UNBLENDED POLYDIMETHYLSILOXANE DIELECTRICS FOR
NON-CONTACT ELECTROCARDIOGRAPH BIOELECTRODES

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DEDICATION

This thesis is dedicated to the Muslim Ummah.

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ABSTRACT

Dielectric materials play crucial roles in the non-contact recording of electrocardiograms (ECGs), allowing cardiac physiological signals to be monitored and abnormalities detected early. Previous dielectrics are rigid, unstable, require expensive fabrication, and induce severe noise. PDMS is a polymer with remarkable biomedical properties suitable for non-contact bioelectrodes. However, the poor dielectric capability of PDMS has led to costly attempts to improve it. This research introduces an affordable technique to explore and characterize unblended PDMS films as dielectrics for non-contact ECG bioelectrodes by varying the weight mix ratio of Sylgard 184^{TM} silicone elastomer and its crosslinker. Capacitance and relative permittivity were measured and estimated using the parallel-plate technique in the frequency range of the LCR meter. Skin impedance values were also measured for different skin conditions using the Hioki impedance analyzer test frequency of 4 Hz - 1 MHz. Non-contact ECG measurements are affected by morphology of the dielectric, bioelectrode conductor, contact area, and skin conditions. To investigate the impact of these factors on skin impedance, the skin bioelectrode interface was modeled using equivalent circuits for wet, direct contact, and non-contact modes and analyzed using least-squares non-linear curve fit. Finally, the proposed approach was verified by recording ECG using different bioelectrodes and dielectrics. Due to susceptibility to motion artifacts and electrical interferences, digital filters were introduced to improve the quality of ECG recorded. The results demonstrate the effectiveness of the proposed method for improving the dielectric performance of unblended PDMS films, with a steady increase in capacitance (50.53 pF to 102.86 pF) and relative permittivity (0.19 to 0.69) observed with an increase in the proportion of the crosslinker. Good agreement was found between the measured skin impedance and equivalent circuit models, but differences exist in estimated circuit parameters. The unblended PDMS films with the highest mix ratio (10:2) successfully recorded visible P-QRS-T peaks in ECG. Skin conditions, bioelectrode conductivity, dielectric thickness and porosity, and the OpenBCI board filtering parameters affect the ECG recordings. This study confirms the potential of unblended PDMS films for non-contact bioelectrodes to detect heart abnormalities early.

ABSTRAK

Biolektrod tanpa sentuh telah digunapakai bagi pemantaun isyarat fisiologi jantung disamping mengesan keabnormalan jantung. Di mana, bahan dielektrik merupakan komponen utama yang digunakan dalam merakan isyarat elektrokardiogram (ECG) tanpa sentuh. Sebelumnya, dielektrik bukan polimer adalah bersifat tegar, tidak stabil, memerlukan kos fabrikasi yang mahal serta menghasilkanhingar yang teruk. Satu bahan dielektrik yang dipanggil sebagai polimer polydimethylsiloxane (PDMS) adalah tidak berbahaya kepada tisu manusia, memberikan sentuhan kulit ergonomik yang lebih baik terhadap artifak gerakan dan cas triboelektrik, dan ianya berpotensi untuk digunakan sebagai dielektrik bioelektrod ECG tanpa sentuhan. Oleh kerana PDMS mempunyai pemalar dielektrik yang rendah, beberapa percubaan untuk memperbaiki sifat penebat PDMS telah dibuat. Penyelidikan ini telah meneroka dan mencirikan filem PDMS yang tidak dicampur sebagai dielektrik untuk digunakan sepagai bioelektrod ECG tanpa sentuhan. Dengan mengubah nisbah campuran berat elastomer silikon Sylgard 184TM dan penyambung silangnya, filem PDMS didepositkan secara manual pada Mylar dan kesan dielektrik bagi ketebalan filem yang berbeza akan diperiksa dan dianalisis. Jumlah kapasitan dan kebolehtelapan relatif adalah antara parameter dielektrik penting yang diukur dan dianggarkan menggunakan teknik plat selari untuk frekuensi ujian yang berbeza dalam julat frekuensi meter LCR pratetap (100 Hz - 100 kHz). Disamping itu, nilai impedans kulit diukur untuk keadaan kulit yang berbeza iaitu dengan menggunakan julat impedans Hioki antara 4 Hz hingga 1 MHz. Antra faktor penting yang akan mempengaruhi pengukuran fisiologi bukan sentuhan adalah seperti morfologi dan keliangan dielektrik, jenis konduktor bioelektrod dan kawasan sentuhan, dan keadaan kulit. Bioelektrod kulit telah dimodelkan sebagai litar setara bagi beberapa mod ECG seperti basah, sentuhan langsung dan bukan sentuhan bagi mengkaji kesan faktor biopenderiaan pada impedan kulit, dan dianalisis menggunakan kesesuaian lengkung tak linear kuasa dua terkecil. Akhir sekali, ECG telah direkodkan dengan menggunakan bioelektrod yang mempunyai dielektrik yang berbeza untuk pengesahan prestasi. Oleh kerana rakaman ECG mudah dipengaruhi oleh artifak gerakan dan gangguan elektrik, maka penapis digital telah diperkenalkan untuk memproses dan memeriksa kualiti ECG yang direkodkan oleh papan biopenderia OpenBCI Cyton pada kulit bergel, kering dan berpeluh. Hasil kajian menunjukkan keberkesanan kaedah yang dicadangkan untuk meningkatkan prestasi dielektrik filem PDMS yang tidak dicampur. Sepanjang julat frekuensi 100 Hz hingga 100 kHz, peningkatan yang stabil dalam bahagian pemaut silang mengakibatkan peningkatan nilai kapasitan daripada 50.53 pF kepada 102.86 pF dan kebolehtelapan relatif daripada 0.19 kepada 0.69. Selain itu, kesesuaian yang baik dengan model litar setara telah diperoleh untuk impedan kulit yang diukur, akan tetapi terdapat perbezaan antara parameter litar yang dianggarkan. Perbandingan prestasi antara dielektrik dan bioelektrod telah disahkan dengan melihat rakaman ECG dengan puncak P-QRS-T yang boleh dilihat menggunakan filem PDMS yang tidak dicampur dengan nisbah campuran tertinggi (10:2). Hasil kajian juga menunjukkan bahawa keadaan kulit, jenis bioelektrod dan kekonduksian mempengaruhi kadar dielektrik dan ukuran fisiologi. Pada masa yang sama, kualiti ECG yang ditapis didapati bergantung pada masa penderia OpenBCI, frekuensi pensampelan dan potongan serta susunan penapis. Buat pertama kalinya, kajian ini telah mencadangkan dan mengesahkan bahawa potensi dielektrik PDMS yang tidak dicampur boleh digunakan dalam bioelektrod bukan sentuhan untuk pemantauan jangka panjang dan pengesanan awal keabnormalan jantung.

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LIST OF ABBREVIATIONS

AAMI	-	American Association for Medical Instrumentation
Ag-NW	-	Silver Nano-Wire
ANSI	-	American National Standards Institute
BCG	-	Ballistocardiograph
CB	-	Carbon Black
CuNi	-	Copper-Nickel
CVD	-	Cardiovascular Disease
ECG	-	Electrocardiogram
EDX	-	Energy dispersive X-ray spectroscopy
EEG	-	Electroencephalogram
EMG	-	Electromyogram
FESEM	-	Field Emission Scanning Electron Microscope
FFT	-	Fast Fourier Transform
FIR	-	Finite Impulse Response
GO	-	Graphene Oxide
GUI	-	Graphic User-Interface
IIR	-	Infinite Impulse Response
IRT	-	Infrared Thermography
MEMS	-	Micro-Mechanical Systems
MIM	-	Magnetic Induction Monitoring
MNB	-	Micro-Needle Bioelectrode
MWCNT	-	Multi-walled Carbon Nano-Tube
OpenBCI	-	Open Brain-Computer Interface
PCB	-	Printed Circuit Board
PDMS	-	Polydimethylsiloxane
PMMA	-	Poly(methyl methacrylate)

PPGI	-	Photoplethysmography Imaging
PVC	-	Polyvinyl Chloride
SC	-	Stratum Corneum
WHO	-	World Health Organization

LIST OF SYMBOLS

A	-	Contact area
C	-	Capacitance
C_B	-	Capacitance of gel-bioelectrode interface
C_E	-	Capacitance of epidermis
C_n	-	Capacitance of non-porous dielectric
C_p	-	Capacitance of porous dielectric
d	-	thickness of dielectric
E_v	-	Half-cell potential
f_c	-	Cut-off frequency
$f_s amp$	-	Sampling frequency
k	-	Dielectric constant
M	-	Filter length (number of filter coefficients)
R_B	-	Resistance of gel-bioelectrode interface
R_C	-	Resistance of conductive gel
R_E	-	Resistance of epidermis
R_n	-	Resistance of non-porous dielectric
R_p	-	Resistance of porous dielectric
R_p	-	Resistance of dermis and subcutaneous tissue
Z	-	Impedance
ε_o	-	Relative permittivity of free-space
ε_r	-	Relative permittivity of dielectric material
δp	-	Pass-band ripple
δs	-	Stop-band ripple
Δf	-	Normalized transition width

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CHAPTER 1

INTRODUCTION

1.1 Problem Background

Cardiovascular diseases (CVDs) are non-communicable lifetime illnesses and the highest cause of deaths globally [1]. At present, CVD has a global presence as the supreme health threat to all demography of people across different genders and ages [2]. In 2019, the World Health Organization (WHO) reported ischaemic-CVD as the most deadly disease (Figure 1.1(a)) [3]. Furthermore, in 2020, the Malaysian Department of Statistics reported that ischaemic-CVD is the leading cause of death at the national level, as shown in Figure 1.1(b) [4]. Consequently, many countries bear a heavy financial burden to manage affected citizens with heart abnormalities [5–8]. In 2022, the the Ministry of Health Malaysia published a report confirming that the country spent RM3.93 Billion as the total healthcare cost for CVDs [9].

Heart diseases are silent-killers. In light of the primary health concerns, patients with a high risk of CVD are highly recommended to regularly monitor their health status to detect heart dysfunction and prevent sudden heart failure or damage of organs that rely on the cardiac conduction system [10, 11].

Electrocardiography is a non-invasive medical procedure for acquiring bioelectrical signals generated by the heart in the form of an interpretable waveform [12]. To date, Cardiologists use the electrocardiogram (ECG) to identify heart problems such as infarction, ischemia, and other cardiac abnormalities [13–15]. Since the coronavirus was reported in 2020, the ECG has been attracting the interest of the healthcare communities. For example, an irregularity in the ECG waves pattern indicates infection by the COVID-19 [16]. Therefore, there is a universal increase in the application of non-invasive ECG recording systems to evaluate the potential threat and influence of cardiac diseases on the COVID-19 prognosis [17–22].

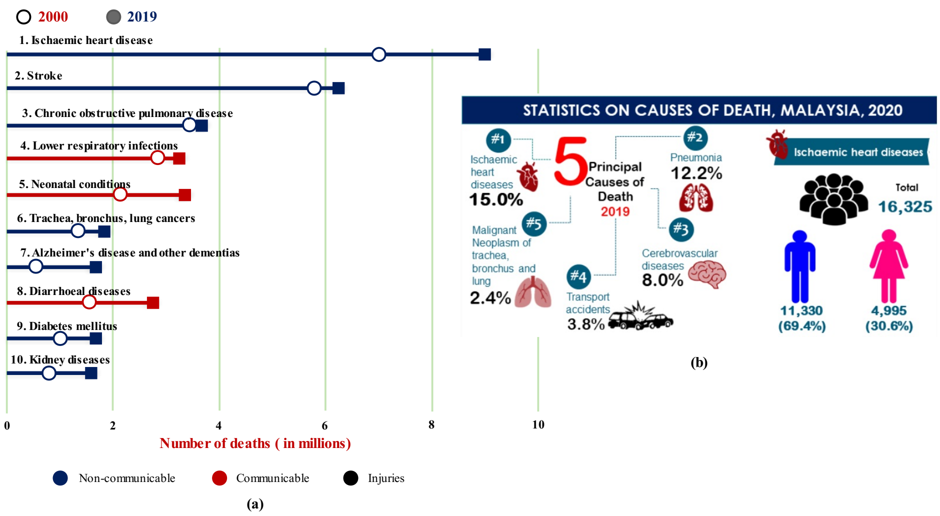

Figure 1.1 The Ischaemic-CVD is the principal cause of death (a) worldwide as reported by the WHO in 2019 [3] and (b) domestically by the Malaysian Department of Statistics [4] (modified).

Over a century since Augustus Waller invented the first heart monitoring device to record heart rhythms using five zinc bioelectrodes [23]. In 1893, the Dutch Physiologist improved the work of Waller and successfully reduced the number of ECG bioelectrodes from five to three [24, 25]. The conventional approach of measuring heart biopotential is the 12-lead ECG system [26] and Holter monitor [27]. Despite the advantages of these clinical ECG systems, they rely on non-invasive bioelectrodes that are effective for a short period and restricted to clinical settings. As such, patients that require constant monitoring are at high risk when they experience sudden heart failure [28].

In physiological signal monitoring, non-invasive bioelectrodes are necessary to acquire the electrical potentials on the skin. Regarding the placement of bioelectrodes, bioelectrodes are classified as invasive when embedded within human tissue to record physiological signals and non-invasive when attached to the outer skin layer [29]. The focus of this study is entirely on non-invasive bioelectrodes. Non-invasive ECG bioelectrodes are sub-classified, based on skin conditions, into wet and dry bioelectrodes. The wet silver/silver-chloride (Ag/AgCl) bioelectrodes were the foremost and the most popular bioelectrodes for acquiring heart electrical signals [30, 31]. The wet bioelectrodes require conductive gel before use. The gel creates a low impedance path for easy movement of ionic current from the cardiac tissues to produce a detectable electrical potential with high resolution.

Over the past years, there has been a growing effort towards long-term ECG monitoring. The long-term monitoring ensures physicians do not miss any useful indicators of heart abnormalities and allows early diagnosis of critically affected patients [32–36]. In long-term ECG monitoring, the performance of the wet bioelectrodes is greatly restricted. The conductive gel may dehydrate after extended use, leading to severe signal instability and attenuation [30]. Besides, the placement of wet bioelectrodes requires clinical experience and sensitive skin abrasion. When the ECG is frequently recorded, the chemical reaction between the gel and metal conductor can result in skin irritations and contact dermatitis infections [37, 38]. Since wet bioelectrodes are not often reusable and biodegradable, their waste can contribute to substantial environmental pollution [36, 39].

Studies have proposed that by constantly monitoring the ECG, many heart patients could be better examined and protected from sudden death [40–43]. Unfortunately, the wet bioelectrodes are constrained by the problems mentioned earlier. Dry bioelectrodes are biopotential sensors capable of operating without skin abrasion and conductive gels. In the last decade, non-invasive bioelectrodes have been researched to support telemedicine and real-time monitoring of physiological activities [44,45]. In addition, patients benefit from constant health monitoring and enjoy better comfort [46]. Dry bioelectrodes are of three types: direct contact, micro-needles, and non-contact (capacitive), as classified in Figure 1.2.

Direct contact bioelectrodes are alternative to the wet bioelectrodes and 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 [47, 48]. Several studies have explored direct contact bioelectrodes utilizing diverse materials. Some of the bioelectrodes were implemented using stainless-steel [49, 50], brass [51], silver, gold [52], and silver-coin [53, 54]. Despite being good conductors, metal-based ECG bioelectrodes have drawbacks. They are rigid, un-ergonomic, induce motion artifacts, and are highly vulnerable to

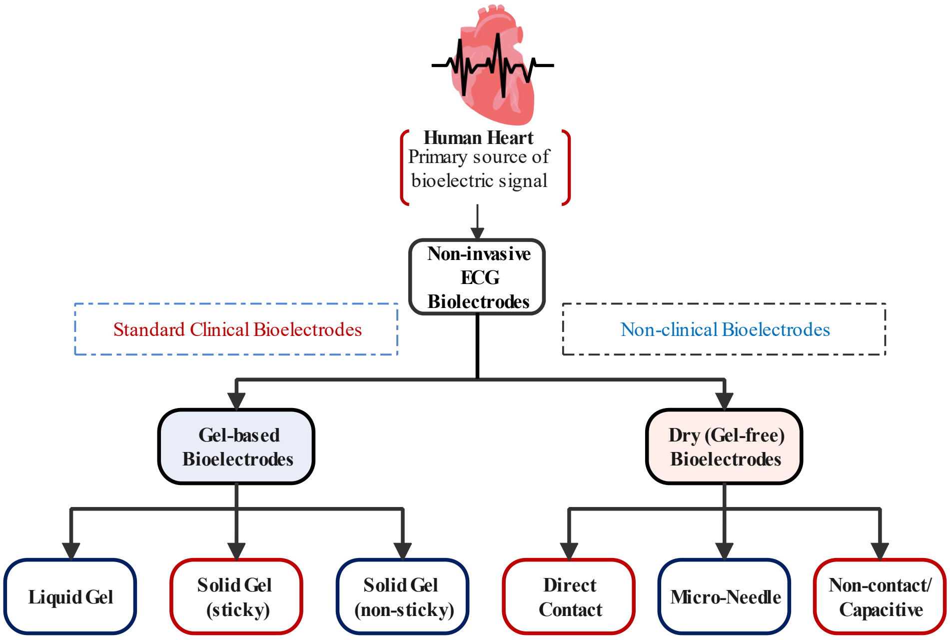

Figure 1.2 Categories of non-invasive ECG bioelectrodes

electrical interference [30]. Furthermore, the performance demonstrated by direct contact bioelectrodes is dependent on skin condition. For instance, dry skin exhibit high impedance than when moisturized by sweat, which is responsible for unstable biosignal quality in dry ECG bioelectrodes [55].

Fabrics are the most accessible and used materials by humans. Fabrics have been investigated by many researchers for wearable and long-term ECG monitoring, such as conductive-threads, inks, and printed-fabrics [56]. Many other bioelectrodes include yarns of silver-coated threads stitched to fabrics [57, 58] and screen-printed fabrics using conductive inks [59, 60]. Fabric-based bioelectrodes have the advantage of being soft and flexible. However, they still require excessive pressure to attach them to the skin firmly. Furthermore, skin exudates could produce unpleasant skin reactions [61] and cause the coated ink to fade, resulting in low and unstable ECG quality [62, 63]. Also, triboelectric charges are more frequent in fabrics.

Materials and fabrication techniques have greatly influenced bioelectrode technology in the past decade. Polymers have inspired the development of polymeric materials and nano-metallic conductors with special properties for biomedical applications. Typical examples are the direct contact bioelectrodes fabricated using multi-wall carbon nanotube (MWCNT)/polydimethylsiloxane (PDMS) [64, 65]; polypyrrole (PPy)/patterned-vertical carbon nanotube [66]; silver nanowires, MWCNT, and adhesive PDMS [67]; graphene-oxide (GO)/Ag-NWs [68]; and polyvinyl butyral/Ag-NWs [69]. The disadvantages of nano-materials are that fabrication methods are expensive and require hazardous laboratory experiments.

In skin anatomy, the stratum corneum (SC) is the outer-most layer of the epidermis. When bioelectrodes are applied in direct contact mode, they are placed on the surface of the SC. The SC has a dry structure, short lifespan, and the ability to regenerate [70]. The thickness of the SC varies with the region of the body (6 – 40 µm) [71] and is weakly conductive [72]. This makes the SC the primary cause of high skin impedance that hinders biopotential signal recording [73, 74]. In wet ECG, the SC is usually abraded to minimize its influence. Dry bioelectrodes with micro-needles can overcome the restrictions of the SC layer. The micro-needles are invented to penetrate the SC layer and minimize its influence on the skin impedance. In addition, the micro-needles ensure stable contact with the skin and prevent motion artifact [75, 76]. Examples of the dry micro-needles bioelectrodes reported can be found in the studies conducted by [77–81]. However, the micro-needles can cause inconveniences such as pains, bleeding, and infection in prolonged use. Moreover, the fabrication of micro-needles requires stringent laboratory procedures and expensive facilities [75].

Non-contact ECG bioelectrodes were researched long ago and are presently investigated because of their potentials [82–84]. In contrast to the dry, direct skin contact bioelectrodes, the non-contact bioelectrodes are capable of recording ECG from unabraded skin via dielectric materials. Dielectrics provide non-contact bioelectrodes the advantage to operate for an extended period with much convenience and assurance to detect heart abnormalities in good time [85]. Table 1.1 shows a comparison of different ECG bioelectrodes.

Table 1.1 Comparison of ECG bioelectrodes

Bioelectrodes	Skin Condition	Advantages	Disadvantages
Wet	Gelled	• Records high-quality biosignal • Suitable for short-term ECG recording • Acceptable in clinical practice	• Gel causes skin allergies • Requires skin abrasion • Dehydration of gel affects quality of biosignal • Usage is restricted to clinical settings
Direct contact	Dry	• No gel or skin abrasion is required • No skin allergies • Suitable for long-term ECG recording and wearable biosensors	• Biosignal quality depends on skin condition • Requires special front-end bio-amplifiers with a high input impedance • Prone to powerline noise and motion artifacts
Micro-needles	Dry	• Requires no skin preparation • Ensures proper skin contact and low impedance • Good ECG quality	• Inconvenient • Could cause skin infection in long-term use • Fabrication is expensive
Non-contact	Dry	• No gel or skin abrasion is required • No skin allergies • Suitable for long-term ECG recording and wearable biosensors • Dielectrics provide better skin comfort	• Biosignal quality depends on the type of dielectric material • Requires front-end circuits with a high input impedance • Susceptible to powerline noise, motion artifacts, and triboelectric charge effect

In non-contact ECG sensing, the performance of the bioelectrodes is greatly influenced by the dielectric material, which makes the dielectric material worth investigating. From non-contact ECG bioelectrodes developed previously, three dielectrics were most common. They are metal-oxides, fabrics, and polymers. Metal oxides were employed as dielectrics in non-contact ECG bioelectrodes because of their high dielectric constant [82–84]. However, their rigid structure induces severe motion artifacts and electrical interference [30].

Alternatively, previous research investigated natural and synthetic fabrics as dielectrics [54, 86–89]. Fabric dielectrics are soft, breathable, and skin conformable. Although, the aforementioned properties are desired in dielectrics used for wearable medical bioelectrodes. However, in long-term ECG monitoring, fabrics can exhibit unstable behaviour when they are moisturized by sweat and skin exudates. Also, they can introduce triboelectric charges strong enough to distort weak biopotential signals such as the ECG.

Among the enormous materials on the earth, polymers are well-known. Polymers are primarily used in electronics as insulators because they are non-conductive materials [90, 91]. Recent years have seen a high increase in the characterization and application of polymers as biomaterials and biomedical electronics for several reasons [92–97]. Polymers are exciting materials. They are easy to synthesize and characterize to produce degradable medical devices [98]. Polymers are easy to fabricate into any desired shape [99]. They are inert against most chemicals, light weight, stretchable, skin conformable, and harmless to human tissue [100, 101].

Polydimethylsiloxane (PDMS) belongs to the siloxane group of polymers that has gained broad research interest in biomedical engineering because of its outstanding properties and compatibility [102–106]. More interestingly, PDMS has the following qualities and potentials; (a) it is relatively cheap, (b) resembles and harmless to human tissue when characterized and moulded [107], (c) flexible and stretchable, (d) can be fabricated using a simple method to conform with skin structure, (e) bio-durable and compatible with other polymers and nano-particles. Among the poly-siloxanes group, PDMS has been extensively employed as a significant composite constituent in developing polymer biosensors [65, 108], micro-fluidic devices [109–112], and substrate materials [113–115].

In non-contact ECG measurement, the following qualities are essential in a dielectric material; high dielectric constant, biocompatible, bio-durable, and ability to conform with the skin. Another essential characteristic of a biomedical dielectric is the capacity to protect any front-end electronics from short-circuit by sweat and enable high coupling capacitance. Also, dielectric materials that induce triboelectric (electrostatic) charges are unsuitable since they can generate charges that can distort the ECG signal.

1.2 Problem Statement

Like most polymers, several investigators have explored PDMS to serve as insulants in electronics [116, 117]. 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 nano-particles as composites for non-biomedical applications [111, 118–121]. However, the techniques employed require expensive nano-particles, 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 [122], leading to a waste of time and resources. When undertaking the current study, we are yet to find literature on composite PDMS dielectrics for capacitive bioelectrodes.

PDMS is commercially available as a single kit but in two parts; elastomer base and curing agent (crosslinker). 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 conducting nano-particles. In this study, the term "unblended" means without electrical conducting nano-particles. By adding more of the crosslinker, the polymer chains get shorter and structurally affected [123–125]. This study intends to explore the dielectric performance of unblended PDMS films for non-contact ECG sensing applications. By employing the manual deposition technique, PDMS dielectric films can be characterized by varying the proportion of the silicone elastomer (Sylgard 184^{TM}) base and its curing agent. To our knowledge, no direct study investigating the dielectric performance of unblended PDMS for non-contact ECG bioelectrodes has been reported in the literature. Exploring unblended PDMS elastomer for dielectrics will open up and better understand its performance when applied in non-contact ECG bioelectrodes.

1.3 Research Objectives

The objectives of the present investigation are:

(a) To experiment and fabricate by varying the polymer mix ratios, unblended PDMS dielectric films.

(b) To characterize the electrical and dielectric properties of unblended PDMS films.

(c) To model and simulate the bioelectrode skin interface equivalent circuit for a porous and non-porous dielectric for non-contact ECG bioelectrode.

(d) To assess the dielectric performance of unblended PDMS dielectrics in non-contact ECG recordings.

1.4 Scope of the Research

The primary focus of this research is to modify and fabricate by manual deposition the dielectric properties of PDMS films. Thin PDMS films were characterized by varying the proportion of the silicone elastomer liquid and its crosslinker. The polymer mix ratios affect PDMS baking temperature. Consequently, a trade-off was made between the PDMS mix ratios baking and its baking temperature that can be tolerated by the glass substrates. Considering these factors, the weight ratio method is applied on the elastomer and its crosslinker to generate three polymer mix ratios; 10:1, 10:1.5, and 10:2. Also, the minimum thickness of the proposed PDMS films is defined by the thickness of the spacer. The effects of polymer composition ratio on PDMS dielectric properties are the main interest within the research scope. The dielectric parameters are the capacitance and relative permittivity of the PDMS films. PDMS is a low-k dielectric material. The parallel-plate technique is considered for measuring the generated capacitances with an Agilent LCR meter of a preset frequency range of 100 Hz to 100 kHz. Also, the instrument applied for measuring impedance has fixed and restricted test frequencies. The possible range of frequency allowed is 4 Hz to 1 MHz. The equivalent circuit model for a non-porous PDMS dielectric was simulated in Matlab 2019b to determine circuit parameters.

Furthermore, the relative permittivity of each film sample was estimated by the general expression for parallel-plate capacitance. PDMS films with reliable dielectric characteristics were selected and experimented. Separately, ECG waveforms were acquired with the Ambu Ag/AgCl, direct contact, and non-contact bioelectrodes with fabric (porous) and PDMS (non-porous) dielectrics.

1.5 Significance of the Research

The study will provide benefits in the following ways:

(a) The model and simulation results of the equivalent circuit for a porous and non-porous dielectric will provide invaluable information for designing biopotential amplifiers and other front-end biosignal processing circuits.

(b) This is the first occasion the dielectric performance of unblended PDMS films from Sylgard 184^{TM} elastomer is investigated for non-contact ECG sensing. The results obtained will provide relevant insights on the characterization of unblended PDMS as a safe biomedical dielectric without expensive nanoparticles.

(c) Non-porous PDMS dielectric film is flexible, non-toxic, harmless to the skin, and can satisfactorily protect the front-end circuit from skin exudates. The recorded ECG signals with visible P-QSR-T peaks confirm the feasibility of reliable non-contact measurement physiological signals for early detection, diagnosis of cardiac abnormalities, and preventing sudden death of heart patients.

1.6 Thesis Outline

The organization of the different chapters of the thesis is as follows:

In Chapter 1, essential information on cardiovascular diseases and the health challenges are introduced. This is followed by a description of the techniques used to record ECG signals and performance limitations of bioelectrodes in long-term ECG measurement. Next, the problem statement and research objectives are outlined. The remaining sections of the chapter define the scope and significance of the research.

In Chapter 2, the literature review covers relevant aspects of the human cardiac system, skin physiology, and the ECG extraction points. Further, an extensive review is conducted on non-invasive ECG bioelectrode technologies, the skin bioelectrode interface, and equivalent electrical models. The focus of the study is non-contact bioelectrodes and biomedical dielectrics. Therefore, the final part of the chapter provides a detailed assessment of three categories of dielectric materials and their crucial role in non-contact ECG recording.

Chapter 3 elaborates the experimental procedures employed to modify and boost the dielectric performance of pure Sylgard 184^{TM} PDMS films. The chapter starts with a schematic narration of the experimental design, fabrication of PDMS dielectrics, and morphological analyses. Next, the electrical and dielectric parameters of selected samples PDMS films were measured using the parallel plate technique. Equivalent circuit models for both porous and non-porous dielectrics are presented. Matlab simulation was implemented to observe the output response of the proposed capacitive biosensing system using the transfer function. The performance of the proposed PDMS films is tested through a non-contact method to record ECG from a human subject, processed using the digital filters, and compared with other types of dielectric materials.

In Chapter 4, the results presented comprise the surface morphological analyses of unblended PDMS films made from different mix ratios, dielectric properties, equivalent circuit model simulations, ECG signal recordings, and denoising ECG signal techniques. In addition, a detailed analysis is made and the implication of the results in capacitive bioelectrodes and extended ECG monitoring.

Lastly, Chapter 5 completes the thesis with a conclusion and offers suggestions for future research.

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LIST OF PUBLICATIONS

Journal with Impact Factor

1. A. H. Umar, M. A. Othman, F. K. C. Harun and Y. Yusof, ”Dielectrics for Non-Contact ECG Bioelectrodes: A Review,” in IEEE Sensors Journal, vol. 21, no. 17, pp. 18353-18367, 1 Sept.1, 2021, doi: 10.1109/JSEN.2021.3092233. **Q2, Impact Factor 3.301**

Indexed conference proceedings

1. A. H. Umar, F. K. C. Harun and Y. Yusof, ”Characterization of Polydimethylsiloxane Dielectric Films for Capacitive ECG Bioelectrodes,” 2020 7th International Conference on Electrical Engineering, Computer Sciences and Informatics (EECSI), 2020, pp. 127-131, doi: 10.23919/EECSI50503.2020.9251882. **Awarded Best Conference Paper**