A CAPACITIVE BIOELECTRODE FOR RECORDING ELECTROPHYSIOLOGICAL SIGNALS*

UN BIOELECTRODO CAPACITIVO PARA MEDIR SEÑALES ELECTROFISIOLÓGICAS*

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In this paper we describe a gel-free sensor with on-board electrode design, which capacitive couples to the skin to detect the electrical activity in the body. The integrated sensor is manufactured on a standard printed circuit board within 2.2 cm diameter enclosure that can operate through fabric or other insulation. The electrode includes amplification (60 db gain) and passive bandpass filtering (0.5 to 100 Hz). Active shielding surrounding the sensor plate is used to reduce noise pickup. The input referred noise, measured over the electrode bandwidth is 4 μVrms at 0.2 mm sensor distance, and 16 μVrms at 1.2 mm distance trough two cotton cloths. The bioelectrodes were coupled to the scalp trough hair for EEG signals (with 80 db gain), and coupled to the chest through clothing for ECG signals. The recorded signals show well performance of the designed bielectrode.

En este artículo describimos un sensor libre de gel diseñado con un electrodo en la placa, que se acopla capacitivamente a la piel para detectar la actividad eléctrica en el cuerpo. El sensor integrado fue manufacturado en un circuito impreso estándar dentro de un diámetro de 2.2 cm y puede operar a través del aislamiento de fábrica u otro. El electrodo incluye amplificación (ganancia de 60 db) y un filtrado pasa-banda pasivo (0.5 a 100 Hz). Un escudo activo rodea el sensor para reducir los efectos de los picos del ruido. El ruido a la entrada medido en el ancho de banda del electrodo es 4 μVrms a una distancia de 0.2 mm del sensor y 16 μVrms a 1.2 mm a través de dos capas de algodón. Los bioelectrodos se acoplaron al cuero cabelludo a través del cabello para obtener las señales de EEG (con 80 db de ganancia) y al pecho a través de la ropa para obtener las señales de ECG. Las señales grabadas muestran bien el rendimiento del bioelectrodo diseñado.

PACS: Skin effect, 72.30.+q, Potential models, 12.39.Pn, biopower systems, 88.20.M-, measurement of Impedance, *43.58.Bh, electrical Impedance, 84.37.+q, Noise in biological complexity, 87.18.Tt, in electrical circuits, 07.50.Hp

INTRODUCTION

The time-varying potentials on the surface of the human body result from the summation of many thousand individual sources underlying the skin. These sources are volume sources, which produce ion currents and electric fields through volumetric conductors. The recording of these electrical signals on the body´s surface like ECG, EEG, EOG and EMG, among others, are an important tool for research and clinical purposes, they provide useful information regarding the physiological state of a subject. All of these biopotentials are microvolt level to millivolt amplitude on body surface, and they vary relatively slowly with signal bandwidths from 0.01 to 100 Hz and up to 4 kHz for EMG signals. Amplification of these potentials is not difficult but their detection from tissue or skin is not a straightforward processes. The placement of electrodes on the skin often creates problems such as baseline drifts, motion artifacts, and interference originating from power wiring and radiofrequency sources.

For detecting bioelectric signals from the body surface the standard wet Ag/AgCl electrode is still almost universally used. This electrode requires real charge current contact to the surface of the body by an electrolytic gel. The gel is used as an electrical transducer to convert the ionic current flow in the surface of the skin, into an electron flow which can then be detected by an electronic amplifier. For long recording periods the gel causes skin irritation and discomfort, and it tends to dry out which limits the recording time. Besides, the application area has to be properly cleaned, and sometimes also shaved. Non-contact capacitive electrodes do not require an ohmic connection to the body. Capacitive sensing relies on capacitive coupling, involving the skin and a metal sensor with a thin dielectric film on the contact surface. For body sensor applications, this offers numerous advantages since non-contact electrodes require zero preparation, are insensitive to skin conditions and can be embedded within comfortable layers of fabric. Capacitive electrode technology permits a continuous monitoring of patients for long periods of time, leading to very wearable body sensor network, which can include a wireless data transmission for real patient monitoring [1-6].

When biopotentials are picked up through clothes, coupling

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capacitances are of a few pF, increasing up to hundreds of pF when the dielectric film is directly placed over the skin. In order to ensure a low frequency response and to avoid signal attenuation, such capacitances call for ultra-high input impedance amplifiers ≥ 10^{15} Ω. When using insulating electrodes, a path for the amplifier's bias current must be provided because it cannot flow through the patient, as is the case when wet electrodes are used. The ultra-high input impedance node is susceptible to any stray interference and motion induced artifacts. The design of low-noise amplifiers for insulating electrodes consists in achieving low noise levels in front of source impedances of a few pF, preserving input impedances of 10^{15} Ω. In this paper we report the design and characterization of a low-noise capacitive electrode, as well as some measurements recorded of ECG, EOG, and EEG signals.

BIOLECTRODE DESIGN

Figure 1 shows a simple model of the bioelectrical potential on the skin, coupled to the input amplifier through a skin-sensor capacitance *Cc*. For ultra high input resistance amplifiers ($Ri \geq 10^{15}$ Ω), the input impedance *Zi* is dominated by the input capacitance at low frequencies ($\omega RiCi >> 1$), and the circuit has an essentially flat-band response given by $Vi = VsCc/(Cc + Ci)$. For a good coupling $Cc >> Ci$, such that $Vi \approx Vs$, yielding a near ideal detection of the electric biopotential on the skin.

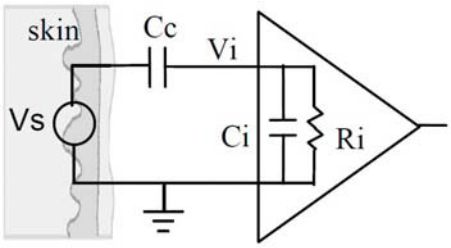

Figure 1. Skin-amplifier capacitive coupling interface.

Low signal frequencies (1-100 Hz) result in extremely high capacitive source impedance ($1/j\omega Cc$) on the same order as the input impedance of the amplifier ($\approx 1/j\omega Ci$). Therefore, even small variations in the coupling capacitance and distance can lead to large amounts of distortions due to signal attenuation, poor rejection to interferences and motion artifacts. To minimize these undesired distortions, the amplifier input capacitance must be as low as possible. A small amount of positive feedback can be applied to the input node, trough a feedback capacitor, to neutralize the effect the amplifier input capacitance [3, 7, 8]. To operate the amplifier an input bias current must flow into the terminals at the amplifier input. However, in the circuit shown in Figure 1 there is no path to provide the input bias current, and so it charges up the capacitance *Cc*. The result is that the input has a steady voltage drift which can increase with time, saturating the amplifier output. To overcome the drift problem, some stabilization and feedback techniques have been developed, providing the ultra-high resistive path to ground at the input, as required for a properly bipotentials detection [2, 3, 9].

The designed electrode consists of two round standard printed circuit boards (PCBs) of 26 mm diameter, which are electrically connected by one cooper surface of each PCB, forming a shielding plane between the boards. A photograph of the assembled electrode is shown figure 2. Biopotentials on the skin are sensed through a 17 mm diameter cooper disc at the bottom of the lower PCB, covered with solder mask for electrical insulation. The sensor disc is shielded from external interference by an outer cooper ring, which is connected to the shielding plane above the sensing surface. The upper side of the second PCB contains the front end amplifier as well as gain amplifiers and passive components. As front end amplifier we use the electrometer grade operational amplifier LMP7721 (National Semiconductor) configured as unit-gain voltage follower. This device is designed with an input CMOS technology (patent pending) suitable for ultra-high impedance sensors. The LMP7721 has an ultra low input bias current of 3 fA at 25°C, a low input voltage noise of 6.5 nV/√Hz, and an input current noise of 0.01 pA/√Hz. In figure 2, an active shield is implemented by surrounding the critical input node by driving the sensor shield from the follower voltage output. Because it is actively driven to duplicate the input voltage, it avoids parasitic capacitance division of signal. Active shield guards high-impedance input from interference by other sources [3, 4, 8].

It was found experimentally that no external input biasing network is necessary with the LMP7721. The critical input node was left floating and the inputs consistently charge and stay within the rail-to-rail input range during use. The input can reliably self-bias purely through the device's internal ESD protection structure and other parasitic leakages. This achieves optimal performance since any bias network necessarily adds noise and degrades the input impedance. Figure 2 shows the bioelectrode schematic, where the lack of a bias network at sensor input, results in an undefined (although full usable) DC operating point. To remove this offset as well as low frequency noise drift, a passive RC high pass filter with a corner frequency of 0.5 Hz is used to center the signal around *Vref*. To amplify the signal we use the micro power precision dual operational amplifier LTC 6078 (Linear Technology). Each operational in the package is configured as non-inverting amplifiers, approximately with a gain of 20 for the first stage and 50 for the second stage. A passive RC low pass filter with a corner frequency of 100 Hz is used at output of the first amplification stage, to obtain an overall electrode bandwidth of 0.5 – 100 Hz with 1000 gain.

CHARACTERIZATION

To characterize the performance of the sensor at various distances and dielectric mediums, a frequency sweep was applied to a cooper disc from a precision function generator (Fluke PM5139). This source electrode was coupled to the sensor trough the smallest distance that of the solder mask, which is estimated to be 0.2 mm. Larger distances between the source electrode and the sensor plate, are formed with spacers

made of acrylic plastic and cotton fabric. The output voltages of the first amplification stage (gain of 20) were recorded with a digital oscilloscope (Tektronix TDS 2022B), and the signals measured were processed with a LabVIEW® program. Figure 3 shows the measured gains over the range of frequencies, for three different distances between the signal generator electrode and the sensor plate: 0.2 mm (solder mask), 0.6 mm (cotton fabric), and 1.2 mm (cotton fabric). As the distance is increased, the input coupling capacitance is reduced, as is the gain, and the input noise is increased.

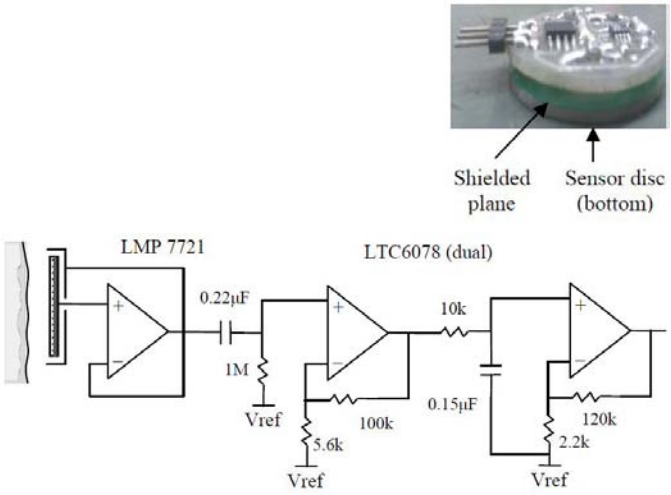

Figure 2. Schematic and photograph of the capacitive bioelectrode.

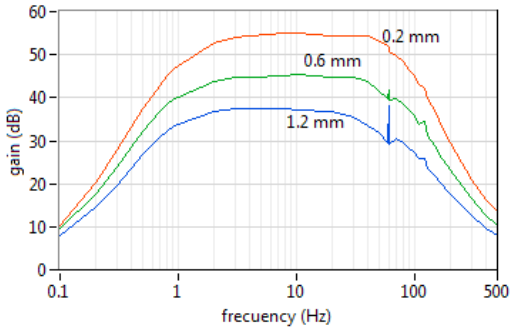

Figure 3. Gain versus frequency for three distances: 0.2 mm (solder mask), 0.6 mm and 1.6 mm (cotton).

Noise levels were measured by removing the signal generator and connecting the source electrode to the reference voltage of the circuit. The input referred noise is computed by taking the total output noise spectrum and dividing by the previously found mid-band (10 Hz) gain at the same separation distance. The calculated referred to input voltage noise of the amplifier in the frequency band is about 4 µVrms for 0.2 mm and 16 µVrms for 1.2 mm (cotton). The best estimated SNR is approximately of 72 at 0.2 mm (solder mask). For cotton the SNRs are approximately 35 at 0.6 mm, 17 at 1.2 mm, 4 at 1.8 mm, and degrades significantly for further sensing distances.

PHYSIOLOGICAL RECORDINGS

When electrodes are placed on the body, common mode voltage (*Vcm*) results between the average potential of the body to the amplifier reference point. An effective way to reduce *Vcm* is by means of a common mode negative feedback, known as driven right leg circuit (DRL) [9]. We assembled a DRL circuit with the dual operational amplifier LTC6078 (Linear Technology) with -40 dB gain. This gain attenuates the *Vcm* by approximately 0.01.

For ECG experiments two electrodes were pressed against a subject's chest using an adhesive band. One was located above the heart while the second was placed over the opposing ribcage. The differential voltage between the two electrodes is recorded. When the recording is made over two cotton shirts (1.2 mm), some 60 Hz line noise is introduced as a result of capacitive mismatch due to the larger separation distance and corresponding smaller coupling capacitance, but the a signal caused by the beating heart is clearly visible, as figure 4 shows.

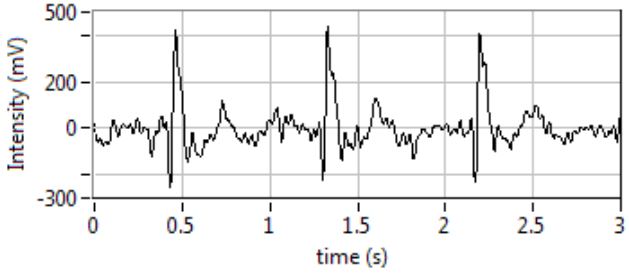

Figure 4. ECG recording taken over a subject´s chest with the electrode placed through two cotton shirts.

For EOG experiments two electrodes were manually placed on the surface of the skin, each one beside of each eye (points L-R in the 4/5 system). Figure 5 shows difference voltage signal between the two electrodes resulting from left and right eye movements. Similar results were obtained placing the electrodes in points U-D (up and down of each eye) for up and down eye movements.

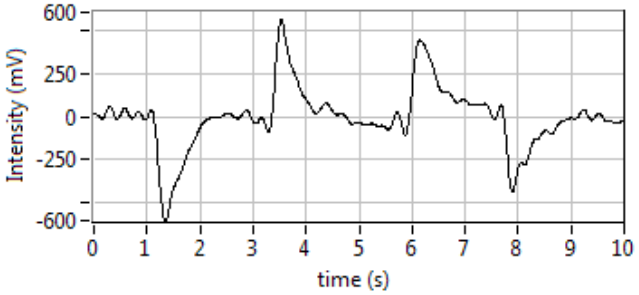

Figure 5. EOG signal recorded with electrodes placed over points L-R in the 4/5 system, showing deflections resulting from left and right slow eye movements.

EEG experiments were performed with two electrodes (overall gain of 5000) pressed against a subject's head using a headband. They were placed through hair approximately in the 'P3' and 'O1' positions referring to the international standard 10-20 system. The power spectral densities of the data recorded with open and close eyes are shown in Figure 6. Increased power in the alpha band of frequencies around 10 Hz can clearly be observed when the eyes are closed, which disappears when the eyes are opened, commonly observed in EEG experiments [4, 5, 8].

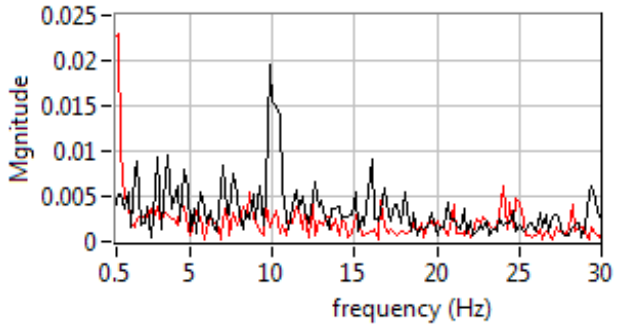

Figura 6. Spectra of EEG signals showing an alpha band with open eyes (black trace), which is not present when eyes are closed (red trace).

CONCLUSION

We have presented a non-invasive capacitive electrode for biopotentials sensing over body´s surface. Biopotentals are capacitive coupled to a cooper disc sensor made on standard PCB, which integrates an ultra-high input impedance front-end, based on the latest CMOS technology operational amplifier LMP7721. The high input impedance level (1015 Ω parallel 0.2 pF) means that the sources of body electrical signals are not loaded down by the coupling capacitance. The signals attenuation is less than 2% for 10 pF coupling capacitance. The electrode design includes two amplification stages with 60 dB overall gain, and two passive filter with 0.5 – 100 Hz bandwidth. In non-contact sensing the effect of capacitive gain errors due to variable strength coupling, introduces errors in the detected signals and increases input noise. In our experiments we found problems to sense EEG signal through the hair, due the inherent sensitivity to motion/friction artifacts. Future work involving active bootstrapping is expected to mitigate these effects The signals recorded show a well performance of the bioelectrode designed.

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