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Flexible Electrodes for Smart Bandages: Feasibility Exploration

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Abstract

Flexible electrodes are revolutionizing the field of wearable health-monitoring and therapeutic devices by enabling the production of large, lightweight, and thin gadgets. These electrodes are incredibly beneficial for collecting bioelectric signals from the human body. They offer stable, high-quality signals while ensuring breathability and skin-friendly contact. Products for which flexible electrodes are actively being developed include innovative wearable devices, portable medical equipment, and brain-computer interfaces.

Wearable medical technology has made leaps and bounds in recent years with notable advancements in consumer health-monitoring gadgets such as smartwatches and clinical-grade sensors. Various types of sensors have been showcased, capable of measuring body temperature, heart rate, blood-oxygen levels, and metabolites in sweat. For a device to qualify as ‘wearable,’ it should be comfortable, lightweight, thin, shape-conforming, and wireless.

Wearable medical devices necessitate the integration of electrodes, power sources, and microcontroller chips. These components should all adhere to the same comfort standards as the electrodes themselves. Flexible electrodes offer several advantages, namely, flexibility, comfort, biocompatibility, and superior signal quality. Conventional wet electrodes capture and measure the human body’s electrical signals while preserving high signal quality. Contrastingly, traditional dry electrodes provide superior comfort and facilitate long-term monitoring. Flexible electrodes made using conductive ink and a polyurethane film with an adhesive layer are capable of long-term monitoring while maintaining high signal quality.

The primary objective of this study is to refine the design of flexible electrodes used in wearable health-monitoring and therapeutic devices. By fabricating micro-perforated structures with various aperture sizes and spacings and applying silver ink on Z-conductive electrodes, the aim is to identify the optimal combination of aperture size and spacing. To this end, a measuring and fitting process is employed. We discovered that Z-conductive electrodes with a hole spacing of 0.28 mm exhibited the lowest impedance values in the low-frequency range of 5 kHz-50 kHz. Comparatively, holes with a spacing of 0.4 mm had the lowest impedance in the high-frequency range of 100-500 kHz. These findings may facilitate future mass production efforts for our industrial partner, Ti2 Pty Ltd. This research contributes to innovation in wearable medical technologies by enhancing the performance of flexible electrodes, thereby improving the quality of biometric signal collection and the comfortability of wearable devices.

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Chapter 1

Introduction

This study aims to identify the optimal design configuration of Z-conductive electrodes that yield the lowest impedance. Flexible electrodes have become a dynamic research area in recent years due to their potential applications in smart bandages for wound monitoring and treatment.

Currently, wound dressings primarily seal and protect the injury site. Some wound dressings release compounds or drugs that prevent infection and accelerate healing [7]. However, a key limitation of these dressings is their inability to provide information regarding the wound's healing status and the wound environment conditions such as pH levels, bacterial load, tissue oxygenation, and inflammatory markers. Information from sensors within the wound environment could provide crucial insights to hasten the decision-making process in wound care [27].

However, existing flexible electrodes present design complications, necessitate prolonged preparation for use and require complex manufacturing processes for mass production. Contrastingly, Z-conductive electrodes offer the advantages of lower manufacturing costs, as well as the potential for mass production and rapid deployment. This makes Z-conductive electrodes a promising candidate as electrode carriers for affordable smart bandages in the future.

1.1 Research Motivation

Providing medical care for wound management has burgeoned into a daunting problem within the global healthcare landscape. New burn wounds that are sufficiently severe to

warrant professional medical intervention occur daily at a rate of over 30,000 globally. This statistic extrapolates to an alarming annual figure of approximately 11 million new burn incidents internationally [47]. It has been estimated that between 1% and 2% of people in developed nations will be confronted with a chronic wound at some point in their lifetime [6]. In the United States, chronic wounds reportedly affect a patient population of \$6.5 million and engender an annual fiscal burden on the healthcare system of over \$25 billion to manage complications related to wound pathology [35]. The sheer magnitude of this healthcare challenge underscores the necessity for effective and innovative solutions in wound management and care.

Currently, the primary function of wound dressings is to protect and encapsulate injury sites. Some variants dispense substances or drugs capable of inhibiting infections and promoting wound healing. However, a significant shortcoming of these products is their inability to relay data concerning the wound's healing status and environmental conditions with parameters such as pH, bacterial presence, tissue oxygen levels, and the degree of inflammation [15]. Data obtained from sensors embedded in the wound dressing could offer pivotal insights to hasten the decision-making process in wound management.

Smart bandages could offer an effective solution to challenges in wound management, and flexible electrodes are a key component in designing such innovative wound dressings [15]. However, the prevailing landscape of flexible electrodes confronts a series of issues, namely:

- 1) Flexible electrodes designed for smart bandages are primarily fabricated by adhering silicon onto a polydimethylsiloxane (PDMS) base. The electrode is then manufactured on top of the PDMS base using thin-film silicon. This process necessitates the placement of 100 nm thick single-crystal silicon on a 20 μm thick plastic substrate [29].
- 2) High Cost: Smart bandages used for wound care and healing are expected to be used in large quantities. However, electrodes that come into contact with wounds are non-reusable [41]. This necessitates sufficiently low manufacturing costs.
- 3) Preparation Time: Most smart bandages employ hydrogel-type wet electrodes as sensors that require preparation before use. Such electrodes are unsuitable for use over extended periods because hydrogels tend to dry out over time in indoor conditions [41, 26].

1.2 Problem statements

The prior experimental report from Ti2 on Z-conductive electrodes provided evidence of their electrical conductivity under varying curing temperatures of the conductive ink. Additionally, the study verified that, upon coating and curing, the Z-conductive electrodes allow anisotropic conductivity throughout the bandage but not transversally through the bandage's cross-section. This renders the dressing appropriate for bioimpedance device applications [13]. Nevertheless, there remain three elements of the hole design that require additional investigation, namely:

- 1) Holes with differing diameters do not consistently retain the same proportion of z-oriented silver micro towers, which could lead to inconsistencies in impedance measurements during application.
- 2) The anisotropic conductivity of deposited silver in holes with disparate inter-hole distances could also result in inconsistent impedance measurements during application.
- 3) The potential effects of various hole arrangement methods on measurement outcomes remain undetermined.

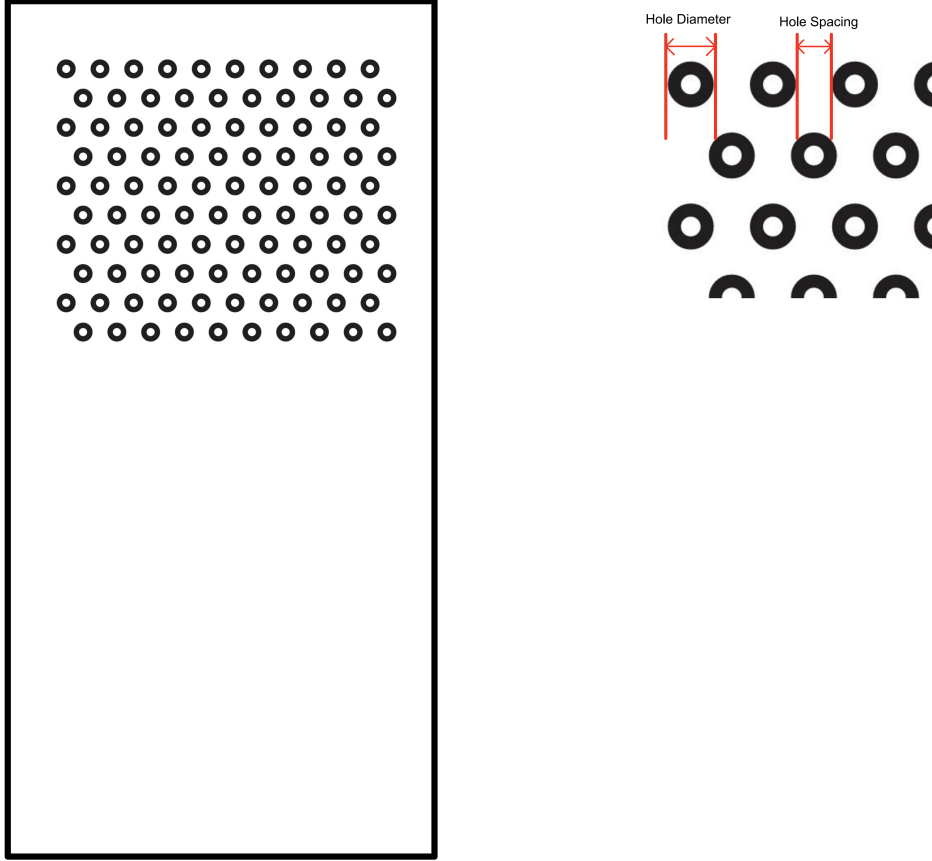


Figure 1.1: Hole Diameter and Hole Spacing Schematic

1.3 Research questions

This study aims to ascertain the optimal hole diameter and inter-hole spacing configuration by way of conducting empirical assessments.

1. With a specific focus on hole diameter and inter-hole distance configurations, what strategies can be employed to optimise the construction of Z-conductive electrodes?
2. What approaches can be utilised to enhance the conductivity of Z-conductive electrodes, thereby reducing their functional impedance?

1.4 Thesis organisation

This thesis consists of six chapters, including an introduction in Chapter 1 and a literature review in Chapter 2. Chapter 3 introduces the methodology, and Chapter 4 and Chapter 5 presents the results and a discussion. Chapter 6 concludes the thesis and discusses possible future research directions. In Chapter 2, the definition and types of chronic wounds are first introduced, and the role that smart bandages play in wound monitoring and healing is also highlighted. The current applications of flexible electrodes in smart bandages are also emphasised. Thereafter, the potential of Ti2's Z-conductive electrodes for mass production is analysed. A review of previous experimental results from Ti2 is also provided to clarify the purpose of this research. Chapter 3 describes 15 different designs with varying combinations of hole diameters and spacings for Z-conductive electrodes. The coating and measurement methods and experimental design are also introduced. Chapter 4 and 5 presents the experimental data which mainly consists of bioimpedance measured on the left arm under real conditions. The experimental data are subsequently analysed, and the two questions posed above are answered. Chapter 6 summarises the entire research and provides suggestions and questions for possible future research.

Chapter 2

Literature Review

2.1 Wounds and wound types

To the standard anatomical structure and function of a tissue, a wound is characterised as an interruption or impairment. Such a disruption can be as minor as a superficial tear in the skin's epithelial layer or as severe as deep trauma that damages the subcutaneous tissue, muscles, tendons, blood vessels, nerves, internal organs, or even bones [23, 50]. Wounds can be the outcome of external factors like accidents or trauma, or they can result from internal pathological conditions (e.g., diseases). Regardless of origin, all wounds cause tissue damage and upset the local environment [22].

The wound healing process is a dynamic and intricate progression of interrelated events [21]. It commences with haemorrhaging followed by the activation of the coagulation cascade, then the onset of an acute inflammatory response and the regeneration, migration, and proliferation of connective tissue and paraneural cells [31]. Additionally, the synthesis of proteins in the extracellular matrix, tissue remodelling, and collagen deposition collectively contribute to restoring the wound's strength [50]. The mending process is dependent on the participation and involvement of resident and migratory cell populations, extracellular matrix components, and soluble mediators [2, 22]. Equally important, at the time of injury, healing mechanisms, which comprise inflammatory mediators, growth factors, and cell-cell/cell-matrix interactions that regulate cell proliferation, migration, and differentiation, are activated and persist throughout the repair process [21].

Although wound healing mechanisms are generally similar across different tissues, the specific pathways and timelines may vary depending on the type of tissue and

the severity of the injury [23, 38]. In some cases, it may take over a year for a wound to heal completely [42]. Healing occurs when the affected area has regained its normal anatomical structure, function, and appearance within a reasonable period [33]. However, both systemic and local factors can impede the coordinated wound healing process, causing chronic, non-healing wounds [22, 2].

From a medical standpoint, the categorisation of wounds as either acute or chronic is determined based on their anticipated healing duration [48, 14]. Acute wounds tend to heal within the expected time frame, whereas chronic ones exhibit delayed or incomplete healing.

2.1.1 Acute Wounds

Acute wounds, which may result from traumatic tissue injury or surgical procedures, are classified as those that heal in a timely, predictable manner, with the restoration of normal tissue structure and function [48]. Patients typically recover within a period of 5-10 days, up to a maximum of 30 days. More specifically, surgical excision of a soft tissue tumour that leaves a large defect may also leave an acute wound that cannot be closed straight away but will still heal in an orderly manner. Similarly, traumatic wounds involving soft tissue or bone fractures are usually acute. These combined injuries are classified by systems like the AO Foundation system, accounting for the extent of damage to muscle and tendons that impacts healing. A valuable benefit is a prognostic information based on wound severity [50].

2.1.2 Chronic wounds

Wounds that remain unhealed for more than a month are conventionally known as chronic wounds, a common issue worldwide. They are often referred to as complex wounds, a term that can be misleading, since even acute and traumatic wounds can be complex, to some extent. Chronic wounds are identified by superficial, partial, or full-thickness skin loss that heals through secondary intention [25].

Wound healing normally undergoes a series of continuous, interlocked phases: immediate haemostasis, followed by inflammation, proliferation, and remodelling. However, this continuous, overlapping stage paradigm does not accurately reflect the non-linear organisation of the tissue repair process in chronic wounds [32].

Depending on the underlying cause, chronic wounds are divided into four categories: pressure ulcers, diabetic ulcers, venous ulcers and arterial insufficiency ulcers [35].

Research suggests that between 1% - 2% of the population in developed countries will likely experience a chronic wound during their lifetime [43]. The elderly usually sustain such wounds, and as their population continues to increase, these figures will continue to climb.

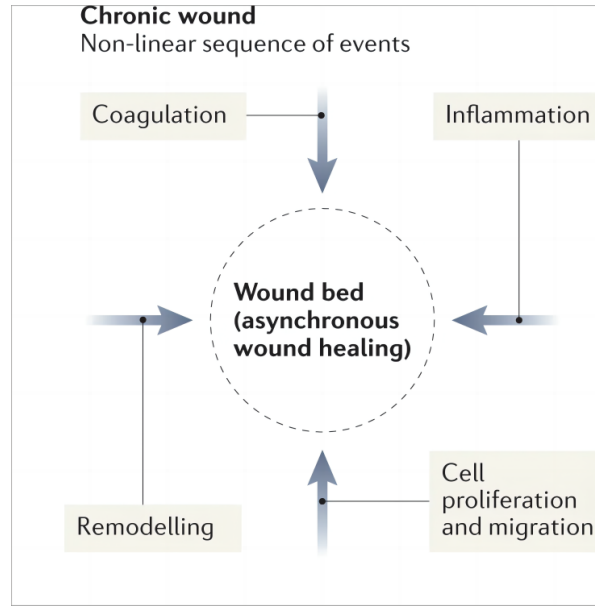


Figure 2.1: Chronic wound recovery process [17]

Chronic wounds present significant heterogeneity in terms of aetiology, pathogenesis, size, body location, incidence, risk of limb amputation, and host factors, among several other variables. In less developed countries, the root causes may differ significantly from those in advanced nations, including nutritional deficiencies, parasitic and chronic fungal infections, and leprosy. Irrespective of the varying causes of chronic wounds, they all present with impaired healing, with some authors and clinicians referring to them as ‘non-healing’ [17].

2.1.3 Complex wounds

Infection and tissue damage coexist in complex wounds, setting them apart from other categories [1]. The healing process is constantly obstructed by the presence of infection, while tissue damage can result from trauma, surgical procedures, or underlying health issues. All wounds are exposed to some level of contamination, yet the emergence of a visible infection is dependent on the virulence and quantity of microbes, as well as the

host's resistance and wound perfusion. Possible indications of an infection are redness, heat, pain or discomfort, swelling, and functional impairment. The risk of contracting an infection can vary according to the nature and site of the wound, and any surgical interventions [4].

Taking into account origin, contamination level, shape, and any connection to internal organs, various wound classification systems have been developed. These origins could include contusions, scrapes, cuts, puncture wounds, crush injuries, gunshot wounds, or burns. The degree of contamination can classify wounds as aseptic (e.g., orthopaedic), contaminated (e.g., gastrointestinal), or septic (e.g., abscesses). Additionally, wounds may be categorised as closed (with unbroken skin) or open (with compromised skin) [4]. These factors aid in determining the most suitable wound care and in anticipating healing outcomes.

2.1.4 Wound Healing Process

The wound healing process is continuous and involves coordinated interactions between immunological and biological systems. The physiological events in the wound and surrounding tissue can be understood by breaking this process down into distinct phases. These phases consist of the coagulation and homeostasis phase, the inflammatory phase, the proliferate phase, and the remodelling phase [21].

Coagulation and haemostasis Phase The injury initiates the coagulation and haemostasis phase to prevent blood loss. This phase safeguards the integrity of the vasculature for organ perfusion and provides a matrix for cell migration.

Inflammatory Phase It establishes an immune defence against microbes. Activation of the complement system in the early stages of inflammation induces the recruitment of neutrophils to prevent infection.

Proliferative Phase Once haemostasis and inflammation are underway, acute wounds switch into the repair phase. In a matter of days, the proliferative phase is initiated with the migration of fibroblasts and the deposition of the extracellular matrix.

Remodelling Phase The process entails the creation of new epithelium and the final scar tissue. Synthesis of the extracellular matrix begins with the formation of granulation tissue and may extend for years.

Healing by Primary and Secondary Intention Normal wound healing relies on the coordinated functioning of multiple processes, cells, mediators, and structures. The cell patterns vary by the type and extent of the injury. Closed or sutured incisions with minimal tissue damage heal relatively fast through primary intention. Moreover, healing extensive defects by secondary intention takes longer due to the formation of a large amount of granulation tissue. Impaired healing is a consequence of pathophysiological/metabolic factors that alter the local environment. This delays epithelialization, inflammation, fibroplasia, and matrix deposition, converting acute wounds into chronic ones.

2.2 Smart Bandage

Current wound dressings primarily seal and protect the injury site, some of which are designed to release compounds or medications that can prevent infection and expedite healing. However, one major drawback is their inability to provide information about the healing progress and conditions of the wound site, including the pH, bacterial load, tissue oxygenation, and inflammation levels. Utilising data collected from sensors within the wound environment, crucial insights can be obtained, thereby accelerating the decision-making process in wound care.

For instance, Mostafalu et al. [41] designed a smart bandage equipped with sensors (for pH and temperature), a miniature heater, thermosensitive drugs embedded in a hydrogel patch, and wireless electronic devices for data reading. This platform's design makes the sensor module and integrated heater economical and disposable, with the entire assembly, affixed to transparent medical tape and the electronic devices being reusable. 2.2 illustrates how an externally connected reusable electronics module powered and provided the entire smart bandage system with Bluetooth connectivity. The smart bandage, by regulating the pH and temperature sensors, can be used to heat the drug patch to release drugs at specific intervals throughout the wound healing process, thereby promoting wound healing.

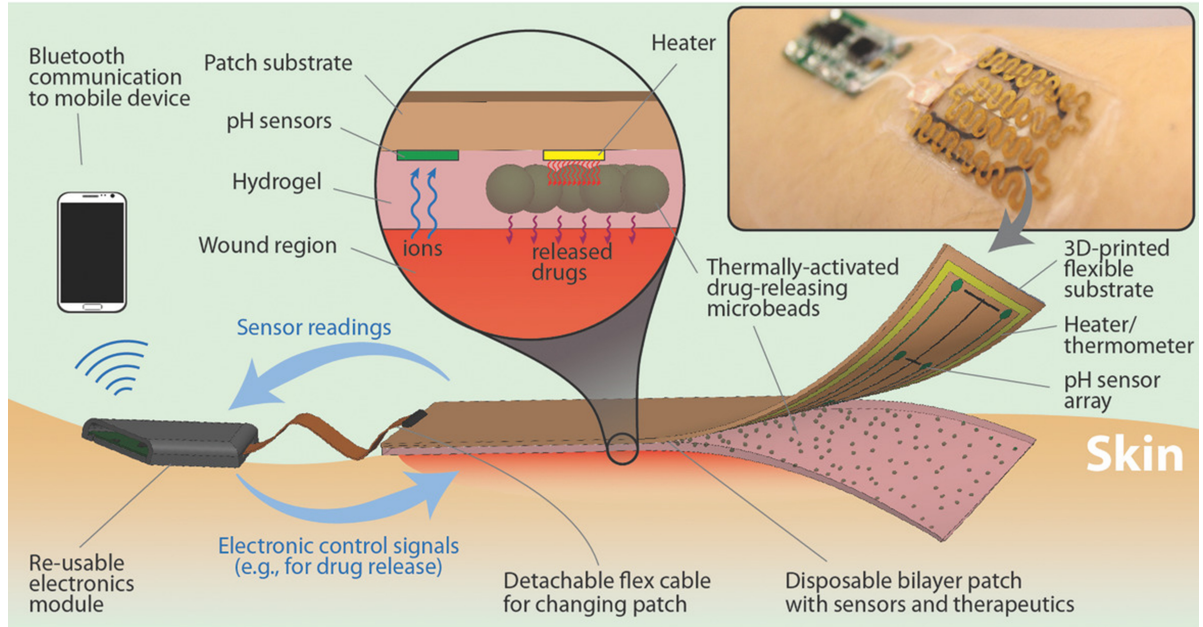


Figure 2.2: Schematic and conceptual view of the automated smart bandage [41].

Jiang et al. [26] developed a battery-free flexible smart bandage system with wirelessly powered sensors and stimulation circuits. It features dual-channel continuous surveillance of wound impedance and temperature, as well as a parallel stimulation circuit to deliver programmed electrical signals for expediting wound healing. Figure 2.3 depicts that this smart bandage features a wirelessly powered flexible printed circuit board (FPCB) and a tissue interface conductive adhesive hydrogel interface. An RF energy collection antenna can supply energy to apply an electric bias on the wound for programmed treatment, making it possible for the near-field communication (NFC) transmitter in the microcontroller unit (MCU) to continuously observe wound impedance and temperature. Overall, this wireless smart bandage can actively monitor the wound, providing a closed-loop treatment, which promotes wound healing through electrical stimulation.

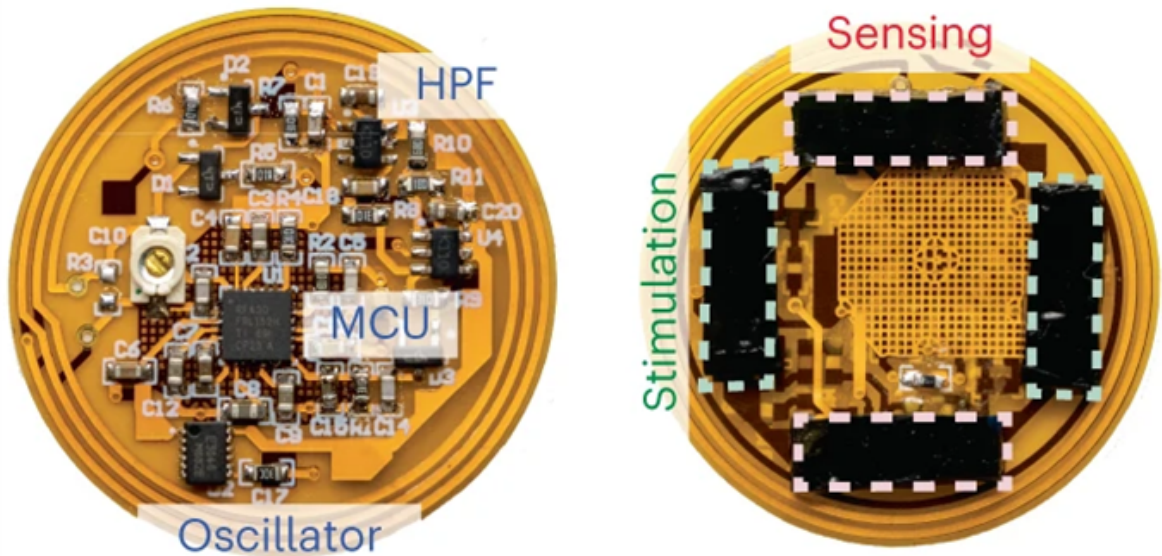


Figure 2.3: Photographs of the front (left) and back (right) sides of the smart bandage showing the MCU, crystal oscillator, HPF and stimulation and sensing electrodes [26]

Escobedo et al. [16] smart bandage for wound information monitoring is also worth noting. It can survey the condition of the wound through pressure and temperature sensors and transmit relevant data to a smartphone application via battery-free NFC. As shown in Figure 2.4 depicts how the entire smart bandage system captures wound data through temperature and pressure sensors, whilst the NFC chip transmits power and data. This smart bandage pays special attention to functions related to monitoring that only require minimal power consumption.

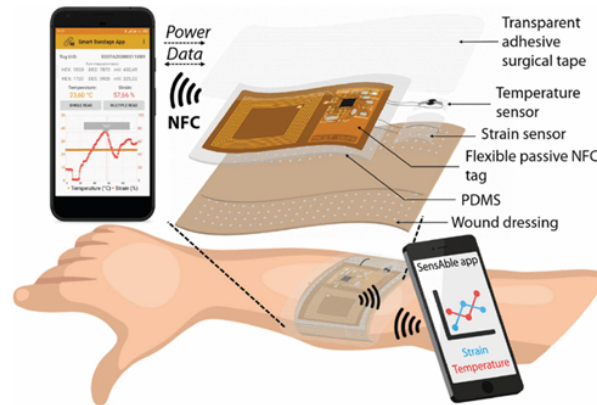


Figure 2.4: A synopsis of the smart bandage equipped with NFC for wireless, real-time monitoring of strain and temperature. The bandage utilizes a specially developed smartphone application for power supply and two-way communication [16]

These innovations in smart bandages represent a major breakthrough in wound care, offering real-time, in-situ monitoring of wound conditions and facilitating more effective and personalised treatment strategies.

2.3 Flexible Electrodes

2.3.1 Wet Electrodes

Bioelectric sensors of the wet electrode type utilise a conductive gel to form an electrolytic contact with the skin, thereby minimising impedance and enhancing the quality of recordings [51]. Through the close contact between the electrode and the skin's high amplitude, information-rich signals across a wide frequency range can be captured. Consequently, wet electrodes have become the preferred option in numerous clinical and research settings (e.g., sleep studies, neuropsychiatry, and cardiology) [46].

However, despite their widespread use, wet electrodes have several limitations. Sensitive individuals are often prone to skin irritation and discomfort, and can sometimes have allergic reactions due to the use of conductive gel and adhesives [30]. Along with that, the gel can dry out over time, thereby increasing the impedance between the electrode and skin while potentially compromising the signal's quality [12]. This issue also influences the long-term wearability of wet electrodes, rendering them less suitable for extended studies [49].

The need to address maintenance-related challenges has inspired the creation of dry electrode technologies, even though wet electrodes provide superior baseline signal quality. These emerging technologies, including advances in microneedles, nanomaterials, and microfabrication, strive to deliver performance on par with wet electrodes while eliminating the need for a conductive gel.

2.3.2 Dry Electrodes

In comparison to wet electrodes, dry electrodes have direct contact with the skin, bypassing the need for conductive gels, which eliminates the often lengthy preparation process typical of wet electrodes [20]. Recent technological advancements, particularly in the areas of materials, coatings, and microfabrication, have enabled dry electrodes to match, and in some cases, exceed the signal quality of conventional wet electrodes in various bioelectric monitoring applications [30].

Dry electrodes have numerous benefits, making them especially suitable for long-term, continuous physiological tracking [20]. Wearable devices are particularly advantageous due to their user-friendly design and the absence of skin irritation from conductive gels, thereby providing comfort during extended wear. With no prior skin preparation required, the application and removal are effortless, a factor that significantly increases user compliance [49]. Utilising dry electrodes, wearable devices are an optimal choice for gathering bioelectric data in real-world conditions, widening the scope of clinical or laboratory settings. Additionally, eliminating gels mitigates the risk of skin allergies and reactions, a common issue with prolonged wet electrode use among individuals with sensitive skin [30].

Despite these advantages, challenges and limitations associated with dry electrode technologies remain to be addressed. A crucial problem is that dry electrodes necessitate a certain degree of sustained contact pressure on the skin to acquire quality signals, potentially causing discomfort over extended periods of use [30]. Furthermore, motion artifacts are more pronounced with dry electrodes compared with wet electrodes, causing a severe degradation in signal fidelity. Ongoing attempts to mitigate these limitations are underway, with advances in conductive coatings, novel nanomaterials, and microstructured interface designs showing promise [44]. However, the issue of motion artifact susceptibility remains a major hurdle for dry electrodes to overcome to equal the performance benchmarks of traditional wet electrodes for high-fidelity bioelectric signal recording [12].

2.3.3 Flexible Electrodes

As mentioned, flexible electrodes are an integral part of the sensors and various electronic components, which are inseparable from the advancements in smart bandages. Flexible electrodes require a high degree of mechanical flexibility and stretchability, and ensuring such features on electronic devices and sensors require soft and hard materials. These are regularly integrated into novel composite combinations or geometric patterns, enabling the system to withstand heavy strains without faltering [29]. Metals and composite materials can be transferred and printed onto elastomer substrates as a type of flexible electrode [36].

Flexible electrodes can also be developed through economical and straightforward manufacturing processes while being environmentally friendly. The conductive material chosen for the base of flexible electrodes is usually conductive silver ink. Its potential in flexible electronic devices for bio-sensing applications has been substantiated. Fernandes et al. tested a variety of conductive silver ink coatings on polyethylene (PET) substrates. They used inkjet printing to coat the conductive ink containing silver nanoparticles onto the surface of this substrate [18]. Their findings highlighted that different curing temperatures influence the impedance of the produced flexible electrodes, which is consistent with the previous experiments conducted by the Ti2 company.

The potential for printed electrodes in wearable health monitoring applications is significant. Several conductive materials have been printed onto textile substrates, and their performance has been assessed. However, improving sensor performance is reliant on various parameters of printing, textile structure, and the conductive material used. Through innovative device design and the use of novel materials, sensitivity and linearity can be customised. Compared with biochemical sensing, printed electrodes have been extensively used for bioelectric potential monitoring. Although the functionality of printed wearable health monitoring has substantially improved, data reliability needs further enhancement. While the electronic-based manufacturing of electrodes gives better data reliability, printed electrodes possess the potential for mass production and cost reduction, making the device accessible to the general public [24].

Table 2.1: Comparison of Wet, Dry, and Flexible Electrodes

Electrode Type	Advantages	Disadvantages
Wet Electrodes	• High-quality recordings • Preferred in clinical settings • Captures information-rich signals	• Potential skin irritation and allergies • Gel can dry out over time • Unsuitable for long-term wear
Dry Electrodes	• Eliminates preparation process • Suitable for long-term monitoring • User-friendly design • Less skin irritation	• Requires sustained contact pressure • More pronounced motion artifacts
Flexible Electrodes	• High degree of mechanical flexibility • Cost-effective manufacturing • Environmentally friendly • Potential in flexible electronic devices	

2.4 Anisotropic Conductive Dressings by Ti2 Company

In previous experiments by Ti2, an anisotropic conductive dressing was successfully developed through laser perforated Aero Film dressings impregnated with silver. The dressing, labelled Aero Film 1827, was constructed with layers of polyethylene, polyurethane, and acrylic, and was laser perforated with holes of varying diameters and spacings. Using a commercial screen-printing unit configured as a doctor blade coater, this study used a silver ink called Creative Materials silver flexo ink Type 124-02. Afterwards, the ink was then dried, and the percentage of successful hole filling (i.e., silver impregnation) was determined. Resistance measurements and imaging of the various bandage layers were used to ascertain the anisotropic conductivity of the dressing [13].

The study discovered that 120 μm was the most ideal hole diameter due to its high percentage of hole filling (as much as 95%), satisfactory silver retention during handling, and high anisotropic conductivity. Research revealed that 50 °C was the optimal curing temperature for the silver ink, as it maintained the dimensional stability of the Aero Film dressings. At higher temperatures, shrinkage of the dressing's support layer was observed [13].

The dressing developed was conductive through the bandage but not sideways, making it suitable for use as a bio-impedance device. Nevertheless, the study recommended that further optimisation of hole manufacturing parameters and detailed imaging of the holes would be beneficial, specifically by improving silver retention and product uniformity. It was also proposed to further examine the factors leading to the undesirable detachment of the support layer prior to the release layer.

2.5 Bioimpedance Spectroscopy and Cole–Cole model

Two components make up the complex impedance (Z) of biological tissues: the real component, resistance (Z'), and the imaginary component, reactance (Z''). The impedance measurements for each frequency in Bioimpedance Spectroscopy (BIS) are represented on a (Z' , Z'') plane, forming a semicircle [45]. The mathematical model for biological tissues, put forth by Cole and Cole in 1941 [10] proposed a mathematical model for biological tissues, which includes three theoretical elements of the circuit: the resistance at low frequency (R_0), the resistance at high frequency (R_∞), and the

Constant Phase Element (CPE). The CPE functions similarly to a fractional capacitor, with its impedance defined as $Z_{CPE} = 1/(j\omega C)^\alpha$, where C is the capacitance and α is an exponent within the range of $0 < \alpha \leq 1$. [34]. The following equation describes the Cole-Cole impedance:

$$Z = R_\infty + \frac{R_0 - R_\infty}{1 + (j\omega\tau)^\alpha} = Z' + jZ''$$

OR

$$(j\omega)^\alpha = \omega^\alpha [\cos(\alpha\pi/2) + j \sin(\alpha\pi/2)]$$

Analysis of body composition analysis with BIS is premised on the variability of the body's electrical properties, which are dependent on the ratio of bodily fluids and tissues. Materials like blood or muscle exhibit higher electrical conductivity than skeletal or adipose tissue. The cellular membrane separates the body's water content into intracellular water (ICW) and extracellular water (ECW). Both ECW and ICW primarily serve as resistive entities, while the cell membrane, owing to its lipid layer, has an insulative (capacitive) behaviour. Taking this into account, the path of the injected current shifts between low and high frequencies: at low frequencies, the current exclusively circulates through the extracellular fluid (ECW, denoted by R_e), whereas at high frequencies, it also penetrates the cell membrane and intracellular fluid (ICW, denoted by C_m, R_i). Figure 2.5 illustrates this phenomenon as represented by the electrical model [3].

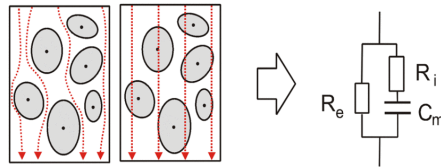


Figure 2.5: Low and high-frequency current flow through body tissue

2.6 Discussion

In the burgeoning era of wearable devices, the demand for seamless health monitoring has been escalating. Traditional sensor technology, however, is becoming increasingly inadequate to fulfil the requirement of this pervasive, large-scale monitoring. Smart bandages epitomise this evolution—being able to not only monitor wounds but also administer treatment during the healing phase, heralding a transformative shift in the wound management sector.

Although the market has yet to encounter a fully matured smart bandage, research in this domain is thriving. Mostafalu et al. [41] have pioneered smart bandages capable of drug release and wound monitoring. Similarly, Jiang et al. [26] have put forward designs that incorporate electrical stimulation to expedite wound healing, complemented by monitoring features. On the other hand, Escobedo et al. [16] have a more singular focus on wound monitoring. Collectively, these innovations underscore the substantial advancements made in wound monitoring, highlighting the imperative for long-term wearable sensors and flexible electrodes, particularly for impedance and temperature measurements.

However, a major obstacle that persists is the large-scale, cost-effective production of these flexible electrodes. Today’s research landscape has not adequately resolved the issues of mass production and the simplification of manufacturing processes. In this setting, the z-conductive electrode designed by Ti2 emerges as a promising solution. This design is ideal for industrial-scale production, with conductive holes formed using a thermal needle technique and the surface embellished with screen-printed conductive silver ink. Ensuring that flexible electrodes are efficient and cost-effective is key. Utilising readily available materials and streamlined manufacturing processes, like screen printing, will bolster the feasibility of these smart bandages to achieve widespread adoption. In essence, the future of health monitoring hinges on the advancement of sensors that are both enduring and accurate. As the landscape evolves, the emphasis must shift towards producing flexible electrodes that can firmly adhere to human skin while ensuring they are also amenable to cost-effective, large-scale production.

This research principally considered two aspects:

1. Extending the experimental foundation established by Ti2 Company, this study conducted a comprehensive analysis of how different hole spacings in z-conductive electrodes affect their impedance. The objective was to discern how variations in the distance between the holes influence the overall performance of the electrodes.

2. In continuation of the first point, the research also explored the impedance variations associated with different hole diameters in z-conductive electrodes. The goal here was to determine the ideal hole diameter that would result in the lowest impedance, contributing to the identification of the most effective electrode design for future large-scale production.

Chapter 3

Methodology

3.1 Electrode Design

The electrodes utilised a patented patch from Ti2 in Australia as the substrate. The patch was perforated using a laser to construct a micro-pylon structure. Figure 3.1 depicts the microstructure of this flexible electrode substrate. In line with this structure, the silver ink was plated onto it to serve as the conductive structure.

3.1.1 Z-conductive Electrodes

The patented technology from Ti2 was based on the use of multilayer films as substrates (Figure 3.1). One such example is the first layer, a polyethylene (51 μm) PE carrier that works as a protective layer prior to the application of the overall dressing. A 25 μm thick polyurethane (PUR) film is the second layer, which acts as the substrate for the conductive ink coating. The third layer consists of an acrylic adhesive (19 μm), which functions as a binder for the dressing and allows the eventual flexible electrode device to adhere to the human body. The final layer, a PE-coated paper liner (178 μm), is the liner for the dressing. All four layers, when combined, form a multi-layer structure that is fundamental to the functionality of the flexible electrode.

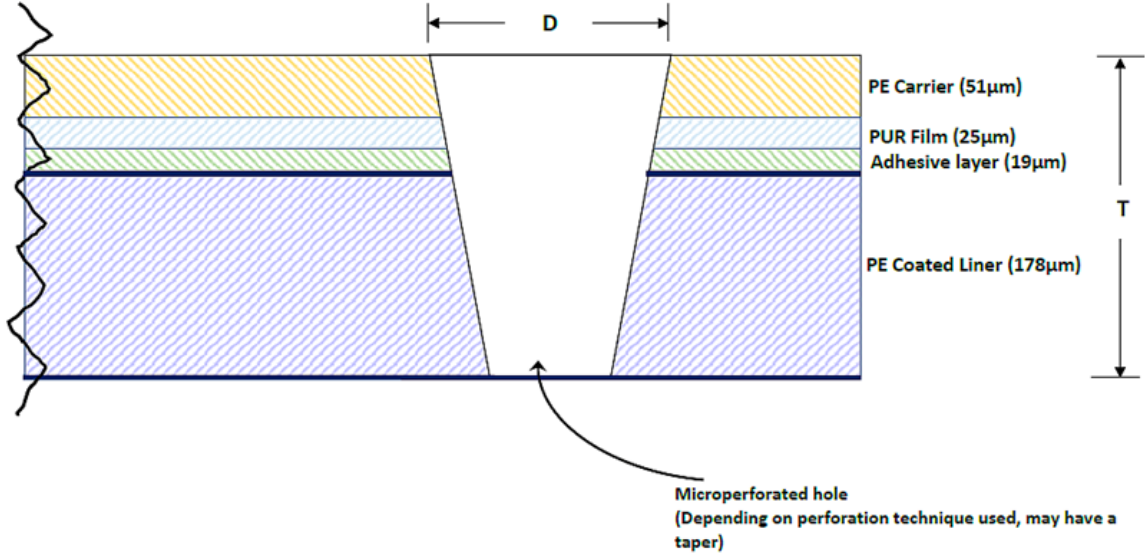


Figure 3.1: Cross-sectional view of the multi-layer film substrate used in this study to prepare Z-conductive electrodes

3.1.2 Manufacturing Process

The fabrication of this flexible electrode was built upon the multi-layer film provided by Ti2. The design process began with CorelDRAW, a vector graphics editor software, to devise a micro-pylon structure design for an A4-sized dressing. Upon completing the design file, it was submitted to the Australian National Fabrication Facility (ANFF), where a laser was used to perforate the dressing based on the design, thereby forming a micro-pylon array structure. Creative Materials 118-38 (EU) silver ink was then applied to the perforated film with the Doctor Blade Coating technique. The coated material was then oven-dried at 80 °C for 10 minutes prior to being cut into the appropriate dimensions to complete the fabrication of the flexible electrodes.

3.1.3 Electrode Hole Design

For this study, CorelDRAW, a graphic design software, was used to draw the locations and sizes of the perforations in the substrate. There were three different hole spacings and five different hole diameters combined. Each software file template specified a uniform hole spacing along with five different hole diameters. Figure 3.2 displays the splitting of the sample with a hole spacing of 1 mm into five columns, each with five

strips allocated for designated electrodes with the same hole spacing and diameter. Each designed electrode adopted a hexagonal arrangement (Figure 3.3). All micro-pylon electrodes were formed from 100 holes, laid out in a 10x10 grid with a hexagonal pattern and a 1 mm centre distance between each one.

Moreover, Figure 3.3 shows the combination of a 1 mm inter-hole distance and a 0.28 mm hole diameter. The designed micro-pylon structure and the surrounding patch, which was 1 mm on the left and right, 5 mm on the top, and approximately 25 mm on the bottom, form a flexible electrode unit roughly 12-17 mm wide and 40 mm long. The area of each flexible electrode unit was not uniform because of the different hole spacings.

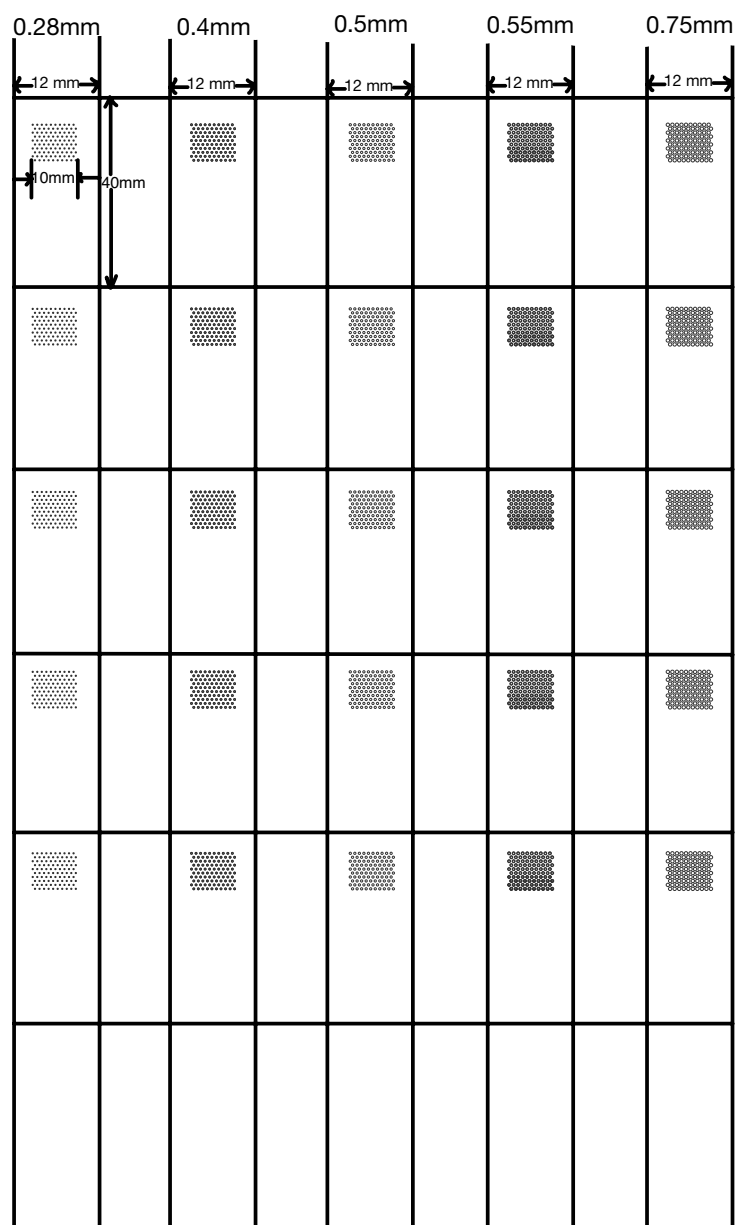


Figure 3.2: Sample design drawings overall

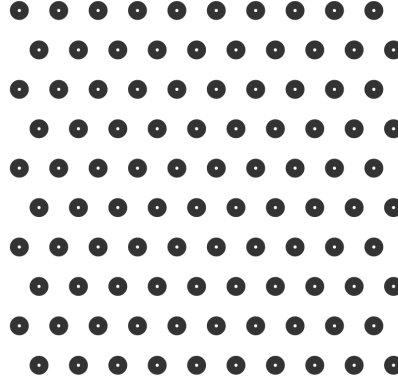


Figure 3.3: Hole Sample

3.1.4 Manufacturing Method

Upon completing the sample design, the samples were forwarded to the Laser Manufacturing Department of the ANFF at Macquarie University to have the material processed by laser perforation. Currently, the more expensive laser perforation method is the only option available for small-scale manufacturing. Once the optimal combination of hole spacing and hole diameter is identified, scaling up production will become possible. For mass production, the hot needle method, also known as thermal perforation, will be utilised. It is a technique that uses a heated needle array to puncture holes in a material. The needle array is heated to a high temperature and then pressed against the material, melting or burning a hole through it. This method is more economical and ideally suited for large-scale industrial production.

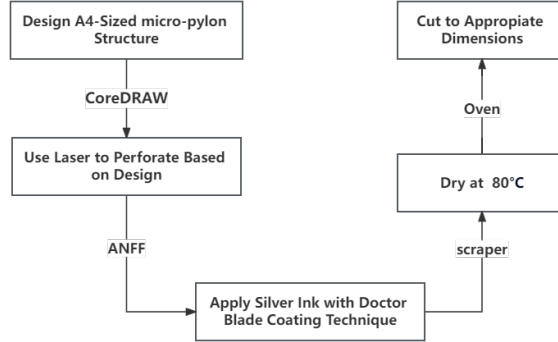


Figure 3.4: Ti2 Dressing Fabrication Process

3.2 Printing Technology

3.2.1 Ink

The market has an extensive selection of conductive inks, each tailored for specific printing applications. However, the range of options available for the fabrication of bio-flexible electrodes for human use is limited due to considerations such as biocompatibility, conductive efficiency, and cost-effectiveness. Consequently, the main alternatives are narrowed down to silver, gold, carbon, and conductive polymer inks [9].

Printed electronics frequently utilise silver for its superior conductivity relative to carbon and its cost-effectiveness in comparison to gold. Furthermore, silver’s superior biocompatibility makes it an optimal choice. Silver inks are crafted from silver in the form of nanoparticles or flakes, combined with a polymer binder and dissolved in a solvent [52].

Certain additives may be added in small amounts to some inks to augment certain characteristics. For example, such additives could be used to boost adhesion to specific substrates or increase resistance against certain chemicals. Table 3.1 itemises the key parameters of frequently used inks.

Since the silver in the ink exists as distinct, solid nanoparticles dispersed in a solvent, implementing a heat treatment procedure following the printing process is necessary. This thermal operation allows the solvent to evaporate, and the individual silver nanoparticles to merge into a conductive trace. Using an oven, the samples were subjected to heat treatment in an oven, with the optimal annealing temperature ranging

Printing Technique	Example	Sheet Resistivity max. (Ω /sq/mil)	Consistency	Crease Resistance	Thermal Stability ($^{\circ}\text{C}$)
Screen printing	Creative Materials 124-02	0.05	Thin Liquid	Excellent	175
Screen printing	Creative Materials 118-38(EU)	0.055	Liquid	Excellent	175
Spray coating	Creative Materials 127-10	0.1	Low Viscosity Liquid	Good	150
Inkjet printing	Novacentrix JS-A102	25	Liquid	Good	100

Table 3.1: A comparison of different printing techniques.

between 55-175 $^{\circ}\text{C}$, depending on the composition of the substrate and the properties of the ink. It is of paramount importance that the annealing temperature does not adversely affect the substrate's properties, as excessive heat could potentially cause substrate deformation [11].

3.2.2 Printing and Coating Methods

This research conducted a comparative evaluation of four different diverse printing and coating fabrication techniques, namely inkjet printing, doctor blade coating, stencil printing, and screen printing. Every method had its own merits and drawbacks. Choosing the coating technique was contingent upon the specific phase of the project. For instance, during the preliminary experimental stage, doctor blade coating was favoured for its cost-effectiveness. However, after successfully determining the optimal electrode configuration, a transition to screen printing was envisaged to facilitate mass production.

Inkjet coating Also known as inkjet printing, inkjet coating is a non-contact additive manufacturing process where ink is loaded into the cartridge of an inkjet printer. In a controlled manner, the printer deposits the ink onto a substrate. This technique affords significant flexibility and is capable of handling various materials and creating intricate, detailed designs. Its precision stems from its capacity to deposit material droplet by droplet, therefore crafting finely detailed structures. Since inkjet printing

is a non-contact process, it can print on fragile or flexible substrates without causing damage, but it also has its limitations. One major challenge is the production speed, considering that the pattern is done droplet by droplet.

Spray coating Also known as spray deposition, spray coating is a versatile coating technique that can uniformly apply a diverse range of materials to various substrates. Its advantage involves the ability to achieve uniform coatings over wide areas and complex geometries, making it suitable for a variety of applications, including the manufacturing of flexible electrodes. However, spray coating also has its limitations. The printed pattern's quality can be influenced by different variables, such as ink viscosity, nozzle size, and the distance between the nozzle and the substrate. Therefore, to ensure the required print quality, these parameters must be carefully controlled [37].

Doctor Blade Coating Doctor Blade Coating, or blade coating, is used across several different industries when applying a thin material film onto a substrate. It uses a blade to spread a liquid material evenly across a surface. The blade, also referred to as the 'doctor blade', is set at a specific angle and moves across the surface, spreading the material and scraping off any excess to ensure a uniform coating. Doctor Blade Coating offers a number of advantages, particularly in terms of simplicity and cost-effectiveness. The process is relatively straightforward and does not require the use of complex machinery or high energy consumption. As a result, it has become popular for many applications, specifically in producing thin films for electronics and other high-tech industries. The process also provides a remarkable level of control over the thickness of the coating, which can be modified by adjusting the angle and speed of the blade. Nevertheless, the limitations of Doctor Blade Coating involve several factors, including the material's viscosity, the blade's speed, and the substrate's smoothness, which can influence the quality of the coating. To guarantee a uniform and defect-free coating, these factors must be carefully controlled [8].

Screen Printing Screen printing, otherwise known as silk-screening or serigraphy, is a versatile printing technique that involves creating a stencil (known as a 'screen') and layering ink onto the printing surface. Performing the process requires placing a stencil over a screen and forcing ink through the porous fabric onto the printing area. One of its several advantages is its versatility and capacity to be used on different materials, such as textiles, ceramics, wood, paper, glass, and metal, thereby making it suitable for a wide range of applications. Next, it is cost-effective for large batches, where the same stencil

can be used to replicate a design multiple times. However, its disadvantages must also be noted. The initial setup cost can be high due to the demand for screens and stencils. This process is not suitable for high-resolution, detailed images. In cases where colour is vital and multiple coloured inks are to be used, the process can be time-consuming as each colour needs to be applied and dried individually. Lastly, it is not cost-effective for small production runs due to the high setup cost [5].

3.2.3 Coating Processes

Due to the production scale and cost limitations of the current research stage, the doctor blade method was the preferred method for coating. This is because screen printing is an unfeasible choice for coating. More specifically, the chosen method requires a significant level of expertise to be properly executed, necessitating many practice runs on blank paper to achieve a uniform coating thickness on the patch. Additionally, since the surface of the patch is made of polyurethane, using a metal blade was not an option since it could damage the surface structure. Therefore, a silicone spatula purchased from Woolworths for cooking was used for coating. If large-scale production is to be undertaken, screen printing could be the coating method of choice.

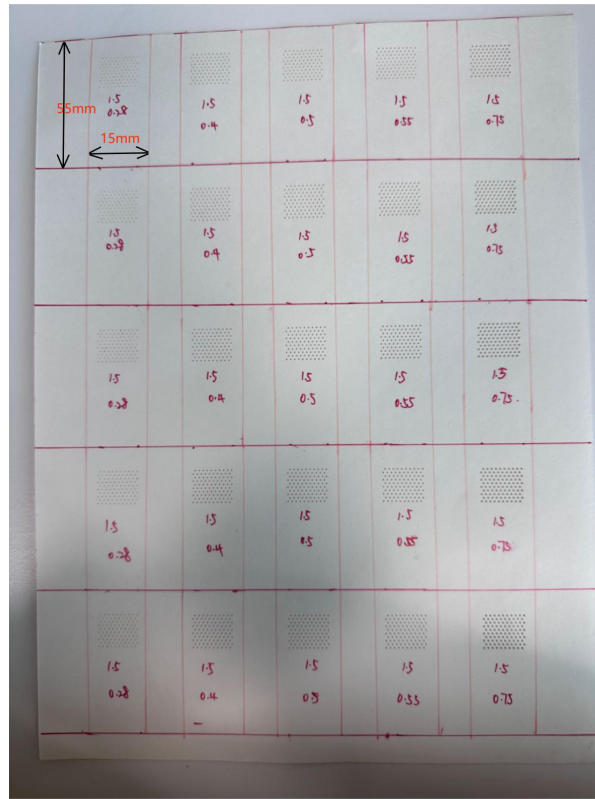


Figure 3.5: Front side of micro-perforated array patterns prior to cutting out of individual rectangles and the coating step to form Z-conductive electrodes

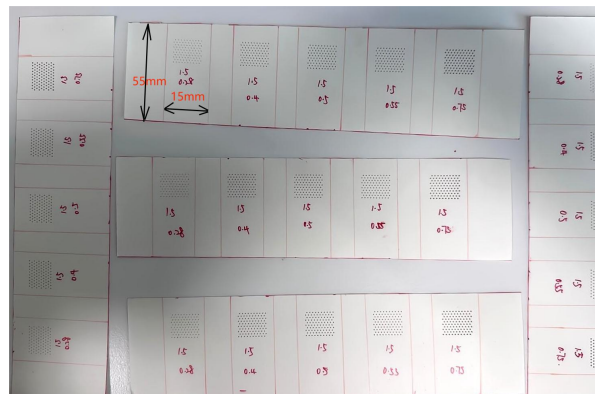


Figure 3.6: Back side of micro-perforated array patterns after cutting out of individual rectangles and prior to the coating step to form Z-conductive electrodes

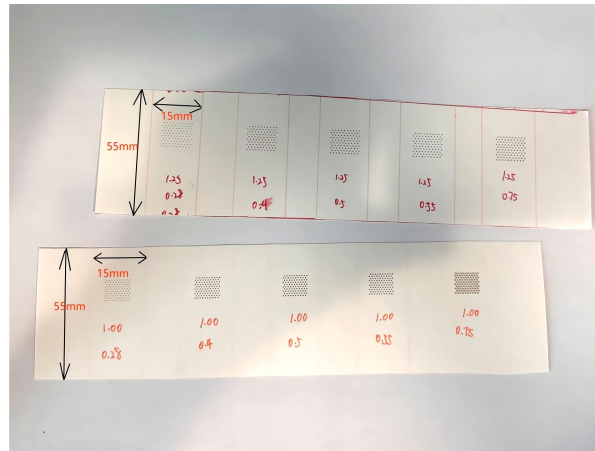


Figure 3.7: Back side of micro-perforated array patterns after cutting out of individual rectangles and prior to the coating step to form Z-conductive electrodes close-up view



Figure 3.8: The front side of two Z-conductive electrodes after coating and drying

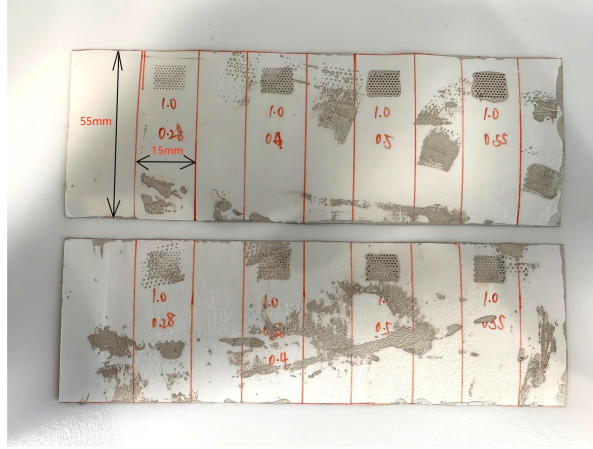


Figure 3.9: The backside side of two Z-conductive electrodes after coating and drying

After dividing and cutting the sample into five parts using a pair of scissors, each part was coated separately. Creative Materials 118-38 (EU) silver ink was applied to the surface of the patch and spread evenly with the silicone spatula. The objective was to make sure that the silver ink deposited on the material's surface was as wrinkle- and bubble-free as possible and that all polyurethane surfaces were completely covered with silver ink. To prevent silver ink nanoparticles from contaminating the environment, this step was conducted inside a fume hood, which maintained negative pressure. Afterwards, the patch with the applied silver ink was placed in a preheated oven. The temperature was maintained at 80 °C for 10 minutes to dry the silver ink. Using an electronic probe thermometer, the actual temperature in the oven was monitored to check that it was kept at 80 °C $\pm$ 5 °C. This temperature ensures the solvent was dried and that the polyurethane material was not severely deformed. Then, the material was removed from the oven to complete the coating.

3.3 Experimental Design

3.3.1 Measurement Tools

The ImpediMed SFB7 BioImpedance Spectroscopy (BIS) device is a vital tool in bioimpedance research. This state-of-the-art equipment is specifically engineered to deliver accurate bioimpedance measurements. It is configured to provide multi-frequency measurements ranging from 3 kHz to 1 MHz, thereby enabling a detailed examination of the impedance attributes inherent to biological tissues.

A salient feature of the SFB7 is its capability to detect both resistance (R) and reactance (X_c) - the dual facets of impedance. This feature enables a comprehensive understanding of the tissue's electrical properties. Moreover, the device is adept at procuring phase angle measurements, which further reveal cellular vitality and structural integrity.

In diminishing the influence of skin and electrode contact impedance, the tetrapolar electrode configuration employed by the SFB7 is pivotal, thus refining measurement accuracy. This setup comprises two electrodes dedicated to current injection and another pair for voltage measurement, strategically positioned on the skin.



Figure 3.10: The ImpediMed SFB7 BioImpedance Spectroscopy (BIS) Device

Compared with other similar options on the market, the SFB7 distinctly emerges superior, courtesy of its high-resolution, multi-frequency capabilities coupled with its ability to measure both resistance and reactance. This holistic approach to bioimpedance measurement furnishes an accurate and detailed insight into tissue characteristics. The device is complemented by an intuitive interface and specialised software, streamlining the processes of data acquisition and interpretation [19].

3.3.2 Measurement Method

The ImpediMed SFB7 was utilised in this study to measure bioimpedance through the two-electrode method. It was completed by connecting flexible electrodes to the dorsal and volar surfaces of the test subject's lower arm and connecting them to the SFB7.

The two-electrode measurement method, also known as the bipolar method, is a fundamental technique adopted in bioimpedance analysis. It plays a crucial role in the development of flexible electrodes for smart bandages. An excitation current is introduced into the biological tissue with the help of a pair of electrodes, and the resulting voltage drop across the tissue is measured.

In this study, the two-electrode method was employed to assess the bioimpedance of the skin under various conditions. Two flexible electrodes, designed with a microperforated structure, were affixed to the skin, one on each side of the left lower arm. Additionally, a current-controlled alternating current was applied through the SFB7 device. The device software then calculated the bioimpedance from the resultant voltage drop according to Ohm's law, considering the phase difference between the current and the voltage [28].

One of the main advantages of the two-electrode method is its simplicity, making it especially suitable for applications that require minimal setup like wearable health monitoring devices. However, the contact impedance at the electrode-skin interface affects this method. In this study, this issue was, to some extent, controlled by using adhesive electrode washers together with flexible electrodes. It sought to ensure well-defined contact with the skin and minimise variation in contact impedance [40].

In spite of its straightforward nature, the two-electrode method can generate complex results due to the inherent heterogeneity and anisotropy of biological tissues. In this case, meticulous interpretation of the results is necessary. In this study, a comprehensive bioimpedance analysis was conducted using a range of frequencies in a technique called impedance spectroscopy [39]. This can provide valuable insights into the electrical properties of the skin and the effectiveness of the flexible electrodes [28].

An indispensable tool in bioimpedance research, the two-electrode measurement method provides a direct and effective means of evaluating the performance of flexible electrodes in bioimpedance measurement. Its simplicity and versatility make it suitable for smart bandages and other wearable health monitoring devices. The insights acquired through this method are of great importance to the understanding of skin bioimpedance and the optimisation of electrode design for bioimpedance applications.

3.3.3 Measurement Operation

The measurements began with an inter-hole distance of 1.00 mm, followed by 1.25 mm, and concluded at 1.50 mm. In ascending order, the hole diameters were also

systematically varied, commencing with 0.28 mm, followed by 0.4, 0.5, 0.55, and 0.75 mm. That is, the combinations started from an inter-hole distance of 1.00 mm paired with a hole diameter of 0.28 mm, progressing to an inter-hole distance of 1.50 mm paired with a hole diameter of 0.75 mm. For each measurement, the selected frequencies were 5, 10, 20, 50, 100, 200, 300, and 500 kHz.

Throughout the measurements, the SFB7 would automatically and continuously record data at the selected frequencies. Each conductive electrode experimental group underwent 15-20 sets of measurements. Every set contained data for the 8 selected frequencies, with each frequency measurement generating values for resistance (Z') and reactance (Z''). Upon completing the measurements, the collected data were manually transferred from the SFB7 to MS Excel for editing and statistical analysis. To calculate the mean, median, variance, and standard deviation, all data sets were processed accordingly. All frequency data were then compiled into a single spreadsheet for ease of reference.

3.3.4 Measurement Subject

This thesis used conductive rubber as the initial substrate for measurements, employing both the two-electrode method and the four-electrode method to gauge the transverse impedance between the flexible electrodes and the conductive rubber. Moreover, the inherent impedance of the flexible electrodes was ascertained by comparing the data differences between the experimental electrode group and the control electrode group. In another phase, simulations of more practical scenarios were performed by testing on the human arm, and the inherent impedance of the flexible electrodes under real-use conditions was established by, again, comparing the data obtained with a control electrode group.

Measurement on Rubber Sheet The use of conductive rubber sheets, despite failing to provide an accurate phantom material for skin simulation, was essential in this research. This substrate was anticipated to be useful in screening future adhesive Z-conductive electrodes. Generally, this approach aimed to identify the combinations of hole size, hole spacing, and ink formulation that would yield the lowest transverse impedance. It was imperative that the mechanical and adhesive properties of the electrodes were not significantly compromised during this process. The presumption was that there would be a more or less positive correlation between the contact impedance of several electrodes on conductive rubber and that on skin. This assumption could be

verified once candidate Z-conductive electrodes become available for testing at a later time.



Figure 3.11: The adhesive washer on square rubber sheet



Figure 3.12: An adhesive washer and flexible electrode affixed to the opposite side of the rubber square

During the measurement process, a set of a 22 mm outer diameter and 8 mm inner hole diameter adhesive washer was affixed to a 22 x 22 x 1.5 mm square of conductive carbon-loaded silicone rubber sheet (UVOX Ltd., UK). Figure 3.10 shows the conductive

rubber sheet with the adhesive washer attached. The flexible electrodes were then adhered to the adhesive washers, making certain that the microperforated structure was centred over the hole in the washer (Figure 3.11). An adhesive washer and flexible electrode with the same specifications were also affixed to the opposite side of the rubber square.

Arm Measurement Skin impedance testing can be demonstrated using actual arm measurements. Comparable to the rubber measurements, the two-electrode method was utilised. By testing in real-world scenarios, we could gain a more realistic insight into the impedance between the flexible electrodes and the skin, as well as the potential issues that may arise during actual usage. This will aid in identifying the best-performing experimental group and optimise the use scenarios for the flexible electrodes.

Arm measurements are taken into account to cover the inherent variability and complexity of human skin, as well as to provide a more realistic testing environment. This includes factors such as skin moisture, thickness, and individual physiological differences, which can all influence the performance of the flexible electrodes. By performing measurements on the arm, the research can better anticipate and manage these variables, thus producing more robust and reliable electrode designs. Furthermore, such measurements can yield valuable insights into the comfort and wearability of the electrodes, which are indispensable for real-world applications.

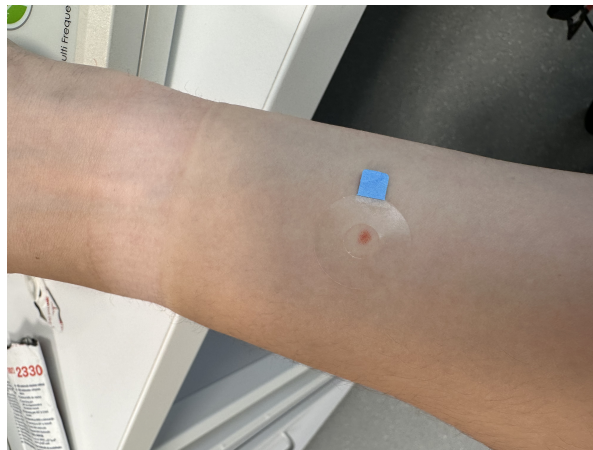


Figure 3.13: The volar surface of left lower arm with the affixed washer

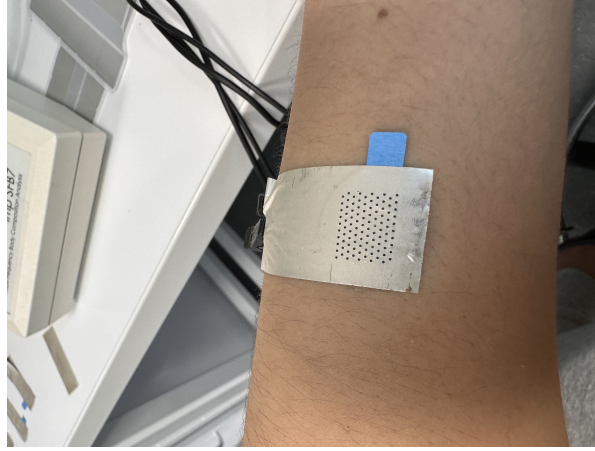


Figure 3.14: The arm with an attached flexible electrode

During the measurement process, an adhesive washer with an outer diameter of 22mm was placed in the middle of the subject's left forearm, and another was affixed to the opposite side of the arm. Afterwards, the flexible electrodes were then adhered to the washers on the arm in such a way that the electrode hole arrays were centred within the washers. Figure 3.12 presents the arm with the affixed washers, and Figure 3.13 exhibits the arm with the attached flexible electrodes ready for measurement.

3.4 Preliminary Hypothesis

For the forthcoming research on Z-conductive electrodes, the main hypothesis will revolve around the optimal combination of hole diameter and hole spacing to realise the most effective electrode performance. Drawing from preliminary assessments and theoretical considerations, the predictions are as follows:

1. Hole Diameter: Electrodes with smaller hole diameters are expected to demonstrate superior performance. The underlying rationale is that a reduced hole diameter may increase the capacity to retain conductive materials, such as silver, within the holes. This could likely cause improved contact with skin or materials, thereby reducing impedance values.

2. Hole Spacing: Larger hole spacings are anticipated to boost electrode performance. An increased inter-hole distance could reduce the risk of structural deformities or fractures, especially when the electrode is applied to dynamic surfaces. Additionally,

a wider spacing could also mitigate interference between adjacent holes, potentially generating more precise and consistent measurements.

In essence, the hypothesis implies that the most optimal combination for Z-conductive electrodes would be configurations with the smallest feasible hole diameter combined with the largest possible hole spacing. Such a combination is projected to balance mechanical robustness with optimal performance, and the succeeding experiments aim to validate or refute this hypothesis.

Chapter 4

Results

In order to determine the optimal combination of hole diameter and hole spacing for Z-conductive electrodes, which leads to the lowest impedance for future mass production, measurements need to be conducted on various substrates to observe the changes in impedance. Initially, measurements are taken on rubber to identify the combination that results in the lowest impedance with the least amount of interference from external factors. This involves affixing the electrodes onto a rubber substrate and connecting them to an SFB7 to measure their impedance. This process will accumulate experience in impedance measurement using Z-conductive electrodes.

Following this, measurements of human body impedance will be conducted using Z-conductive electrodes to determine the relationship between the lowest impedance of Z-conductive electrodes in practical use and the combination of hole diameter and hole spacing. The results of the experiments will elucidate the impact of hole spacing and hole diameter on impedance.

This information will be summarized in the discussion section, assisting in identifying the optimal combination of hole diameter and hole spacing to facilitate future mass production of Z-conductive electrodes.

4.1 Rubber as Test Substrate

Figure 4.1 illustrates the impedance values of the conductive rubber when measured using a coated Z-conductive electrode across various low frequencies. The 3M Red Dot wet electrode control group is also included in the graph for comparative analysis.

Evidently, there is a pronounced decrease in the overall impedance value alongside an increased measurement frequency.

At 5 kHz, the Z-conductive electrode configuration with a hole spacing of 1.25 mm and a hole diameter of 0.28 mm registered the lowest impedance value of 745.65 Ω . At 10 kHz, the same electrode configuration (with a spacing of 1.25 mm and a diameter of 0.28 mm) exhibited the lowest impedance, registering 589.24 Ω . At 20 kHz, the Z-conductive electrode with a hole spacing of 1.25 mm and a hole diameter of 0.5 mm registered the lowest impedance value of 541.42 Ω . Interestingly, at 50 kHz, the electrode with a hole spacing of 1.25 mm and a hole diameter of 0.28 mm exhibited the lowest impedance, registering 725.7 Ω .

It is noteworthy that the measurement taken at a frequency of 20 kHz showed a slight increase in the impedance value rather than a decrease. This deviation from the expected trend suggests that there are factors that may potentially influence the impedance measurements at this specific frequency.

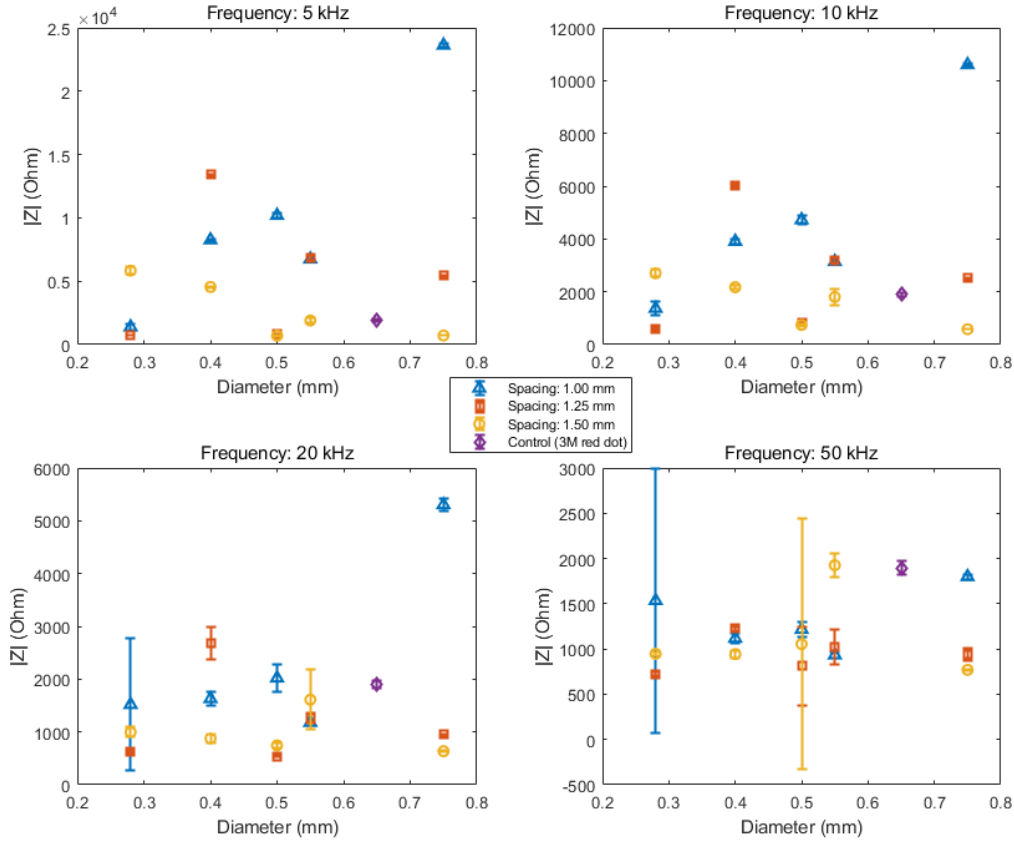


Figure 4.1: Impedance Analysis of Z-conductive Electrodes on Rubber Substrate at 5, 10, 20, and 50 kHz **Note:** For comparative purposes, the data for the 3M Red Dot control group has been plotted at the 0.65mm position. This does not indicate that it has holes.

Figure 4.2 illustrates the impedance measurements of the conductive rubber using the coated Z-conductive electrode at higher frequencies. The 3M Red Dot wet electrode control group is plotted for easy comparison. A distinctly observable trend is the pronounced reduction in impedance alongside higher frequencies.

In the high-frequency range of 100 kHz to 500 kHz, the Z-conductive electrode configuration with an inter-hole spacing of 1.5 mm and a hole diameter of 0.28mm consistently exhibited the lowest impedance. Specifically, this configuration achieved an impedance value of 53.43Ω at a frequency of 300 kHz.

It is worth noting that, as the frequency increased to 500 kHz, the impedance value

for the same electrode configuration (with a spacing of 1.5 mm and a diameter of 0.28 mm) increased slightly, registering 13.95Ω . This deviation from the expected trend (where impedance typically decreases alongside higher frequencies) suggests that there are factors or anomalies that may potentially influence the impedance measurements at this specific frequency.

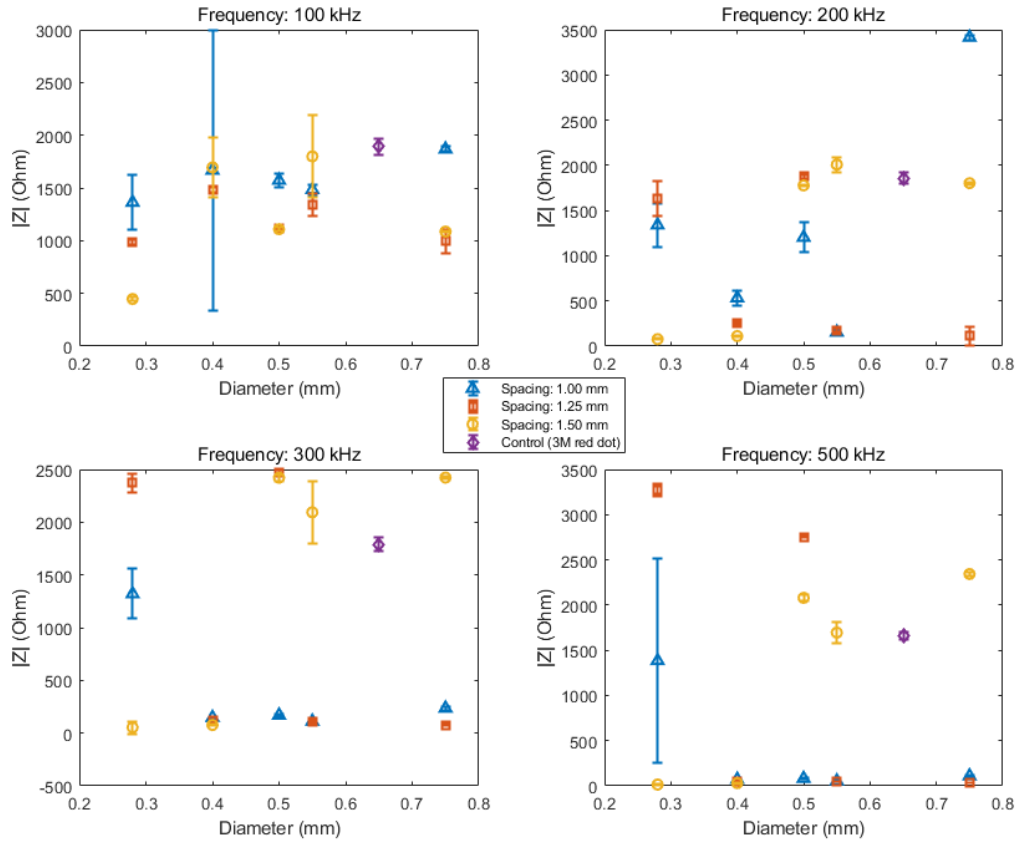


Figure 4.2: Impedance Analysis of Z-conductive Electrodes on Rubber Substrate at 100, 200, 300, and 500 kHz **Note:** For comparative purposes, the data for the 3M Red Dot control group has been plotted at the 0.65mm position. This does not indicate that it has holes.

Figure 4.3 illustrates the impedance values of conductive rubber (measured using a Z-conductive electrode coated with silver ink) across varying frequencies after undergoing a natural logarithmic transformation. The data demonstrate the effect of varying hole diameters on the impedance. Several observations can be made from the presented

results:

For the samples with a hole diameter of 0.28 mm, the impedance of the Z-conductive electrodes with a hole spacing of 1.50 mm exhibited the most pronounced decline.

Notably, at lower frequencies, the Z-conductive electrodes with a hole spacing of 1.25 mm registered the lowest impedance.

For the samples with a hole diameter of 0.40 mm, the impedance values of all the Z-conductive electrodes consistently decreased alongside higher frequencies. Among these samples, the electrodes with a hole spacing of 1.50 mm consistently exhibited the lowest impedance values across all spacings.

For the samples with a hole diameter of 0.50 mm, the Z-conductive electrodes with a spacing of 1.00 mm demonstrated the most pronounced reduction in impedance at higher frequencies. They also registered the lowest impedance at higher frequencies. However, the samples with a spacing of 1.50 mm exhibited a marginally lower impedance at lower frequencies.

For the samples with a hole diameter of 0.55 mm, the impedance curves for the Z-conductive electrodes with spacings of 1.00 mm and 1.25 mm were strikingly similar. Both displayed a marked decrease in frequency.

For the samples with a hole diameter of 0.75 mm, the Z-conductive electrodes with spacings of 1.00 mm and 1.25 mm followed a similar declining trend. However, in terms of their overall impedance values, the samples with a spacing of 1.25 mm consistently registered lower impedance values.

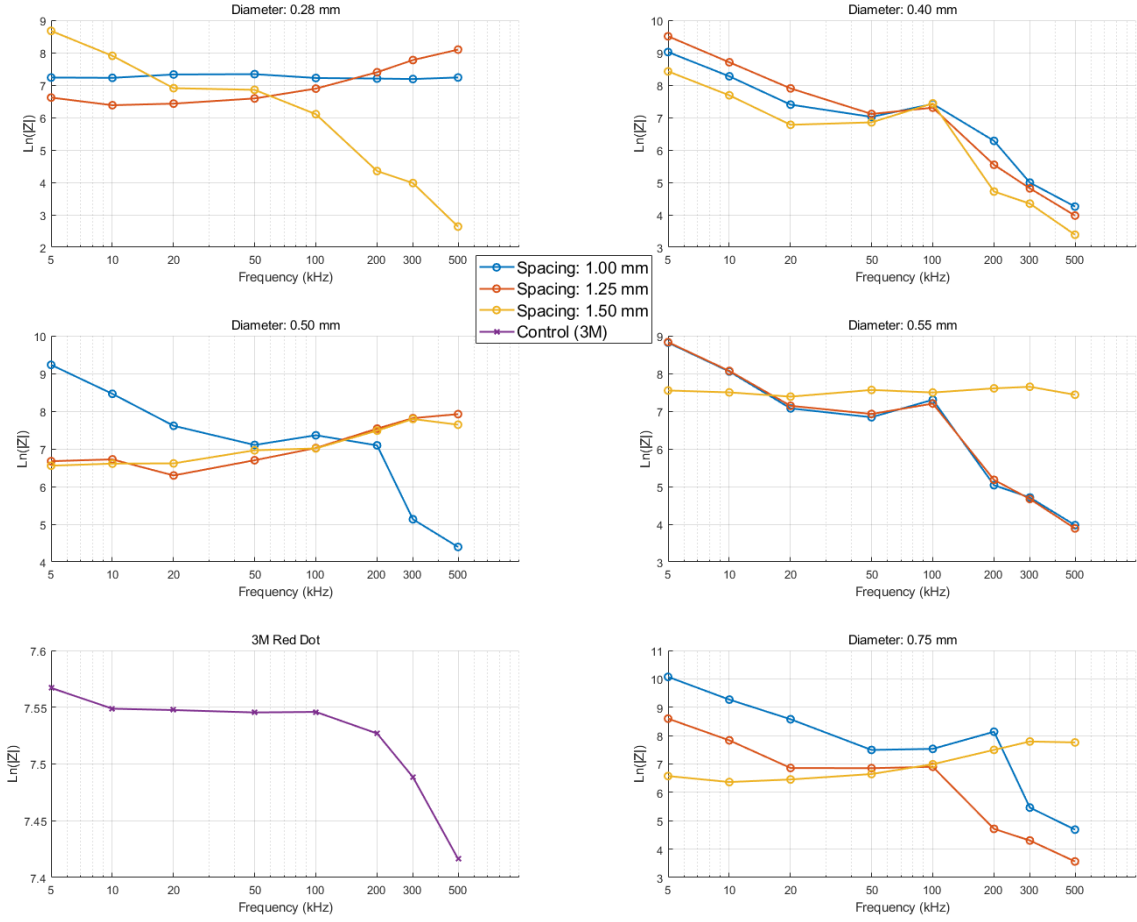


Figure 4.3: Logarithmic Impedance Analysis of Z-conductive Electrodes on Rubber Substrate Across Logarithmically Scaled Frequencies with Varying Hole Diameters

Figure 4.4 illustrates the impedance values of conductive rubber, measured using Z-conductive electrodes coated with silver ink after undergoing a natural logarithmic transformation, across various frequencies. The data reveals the influence of different hole spacings on impedance. Several conclusions can be drawn from these results:

Most of the samples exhibited a similar impedance variation trend, namely that the impedance decreased alongside higher frequencies. Notably, there was a slight increase in impedance from around 100 kHz to 200 kHz, after which it continued to decline. All the samples with a hole spacing of 1.00 mm (except for the sample with a hole diameter of 0.28 mm) followed an identical impedance variation trend. Notably, the samples with a hole diameter of 0.55 mm consistently demonstrated the lowest impedance values. For

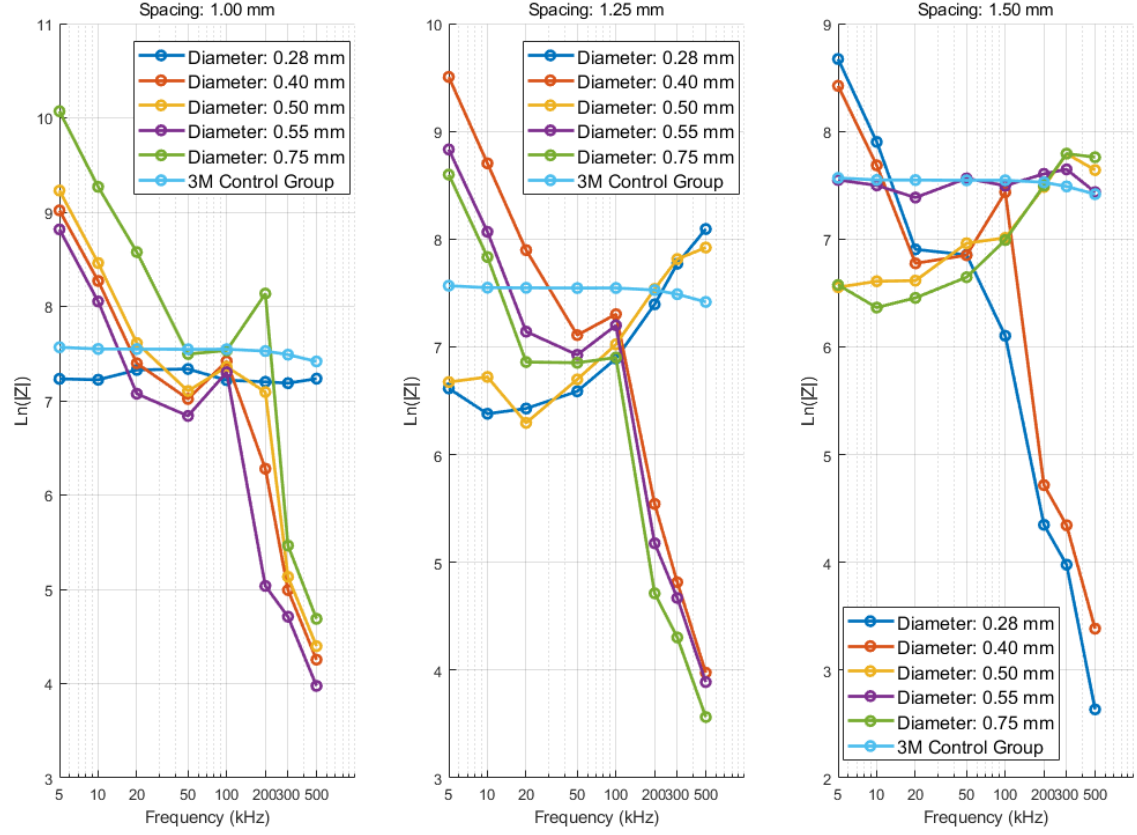


Figure 4.4: Logarithmic Impedance Analysis of Z-conductive Electrodes on Rubber Substrate Across Logarithmically Scaled Frequencies with Varying Hole Spacings

the samples with a hole spacing of 1.25 mm, those with hole diameters of 0.40 mm, 0.55 mm, and 0.75 mm exhibited similar impedance variation trends. For the samples with a hole spacing of 1.50 mm, those with hole diameters of 0.28 mm, 0.40 mm, and 0.75 mm also followed a comparable impedance variation trend.

4.2 Arm Experiment

Figure 4.5 presents the impedance values of a left human arm measured in the first experiment using coated Z-conductive electrodes at low frequencies. The 3M Red Dot wet electrode control group consistently exhibited the lowest impedance values across all frequencies. Among the Z-conductive electrodes used, the configuration with a hole spacing of 1.00 mm and a hole diameter of 0.28 mm exhibited the lowest impedance

value of 5085.98Ω at a frequency of 5 kHz. At 10 kHz, the Z-conductive electrode with a hole spacing of 1.00 mm and a hole diameter of 0.28 mm again showed the lowest impedance, registering 2364.88Ω .

Similarly, at 20 kHz, the same electrode configuration exhibited the lowest impedance value of 912.59Ω . As the frequency increased, there was a general decline in impedance values across all the electrode configurations, with the aforementioned configuration consistently exhibiting the lowest impedance.

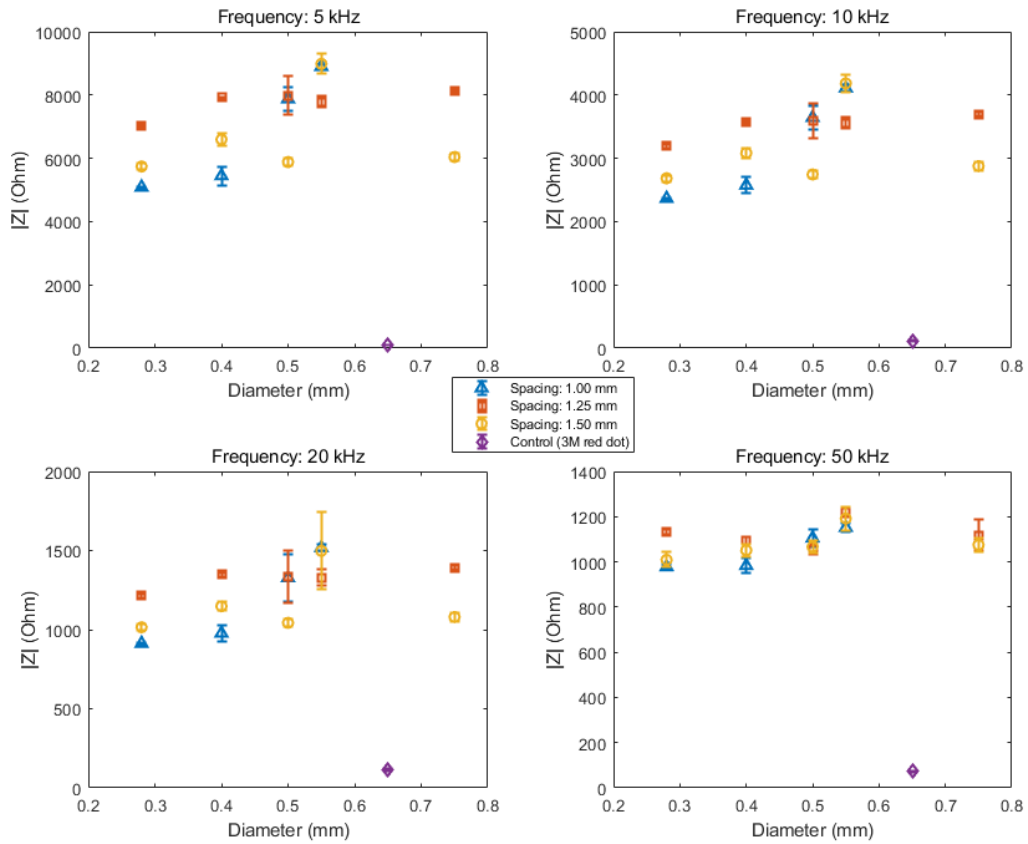


Figure 4.5: First Experiment: Impedance Analysis of Z-conductive Electrodes on Human Arm at Frequencies of 5, 10, 20, and 50 kHz. **Note:** For comparative purposes, the data for the 3M Red Dot control group has been plotted at the 0.65mm position. This does not indicate that it has holes.

Figure 4.6 presents the results from the second experiment where the impedance measurements were taken on a left human arm using coated Z-conductive electrodes at

lower frequencies. The 3M Red Dot wet electrode control group is also plotted on the graph for ease of comparison.

Within the frequency range of 5 kHz to 50 kHz, the Z-conductive electrode configuration with an inter-hole spacing of 1.5 mm and a hole diameter of 0.28 mm consistently demonstrated the lowest impedance values.

Specifically, at 20 kHz, this configuration achieved an impedance of 700.05 Ω . Interestingly, as the frequency increased to 50 kHz, the impedance value for the same electrode configuration (with a spacing of 1.5 mm and a diameter of 0.28 mm) increased slightly, registering 1358.25 Ω . This deviation from the expected trend (where impedance typically decreases alongside higher frequencies) suggests that there are factors or anomalies that may potentially influence the impedance measurements at this specific frequency.

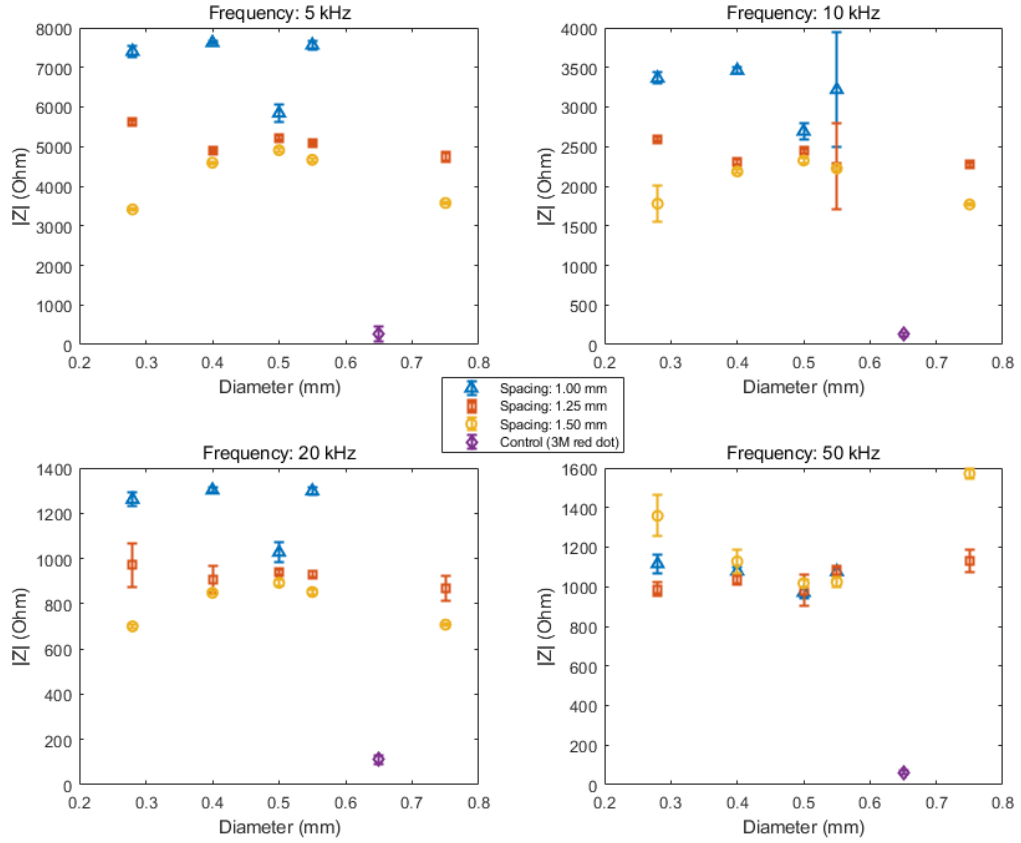


Figure 4.6: Second Experiment: Impedance Analysis of Z-conductive Electrodes on the Human Arm at Frequencies of 5, 10, 20, and 50 kHz. **Note:** The 3M Red Dot control group data, plotted at 0.65mm, does not indicate holes.

Figure 4.7 shows the results of the first experiment where the left human arm impedance measurements were captured using coated Z-conductive electrodes at higher frequencies. The 3M Red Dot wet electrode control group is also plotted for ease of comparison.

Across the frequency range of 100 kHz to 500 kHz, the Z-conductive electrode configuration with a spacing of 1 mm and a diameter of 0.5 mm exhibited significant fluctuations in impedance values. Specifically, at 100 Hz, this configuration registered an unusually high impedance of 1762.8 Ω .

As the frequency increased to 200 kHz, the impedance values generally declined. The configuration with a diameter of 0.4 mm and a spacing of 1 mm registered the lowest

impedance of 79.37Ω . Conversely, the configuration with a spacing of 1.5 mm and a diameter of 0.5 mm registered a high impedance of 455.9Ω .

At a frequency of 300 kHz, the impedance measurements remained relatively consistent across all different configurations. The 3M Red Dot control group consistently exhibited the lowest impedance at 40.63Ω .

The impedance values decreased further at a frequency of 500 kHz. The configuration with a diameter of 0.4 mm and a spacing of 1 mm registered the lowest impedance of 14.13Ω . The 3M Red Dot control group maintained its low impedance trend, registering a value of 37.06Ω .

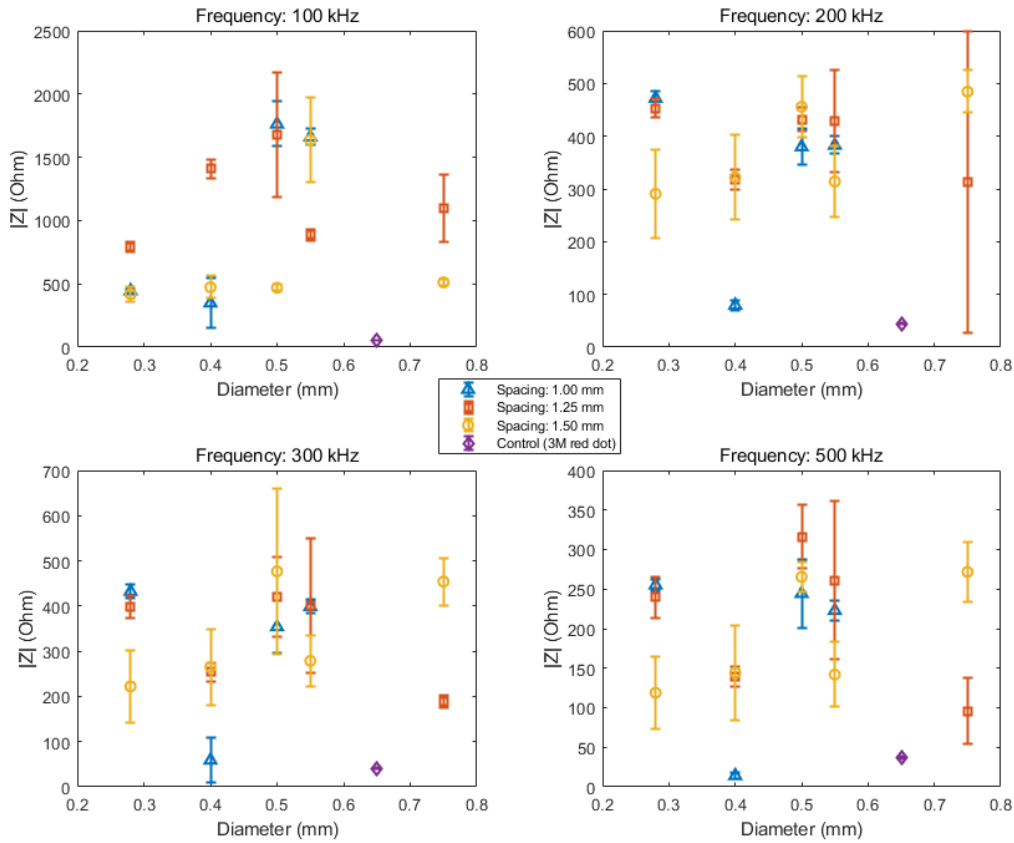


Figure 4.7: First Experiment: Impedance Analysis of Z-conductive Electrodes on Human Arm at Frequencies of 100, 200, 300, and 500 kHz. **Note:** For the purpose of the 3M Red Dot control group, data has been plotted at the 0.65mm position. This does not indicate that it has holes.

Figure 4.8 presents the outcomes of the second experiment. Specifically, it shows the impedance measurements of a left human arm when utilizing a coated Z-conductive electrode at higher frequencies.

At a frequency of 100 kHz, the Z-conductive electrode with a spacing of 1 mm and a diameter of 0.5 mm exhibited an impedance value of 386.67 Ω . This configuration offers insights into the potential biomedical applications of Z-conductive electrodes.

At 200 kHz, the impedance values for most of the Z-conductive electrode configurations decreased. Notably, the configuration with a 0.4 mm diameter electrode and a spacing of 1 mm registered an impedance of 257.58 Ω . This suggests that smaller electrode diameters at closer spacings may be more effective at this particular frequency.

At a frequency of 300 kHz, the Z-conductive electrode with a 0.4 mm diameter at a 1 mm spacing continued to perform well, achieving an impedance of 206.32 Ω . This consistent performance across various frequencies highlights the reliability of Z-conductive electrodes.

When the frequency was increased to 500 kHz, the Z-conductive electrodes' impedance values generally decreased further. The configuration with a 0.4 mm diameter electrode at a 1 mm spacing performed the best, registering an impedance of 107.64 Ω .

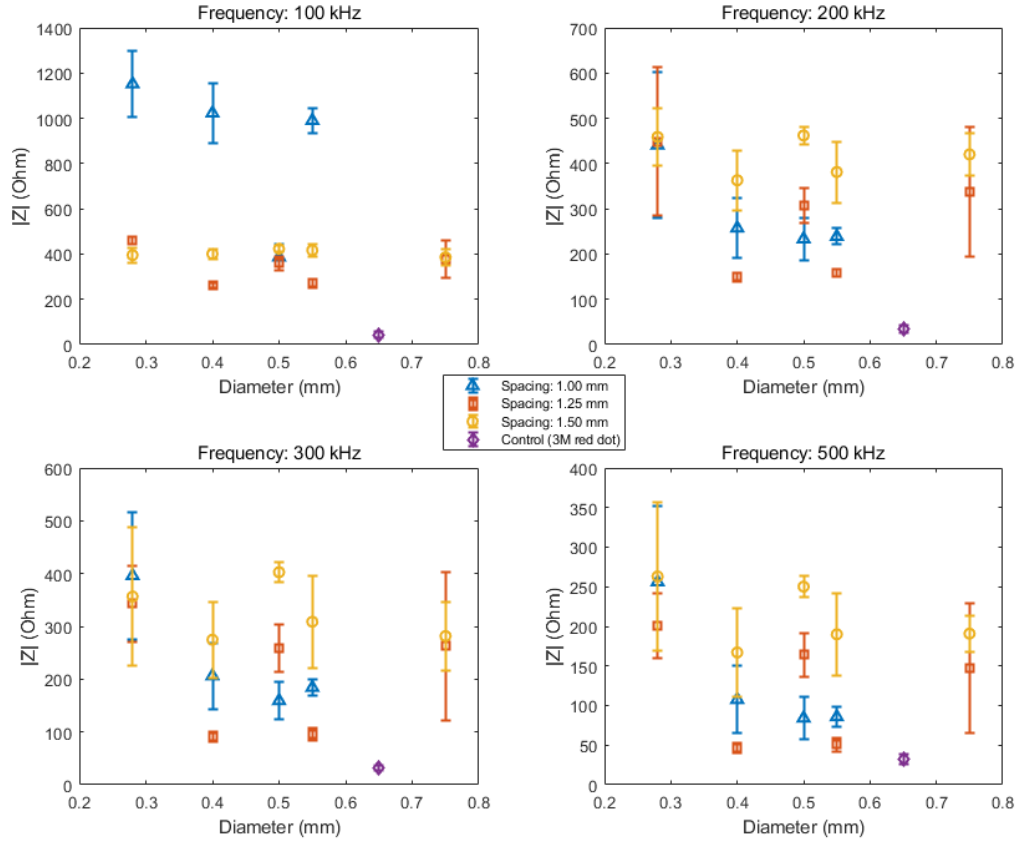


Figure 4.8: Second Experiment: Impedance Analysis of Z-conductive Electrodes on Human Arm at 100, 200, 300, and 500 kHz Frequencies. **Note:** The 3M Red Dot control group data, plotted at 0.65mm, does not indicate holes.

Figure 4.9 presents the impedance values of the human arm, measured across different frequencies using Z-conductive electrodes coated with silver ink, after undergoing a natural logarithmic transformation. These data illustrate the effect of varying hole diameters on impedance. Several observations can be made from the presented results:

All the Z-conductive electrode samples with a hole diameter of 0.28 mm exhibited similar impedance variation curves. However, at lower frequencies, the samples with a spacing of 1.00 mm exhibited the lowest impedance. At higher frequencies, the samples with a spacing of 1.50 mm registered the lowest impedance.

The samples with a hole diameter of 0.40 mm all demonstrated a consistent trend at lower frequencies. At higher frequencies, however, the impedance values for the samples

with a spacing of 1.00 mm declined most precipitously.

The samples with hole diameters of 0.50 mm and 0.55 mm showed no significant differences across all frequencies, aside from a slight deviation at 100 kHz.

Among the samples with a hole diameter of 0.75 mm, those with a spacing of 1.25 mm exhibited a rapid decline in impedance at frequencies above 100 kHz.

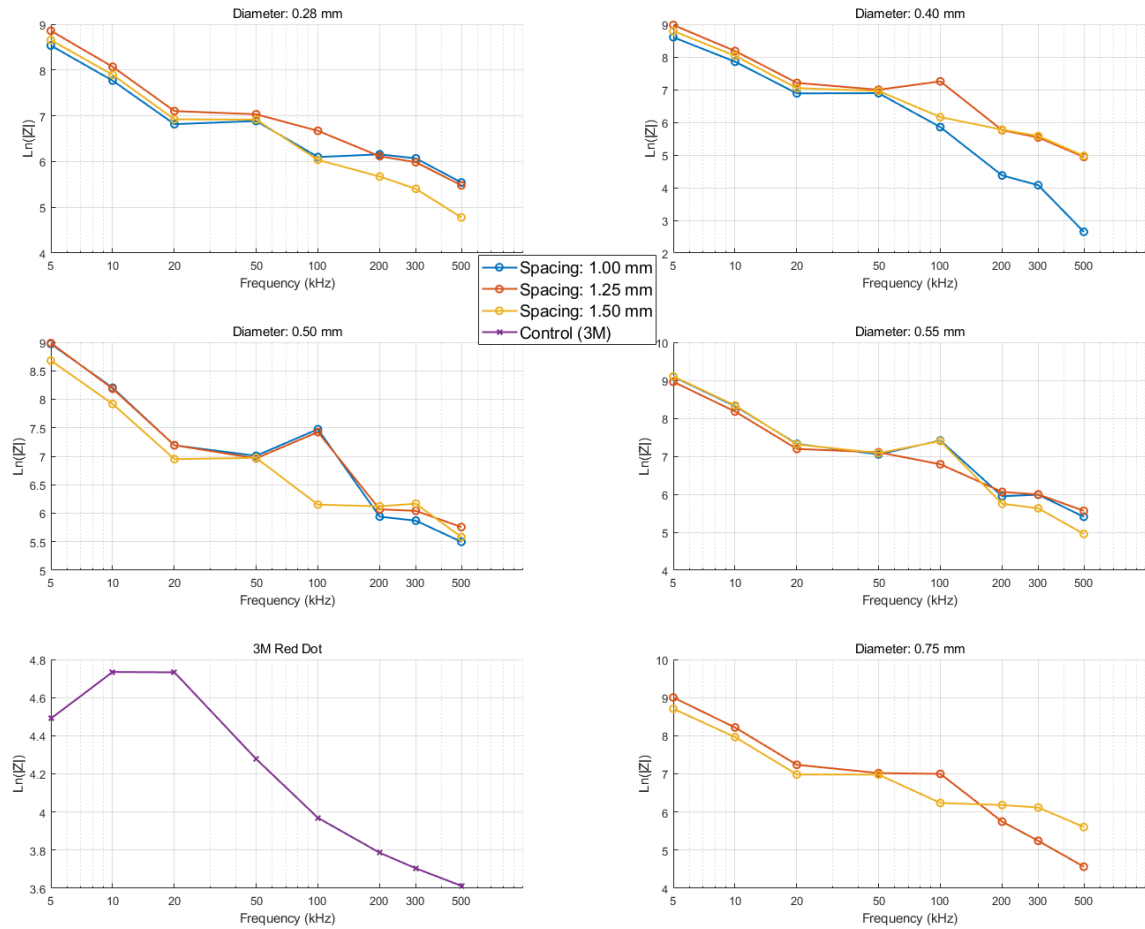


Figure 4.9: First Experiment: Logarithmic Impedance Analysis of Z-conductive Electrodes on Human Arm Across Logarithmically Scaled Frequencies with Varying Hole Diameters

Figure 4.10 illustrates the impedance values of a human arm obtained during a second measurement using Z-conductive electrodes coated with silver ink across various frequencies after undergoing a natural logarithmic transformation. These data

demonstrate the effect of varying hole diameters on the impedance values, revealing several noteworthy patterns:

The Z-conductive electrode samples with a hole diameter of 0.28 mm all showed similar impedance variation curves. Notably, at lower frequencies, the samples with a spacing of 1.50 mm exhibited the lowest impedance. At higher frequencies, the samples with a spacing of 1.25 mm registered the lowest impedance values.

The samples with a hole diameter of 0.40 mm showed a similar impedance variation trend. At higher frequencies, the samples with a spacing of 1.25 mm consistently exhibited the lowest impedance.

The impedance variation curves of the samples with a hole diameter of 0.50 mm are largely similar. At higher frequencies, however, the samples with a spacing of 1.00 mm exhibited the lowest impedance.

Regarding the samples with hole diameters of 0.55 mm and 0.75 mm at higher frequencies, the samples with a spacing of 1.25 mm exhibited the lowest impedance values.

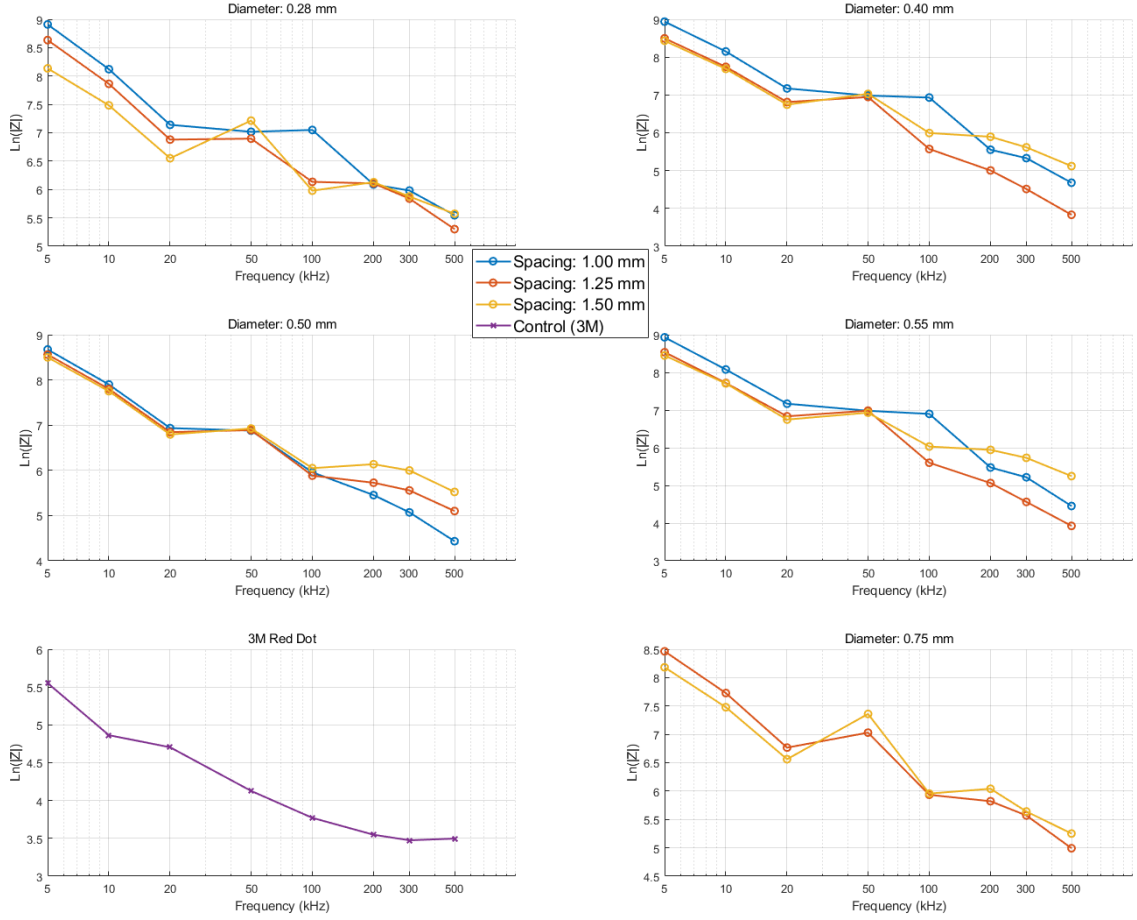


Figure 4.10: Second Experiment: Logarithmic Impedance Analysis of Z-conductive Electrodes on Human Arm Across Logarithmically Scaled Frequencies with Varying Hole Diameters

Figure 4.11 presents the impedance values of a human arm, measured in the first experiment using Z-conductive electrodes coated with silver ink. After undergoing a natural logarithmic transformation, these values are plotted against various frequencies to highlight the effects of different hole spacings on impedance. Several insights can be derived from the presented data:

All the samples exhibited a consistent trend, namely that the impedance decreased alongside higher frequencies, apart from a minor increase at around 100 kHz or 200 kHz followed by a continued decline.

Among the samples with a 1.00 mm hole spacing, the ones with hole diameters of

0.40 mm and 0.28 mm displayed nearly identical impedance values at lower frequencies (the lowest apart from the control group). However, at higher frequencies, the 0.40 mm sample demonstrated a pronounced reduction in impedance, distinctly lower than its counterparts. All the samples with a hole spacing of 1.25 mm showed nearly identical impedance values at lower frequencies. Nevertheless, the sample with a hole diameter of 0.75 mm stood out with a significant decrease in impedance at higher frequencies. Although the impedance values were nearly uniform at lower frequencies among the samples with a hole spacing of 1.50 mm, the sample with a 0.28 mm diameter exhibited the lowest impedance at higher frequencies.

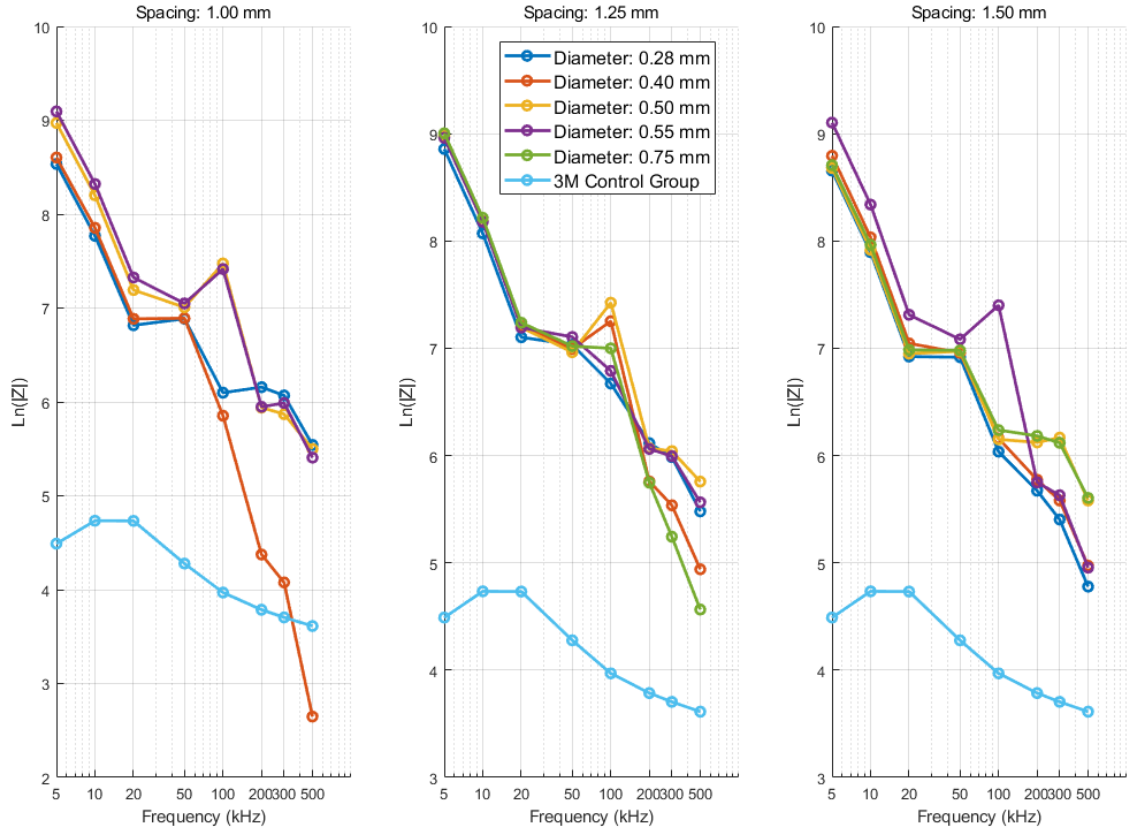


Figure 4.11: First Experiment: Logarithmic Impedance Analysis of Z-conductive Electrodes on Human Arm Across Logarithmically Scaled Frequencies with Varying Hole Spacings

The results presented in Figure 4.12 illustrate the second experiment's impedance values of a human arm, measured using Z-conductive electrodes coated with silver

ink. After a natural logarithmic transformation, the values are plotted across varying frequencies to underscore the nuanced influence of different hole spacings on impedance.

Across the board, a consistent trend emerges, namely that the impedance generally decreases alongside higher frequencies, apart from a minor uptick at around the 100 kHz or 200 kHz mark. Among the samples with a hole spacing of 1.00 mm, those with a hole diameter of 0.50 mm consistently outperformed their counterparts, registering the lowest impedance across all frequencies. In contrast, the samples with a 1.25 mm hole spacing displayed nearly identical impedance values at the lower end of the frequency spectrum. At higher frequencies, the samples with hole diameters of 0.40 mm and 0.55 mm distinguished themselves with notably low impedance values. Meanwhile, among the samples with a hole spacing of 1.50 mm, similar impedance values were noted at higher frequencies. In the lower frequency range, the samples with hole diameters of 0.28 mm and 0.75 mm emerged as the frontrunners, registering the lowest impedance.

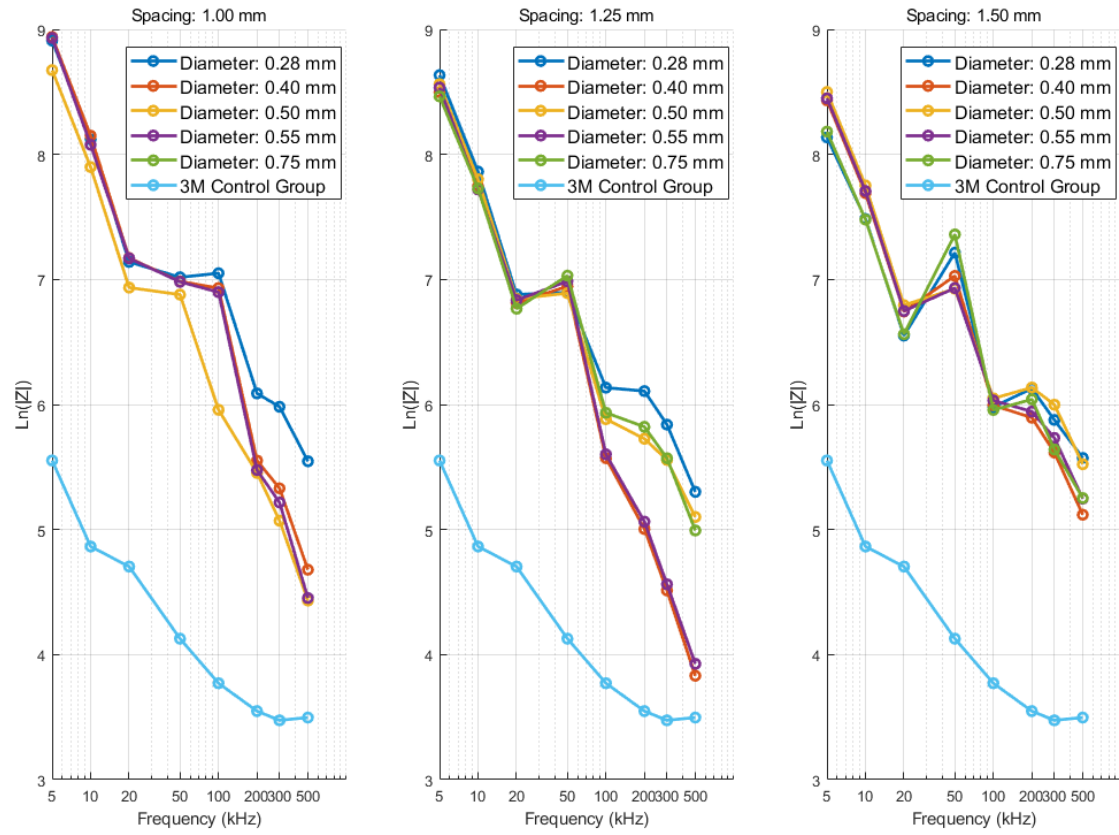


Figure 4.12: Second Experiment: Logarithmic Impedance Analysis of Z-conductive Electrodes on Human Arm Across Logarithmically Scaled Frequencies with Varying Hole Spacings

Chapter 5

Discussion

This section will delve into the similarities and deviations observed in the results obtained in the preceding section. This section will also explore the potential reasons for these deviations and propose improvements for future experiments.

5.1 Rubber Experiment

Comparing the experimental data in Figures 4.1 and 4.2, it is evident that there was no consistently low impedance for specific Z-conductive electrodes during low-frequency measurements. However, the electrodes exhibiting the lowest impedance values were consistently those with a spacing of 1.25 mm.

Among these, the configuration with a spacing of 1.25 mm and a hole diameter of 0.28 mm most often exhibited the lowest impedance. During high-frequency measurements ranging from 100 kHz to 500 kHz, the Z-conductive electrode with a spacing of 1.5 mm and a hole diameter of 0.28 mm consistently demonstrated the lowest impedance.

These observations suggest that the Z-conductive electrodes' configuration significantly influences its impedance, especially at higher frequencies. Measurements taken on conductive rubber indicate that configurations with smaller hole diameters and spacings (ranging from 1.25 mm to 1.5 mm) may provide lower impedance values.

Further examination of Figures 4.3 and 4.4 (where the impedance values are logarithmically transformed) highlights the differences between the samples. This reinforces the conclusions drawn from Figures 4.1 and 4.2 that, for measurements taken

on conductive rubber, hole spacings ranging between 1.25 mm and 1.5 mm yielded lower impedance values.

Predominantly, the electrode configurations with a hole diameter of 0.28 mm consistently demonstrated the most favourable impedance outcomes.

5.2 Arm Experiment

During the bioimpedance measurements on the arm, the Z-conductive electrode with an inter-hole distance of 1.0 mm and a hole diameter of 0.75 mm experienced some deformation and even fracturing while being removed from its protective layer and adhered to the arm. As depicted in Figure 5.1, all the central holes have fractured. This could be attributed to the inadequate strength of the polyurethane between the holes, compromising the structural stability, or possibly due to the thinness of the PUR layer. Consequently, the experimental results do not include data from the Z-conductive electrode with an inter-hole distance of 1.0 mm and a hole diameter of 0.75 mm, and this configuration was subsequently excluded from the study.



Figure 5.1: Broken Z-conductive Electrode with a Hole Spacing of 1.0 mm and a Hole Diameter of 0.75 mm.

Upon comparing the data from two separate low-frequency experiments, a consistent impedance pattern for the Z-conductive electrodes emerges. In the first experiment, the electrode with a 1.0 mm spacing and 0.28 mm hole diameter consistently exhibited the lowest impedance. Contrastingly, in the second experiment, while the electrode with a 1.0 mm spacing and a 0.5 mm hole diameter showed the lowest impedance at 50 kHz, the electrode with a 0.28 mm hole diameter outperformed its counterparts at other low frequencies (below 100 kHz). This highlights a recurring trend, namely that the electrodes with a 0.28 mm hole diameter tended to perform best at lower frequencies. This finding aligns with the results from the conductive rubber measurements. However, the consistency in spacing observed in the conductive rubber measurements was not mirrored in the human arm measurements.

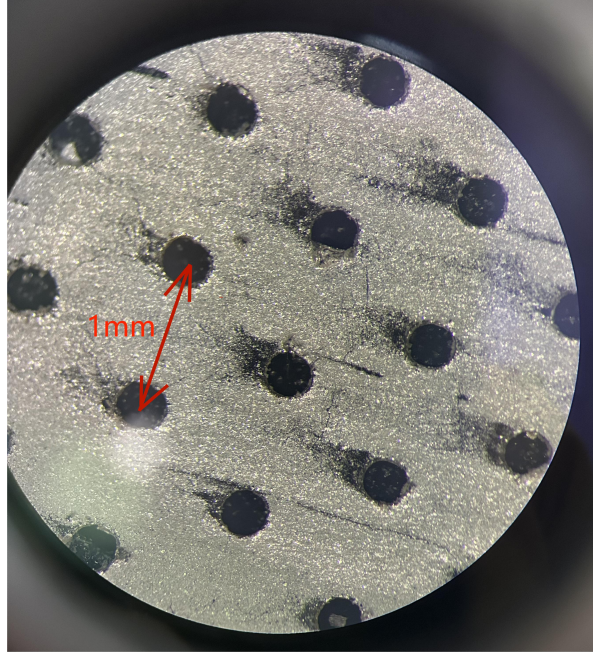


Figure 5.2: Z-conductive Electrode with a Hole Diameter of 0.28 mm Under the Microscope

One plausible explanation for this phenomenon might be the higher retention rate of silver within smaller-diameter holes. This allows the silver nanoparticles on the surface to have better skin contact, thereby leading to lower impedance values. This is further corroborated by Figure 5.2, which shows that the areas surrounding the holes were also coated with silver ink and surface-plated, potentially enhancing the conductivity.

Upon analysing the high-frequency measurements from two distinct experiments, several observations are evident. In the first experiment, the Z-conductive electrode with a spacing of 1.0 mm and a hole diameter of 0.4 mm consistently demonstrated the lowest impedance. Meanwhile, the electrode in the second experiment with a 1.25 mm spacing and the same hole diameter of 0.4 mm maintained the lowest impedance values. Across both experiments, the Z-conductive electrodes with a 0.4 mm hole diameter consistently exhibited superior performance at high frequencies. This trend underscores the efficiency of the 0.4 mm hole diameter configuration in Z-conductive electrodes operating at high frequencies. Notably, these findings diverge from the results obtained from the measurements taken on conductive rubber. Nonetheless, these observations align with the initial hypothesis.

Examining Figures 4.9 and 4.10,, it becomes clear that the logarithmic

transformation of the impedance values accentuates the disparities among the samples. However, even with this enhanced differentiation, discerning the impact of varying hole spacings on impedance across different frequencies remains challenging. Neither of the two figures provides a specific spacing configuration that exhibits significantly lower impedance. This observation aligns with our initial hypothesis, suggesting that identifying an optimal hole spacing may be more problematic than anticipated.

Detailed scrutiny of Figures 4.11 and 4.12 underscores the nuanced differences among the samples. In the initial experiment, the samples with a spacing of 1.00 mm (specifically those with hole diameters of 0.28 mm and 0.40 mm) exhibited remarkably similar and notably low impedance values at lower frequencies compared to other configurations. Conversely, the sample with a spacing of 1.00 mm and a hole diameter of 0.40 mm demonstrated exceptionally low impedance at higher frequencies, thus aligning with our pre-experimental hypotheses. In the second experiment, the sample with a hole diameter of 0.28 mm exhibited the lowest impedance at lower frequencies. The sample with a hole diameter of 0.40 mm prevailed at higher frequencies, thus aligning with our anticipatory assumptions.

In summary, a comprehensive analysis of the bioimpedance measurements taken on conductive rubber and a human arm has provided valuable insights into the performance of Z-conductive electrodes with varying configurations. The structural integrity of certain electrode configurations (particularly those with a 1.0 mm hole spacing and a 0.75 mm hole diameter) was found to be compromised, resulting in their exclusion from the study. The consistent performance of electrodes with a 0.28 mm hole diameter at lower frequencies and those with a 0.4 mm hole diameter at higher frequencies was a recurring theme across both experiments. This is suggestive of the potential role of silver retention in smaller-diameter holes and the consequent enhanced skin contact in determining Z-conductive electrodes' impedance values. Despite the logarithmic transformation of the impedance values, the challenges in discerning the optimal hole spacing highlight the intricacies involved in such measurements. This study's findings validate several of our pre-experimental hypotheses and pave the way for further research into optimising the design and performance of Z-conductive electrodes.

5.3 Performance Analysis

The Z-conductive electrodes utilised in this study are constructed with a polyurethane (PUR) film, which has a thickness of merely 25 μm . This ultra-thin design, while

advantageous for flexibility, presents significant challenges in terms of structural stability, especially when the inter-hole spacing is minimal and the hole diameter is substantially large. The insufficient strength of the surrounding polyurethane material under such configurations can compromise the electrode's structural integrity.

The surface of the Z-conductive electrodes is coated with a layer of silver ink. After the coating process, the silver layer adheres well to the surface of the electrodes, endowing the entire electrode with excellent bending stiffness. This ensures that the electrode can conform seamlessly to the contours of the skin. However, it is noteworthy that while the electrode demonstrates commendable bending stiffness, its tensile stiffness is not as impressive. When subjected to stretching, the silver coating on the surface tends to peel off, reflecting a potential limitation in applications requiring extensive elasticity.

In terms of fatigue performance, the Z-conductive electrodes showcased remarkable resilience. Throughout the experimental measurements, no noticeable changes were observed in the performance of the electrodes, and the silver coating remained intact without any signs of peeling or alteration. This is indicative of the electrodes' robustness and durability, even under repetitive use.

However, a noteworthy point of consideration is the use of acrylic adhesive in the construction of the Z-conductive electrodes. While this adhesive ensures a secure application of the electrode to the skin, it also renders the electrodes prone to sticking together post-use, which significantly hampers their reusability. Given this characteristic, combined with the potential issues arising from stretching, the Z-conductive electrodes have been designed and are best suited for single-use applications.

Chapter 6

Conclusions

6.1 Addressing the Research Questions

After conducting extensive bioimpedance studies and analyses, we are now equipped to address the initial research questions posed in this study. This section will outline our key findings and discuss how these findings answer our research questions.

1. How Can We Optimise the Construction of Z-Conductive Electrodes with a Specific Focus on Hole Diameter and Inter-Hole Spacing Configurations?

Our research has revealed the significant impact of hole diameter on the performance of Z-conductive electrodes, especially in lower frequency ranges. Through empirical observations, we found that electrodes with smaller hole diameters, particularly 0.28 mm, consistently outperformed their counterparts in achieving lower impedance values. This trend was evident in measurements taken on both conductive rubber and the human arm, providing a clear direction for optimising electrode performance through reducing hole diameter.

However, the impact of hole spacing appears to be more nuanced. While conductive rubber measurements highlighted optimal performance from electrodes with hole spacings ranging between 1.25 mm and 1.5 mm, this trend was not consistent in human arm measurements. This indicates that the relationship between hole spacing, hole diameter, and impedance is influenced by the material upon which measurements are conducted, necessitating further investigation and optimisation.

2. How Can We Enhance the Conductivity of Z-Conductive Electrodes to Reduce Their Functional Impedance?

Our study indicates that selecting the appropriate hole diameter is key to enhancing the conductivity of the electrodes. Smaller hole diameters, such as 0.28 mm and 0.4 mm, demonstrated consistent performance across different frequencies, suggesting their potential advantage in increasing the electrode's contact area with the skin. Furthermore, the enhanced retention of metal in smaller holes may also play a role in reducing impedance values.

6.2 Conclusion

Through our thorough investigation of Z-conductive electrodes, we have not only validated some of our initial hypotheses but also uncovered the complex dynamics influencing electrode performance. While significant progress has been made, many questions remain unanswered. Future research should continue to explore the optimal combination of hole diameter and spacing, considering the use of machine learning or advanced optimisation techniques to further refine electrode design. Additionally, exploring the use of different materials will be key to enhancing electrode performance. We believe that the insights gained from this study will provide a solid foundation for future research in this domain and contribute to the ongoing endeavour to optimise the design and functionality of Z-conductive electrodes.

6.3 Future Research Directions

Given the current study's findings, future research should consider several critical areas for further exploration and validation. Firstly, the inconsistent correlations between hole diameter and impedance values across different frequencies and experiments suggest that the influence of hole diameter on the performance of Z-conductive electrodes is not uniform. Therefore, future experiments should investigate this relationship further, potentially considering a wider range of hole diameters and more diverse experimental conditions.

Secondly, while smaller hole spacings were found to improve electrode performance at specific frequencies, the optimal combination of hole spacing and diameter that offers low impedance while ensuring the electrode's structural integrity for prolonged use has

yet to be identified. Future research should focus on identifying parametric optimisation of hole spacing, potentially by employing advanced optimisation techniques or machine learning algorithms.

The materials used in the electrodes' fabrication process could also be explored further. Different materials may offer varying levels of conductivity, flexibility, and durability. Understanding the trade-offs and benefits of alternative materials could lead to the development of more efficient and longer-lasting electrodes. Lastly, the current study primarily focuses on low and medium frequency ranges.

Future research should explore higher frequencies to better understand the performance of Z-conductive electrodes across wider frequency ranges.

In summation, this study has provided valuable insight into the performance dynamics of Z-conductive electrodes and has paved the way for more advanced and specified research in this domain. While perfecting the design and functionality of Z-conductive electrodes is an ongoing process, the insights gleaned from this study will undoubtedly serve as a foundation for future research endeavours.

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