Portable cost-effective EEG data acquisition system

Article *in* Journal of Medical Engineering & Technology · March 2011

DOI: 10.3109/03091902.2011.560701 · Source: PubMed

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Portable cost-effective EEG data acquisition system

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(*Received 29 October 2010; revised 25 January 2011; accepted 2 February 2011*)

Neuro-cognitive dysfunctions are common clinical abnormalities found in society. They require objective analysis by various instruments; an important technique involves monitoring electroencephalogram (EEG) signals. To date, EEG machines have been robust, costly and require patients to come to a hospital for test. Therefore, we have constructed a simple, cheap and portable EEG instrument for wider patient use. It consists of two active digital EEG probes with two channels each, making it a four-channel portable acquisition system. It is further connected through a two-wire serial bus to the acquisition unit, which comprises an analogue to digital converter (ADC) and an ARM board processor with 2 GB memory and USB interface. The whole system is placed in a small box making it highly portable for wider use in clinical settings.

Keywords: Cognition; EEG amplifier; Driven right leg; ORCAD design; HPF; LPF

1. Introduction

Neuro-cognitive dysfunctions such as depression, stress, frustration, memory loss, degree of interest, excitement, attention and drowsiness have been frequently reported in society [1–5]. Previously, doctors used questionnaires and facial expressions to diagnose the patient. These methods are based on assumptions, making them very inaccurate. Therefore, an objective analysis needs to be performed, which can only be done by monitoring real-time bio-signals electroencephalogram (EEG). In the past, EEG machines were robust and costly, and patients had to attend major hospitals in cities for their EEG to be measured. Therefore an effort was made to reduce the size of the EEG machine so that it could be used more frequently and in a wider range of clinical circumstances. Portable EEG machines are currently available in the market; although they solve the issues related to size to some extent, they are still expensive for wider usability in the field. Thus we decided to make an integrated EEG (figure 1) system with data logger, that is simple, cheap, portable and convenient for patient use.

2. Method and materials

The amplitude of the EEG signal is in the order of few microvolts, making it difficult to acquire and measure. Hence an amplifier is required to strengthen the bioelectric potentials to a desired or recognizable level. Further, due to the small amplitude, these signals are very susceptible to any background noise, which can be picked up by the human body through the ground path or the cables used for the recording unit of the electrodes [6]. Thus an integrated EEG system was designed with a bandwidth of 0.16–47 Hz for patient monitoring purposes [7]. For recording EEG signals, a four-channel portable acquisition system was developed. The electrodes were connected to the patient's scalp. In this way, EEG readings can be acquired when the patient is performing mental tasks. The

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Journal of Medical Engineering & Technology
ISSN 0309-1902 print/ISSN 1464-522X online © 2011 Informa UK, Ltd.
http://www.informahealthcare.com/journals
DOI: 10.3109/03091902.2011.560701

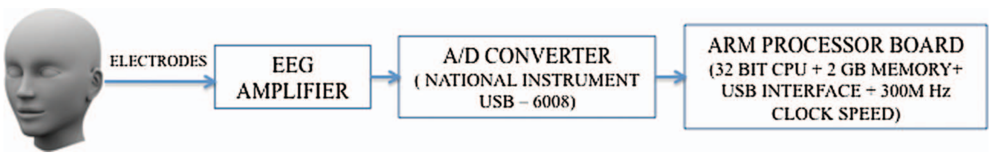

Figure 1. Block diagram of the integrated EEG system.

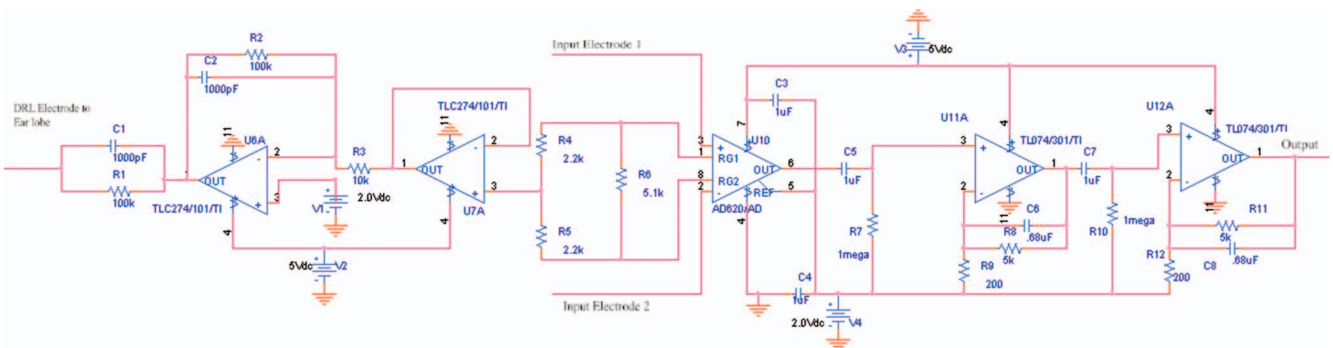

Figure 2. Schematics of the EEG amplifier.

system consists of two active digital EEG probes, with two channels each, connected by a two-wire digital serial bus to the acquisition unit. The acquisition unit comprises of an analogue to digital converter (ADC) and an ARM processor board, which stores the data onto memory. The latter can be accessed later through a USB interface. The amplifier and the acquisition unit were placed in a small box ($5.1 \times 6.4 \times 12.7$ cm), which is sufficiently portable. So the system first acquires the EEG signal and then stores it in the inbuilt memory, which can later be loaded in the main computer for further processing and diagnosis.

The design consists of four subsections, as shown in figure 2: an instrumentation amplifier, a driven right leg amplifier, a filter section and the power source. The overall design was tested in ORCAD (9.1 student version) and was implemented on printed circuit board (PCB). The EEG acquisition was carried out by placing two electrodes onto the scalp and one (driven right leg electrode) onto the ear [8]. The signal was amplified, filtered and the results were viewed on the oscilloscope successfully. Further noise analysis was performed and it was found that AD620 provides acceptable levels of noise.

2.1. *Instrumentation amplifier*

The instrumentation amplifier amplifies the signal difference and rejects input signals common to both input leads. This is very important as the noise level is practically the same on both the input leads of the instrumentation amplifier, so due to its ability to reject input signals common to both inputs it will reject the noise. Brain potentials are different on each electrode so it will amplify the potential difference of input signals.

The EEG has a magnitude of around 100–500 μV, so a reasonably high gain and a high quality bio-potential amplifier is needed to amplify the signal [9]. Figures 3 and 4 show the circuit diagram and the output of the amplifier for a 100 μV input signal. Instrumentation amplifier AD620 is cheap, indigenously available and consumes low power, and so is suitable for portable medical instrumentation application. This instrumentation amplifier offers excellent accuracy that rejects common signal with 100 dB. It also has low offset voltage of 50 μV and a low drift voltage of 0.6 μV/°C.

The differential gain of AD620 was set to 23 using three resistors, two of 2.2 kΩ and one of 5.1 kΩ. As shown in the circuit, the two 2.2 kΩ resistors were in series, making 4.4 kΩ. This is in turn parallel with 5.1 kΩ. So the net resistance (R_G) is 2.3 kΩ. Using the gain formula for AD620:

$$G = 49.4\,\mathrm{k\Omega}/R_G + 1, \qquad (1)$$

or

$$R_G = 49.4\,\mathrm{k\Omega}/G - 1. \qquad (2)$$

Reference pin 5 was connected to the virtual ground (2 V), while pin 4 was connected to the ground. The bypass capacitors at pin 7 and pin 4 were used to reduce the ripple in a circuit.

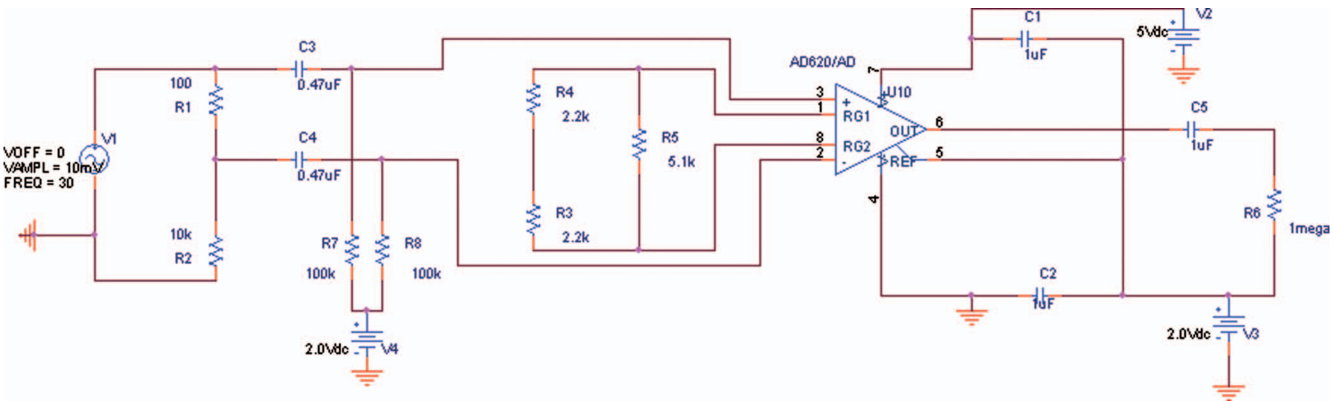

Figure 3. Circuit diagram of the instrumentation amplifier.

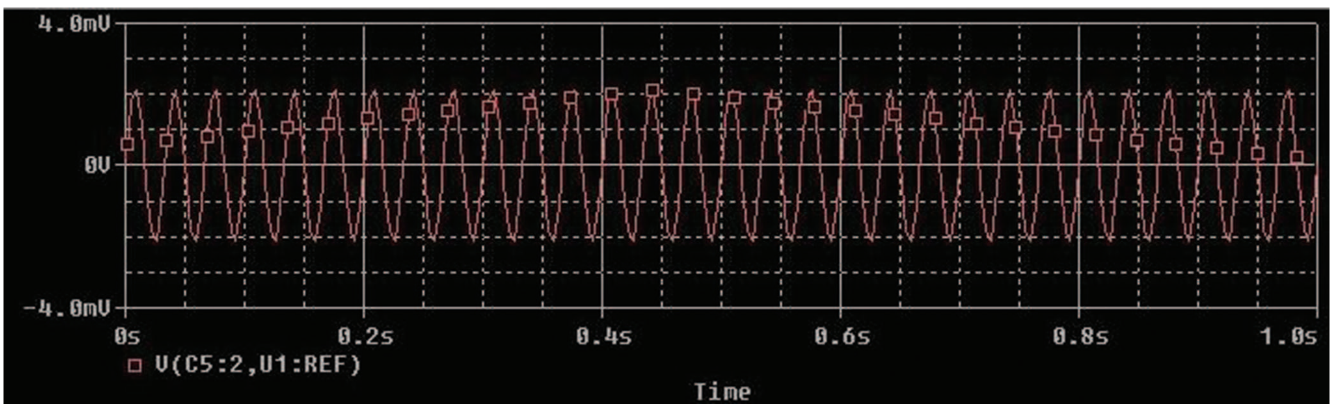

Figure 4. Output of the instrumentation amplifier.

2.2. Drive right leg

A driven right leg (DRL) circuit is used as a connection between the signal source and the amplifier common (0 V, the midpoint between the supply voltages). It reduces the common mode voltage by driving from the instrumentation amplifier actively to the potential of the amplifier common [10]. Therefore a DRL is required in every biomedical recording system for interference suppression and user safety.

This design of the DRL circuit used a TLC274, due to its high input impedance and low power consumption (figure 5). The capacitor in the feedback loop was chosen to maintain the stability of the right leg drive loop [11]. It limits high-frequency gain and helps to prevent oscillation. The output of TLC274 was then connected to the right leg electrode, which in turn is connected to the ear (for EEG measurement).

2.3. Filter section

The filter section is mandatory for biopotential signals such as EEG, as they have a particular bandwidth. For an EEG this range is between 0.16 Hz and 47 Hz. Therefore an appropriate filter has to be designed keeping these ranges in mind. The operational amplifier used for this stage was TLC274, which is a quad amplifier with high impedance and low noise.

The filter section contains two types of filter arranged in a cascade (figure 6). One was the passive high-pass filter (HPF) and the other an active low-pass filter (LPF). Both of them were of order one. The passive high-pass filter passes high frequencies signals but attenuates frequencies lower than the cut-off frequency. It was applied after the instrumentation amplifier to remove DC offsets. Its cut-off frequency was set to 0.16 Hz. After the instrumentation amplifier, there was a second high-pass filter stage, which was applied just after the first operational amplifier (having same cut-off frequency) because amplification brings some new DC components [12] to the signal, causing some offset and drift voltage.

$$\text{Cut-off frequency (HPF)} = 1/2\Pi RC = 0.16\,\text{Hz}, \qquad (3)$$

where $R = 1\ M\Omega$ and $C = 1\ \mu F$.

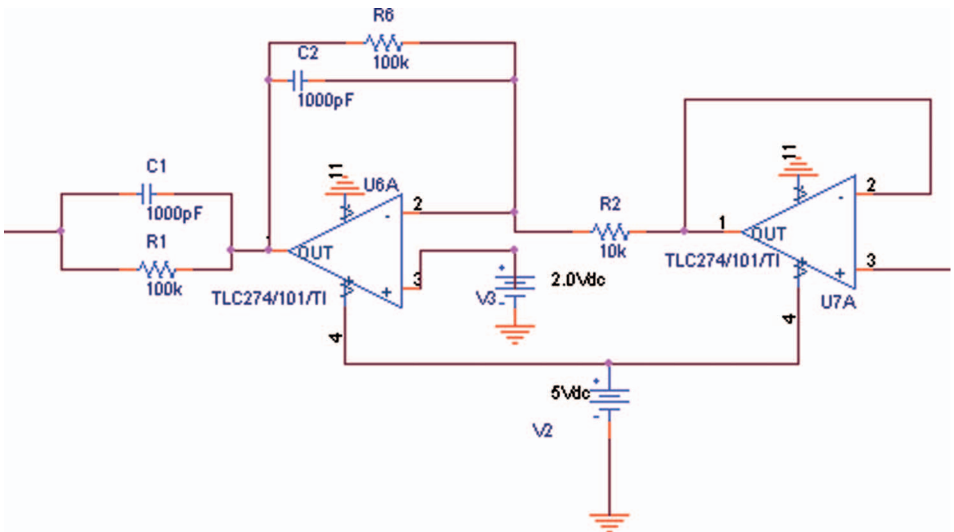

Figure 5. Circuit diagram of the drive right leg circuit.

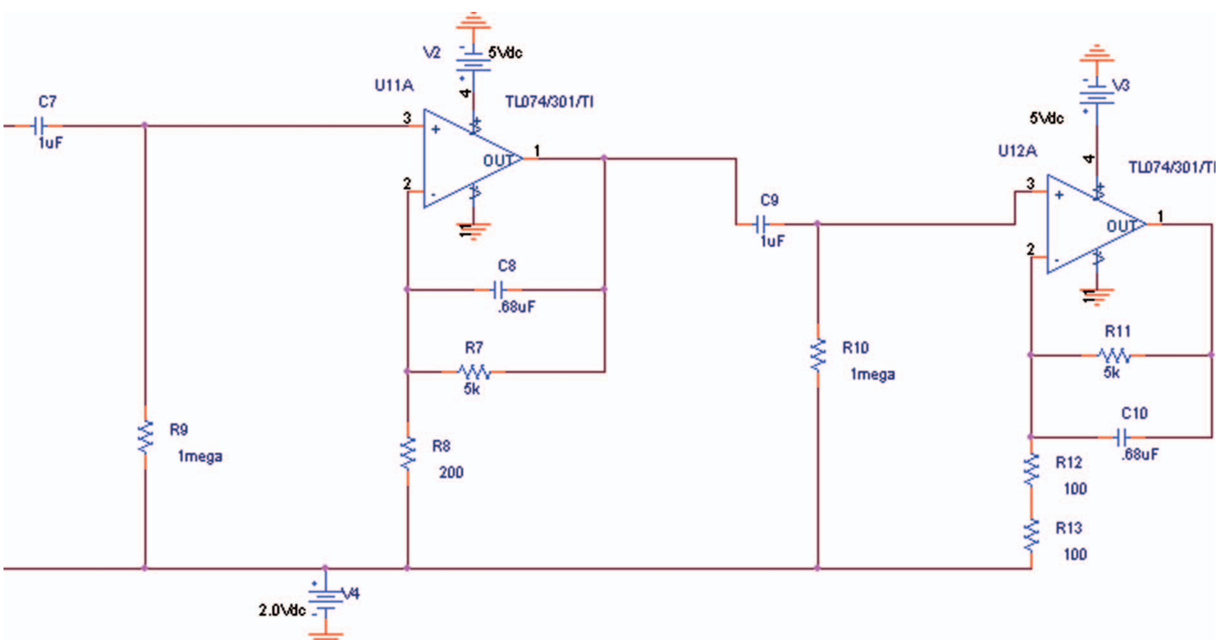

Figure 6. Circuit diagram of the filter section.

The active low-pass filter passes low-frequency signals but attenuates frequencies higher than the cut-off frequency. The fixed low-pass filter limits the frequency up to 47 Hz. It was chosen to filter out unnecessary signals larger than 47 Hz, including power line interference.

$$\text{Cut-off frequency (LPF)} = 1/2\Pi\text{RC} = 47\,\text{Hz}, \qquad (4)$$

where R = 5 kΩ and C = 0.68 μF.

The LPF filters also had a gain of 26 each, making the amplification of the filter section 676 (26 $\times$ 26). With the gain of AD620, the total amplification was 15548. Therefore the EEG signal will have an output strength of nearly 1.5 V if a 100 μV differential signal is given as an input.

2.4. Power source

A 6 V battery could be used to power the entire hardware, if it provided a constant voltage. To avoid voltage fluctuation a L07805 voltage regulator was used. It supplied a continuous voltage of 5 V from a commonly available 9 V battery.

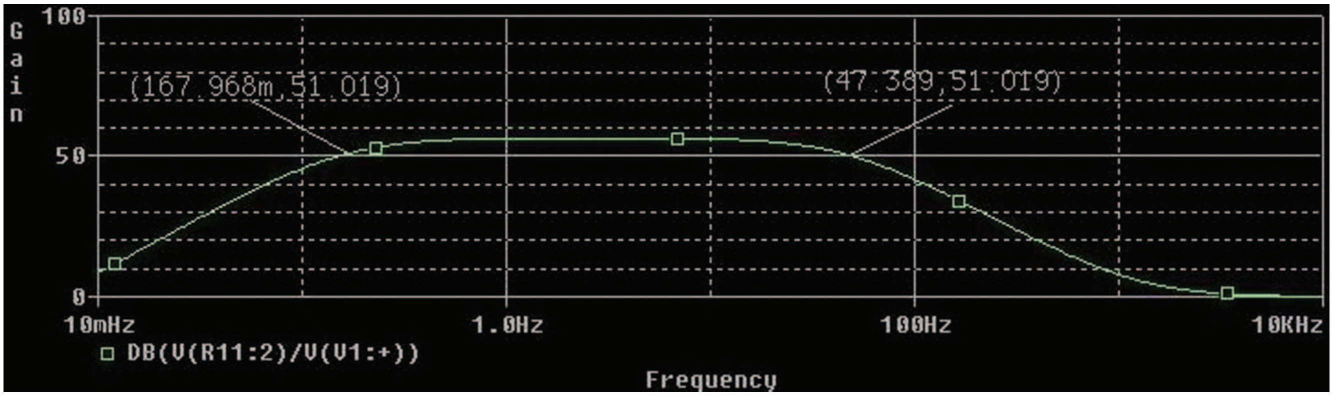

Figure 7. Frequency response of the filter section.

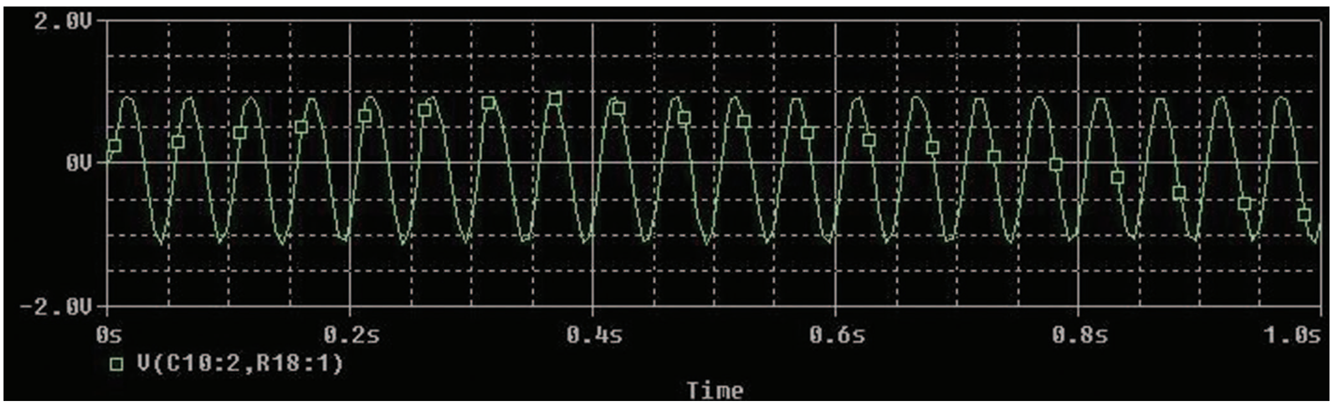

Figure 8. Output if 100 μV is given to the system.

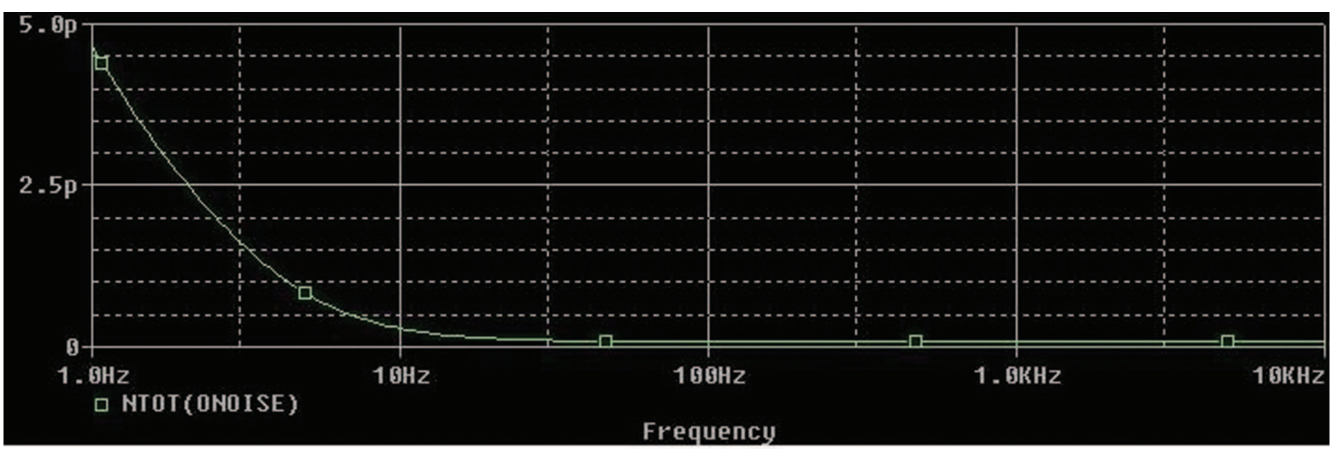

Figure 9. Noise analysis using AD620.

A common problem in analogue electronics is a requirement for a dual-voltage supply (e.g. ± 5 V) but only having a single supply available, such as a battery. Dual battery split into two is not a very efficient idea, as sometimes one of the batteries drains faster, causing DC offset [13]. Therefore, in signal-conditioning applications such as this, we used a single power source of 5 V and a reference voltage (virtual ground) of 2 V. Using the concept of voltage division, 5 V was converted into 2 V using a 330 kΩ and a 220 kΩ resistor. The output of this was fed to a voltage follower, which was further connected with suitable capacitors for buffering.

3. Results

Following a 100 μV signal to the system, the frequency response of the filter section (figure 7) and the output (figure 8), were studied and plotted successfully in ORCAD. Noise analysis of the circuit with AD620 as the instrumentation amplifier using ORCAD is shown in figure 9. The signal-to-noise ratio (SNR) was also calculated using the noise analysis data for minimum frequency and the noise was found to be acceptable.

For the SNR calculation:

$$\mathrm{SNR_{dB}} = 20\log([\mathrm{Signal\ RMS}]/[\mathrm{Noise\ RMS}]) = 146.02. \tag{5}$$

The EEG test was performed only when a perfect electrocardiogram (ECG) waveform was acquired. This showed that the instrumentation amplifiers, operational amplifiers and filters were working well and also all noises and artifacts were at the lowest level. However, with the present gain, testing ECG would lead to saturation of the signal, therefore output of one stage prior to the final one was obtained for the ECG.

After testing it for amplification, noise and cut-off frequency in ORCAD, it was physically implemented onto the printed circuit board (PCB) as shown in figure 10. The dimensions of the PCB were 5.1 × 6.4 cm. It represents a single channel only and to make it four channels, four such PCBs were stacked on top of each other. The PCBs along with the ADC, memory card and USB interface were placed in a box, making it a practical portable acquisition system.

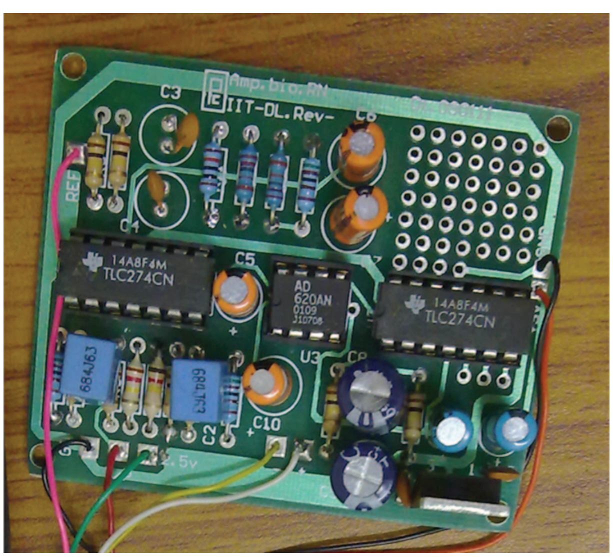

Figure 10. PCB representing a single channel.

4. Conclusion

In conclusion, a data logger for the acquisition of raw surface EEG signals detected with electrodes has been designed and successfully tested. The entire system is simple, battery powered, portable, and has a USB interface. The system can be used for field applications to study EEG signals. Besides EEG data, this hardware also can store ECG data by reducing its gain from 15 548 to 1000 in our future multi-parameter portable system.

Declaration of interest: The authors report no conflict of interest.

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