A wearable low-power and low-cost electromyographic sensor for arm prosthesis

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Abstract— This work presents the design considerations for an electromyographic (EMG) sensor destined to be integrated in a low-cost robotic arm prosthetic for below-elbow amputation, which is being developed in Paraguay by a multidisciplinary teamwork. The design exposed on this paper is addressed to optimize the embedded 14 bits ADC of the MSP432P401R microcontroller. It achieves the amplification and filtering of EMG signals with an input-referred noise of 1.35 µVRMS, and a CMRR greater than 95 dB, with a very low quiescent current consumption of 40 µA. The sensor also includes a common-mode feedback circuit that drives a reference electrode.

***Keywords*— electromyographic sensor, biopotentials amplifiers, arm prosthetic.**

***Resumen*— Este trabajo presenta las consideraciones de diseño para un sensor de señales de electromíograma (EMG), diseñado para ser integrado en una prótesis robótica de bajo costo, para amputaciones bajo codo, que está siendo desarrollada en Paraguay por un equipo de trabajo multidisciplinario. El diseño presentado está dirigido a optimizar el rango dinámico del ADC embebido en el microcontrolador MSP432P401R de Texas Instruments. El sensor efectúa la amplificación y el filtrado de las señales de EMG con un ruido referido a la entrada de 1.35 µVRMS, un CMRR mayor a 95 dB y un muy bajo consumo de corriente, menor a 40 µA. Este también incluye un circuito de realimentación de modo común.**

***Palabras clave*— sensor electromiográfico, amplificadores de biopotenciales, prótesis de brazo.**

I. Introduction

The electromyographic (EMG) signals are of usefulness to diagnose nervous system or muscular diseases [1] but are also used as control signals for applications so different as PC peripherals [2], rehabilitation devices [3] or prosthesis [4,5]. The major advance of the EMG signals against other biomedical signals is the possibility of volitionally modulate its amplitude with relative easiness, due to a correlation between muscle contraction strength and EMG amplitude.

The lack of a lower or upper limb, of the body is quite frequent nowadays. The causes of this condition are due to congenital amputations, illnesses or traumas caused by accidents. In Paraguay, according to the Ministry of Public Health and Social Welfare (MSPBS, for its acronym in Spanish), per day, on average, up to two people suffer loss of lower or upper limbs as a result of car or motorcycle accidents (MSPBS Web site, 14 / 01/2014) [6]. Likewise, the National Secretariat for the Human Rightsof Persons with Disabilities (SENADIS, by its acronym in Spanish), daily receives on average four patients amputated as a result of accidents (Ultima Hora newspaper, digital edition, 01/14/2014). The statistics, up to January 2014, shown that the number of people assisted by SENADIS due to amputations, reached 91,000, who also receive rehabilitation services.

The loss of a limb has an important impact on the life of an individual human being, as it experiences a decrease in their ability to perform daily activities and generally require the care or the accompaniment of a family member or the help of a person to perform certain tasks. Therefore, the improvement of the quality of life of the individual who experienced the amputation of a member requires the use of a prosthesis that facilitates their daily activities. However, the vast majority of existing prostheses on the market (Paraguay) are cosmetic, that is, their primary objective is to cover the absent limb without providing any functionality to the user, or they are merely mechanical, which implies a reduced capacity of movements and generally require the movement of some part of the body to obtain a certain action of the prosthesis. On the other hand, robotic prostheses, those that use electromyography to achieve movements, are more precise and varied, which is possible through the integration of sensors, processors, actuators and complex control algorithms. Although prostheses with these characteristics are available, they have a high cost due to the materials used in their manufacture and the use of proprietary technologies, so people with limited economic resources cannot access the prostheses.

Therefore, a development project of electromyographic arm prostheses for below-elbow amputation is being

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conducted, funded by the National Committee of Science and Technology of Paraguay which seeks to provide an accessible, functional and low-cost prosthesis model to improve the quality of life of the individuals who live with amputees.

A key block of the prosthetic arm is the analog front-end, that defines the quality of acquired signals and that can subsequently affect its performance. There are several commercial sensors for myoelectric signals, from development kits for makers as the MyoWare Muscle Sensor Kit (from Sparkfun), to professional products for researchers from companies as Delsys or Biometrics Ltd. Nevertheless, these products are too expensive for the desired low-cost design and makes it dependent of a third-party technology. Therefore, a development of a custom low-cost EMG sensor was conducted as a part of the development project of the prosthetic arm.

II. Materials and Methods

The project aims to provide a prosthesis that can perform the following basic functions: cylindrical grasp, hook, palmar and lateral grasps. The wrist to be used is a passive mechanism that can be rotated manually.

The socket itself will be made by a prosthetics and orthopedics company (Conforpes Paraguay) associated to the project and will be done specifically for each amputee. On top this socket, a 3D printed, PLA (PolyLactid Acid) forearm structure will be attached. Holes for the placing of EMG sensors will be placed at specific locations for each amputee. The forearm of each participant will be scanned using a 3D laser scanner, to be used as a base model for the 3D printed prosthetic forearm. Inside this structure, compartments will be designed and accommodated depending on the length of the remaining limb of the participant. Inside these compartments, components such as battery, MCU and cabling will be placed.

A functional block diagram of the prosthetic arm under development is presented in Fig. 1. User's EMG signal is obtained by the sensors and after being amplified, it is digitized by the ADC embedded in the microcontroller unit (MCU) for processing.

The MCU MSP432P401R from Texas Instruments was chosen for features such as its ultra-low power performance, its number of differential analog inputs, high resolution embedded ADC, variety of serial communication options, the ability to use floating point unit, and DSP capabilities. Furthermore, having a flexible MCU like this one, gives developers leeway to explore possibilities such as adding more sensors, or trying different filtering algorithms.

If a voluntary contraction by the user is detected, the MCU sends a command to the motor controller to activate the linear actuators inside the prosthetic 3D printed hand, thus allowing the fingers to open or close. While in motion, force sensing resistors placed at the fingertips will stop the movement of motors after a defined amount of force is measured in the hand, reducing energy waste.

To keep the weight of the prosthesis low, small 15 grams linear actuators where chosen. The linear actuator's motion reduces mechanical complexity inside the palm of the 3D printed hand to control finger movement and provides a force of 18 to 45 Newtons depending on its gear ratio. A 7.4v 2100 mAh lithium-ion battery supplies the power, which can meet the requirements for a prosthesis with many actuators to last for a day of use [7].

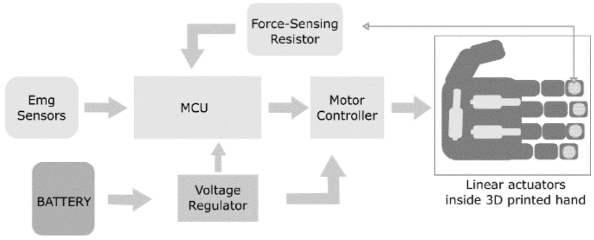

Fig. 1: Prosthetic arm functional diagram.

In this context, it is needed an EMG sensor that concentrates the electrodes and the electronic front-end, conditioning the EMG signal for the dynamic range of the embedded ADC. At the same time the sensor's outputs must be compatible with the 3.3 V tolerant inputs of the MCU. The use of wearable sensors without disposable electrodes and easy to wear is fundamental to achieve maximal comfort and practicality for the end user. In addition to these characteristics, wearable devices have important features, such as, small-size, lightweight, low-power consumption that are crucial for the effective operation of the prosthesis [8].

Hence as part of the aforementioned, this paper presents the design and development of an electromyographic sensor with the aim of integrating it into a functional, low cost arm prosthesis. The design of the sensor is based on commercial off-the-shelf components and, as it is depicted in Fig. 2, it was divided in two sections: the differential amplifier and the common mode reference.

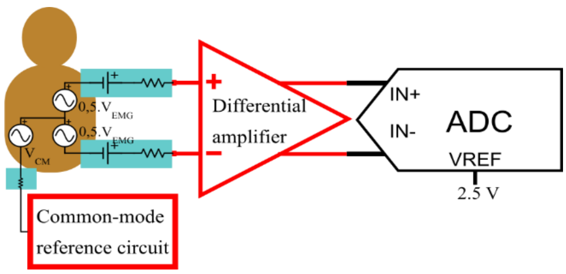

Fig. 2. Instrumentation model. In red are shown the EMG sensor blocks.

A. *Differential amplifier*

The aim of this circuit is conditioning the EMG voltage signal, that is differentially measured between two electrodes, to take advantage of the dynamic range of the ADC. This stage must also deal with the high impedance of the electrodes, offering a high input impedance, and must be able to drive the impulsive input current pulses of the ADC.

The ADC is a 14-bit successive approximation converter, with differential inputs and a 2.5 V dedicated voltage reference. Thus, the resolution of ADC is Δ = 300 µV. Considering a minimum desirable resolution of EMG signal of 1 µV, the amplification gain must be at least 300 V/V, meanwhile the maximum gain must be 1000 V/V to allow measuring EMG signals up to ±2.5 mV without saturation.

The proposed circuit is shown in Fig. 3. It is a fully differential amplifier, based on two operational amplifiers (OAs) which addresses the aforementioned specifications. This topology amplifies the input differential voltage $v_i = v_{iH} - v_{iL}$ to the differential output voltage $v_O = v_{OH} - v_{OL}$ by the gain,

$$G_{DD} = 1 + 2.\frac{R_1}{R_2}.\left[\frac{s.\tau_{HP}}{1 + s\,\tau_{HP}}\right].\left[\frac{1}{1 + s\,\tau_{LP}}\right], \qquad (1)$$

where $\tau_{HP} = R_1.C_1$ and $\tau_{LP} = R_2.C_2$. The time constant τ_{HP} sets the AC coupling frequency $f_{HP} = (2\pi\tau_{HP})^{-1}$ = 25.7 Hz, necessary to avoid the saturation of amplifier due to the electrode offset voltages and to reject the low frequency movement artifacts [9]. The amplifier also limits its bandwidth for frequencies higher than $f_{LP} = (2\pi\tau_{LP})^{-1}$= 219 Hz. Within the signal bandwidth, the designed gain can be simplified by $G_{DD} \approx 1 + 2\frac{R_2}{R1}$ = 726 V/V. The used OAs are both integrated in the dual OPA2333 from Texas Instruments, that has CMOS-based inputs and a quiescent current of just 17 μA per OA.

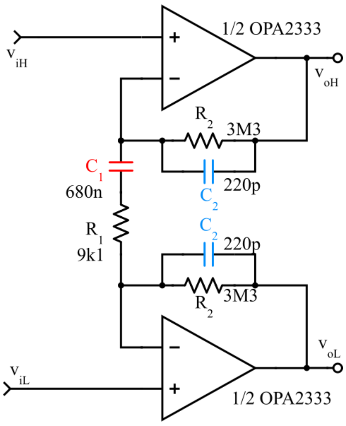

Fig. 3: Fully differential amplifier. The inputs v_{iH} and v_{iL} are connected to the electrodes through protection 100kΩ resistors (not shown in this figure). The outputs v_{OH} and v_{OL} are connected to the MCU analog inputs.

B. Common-Mode reference circuit

A third electrode in Fig. 2 is used to establish the common mode voltage of the user to a DC reference voltage, which for single supply applications is usually at the midpoint of power supply. The reference electrode is driven by a negative feedback circuit [10]. The Fig. 4 depicts this circuit, that estimates the input common mode voltage v_{CM} averaging both amplifier outputs v_{OH} and v_{OL}, compares it with the desired DC common mode VREF, and amplifies the error signal which drives the reference electrode. The chosen OA for this circuit was the single OPA333 from Texas Instruments.

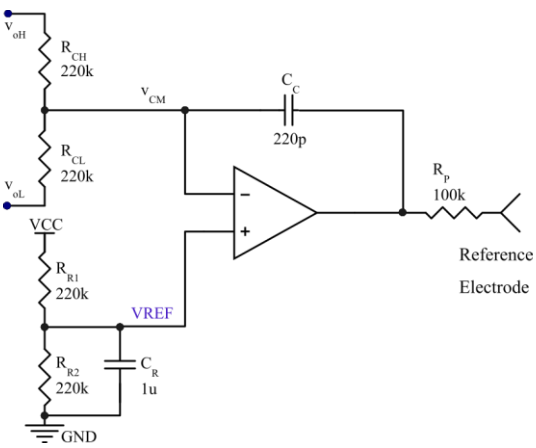

Fig. 4: Common mode feedback circuit.

C. Printed Circuit Board

The two parts were implemented in a divisible board. A built prototype is shown in Fig. 5. The electrodes are attached to the board conforming a whole sensor, avoiding complications associated with electrode cables as the coupling of electromagnetic interference [11]. The PCB is based on a single layer design of 42 mm by 34 mm, with an inter-electrode distance of 20 mm. All electronic components are of surface mount type and the electrodes are metallic pieces of nickel-plated stainless steel. A four conductor ribbon cable links the sensor with the microcontroller. The implemented PCB includes protection resistors of 100 kΩ in series with measurement electrodes and power supply decoupling capacitors for OAs, that are not shown on schematics of Figs. 3 and 4.

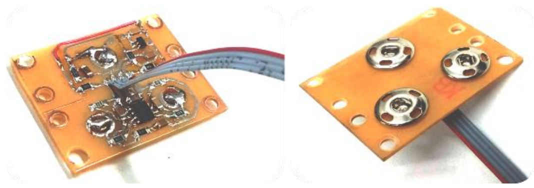

Fig. 5. Designed PCB.

The PCB is encapsulated by a 3D-printed PLA housing and the side in contact with the user's skin is covered by a medical grade silicone rubber (shown in Fig. 6).

Fig. 6. Encapsulated sensor. Left: a sensor with its 3D-printed housing. Right: sensor coated by medical grade silicone rubber.

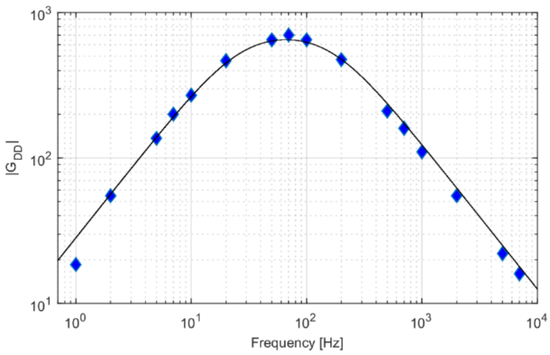

Fig. 7. Frequency response of differential gain GDD. Black continuous line is the expected frequency response and blue marks are the measured response.

III. Results

A. Frequency response of G_{DD} and CMRR

The frequency response of the differential gain G_{DD} and also of the common mode rejection ratio (CMRR) were assessed on a test-bench, exciting inputs with a signal generator and measuring outputs with an oscilloscope that allows to compute the differential signal between two single ended channels. The frequency response of G_{DD} is depicted on Fig. 7, where can be seen how the observed frequency response resembles the expected one. The CMRR was higher than 95 dB in the signal bandwidth.

B. Noise and Myographic signal measurement

The acquisition and register of signals were done with a battery powered laptop and a MSP432P401R LaunchPad, connected by USB to the computer. The built prototype of sensor was connected to a differential ADC input of the microcontroller, which ran a program that samples the signal at 1000 samples per second and sends it by UART/USB to the laptop.

To measure the electronic noise level of the amplifier the three electrodes of the sensor were shorted. The effective input-referred noise computed over a 10 seconds register was 1.35 μV_{RMS}.

On Fig. 8, a 10 seconds of EMG signal is depicted; this signal was measured over the forearm of one of the authors which performed two one-second contractions with different strength.

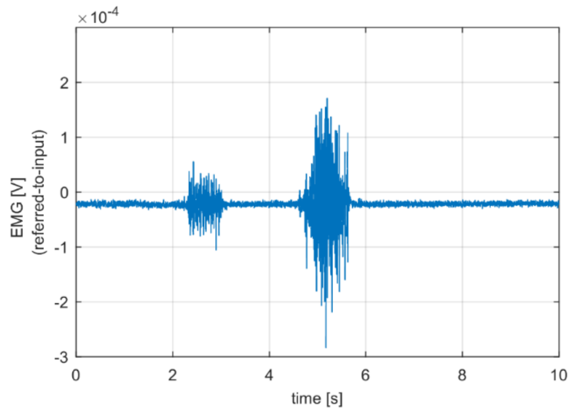

Fig. 8. A 10 seconds record of EMG signal over the forearm. There were shown two different contractions.

C. Current consumption

The current consumption measured with the three electrodes shorted was of 40μA.

IV. Discussion

The implemented sensor reached performance parameters comparable to other commercial products (see table 1) with a prototype cost below u$s 3.50 (board, electronic components, electrodes, PLA, and medical grade silicone rubber). That is more than ten times cheaper than the MyoWare sensor and hundred times cheaper than the SX230 from Biometrix.

But, the main achievement of this development was the increase of technological independence of the prosthetic arm development project with respect to closed and expensive commercial solutions.

V. Conclusions

A low power, low cost and wearable EMG sensor was designed, implemented and tested. It achieved CMRR and referred-to-input noise levels comparable to commercial devices.

The fully differential topology takes advantage of the input dynamic range of the ADC embedded on the MSP432P401R, with a minimum quantity of commercial off-the-shelf components. Besides the differential amplification of 726 V/V, this amplifier performs the band-pass filter which reject motion artifacts at low frequencies and thermal noise at high frequencies. A key advantage of this circuit is its high CMRR which does not depend of component tolerances and mismatching.

The built prototype was tested, achieving the expected differential gain frequency response and performing the acquisition of EMG signals with noise levels on the order of the μV.

TABLE I
Comparation between our sensor and commercial alternatives

Parameter	This sensor	Bagnoli DE-2.1. (Delsys)	SX230 (Biometrix Ltd.)	MyoWare (Sparkfun)
Gain	726	10	100/1000	Adjustable
CMRR	> 95 dB	92 dB	96 dB	110 dB
RMS noise	1.35 μV	1.2 μV	<5 μV	N.S.
Power consumption	0.12 mW	20 mW	N.S.	30 mW
Cost	< U$S 3,50	N.S.	U$S 325	U$S 38

N.S.: Not Specified.

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