Medical Engineering and Physics 38 (2016) 286–289

Contents lists available at ScienceDirect

Medical Engineering and Physics

journal homepage: www.elsevier.com/locate/medengphy

Technical note

A simple and reproducible capacitive electrode

Enrique Spinelli[a,b,*], Federico Guerrero[a,b], Pablo García[a], Marcelo Haberman[a,b]

[a] LEICI – Departamento de Electrotecnia, Universidad Nacional de La Plata, CC 91 (1900) La Plata, Argentina
[b] Consejo de Investigaciones Científicas y Técnicas (CONICET), Argentina

ARTICLE INFO

Article history:
Received 3 July 2015
Revised 7 November 2015
Accepted 5 December 2015

Keywords:
Insulating electrodes
Active electrodes
Non-contact measurements

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 μV_{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.

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 [1,2] clothing [3,4], or without any physical contact with the patient [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) [6]. This is reduced to a simple AC-coupled amplifier [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 [4]. In order to achieve the very low cut-off frequencies that biomedical signals require for an ECG [8], these small C_S values demand bias resistors R_B as high as 0.1–1 TΩ. Electronic requirements relax for large C_S values, when very thin dielectric layers [2,7] or large-area CEs are used [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 [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 [6].

Capacitive electrodes present noise levels greater than their 'wet' counterparts. The noise Power Spectral Density (PSD) e_O for frequencies above the cut-off frequency f_N, is given approximately by [6]:

$$e_O^2 \approx e_{R_B}^2 (f_N/f)^2 + i_n^2 {R_B}^2 (f_N/f)^2 + \alpha e_n^2, \tag{1}$$

where i_n, e_n denote the Operational Amplifier's (OA) current and voltage noises respectively, e_{RB} is the thermal noise of R_B, factor α represents effects of neutralization and guarding circuits that amplify e_n, and f_N denotes the cut-off frequency:

$$f_N = (2\pi R_B C_S)^{-1}. \tag{2}$$

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 [6].

Expression (1) does not include noise sources outside the CE itself, such as those produced by clothing or skin layers [4]. Replacing $e_{R_B}^2$ by the Nyquist expression ($e_{R_B}^2 = 4kTR_B$) and f_N by (2), results in:

$$e_O^2 \approx \frac{kT}{(\pi C_S f)^2}\frac{1}{R_B} + \frac{i_n^2}{(2\pi C_S f)^2} + \alpha e_n^2. \tag{3}$$

As can be observed in (3), to reduce e_O a low noise OA should be used, and — less obviously — the value of R_B should be the highest possible [3]. Resistors R_B 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 [2,10] and 'gimmick' resistors implemented from insulated cables' leakages [6]. Moreover, these high-value R_B values impose large time constants to discharge C_S when

* Corresponding author. Tel.: +54 2214259306.
E-mail address: spinelli@ing.unlp.edu.ar (E. Spinelli).

http://dx.doi.org/10.1016/j.medengphy.2015.12.006
1350-4533/

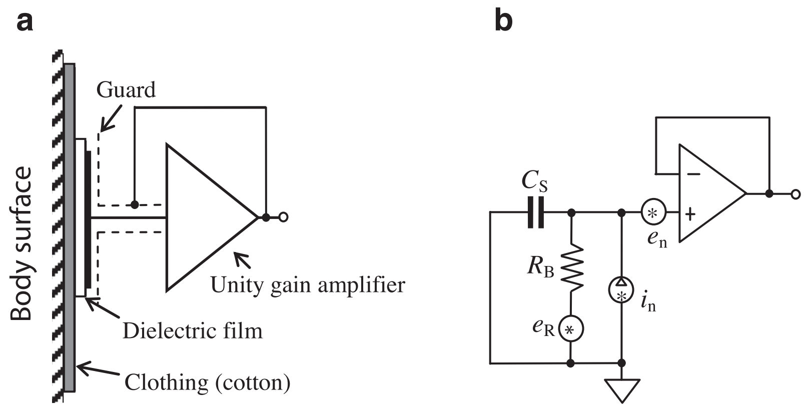

Fig. 1. (a) General scheme of a capacitive electrode and (c) its equivalent circuit including intrinsic noise sources.

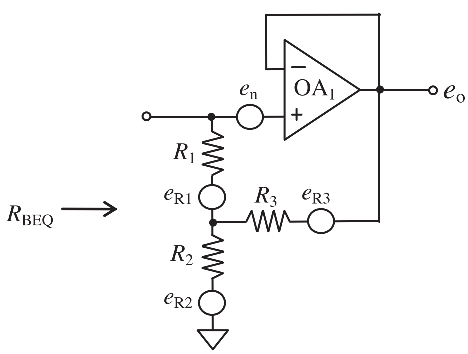

Fig. 2. Bootstrap circuit to high-input impedances by 'multiplying' resistor R_1 value. This figure also includes circuit's noise sources related to resistors and to the operational amplifier.

a high-amplitude artefact saturates the front-end, thus requiring additional circuits elements to recover the baseline in reasonable times.

It is possible to 'simulate' ultra-high resistance values by moderate value resistors, using bootstrapping techniques. A typical circuit, shown in Fig. 2, works as an equivalent resistor R_{BEQ} given by:

$$R_{BEQ} = R_1 + R_2 + R_1R_2/R_3. \tag{4}$$

In general, the last term in (4) is dominant and $R_{BEQ} \approx R_1R_2/R_3$: the bootstrap circuit increases R_1 value by R_2/R_3 times. However, this technique increases resistor R_1 value at the expense of increasing OA's noise and offset voltage effects [11]. Voltage offset amplification can be avoided by replacing R_3 with a capacitor [3], but it introduces a singularity in the transfer function. Thus, the input impedance becomes inductive [12], and it is difficult to fulfil the strict transient response that biomedical standards demand [8]. Nowadays, OAs suitable for CEs with low offset input voltages (i.e. the OPA320 of Texas Instruments) are available, and the fully resistive circuit of Fig. 2 is feasible. Herein, a very simple CE based on this circuit, is proposed.

2. Material and methods

Bootstrapping allows increasing resistor R_1 to R_{BEQ}, but R_{BEQ} presents a noise voltage e_{RBEQ} greater than that of a real resistor of the same value. As is depicted in the Appendix, the noise of the equivalent resistor R_{BEQ} is:

$$e^2_{R_{BEQ}} \approx 4kTR_{BEQ}(R_2/R_3). \tag{5}$$

The bootstrap circuit in Fig. 2 increases R_1 by the factor (R_2/R_3), thus reducing f_N to $(2\pi R_{BEQ}C_S)^{-1}$, but the noise of R_{BEQ} is $\sqrt{R_2/R_3}$ 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 $f_N = (2\pi R_{BEQ}C_S)^{-1}$, Eq. (1) becomes:

$$e^2_o \approx \frac{kT}{(\pi C_S f)^2}\frac{1}{R_1} + \frac{i^2_n}{(2\pi C_S f)^2} + \alpha e^2_n; \tag{6}$$

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 GΩ (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_{BEQ}=100 GΩ by R_1=10 GΩ, and a bootstrap ratio R_2/R_3=10. This R_B 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 GΩ/square, instead of using more expensive substrates materials such as Teflon™, that present SR values of 1 TΩ/square 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 OA_1, and an input capacitance neutralization circuit implemented by OA_2 through capacitor C_C, according to [6]. The corresponding PCB design is shown in Fig. 3(b), where it can be observed that the neutralization capacitance C_C is implemented by a PCB area. A dual low-bias current operational amplifier OPA2320 (by Texas Instruments™) was used. The non-inverting input of OA_1 — the most vulnerable node of the circuit — is not soldered to the PCB, but bent upwards, and capacitor C_A 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 $R_{BEQ}C_S$, but short-circuiting the output of OA_1 (it must be output-protected), reduces it to $(R_1+R_3)C_S$, thus providing a way to discharge C_S. 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 GΩ and a low current noise OA as the OPA320, the noise PSD e_O is dominated by the first term in (6), decreasing with frequency f according to:

$$e^2_O \approx \frac{kT}{(\pi C_S f)^2}\frac{1}{R_1}; \tag{7}$$

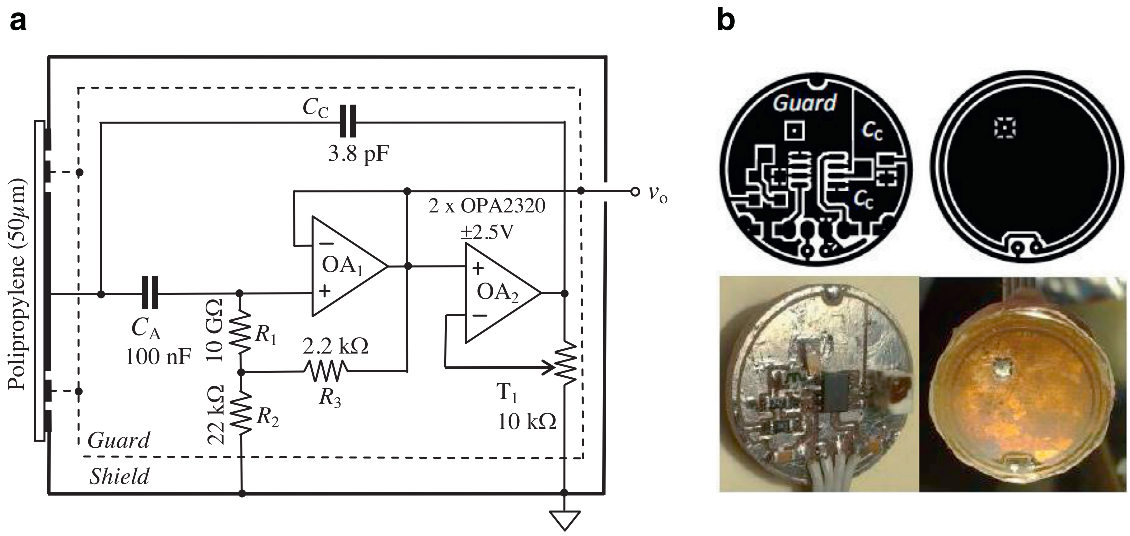

Fig. 3. (a) Complete circuit of the proposed CE and (b) printed circuit board and photo of the built prototype. Note that the proposed CE requires a reduced number of parts and includes a guard ring, plus a shield (ground) ring.

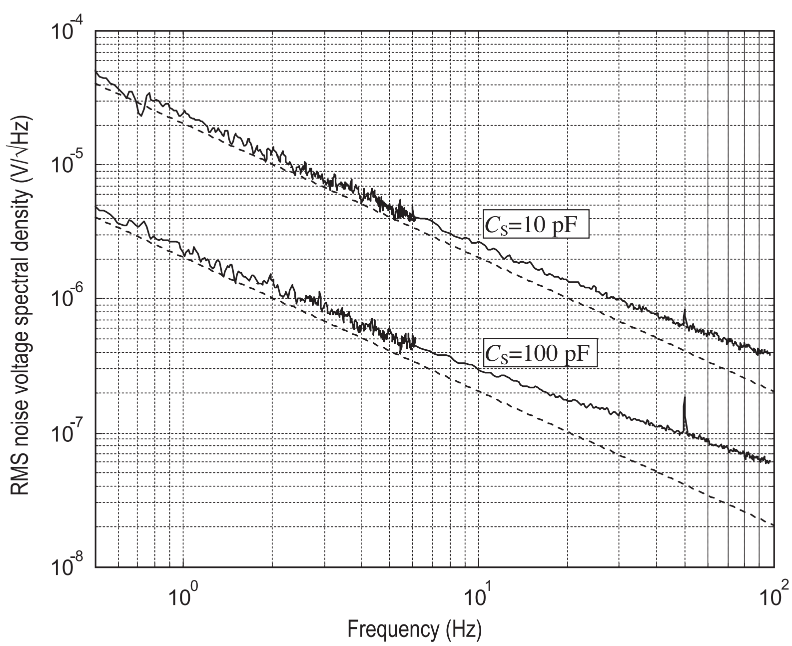

Fig. 4. Noise Power Spectral Density (PSD) of the proposed capacitive electrode for C_S=10 pF and C_S=100 pF. The expected noise due to resistor R_1, given by (8) for each C_S value, is indicated in dashed line.

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:

$$E_O^2 \approx \frac{kT}{(\pi C_S)^2}\frac{1}{R_1 f_1}. \tag{8}$$

Considering R_1=10 GΩ, T=300 K, C_S=100 pF and f_1=0.05 Hz, the total noise E_O results in 9 μV_{RMS}, which is a reasonable value for ECG signals. If f_1=0.5 Hz is considered, the noise is limited to 2.9 μV_{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 μV_{RMS}. This value is a little higher than the 2.9 μV_{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 μm 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.

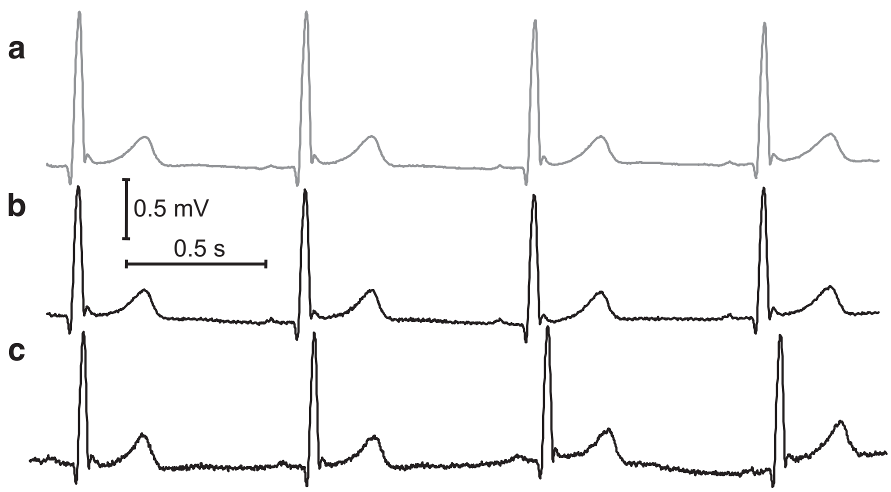

Fig. 5. ECG signals picked: (a) by standard disposable wet electrodes; (b) by the built capacitive electrodes placed on the skin; and (c) through a cotton T-shirt. Records in (a) and (b) were acquired simultaneously, and (c) corresponds to a different trial. Signal bandwidth was limited to 0.05–100 Hz and linear trends subtracted.

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_{BEQ} 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_{B}. 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:

$$e_{\mathrm{o}}^2 \approx e_{\mathrm{n}}^2(1+R_2/R_3)^2 + e_{\mathrm{R1}}^2(1+R_2/R_3)^2 + e_{\mathrm{R2}}^2 + e_{\mathrm{R3}}^2(1+R_2/R_3)^2; \tag{9}$$

and replacing resistors' noise PSDs by the Johnson–Nyquist formula (e_R=4kTR):

$$e_{\mathrm{o}}^2 \approx e_{\mathrm{n}}^2(1+R_2/R_3)^2 + 4\mathrm{k}TR_1(1+R_2/R_3)^2 + 4\mathrm{k}TR_2 + 4\mathrm{k}TR_3(1+R_2/R_3)^2. \tag{10}$$

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

$$e_{R_{\mathrm{BEQ}}}^2 \approx 4kTR_1(R_2/R_3)^2; \tag{11}$$

which can be written as:

$$e_{R_{\mathrm{BEQ}}}^2 \approx 4kTR_{\mathrm{BEQ}}(R_2/R_3). \tag{12}$$

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. Electrocardiografic and Bioelectric Capacitive Electrode. U S Pat 3 1970;500:823.
[3] Prance RJ, Debray A, Clark TD, Prance H, Nock M, Harland CJ, Clippingdale AJ. An ultra-low-noise electrical-potential probe for human-boy scanning. Meas 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. Meas Sci Technol 2000;13:163–9.
[6] Spinelli 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. Insulated Electrocardiogram electrodes. Med Biol Eng Comput 1972;10:742–51.
[8] Association for the Advancement of Medical Instrumentation. Ambulatory electrocardiographs American National Standard ANSI/AAMI EC38; 1999.
[9] Lee SM, Sim KS, Kim KK, Lim YG, Park KS. Thin and flexible active electrodes with shield for capacitive electrocardiogram measurement. Med Biol Eng Comput 2010;48:447–57.
[10] Prance RJ, Clark TD, Prance H, Clippingdale AJ. Non-contact VLSI imaging using a scanning electric potential microscope. Meas Sci Technol 1998;9:1229–35.
[11] Lanyi S. The noise of input stages with low parasitic capacitance. Meas Sci Technol 2001;12:1456–64.
[12] Pallas Areny R, Colominas J, Rosell J. An improved buffer for Biolectric Signals. IEEE Trans Bio Med Eng 1989;36:490–3.
[13] LPC662 Low Power CMOS Dual Operational Amplifier datasheet. Texas Instruments Incorporated 2000. Available at www.ti.com/cn/lit/gpn/lpc662.
[14] Spinelli E, Haberman M, García P, Guerrero F. 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.