First steps towards an implantable electromyography (EMG) sensor powered and controlled by galvanic coupling

Laura Becerra-Fajardo[1[0000-0002-5414-8380]] and Antoni Ivorra[1,2[0000-0001-7718-8767]]

[1] Department of Information and Communication Technologies, Universitat Pompeu Fabra, Barcelona 08018, Spain
[2] Serra Húnter Fellow, Universitat Pompeu Fabra, Barcelona 08018, Spain
laura.becerra@upf.edu

Abstract. In the past it has been proposed to use implanted electromyography (EMG) sensors for myoelectric control. In contrast to surface systems, these implanted sensors provide signals with low cross-talk. To achieve this, miniature implantable devices that acquire and transmit real-time EMG signals are necessary. We have recently *in vivo* demonstrated electronic implants for electrical stimulation which can be safely powered and independently addressed by means of galvanic coupling. Since these implants lack bulky components as coils and batteries, we anticipate it will be possible to accomplish very thin implants to be massively deployed in tissues. We have also shown that these devices can have bidirectional communication. The aim of this work is to demonstrate a circuit architecture for embedding EMG sensing capabilities in our galvanically powered implants. The circuit was simulated using intramuscular EMG signals obtained from an analytical infinite volume conductor model that used a similar implant configuration. The simulations showed that the proposed analog front-end is compatible with the galvanic powering scheme and does not affect the implant's ability to perform electrical stimulation. The system has a bandwidth of 958 Hz, an amplification gain of 45 dB, and an output-referred noise of 160 μV_{rms}. The proposed embedded EMG sensing capabilities will boost the use of these galvanically powered implants for diagnosis, and closed-loop control.

Keywords: galvanic coupling, microsensors, microstimulator.

1 Introduction

Apart from diagnosis, electromyography (EMG) has been extensively used as a source of control for exoskeletons and prostheses [1], [2], and has been proposed for closed-loop control in electrical stimulation. Implanted EMG sensors are an alternative to superficial EMG as they provide signals with lower cross-talk for myoelectric control, potentially helping to control devices with multiple degrees of freedom, which is not possible nowadays with superficial EMG [3]. These sensors have been tried as implantable central units wired to electrodes [4], which present surgical com-

plexity; and as wireless implant capsules (e.g. IMES) [2]. Even though these capsules have accomplished high miniaturization levels, they are still too stiff and bulky to be massively deployed in tissues, hampering the development of a network of microdevices with high selectivity for stimulation and sensing. This can be explained as they require voluminous components as coils and batteries for power transfer/generation to produce the current magnitudes required for neuromuscular stimulation. In the case of inductive coupling, the implant's diameter is limited by the coil used; while in existing battery technologies, the implant's size is limited by their energy density.

In [5] we proposed a heterodox method to create ultrathin implants that lack coils and batteries. The implants act as rectifiers of innocuous high frequency (HF) current bursts (≥ 1 MHz) supplied to the tissues by galvanic coupling using superficial electrodes (Fig. 1). We *in vivo* demonstrated microcontrolled injectable stimulators that could be galvanically powered and that could deliver low frequency (LF) currents capable of stimulating excitable tissues [6]. These implants (diameter = 2 mm), made of commercially available components, are the first step towards future ultrathin and flexible implants (diameter < 1 mm) based on an application-specific integrated circuit (ASIC). In [7] we *in vitro* demonstrated that bidirectional communication is feasible in this method, allowing closed-loop control. The aim of this work is to demonstrate a signal conditioning electronics architecture (i.e. an analog front-end) for embedding EMG sensing capabilities in our galvanically powered implants. This will boost their use in diagnosis, closed-loop control in neuroprostheses, and man-machine interfaces as those used for prostheses control.

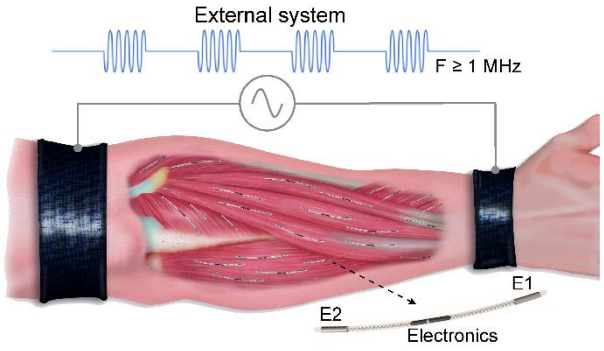

Fig. 1. Basic scheme of the method to galvanically power implants. An external system delivers high frequency current (≥ 1 MHz) bursts, which flow through the tissues by galvanic coupling. The implants are envisioned as elongated, flexible and ultrathin devices with two electrodes at opposite ends (E1 and E2) to pick up the high frequency current and rectify it for power and for electrical stimulation.

2 Methods

2.1 Proposed electronic architecture

The core architecture of the microcontrolled injectable stimulators was described in [6]. Briefly, implant electrodes E1 and E2 pick up a portion the HF current delivered to the tissues by the external system (Fig. 2). This HF current is full-wave rectified by a bridge rectifier. A regulator subcircuit stabilizes the rectified voltage to power the control unit (CU) and the rest of the electronics, and it is followed by a capacitor that powers the electronics during sleep mode (i.e. when no HF current is delivered by the external system). Current consumption is limited in this mode to keep the sleep capacitor's size small. The circuit includes a demodulator that is capable of extracting information amplitude modulated in the same HF current used for galvanic powering. This information is used to independently address each implant. The CU drives two current limiters that deliver LF current pulses for electrical stimulation. These current limiters are also used to modulate the HF current to send information from the implantable circuit to the external system [7].

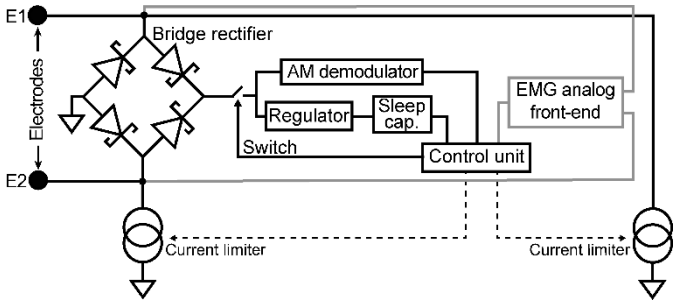

Fig. 2. Core architecture of the implantable device based on commercially available components. The analog front-end for EMG acquisition proposed here is shown in gray. The sleep capacitor powers the electronics, including the analog front-end, when no HF current is delivered by the external system.

In here we propose to add an analog front-end for EMG acquisition to this core architecture (Fig. 3A). The implant senses the EMG signals as the difference between the voltage obtained from the two implant electrodes E1 and E2 when no HF current is delivered by the external system. The first stage consists on analog filters: 1) a first-order low-pass filter (F_c = 342 kHz) used to protect the front-end's amplifier from the HF current used for powering and 2) a first-order high-pass filter to prevent possible dc components seen across the implant electrodes (F_c = 73 Hz). An operational amplifier (OPAMP) configured as a differential amplifier is used to amplify the voltage difference and suppress voltages that are common to both E1 and E2. A single-supply, low power OPAMP is used for this purpose as the implant's regulator delivers a single-supply voltage (VCC), and the front-end powers from the sleep capacitor. Additionally, the OPAMP must ensure low noise, very low input bias current, and very small package. The ADA4691-2 (by Analog Devices Inc.) is a 1.21 × 1.22 × 0.4 mm

dual OPAMP that provides all these features and has a shutdown mechanism for further power reduction. A voltage divider is used to set a steady state of VCC/2 V at the output of the differential amplifier when no EMG signal is picked-up by the implant electrodes. The second OPAMP of the package is used to further amplify the obtained voltage, and it is followed by a low-pass filter (F_c = 1.25 kHz) to avoid aliasing during digitalization.

2.2 Computer simulations

Computer simulations were performed in a SPICE simulator (LTspice XVII by Linear Technologies) using the setup shown in Fig. 3B. A Thévenin equivalent is used to model the coupling between the tissue and implant electrodes E1 and E2, which have a diameter D and a separation distance L. If the implant is aligned with the electric field ($\mathrm{E}(t)$), then $\mathrm{V_{Th}} = \mathrm{LE}(t)$. The equivalent resistance ($\mathrm{R_{Th}}$) is the resistance of the dipole formed by the two electrodes. If $\mathrm{L} \gg \mathrm{D}$, $\mathrm{R_{Th}} = 1/\pi\sigma\mathrm{D}$, where σ is the electrical conductivity of the tissue (S/m). The impedances of the implant electrodes are neglected for simplicity. For this study, we supposed the same length and diameter of the implant we demonstrated in [6] (L = 5 cm; D = 2 mm), an electric field of 240 V/m, and an electrical conductivity of 0.5 S/m which approximately corresponds to the admittivity magnitude of skeletal muscle at 1 MHz [8]. A voltage source connected in series with the Thévenin equivalent is used to simulate the EMG signals that appear across the implant electrodes. They are based on a 500 ms sample of a digitized intramuscular EMG signal obtained from an analytical infinite volume conductor model that used an implant configuration with two electrodes at opposite ends [3].

Transient, ac and noise analyses were used to evaluate the behavior of the proposed circuit, including if the front-end affected the ability of the implant to perform electrical stimulation, and to assess its gain and bandwidth.

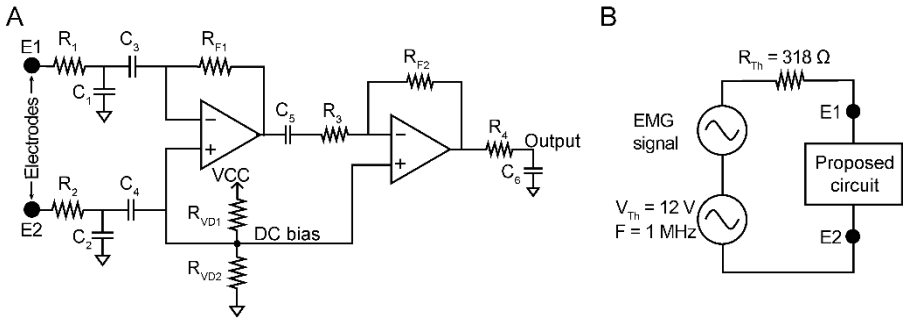

Fig. 3. A) Analog front-end scheme added to the core architecture of the implants (Fig. 2). The circuit includes a filtering stage, differential amplification, and antialiasing filter. B) Simulation setup used to evaluate the front-end proposed here. A Thévenin equivalent is used to model the coupling between muscle tissue and the implant electrodes E1 and E2. EMG signals are applied through the EMG voltage source, and a sinusoidal signal of 1 MHz is used as the HF current delivered for galvanic coupling (V_{Th}).

3 Results

The simulations show that the proposed front-end does not unbalance the load of the circuit on any electrode, and the circuit is able to deliver symmetrical pulses of 2 mA, as those obtained in previous demonstrations [6].

Fig. 4A shows the intramuscular EMG signal applied (top) and the output of the proposed front-end (bottom). The picked-up EMG signal was effectively amplified and biased over the 1.65 V established in the differential amplifier (VCC = 3.3 V). To analyze the delay between the picked-up EMG signal and the output of the analog front-end, the signals were normalized and resampled and a cross correlation was applied between them. The results show that the lag between the output and the original EMG signal was 58 µs.

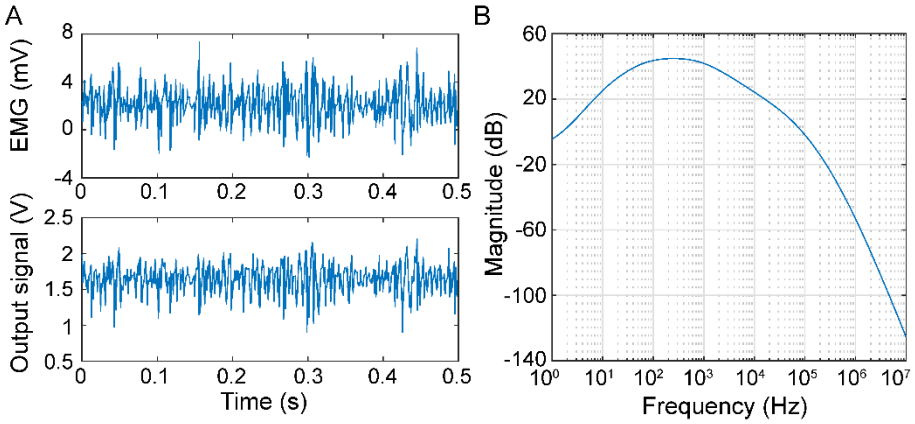

Fig. 4. A) Sample of simulated intramuscular EMG signal detected by a myoelectric sensor [3], and corresponding output signal of the analog front-end proposed. The lag between signals is 58 µs. B) Frequency response of the analog front-end proposed.

Fig. 4B shows the results obtained from the ac analysis. The acquisition system has a bandwidth of 958 Hz. The maximum gain of the EMG acquisition system is 45 dB. The output-referred noise of the analog front-end is 160 μV_{rms} (from 1 Hz to 10 MHz). Having in mind that the output signal without dc bias is 0.21 V_{rms}, the calculated signal-to-noise ratio (SNR) for the signal conditioning circuit is 62.4 dB.

4 Conclusion

In here we have proposed an analog front-end architecture for an existing microcontrolled injectable stimulator based on commercially available components and that is galvanically powered using HF currents. According to simulations, the proposed signal conditioning circuit is compatible with the galvanic powering scheme proposed and demonstrated in the past. The simulations also show that the circuit filters and amplifies the expected intramuscular EMG signals that are going to be picked up by

the implant electrodes, as those obtained from an analytical model that used a similar implant configuration [3]. Additionally, the gain, bandwidth and noise parameters obtained with the simulations demonstrate that the proposed circuit has a similar behavior to that previously reported by implantable wireless EMG capsules [2]. In contrast to these capsules based on inductive coupling, our future galvanically powered implants based on ASICs will be potentially thinner and more flexible.

The proposed embedded EMG sensing capabilities will boost the use of these galvanically powered implants for diagnosis, closed-loop control in neuroprostheses, and man-machine interfaces as those used for prostheses control.

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6 Conflict of interest

The authors declare no conflict of interest.

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