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ORIGINAL ARTICLE

# Comments on "A novel high input impedance front-end for capacitive biopotential measurement"

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**Abstract**
A front-end for biopotential sensing in wearable medical devices has been recently proposed which is claimed to provide 100 GΩ input impedance by manually matching two resistor pairs in a positive- and a negative-feedback loop around an operational amplifier (op amp); the cost being that the equivalent input noise voltage doubles with respect to a simple non-inverting amplifier. The ECG acquired with capacitive (sic) electrodes through a cotton shirt is presented as a proof of the performance of the proposed circuit. It turns out, however, that the analysis ignores op amp's input capacitance hence the effort to achieve a very high input resistance seems futile. Further, cotton is highly hygroscopic hence not an appropriate dielectric, so that there is no proof that the electrodes tested were actually capacitive. This comment addresses these two problems and some additional conceptual and methodological inaccuracies found in the paper.

**Keywords** Biopotential sensing · Capacitive electrodes · Voltage-loading effect · Noise analysis

## 1 Impedance model for capacitive electrodes

At any given frequency, electrode impedance, the same as the impedance of any other material, can be modeled by a resistance $R_s$ in series with a capacitance $C_s$ or a resistance $R_p$ shunted by a capacitance $C_p$. For electrodes wherein DC can flow through, the parallel model seems more appropriate because, in the series model, capacitance $C_s$ blocks DC. The relationship between the parameters of the two models is

$$R_s = \frac{R_p}{1 + (\omega R_p C_p)^2} = \frac{R_p}{1 + (\omega/\omega_p)^2} \quad (1)$$

$$C_s = C_p\left[1 + \frac{1}{(\omega R_p C_p)^2}\right] = C_p\left[1 + \frac{1}{1 + (\omega/\omega_p)^2}\right] \quad (2)$$

where $\omega_p = 1/\tau_p = (R_p C_p)^{-1}$. These equations show that, even if $R_p$ and $C_p$ are constant within a broad frequency range, the value of $R_s$ and $C_s$ will change at each signal frequency being considered inside that range. At (angular) frequencies smaller than $\omega_p$, $R_p$ and $R_s$ and $C_p$ and $C_s$ will be very close. Therefore, a small $R_s$ at a given frequency requires a very small $\omega_p$ as compared with that frequency, hence a very large $\tau_p$; a large $R_p$ is not enough to guarantee a small $R_s$. Consequently, the statement in [1] "*If the equivalent model between the body and the electrode is simplified as a coupling capacitor, the equivalent input impedance of the post front-end should be at least 100 GΩ to detect 0.1 Hz low frequency signal for the coupling capacitance as low as several pF*" needs some discussion relative to the electrode model and to the ECG signal frequency.

With regard to electrode impedance, $R_p$ would become redundant if its impedance were much larger than that of $C_p$ at the signal frequency $\omega$, i.e., $R_p >> 1/\omega C_p$, hence $\omega_p << \omega$. With regard to ECG signal frequency, its fundamental component is usually higher than 0.5 Hz (30 beats per minute). The 0.05 Hz corner frequency in some ECG diagnostic standards is intended to prevent waveform distortion. Therefore, electrode impedance at 0.5 Hz will be capacitive whenever $\tau_p = R_p C_p >> 1$ s/π = 318 ms, which is quite large for biopotential electrodes. For example, a cotton-based "non-contact electrode" described in [2] had 305 MΩ shunted by 34 pF, hence $\tau_p = 10.4$ ms ($\omega_p \approx$ 96 rad/s), which means that its reactance at 0.5 Hz (3.14 rad/s) is more than 30 times larger than its resistance, and at 10–15 Hz, wherein most of ECG power is, electrode reactance and resistance will be close. Then, from Eq. (1), it follows $R_s \approx R_p/2$; hence, it cannot be neglected at all.

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Moreover, even if the reactance predominated, the electrode would not be "capacitive" if amplifier input bias current could flow through $R_p$ [3]. Consequently, if input bias currents are very low, the insulation of capacitive electrodes must be extremely good. The LMP770x used in [1], with $I_b = \pm$ 200 fA, needs insulation better than 1 TΩ, which is not easy to achieve in printed circuits (hence the use of "air wiring"), less with common fabrics between the electrode plate and the skin. The electrodes described in [1] were circular metal plates 4 cm in diameter on a cotton T-shirt. If the metal-skin gap is assumed to be about 0.25 mm, and the relative permittivity for cotton is between 1.3 and 1.4, then the electrode capacitance is about 58 to 72 pF. Table 1 in [1] assumes $C_{eq} = 470$ pF, and later calculations use 46 pF. No insulation resistance was specified but achieving 1 TΩ with cotton is unrealistic.

## 2 Voltage-loading effect

The target 100 GΩ input impedance for the proposed font-end is meant to avoid undesired voltage attenuation. For an electrode with equivalent impedance $Z_{eq}$ connected to a front-end with equivalent input impedance $Z_{in}$, the voltage attenuation will be

$$A = 1 - \frac{Z_{in}}{Z_{in} + Z_{eq}} \tag{3}$$

For the proposed front-end, $Z_{in}$ comprises the equivalent resistance $R_{eq}$ (termed "impedance" in [1]) between the op amp's non-inverting terminal and ground, shunted by the capacitance $C_c$ between that terminal and ground (common-mode input capacitance), which is unaffected by feedback networks around the op amp. For FET-input op amps, $C_c$ plus the layout capacitance can expected to be at least 1 to 5 pF, and for the CMOS-based LMP770x, $C_c$ can easily reach about 30 pF. A PCB ground plane like that used in [1] will add a few more picofarads. Since most ECG power is between 10 and 15 Hz, at this frequency, $C_c$ will predominate over $R_{eq}$ and for an electrode (capacitive or not) with equivalent capacitance $C_{eq}$ and a very high $R_p$, we would have $A = C_{eq}/(C_{eq} + C_c)$. For a 34-pF electrode, $A \approx 0.5$. Coincidentally, in [1], the ECG in Fig. 14, obtained with cotton electrodes, is below 50% of that in Fig. 13, obtained with dry electrodes, which capacitance can be much larger. Therefore, since the input resistance of FET or CMOS op amps can be higher than 10 TΩ, there is no need to try to improve it in order to avoid undesired signal attenuation. In fact, any external network will probably degrade the very high input resistance of the device. What limits the input impedance of analog front ends for biopotentials is the equivalent input capacitance of the amplifier (off-the-shelf or specific IC), circuit layout, and electric shields if used. Table 1 in [1] provides only input resistance measurements, not actual input impedance measurements that could have revealed the presence of the input capacitance.

## 3 Amplitude frequency response

Since, on the one hand, the electrode is assumed to be purely capacitive ($C_{eq}$) and, on the other hand, the input impedance of the front-end is assumed to be purely resistive ($R_{eq}$), the frequency response is high pass and, from Table 1 in [1], the time constant will be $\tau = 111$ GΩ × 470 pF = 52.2 s. This means that the settling time after any transient change will be of the order of several minutes, which is inconvenient in wearable devices. Fig. 2 in [1] shows quite correctly that the (−3 dB) corner frequency is 0.003 Hz ($\tau \approx 53$ s), which implies that low-frequency noise (electronic and undesired physiological signals) will add to the desired ECG. The dependence of this corner frequency on electrode impedance, which depends itself on the body site where electrodes are placed, means that each electrode signal may see a different corner frequency. This is a demerit for any differential measurement.

## 4 Noise model and analysis method

In ECG recordings with common conductive electrodes, noise other than motion artifacts has been demonstrated to be larger than the thermal noise of the electrode impedance [4]. Electrode noise mainly originates in the electrolyte-skin interface and its rms value ranges from 1 to 20 μV depending on the electrode gel and the skin properties of the subject [5]. For wearable devices, EMG noise and motion artifacts will probably predominate because in ambulatory monitoring it has been demonstrated that "*the electrode-tissue impedance can correlate with the motion artifacts for local disturbance of the electrodes*" [6]. Therefore, front-ends need only a basic noise analysis in order to check that their noise is commensurate with the expected EMG and motion artifact noise.

The focus in [1], however, is on the relationship between the equivalent input resistance of the proposed front-end circuit and the equivalent input noise. This does not need any effort as it was demonstrated, long ago, that a larger input resistance in voltage measurements does not increase the total output noise in spite of the increased spectral density of the thermal noise of the resistor [7]. For active electrodes in particular, it has been demonstrated that "*driving the input impedance to infinity minimizes the noise figure for the sensor, irrespective of the source impedance*" [8]. Input current noise contributions can be kept small in any case by selecting an appropriate amplifier.

Nevertheless, the noise model and analysis method used in [1] deserve some additional cautionary words. First, the following sentence in the "Introduction" "*the input referred noise*

*in 0.7−100 Hz frequency band is 3.8 $\mu V_{RMS}$ with the gain of 46 dB*" (sic) is somewhat misleading as it associates input referred noise to amplifier gain. Input referred voltage noise is the output voltage noise divided by the voltage gain; hence, it is independent of the gain. Second, output noise power spectral density must be calculated by multiplying the input noise power spectral density by the square of the modulus of the transmittance but Eqs. (23) and (32) in [1] show input and output voltage noise related by the transfer function as if these were deterministic signals. Third, it is not clear how or what determines the noise bandwidth in noise calculations in [1]. It seems that the noise bandwidth considered is 1–100 Hz for all noise sources, when in fact, the actual noise bandwidth is not the same for all resistors. Finally, there is no clear explanation about how were electrodes placed during noise measurements: where they in contact with a dry T-shirt? These facts cast serious doubts about the validity of the noise analysis and noise measurements performed.

## 5 Conclusions

Capacitive biopotential electrodes are expected to form a capacitive voltage divider with the input capacitance of the circuit they are connected to. This is because above, say, 0.5 Hz, the input reactance of high-input-resistance amplifiers is smaller than their input resistance as 1 pF has "only" 340 GΩ whereas input resistance can exceed 10 TΩ. But if the input bias/leakage current of the front-end circuit actually flows through the electrode, this means that the insulation resistance is not high enough and the equivalent impedance for the electrode is that insulation resistance shunted by some capacitance; hence, the electrode can no longer be considered capacitive. This is often the case of through-clothing ECG measurements as sweat (an electrolyte) bridges the gap between the electrode plate and the skin so that the resistance between them will seldom reach 1 GΩ.

Further, the input impedance of the front-end proposed in [1] is limited by the common-mode input capacitance of the op amp, the same as any other of the several front-ends cited in [1], and no external feedback can reduce that capacitance. Consequently, the electrode model in Fig. 1 of [1], which includes only a capacitance and the omission of the op amp's common-mode input capacitance in the same figure are unjustified oversimplifications.

A widely used solution for through-clothing ECG measurements is a simple voltage buffer, with a commensurate resistor connected to signal ground to provide a path for input bias currents. This could be the reference design to assess the performance of alternative front-end circuits such as the one proposed in [1].

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