipolar active electrodes. The stainless steel metal pins with Ag/AgCl coated cotter pin of the 3M electrodes were insulated and coupled to *Snap Connect Adapters* (Fig. 3(a)) soldered to the circuit board as shown in Fig. . 3.

The acquisition and filtering circuit design developed and detailed in previous works (Morais, 2016) and Morais (2019), was adapted for surface mounting aiming at its reduction (final dimension of 1.4×3.0 cm).

The signal acquisition and the first amplification stage were performed with a Burr-Brown INA118P instrumentation amplifier, with a gain of 501. The signal was filtered by a bandpass filter composed in two stages by a passive high-pass filter and by a first-order inverting active low-pass filter, with cut-off frequencies of 50 Hz and 500 Hz, respectively. The low-pass filter was implemented with the LF351 low-frequency operational amplifier with a gain of 4.39 and an offset set to 1.6V. So the total combined gain is 2199.5. In this way, the signal was properly adjusted for the microcontroller processing. The scheme of the pre-processing circuit parts is shown in 1.

System power is provided by a single 2-cell, 7.4V Li-Po battery, connected to the 7660 voltage inverter, generating -7.4V. The voltage regulated to +5V and -5V by the ICs 78L05 and 79L05, respectively, supply the amplifiers. A third 78L33 regulator adjusts the voltage to 3.3V to power the microcontroller. The final circuit was placed in a 3D-printed box and incorporated into the bracelet (Fig. 2-A).

2.2 Microcontroller and Computer Processing

The microcontroller was used to capture the myoelectric signal through the A/D converter and to control the prosthesis. In this project we used the STM32 microcontroller with ARM 32 Cortex-M3 processor, produced by *STMicroelectronics* embedded in the STM32F103c8t6 board, programmed in the Mbed environment. The A/D converter has 10 channels with 12-bit resolution and an operating voltage range from 0 to 3.3V.

The conditioned signal was sampled at a frequency of about 1560 Hz and stored in a vector containing 512

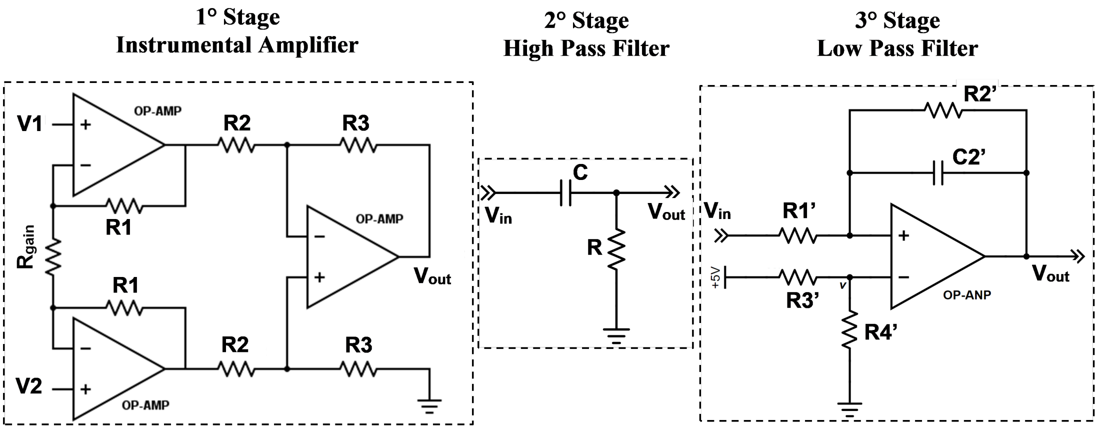

Figure 1. Pre-processing circuit.

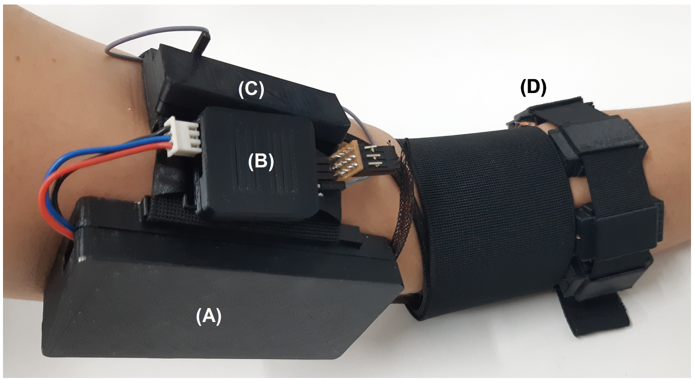

Figure 2. Electromyographic bracelet: (A) Battery, (B) Power Supply, (C) STM32 Microcontroller, (D) Electrodes

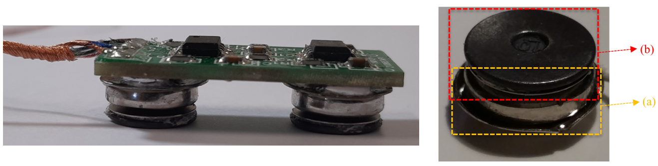

Figure 3. Pre-processing board with myoelectric sensor: (a) Snap Connect Adapters and (b) electrode pin.

samples from each of the three channels. This vector is transferred via serial communication to the computer and received by a serial terminal developed in *Python 2.7.1*. In addition to dealing with incoming serial data streaming, another feature of this application allows the researcher to manually insert markers corresponding to the movement performed by the volunteer. This function is important for the identification of the different tasks in the analysis of the collected myoelectric signal.

The data stream is then transferred from *Python* to *MATLAB R2015a* using the *Lab Streaming Layer* communication protocol. This system synchronizes streaming data for analysis, simultaneous viewing of the three channels in real time, and data recording. This feature helps the researcher to ensure that all electrodes are measuring correctly during the assessment.

The *Lab Streaming Layer* (LSL) is a powerful tool for experimental research involving time series measurements from multiple sources. It allows for unified collection by creating a data communication network, performing time synchronization, (almost) real-time access, as well as optionally centralized collection, visualization and disk writing.

LSL basically works by creating two types of *streamings* (flows) of data: *Stream Outlet* (flow out) and *Stream Inlet* (flow in). A *Stream Outlet* makes data streams available on the laboratory network. The data can be inserted in three different ways: sample by sample, by batches, or with a constant rate.

3. RESULTS

Tab. 1 shows the profile of the volunteers who participated in the study. Among them, five were men and seven women, almost all of them had dominance of the right hand with the exception of one, and half of the group had some muscular physical preparation.

During the collections with the volunteers, it was possible to observe some characteristics of the developed sensors and analyze their performance. Although the protocol established in compliance with SENIAM's recommendations requires trichotomy in the measurement region, the electrodes work without shaving the hairs, therefore, it is not mandatory.

Furthermore, the factors identified that most affected the sensitivity of the electrodes were positioning, amount of skin or adipose tissue and sweat. Often when the sensors did not capture the myoelectric signal, a small longitudinal displacement to the muscle of the order of millimeters induced the function. Another notable point was that signal detection was markedly more difficult in people with more skin or fatty tissue on the forearm. In these cases, it was important to control the tightness of the elastic tape so as not to form protuberances of skin between the electrodes that would cause them to move during the

Table 1. Profile of the volunteers participating in the research.

ID	Hand dominance	Sex	Age	Height	Weight	BMI	Physical Activity	Start
1	Right	Male	28	1,81	85	25,95	No	Left
2	Right	Female	28	1,65	70	25,71	No	Right
3	Right	Female	27	1,70	60	20,76	Yes	Right
4	Right	Male	28	1,88	92	26,03	Yes	Left
5	Left	Male	30	1,72	80	27,04	No	Left
6	Right	Female	26	1,59	66	26,11	Yes	Right
7	Right	Female	45	1,55	63	26,22	Yes	Left
8	Right	Female	36	1,60	48	18,75	Yes	Left
9	Right	Male	33	1,79	93	29,03	No	Left
10	Right	Female	44	1,65	83	30,49	No	Left
11	Right	Male	53	1,82	81	24,45	Yes	Left
12	Right	Female	52	1,59	86	34,02	No	Right

execution of the task. It was also important to ensure that the skin was taut between the two measurement points of each electrode. Collection was also considerably more difficult in people with more sweat, especially on warmer days. As the measurements in each arm took a long time, it was necessary for these cases to repeat the skin cleaning, ensuring the correct capture of the signals throughout the experiment.

This observed behavior was widely discussed in the work of Cömert et al. (2013) and is related to motion artifacts. The electrical modeling of the skin and electrode, shown in Fig. 4, helps to understand how motion artifacts result in myoelectric signal attenuation.

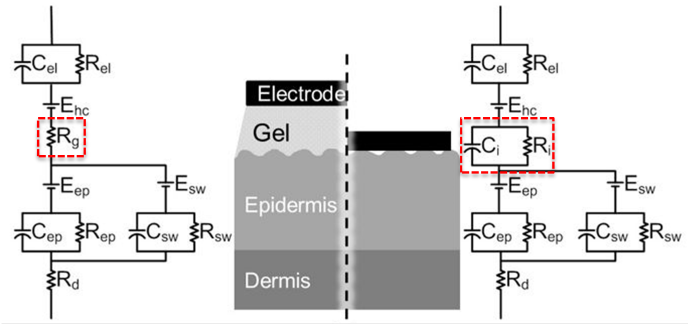

Figure 4. Electrical modeling of Dermis (Rd), Epidermis (Eep, Rep, Cep), sweat (Esw, Rsw, Csw), electrode-skin interface (Rg or Ri, Ci) and electrode (Ehc, Rel, Cel). Highlight in red the difference between circuits with and without electrolytic gel. (Cömert et al., 2013)

The main component of motion artifacts comes from the change in potential difference of the epidermis Eep. Deformation of the skin by lateral stretching or application of vertical pressure alters the ionic gradient, and hence the electrical potential, between the layers of the epidermis.

In the case of dry electrodes, motion artifacts are also associated with the electrode-skin interface. The application of pressure and relative movements between skin and electrode considerably alters the contact area, affecting the electrical properties of the interface. As the interface is a thin and uneven layer of moisture with eventual air bubbles, small changes in the location of the electrode will cause large disturbances in the ionic concentration near the electrode, altering the half-cell potential. The interface impedance was modeled as a capacitor Ci in parallel with a resistor Ri. This reactance is not observed in the presence of electrolytic gel, which associated with the adhesive tape, forms a kind of weld with the skin, making the electrode physically stable in relation to lateral and vertical changes.

In addition, the results of the Cömert et al. (2013) research determined a decrease in movement artifacts by applying vertical pressure between 15 mmHg and 20 mmHg, although the increase in pressure was shown to be harmful to the quality of the signal. Thus, as clearly observed during the collections, excessive tightening of the elastic bracelet, in addition to causing discomfort and superficial skin lesions, did not result in better capture quality.

Also, although the model proposed in Cömert et al. (2013) suggests that sweat can act as an electrolyte, it can also short-circuit the measurement points of the differential electrode. This fact is a possible explanation for the difficulty of measuring in volunteers with sweaty skin.

The subcutaneous fat is in between the dermis and the muscle and has high resistance, directly proportional to the thickness of the layer (Petrofsky, 2008). This resistance is highly subject to variations in blood flow in the skin and fat caused by a local and neurological control, since blood has high electrical conductivity (Sadick and Makino, 2004). Thus, the high resistance explains the difficulty of detection in people with more adipose tissue in the forearm, increased noise and considerable attenuation of the myoelectric signal for these cases, as seen in Fig. 5 for the individual #12. However, these effects did not occur in a standard way for all volunteers with high BMI (Body Mass Index), given that the same resistance varies depending on individual conditions.

In a first analysis of the collected signals (Fig. 5), the variation in the noise amplitude between the muscle groups of the same individual and between volunteers with different body types was highlighted. The individuals #8, #6 and #12 selected for this comparison have a BMI equal to 18.75; 26.11 and 34.02 respectively. Although the differences in noise amplitude apparently did not follow a fixed pattern, some trends could be noticed. In addition to the adipometric issues already discussed, it was observed that the finger flexor measurements were generally the noisiest and had the lowest signal-to-noise ratio. This result is due to the fact that the flexor of the fingers is in the deeper layers of the forearm, in such a way that the superficial electromyographic signal suffers more attenuation and is subject to greater interference.

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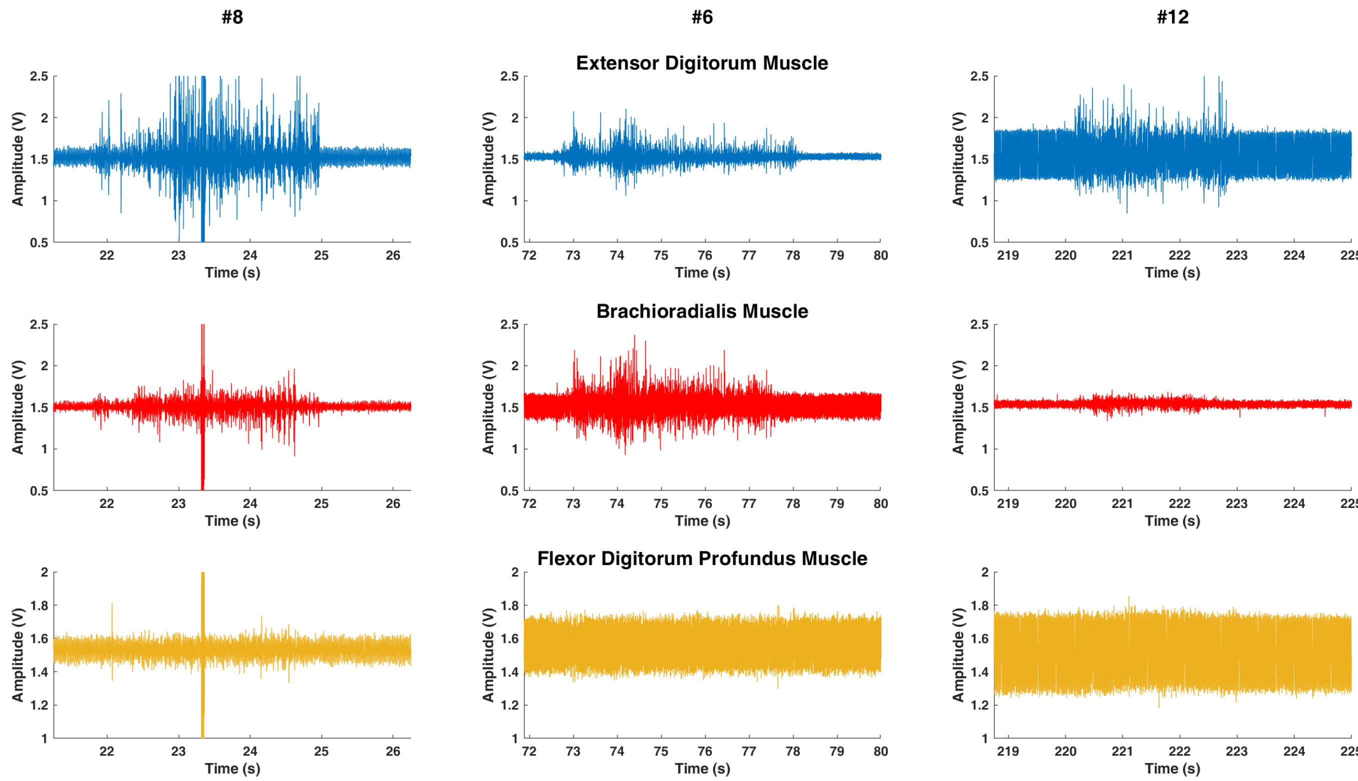

Figure 5. Comparison of the myoelectric signal of the M3 movement between volunteers with different body types (#8 – BMI 18.75 ; #6 - BMI 26.11; #12 - BMI 34.02) and between muscle groups . There is a trend of increased noise and signal attenuation the greater the individual's BMI / amount of adipose tissue. Furthermore, the flexor muscle measurements are the noisiest and have the lowest signal-to-noise ratio due to their depth in the forearm.

Another issue to be discussed is the use of the same bottle with constant load for all volunteers. As simple tasks were evaluated using a relatively light load, some volunteers did not have the musculature sufficiently required, generating low amplitude activation peaks, not observing the clear and differentiable activation intervals in the myoelectric signal during the tasks.

The electromyographic bracelet was tested with a 1DOF prosthesis. The trigger for opening and closing the grip was implemented using only one channel for measuring the myoelectric signal.

The STM32 microcontroller operates in continuous cycles, making the analog-to-digital conversion of 512 (equivalent to 641 ms) each cycle. For each measured interval, the signal modulus was then calculated and the DC level was subtracted. The signal strength per window was then obtained by summing the rectified signal. Two intensity thresholds, τ_1 and τ_2, operate the gripper movement. The first threshold τ_1, set slightly above the noise intensity, is exceeded by a slight muscle contraction. It triggers the consecutive closing and opening of the gripper. The second threshold τ_2, higher than τ_1, is reached by a strong contraction. It operates the closing of the gripper by holding it in this position or holding an object. To open it again, the system must be triggered by a new strong contraction command. It is also worth mentioning that the thresholds must be adjusted per user according to the noise baseline level.

The operational simplicity of the proposed algorithm favored a quick response of the prosthesis to the user's intention to move. The system was also tested identifying the muscle activation interval by the moving average, but the computational complexity was reflected in the activation response time.

4. CONCLUSION

The work demonstrated the feasibility of reusing passive electrodes with electrolytic gel adapted as dry active electrodes. However, in the performance analysis of the electromyographic bracelet implemented, it was observed greater difficulty in positioning and measuring in individuals with more adipose tissue and sweating. This should be considered for improving the device in future works.

Some design details of the electromyographic bracelet related to the reduction of movement artifacts, proved to be crucial in the performance of the system, highlighting the immobilization of each electrode at the measurement point by pressing it on the skin and the preference for soldering to connect the wires.

Superficial myoelectric signals from three different muscles were collected in 12 non-amputation volunteers performing daily tasks with a bottle. The myoelectric activation of complete sets of movements that make up a task (M1, M2 or M3) performed naturally were evaluated.

The collections also allowed the analysis of the general performance of the electromyographic bracelet developed. It was noted that the measurements were highly influenced by the physical characteristics of each volunteer, especially those related to physical fitness, amount of skin or adipose tissue of the forearm and sweating. Furthermore, the electrodes also showed high sensitivity to positioning and movement artifacts at the electrode-skin interface. Especially, difficulties were observed with some of the volunteers in the initial positioning of the electrodes, guaranteeing the simultaneous measurement in the three channels. These challenges will need to be considered in future releases.

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