Technical note

A simple and reproducible capacitive electrode

Enrique Spinelli\textsuperscript{a,b,*}, Federico Guerrero\textsuperscript{a,b}, Pablo García\textsuperscript{a}, Marcelo Haberman\textsuperscript{a,b}

\textsuperscript{a}LDEI – Departamento de Electrotecnia, Universidad Nacional de La Plata, CC 91 (1800) La Plata, Argentina
\textsuperscript{b}Correo de Investigaciones Científicas y Técnicas (CONICET), Argentina

\textbf{ARTICLE INFO}

\textbf{Abstract}

Capacitive Electrodes (CE) allow the acquisition of biopotentials through a dielectric layer, without the use of electrolytes, just by placing them on skin or clothing, but demands front-ends with ultra-high input impedances. This must be achieved while providing a path for bias currents, calling for ultra-high value resistors and special components and construction techniques. A simple CE that uses bootstrap techniques to avoid ultra-high value components and special materials is proposed. When electrodes are placed on the skin; that is, with coupling capacitances $C_s$ of around 100 pF, they present a noise level of 3.3 $\mu$V$_{\text{rms}}$ in a 0.5–100 Hz bandwidth, which is appropriate for electrocardiography (ECG) measurements. Construction details of the CE and the complete circuit, including a fast recovery feature, are presented. © 2015 IPEM. Published by Elsevier Ltd. All rights reserved.

\section{1. Introduction}

Capacitive Electrodes (CE) do not require the use of electrolytes. They acquire biopotentials through a dielectric layer by just placing them on the skin \cite{1,2} clothing \cite{3,4}, or without any physical contact with the patient \cite{5}. They avoid skin irritation, are simple to install, and appropriate for long-term patient monitoring.

The general scheme of a CE measurement set-up is shown in Fig. 1(a) and its equivalent circuit in Fig. 1(b) \cite{6}. This is reduced to a simple AC-coupled amplifier \cite{7}, but for ‘coin-size’ CEs, coupling capacitance $C_s$ can be as low as a few tens of pF (10–30 pF), when biopotentials are picked up through clothing, or hundreds of pF (100–300 pF), when CEs are placed on the skin with a dielectric film \cite{4}. In order to achieve the very low cut-off frequencies that biomedical signals require for an ECG \cite{8}, these small $C_s$ values demand bias resistors $R_0$ as high as 0.1–1 TΩ. Electronic requirements relax for large $C_s$ values, when very thin dielectric layers \cite{2,7} or large-area CEs are used \cite{9}.

Since CEs must work with ultra-high impedances, they are vulnerable to electric-field interference and sensitive to circuit leakages, requiring high-quality Printed Circuit Board (PCB) substrates, and careful guarding and shielding techniques to keep unavoidable leakages and couplings under control \cite{3,4}. To deal with this, a practical CE circuit includes a guard-driver, and a neutralization circuit to reduce the effects of PCB and amplifier input capacitances. Details of how these sub-circuits work can be found in \cite{6}.

Capacitive electrodes present noise levels greater than their ‘wet’ counterparts. The noise Power Spectral Density (PSD) $e_0$ for frequencies above the cut-off frequency $f_N$, is given approximately by \cite{6}:

\begin{equation}
e_0 \approx e_0^n (f_N/f)^2 + e_0^r R_0^2 (f_N/f)^2 + \alpha e_0^r.
\end{equation}

where $e_0$, $e_0^n$ denote the Operational Amplifier’s (OA) current and voltage noise respectively, $e_0^r R_0$ is the thermal noise of $R_0$, factor $\alpha$ represents effects of neutralization and guarding circuits that amplify $e_0^n$, and $f_N$ denotes the cut-off frequency:

\begin{equation}
f_N = (2\pi R_0 C_s)^{-1}.
\end{equation}

Eq. (1) shows that reducing $f_N$ decreases the electrode noise PSD. Then, the noise cut-off frequency $f_N$ must be set below the signal pass-band (as far below as possible) in order to limit the effect of low-frequency noise \cite{6}.

Expression (1) does not include noise sources outside the CE itself, such as those produced by clothing or skin layers \cite{4}. Replacing $e_0^r$ by the Nyquist expression ($e_0^r = 4kT R_0$) and $f_N$ by (2), results in:

\begin{equation}
e_0 \approx \frac{kT}{(\pi C_s f)^2} \frac{1}{R_0} + \frac{e_0^r}{(2\pi C_s f)^2} + \alpha e_0^r.
\end{equation}

As can be observed in (3), to reduce $e_0$, a low noise OA should be used, and — less obviously — the value of $R_0$ should be the highest possible \cite{3}. Resistors $R_0$ of the order of TΩ are desirable, but they are not easy to obtain and handle. Some techniques to achieve ultra-high value resistors have been proposed, such as using reverse polarized diodes $\cite{2,10}$ and ‘gimmick’ resistors implemented from insulated cables’ leakages \cite{6}. Moreover, these high-value $R_0$ values impose large time constants to discharge $C_s$ when
times greater than that of a ‘real resistor’ of the same value. Replacing in (1) \( e_{RB} \) by the \( e_{RBEQ} \) expression given by (5), and \( f_N=(2\pi R_{RBEQ}C_S)^{-1} \), Eq. (1) becomes:

\[
e^{2}_o \approx kT \left( \frac{1}{(\pi C_S)^2 R_1} + \frac{\varepsilon^2}{(2\pi C_S)^2} + \alpha e^2_n \right);
\]

which corresponds exactly to (1) with \( R_B=R_1 \). Hence, the bootstrapping multiplies \( R_1 \) allowing to achieve a very low cut-off frequency \( f_N \) and proper transient responses, but it also amplifies the noise of \( R_1 \). As a result, the CE noise is the same as using \( R_1 \) in place of \( R_B \). However, an \( R_1 \) of a few \( \Omega \) (a high but accessible value) is high enough to acquire good-quality ECG signals, even picking them up through cotton clothes. The circuit herein proposed implements \( R_{RBEQ}=100 \Omega \) by \( R_1=10 \Omega \), and a bootstrap ratio \( R_2/R_3=10 \). This \( R_1 \) value is enough to achieve time constants of a few seconds and allows building the CE with standard FR4 PCB material, which has a superficial resistivity (SR) of around 50 \( \Omega \)\text{cm}, instead of using more expensive substrates materials such as Teflon\textsuperscript{TM}, that present SR values of 1 \( \Omega \)\text{cm} and more.

2.1. Description of the proposed circuit

The complete circuit shown in Fig. 3(a) includes a guard driven by the output of OA1, and an input capacitance neutralization circuit implemented by OA2, through capacitor \( C_e \), according to [6]. The corresponding PCB design is shown in Fig. 3(b), where it can be observed that the neutralization capacitance \( C_e \) is implemented by a PCB area. A dual low-bias current operational amplifier OPA2320 (by Texas Instruments\textsuperscript{TM}) was used. The non-inverting input of OA1, — the most vulnerable node of the circuit — is not soldered to the PCB, but bent upwards, and capacitor \( C_e \) and \( R_1 \) are soldered directly to it [13]. No solder-mask was used, in order to reduce superficial leakages.

The proposed CE itself, without additional elements, provides a ‘fast recovery’ mechanism to restore the baseline when high-amplitude artefacts saturate the amplifier. The circuit time constant in normal operation is \( R_{RBEQ}C_e \), but short-circuiting the output of OA1 (it must be output-protected), reduces it to \( (R_1+R_3)C_e \), thus providing a way to discharge \( C_e \). The recovery of the baseline is not as fast as using the circuit proposed in [14], but is much simpler to implement.

Using resistors \( R_1 \) of around 10 \( \Omega \) and a low current noise OA as the OPA2320, the noise PSD \( e^2_o \) is dominated by the first term in (6), decreasing with frequency \( f \) according to:

\[
e^2_o \approx kT \left( \frac{1}{(\pi C_S f)^2} + \frac{\varepsilon^2}{(2\pi)^2} + \alpha e^2_n \right);
\]
The total noise that is obtained by integrating (7) in a bandwidth from $f_1$ to $f_2$, for $f_2 > f_1$ becomes independent of $f_2$ [6] and is given by:

$$E_0^2 \approx \frac{kT}{(\pi C_S)^2 R_1 f_1}.$$  \hspace{1cm} (8)

Considering $R_1=10$ GΩ, $T=300$ K, $C_S=100$ pF and $f_1=0.05$ Hz, the total noise $E_0$ results in $9 \mu V_{\text{RMS}}$, which is a reasonable value for ECG signals. If $f_1=0.5$ Hz is considered, the noise is limited to $2.9 \mu V_{\text{RMS}}$.

3. Experimental results

The circuit of Fig. 3(a) was built with a diameter of 25 mm and shielded. At first, a sinusoidal signal of ± 100 mV, 1 kHz was applied through a 10 pF capacitor working as $C_S$, and neutralization was adjusted by the trimpot $T_1$ to a unity gain. Then, the noise PSD was measured for $C_S=10$ pF and $C_S=100$ pF, resulting in the curves presented in Fig. 4. The CE total noise in the bandwidth 0.5–100 Hz for $C_S=100$ pF, obtained by integrating the respective PSD, is $3.3 \mu V_{\text{RMS}}$. This value is a little higher than the $2.9 \mu V_{\text{RMS}}$ predicted by (8), because of additional noise sources not considered in this equation, such as OA current noise and other effects that amplify the voltage noise of the OA [6].

Finally, the CE was insulated with a $50 \mu \text{m}$ auto-adhesive polypropylene film, and real ECG signals were acquired from a volunteer. Records were performed simultaneously by two CEs placed on the subject’s chest, using a pair of standard disposable wet electrodes (3M™2223). Signals were acquired using an 8-channel biopotential acquisition system, based on the IC ADS1298 of Texas Instruments™. Monopolar channels were used, in order to verify that each electrode worked properly, and bipolar (differential) signals shown in Fig. 5 were obtained digitally by subtraction.
4. Conclusions

By using bootstrapping it is possible to implement CE avoiding the use of ultra-high bias resistors, simulating them with moderate value ones, but the CE noise is the same as when using the circuit’s higher value resistor as a bias path. As can be observed in Fig. 5, the proposed capacitive electrode allows ECG signals to be acquired with a good signal-to-noise ratio, even picking them up through clothing.

The proposed CE does not require the special substrates and fabrication techniques needed in [3]. It does not demand ultra-high value resistors as the one presented in [6], and provides a fast recovery feature with a simpler circuit than those in [14] or [15]. The noise level of the CE is slightly higher than that in its previous version [6], but it is easier to build and replicate. Complete circuits and construction details were provided, thus placing this work within a reproducible research framework.

Conflict of interest

No conflict of interest.

Acknowledgments

This work has been funded by the Universidad Nacional de La Plata (UNLP) by Project I-167 and Agencia Nacional de Promoción Científica y Técnica through Project PICT-2012/0037. The authors also acknowledge the technical support of Sergio Rodriguez.

The experimental setup of this work corresponds to the project “Instrumentation and Control for neuroprosthesis”, approved in February 2012 by the Bio-Ethics Committee of the Universidad Nacional de La Plata.

Appendix. Noise analysis of the bootstrap circuit

The simulated resistor $R_{EQ}$ presents a noise voltage greater than that of a real resistor of the same value. To estimate this, the noise of each resistor and OA composing the circuit in Fig. 2(b) must be considered. The voltage source $e_n$ represents the OA voltage noise and $e_{R1}$, $e_{R2}$, $e_{R3}$ are the noise of resistors $R_1$, $R_2$ and $R_3$, respectively. The amplifier current noise $i_n$ is not included, because the effect that $i_n$ produces on $R_{EQ}$ is the same as that it produces on a real $R_3$. This is already considered in the CE noise analysis that yields (1). Solving the circuit of Fig. 2, the overall noise at the output results:

$$e_n^2 \approx e_{R1}^2 (1 + R_2 / R_1)^2 + e_{R2}^2 (1 + R_3 / R_2)^2 + e_{R3}^2 (1 + R_3 / R_2)^2;$$

and replacing resistors' noise PSDs by the Johnson–Nyquist formula ($e_n^2 = 4kT$):

$$e_n^2 \approx e_{R1}^2 (1 + R_2 / R_1)^2 + 4kTR_1 (1 + R_3 / R_2)^2 + 4kTR_2 + 4kTR_3 (1 + R_3 / R_2)^2.$$  (9)

Given that $R_1 \gg R_2$, $R_3 / R_2 \gg 1$ and $e_n \ll e_{R1}$; Eq. (10) can be approximated by:

$$e_{R_{EQ}}^2 \approx 4kTR_1 (R_1 / R_2)^2;$$

which can be written as:

$$e_{R_{EQ}}^2 \approx 4kTR_{EQ} (R_1 / R_2).$$  (11)

References

[1] Griffith M, Portnoy M, Stotts L. Improved capacitive electrocardiogram electrodes for burn applications. Med Biol Eng Comput 1979;17:641-6.
[2] Richardson P, Lopez A. Electrocardiographic and Bioelectric Capacitive Electrode. U S Pat 3 1970:500:823.
[3] Pearce RJ, De Bray A, Clark TD, Prance M, Nock M, Harland CJ. Clippingsdale AJ. An ultra-low-noise electrical-potential probe for human body ECG measurement. Med Sci Technol 2000:11:291–7.
[4] Chi Y, Jung T, Cauwenberghs G. Dry-contact and noncontact biopotential electrodes: methodological review. IEEE Rev Biomed Eng 2010:3:106–19.
[5] Harland C, Clark T, Prance R. Electrical potential probes-new directions in the remote sensing of the human body. Med Sci Technol 2000:13:83-9.
[6] Spinnelli E, Haberman M. Insulating electrodes: a review on biopotential front-ends for dielectric skin-electrode interfaces, Physiol Meas 2010:31:183-98.
[7] David R, Portnoy M, Insulating Electrocardiogram electrodes, Med Biol Eng Comput 1977:15:742-51.
[8] Association for the Advancement of Medical Instrumentation, Ambulatory electrocardiographs American National Standard ANSI/AMI EC38, 1999.
[9] Lee SM, Kim KS, Lim BK, Lim YG, Park KS. Thin and flexible active electrodes with shield for capacitive electrophrocardiogram measurement. Med Biol Eng Comput 2010:48:447–57.
[10] Prance RJ, Clark TD, Prance H, Clippingsdale AJ. Non-contact VLSI imaging using a scanning electric potential microscope. Med Sci Technol 1998:5:1229–35.
[11] Langi S. The noise of input stages with low parasitic capacitance. Med Sci Technol 2001:12:1456–64.
[12] Pallas Areyn R, Colominais J, Rosell J. An improved buffer for Biologic Signals, IEEE Trans Bio Med Eng 1989;36:490–9.
[13] LPC662 Low Power CMOS Dual Operational Amplifier datasheet, Texas Instruments Incorporated 2000, Available at www.ti.com.cn/lit/gpn/lpc662.
[14] Spinnelli E, Haberman M, Garcia P, Guerrero E. A capacitive electrode with fast recovery feature. Physiol Meas 2012:33:1277–88.
[15] Sullivan T, Deiss S, Cauwenberghs G. A Low-Noise, Non-Contact EEG/ECG Sensor, IEEE Conf Biomed Circuits Syst BIOCAS 2007:154–7.