Surface EMG Multichannel Measurements Using Active, Dry Branched Electrodes

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Abstract— **A surface electromiography (sEMG) multichannel aquisition system was implemented using dry electrodes. Branched electrodes were used to obtain spatially selective EMG measurements. Active electrodes coupled with a high dynamic range converter were used to achieve good electromagnetic interference rejection and measurement robustness. Functioning parameters were measured. The device has an equivalent input referred noise less than 1 μV_{rms} with a voltage gain of 4 , allowing for a $\pm$550 mV electrode offset. The device bandwidth ranges from DC to 500 Hz with a 2000 sps sampling rate. The CMRR is 90 dB further improved by an independent DRL circuit with gain above 30 dB at 50 Hz. Multichannel EMG measurements were successfully conducted and example results are shown.**

Keywords— **surface electromiography; branched electrodes; multichannel measurements; sigma-delta converters.**

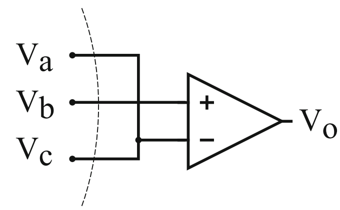

Fig. 1: Branched electrode configuration.

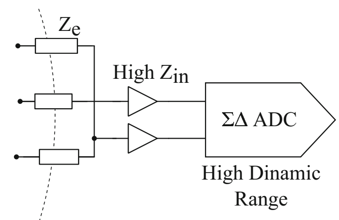

Fig. 2: Proposed implementation to manage the large impedance Z_e, voltage offset, and baseline drift of dry electrodes: high input impedance (Z_{in}) active electrodes mitigate EMI effects and a high dynamic range A/D converter allows for low gain stages.

I INTRODUCTION

Surface electromyography (sEMG) measurements are a non-invasive means to obtain information about muscle activity, delivering signals that can be of use for medical diagnosis, prosthetics and rehabilitation devices. Improving the patient comfort and simplifying the use of the measurement system are desirable goals because they encourage a more frequent use of the device, and reduce the training required of the user. For this purpose, dry electrodes are an attractive solution compared with traditional wet-gel electrodes. They do not require any electrolyte other than the natural moist of the skin so they can be easily put into place, and they have no disposable parts contributing to their ease of use and simple maintenance.

The most common electrode configuration for sEMG acquisition is a simple differential pair. Many other spatial configurations have been proposed with the aim of crosstalk reduction; the double-differential variety is found in many devices. In this work, branched electrodes (as defined in [1]) were used because they achieve a good selectivity comparable with that of double-differential electrodes [2], but with a simpler implementation. The branched electrode basic design can be seen in Fig. 1. It is recommended in [3] to build the electrodes with rectangular 10 mm x 1 mm rods with a separation between 20 and 10 mm. A 10 mm inter-electrode distance is found to produce good crosstalk reduction in [4].

As with any electrode, the interface between the metal contacts and the skin produces an electronic connection with non-zero impedance. In the case of dry electrodes placed on the skin without preparation, this impedance is of relatively high magnitude and variability, making them susceptible to electromagnetic interference (EMI). This problem can be overcome through electronic design strategies. Interference mechanisms are well explained in [5]: common mode signals can interfere through mode transformations caused by electrode impedances (through the "potential divider effect") and component mismatch. The baseline of signals obtained with dry electrodes takes more time to stabilize. In this work this is overcome by the use of active electrodes [6]. Large DC offsets and high amplitude artifacts can be tolerated if the amplifier input has sufficient dynamic range (DR). A DC coupled approach using Sigma Delta converters can reach a 120 dB DR, enough for EMG measurements accepting a $\pm 5V$ excursion of the baseline signal. The diagram in figure 2 summarizes the proposed approach.

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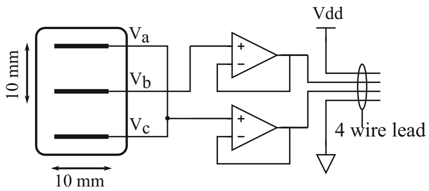

Fig. 3: Active electrodes layout.

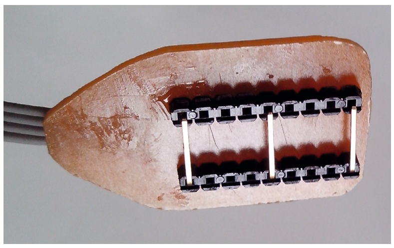

Fig. 4: The implemented active electrode. Three parallel gold-plated rod electrodes with a 10 mm inter-electrode distance were fixed to a PCB board that also holds the operational amplifiers. They provide a buffered differential output. Additional power wires are needed.

II MATERIALS AND METHODS

A System implementation

The electrodes were manufactured with gold-plated standard pin connectors, as they have the exact dimensions required and are robust, with a durable surface, and commonly available. Three of these connectors were soldered parallel to each other on a printed circuit board (PCB) to implement each branched electrode. Caution was taken to make sure that the solder would not come into contact with the skin, because it was observed in previous experiences that the “solder-skin interface” was highly noisy, hampering measurements. The electrode layout can be seen in figure 3. To accomplish the requirement of high input impedance and maintain a low noise floor, operational amplifiers (op-ams) with unity gain feedback configuration were used as buffers on the electrode itself. This has the additional advantage of driving to ground interference currents capacitively coupled to the electrode wires running from the main circuit [7], which would otherwise result in significant interference.

Thus, each electrode presents a fully differential channel that extends to the converter. Bacause mismatch between components on each rail of the channel degrade the CMRR, it is impractical to strive for a ratio above 90 dB when using discrete off-the-shelve components. High quality measurements however require a CMRR greater than 100 dB [8], so a DRL circuit was used to boost the CMRR. Traditional DRL circuits average the input from every measurement channel to calculate the common mode signal

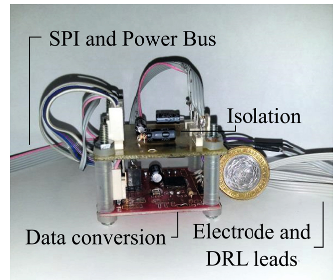

Fig. 5: Data conversion and isolation module, composed of two 5cm x 5cm boards.

to be canceled. Since dry electrodes can be accidentally detached more easily than the wet-gel kind, this is impractical as failure of one electrode would compromise the entire set of measurements. Hence, an independent DRL circuit was used, with its own measurement and feedback channels. Stability was ensured by dominant pole compensation, yielding a gain, and therefore a CMRR improvement, of nearly 30 dB at 50 Hz.

The acquisition system used in this work was based on the design presented in [9]. The modules of that device were adapted for improved portability and modified to hold a ADS1299 analog front-end from Texas Instruments. This front-end provides adjustable gain and 24-bit Sigma-Delta converters for each of its 8 channels. Two 5×5 cm circuit boards were implemented for the front-end and the isolation circuitry respectively, as can be seen in figure 5. A voltage reference is also available for the DRL. Digitized data is transmitted through a SPI bus. Isolation was implemented with two ADUM6401 ICs that provide both an isolated SPI data bus and an isolated power bus. One of them provides the 3.3V digital supply that the ADS1299 requires and the other one its 5V analog supply. A microcontroller receives the data as host of the SPI bus and relays it to a PC through a USB real-time connection to the PC. A C# application was created with a simple waveform visualization screen and a configuration panel to set the channel gain and sampling frequency, together with a logging system that creates a document with details of the acquisition sessions. Real-time visualization was included in order to have feedback over the tests (e.g., check for electrode detachment, biofeedback of muscle activation). Recorded signals were processed with Matlab.

B Measurement methods

The system was characterized by measuring parameters of interest: noise, bandwidth, and CMRR.

Noise was measured short-circuiting the branched

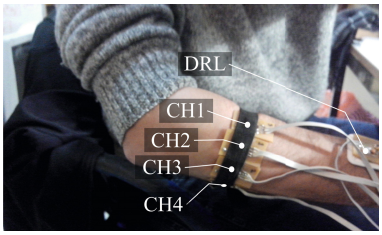

Fig. 6: Electrode array placement on the rigth forearm.

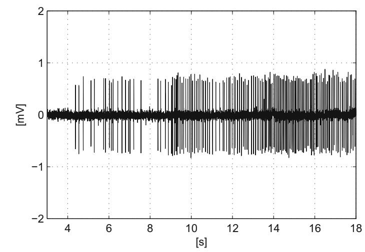

Fig. 7: EMG activity caused by index finger flexion with increasing force.

electrodes input to the low noise voltage reference used for the DRL. Records of every channel were taken with a $100s$ duration and processed off-line applying a bandpass filter between 5 and 450 Hz. This was done with different channel gains configured for the ADS1299 programmable amplifiers.

The device's frequency response was measured connecting a function generator's ground and the system's reference to one input of the branched electrode, and the generator's signal line to the other. The output amplitude was estimated off-line from the digital recordings.

The CMRR was calculated for different conditions. First a 50 Hz signal was applied directly to the short-circuited branched electrode inputs. Then, different unbalanced impedances were applied between the generator and the inputs, including that recommended in [10]: 51 $k\Omega$ in parallel with 47 nF.

EMG measurements were conducted to test the performance of the system. An elastic band was used to secure the electrodes in place. First, a single branched electrode was placed on the right forearm in the position of channel 3 of Fig. 6, and the index finger was flexed with progressively growing strength (setup *a*). Then, an array of four electrodes was placed on the forearm, as shown in Fig. 6 (setup *b*). Different hand and wrist movements were executed.

III RESULTS

The experimental setups described in section II B allowed to check the functioning parameters of the implemented system as a preliminary validation.

The equivalent voltage noise referred to the input (RTI) resulted in less than 3 μV_{rms} with a gain of 1. This noise is dominated by the RTI noise of the ADC. Using higher gains yields lower noise values, with the lower attainable bound being the equivalent noise sources of the operational amplifiers used on the active electrodes. A gain of 4 times and above yields a voltage noise lower than 1 μV_{rms}.

The device's frequency response was found to match very closely the $\Sigma-\Delta$ converter digital filter, which has a sinc^3 shape. resulting in a $3dB$ bandwidth from DC to 512 Hz when for a sampling rate of $2000sps$.

The CMRR resulted to be of 90 dB with no impedance imbalance. When 51 $k\Omega$//47 nF was applied, the CMRR was of only 46 dB. 90 $k\Omega$ and 9 $k\Omega$ imbalances were tested yielding CMRRs of 56 and 79 dB respectively.

A sample of the EMG measurements for setup *a* is shown in figure 7. An action potential train with increasing fire rate as force increased can be seen. Results for setup *b* can be seen in figure 8. All hand fingers were extended and flexed, and the resulting EMG signals are shown in the first and second columns respectively. The third column shows the signal resulting from an ulnar flexion of the wrist. The labels of each row correspond to the labels of each electrode shown in Fig. 6.

Electrodes were easily put into place, although the ribbon cables used were too stiff and caused movement artifacts. It was observed that a longer time was required for the baseline of the signal to stabilize compared with the case when wet-gel electrodes were used, and during that time spurious spikes of high amplitude were produced. After this initial period, the system remained stable during hour-long measurement sessions. Measurements without the DRL were not possible due to EMI, but when using the DRL interference was not a problem in any of the measurement setups.

IV DISCUSSION

The implemented dry electrodes successfully measured the EMG signals. They showed higher low frequency noise and more tendency to produce artifacts than wet-gel electrodes, so the DC-coupled, high dynamic range design was well suited for them. The system parameters were appropriate for EMG measurement. Noise of 1 μV_{rms} with a gain of 4 provides enough resolution while allowing a DC offset of ±550 mV. Bandwidth is in accordance with preconditioning filters typically used in EMG amplifiers [1].

Because the DRL circuit contributes 30 dB to the common

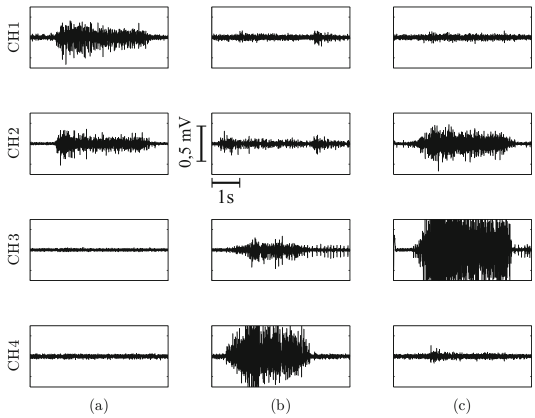

Fig. 8: Signals from rigth forearm muscles elicited from: (a) the extension of all hand fingers, (b) finger flexion, (c) wrist ulnar flexion.

mode signal attenuation, a CMRR figure of 70 dB is enough to reach the 100 dB target. Highly mismatched electrode impedances would result in rejection values below this limit, because of the potential divider effect. The TLC2202 input impedance is not listed on its datasheet, but because of the observed CMRR degradation it can be estimated to be dominated by an input capacitance of roughly 30 pF. Because the contacts of one electrode have very similar geometry and are close to each other, the impedance imbalance in the obtained measurements was within acceptable limits. However, op-amps with lower input capacitance would be recommendable for future designs, or a higher DRL gain.

The wired link maintained with the PC provided power to the device, and together with the use of dry electrodes resulted in a zero-maintenance device. However, it is only suitable for measurement situations where the patient is static relative to the PC and only limited-range movements are carried out.

V CONCLUSION

A multichannel sEMG acquisition system with dry branched electrodes was implemented and tested. It allowed to measure EMG activity reliably with a simple electrode setup, with functioning parameters in agreement with those reported in the literature [1] [11] .

Active dry electrodes and a high dynamic range A/D converter conformed a robust measurement system allowing high quality measurements disregarding the poor contact impedance and baseline drift characteristics of dry electrodes. EMI was successfully rejected with the complementary use of an independent DRL circuit.

The prototype boards had a small form factor, being suitable for compact enclosing. Integrated application-specific standard products (ASSPs), such as the ADS1299 front-end and ADUM6401 isolation chip used in this work are useful building blocks that simplify the design and implementation of flexible systems.

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