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Evaluating the Dielectric Performance of Unblended PDMS Films for Non-contact ECG Sensing Applications

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Biocompatible polymer, Dielectrics, Non-contact ECG, Bioelectrodes, Unblended PDMS films.

ABSTRACT

Non-contact electrocardiogram (ECG) bioelectrodes need biocompatible dielectrics that can record cardiac signals for the long-term without skin irritation. Existing biomedical dielectrics used in non-contact ECG bioelectrodes exhibit rigidity, require expensive experiments, and induce noise interference during recording. This study explores unblended polydimethylsiloxane (PDMS) films as alternative biocompatible dielectrics for non-contact bioelectrodes. Unblended PDMS films were characterized using a simple deposition technique by increasing the proportion of the PDMS cross-linker while the polymer elastomer was kept constant. Next, energy dispersive X-ray (EDX) was performed to determine the elemental components for each PDMS film sample as well as their dielectric constants. The dielectric performance of PDMS film with the highest dielectric constant alongside fabric dielectrics and wet bioelectrodes were experimented in recording ECG signals. The EDX analysis revealed different compositions, by percentage of carbon, silicon, and oxygen, which varied with the increase in proportion of the cross-linker. Moreover, ECG measurements confirmed that the unblended PDMS dielectric film with the highest cross-linker ratio (10:2) performed sufficiently comparable with other bioelectrodes experimented with, demonstrating their potential as a cost-effective and biocompatible solution for non-contact bioelectrodes in long-term ECG monitoring.

1. INTRODUCTION

Cardiac diseases are non-communicable lifetime illnesses that affect all demography of people across different genders and ages. Heart abnormalities remain a health challenge and the highest cause of death globally [1, 2]. Patients suffering from heart problems are medically advised to continuously monitor their health status to early detect heart dysfunction to avoid sudden heart failure or harm to organs that rely on the cardiac conduction system [3, 4]. Electrocardiography is the clinical technique used to acquire the bioelectrical signals generated by the heart in the form of an interpretable waveform [5]. To date, Cardiologists use the electrocardiogram (ECG) as a medical diagnostic tool to provide remedy for heart problems.

ECG bioelectrodes are of various types: wet, direct contact, micro-needles, and non-contact (capacitive), as classified in Figure 1. The silver/silver-chloride (Ag/AgCl) bioelectrodes were the foremost and till today the most popularly employed non-invasive biopotential sensors for acquiring heart electrical signals [6, 7].

Over the past years, there has been a growing effort towards long-term ECG monitoring for patients with critical heart problems. Monitoring of heart activities for extended period ensures physicians do not miss any useful indicators of irregular heart functioning [8-11].

The United States (US) Food and Drug Administration (FDA) has recommended that "any biomedical electronic for health monitoring needs to be biocompatible in the environment applied" [12]. Therefore, the regulation entails that the bioelectrode skin region (BSR) which is the point for sensing biopotentials, should be free from unhealthy reactions. However, in long-term monitoring of ECG, environmental factors would cause the skin to generate exudates and moistness [13].

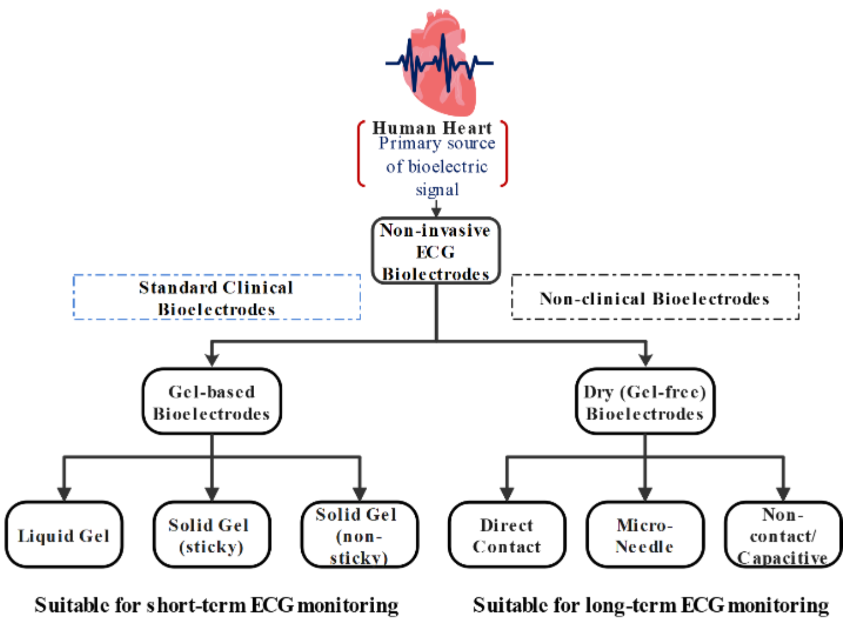


Figure 1. *Categories of non-invasive ECG bioelectrodes*

Dry, direct contact bioelectrodes operate through physical contact with the skin. They can be used in long-term ECG monitoring because they guarantee ease without skin preparations and offers the safest approach to prevent skin irritations [14, 15]. For the non-contact or capacitive biosensing, ECG is measured via a dielectric material. The bioelectrodes can effectively operate without direct skin contact and need for stringent clinical procedures.

Research on non-contact mode of acquiring ECG is not recent. Early examples of dry capacitive medical bioelectrodes were demonstrated by Richardson *et al.* in 1968 [16], Lopez and Richardson in 1969 [17], and Potter and Menke in 1970 [18]. The earlier research on non-contact ECG bioelectrodes utilised metal compounds with high dielectric constant as dielectrics [16,19-22]. However, majority of these metal-based dielectrics involve complex methods such as anodic reaction [23], sputtering [24] and chemical deposition [25, 26], and required expensive equipment.

An important aspect of non-contact ECG recording is high coupling-capacitance is essential to obtain a biosignal with good resolution. Theoretically, equation (1) shows that the coupling-capacitance largely depends on the properties of the dielectric material [27], as well as the active contact area of the bioelectrode.

$$C = \frac{\varepsilon_o \varepsilon_r}{d} \quad (1)$$

Where, C is the coupling capacitance, ε_o is the permittivity of air, ε_r is relative permittivity of the dielectric material, and d is the thickness of dielectric.

Cotton is another commonly used fabric dielectric material. However, its porosity causes unstable dielectric behavior in long-term ECG recording thereby causing unreliable biosignal measurements [28]. Other types of biomedical dielectrics are polymers. Polymers are inert materials with low dielectric constants and are employed as insulators [29, 30]. Currently, there is an increase in the exploration of polymers for different biomedical applications because they are stretchable, skin conformable and non-toxic to human tissue [31-33].

PDMS is a good insulator, but it has poor dielectric performance. Many techniques have been proposed to enhance the weak dielectric properties of PDMS using nanoparticles as composites for non-biomedical applications [34-38]. However, the techniques employed require expensive nanoparticles, hazardous laboratory procedures, and expensive equipment. Strict safety precautions are also required to characterize these materials. In some instances, the experiment might fail, as reported in [39], leading to a waste of time and resources.

PDMS is commercially available as a single kit but in two parts: elastomer base and curing agent (cross-linker). Mixing of these two parts results in a change in molecular structure and dielectric properties of PDMS. It is feasible to modify the dielectric properties of unblended PDMS without the need for expensive nanoparticles. By adding more of the cross-linker, the polymer chains get shorter and structurally affected. This study intends to explore the dielectric performance of unblended PDMS films for non-contact ECG sensing applications.

2. MATERIALS AND METHODS

The unblended PDMS dielectrics employed in this study were characterized by manual deposition outlined in our previous work [40]. The electrical characterization demonstrated PDMS films with dielectric performance well-suited for this study. We provided a schematic of the experimental procedures in Figure 2 and harnessed the PDMS dielectrics for non-contact ECG recording.

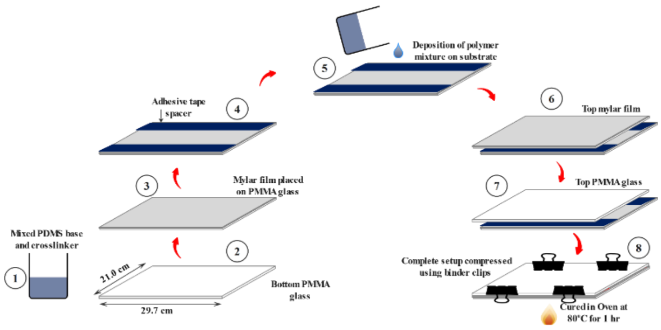


Figure 2. *Fabrication process of thin PDMS dielectric films [40]*

Energy-dispersive X-ray spectroscopy (EDX) was included in this study for elemental composition to confirm that the unblended PDMS has a characteristic chemical structure consisting of silicone, oxygen, and carbon. EDX analysis was employed to confirm the presence of these elements and their relative proportions. This verification step was essential to ensure the integrity of the PDMS films and rule out the presence of significant impurities that could influence PDMS dielectric properties and experimental results. Table 1 summarizes the weight-base scale determining each polymer mix ratio used in fabrication of the PDMS dielectrics of various thickness.

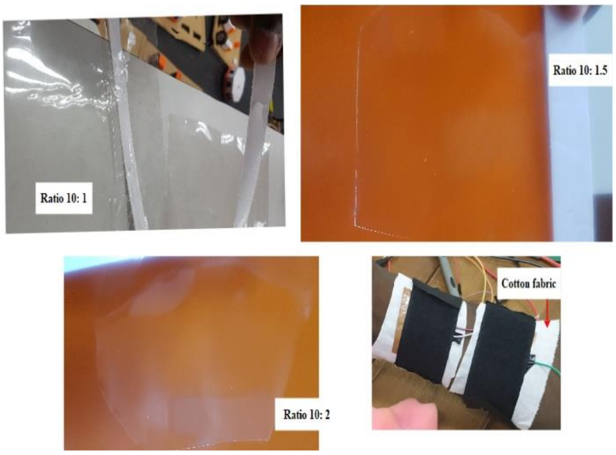
Table 1. *Preparation ratio of PDMS dielectrics*

Ratio	Weight of Base (g)	Weight of Crosslinker (g)	Thickness of Dielectrics (mm)
10:1	38.065	3.807	0.14; 0.28;
10:1.5	27.190	4.079	0.42; 0.56;
10:2	54.862	10.972	0.60

2.1. Experimentation of PDMS Dielectrics and Recording of ECG

ECG data were acquired in three different modes; wet, dry, and capacitive using the OpenBCI 8-channel Cyton biosensing board designed for recording biopotential signals. The board offers a graphic user interface (GUI) tool for recording, visualizing, and streaming biopotential signals from it. Figure 3 illustrates the setup for non-contact ECG recording via fabric and PDMS dielectrics.

In non-contact (capacitive) ECG recording, a dielectric which enables a high coupling



capacitance is essential, as expressed in (1). As reported in [40], PDMS dielectrics of 0.14 mm thickness characterized from the polymer mix ratio of 10:2 yielded the highest coupling capacitance. Therefore, it was employed in this study. Next, we evaluated the ECG recorded by our proposed PDMS dielectrics compared with those obtained from fabric dielectrics and the standard Ambu Ag/AgCl whitesensors. The Ambu bioelectrodes used in this experiment are from Medico International Electrodes, Ltd, Denmark. Table 2 provides the specifications of the bioelectrodes, and setup employed in the current work.

Prior to ECG recording, no skin preparation was performed on the subject, a male of 42 years old. Furthermore, we separately conducted the experiments under dry and sweaty skin conditions to examine the influence skin exudates on ECG recordings.

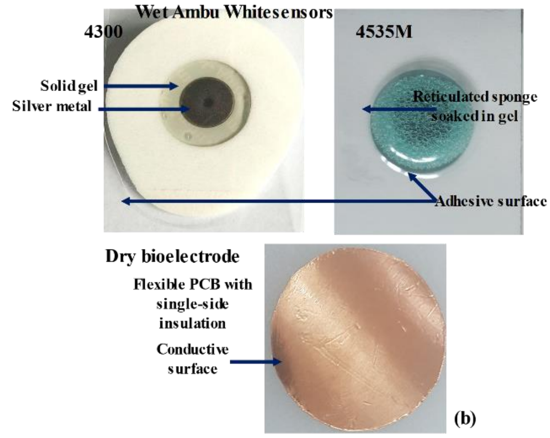


Figure 3. *(a) ECG recording setup and (b) bioelectrodes (wet and customized flexible PCB)*

3. RESULTS AND DISCUSSION

Here, we provide some experimental evaluation on performance of the proposed unblended PDMS dielectrics. Figure 4 shows samples of the PDMS fabricated from different polymer mix ratios.

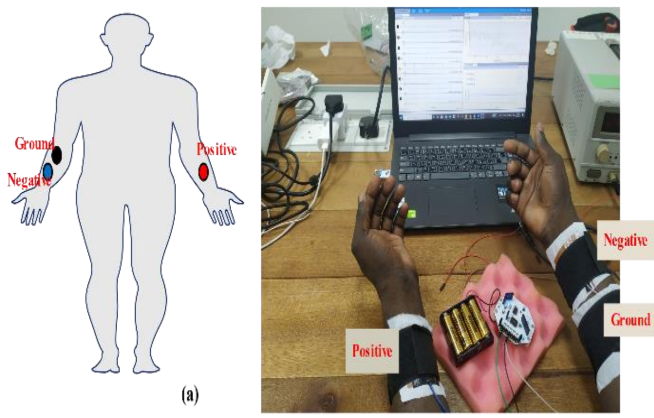


Figure 4. *Samples of PDMS films fabricated from weight mix ratio of silicone elastomer and its curing agent*

EDX elemental analysis performed on unblended PDMS films show similar surface morphology confirming absence of contaminants, as shown in Table 2.

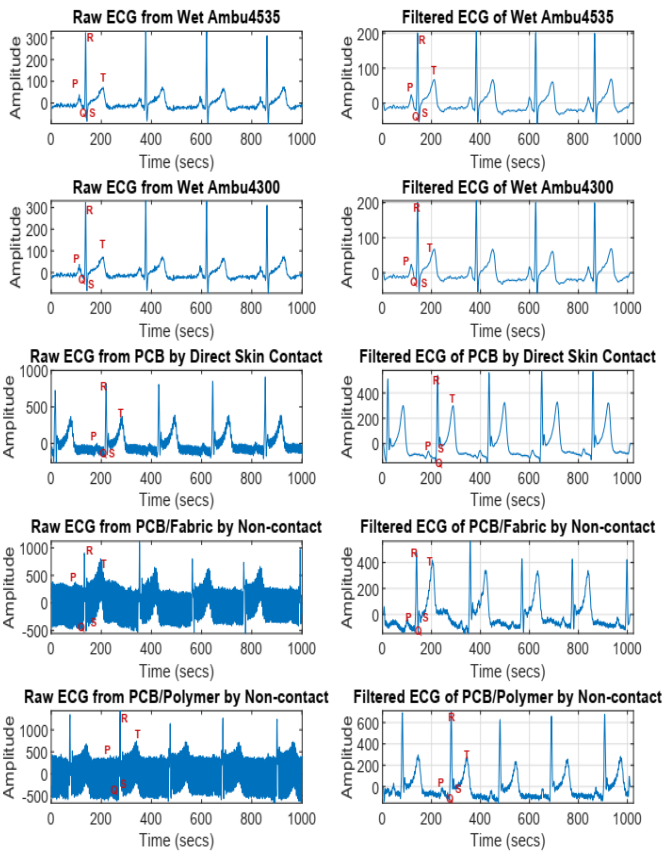
Table 2. *Elemental percentage in unblended PDMS dielectric films*

PDMS Mix Ratio	Carbon C (%)	Silicone Si (%)	Oxygen O (%)
10:1	38.9	34.6	26.6
10:1.5	38.2	34.3	27.6

10:2	39.3	31.3	29.4

The elemental constituents of PDMS based on percentage weight (wt %) indicate a difference in quantity of silicon (Si), carbon (C), and oxygen (O). Parvez, *et al.* [41] also reported comparable observations in their experiment. The EDX analysis shows that silicon is the dominant element, followed by oxygen then carbon. Besides, there are slight differences in the percentage concentration of elements as the proportion of the curing agent increases. However, this study is limited because the maximum PDMS mix ratio for which PDMS dielectric films can maintain flexibility was not ascertained.

Next, Figure 5 shows the ECG signal recorded by the reticulated Ambu 4535M and the solid gel whitesensor 4300 bioelectrodes after filtering of artifacts and electrical interferences. Since both



bioelectrodes use conductive gels, they generated a less noisy ECG with visible P-QRS-T peaks.

Figure 5. *ECG signals recorded with different bioelectrodes under dry skin condition.*

The study also examined the performance of the proposed unblended PDMS dielectrics for non-contact ECG bioelectrodes and compared them with direct contact mode and fabric dielectric under dry and sweaty skin conditions. Figure 5 displays the ECG recorded for dry and unprepared skin for direct contact and non-contact mode with fabric and unblended PDMS applied as the dielectrics.

The appearance of the unprocessed (raw) ECG signals indicates the severe impact of the powerline interference on the biosignal recorded by the fabric and unblended PDMS dielectrics. The electrical noise picked up by bioelectrode cables can be seen raw ECG (Figure 5) but was attenuated. Although the ECG is of good quality, it exhibits low frequency noise resulting from the hardware recording device.

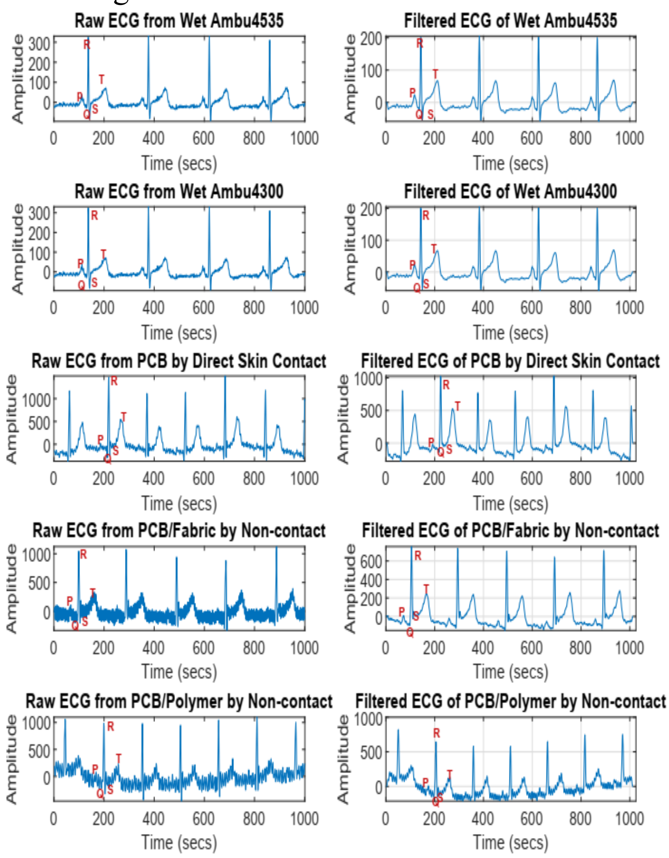


Figure 6. *Samples of ECG recorded with different bioelectrodes on sweaty skin condition.*

The appearance of the unprocessed (raw) ECG signals indicates the severe impact of the powerline interference on the biosignal recorded by the fabric and unblended PDMS dielectrics. The electrical noise picked up by bioelectrode cables can be seen raw ECG (Figure 5) but was attenuated. Although the ECG is of good quality, it exhibits low frequency noise resulting from the hardware recording device.

It has been demonstrated that sweat has a strong influence on skin impedance and alters the behavior of dielectric materials. Figure 6 shows the ECG recorded on sweaty skin. Human sweat (a weak electrolyte) can minimize skin impedance. Also, when absorbed by a porous material such as fabric, can affect its electrical and dielectric characteristics. Except for the wet ECG, the raw and filtered ECG recorded in direct contact mode and fabric dielectric showed less noise. Unlike the porous fabric dielectric, the ECG recorded by the non-porous PDMS dielectric revealed noisy biosignal with some baseline drift. Even after filtering, the ECG still contains low-frequency noise and baseline drift. As explained previously, PDMS dielectrics under certain conditions showed high skin impedance than fabric dielectrics due to their poor porosity.

Also, the PR, QR, QT, and RR are vital ECG parameters that are used screening cardiac abnormalities. Table 3 presents the intervals acquired via the OpenBCI graphic user interface (GUI).

Table 3. *Comparison of ECG parameters recorded by the OpenBCI Cyton board*

ECG Mode	**Bioelectrode**		**Skin Condition**			**ECG Parameters**			
	Type	Area cm^2	Gel	Dry	Sweaty	PR ms	QR ms	QT ms	RR ms
Wet	Ambu whitesensor	0.79	Yes	-	-	50.0	13.0	95.0	208.0
			Yes	-	-	35.0	24.0	81.0	204.0
Direct contact	Flexible PCB	31.50	-	Yes	-	36.5	18.0	86.5	206.0
Non-contact	PCB/Fabric		-	Yes	-	33.5	29.5	98.5	206.0
	PCB/PDMS		-	Yes	-	31.0	18.0	103.5	208.0
	Flexible PCB	31.50	-	-	Yes	40.0	23.0	81.0	162.0
	PCB/Fabric		-	-	Yes	31.0	21.0	120.0	159.9
	PCB/PDMS		-	-	Yes	29.0	21.0	84.5	156.0

There are clear disparities of duration for each ECG segment which could be attributed to the physiological condition of the subject, powerline noise, and the ECG recording system that was employed.

4. CONCLUSION

It is now proven that the dielectric performance of unblended PDMS films can be enhanced by varying polymer mix ratios. The proposed PDMS dielectrics showed comparable performance to fabric dielectrics. However, the non-porousness of the PDMS dielectrics creates a sweat layer that accumulates underneath the dielectric. This layer of sweat influences the coupling distance between the bioelectrode and skin. That said, the performance of the PDMS dielectric could be compensated by decreasing its thickness, using bioelectrode with better conductivity and a large contact area.

Furthermore, this research has provided invaluable insights to enhance the performance of a biocompatible polymer dielectric for medical applications without the use of expensive enhancement techniques. This research showed that varying the concentration of the polymer crosslinker affects the morphology and dielectric performance of the unblended PDMS. Experimental results on ECG recordings also indicated that PDMS dielectrics fabricated from a high crosslinker ratio can perform comparably to fabric dielectrics. Nevertheless, the performance of the bioelectrode was strongly influenced by the porosity of dielectric material, skin condition, type of bioelectrode, and contact area.

Future work to improve on this new development is suggested exploring higher mix ratios of the silicone elastomer and its crosslinker as well as increasing the recording time for the ECG to examine the changes in the performance of PDMS dielectrics.

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Conflict of Interest Declaration

The authors declare no conflict of interest.

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