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A capacitive electrode with fast recovery feature

Enrique Spinelli[1,2], **Marcelo Haberman**[1,2], **Pablo García**[1] **and Federico Guerrero**[1,2]

[1] LEICI—Departamento de Electrotecnia, Universidad Nacional de La Plata CC 91 (1900) La Plata, Argentina
[2] Consejo de Investigaciones Científicas y Técnicas (CONICET), Buenos Aires, Argentina

E-mail: spinelli@ing.unlp.edu.ar

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Abstract
Capacitive electrodes (CEs) allow for acquiring biopotentials without galvanic contact, avoiding skin preparation and the use of electrolytic gel. The signal quality provided by present CEs is similar to that of standard wet electrodes, but they are more sensitive to electrostatic charge interference and motion artifacts, mainly when biopotentials are picked up through clothing and coupling capacitances are reduced to tens of picofarads. When artifacts are large enough to saturate the preamplifier, several seconds (up to tens) are needed to recover a proper baseline level, and during this period biopotential signals are irremediably lost. To reduce this problem, a CE that includes a fast-recovery (FR) circuit is proposed. It works directly on the coupling capacitor, recovering the amplifier from saturation while preserving ultra-high input impedance, as a CE requires. A prototype was built and tested acquiring ECG signals. Several experimental data are presented, which show that the proposed circuit significantly reduces record segment losses due to amplifier saturation when working in real environments.

Keywords: insulating electrodes, active electrodes, non-contact measurements

(Some figures may appear in colour only in the online journal)

1. Introduction

Biopotential acquisition systems are intended to pick up very small biopotentials (lower than a few mV) but, under some conditions, artifacts up to a few volts are produced, thus saturating the biopotential amplifier. Many events such as electrode movements (Searle and Kirkup 2000), change of electrodes or defibrillation shock pulses (Neycheva and Krasteva 2003) can produce these high amplitude artifacts. Given that biomedical signals demand to acquire very low frequency components, biopotential amplifiers present transient responses

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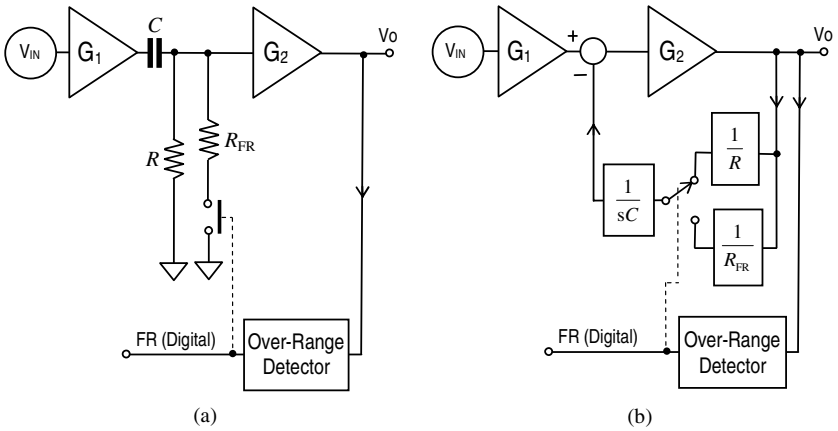

Figure 1. (a) Fast recovery passive scheme based on switching resistors and (b) by modifying integrator's time constant in an active dc suppression approach.

with time constants of several seconds. Once saturated, amplifiers remain in this state for long periods of time, during which the biopotential signal is lost. Therefore, equipment for clinical practice usually provides a mechanism for rapid restoration to a proper operating point. This feature known as fast recovery (FR) or rapid baseline recovery (RBR) (Neycheva and Krasteva 2003, Seguine *et al* 1998, Spinelli *et al* 2004) is an important issue, mainly in autonomous unsupervised acquisition systems such as holters and other long-term biomedical recording devices.

The FR function is implemented in the biopotential amplifier's ac-coupling stage, usually placed between two gain blocks G_1 and G_2, as figure 1 shows. Actually, FR is just an ac-coupling stage with a selectable low cut-off frequency f_{CL}. Under normal operating conditions, very low f_{CL} (i.e. 0.05 Hz) must be used to avoid low-frequency distortions (Daskalov *et al* 1997), thus involving time constants of 3–4 s and bigger. Therefore, if an artifact saturates the amplifier, then a longer time is required to restore a proper dc voltage. A FR system works switching time constants. So, when the amplifier gets saturated, a small time constant τ_{FR} is selected until its output is in range, and then the nominal time constant τ is restored. Switching between these time constant is managed by a supervisor circuit that verifies whether the amplifier's output signal is within its operating range. An additional desirable feature of a FR system is the generation of marks that indicate when the FR was activated. This extra signal helps in avoiding medical misinterpretations.

The two main methods to achieve FR are using a passive network (Dotsinsky *et al* 1985) or subtracting the dc input voltage by using an integrator to feed back the dc output voltage (Seguine *et al* 1998, Smit *et al* 1987, Spinelli *et al* 2004) (figures 1(a) and (b) respectively). Both approaches restore the baseline level by switching time constants as was described before. The second alternative can also reject any dc perturbations such as operational amplifier (OA) offset voltages, bias currents or electrode offset potentials, and ensures a zero mean signal at the amplifier's output

The FR system prevents the saturation of amplifier G_2, but the gain G_1 of the first stage must be low enough to ensure that, even in the worst case, it will not saturate. It is worth noting that FR systems do not avoid the front-end saturation due to high amplitude artifacts. They just restore the amplifier baseline level after the artifact vanishes.

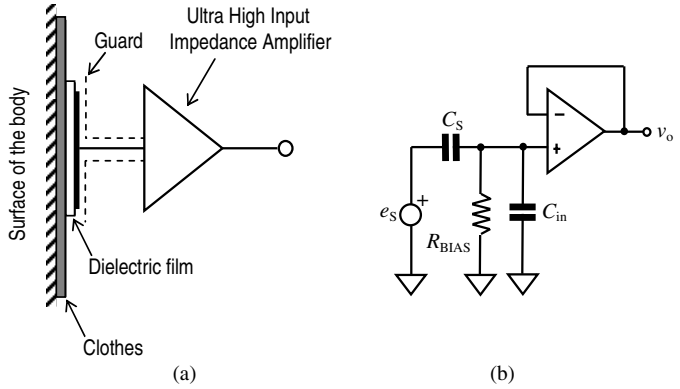

Figure 2. (a) General scheme of a capacitive electrode and (b) its simplified equivalent circuit.

When wet electrodes are used the main artifact is the electrode polarization voltage, which could be of hundreds of mV, increasing residual polarization voltages up to 1 V when a defibrillator is applied (Neycheva and Krasteva 2003). Then, considering a ± 5 V power supply, G_1 must be lower than 5 times. An amplifier merit factor is the highest artifact amplitude (or dc voltage) at the input (V_{iDCMAX}) that it can reject. It depends on G_1 and also on the available power supply voltage $\pm V_{CC}$:

$$V_{iDCMAX} = \frac{V_{CC}}{G_1}. \tag{1}$$

The situation is quite worse for capacitive electrodes (CEs), because part of the ac-coupled stage moves outside the front end: the body-electrode capacitance C_S acts as a coupling capacitor (figure 2(b)). The stage is completed by a grounded resistor R_{BIAS}, included to provide a path for amplifier bias currents. Nowadays, OAs present very low bias currents and, properly managed, printed circuit board (PCB) and isolation layer leakages are enough to provide a path for these currents (Prance *et al* 2000, Chi *et al* 2010). Nevertheless, in order to restore a proper operation point when a motion artifact or any sudden charge injection occurs, a mechanism to set this dc level must be included (Sullivan *et al* 2007).

CEs demand ultra-high input impedances, and they usually include circuitry to neutralize OA input capacitance C_{IN} and for shielding. These techniques were used by Prance *et al* (1998) for non-contact electric potential measurements, and then extended to biomedical signals (Harland *et al* 2002, Prance 2011, Spinelli & Haberman 2010). Recently, Chi *et al* (2012) solved all the instrumentation problems associated with CE front ends introducing bootstrapping and shielding techniques (Prance *et al* 1998) inside the integrated circuit itself, achieving input capacitances as low as 60 fF. This avoids manually adjusting the input capacitance neutralization and other procedures frequently used to achieve working CEs. However, amplifier saturation and latch-up due to high amplitude artifacts remain as problems to be solved (Chi *et al* 2010).

The coupling capacitor C_S ranges from hundreds of pF when the electrodes are placed over the skin to tens of pF when placed over clothing. Under this last condition, C_S also significantly varies with motion, pressure or other mechanical factors (Luna-Lozano and Pallas-Areny 2010). As a result, the bias resistor R_{BIAS} must be very large (up to TΩ) in order to fulfil the low cut-off frequencies required in biomedical applications, and also to achieve low noise levels (Prance *et al* 2000, Spinelli and Haberman 2010).

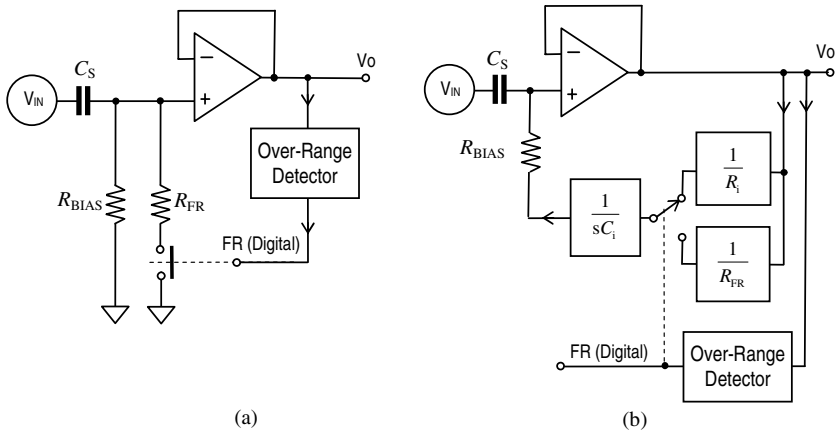

Figure 3. Adaptation of the FR schemes of figure 2 to capacitive electrodes.

Working with CEs, the amplifier saturates due to the accumulation of charges in C_S, which is not part of the electronic circuit. Since C_S can be as low as a few pF, a small amount of injected electrostatic charge can easily produce artifacts of several Volts and higher, thus saturating the amplifier. Restoring a proper operation point demands working directly on C_S's charge.

The schemes of figures 1(a) and (b) can be adapted for CEs, resulting in the circuits of figure 3. The dc suppression approach of figure 3(b) works properly under normal conditions (Eilebrecht *et al* 2010, Spinelli and Haberman 2010), but it is not suitable for implementing the FR function because R_{BIAS} limits the maximum discharge current to V_{CC}/R_{BIAS}. So, if a large charge is accumulated in C_S, long times are required to restore the baseline. Moreover, the integrator in the feedback loop introduces an additional pole in the transfer function and very large integrator's time constants are required to fulfil the transient response imposed by the biomedical standards.

The passive method of figure 3(a) is simpler and effective; it consists of a switch in parallel with R_{BIAS} to discharge C_S when its voltage saturates the amplifier. The challenge is to implement this switch without degrading the ultra-high input impedance that CEs require. One alternative is to use reed-relays, but this is not well suited for integrating it in the electrode itself. This problem was solved by Sullivan *et al* (2007) using a bipolar transistor at the amplifier input to discharge C_S when the amplifier goes out of range. The transistor is bootstrapped to the input voltage to avoid the effects of its leakage currents when it does not work. This ingenious solution, however, brings about some difficulties in generating the proper signals to drive the transistor. It either demands several components, increasing the electrode circuit complexity, or requires wiring almost two individual lines to each electrode if the FR circuit is implemented in the main board. It also presents some problems to discharge negative voltages on C_S. The circuit proposed in this paper is intended to overcome these problems, adding just a few components inside the CE.

2. The proposed scheme

The proposed scheme is presented in figure 3(a). It is based on the idea proposed by Sullivan *et al* (2007) to implement the basic scheme of figure 4(a), but it places the FR resistor R_{FR} between the amplifier input and output. Ideally, during normal operation, the potential

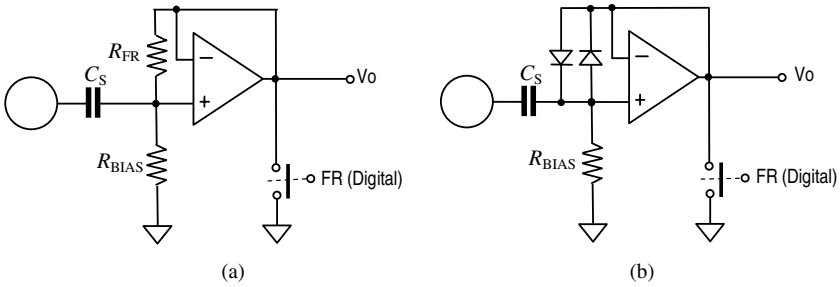

Figure 4. (a) Proposed FR circuit. (b) Replacing R_{FR} by two back-to-back diodes to avoid the OA offset voltage amplification and the use of high value R_{FR} resistors.

difference across R_{FR} is zero, and it does not affect the amplifier's behavior. When the voltage on C_S saturates the amplifier, the amplifier's output is short-circuited and R_{FR} discharge C_S. Then, the short circuit is removed and the CE starts working normally. A resistor can be placed in series to the OA output to limit the output current during the reset, but it can be omitted if output-protected OAs are used.

An important advantage of the proposed scheme is that the circuitry for the detection of the 'out-of-range' condition can be placed in the electrode itself or in the main board containing the ADCs and interface circuits. Given that amplifier reset is performed through the preamplifier output V_O, no additional wires are required if the FR system is moved from the electrodes to the main board. This reduces the number of components on the electrodes, thus allowing us to reduce their size and/or increase free PCB areas for shielding. Moreover, the latter alternative also permits us to generate marks on the record to know when the FR has been activated, helping us to prevent misinterpretations of the ECG signals.

The circuit is very simple and works as described above; but it presents some problems related to factors not included in the simplified circuit of figure 4(a), such as the OA offset voltage and certain PCB stray capacitances.

2.1. Operational amplifier offset voltage effect

The resistor R_{FR} in figure 4(a) is connected between two nodes that are at the same potential (virtual ground), and it should not affect circuit variables, but dealing with the real OA, an equivalent input offset generator V_{OS} appears over R_{FR}. So, a current V_{OS}/R_{FR} flows through it and continues through R_{BIAS}, acting like a bias current (i_{BIAS}) and resulting in an output voltage V_O given by

$$V_O = V_{OS}\left(1 + R_{BIAS}/R_{FR}\right). \tag{2}$$

So, to achieve a fast discharge of C_S a low value R_{FR} resistor is required, but it amplifies the offset voltage as given by (2). This compromise between a FR and OA's offset voltage amplification can be broken replacing R_{FR} with two back-to-back low leakage diodes, as is shown in figure 4(b). Under normal working conditions both diodes are off, they present high impedance and do not significantly amplify V_{OS}, but when the FR mechanism is activated, this impedance is reduced to hundreds of ohms, thus providing the desired fast C_S discharge. If no R_{BIAS} resistor is included, the effect of R_{FR} is equivalent to increase i_{BIAS} in V_{OS}/R_{FR}. The diodes provide a fast capacitor discharge down to 0.6 V. The remaining dc voltage can be extracted by a second ac-coupling stage or by digital signal processing, if high dynamic range ADCs are used.

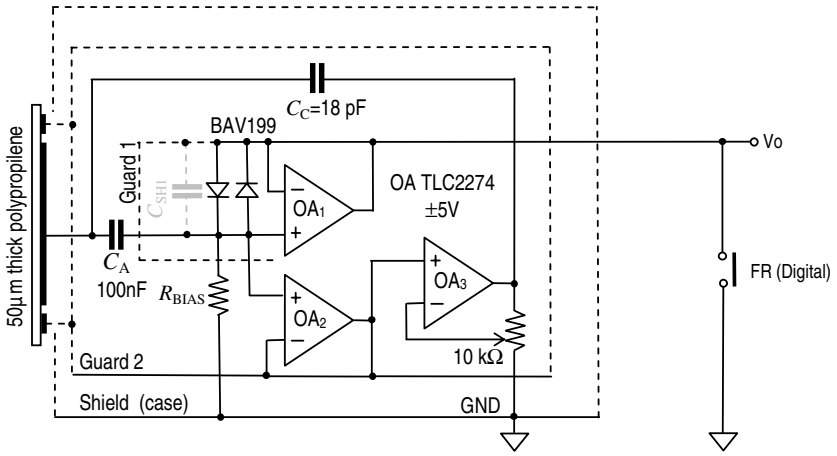

Figure 5. Schematics of the complete circuit, including guards, shield and input capacitance neutralization circuit (AO_3).

OA's offset effects can be avoided using ac-coupled bootstrapping techniques (Pallas Areny *et al* 1989, Prance *et al* 1998), but this approach leads to inductive input impedances, making it difficult to achieve damped transient responses as the ECG AAMI standard requires.

The nonlinear impedance of diodes has also been proposed to provide a bias current path (Prance *et al* 1998, Richardson and Lopez 1970, Potter and Menke 1970) and to clamp the electrode voltage (Chi and Cauwenberghs 2009). In the last case, the electrode voltage is forced to remain within around ± 0.6 V (diode forward voltage drop V_{FDIODE}), but it does not work as a FR circuit. Signals can remain at an improper operational point (low R_{BIAS} value), being distorted.

2.2. Stray PCB capacitances effects

When the OA output is short-circuited to ground, the capacitance C_{SH1} between its input (in parallel with the diodes) can be charged to hundreds of mV (up to V_{FDIODE}). Therefore, when the circuit returns to work normally, C_{SH1} is discharged and its charge is transferred to C_S to charge it: exactly what the proposed circuit tries to avoid. In order to minimize this effect, the capacitance C_{SH1} must be reduced. To achieve this, a dual guard was used: guard 1 just for surrounding a small area and capacitance C_{SH1}, and guard 2 for the rest of the circuit (see figure 5). The price to be paid is that this additional buffer increases the current noise $\sqrt{2}$ times and i_{BIAS} two times. These are not serious problems for ECG signals, but could jeopardize its use for very low noise applications as EEG. The buffer also increases the input capacitance, but it is neutralized jointly with the OA input capacitance C_{IN} by AO_3.

2.3. Practical circuit

The final and complete circuit is shown in figure 5. It includes the neutralization of C_{IN} and the guards (Prance 2011, Spinelli and Haberman 2010), with being a typical CE discrete implementation. The circuit is really simple; the main and sometimes hidden component is the PCB design (figure 6).

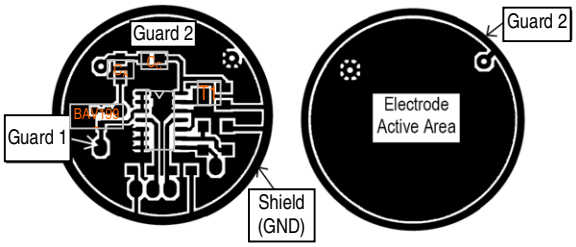

Figure 6. Printed circuit board design of the built CE.

As was stated previously, due to the diodes and i_{BIAS}, a residual dc voltage could remain on C_S after the FR is activated. This can be solved by a second high-pass stage (Chi *et al* 2010, Oehler *et al* 2008, Fuhrhop and Heuer 2009) or by digital signal processing when signals are acquired by a high dynamic range ADC, as was done in the test runs to be presented in the following section. A 24 bit sigma-delta ADC (ADS1259 from Texas Instruments) was used. It provides an input range of ± 1.25 V with an input referenced noise of 1.4 μV_{RMS} for a 127 Hz bandwidth, which is enough to acquire good quality ECG signals.

The circuit was mounted on a standard double layer FR4 PCB, as shown in figure 6, where the guards 1 and 2 are indicated. No solder mask was used in order to minimize superficial leakage currents. The non-inverting inputs of the buffers AO_1 and AO_2 were not soldered to the PCB but left 'floating in the air' with the coupling capacitor C_A. If necessary, a wire-wrap wire can be strapped on this connection to produce a gimmick R_{BIAS} of around 1 TΩ as was used in Spinelli and Haberman (2010). Having a FR circuit, R_{BIAS} can be omitted; OAs usually find a bias path through parasitic leakages (Chi *et al* 2010) and if it is not enough, the FR acts keeping the CE in range. Nevertheless, it is a good practice to include R_{BIAS} when working with low C_S values (20 pF or lower) as when acquiring biopotential through clothing in order to avoid the frequent activation of the FR circuit. A 50 μm thick polypropylene film was used as the isolation layer.

3. Experimental results

The proposed CE was built and the input capacitance neutralization of each CE was adjusted as in Prance *et al* (2000), aiming for a unity gain when a 1 kHz sine wave was applied through a 10 pF capacitor (simulating C_S). Several tests were performed acquiring real ECG signals with the experimental setup as depicted in figure 7. This setup allows acquiring the differential voltage between electrodes A and B and is able to produce manual (switches SW) and automatic activations of the FR circuit (by means of the transistors). On start-up, if the output voltage of some electrodes is out of range, an automatic reset on these channels is performed. It is worth noting that the restoration of each electrode is independently managed.

The common mode and the differential mode of the voltages from the electrodes were conditioned (Spinelli *et al* 2010) and differentially acquired by a high-resolution sigma-delta ADC (ADS1259 from Texas Instruments) with an input range of ± 1.25 V. In the tests, the subject was capacitively grounded by a large capacitance built with approximately 300 cm^2 of 3M^{TM} EMI Shielding Foil Tape placed on a chair and insulated, as the electrodes, by a 50 μm polypropylene film. No wet electrodes were used in these tests.

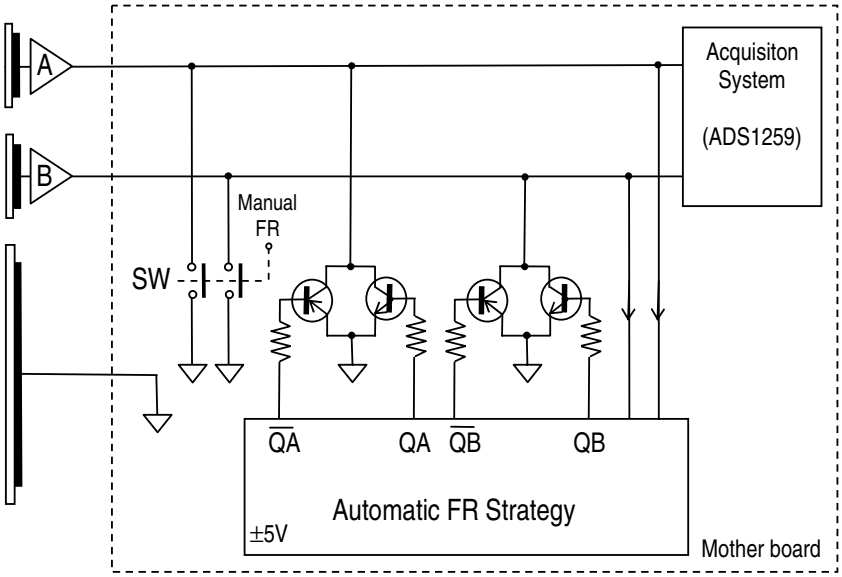

Figure 7. Experimental setup used to acquire ECG signals. Manual activations of the FR are produced by switches SW and automatically by the transistors.

3.1. Motion artifacts

Two electrodes were placed on the chest over a cotton T shirt and held in position by an adjustable band. Strong transversal motion artifacts were produced moving the electrodes by stretching the band. A typical waveform for this condition is shown in figure 8. In this case, the FR was automatically activated for 1 s only for the electrode that came out of range ($\pm$ 1.25 V). Several strategies to manage the FR can be proposed, but they depend on the application and other factors such as the number of electrodes and artifact features, among others. As can be observed in the figure, the artifact saturates at least one of the CEs, and the FR was activated. After this, the ECG signal acquires a baseline level inside the input range. The subject was asked to keep in apnea after the electrode movement.

A second test with motion artifacts was performed. In the first half of the record, shown in figure 9, the subject breathed deeply and stayed in apnea, and a knock (with a piece of wood) was applied to one CE. It can be observed that, in these cases, the large input range of the system is enough to tolerate these moderated breathing and motion artifacts.

3.2. High amplitude electrostatic artifact

A common problem in dry climates is the accumulation of electrostatic charge on people and objects, which can be transferred to the subject. Due to the small capacitances involved, a small amount of charge is enough to produce tens of volts at the buffer's input, thus saturating the front end. In some cases it is enough to wait for tens or hundreds of seconds to restore normal biopotential acquisition conditions (Chi *et al* 2010), but sometimes we observed that if the artifact is very strong, it leads the front end to saturation and it never returns to proper operation by itself, perhaps due to the latch-up effect. This is a serious problem for unsupervised acquisition systems such as small autonomous recording devices.

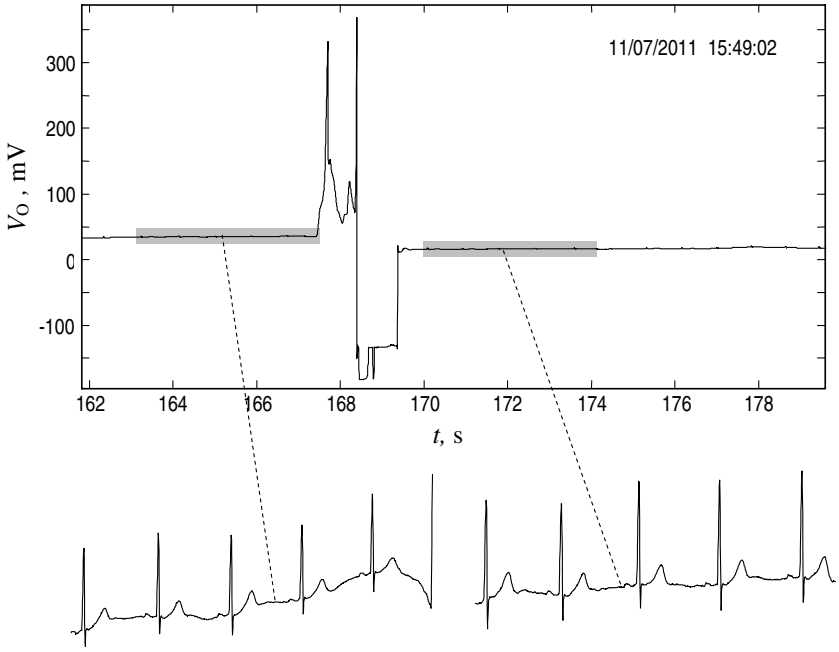

Figure 8. Experimental data obtained by producing strong transversal motion artifacts by stretching the band that hold the CEs over the chest. Below, the ECG signals are shown in detail. For these signals, the linear trends were digitally removed and their bandwidths were limited to 0.05–100 Hz.

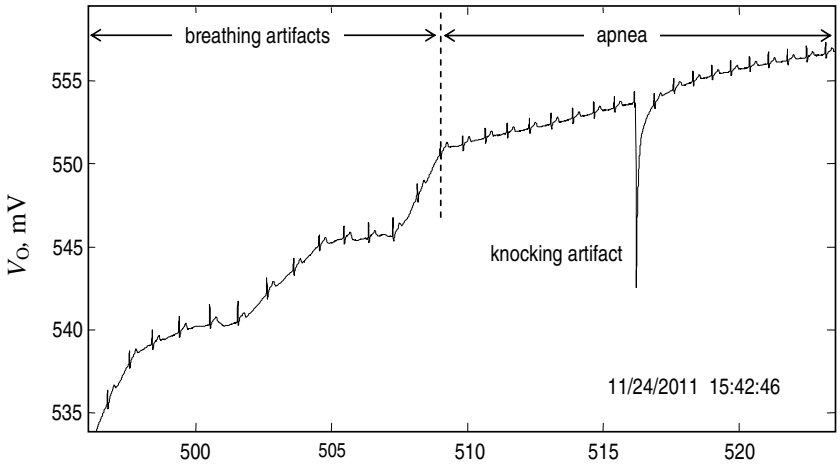

Figure 9. Experimental data obtained producing moderate motion artifacts by deeply breathing and knocking the CE with a piece of wood. The CEs were placed on the chest and held by a band.

In order to reproduce the situation of a sudden charge injection, a 0.47 μF capacitor charged at 9 V was discharged on the subject. The capacitor charge, of around 5 μC, was similar to that of the human boy model capacitance (ESD Association 2007) (100 pF) at 500 V. Figure 10 shows a typical record of this test with CEs placed directly on the chest. This

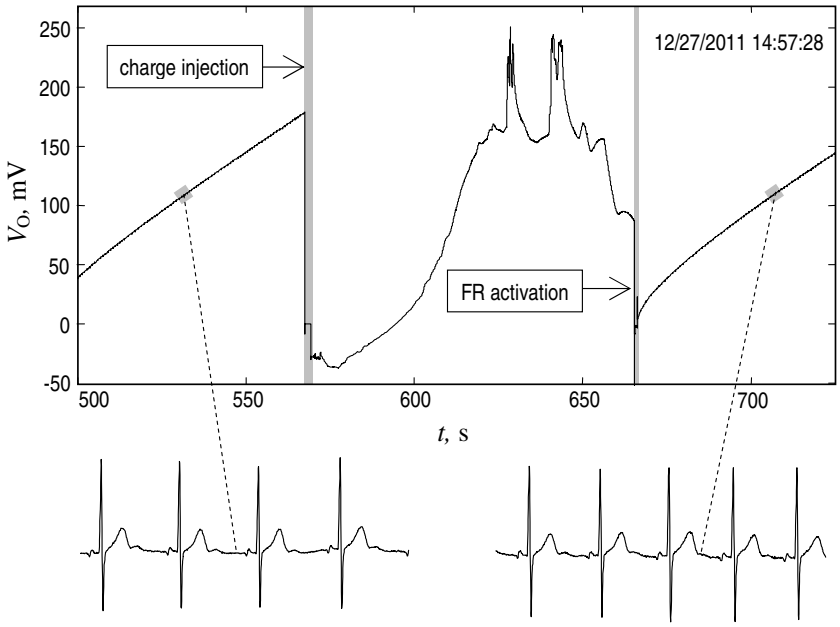

Figure 10. Experimental data obtained by producing a strong artifact by injecting an electrostatic charge. It was injected at 570 s, leading the CE to saturation. It did not return to work until the FR circuit was activated. Below, the ECG signals are shown in detail. For these signals, the linear trends were digitally removed and their bandwidth limited to 0.05–100 Hz.

condition is quite favorable, because the coupling capacitance C_S increases to around 200 pF. As can be observed in this figure, when the charge is injected, the CE saturates and the ECG signal is lost. The CEs do not recover from this state until the FR is activated. In this case, the automatic reset circuit was disabled and the outputs of both electrodes were grounded manually by the switches SW (figure 7)

It is important to note that a supervisor system must examine the output voltages of each electrode individually. For example, in the record of figure 10, the CE saturation cannot be detected from the differential signal.

4. Conclusions

A capacitive electrode (CE) with a fast recovery (FR) feature has been proposed and experimentally tested. It works directly discharging the coupling capacitance C_S and is able to restore the amplifier from saturation due to large amplitude artifacts. The main goal of the circuit is to provide a fast discharge path for C_S without degrading the input impedance.

The circuit is simple and the reset signal can be generated locally in the CE itself or in the main board. It works through the CE output lead; therefore, no additional signals must be wired from the motherboard to the electrodes, thus reducing wiring. It is also possible to generate marks to indicate when the FR circuit has been activated.

The proposed circuit was tested through the acquisition of real ECG signals. It was observed that the FR is not necessary to deal with motion or breathing artifacts when a high input range is provided (± 1.25 V), but it is crucial to achieve a rapid start time when

electrodes are connected–reconnected and to recover the CE from high amplitude electrostatic interferences.

Acknowledgments

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