Investigation of Non-contact Electrodes for Electrocardiogram Monitoring

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***Abstract*— This work investigates the use of an active non-contact electrode to monitor electrocardiogram (ECG) signals in the chest region of the human body across different interface materials. Non-contact electrodes provide an advantage over wet gel electrodes as they do not require direct contact with the skin which can lead to skin irritation. The key aspects and parameters of non-contact electrodes were discussed. Additionally, a printed circuit board (PCB) electrode was designed and prototyped. Experimental evaluations were performed to demonstrate the feasibility of the design as well as the impact of various interface materials. Results showed that R-peaks can be detected through all materials tested. The highest noise levels were observed in polyester, resulting in the highest signal standard deviation, followed by merino and cotton. The patch sensor with just a solder mask as an insulation layer provided the clearest ECG signal, with 100% accuracy on R-peak detection.**

***Clinical relevance*— Non-contact electrodes offer a more comfortable solution for long-term heart monitoring with minimal discomfort due to less skin irritation when compared to conventional electrodes.**

I. INTRODUCTION

An ECG is a test that measures the electrical activity of the heart, providing information about the heart's rhythm. The signal information obtained through ECG monitoring allows healthcare professionals to identify a patient's underlying heart health conditions, such as abnormal heart rate variations and irregular segment patterns. In recent years, the popularity of wearable devices has made ECG monitoring more widely accessible, as it is no longer limited to professional healthcare facilities. These devices come in different forms, including smartwatches, wristbands, and chest straps, making it convenient for individuals to monitor their heart health daily [1]. Wearables are becoming increasingly influential in clinical decisions concerning diagnostic steps, drug prescriptions, and invasive strategies [2]. The ECG measuring method on those wearables is tied to the type of electrode used. Electrodes can be classified broadly into three groups: conductive wet, conductive dry, and non-contact insulated. Conventional cardiac rhythm monitoring employs Ag/AgCl wet electrodes, which offer high accuracy. However, wet electrodes rely on electrolyte gel and adhesive for good contact with the skin, which can cause irritation over time. Also, they cannot be reused, making them less suitable for long-term use [3]. While dry electrodes have direct contact with skin, they still rely on the presence of sweat to mimic the electrolyte gel. In comparison, non-contact electrodes provide several advantages since they are capacitively coupled and do not require direct contact with the skin, which can lead to higher comfort, application versatility and safety levels [4].

There are different methods for constructing non-contact electrodes. Flexible materials are commonly used, where the electrodes can be integrated into chest straps [5], [6]. In some cases, the electrode can be directly integrated into a garment, motivating investigation into the materials that compose these fabrics [7]. Beyond wearables applications, furniture such as chairs can also have non-contact electrodes installed [8], [9], in these cases the dielectric layer is also going to be defined by the clothing fabric being worn.

In this investigation, we have designed a non-contact electrode to conduct tests and evaluate its performance in the measurement of ECG signals in the human chest region. We also investigated the impact on ECG acquisition of different materials placed between the body and the electrode.

II. CAPACITIVELY COUPLED NON-CONTACT ELECTRODES

A. Principle of Operation

A non-contact electrode works based on capacitive coupling between the electrode and the human body.The electrode consists of a sensing plate and an interface layer made of electric insulating material. Unlike wet and dry electrodes, where electric charges are transferred through the skin to the electrode in a resistive path, in a capacitively coupled non-contact electrode, a displacement current moves the electric charges. Thus, DC signals are not transferred to the electrode, creating a high-pass filter (HPF) system where only AC signals will be transferred. The impedance of the non-contact electrodes is higher than that of wet and dry electrodes, so an analog front-end (AFE) with very high input impedance is required to avoid attenuation of the source signal. The use of an operational amplifier, also known as an active buffer, makes it an active electrode.

B. Modelling and Parameters Considerations

Fig. 1 represents the basic system body-electrode when a high-impedance input buffer is present. The electrode coupling capacitance is modelled as C_e. When an additional layer is in between the body-electrode interface, there will

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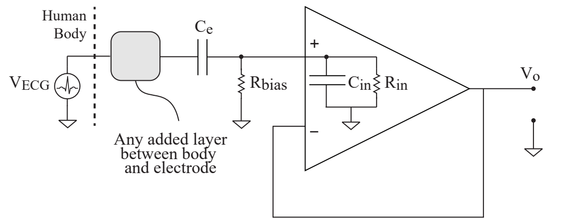

Fig. 1. Modelling of the ECG signal and coupling body-electrode capacitance C_e, with buffer operational amplifier input impedance parameters

be a new body-electrode interface impedance given by the series association of C_e and the impedance of the layer. The interface can also be filled with an air layer when the gap between the electrode and the body increases, making the effective coupling capacitance small, causing signal attenuation as can be seen in Eq. 1.

A very high-impedance input operational amplifier will have the real part of Z_{in} in the magnitude order of 10^{12} Ω, allowing to consider $R_{in} \gg R_{bias}$.

$$\frac{V_o(s)}{V_{ECG}(s)} = \frac{sC_e}{1/R_{bias} + s(C_e + C_{in})} \quad (1)$$

Considering $C_e \gg C_{in}$, R_{bias} and electrode-body capacitance C_e compose a HPF. The cut-off frequency should be set below the frequency of the signal's component of interest. For ECG, P and T waves are among the lowest frequency components. When C_{in} and C_e values are closer in magnitude, further signal attenuation will occur. Also, R_{bias} must be lower than $R_{bias,max}$ to ensure a proper DC operation voltage [10]. Eq. 2 describes the lower and upper bound values for bias resistance R_{bias}. Where $I_{bias,max}$ are the maximum bias current and V_{cc} is the supply voltage of the operational amplifier.

$$\frac{1}{2\pi f_{cut-off} C_e} \leq R_{bias} \leq \frac{V_{cc}}{I_{bias,max}} \quad (2)$$

The input HPF time constant is given by $C_e R_{bias}$ where a high value implies a long settling time, in some applications time constants of seconds are required [10]. Hence, the R_{bias} determination trades-off a good frequency response to ensure a given low cut-off frequency without a long settling time.

III. ELECTRODE DESIGN

A. Active Electrode Board Design

The electrode coupling capacitance, C_e, can be calculated using Eq. 3, assuming only a single dielectric material composes the electrode's insulation layer. The area A is the total surface area of the electrode. The distance d is the distance between the surface of the electrode and the body, ε_r is the relative permittivity of the insulating material and ε_0 is the vacuum permittivity.

$$C_e = \varepsilon_r \varepsilon_0 \cdot A/d \quad (3)$$

The proposed non-contact active electrodes consist of an insulating layer, a driven guard and a buffer amplifier. The electrode was made using a four layers PCB, as shown in Fig. 2. The bottom layer has a coating of solder mask, which acts as a high-impedance insulator. With the chosen sensing plate diameter of 30 mm and considering the solder mask thickness of 80 µm and $\varepsilon_r = 3.4$, from Eq. 3, $C_e \approx 265$ pF. In reality, this value will be smaller since the interface electrode-body is not perfect and air will fill the space in between. For the active buffer, as discussed in section II, the operational amplifier LTC624x (*Analog Devices, MA, USA*) was selected since it has a low bias current (typical 0.2 pA) and high input impedance with low input capacitance ($R_{in} = 10^{12}$ Ω, $C_{in} = 3$ pF). A $R_{bias} = 10\,\text{G}\Omega$ is the value selected for the electrode. An enclosure for the electrode was designed and 3D printed with polylactic acid (PLA) as shown in Fig. 3.

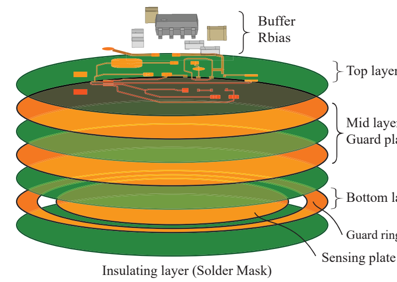

Fig. 2. Electrode PCB layers arrangement. The sensing plate has a diameter of 30 mm. The circular electrode PCB has a diameter of 35 mm and 1.6 mm thickness.

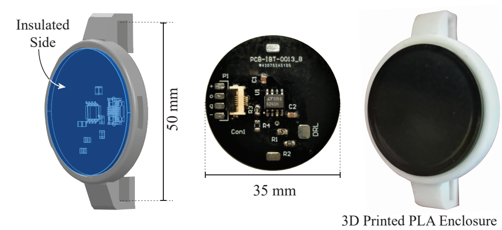

Fig. 3. Assembly of the electrode board and enclosure (left). Components populated in the electrode board (middle) and 3D-printed enclosure with PLA filament (right).

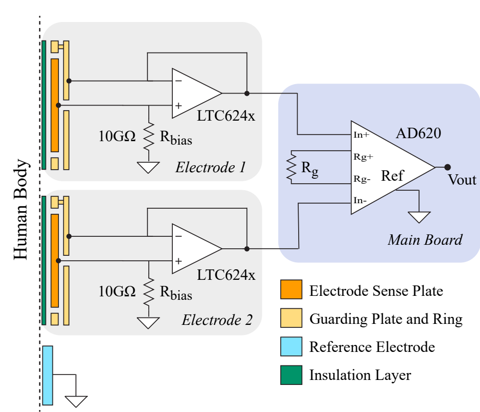

Fig. 4. Placement of electrodes and connection with the instrumentation amplifier. For a gain of 20 dB in the output signal, $R_g = 5.49\Omega$.

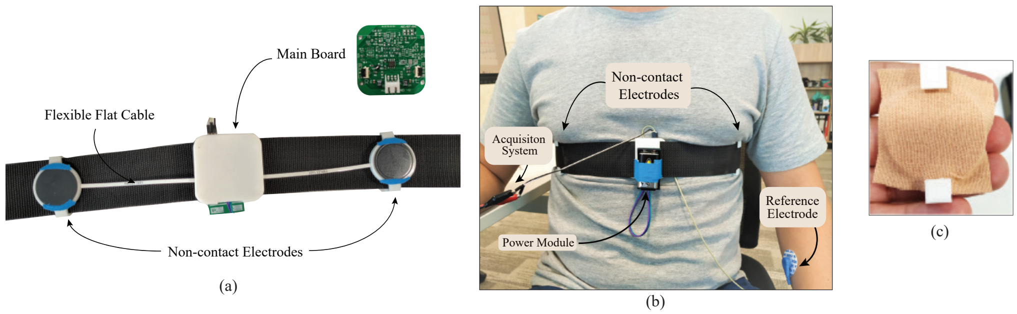

Fig. 5. (a) Internal view of the chest strap. (b) The subject wearing the chest strap. (c) The electrode used as a patch in the chest, with a single side tape applied to the back of the enclosure.

A driven guard is an important feature of a non-contact electrode to avoid further signal attenuation from parasitic capacitances around the PCB traces and components. A driven guard formed by the inner plane of the PCB and a ring around the sensing plate, as shown in Fig. 4, helps to nullify the effects of the parasitic capacitance on the board.

B. Analog Front-End (AFE) Main Board

The two non-contact electrodes, each with their buffer amplifiers, are interfaced with the main board where the signals are combined using an instrumentation amplifier to obtain an ECG signal. The instrumentation amplifier AD620 (*Analog Devices, MA, USA*) was selected. This component has good characteristics of common-mode rejection ratio (CMRR), $100\,\mathrm{dB}$ minimum (when $gain = 20\,\mathrm{dB}$), and gain adjustment. The gain was set to $20\,\mathrm{dB}$ and the output signal was passed directly to an analog-to-digital converter (ADC). The final AFE is shown in Fig. 4 and the main board is shown in Fig. 5(a). A third reference (wet) electrode is connected to the body and the reference ground. It's possible to have a fully non-contact measurement, but for this comparative investigation, the focus was on the impact of the interface of non-contact electrodes. Using a stable reference electrode for all of the scenarios avoids introducing the possible reference electrode variation. The power is supplied by batteries.

IV. PROTOTYPE AND TESTING

A. Chest Strap and Patch

To measure the ECG signal from the human chest region, a chest strap was prototyped using a $50\,\mathrm{mm}$ wide nylon strap. The electrode and AFE boards were attached to the strap using 3D-printed enclosures. Fig. 5(a) illustrates the designed tag shape enclosure that holds the electrode board and how it's integrated. The chest strap components' arrangement and how it is worn are shown in Fig. 5(b). For the patch electrode, a single-sided medical tape was utilized to attach the electrode to the chest, Fig. 5(c).

B. Signal Processing and Acquisition

The digitization of the analog signal coming from the output of the AD620 was carried out by a data acquisition system (DAQ) *ADInstruments PowerLab 2/26T*. The device has an effective 16-bit ADC and a sampling frequency of $1\,\mathrm{kHz}$ was set. The DAQ has its interface *ADInstruments LabChart Pro* that is responsible to acquire and manipulate the data. MATLAB® Version R2019a (*Mathworks, MA, USA*) was used for further data processing and plotting. No further analog or digital filtering is applied through signal acquisition or processing to preserve the performance of the different materials. However, all signals have the same baseline wander signal removed using a Savitzky-Golay 5^{th} degree polynomial interpolation, as shown in Fig. 6. The R-peaks detection is based on the threshold mapping of minimum peaks of $7\,\mathrm{mV}$ height and minimum peak distance of $0.5\,\mathrm{s}$.

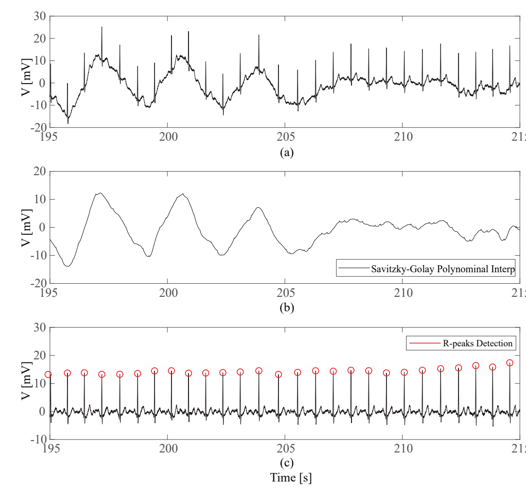

Fig. 6. A segment of the signal from the patch test. The raw ECG (a) has a breathing artifact, which ceases after $205\,\mathrm{s}$ when the subject holds the breath for a brief time. (b) shows the baseline wander signal. The signal on (c) is the subtraction of (a) and (b). Also, (c) shows the detected R-peaks

V. EXPERIMENTAL RESULTS

A subject wears the chest strap and the ECG signal is recorded for 300 seconds while sitting in a resting state. T-shirts made of pure cotton, merino and polyester were worn for each test. The patch was attached directly to the skin. The same subject participated in all the tests.

Fig. 7 shows the segments of the raw ECG signal of each test after subtraction of the baseline wander signal. The electrode was effective in detecting R-peaks in all materials,

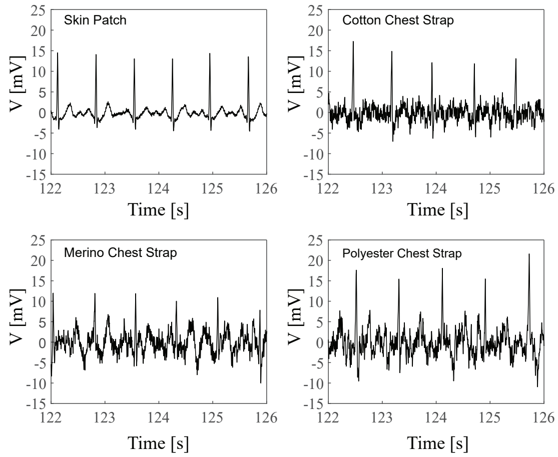

Fig. 7. Segments of ECG signal of each test.

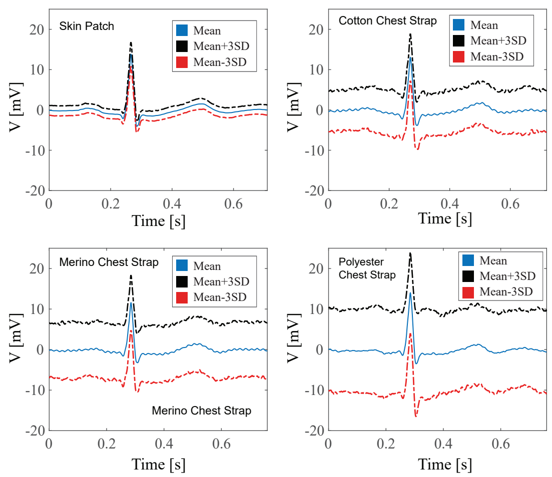

Fig. 8. ECG cycles mean and standard deviation (SD). Polyester presents a high R-peak value but has the highest standard deviation.

at different levels of accuracy. The results presented in Fig. 8 were processed based on the method used in [6]. The peaks' positions are mapped for the whole test signal duration and each ECG cycle is windowed around its peak and superposed. The mean and SD for sets of aligned samples are calculated for all cycles. Fig. 9, shows in the range of 1 Hz to 10 Hz, the close correspondence of the patch and cotton signal power, where polyester and merino have more signal power meaning that the noise is present in the ECG frequency range. Table I shows the comparison of the results between the different materials.

VI. Discussions and Conclusions

The developed non-contact electrode was tested as a patch and in a chest strap over T-shirts made of cotton, polyester, and merino. Common-mode noise and mains interference are present in all tests at different levels but can be improved using notch filters or having a low-frequency filter with a frequency cut-off of around 40 Hz, which is below the mains frequency. The patched electrode associated with conventional low and high-frequency filtering can not only

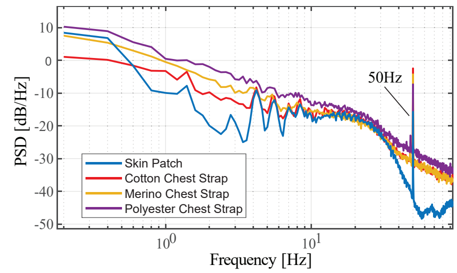

Fig. 9. The power density spectrum (PSD) of each raw signal shows the presence of 50 Hz mains noise.

TABLE I
Results Summary.

Material	Mean R-peak [mV]	Mains PSD [dB/Hz]	Accuracy [%] = (TP/(TP+FP+FN))
Patch	13.8	-15.6	100.0
Cotton	13.1	-2.6	99.5
Merino	11.6	-4.2	97.8
Polyester	14.0	-7.3	91.8

R-peaks: TP - True Positive, FP - False Positive, FN - False Negative

detect R-peaks but also provide clear information about ECG intervals and segments. The developed electrode is modular and might be integrated into different objects. Finally, the non-contact electrode demonstrated a high sensitivity to motion artifacts during all tests. Further investigation will focus on non-contact electrode performance where motion is present, exploring materials and processing algorithms.

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