

Article

Development of a Prototype for the Acquisition of Biopotentials Implementing a New Interconnection Method for Shielding

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Abstract: A biological system can emit signals, and if these signals are correctly acquired, they can provide valuable information about the processes occurring within the system, enhancing our knowledge of the biological system. For this reason, we present a prototype system for acquiring various biopotentials using a main module that integrates amplification, high-pass filtering, band-reject filtering, and offset adjustment stages. This configuration allows for adjustable gain when working with different biopotentials and includes dedicated filtering modules for each biopotential type. We also propose a new topology for the shielded controller used in the interconnection between electrodes and the amplification stage to reduce noise introduced by the electrical network. Biopotentials acquired using the proposed topology show improved noise reduction and signal definition compared to those acquired using other topologies found in the literature. The design of the proposed system utilizes basic electronics, making it a low-cost solution. Ultimately, the system is simple, efficient, and suitable for applications requiring the acquisition of multiple types of biopotentials.



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1. Introduction

In recent years, the monitoring of physiological parameters of the human body has increased, where biopotential signals can directly reflect the physical condition and the health status of an individual [1]. Biopotential signals are increasingly being applied to real-time health monitoring, disease diagnosis, rehabilitation device control, brain–computer interfaces, neuroscience, cognitive psychology, and artificial intelligence research, among others [2,3]. In general, biopotential signals, such as electrocardiogram (ECG), surface electromyography (sEMG), electroencephalogram (EEG), and electrooculogram (EOG) signals, are characterized by high impedance, low frequency, low amplitude, and strong background noise [4]. When a biopotential signal is recorded, the target/result of the recordings is almost completely submerged by the background noise/interference inherent in the recordings, as the signal is usually characterized by a low amplitude [5,6]. Therefore, these noises/interferences mainly fall into the following three categories: first, power line interference, which is composed of a 50 Hz or 60 Hz frequency and its harmonics, commonly defined as electromagnetic noise produced by power supply circuits and electronic equipment [7]; second, motion artifacts, which are caused by poor contact between the electrodes and an individual's skin surface [8], or as a result of wire movement between

the electrodes and the amplifier [9]; and third, physiological artifacts in the form of other signal(s) recorded along with the originally desired biopotential signals [10,11]. Several methods have been proposed to remove the previously described interference/noise from a biopotential signal of interest. To this end, most methods have focused on improving hardware circuitry and using software filtering techniques [12–14]. Many hardware-based methods exist to minimize the interference inherent in biopotential signal recordings, such as twisting the input cables to reduce the area of the loop formed by the cables [15] or making the acquisition devices rely on batteries (direct current) for power [16]. The adoption of a high common-mode rejection ratio amplifier, a driven right-leg circuit, active electrodes, and isolation can further suppress interference [17–19].

Shielded technology is also one of the most widely used hardware methods for suppressing power line interference. The most popular shielded technology is the use of a small coaxial cable, in which the outer layer is connected to a certain type of conducted signal to prevent the signal of the inner cable from being contaminated by extraneous interference. Alnasser showed that unshielded electrodes are responsible for most of the power line interference, and that shielding the electrodes, together with the conducted right leg, is quite effective in reducing such interference [20].

The human body is a volumetric conductor that collects interference currents like an antenna [21]. The electrical grid is the main source of interference currents. The capacitance between the body and ground (neutral wire) is considered to be about 200 pF. The coupling capacitance to the live wire is usually about 2 pF [21,22]. Maximum values of these parasitic capacitances have been measured as high as 4 nF and 5 pF [23,24]. These capacitive couplings conduct a power line interference (PLI) current of nearly 200 nA, which flows through the body to the ground. The PLI currents pass partially through the electrodes directed to the ground signal electrode of the amplifier. Therefore, bioamplifiers with a high common-mode rejection ratio (CMRR) are needed to avoid a voltage drop at common-mode input impedances. The unshielded conductors are a kind of antenna that collects PLI currents. At the highly insulated front ends, electrodes are passed in the direction of the ground through the body's capacitance. Ultimately, the total amount of PLI currents flowing through the electrodes is multiplied by the impedance imbalance of the electrodes, thus producing the PLI differential voltage drop [25,26]. This is negatively amplified along with the useful ECG signal, and then must be filtered by analog or digital band-reject filters.

EEG signals are generated by the brain because of neural activity. EEG recordings have amplitudes of approximately 10 mV on the cortex and 100 μ V on the scalp surface [27]. The latter are challenging to measure due to noise introduced by the electrical grid, electrode movement, and poor contact between the skin and electrode. Applying an impedance-reducing gel to the electrode contact area helps mitigate noise caused by poor contact [28]. The proper selection of the connection cable between the electrode and the system's input circuit (typically an amplification circuit) reduces power line noise at 60 Hz in the EEG signal. Electrodes placed on the scalp capture the electrical activity (potentials) of neurons and are used to record brain activity. Electrode placement typically follows the 10–20 international standard configuration [28–30].

EEG signals commonly include α , β , and δ waves, each distinguished by different frequency ranges that vary with the activity level of the cerebral cortex. β -waves have frequencies between 12 and 33 Hz, though they can reach up to 50 Hz [31]. These waves appear when a person is thinking or responding to stimuli and are primarily recorded in the parietal and frontal regions. The α -waves, with frequencies between 8 and 13 Hz (overlapping the β range), are recorded in healthy individuals who are awake but at rest, with closed eyes, and are primarily located in the occipital region. The δ -waves, which range from 4 to 8 Hz, are common in childhood, but may also appear in adults during

periods of emotional stress or frustration and are recorded in the parietal and temporal areas. Finally, δ -waves, with frequencies below 4 Hz, occur during deep sleep in childhood and in cases of severe organic brain diseases [28,32–36].

Other applications for systems developed for biopotential acquisition include the acquisition and processing of surface electromyographic (sEMG) signals, which are used in rehabilitation with robotic prosthetic legs and in the detection and analysis of nerve conduction disorders for diagnosing conditions such as Guillain–Barré syndrome, myasthenia gravis, and bruxism, as well as for controlling prostheses [15–20]. Electrocardiographic (ECG) signals are also used for clinical diagnosis, providing reliable indicators of the functional and anatomical state of the cardiovascular system [26,37–39].

The acquisition of sEMG signals is performed using electrodes placed on the skin over the specific area to be analyzed. For example, when bending or extending the arm, the brain sends signals to the muscles of the upper limb to initiate movement. This information travels from the brain to the muscle fibers via the spinal cord as an electrical impulse [40,41]. Before reaching the muscle fibers, this impulse passes through a type of neuron called a motor neuron, which is responsible for receiving and transmitting the electrical impulse to the muscle fibers, thereby generating an action potential. Consequently, when one electrode captures the depolarization signal, the other records a resting potential. Thus, the recorded sEMG signal corresponds to the difference in action potentials between the two electrodes. sEMG signals range from 50 μ V to 5 mV, with a frequency range of 20 to 120 Hz [42,43].

The ECG signal acquisition is performed using two or more electrodes placed on the chest at various points, excluding the reference electrode, to obtain the corresponding electrocardiographic leads. The ECG signal has a frequency spectrum of 0.05 to 100 Hz and an amplitude of 0.02 to 5 mV [33–35]. This signal consists of the P, the QRS complex, and the T waves. The signal ends with a T wave of the same polarity as the QRS complex, reaching an amplitude of 0.1 to 0.5 mV [38,44].

This paper presents the design of a prototype for the simultaneous acquisition of different types of biopotentials using the same electronics, without altering specific parameters or the ADC resolution. The proposed system includes improvements in the interconnection method for the shielding controller system, gain variation through both hardware and software, noise reduction from the electrical grid by employing a cascaded notch filter configuration in the signal filtering stage, and an open architecture in software and hardware that eliminates the need for a PC, allowing migration to a minimal system.

The contributions of this paper are as follows: (1) the proposed system includes a modification in the shield driver configuration for biopotential acquisition, enabling 60 Hz noise reduction and improved signal filtering; (2) the system features an open architecture in both software and hardware, distinguishing it from other systems in the literature and commercially available options; and (3) the system allows for the addition or reduction of electrodes based on user needs. This modularity enables the acquisition of more detailed information depending on the specific study being conducted.

2. Materials and Methods

Some works related to biopotential acquisition are presented as follows: (1) The development of a BCI interface for acquiring steady-state visual evoked potentials using the ADS1298 IC is shown in [45]. This BCI uses the Fast Fourier Transform (FFT) for data processing. Its disadvantages include a fixed gain, a lack of modularity, and the inability to adjust the analog-to-digital converter (ADC) resolution. Additionally, it does not specify a configuration for shielding control to reduce common-mode voltage. (2) Based on an MSP430G2553 with a 10-bit resolution ADC, a BCI was developed [46]. This system includes signal pre-amplification using OPA333 operational amplifiers with gains of

1001 and 8.2. Its disadvantages are a lack of modular implementation, and the use of an existing shielding control circuit reported in the literature for reducing common-mode voltages. (3) In [47], an acquisition card based on an ADuC841 with a 12-bit ADC, which increases the computational load of the system, is presented. This BCI incorporates preamplification stages of 1000 and 3300 using a network of TLV2264 and OPA4344 operational amplifiers, respectively. Disadvantages include the absence of a modular architecture to increase the number of acquisition channels and the use of a shielding driver circuit, based on the literature-reported configurations with operational amplifiers, to reduce common-mode voltage. (4) A BCI is implemented using a closed system called Emotiv Epoc [48], which performs the acquisition and analysis of visual evoked potentials with classification algorithms. As a closed system, it has the disadvantage of not allowing electrode reduction or software modifications, limiting potential improvements. (5) Finally, a BCI is implemented using an INA118U instrumentation amplifier and an MC33171DG operational amplifier for common-mode shielding and voltage reduction control. However, it uses a 24-bit ADS1252 ADC, which increases computational load [49].

Biopotential acquisition systems present several drawbacks, highlighting the need for improvements in the following areas: (1) Electronics for the acquisition of different types of biopotentials (EEG, ECG, and sEMG) that could be adapted to a specific application or study, with a modular configuration that could allow additional modules to expand the number of electrodes for signal acquisition. This benefits end users in analyses or applications requiring multiple types of biopotentials, such as single-trial EEG-sEMG coherence analysis to identify progressive muscle fatigue-related alterations in corticomuscular coupling, multimodal fusion of muscle and brain signals for hybrid BCI generation, fused ECG-sEMG signals for physiological rehabilitation devices, fused EEG-ECG signals for epilepsy diagnosis, or prosthetic designs combining EEG and sEMG signals. (2) Scalability with adjustable gain over a wide range, selectable by hardware, as well as ADC resolution, can be changed if desired by the user, either to reduce computational load or to test with different dedicated cards. (3) The enhanced reduction of a 60 Hz noise introduced by the electrical grid, which interferes with acquired biopotentials and produces a louder, higher-amplitude signal that impacts subsequent analysis. Solutions to this issue typically involve using a shielding controller with operational amplifiers to reduce common-mode voltage between the electrode terminals and the system input.

2.1. Methodology

The proposed system features a scalable configuration that allows for the modification of the previously designed filtering stage based on the type of signal to be acquired, as well as the adjustment of the ADC resolution. This flexibility enhances the characteristics of acquired biopotentials, when required by the study, by providing direct access to the filtered and amplified signal output. The system allows for the easy modification of the ADC acquisition card (Arduino Mega 2560) and includes a scalable configuration for increasing the number of electrodes used in acquiring various biopotentials, tailored to specific study requirements.

The biopotential acquisition process is as follows: based on the study, the appropriate filtering module (Filtering Block) for the desired bandwidth is selected. Next, the system gain is adjusted, and the ADC resolution can be configured as needed. The signal is acquired using a silver/silver-chloride (Ag/AgCl) electrode with a 24 mm diameter, 220 Ω impedance, and an offset voltage of 0.2 mV [50]. Figure 1 presents the block diagram of the proposed system, comprising the following components:

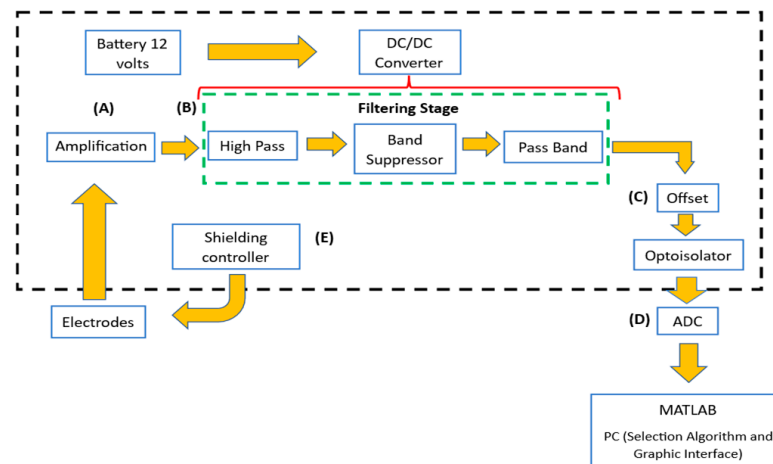


Figure 1. Block diagram of the proposed system, where (A) Signal Amplification Block, (B) Signal Filtering Block, (C) Offset Block, (D) Analog-to-Digital Conversion Block, and (E) Shielding controller.

- (A) Signal Amplification Block:** This stage employs two instrumentation amplifiers. The first amplifier pre-amplifies the signal with a fixed gain of 200, preventing saturation when working with different biopotentials, making the system adaptive and reducing common-mode noise. The second amplifier provides an adjustable gain of up to 10,000 to accommodate the low amplitude of biopotentials. This gain can be modified via hardware using a precision potentiometer to provide gains of 1, 50, 500, 2000, 5000, and 10,000.
- (B) Signal Filtering Block:** The signal obtained from the amplification block is mounted on a DC component with an amplitude of 20 to 50 mV; this is due to the impedance produced by the contact between the skin and the electrode. For this reason, a passive first-order high-pass filter with a cut-off frequency $f_c = 3$ mHz to -3 dB and unit gain is proposed for removing the DC component, followed by a notch filter with a cut-off frequency of 60 Hz, which is used to reduce noise from the electrical grid. Finally, various band-pass filtering modules are implemented to acquire specific biopotential signals (EEG, ECG, sEMG) based on the application, which the user can modify [51].
- (C) Offset Block:** An offset voltage is added to the resulting signal to compensate for the DC component produced by skin and electrode contact, allowing for compatibility with the voltage levels of the Arduino Mega 2560 board.
- (D) Analog-to-Digital Conversion Block:** This block utilizes the Arduino Mega 2560 [52] for data acquisition and further signal processing. The Arduino board is also used as an interface to MATLAB (R2023a), and the entire system is implemented to perform the physical tests; no modules were simulated using MATLAB.
- (E) Shielding controller:** This block is responsible for minimizing common-mode interference, as well as a reference point for obtaining biopotentials; the power line noise is usually not a pure 60 or 50 Hz sine wave but is distorted with artifacts. Therefore, a major advantage of using a RLD (Right-Leg Drive) circuit to attenuate the main hum is that the common-mode signal recorded by the right-leg electrode is actually correlated with the noise in the biopotential recording. The TL082 operational amplifier, which has a low total harmonic distortion of 0.003% and a CMRR of 86 dB, is used to attenuate the common-mode signal recorded by the right-leg electrode. The configuration is based on an Active Shield [53]; the configuration includes an RLD inside the same shield where the lines used for the electrodes are located, also interconnecting through a coupling layer to the ground plane of the device, increasing the CMRR.

Two galvanic isolations are proposed: the first is used to power the system via a DC/DC converter powered by a 12 V battery with symmetrical output voltage, and the second employs an optocoupler to connect the system output to the ADC, thereby eliminating ground loops that may arise from having the entire system connected to a single return point, ensuring user safety. Finally, a proposal for connecting the shielding controller between the electrodes and the system is included, reducing common-mode voltages that could interfere with biopotential acquisition.

2.1.1. Signal Amplification

The amplification block is shown in Figure 2. This stage is designed using two INA128 instrumentation amplifiers. The first amplifier has a gain of 200, a slew rate of 4 V/μs, and a common-mode rejection ratio (CMRR) of 111 dB; the second amplifier has a gain of 10,000, a slew rate of 4 V/μs, and a CMRR of 125 dB, as the voltage amplitude of the EEG signal ranges from 5 to 300 μV [27,28]. The gains of the instrumentation amplifiers are calculated by the following:

$$G = 1 + \frac{R_x}{R_G}, \quad (1)$$

where R_x is a constant whose value is 50 kΩ and R_G represents a resistance value from 0.005 to 50 kΩ to provide a given gain. If $R_G = 1.02$ kΩ, a gain value of 50 is obtained for signal amplification.

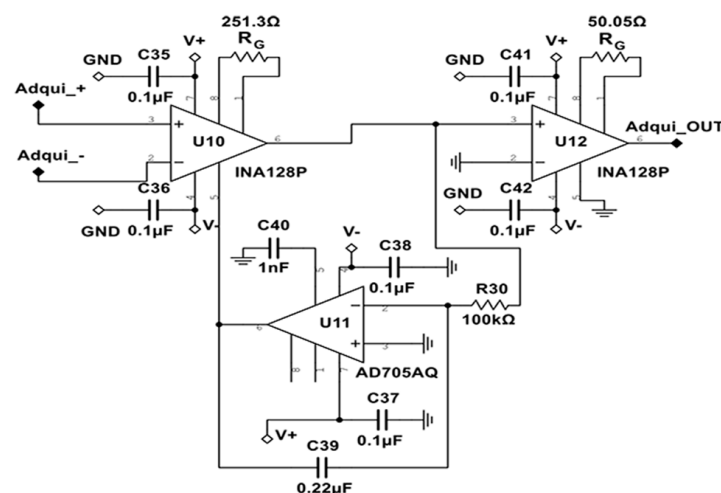


Figure 2. Biopotential amplification block.

An AD705 operational amplifier with a CMRR of 114 dB and a slew rate of 0.15 V/μs is used to stabilize the offset voltage generated at the output of the first instrumentation amplifier.

2.1.2. Signal Filtering

The signal obtained from the amplification block is mounted on a DC component with an amplitude of 20 to 50 mV; this is due to the impedance produced by the contact between the skin and the electrode [33,34]. For this reason, a passive first-order high-pass filter with a cut-off frequency $f_c = 3$ mHz to -3 dB and unit gain is proposed for removing the DC component, using the following:

$$f_c = \frac{1}{2\pi RC}, \quad (2)$$

where $C = 10$ μF is the proposed capacitor and R is the calculated resistance for a given f_c . Solving for R from Equation (2) and substituting the other values gives $R = 5.6$ MΩ. Next, a notch filter with a cut-off frequency of 60 Hz is used to reduce noise from the

electrical grid. This filter is implemented using a TL082 operational amplifier, which has a low total harmonic distortion of 0.003% and a CMRR of 86 dB. The calculation of this filter is performed as follows:

$$\frac{2R_2}{R_1} = \frac{C_1}{C_2}, \quad (3)$$

$$f_r = \frac{1}{2\pi R_1 C_1}, \quad C_0 = 2C_1, \quad (4)$$

where $C_1 = 100 \text{ nF}$ and $C_2 = 100 \text{ nF}$ are the proposed capacitance values for the notch filter, R_1 is the feedback resistance of the operational amplifier, R_2 is the resistance that forms part of the notch filter, C_0 is the grounding capacitance of the notch filter, and $f_r = 60 \text{ Hz}$ is the rejection frequency for which the notch filter is designed.

Solving for R_1 from Equation (4) and substituting the values gives $R_1 = 26.52 \text{ k}\Omega$, which is adjusted to the commercial value of $R_1 = 25 \text{ k}\Omega$. The value of R_2 is calculated from Equation (3), yielding a value of $R_2 = 12.5 \text{ k}\Omega$. The value of C_0 is determined to be a value of 200 nF . Figure 3 shows the schematic diagram used to eliminate the DC component and the 60 Hz frequency noise using the calculated values [54].

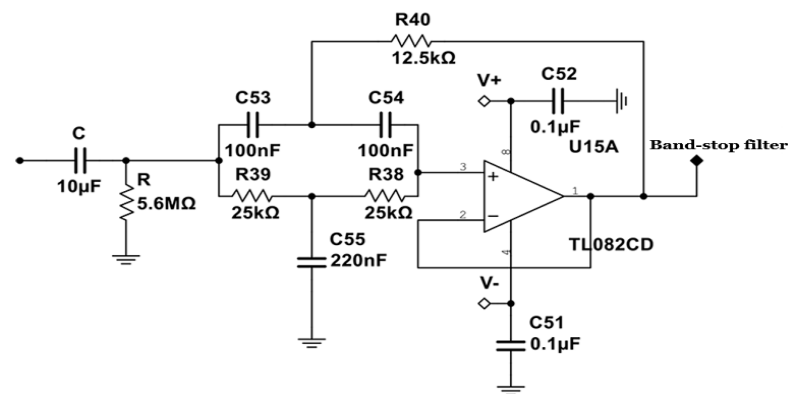


Figure 3. High-pass filtering and band suppressor for biopotentials.

To obtain EEG biopotential signals, active band-pass filters are proposed (one for each frequency range of the analyzed biopotential). For filtering the β signal, a fourth-order Butterworth active band-pass filter with a bandwidth of 12 to 33 Hz is proposed. This filter consists of a low-pass filter stage and a high-pass filter stage connected in series, implemented using a TL084 with a CMRR of 86 dB. This IC contains four operational amplifiers, reducing noise introduced by external components. Similarly, for filtering the α signal, a second active band-pass filter with a bandwidth of 8 to 13 Hz, also implemented with a TL084 with the same characteristics described above, is proposed.

In the design of the low-pass and high-pass filters, the parameters α and k_f , called frequency factors, are used to calculate the damping factor required for the filter order. These parameters are defined as $\alpha_1 = 1.848$, $\alpha_2 = 0.765$, $k_{f1} = 1$, and $k_{f2} = 1$ [54,55]. The low-pass filter is calculated using Equations (5) and (6):

$$C_1 = \frac{2}{\alpha}, \quad C_2 = \frac{\alpha}{2}, \quad (5)$$

$$f_c = f_{3dB} \times k_{f1}, \quad \omega_c = 2\pi f_c R_{1,2} = \frac{1}{w_c}, \quad R_3 = \frac{2}{w_c}, \quad (6)$$

where C_1 and C_2 are the capacitances used to tune the filter to the desired cut-off frequency; α_1 , k_{f1} and α_2 , k_{f2} are the frequency factors used in the first and second section of the low-pass filter, respectively, f_c and w_c are the cut-off frequencies of the filter in Hz and

rad/s, respectively, $f_{3dB} = 33$ Hz is the desing frequency of the filter, and $R_{1,2}$ and R_3 are the resistors used to tune the filter to the given f_c .

Next, the values for the first section of the low-pass filter are calculated, yielding the following: $f_c = 33$ Hz, $C_1 = 108$ nF, $C_2 = 92.4$ nF, $w_c = 207.345$ rad/s, $R_{1,2} = 48.2$ k Ω , and $R_3 = 96.4$ k Ω . The values found for the second section of the filter are $C_1 = 261.4$ nF and $C_2 = 38.25$ nF.

The high-pass filter is calculated using the following:

$$f_c = \frac{f_{3dB}}{k_f}, R_1 = \frac{\alpha}{2w_c}, R_{2,f} = \frac{2}{w_c\alpha}, \quad (7)$$

where f_c is the cut-off frequency of the filter, $f_{3dB} = 12$ Hz is the design frequency of the filter, $C_1 = 1$ μ F and $C_2 = 1$ μ F are the capacitors proposed to tune the filter to the desired frequency, R_1 and $R_{2,f}$ are the resistors calculated to tune the filter to the desired frequency, and α_1 , k_{f1} and α_2 , k_{f2} are the frequency factors used in the first and second sections of the high-pass filter, respectively.

The values for the first section of the high-pass filter are as follows: $f_c = 12$ Hz, $w_c = 75.39$ rad/s, $C_1 = 1$ μ F, $C_2 = 1$ μ F, $R_1 = 12.2$ k Ω , and $R_{2,f} = 14.3$ k Ω ; and for the second section of the filter, the values are as follows: $f_c = 12$ Hz, $w_c = 75.39$ rad/s, $C_1 = 1$ μ F, $C_2 = 1$ μ F, $R_1 = 5.07$ k Ω , and $R_{2,f} = 34.6$ k Ω . Figure 4 presents the configuration used for the band-pass filtering of the EEG β and α signals.

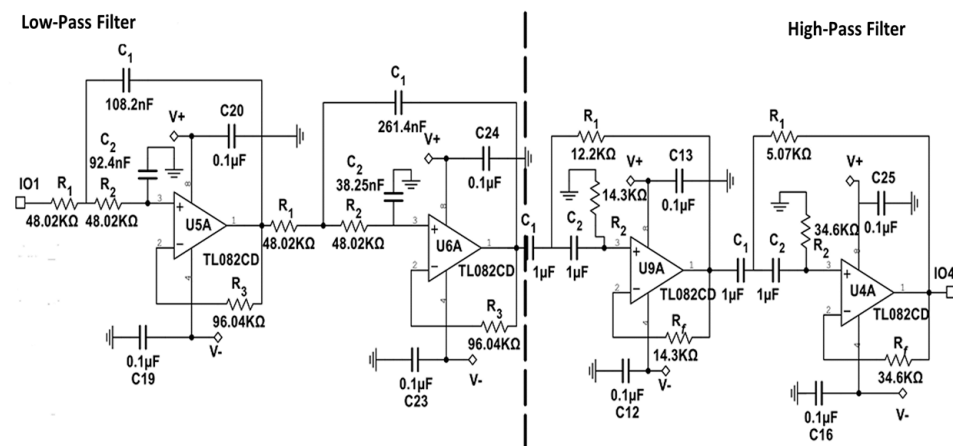


Figure 4. Filtering with bandwidth from 13 to 30 Hz.

It is important to note that for high-pass and low-pass filters, the concept of damping is crucial. In band-pass filters, damping is typically specified by the parameter Q , known as the quality factor, which is the reciprocal of the damping factor. This factor represents the degree of disturbance the filter may introduce, potentially distorting the frequency response of the acquired signals. To determine the cut-off frequency, it is necessary to know the center (or peak) frequency of the filter, which is the point of maximum gain—in our design, this is a unity gain. Our system aims for a wide bandwidth, and based on this, if a filter requires a relatively low Q , around one or less, it is best implemented as a cascade of separate low-pass and high-pass filters [36].

Additionally, tantalum electrolytic capacitors are used for their stability and as decoupling elements for operational amplifiers.

2.1.3. Offset

A DC component is added to the filtered signal to adjust it with the input range supported by the Arduino Mega 2560 (ARDUINO.CC, Scarmagno, Italy) data acquisition board, which has a 10-bit resolution and 16 multiplexed channels, with a sampling rate of 1 MHz for our signal [52]. Figure 5 shows the schematic diagram of the offset generator implemented with two TL082 operational amplifiers (Texas Instruments, Dallas, TX, USA), which have an 86 dB CMRR and a slew rate of 13 V/s. The first amplifier is configured as a voltage tracker, and the second as a non-inverting summing amplifier, both designed with unit gain:

$$v_0 = R_f \left(\frac{v_1}{R_1} + \frac{v_2}{R_2} \right), \quad (8)$$

where v_0 is the output voltage with the DC component added to the signal, R_f is the feedback resistor used to stabilize the bias current, R_1 is the resistor used to combine the signal of the system with the DC component, R_2 is the resistor used to incorporate the DC component into the signal of the system, v_1 is the voltage signal from the system, and v_2 is the DC component voltage incorporated into the system signal [53].

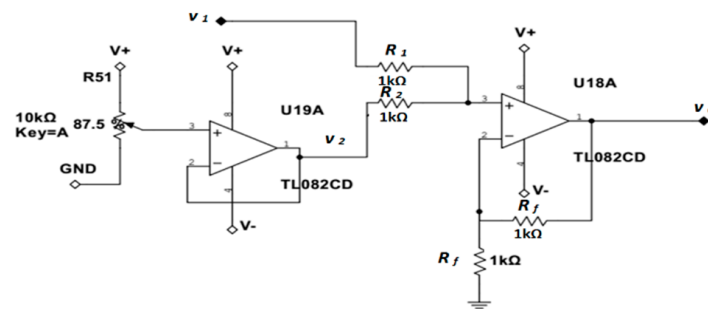


Figure 5. Offset generator block.

To calculate the values of the resistors R_1 , R_2 , and R_f , values of 1 kΩ are proposed to achieve a unit gain in the offset. Substituting these values into Equation (8), we obtain $v_0 = v_1 + v_2$.

Figure 6 shows the system implemented with (a) amplification, filtering, and offset blocks, and (b) the Arduino Mega 2560 ADC. An additional gain can be applied to the acquired signal through a point-to-point multiplicative factor in the software. Additionally, short circuit output protection and low input polarization current are implemented.

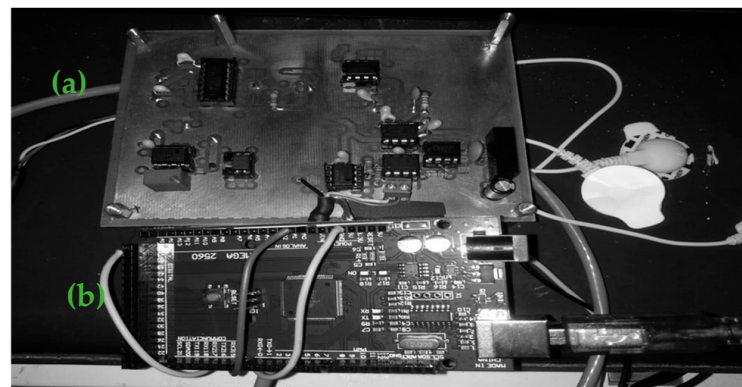


Figure 6. Prototype: (a) card implemented for biopotentials conditioning and (b) Arduino Mega 2560 data acquisition card.

To connect the electrodes to the proposed system, a shielded cable with a configuration different from those used in similar systems is proposed [44–48]. Modifications in the shielding driver configuration have been made to reduce common-mode voltages that can interfere with biopotential acquisition. Unlike other configurations in the literature, which often use a right-leg circuit to reduce common-mode interference, this design minimizes the wiring length that could act as an antenna, thus reducing the body-to-electrode contact that captures electromagnetic noise, particularly the 50/60 Hz noise from the power grid. To address this issue, a configuration is proposed where the conductor length is minimized, and the driver is reconfigured to use a common point across various acquisition configurations for biopotentials. This setup aims to reduce common-mode noise, with the shield and conductor braiding further suppressing electromagnetic interference. Adding a high-gain operational amplifier to the system further decreases electromagnetic noise. Figure 7 illustrates the proposed configuration for connecting the shielding controller to the electrodes and the system.

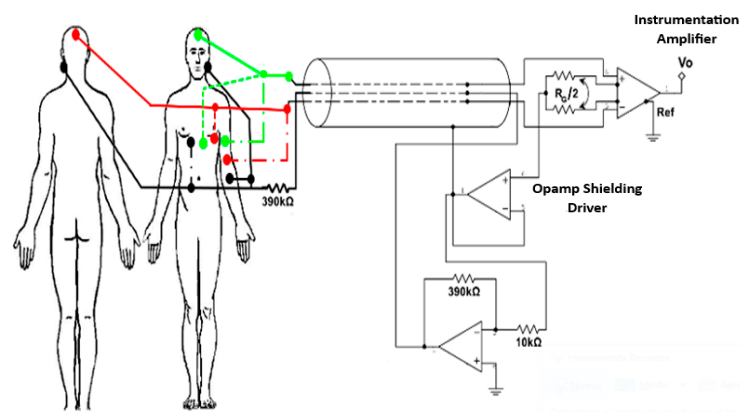


Figure 7. Methodology used to perform the measurement and acquisition of EEG, EMG, and ECG biopotentials whose electrode connections are represented by solid line, dotted line, and dashed line, respectively.

2.1.4. Procedure for Experimental Tests

Experimental tests of the system were performed using a known low-amplitude input signal of 1 pV (V_{pp}) and the gains were obtained with voltage levels within the millivolts to volts. Also, a sinusoidal signal with a peak-to-peak amplitude of 1 mV and a frequency range of 1 Hz to 100 kHz was injected into the system to assess any significant distortion within the frequencies relevant to the biopotentials being measured.

The system is evaluated by acquiring EEG, ECG, and sEMG biopotentials using the proposed shielding driver, applying the Signal-to-Noise Ratio (SNR), which is defined as the ratio between the amplitude of the signal and the amplitude of the noise, expressed in decibels. This can be represented by the following [56–63]:

$$SNR_{dB} = 10 \times \log_{10} \frac{P_{signal}}{P_{noise}} = 20 \times \log_{10} \frac{V_{signal}}{V_{noise}}, \quad (9)$$

where P_{signal} is the power of a signal, P_{noise} is the power of the background noise common gain, V_{signal} is the voltage amplitude of the signal, and V_{noise} is the voltage amplitude of the noise.

To calculate the SNR (Equation (9)) of each type of the acquired biopotential signals, the following measurements were performed [64]: To obtain the V_{signal} value, the difference between the peak to maximum peak voltage of the signal was measured, as shown in Figure 8. Likewise, to obtain the V_{noise} value, the peak to minimum peak voltage of the signal was measured.

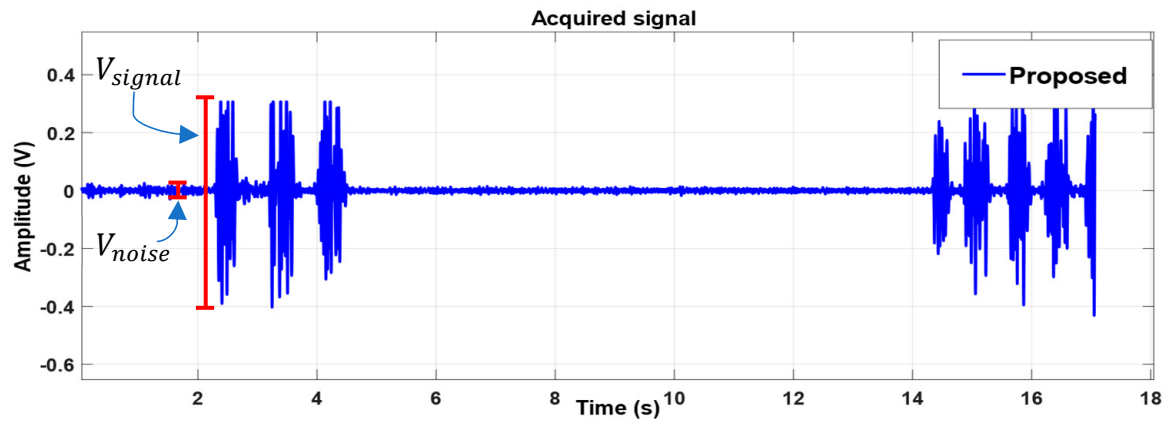


Figure 8. Methodology used to calculate the SNR, obtaining the values of V_{signal} and V_{noise} .

In addition, the CMRR of the system is found using a signal generator to produce a signal with a frequency of 1 Hz and a peak-to-peak voltage of 2 mV, as well as an autotransformer with a frequency of 60 Hz and a peak-to-peak voltage varying from 1 to 200 volts. Figure 9 shows the connections used to make the measurements and obtain the CMRR. The voltage of the autotransformer is varied in steps of one volt and the measurement for the CMRR is performed; the process is continued until 200 Vpp of the autotransformer is reached. Then, the CMRR is computed as follows:

$$CMRR = 20 \times \log_{10} \frac{|A_d|}{A_c} \quad (10)$$

where A_d is the differential gain and A_c is the common gain, both of which are dimensionless [61–63].

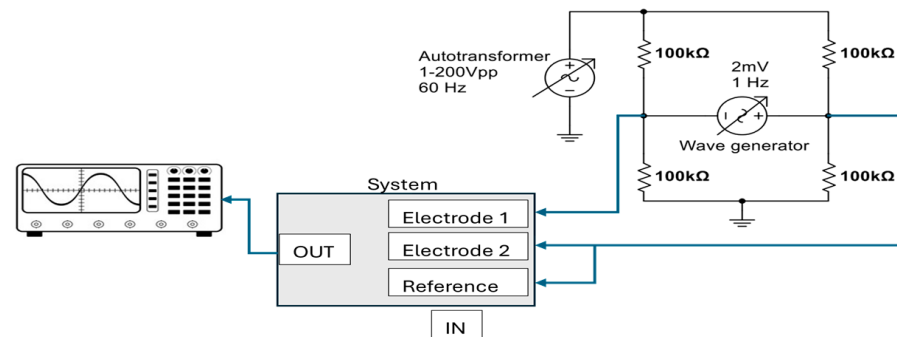


Figure 9. Configuration used to measure the CMRR of the system [37].

To perform this analysis of the acquired biopotentials, an interface is developed in MATLAB® where signal processing and storage are carried out. The FFT is applied to obtain the frequency spectrum of the acquired signal, with the specific analysis of the spectrum at 60 Hz in both acquired signals.

3. Results

The evaluation of the proposed system is performed by implementing the right-leg driving configuration reported in the literature [44,47,50] and the proposed controller on a population of twenty individuals (twenty measurements per configuration). For the acquisition of biopotentials, no digital filtering was applied in MATLAB to the signals obtained with the systems used.

Figure 7 also presents the methodology used for the measurement and acquisition of EEG, sEMG, and ECG biopotentials, with electrode connections represented by a solid line

(β signal in the parietal and frontal regions, α signal in the parietal and occipital regions), a dashed line (signal in the biceps and brachioradialis muscles), and a dotted line (signal from the heart, with electrodes placed between the fourth intercostal space at the right border of the sternum, the fifth intercostal space along the mid-clavicular line, and at the level of the fifth intercostal space on the left anterior axillary line), respectively. The electrode placement is consistent across both the proposed configuration and the configuration reported in the literature for the different types of biopotentials.

The Wilcoxon method [56–60,65] is used to determine if there are significant differences between the signals acquired from the two configurations, while the Bland–Altman method [56–60,65] compares the signals obtained from both configurations to assess the feasibility of using the proposed configuration. Table 1 presents the statistical results of both methods, showing that, according to the Wilcoxon method, the distributions of the two samples are similar. A significance level of $\alpha = 0.05$ (indicated in Table 1) and a 95% confidence interval were used. The average value and standard deviation were calculated to compare the systems. Ideally, the difference in their means should tend to zero; a difference of 0.04 was obtained. To verify system viability, we applied the Wilcoxon method, where the null hypothesis was accepted, indicating no significant difference between the samples, as the p -value obtained was greater than the significance level. This suggests that the samples obtained from the tested system agree with those from the proposed system. In the Bland–Altman method, as shown in Figure 10, a 95% confidence interval (CI) was used, and the difference in means tended toward zero, indicating that the proposed configuration yields similar results to those reported in the literature.

Table 1. Statistic values obtained from two systems.

System	Average (μ)	Standard Deviation (σ)	p -Value Wilcoxon/ α
Literature	0.158	0.088	0.546/0.05
Proposal	0.195	0.106	

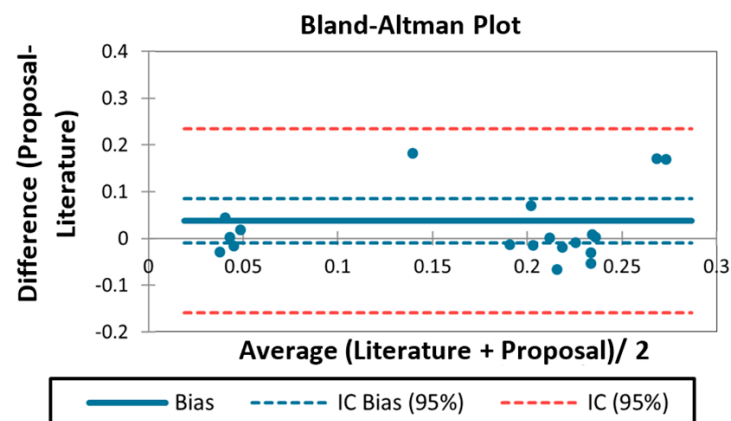


Figure 10. Bland–Altman plot.

Figure 11 presents the experimental results of the CMRR of the proposed system. It is observed that the system achieves a CMRR of 141.767 dB when a peak-to-peak voltage of 180 Vpp is applied, corresponding to an RMS voltage of 127 volts at 60 Hz.

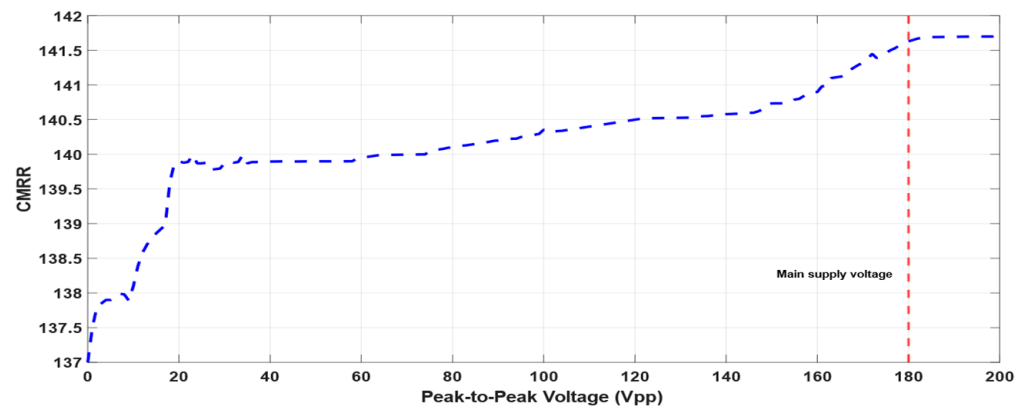


Figure 11. Experimental results of the CMRR of the proposed system as a function of different input voltages.

The CMRR expresses the ability to suppress common-mode signals, which are very large in the human body; therefore, for the overall circuit, the RLD circuit is adopted to minimize the common-mode interference. For this reason, the importance of this work is a new configuration using an Active Shield to improve the noise reduction.

Also, a comparative performance, as a function of the CMRR with different state-of-the-art biopotential readout circuits, is presented [66–77]. Table 2 summarizes the performance comparisons, where it is seen that the proposed system provides the better CMRR result in comparison with other state-of-the-art systems.

Table 2. CMRR-based performance comparisons.

	This System	[41]	[44]	[45]	[46]	[47]	[48]	[49]	[50]	[51]	[52]	[54]	[55]
CMRR (dB)	141.767	66	83	98	80–93	76	110	74	120	100	110	120	60

Figure 12 presents the acquisition and processing of EEG biopotentials for β and α signals. Figure 12a shows the EEG signals resulting from filtering for the β signal in the time domain, and Figure 12b presents these signals in the frequency domain. In both representations, the proposed configuration (blue) shows a better reduction of 60 Hz power line noise compared to the configuration reported in the literature (red). Figure 12c displays the EEG signals after filtering for the α signal in the time domain, while Figure 12d shows the frequency spectra of the signals from Figure 12c. In both representations, the proposed configuration (blue) provides a better 60 Hz noise reduction than the configuration reported in the literature (red).

For sEMG and ECG biopotential acquisition and processing, the resulting signals are presented in Figure 13. Figure 13a shows the sEMG signals in the time domain, and Figure 13b presents the frequency spectrum of the sEMG signals from Figure 13a. An improved reduction of a 60 Hz power line noise can be observed using the proposed configuration (blue) compared to the signal acquired with the configuration reported in the literature (red). Figure 13c,d show the experimental results for ECG signals acquired with the proposed configuration (blue) and the configuration from the literature (red). Similar to the results for EEG and sEMG signal acquisition, the ECG signal acquired with the proposed configuration demonstrates better noise reduction.

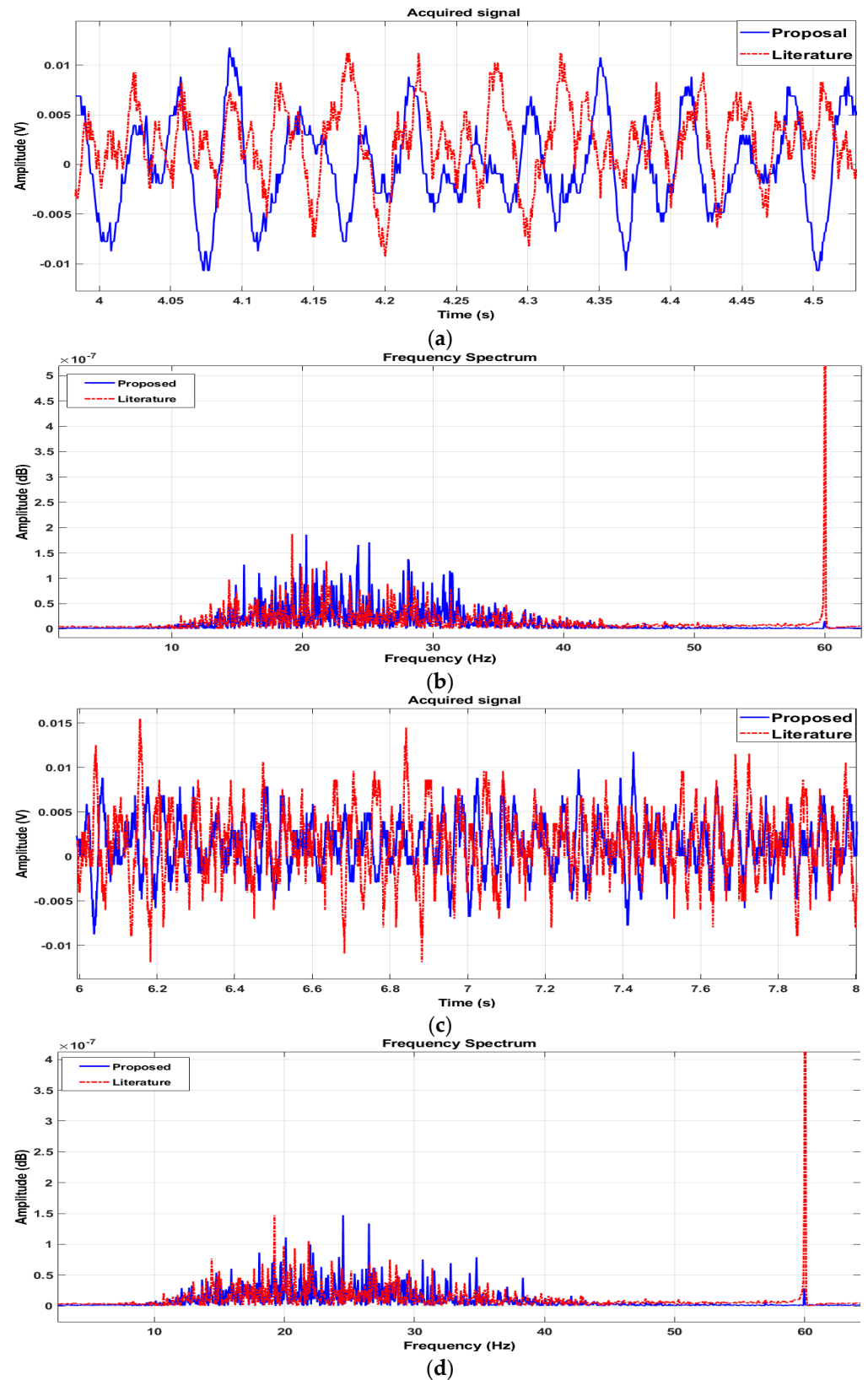


Figure 12. EEG β signal acquisition: (a) time-domain representation of the EEG signal acquired with the proposed configuration (blue signal) and with the configuration reported in the literature (red signal); (b) frequency spectrum of the time-domain signals shown in (a). EEG α signal acquisition: (c) time-domain representation of the EEG signal acquired with the proposed configuration (blue signal) and with the configuration reported in the literature (red signal); (d) frequency spectrum of the time-domain signals shown in (c).

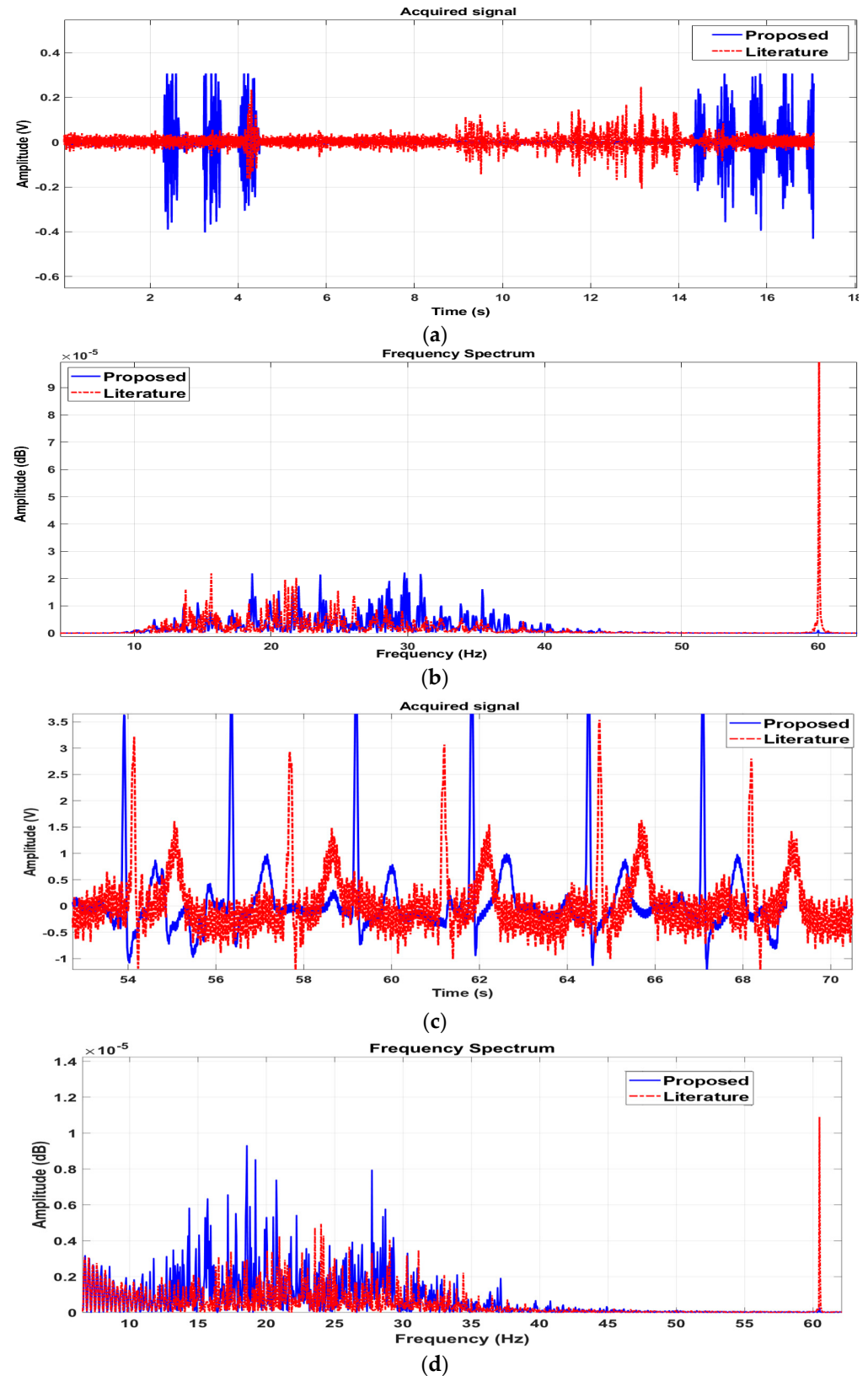


Figure 13. sEMG signal acquisition: (a) time-domain representation of the sEMG signal acquired with the proposed configuration (blue signal) and with the configuration reported in the literature (red signal); (b) frequency spectrum of the time-domain signals shown in (a). ECG signal acquisition: (c) time-domain representation of the ECG signal acquired with the proposed configuration (blue signal) and with the configuration reported in the literature (red signal); (d) frequency spectrum of the time-domain signals shown in (c).

A phase shift was introduced between the two signals to better observe the noise difference in both time and frequency domains. In Figures 12b,d and 13b,d, the blue line at 60 Hz in the frequency domain shows an amplitude difference compared to the red dotted signals. Additionally, in the frequency-domain figures, the blue signal displays only a small peak at 60 Hz. However, since a significant phase shift was not added, the red 60 Hz peak, indicating the signal acquired with the configuration reported in the literature, may appear as if it belongs to the signal obtained with the proposed system.

Finally, the SNR is calculated for the EEG, sEMG, and ECG signals in Figures 12 and 13. The EEG β and α biopotentials have SNR values of 39.4534 dB and 31.5643 dB, respectively. For the sEMG and ECG signals, the SNR values are 30.5287 dB and 29.8654 dB, respectively.

4. Discussion

A new configuration for connecting electrodes to the proposed system is presented; this reduces the noise caused by electrode–skin contact, noise generated by electrode movement, and the intrinsic 60 Hz electrical network noise. Temporal and frequency-domain biopotentials acquired with the proposed configuration show improved external noise reduction compared to those acquired with the configuration reported in the literature. The proposed method presents an improvement in noise reduction at 60 Hz in terms of the CMRR in comparison with the state-of-the-art methods published in the recent literature.

Biopotentials acquired with the proposed configuration demonstrate improved external noise reduction compared to those acquired with the configuration reported in the literature. The data analysis confirms that the proposed configuration accurately acquires signals, making it feasible for use in biopotential applications.

The hardware system can support an increased number of electrodes and offers flexibility in modifying the filtering module based on the specific study, allowing the acquisition of various biopotentials compared to other systems in the literature.

The software system enables the pre-processing of signals at the amplification and filtering stages and allows for the modification of the ADC resolution, facilitating the use of an embedded card to replace a PC.

5. Conclusions

This work presents a system for acquiring EEG, sEMG, and ECG biopotentials. The electronic characteristics of the proposed system—specifically in its amplification, filtering, and compensation blocks—make it simple, efficient, and suitable for various biopotential applications.

The nonparametric Wilcoxon test was used to evaluate differences between samples obtained with the proposed and literature-reported configurations, with a significance level of $\alpha=0.05$. A p -value of 0.546 indicated no significant difference between the data, confirming the feasibility of the proposed configuration. Additionally, the Bland–Altman method was used to compare the signals and determine any deviations between configurations, further validating the proposed configuration by showing that most values lie within the 95% confidence interval on the Bland–Altman plot.

The system can be scalable, allowing for an increased electrode count and adjustable filtering stages, depending on the study requirements. The 10-bit ADC resolution reduces computational load, supporting migration to a dedicated system. The system's resolution can be increased or decreased based on study requirements. Its open architecture enables the partial or complete modification of the MATLAB-based software, allowing for potential migration to other programming languages such as C++ V17, Visual Basic .NET V16.9, or Python V3.10.7. Additionally, the hardware can be modified at the amplification, filtering, and ADC resolution levels, thanks to the modular design of the system. This flexibility

supports the replacement of the PC with an embedded card, FPGA, or compact systems like the Hummingboard-I4, Raspberry Pi, or NVidia Jetson. The proposed system has a low cost (USD 50) compared to commercial systems, which range from USD 700 to USD 1700.

Future work will address current system limitations, including developing classification interfaces for EEG, ECG, and sEMG biopotentials with digital filtering, generating control signals using power electronics, increasing the number of electrodes via software for enhanced robustness, proposing a multiplexing configuration to allow for the software selection of specific filter bands, and migrating the system to an integrated board to create a portable device for medical diagnostics and applications for individuals with physical and motor limitations. Future developments also include an interface for applying various biopotential-processing and digital-filtering techniques.

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