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Designing a Low-noise, High-resolution, and Portable Four Channel Acquisition System for Recording Surface Electromyographic Signal

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ABSTRACT

In current years, the application of biopotential signals has received a lot of attention in literature. One of these signals is an electromyogram (EMG) generated by active muscles. Surface EMG (sEMG) signal is recorded over the skin, as the representative of the muscle activity. Since its amplitude can be as low as 50 μV, it is sensitive to undesirable noise signals such as power-line interferences. This study aims at designing a battery-powered portable four-channel sEMG signal acquisition system. The performance of the proposed system was assessed in terms of the input voltage and current noise, noise distribution, synchronization and input noise level among different channels. The results indicated that the designed system had several inbuilt operational merits such as low referred to input noise (lower than 0.56 μV between 8 Hz and 1000 Hz), considerable elimination of power-line interference and satisfactory recorded signal quality in terms of signal-to-noise ratio. The muscle conduction velocity was also estimated using the proposed system on the brachial biceps muscle during isometric contraction. The estimated values were in then normal ranges. In addition, the system included a modular configuration to increase the number of recording channels up to 96.

Key words: *Portable acquisition systems, sigma-delta analog-to-digital converter, surface electromyography*

INTRODUCTION

Recently, the electrical signals obtained from the human body have been under special attention. One of these important signals is the electrical manifestation of the human muscles known as electromyogram (EMG) signal. EMG signal has different applications such as movement analysis in medical diagnosis, biofeedback systems, rehabilitation sports medicine, and ergonomics.[1-3]

Essentially, EMG signal could be recorded via needle electrodes intramuscular EMG or (iEMG) and surface electrodes (sEMG). iEMG signals have higher amplitude than sEMG; however, the patient is less comfortable. Alternatively, surface EMG is a noninvasive method. The amplitude of the sEMG signal is low and approximately ranges from several tens of microvolts up to few millivolts. The bandwidth of the sEMG signal is generally between 20 Hz and 500 Hz.[4-9] sEMG signal has a small amplitude which can be distorted by other signals or frequency interferences. Therefore, the sEMG signal must be recorded with a suitable system. sEMG signals could be recorded using two different methodologies.[5]

In both of recording systems, the first stage of the amplifier known as the front-end amplifier consists of an instrumentation amplifier. Its gain is as low as 10, preventing the saturation by the movement artifact. In the first methodology, a programmable gain amplifier (PGA) is usually used whose gain is usually manually set to improve the signal-to-noise ratio (SNR) of the recorded signal at low muscle contraction and reduce the gain in high muscle activities to prevent possible amplifier saturation. Waveshapers (filters) are also used to focus on the practical bandwidth of the sEMG signals. The recording structure is ended up with a low-resolution (12–16 bit) analog-to-digital converter (ADC).[4,5,10,11]

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In the second topology, the whole acquisition line is tuned on the high muscle contraction and a high-resolution (24-bit) ADC is used at the final stage.[4,5] Since the resolution of the ADC is high, it is like moving some of the amplification on the ADC instead of the analog circuits. Thus, while considering the possible saturation of the recording system at high muscle contraction levels, the resolution of the ADC is sufficiently high during low muscle activity, and no manual gain tuning is required. In the present design, the second approach was used for the sEMG signal acquisition system. The block diagram of the proposed method is shown in Figure 1. Briefly, the proposed acquisition system employs an alternating current (AC)-coupled instrumentation amplifier for each channel whose output is connected to a differential sigma-delta ADC (ADS1294IPAG) board. Each channel operates at 2 kSPS sampling frequency produced by a 2.048 MHz internal clock. Bluetooth version 2.0 (BLK-MD-BC04-B) transmits the obtained data to the personal computer (PC) with an interface via LabVIEW. To reduce the power-line interference while maximizing patient safety, a 3.7 V lithium ion battery was used [Figure 1].

MATERIALS AND METHODS

The present system can record sEMG signals by using two different methods: Single differential (SD) and monopolar (MP) configurations. The amplifiers are built as differential units, with two electronically symmetrical inputs. In the SD mode, recording system channels consist of two electrodes that are placed on the same muscle with a specified interelectrode distance (IED) and the difference between the two signals is what is actually detected. In MP mode, on the other hand, there is a single electrode that measures the signal with respect to the MP reference electrode.

The acquisition system includes four major parts: The preamplifier board, ADC board, controller board, and power supply board.

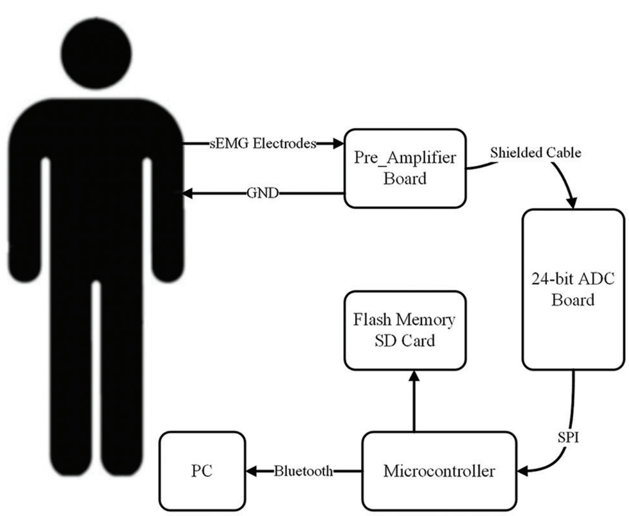

Figure 1: The block diagram of the designed surface electromyogram acquisition system

Preamplifier Board

The sEMG signal obtained from the electrodes is a low-amplitude signal.[5,12,13] As a consequence, it should be amplified carefully. The preamplifier board was designed taking into account the following features: (a) An instrumentation amplifier with a low fixed gain, (b) the first-order active high-pass filter with 8 Hz cut-off frequency, and (c) the first-order antialiasing RC low pass filter with 500 Hz cut-off frequency. The preamplifier board contains an AC-coupled instrumentation amplifier. The output of each AC-coupled instrumentation amplifier was connected to a differential ADC.

The instrumentation amplifier used for the first stage of the circuit must have the following characteristics:

Common mode rejection ratio

This parameter determines the capability of the instrumentation amplifier for eliminating input noises. SENIAM standard recommends the minimum common mode rejection ratio (CMRR) at 96 dB for appropriate attenuation of common mode noise.[14] Most of the current commercial instrumentation amplifiers have proper CMRR between 90 dB and 130 dB which can satisfy the required standards. One of the major noises in the biopotential signal recording is the power-line interference. However, previous studies stated that the required CMRR for attenuating this unwanted signal is from 100 dB to 120 dB.[2,5,15,16]

Referred to input noise

Every semiconductor circuit has an intrinsic noise in its input and output. The referred to input (RTI) noise determines noise level in the input of the instrumentation amplifier. Its power spectrum density is distributed uniformly on all frequencies (such as white noise) and could not be eliminated by filtering. In the recent prototypes, the RTI noise level is negligible respect to sEMG signal amplitude and according to the SENIAM standard, its amplitude should be $<1\ \mu V_{rms}$.[5]

Input impedance

The impedance of each electrode is inversely proportional to its surface area. Needle electrodes have a greater impedance in comparison with surface types. In the sEMG recording system, the input impedance of the instrumentation amplifier should be at least 100 times greater than electrode-skin interface impedance. The impedance of the electrode-skin interface in good condition is approximately 20 KΩ and in the worst case can be as high as 1 MΩ. As a result, input impedance for the appropriate instrumentation amplifier must be higher than 100 MΩ. In the SENIAM standard, it is recommended to use a 1 GΩ impedance for the input of the instrumentation amplifier.[5,13,17,18]

There are other parameters for an acceptable instrumentation amplifier which plays an important role in the sEMG acquisition system in terms of low input bias current, and low input offset voltage. In addition, portable systems should be compact and work with low possible power.

The gain of the instrumentation amplifier was set to 10 to avoid the first stage from saturation. Similar gain in different channels was obtained by using gain resistors with a 5% tolerance. The total RTI noise of the amplifier board was divided into three sections: (a) Output voltage noise of the instrumentation amplifier when its inputs are shorted to the system's ground and was divided by amplifier's gain (RTI), (b) current noise of the instrumentation amplifier that flows in the electrode-skin interface impedance (a resistor was used for simulating this impedance which is maximum 1 MΩ), and (c) electrode-skin interface voltage noise.

AD8295BCPZ (from the analog device) was chosen as the instrumentation amplifier at the first stage of the sEMG acquisition system.[5,15] It was chosen because of its dominant characteristics such as high CMRR (115 dB at 50 Hz), high input impedance (100 GΩ $\|$ 2 pF), and low input voltage noise (8 nV/$\sqrt{Hz}$ maximum at 1 KHz) among other available solutions (e.g., LT1167 from Linear Technology, INA2332 from Texas Instrument).

At the first stage, the gain was set considering two factors: (a) If gain of the first stage is high, total SNR and quality of the recorded signal will be high too, and as a result, the gain of the next stages can be low, (b) when biopotential signals in terms of sEMG is recorded with the surface electrodes, half-cell potential is present up to 300 mV. Setting the high gain at the first stage increases the chance of saturation. To eliminate this problem, two solutions were proposed: Increasing the power supply voltage of the instrumentation amplifier and using a high-pass filter. The first method has higher power dissipation which is not useful for the portable systems while the other solution is given either by placing the filter in front of the instrumentation amplifier or using an AC-coupled configuration.[4,5,16]

AC-coupled instrumentation amplifier can be designed using only one external capacitor with internal operational amplifier and resistor of the AD8295BCPZ. This configuration provides a single-pole high-pass filter. The time constant of the RC group was set to 125 ms, and the cut-off frequency was thus 8 Hz.

The next circuit of the analog board was a single-pole antialiasing low-pass RC filter for each differential output and designed as the following: (a) Any interference in the frequencies around harmonics of the modulator frequency of ADC are attenuated sufficiently and (b) the elimination of such signals beyond the sEMG signal bandwidth. The cut-off frequency of the filter was set to 0.5 kHz for providing at least 40 dB attenuation on the ADC modulator frequency.

The gain of the amplifier board was set as to prevent the amplifiers from saturation; the gain of 10. The amplifier circuit for one channel is shown in Figure 2.

Analog to Digital Converter Board

The ADC board consists of a four channel ADC with a resolution of 24 bit in 5 V full-scale ranges. For this purpose, several ADC devices from different manufacturers were examined in terms of (a) simultaneous sampling of input channels for avoiding timing error, (b) supporting daisy chain configuration for raising number of channels in the future works, (c) low power consumption, and (d) minimum bill of material. Accordingly, ADS1294 (from Texas Instruments) was chosen.

The sampling frequency of ADS1294 can be set by programming related registers. It is between 250 SPS and 32 kSPS and in the designed system was set to 2 kSPS. Multiple ADS1294 could be connected to each other in principal with the same clock. By increasing the number of the ADC boards to 12, the recording channels can be increased up to 96. Simultaneous sampling feature in the selected ADC can eliminate the need of other parts such as sample and holder. The ADC board can be assembled with an eight channel ADC from the same production family without additional changes in the circuit design or printed circuit board.

Each channel of ADS1294 has a PGA whose gain is set by writing appropriate values on the related registers. Onboard PGA in ADS1294 worked in differential mode for each channel. CMRR of each channel was 115 dB for DC to 60 Hz. In the present design, the gain was set to 12 for obtaining maximum attenuation on common mode noises. The protocol for the communication between ADC and controller board was a serial peripheral interface with its speed was set to 1 MHz.

The Controller Board

In the controller board, LPC1768 (from the NXP Semiconductors) was chosen to communicate with ADS1294

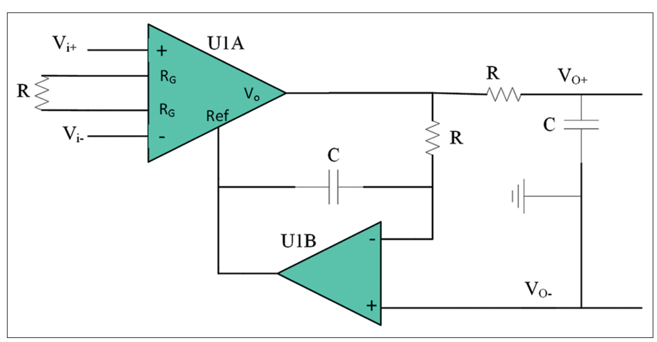

Figure 2: The amplifier board (instrumentation amplifier configured as alternating current-coupled amplifier, the differential output to connect to a differential analog-to-digital converter)

and other modules such as Bluetooth device. LPC1768 is an ARM Cortex-M3 based microcontroller for embedded applications which has special features in terms of the high level of integration and low power consumption. Recorded signals converted by ADS1294 were used for several purposes in terms of stored data in external memories such as secure disk card or other available memories for further processing and illustration of recording signal in liquid crystal display. In the present design, data were sent to a PC-based program via Bluetooth in the raw format.

Transferred data to the PC can be stored and displayed in the user interface without any additional process. The practical frequency range of the sEMG signal is from 20 Hz to 500 Hz, and its dominant energy is between 20 Hz and 150 Hz. According to the Nyquist theory, the sampling frequency must be at least to 1 kHz and was set to 2 kHz in ADC board, as the traditional sampling rate in sEMG signal processing. The data from the output of the ADC board was then sent to PC via universal serial bus (USB) or Bluetooth without any additional processing. For the elimination of the power-line interference and safety of the subject, Bluetooth was used as a communication protocol. Several available Bluetooth models were examined, and finally BC04 (from the Shenzhen Bolutek Technology) was chosen because of its low power consumption.

In the designed system, the recorded signals had an acceptable quality. In the user interface, a digital high-pass filter was utilized for the elimination of direct current voltage and low-frequency noises. For low cut-off frequency, SENIAM standard recommends 10 Hz to 20 Hz and International Society of Electrophysiology and Kinesiology journal endorses 10 Hz.[5-7] In the designed user interface, a rather frequency of 8 Hz was used. Additionally, to eliminate the power-line interference, an optional notch (or Comb) filter was utilized in the user interface.

Power Supply Board

A low-noise power supply voltage is indeed required for a good recording system. The power supply block diagram is shown in Figure 3. The preamplifier board was powered in dual supply mode with ±2.5 V + 2.5 V generated by using a low dropout (LDO) voltage regulator (LM1117-2.5 from Texas Instrument) and −2.5 V was obtained from two cascade regulators: (1) A high efficiency charge pump voltage inverter configured for generating unregulated −5 V in its own output (TPS60403 from Texas Instrument) and (2) A linear low-noise LDO negative voltage regulator with fixed −2.5 V as output (TPS72325 from Texas Instrument).

Power supplies of the ADC board were ±2.5 V for analog power and 3.3 V for digital power. The analog power supplies were obtained from the amplifier board, and digital power was taken from the controller board. The power supply of

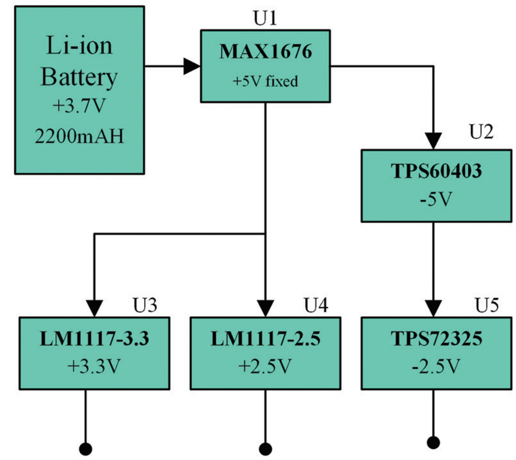

Figure 3: The power supply board (using only one lithium ion battery. Considering the battery discharge, MAX1676 was selected for the first regulator to maintain a +5 V in the output regardless of input voltage variations)

the controller board was 3.3 V and generated by a linear LDO regulator (LM1117-3.3 from Texas Instrument).

In general, the entire power supply of any signal acquisition system could be taken from: (1) External power supply unit, (2) computers USB ports, and (3) external rechargeable batteries. To reduce power-line interference, to maximize the safety of the subject and to enhance recorded signal quality, the external battery was used. For implementing this purpose, a 3.7 V lithium ion battery and a high efficiency, low supply current and compact boost switching regulator with fixed output voltage was used. MAX1676 (from MAXIM) efficiently provides +5 V. MAX1676 has an internal circuit and an output for low-battery detection which could be useful in the portable systems [Figure 3].

Validation

Performances evaluation of the recording system

Three major tests were performed to assess the performance of the proposed system: (1) The amplifier inputs were short circuited to measure the RTI voltage noise, (2) a 20 kΩ resistor was used in the amplifier inputs for analysis of the current and voltage noise, and (3) amplifier inputs were connected to a signal generator for analyzing amplifier frequency response and synchronization among different channels. When a signal is recorded in different channels, it must be the same in the computer, and the timing of the waveforms must not differ. This feature was named as synchronization.

The user interface was used for the data acquisition. The developed interface can record signals, store them and can further implement other processes such as filtering and feature extraction.

System gain and channel synchronization

A signal generator was connected to the inputs of the front-end amplifier to measure the overall gain of the recording system. The sampling frequency for all channels was 2000 Hz. A sinusoid signal with a frequency range from 10 Hz to 500 Hz was fed into the amplifiers simultaneously. A high-resolution alignment was used to check the subsampling timing error. The maximum timing difference between the signals in different channels was 1.4857 e–004 s. This value was less than a sampling period, and synchronization among channels was satisfactory.[19] Thus, the synchronization of the multichannel recording system was satisfactory down to 0.15 sampling interval.

Input noise

The designed system was placed in a shield case, and the entire amplifier inputs were connected to the system ground and the outputs were recorded in 10 s minimum in three separate tests. The power spectral density of the recorded signals is depicted in Figure 4. The input voltage noise of the proposed system with the short circuit inputs and a 20 kΩ resistor were shown in Tables 1 and 2, respectively. The former shows the effect of the pure input voltage noise while the latter takes into account the input current noise, as well. The Input voltage noise of different sEMG recording systems in literature was shown for comparison in Table 3.[12,20,21] The Input voltage noise of our system was 0.5595 μV in worst case compared with what obtained by Marateb *et al*., which was 0.811 μV.[4] Meanwhile, the distribution of the input voltage noise of the four channels was shown in Figure 5. The noise voltage is normally distributed in all channels. In addition, the input noise voltage in all channels was similar [Figures 4 and 5, Tables 1-3].

Experimental data

sEMG signals were recorded from a biceps brachii muscle of a healthy subject during isometric contractions at 20%, 50%, and 80% of maximum voluntary contraction (MVC) in SD mode. Signals were recorded in 90° flexion angle with at least 10 s duration. Figure 6 shows the signals of four recording channels during isometric contractions at 20% MVC. Figure 7 depicts the input noise of the system for the muscles at rest.

Table 1: The input voltage noise for the prototype system with the short circuit input (μV_{rms})

Test	Channel 1	Channel 2	Channel 3	Channel 4
Number 1	0.43141	0.43090	0.31938	0.33001
Number 2	0.43639	0.43291	0.32503	0.33331
Number 3	0.43696	0.42984	0.33219	0.32388

Table 2: The input voltage noise for the prototype system with a 20 KΩ resistor (μV_{rms})

Test	Channel 1	Channel 2	Channel 3	Channel 4
Number 1	0.54770	0.54872	0.33085	0.34794
Number 2	0.55982	0.54886	0.33205	0.35499
Number 3	0.55552	0.54928	0.34043	0.35247

Table 3: The input voltage noise for similar recording systems in literature (μV_{rms})

	Designed system	Previous works		
		Barone and Merletti[12]	Guerrero and Spinelli[20]	Lapatki *et al*.[21]
Minimum	0.31938	2.42187	1	1.92
Maximum	0.43696	4.56770	3	5.24

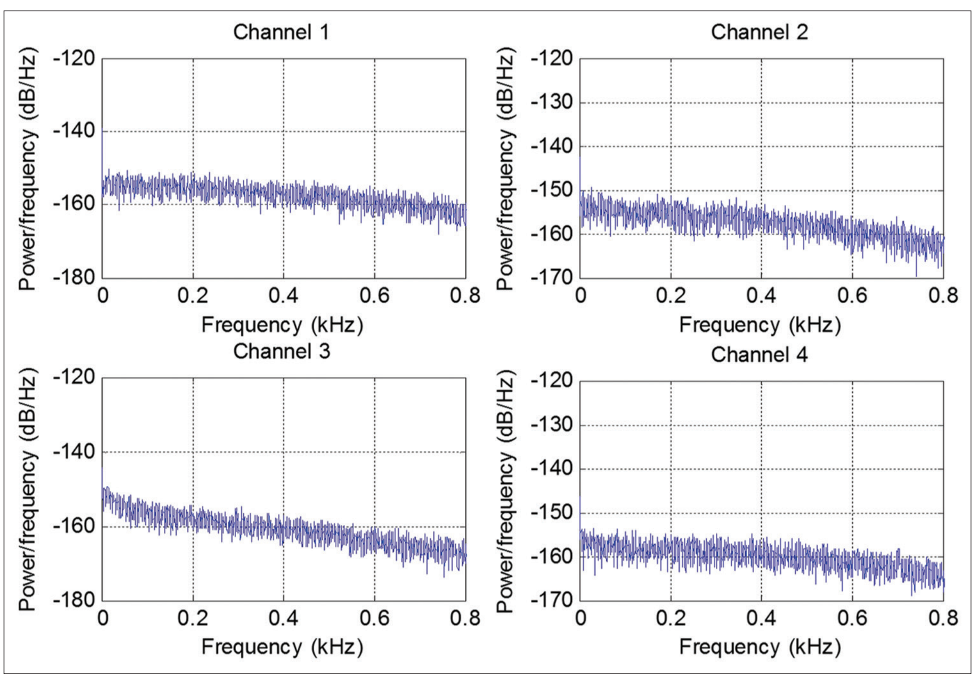

Figure 4: The power spectral density of input voltage noise with short circuit inputs

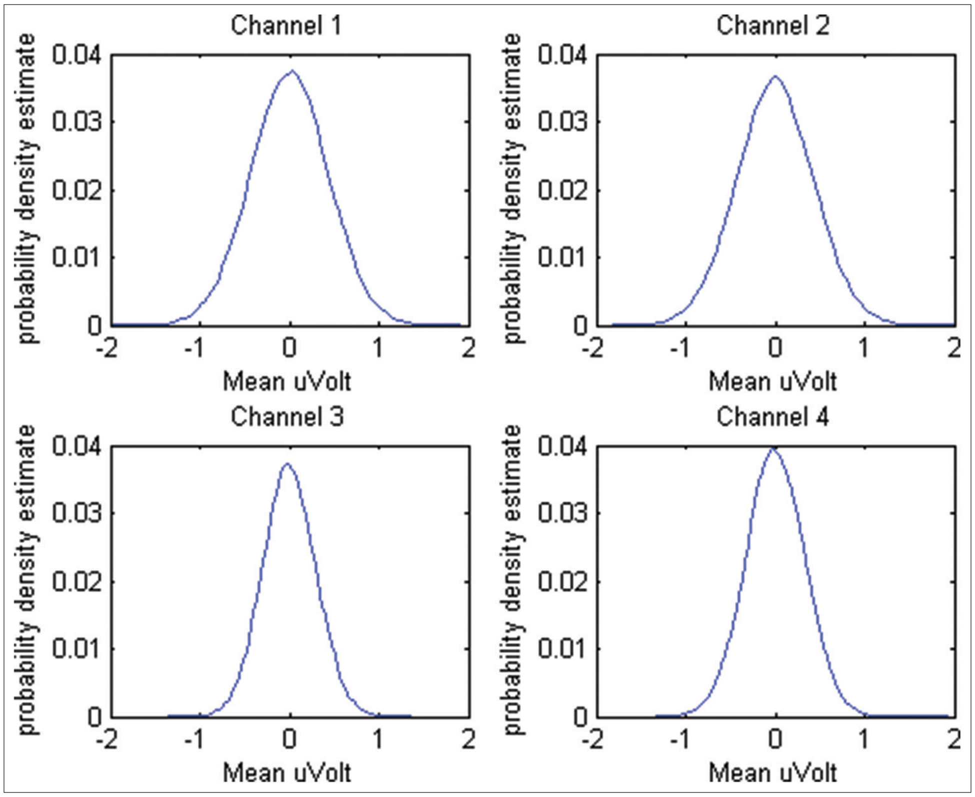

Figure 5: The distribution of the input voltage noise with short circuits inputs

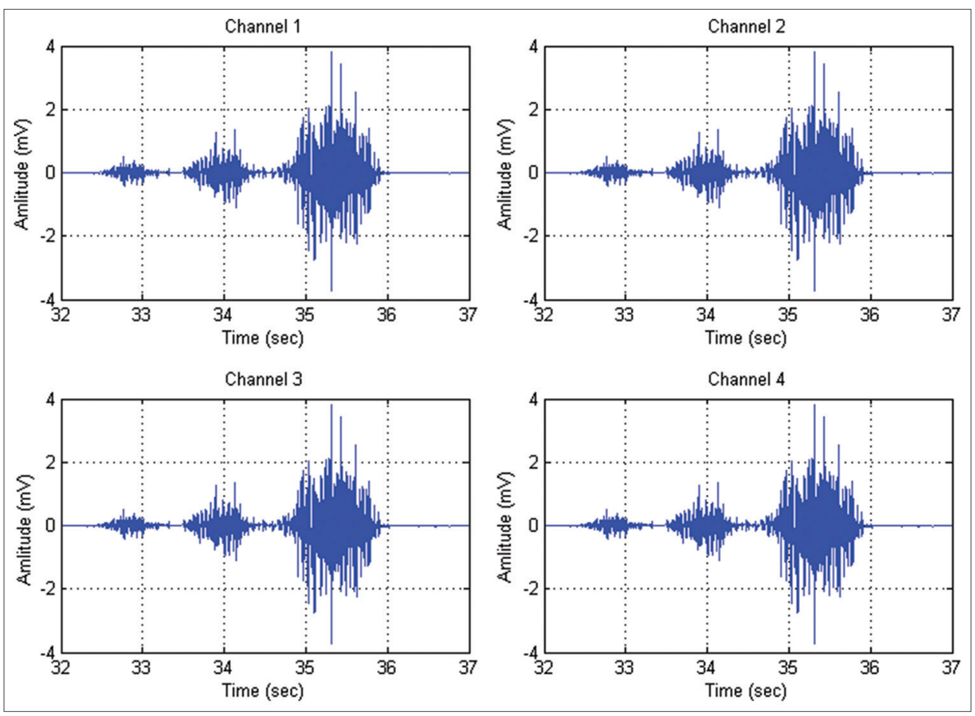

Figure 6: The real surface electromyogram signals of the four recording channels from the biceps brachii muscle of a healthy subject during 20 maximum voluntary contraction

The IED in the designed electrode array was 7 mm. The time delay between recording signals of the two channels was calculated using a high-resolution alignment algorithm.[19] To measure the conduction velocity (CV), IED was divided by the calculated time delay. For each contraction, 5 tests were performed and the average CV for the 20%, 50%, and 80% MVC was 5.41054 m/s, 5.21342 m/s, and 5.55196 m/s, respectively which is in

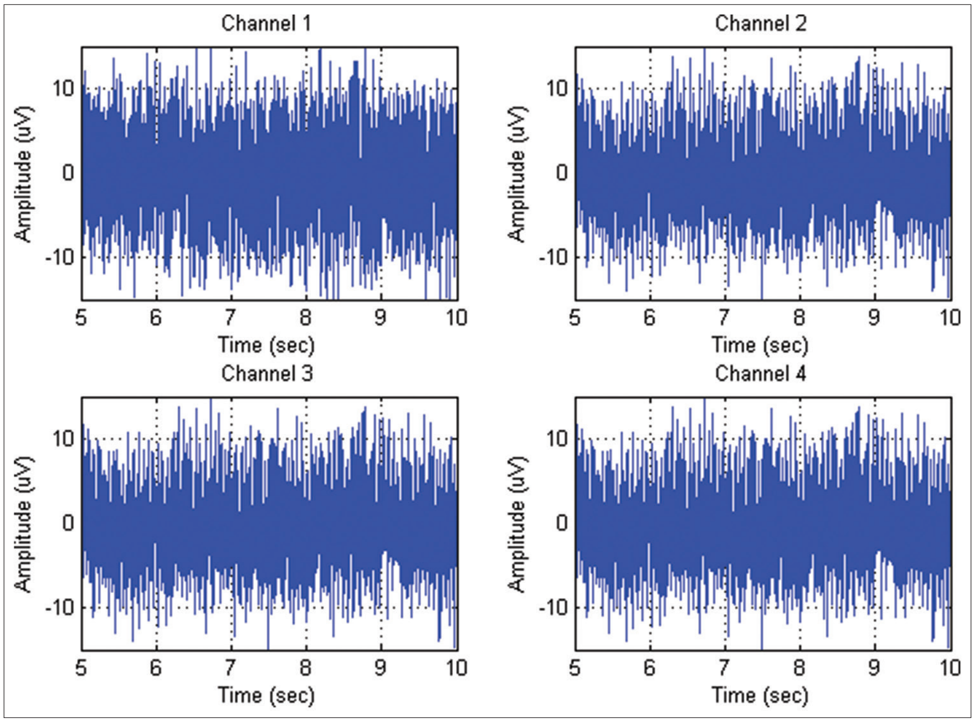

Figure 7: The input voltage noise of the acquisition system during rest

the normal range no muscle fatigue was seen at high MVC contractions [Figures 6 and 7].

DISCUSSIONS AND CONCLUSIONS

sEMG is one of the most important biopotential signals representing the muscles activity. This signal is used for different purposes such as movement analysis in medical diagnosis, biofeedback systems, rehabilitation sports medicine, and ergonomics. Due to low-amplitude and quasi-stationary feature of this signal, the recording system must be low-noise also in many applications; the system is required to be portable. In this paper, we presented a prototype of a portable sEMG recording system with a low voltage and current noise, low power, compact, and low-cost design. The required current for ADC and preamplifier boards was 33 mA which makes it suitable for portable acquisition systems. The total gain was 100 V/V and the low cut-off frequency was set to 8 Hz. The input voltage noise in the prototype was <0.5 μV in the bandwidth of 8 Hz–1000 Hz. The obtained RTI noise in the designed system was lower compared to that of previous systems in literature. The SNR of the recorded system was improved using low-noise components such as instrumentation amplifier, preamplifier close to the electrodes, and a shielded cable.

In the present design, sEMG acquisition system was implemented in four channels. Modular configuration helps to increase the number of recording channels up to 96. This could be done in principal by adding preamplifiers and ADC boards. In the future work, we will increase the number of the recording channels. This system was developed and tested without any additional circuits such as daytime running light circuit and the virtual ground.

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Conflicts of Interest

There are no conflicts of interest.

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BIOGRAPHIES

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