ARTICLE
Bio-amplifier with Driven Shield Inputs to Reduce Electrical Noise and its Application to Laboratory Teaching of Electrophysiology

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We describe a custom-designed bio-amplifier and its use in teaching neurophysiology to undergraduate students. The amplifier has the following features: 1) differential amplification with driven shield inputs, which makes it workable even in electrically unshielded environments, 2) high input impedance to allow recordings of small signals through high signal source impedance, 3) dual fixed frequency bandpass filters (1-340Hz for surface EMG, EEG, local field potential etc and 320Hz – 3.4kHz for neuronal action potential recording) and independent gain controllers (up to x107,000) to allow the recording of different signals from the same source (e.g., local field potential and spiking activity of neurons), and 4) printed circuit board technology for easy replication with consistent quality.

We compared its performance with a commercial amplifier in an electrically noisy environment. Even without any electrostatic shield, it recorded clear electromyographic activity with little interference from other electric appliances. In contrast, the commercial amplifier's performance severely deteriorated under the same condition.

We used this amplifier to build a computer-controlled stimulation and measurement system for electroencephalographic recordings by undergraduate students. The students successfully recorded various sensory evoked potentials with clarity that otherwise would have required costly instruments. This amplifier is a low-cost yet reliable instrument for electro-physiological recording both in education and research.

Key words: amplifier; driven shield; electromyogram; electroencephalography; neuron; physiology

In teaching neurophysiology laboratories, ideally every student experiences recording nervous system activity. Yet most commercially available equipment is not amenable to meeting this ideal. First, the high cost of commercial amplifiers often precludes obtaining sufficient number of units for students' training. This is not a desirable situation because, in group teaching, only a limited number of students have the chance to record neural and/or muscular activity directly. Second, commercial amplifiers are designed for use in electrically shielded environment, which necessitates the use of Faraday cages during recording. For recording from small animals such as rats, shielding is not a major problem. On the other hand, for human subjects, this requires large Faraday cages or shielded room, thereby restricting the place where recording is done. Some manufacturers offer "active electrodes" with a built-in unity gain amplifier to reduce the noise. Yet these electrodes come as surface electrodes, which restricts their application (e.g., recording of EEG, surface EMG). Moreover, most of the commercially available active electrodes are designed to work with specifically designed amplifiers which are again expensive.

Therefore, with affordability and noise attenuation as major requirements, we developed a relatively simple and inexpensive amplifier for students. It features differential amplification which cancels common mode noise, and driven shield technology that protects the inputs against electrostatic interference. We found that these features ensured stable recording in noisy environment, and thus made this amplifier a useful instrument both for students' training and research by faculty members. The gerber files to fabricate this amplifier are available at www.neurophysiology.med.tohoku.ac.jp/work/work08e.html.

MATERIALS AND METHODS

Circuit design

Figure 1 illustrates the circuit diagram of this amplifier. It consists of the 1) initial input stage, 2) broad band amplification stage, 3) gain controller, and 4) final band-pass filtered amplification which are explained in detail below.

1) Input stage

Special attention was paid to this part because it is the most critical part for the overall performance of the amplifier. For the first stage of amplification, we chose Burr-Brown's INA116. This integrated circuit (IC) is a differential amplifier: i.e., it takes inputs from the reference and the indifferent electrodes (pins 6 and 3, respectively) and amplifies their difference, thus subtracting out the noise common to both inputs. A notable feature of this IC is the "driven shield" inputs: i.e., it holds the shield of the input coaxial cable at the same voltage as the electrodes connected to the input through the buffered guard drive pins (pins 2 and 4 for negative input pin, pins 5 and 7 for positive input pin). As a result, the capacitance between the electrode and the shield is cancelled, thus preventing the electrostatic interference through the capacitive coupling between them (Fraden, 2003). Moreover, its exceptionally high input impedance (over 1 peta Ω) and low

input bias current (typical value is ±3 fA) make it a suitable choice to record signals of small amplitude through high signal source impedance.

The 33 MΩ resistor (R1 in Figure 1) connecting the positive and the negative input pins is important. When the electrode is connected to an amplifier with very high input impedance, the capacitive impedance between the electrode and the tissue causes a DC potential, referred to as electrode potential, to develop over time. Because this electrode potential is unstable and varies across electrodes, it causes the baseline of the amplifier's output to fluctuate. The resister between the differential input pins balances this electrode potential by shunting the capacitances developed at the electrodes. The impedance of this resistor must be low enough to allow the electrode potentials to balance against each other, but sufficiently higher than the electrodes' impedance. We found that 33 to 47 MΩ is the best balance for various types of electrodes tested in our study.

The INA116 has only limited slew rate (0.8 V/μs). Therefore, if the gain is too high, its output may be distorted for fast-changing input. For this reason, we limited the gain of this stage to 19.5.

2) Broad band amplifier

This part consists of two-poles band pass filters with gain (x 93.4). Usually, voltage-controlled, voltage-source (VCVS) or multiple feedback (MFB) design is used to construct frequency filters with operational amplifiers. However, we chose non-inverting amplification circuit with RC filter because this design allowed greater degree of freedom over the setting of gain and cutoff frequencies. Also, its output recovers faster when the amplifier is saturated by sudden changes in the DC offset at the input. We used TL072 for this part because of its low cost, availability and relatively good frequency response property. If necessary, the upper and lower cutoff frequencies can be independently changed without affecting the gain by replacing the capacitors C4–C7. In most of the cases, we set the passband between 1Hz to 3.7kHz. This frequency range covered most of the physiological signals of interest (EEG, surface and intramuscular EMG, action potentials, etc).

3) Gain controller

This part has two independent gain controllers, each of which gives the divided voltage of the broad band amplifier to the next band pass filter. It consists of capacitors (C9, C10) to cut the DC offset, a pair of single-throw single position toggle switches (SW1-1, and SW1-2), and two pairs of 910 kΩ fixed (R7, R8) and 100 kΩ variable resistors (VR1, VR2). When the switch is on, the voltage from the previous stage is directly passed to the variable resistor, which gives the attenuated voltage. When the switch is off, the input voltage is divided between the fixed and the variable resistors by the ratio of 9 to 1, the latter of which further attenuates it.

4) Final band pass filtered amplifiers

This stage consists of band pass filters with gain. We chose to incorporate two sets of them to allow users to record two different signals of different frequency ranges

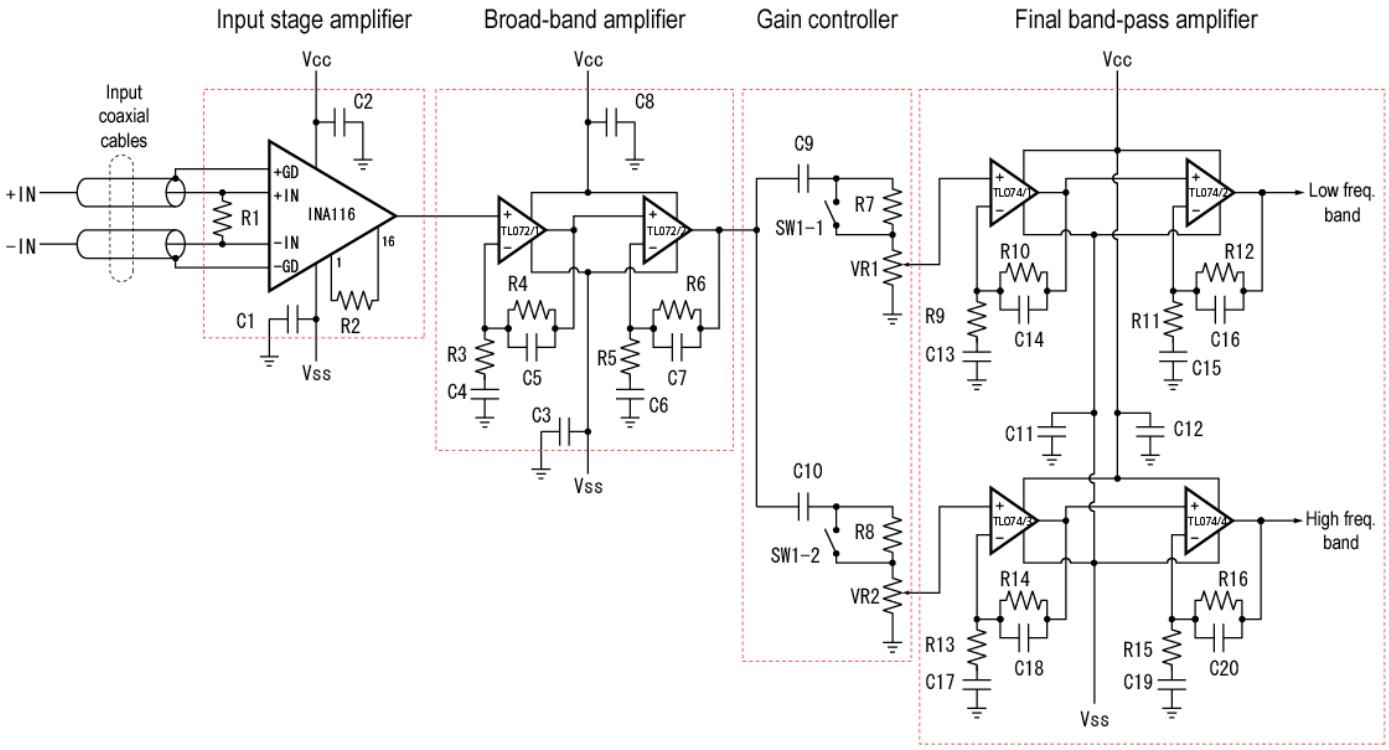


Figure 1. Circuit diagram. See Table 1 for the list of the electronic parts.

Part Number	Description (Manufacturer & Part#)	Package & Size
R1	33MΩ resistor	axial lead
R2	2.7kΩ resistor	SMD 0402
INA116	Instrumentation amp. (Burr Brown, INA116)	16 pin PDIP
C1, C2	1μF capacitator	SMD 1206
C3, C8, C11, C12	0.1μF capacitator	SMD 0603
C4, C6, C13, C15	1μF capacitator	SMD 0805
R3, R5, R9, R11, R13, R15	150kΩ resistor	SMD 0402
C5, C7	33pF capacitator	SMD 0402
R4, R6	1.3MΩ resistor	SMD 0402
TL072	dual op. amp. (Texus Instruments, TL072)	8 pin PDIP
C9, C10	10μF capacitator	SMD 0805
R7, R8	910kΩ resistor	SMD 0402
VR1, VR2	100kΩ variable resistor (Bourns, 3310C-0-0-1-104L)	-
C14, C16	470pF capacitator	SMD 0402
R10, R12, R14, R16	1MΩ resistor	SMD 0402
C17, C19	3.3nF capacitator	SMD 0402
C18, C20	47pF capacitator	SMD 0402
TL074	quad op. amp. (Texas Instruments, TL072)	14 pin SOIC
Power Switch	DPDT toggle switch (NKK Switches Inc., A22-HV)	-
SW1	dual SPST toggle switch (COPAL Inc., CFP-0202C)	-

Table 1. List of the electronic components in Figures 1 and 2. PDIP – plastic dual inline package; SOIC – small outline IC; SMD – surface mount devices. The sizes are in the EIA standards.

from the same electrode (e.g., local field potential and spiking activity of neurons). The gain of this stage is set to 58.8. As in the broad band amplifier, the pass band of each amplifier can be changed by replacing the capacitors (C13-C20), again without affecting the gain. In our present study, we set the passband of these filters to 1-340Hz (for surface EMG, EEG and local field potential) and 320Hz-3.4kHz (for neuronal action potentials).

The final gain of this amplifier could be independently controlled for the low and high frequency outputs, each between 0 to x107,000 when the toggle switch (SW1) is on, and 0 to x10,700 when it is off.

Printed circuit board (PCB)

Figure 2 illustrates the design of the PCB of this amplifier. Special care was taken to minimize the noise in the ground and the power supply. Thus, we used a ground plane to minimize the ground impedance, taking advantage of the PCB technology. Also, each IC has its own bypass capacitors to suppress the noise in the power supply. The PCBs were fabricated by sending the gerber files (available at www.neurophysiology.med.tohoku.ac.jp/work/work08e.html) to a PCB manufacturer. The surface mount components used in this amplifier are difficult to solder by hand, but some companies (e.g., www.pad2pad.com, www.firstchoiceassembly.co.uk) accept orders for PCB assembly as well as fabrication. We ordered the P-ban.com (www.p-ban.com) for the fabrication and assembly of the PCB. The end product was a little smaller than a credit card (80 x 50 mm).

Assembly

Figure 3*a* shows an assembled amplifier. We used variable resistors (VR1, VR2) with shaft and threaded bushing, by which each unit was attached to a chassis or a panel (Figure 3*b*). When multiple units were mounted, the DC power supply and the ground connection were provided via the stack-through connectors inserted in the through holes of the PCB (Figure 3*c*). The stack-through connectors must be long enough to allow at least 15 mm of space between the units. If a proper connector is not available, PC104 type connectors could be used as substitutes.

Input cables

To provide the best protection against the electrostatic noises, shielded cables were used for input, and the shields were connected to the guard drive pins of the INA116.

Power supply

This amplifier works on a DC power supply ranging from ±4.5 to ±18 V. Because its current consumption is rather low (typical value is 10.4 mA per unit on ±15 V), it could be operated either by two 9V batteries or by AC-powered DC power supply units. If the latter is used, use of linear power supply is preferable than switching-mode power supply because the former provides power with fewer ripples than the latter. Throughout this study, we used an AC-powered, regulated DC power supply (±12 V) because it provides stable voltage over hours of continuous operations.

RESULTS

Cost

Table 1 lists the electronic parts for each unit. The cost of the parts was approximately 35 USD. The fabrication and assembly of the 20 units of PCB cost approximately 1000 USD. Therefore, the total cost of each unit was approximately 85 USD. The cost was significantly less than that of the popular commercial amplifiers.

Performance in electrically unshielded environment

We tested the noise resistance of this amplifier in

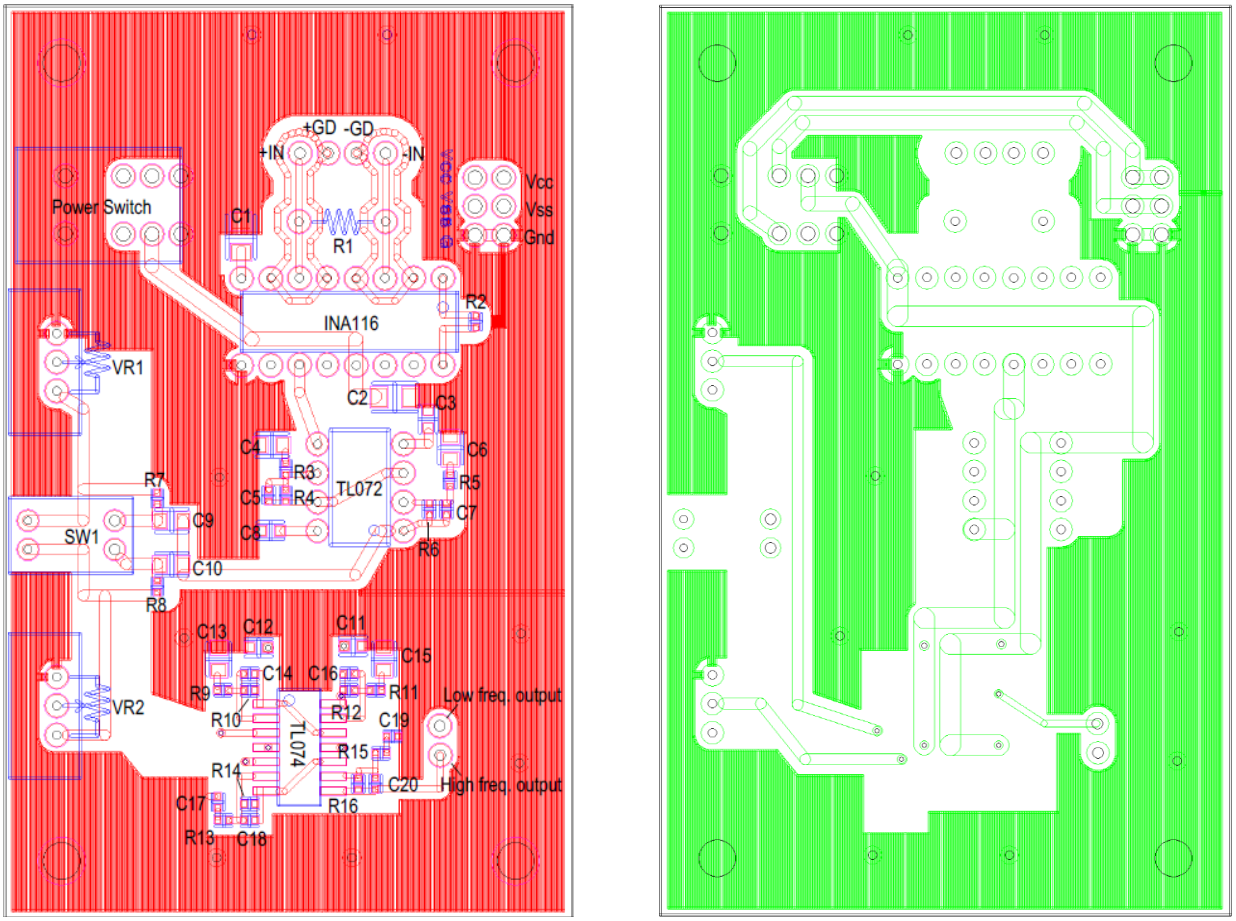
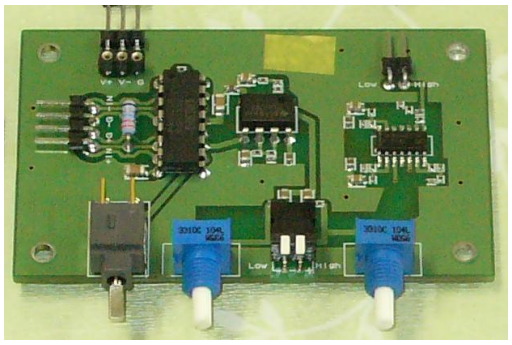


Figure 2. The printed circuit board (PCB). *Left*: the front surface and the electronic parts. Through holes labeled as +IN and –IN are, respectively, connected to positive and negative inputs, and +GD and –GD are connected to their shield conductors. Vcc and Vss are positive and negative DC power supplies. *Right*: back surface. The part numbers correspond to those in Figure 1.

A



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C

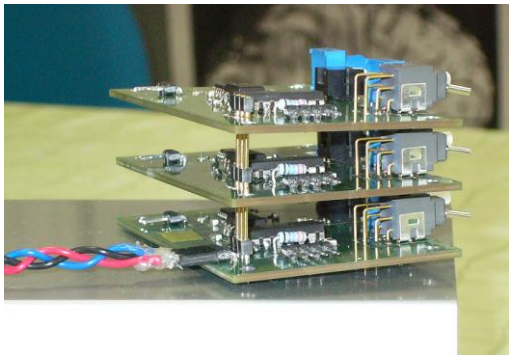

Figure 3. A). An assembled amplifier. All the electronic parts in Table 1 are mounted on the PCB. *B*). The variable resistors (VR1 and VR2) used for gain control have threaded bushing, by which each amplifier is bolted to the chassis. The toggle switch to select the max gain (SW1 of Figure 2) can be accessed through the hole in front. *C*). Multiple PCBs are connected by stack-through connectors, which provides DC power and ground connection.

electromyographic (EMG) activity recording. EMG was recorded by surface electrodes from flexor carpi radialis muscle during wrist flexion movements, then passed to the low frequency filter of our amplifier (1-340Hz, gain x5,000). The recording was done under a fluorescent light without any electrostatic shield. For comparison, we recorded EMG by a commercial amplifier (Nihon Kohden MEG-6100, passband was set to 1.5-300Hz, gain x5000) under the same condition.

Figure 4*a* is the raw EMG recorded by our amplifier and its wavelet spectrum. The EMG was free of the AC power noise (50Hz), and the baseline was stable both before and during the flexion movements. In contrast, the EMG recorded by the commercial amplifier was heavily contaminated by the AC noise (Figure 4*b*). The signal to noise ratio, as calculated by the ratio of the root mean square value of the EMG amplitude during the resting and movement periods, was 16.09 for our amplifier and 1.05 for the commercial model. When the latter amplifier's hum filter was engaged, the AC noise was significantly suppressed, but it attenuated the 50Hz component of the EMG (Figure 4*c*). In addition, the baseline of the EMG recorded by the commercial amplifier tended to fluctuate due to the motion artifact, irrespective of whether the hum filter was used or not.

Performance in electrophysiological experiments

We tested this amplifier in the recording of various bioelectrical signals, including the surface EMGs, electro-encephalography (EEG), local field potential (LFP) and action potentials of single neurons. Figure 5 shows the local field potential (LFP) and the spiking activity of neurons in the cerebral cortex of a primate. These signals were recorded simultaneously through the same metal electrode while the animal was performing key press movements in response to an auditory signal. As shown in the figure, the amplifier recorded both signals clearly.

Application to students' training on EEG recording

We developed a computer-controlled, closed-loop stimulation and measurement system for students' training on electroencephalographic recording (Figure 5*a*). This

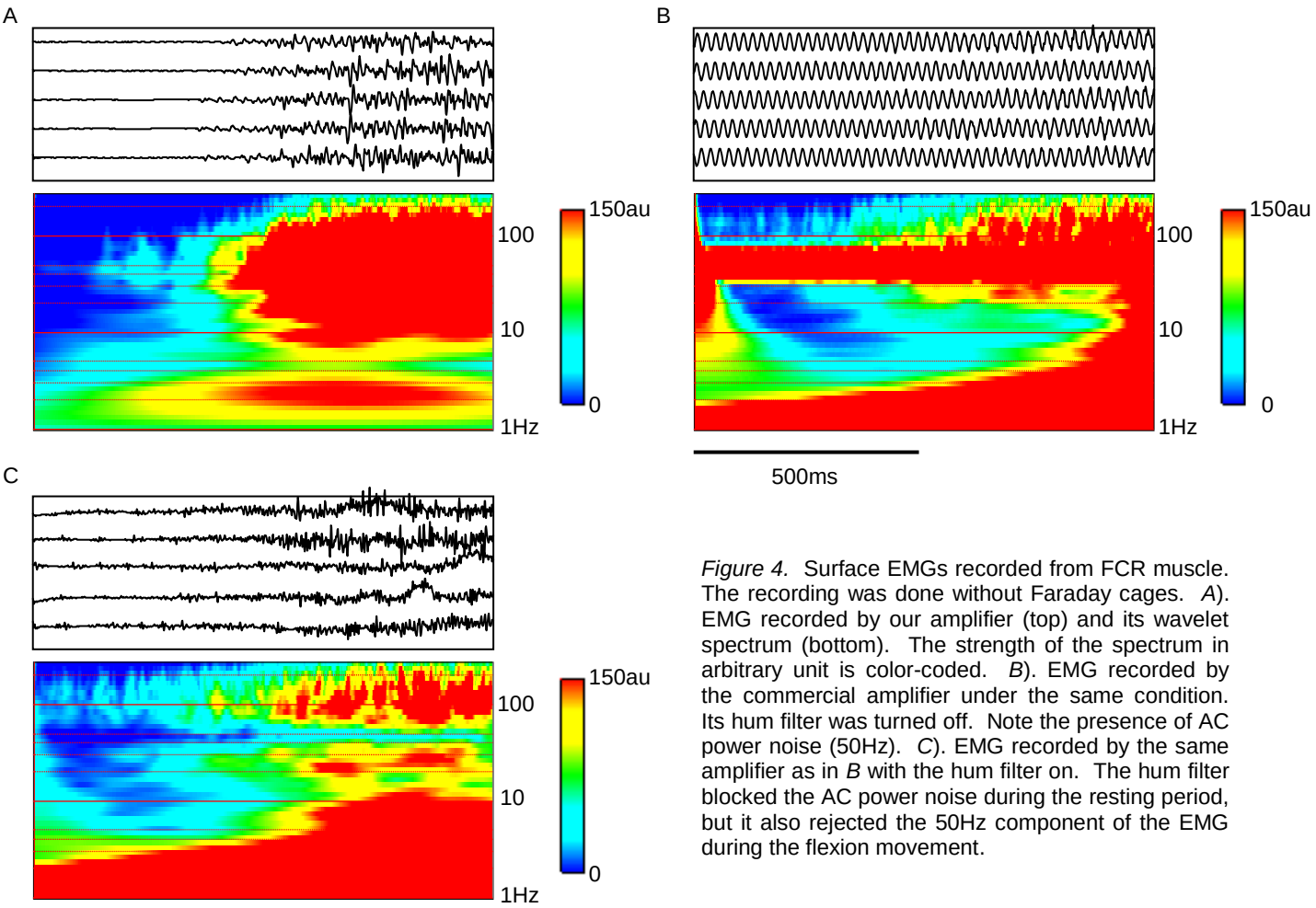

Figure 4. Surface EMGs recorded from FCR muscle. The recording was done without Faraday cages. *A*). EMG recorded by our amplifier (top) and its wavelet spectrum (bottom). The strength of the spectrum in arbitrary unit is color-coded. *B*). EMG recorded by the commercial amplifier under the same condition. Its hum filter was turned off. Note the presence of AC power noise (50Hz). *C*). EMG recorded by the same amplifier as in *B* with the hum filter on. The hum filter blocked the AC power noise during the resting period, but it also rejected the 50Hz component of the EMG during the flexion movement.

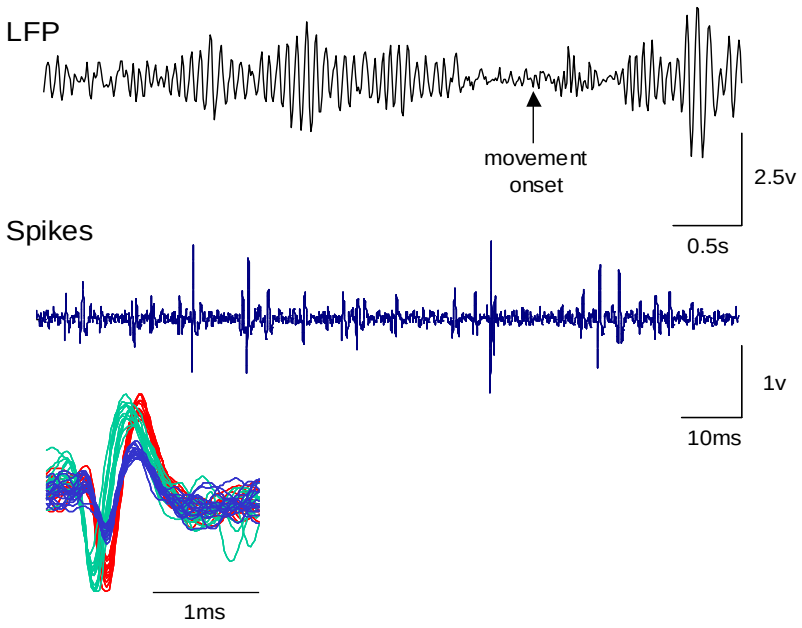

Figure 5. Neural activity of different frequency bands simultaneously recorded from the same electrode. These signals were recorded through a glass-coated elgiloy electrode (impedance = 0.8MΩ at 1kHz) from the supplementary motor area of a primate while it was performing key press movements triggered by an auditory signal. *Top*. Local field potential (LFP) recorded by the low frequency band filter of our amplifier. Gain = x20,000. *Middle*. Action potentials recorded by the high frequency band filter. Gain = x6,000. *Bottom*. Waveforms of individual action potentials of three identified neurons shown in different colors.

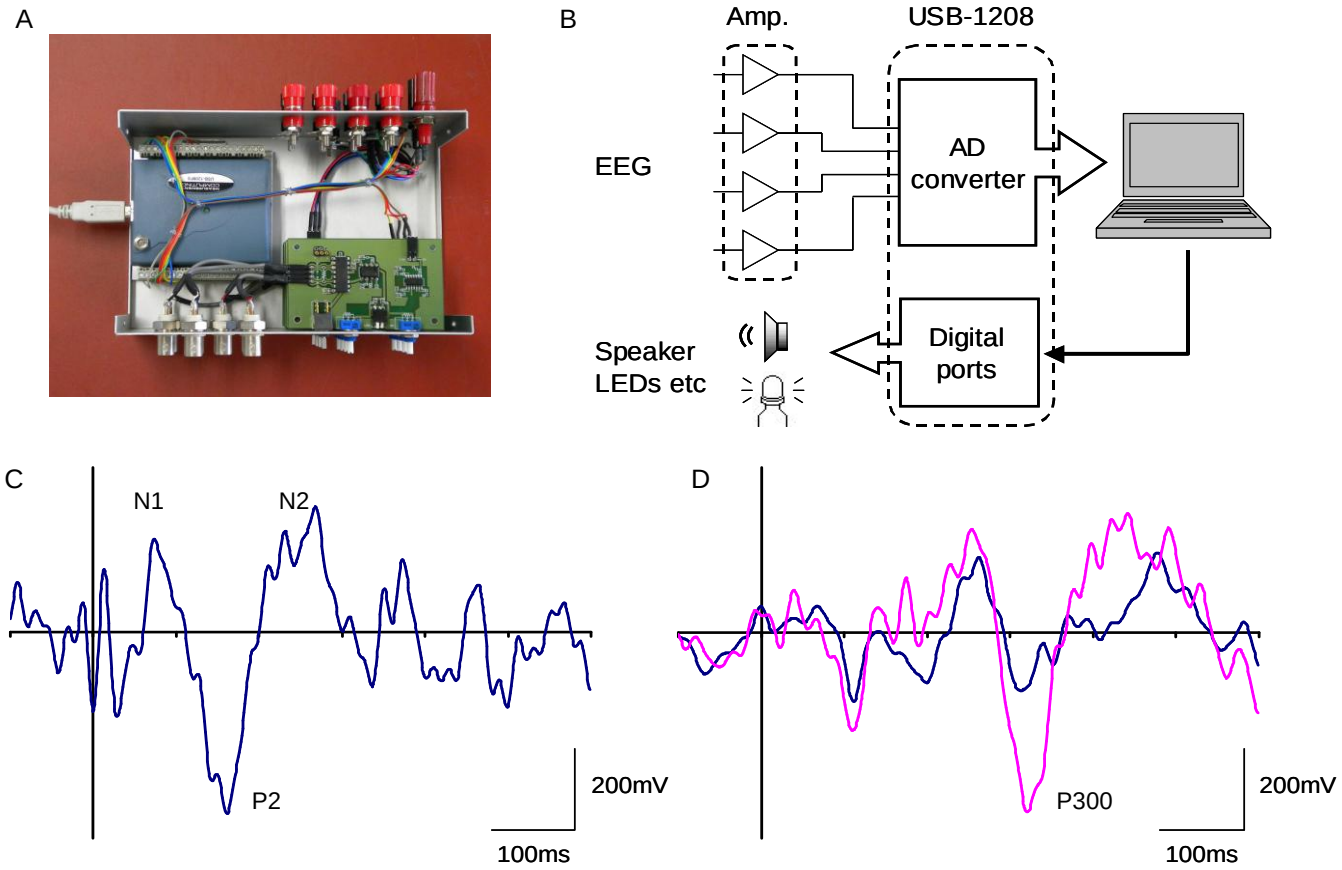

Figure 6. EEG recording system. *A*). The stimulation and recording module. Its lid is removed. The whole device measures 180 x 120 x 75mm. *B*). Block-diagram of the system. The EEG is recorded by the amplifiers, then digitized by the AD converter and sent to the computer. The computer provides the sensory stimuli through the digital ports while continuously saving the data to the disk. *C*). Auditory evoked potential recorded by this system. N1, P2 and N2 components are visible. The vertical line corresponds to the stimulus onset. The up and downward deflections correspond to negative and positive signals, respectively. *D*). P300 of the visually evoked potential. Legends are the same as in *C*. The blue and magenta lines are the responses to standard and target stimuli, respectively.

system incorporated four of these amplifiers, and a multifunction data acquisition device that featured an AD converter and digital outputs (USB-1208FS, Measurement Computing, MA). The USB-1208FS was used to digitize the EEG recorded by the amplifier. It also drove the instruments (speakers, LED, etc.) to present sensory stimuli (Figure 5*b*). A computer program was written to control the sensory stimuli presentation, as well as for the acquisition, storage, and online monitoring of event-related potentials. Figure 5*c* and *d* are, respectively, the long latency auditory evoked potential and P300 of visually evoked potential recorded by this system. Our students succeeded in recording these signals with clarity despite the fact that the recording was done in a room without electric shield.

DISCUSSION

The amplifier described in this paper was developed for both students' training and research by faculty members. In addition to its affordability compared to commercial amplifiers, it was designed to function properly even under noisy environments. The latter feature makes it suitable for both students' training as well as classroom demonstrations that often lack adequate electric shielding.

In practice, electrophysiological recording is frequently impaired by electromagnetic interferences. The most common countermeasure is electrically shielding the recording apparatus and the subjects. However, this requires a dedicated recording room equipped with such shields, which thereby restricts the environments where recording is done. Some commercial amplifiers feature a 'hum filter' that eliminates the AC power noise. A problem with hum filter is that it also attenuates the corresponding frequency band of the original physiological signal, thereby inappropriately filtering out part of the signal (Figure 4*c*). This is problematic because the AC power frequency often overlaps with physiological signals of interest (e.g., EEG, EMG, local field potential). The driven shield inputs of our amplifier effectively blocks the interference without resorting to a hum filter, thus preserving the frequency band of the original inputs. One may argue that electromagnetic interference can be eliminated by simply grounding the shield conductor of the input coaxial cable. This solution works only for relatively short cables. For long cables, even by grounding the shield, the capacitance between the shield and the core conductor remains which leaves the input vulnerable to interference. In addition, such stray capacitance at the input can attenuate the high frequency component of the input signal (Yamamoto et al., 1985). The driven shield input resolves this problem too by canceling the capacitance at the input.

An important issue with electronic devices is safety when they are used with human subjects. Several features make this device unlikely to cause electrical shock. First, the high input impedance of the amplifier (INA116) at the input stage, which directly makes contact with human body, allows little current to flow through the input cables. Second, the PCB of this amplifier is insulated by non-conductive photoresist material. And finally, the DC voltage used for this amplifier (less than ±18 V) is low enough not to cause electrical shock when touched by dry skin. Yet it is advisable to take common safety precautions (e.g., grounding the chassis, avoiding wet conditions). Use of an isolation transformer will also contribute to additional safety.

Several other studies have published the circuit design of neuron amplifiers (Millar and Barnet, 1994; Cheney et al., 1998; Land et al., 2001). However, building an amplifier is a time consuming process. Thus it is impractical to build them in larger numbers by hand wiring. More importantly, the performance of an amplifier depends largely on the circuit layout. Therefore, an amplifier built by hand wiring may not function properly. For these reasons, faculty tend to rely on costly commercial equipments. These problems could be resolved if the optimized circuit layout is freely available. Taking advantage of the printed circuit board, one can reproduce the amplifiers in any number with consistent quality. The resulting product should be useful for both neuroscience education and research that involve electrophysiological recordings.

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