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PROGRAMA DE PÓS-GRADUAÇÃO ENGENHARIA ELÉTRICA - PPGEEL

ELKIN ROMAN-GOMEZ

**BIOLOGICAL SKIN PHANTOM FOR ELECTRICAL IMPEDANCE
SPECTROSCOPY: MODELING AND VALIDATION**

JOINVILLE

2026

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**FANTOMAS BIOLÓGICOS DE PELE PARA ESPECTROSCOPIA DE IMPEDÂNCIA
ELÉTRICA : MODELAGEM E VALIDAÇÃO**

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Masters thesis presented to the Graduate Program in Electrical Engineering of the Santa Catarina State University, as requirement for obtention of the title of Master in Electrical Engineering.

Supervisor: Pedro Bertemes-Filho,
Dr

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To Joyce, for reasons she already knows.

“I would rather have questions that can’t be
answered than answers that can’t be
questioned.” Richard P. Feynman

“Life is not what one lived, but what one
remembers and how one chooses to tell it.”
Gabriel Garcia Marquez

ABSTRACT

Electrical Impedance Spectroscopy (EIS) has emerged as a promising non-invasive technique for the characterization of biological tissues and the early detection of skin abnormalities. However, the heterogeneous and multilayer structure of human skin, together with the influence of electrode geometry on current distribution, makes it difficult to evaluate the sensitivity of surface impedance measurements under controlled experimental conditions. In this context, multilayer biological phantoms provide a reproducible platform for investigating the electrical response of skin-like structures. This thesis presents the development, calibration, characterization, and experimental evaluation of multilayer skin phantoms designed to reproduce the passive electrical properties of the epidermis, dermis, and hypodermis. Calibration procedures were established to determine the appropriate combinations of gelatin, agar, sodium chloride concentration, and temperature required to achieve target tissue resistivities. The study demonstrated that gelatin provides a wider resistivity range while preserving suitable mechanical properties, whereas agar is more appropriate for highly conductive tissues. The results also showed that temperature and the intrinsic ionic content of commercial gelatin significantly affect the electrical properties of the phantoms, highlighting the importance of performing a complete calibration procedure for each material formulation. Three multilayer phantoms were subsequently fabricated by maintaining constant the electrical properties of the epidermis and hypodermis while modifying only the resistivity of the dermis. Electrical Impedance Spectroscopy measurements were performed from 200 Hz to 100 kHz using an EmStat4R potentiostat operating in tetrapolar mode. Three linear and two coplanar electrode configurations were experimentally evaluated to investigate their sensitivity to changes in dermis resistivity. The experimental results demonstrated that electrode geometry has a significant influence on the measured impedance spectra and on the sensitivity to subsurface electrical variations. Relative Difference (%RD) and Bland–Altman analyses showed that the different electrode configurations exhibit distinct sensitivity profiles, with Linear Arrangement 1 providing the greatest discrimination between the proposed phantoms. The combination of spectral and statistical analyses allowed the systematic evaluation of the response of each electrode configuration under controlled conditions. The proposed methodology provides a reproducible and low-cost experimental platform for the development, validation, and optimization of bioimpedance systems intended for non-invasive skin assessment. The developed phantoms constitute a useful tool for future studies involving electrode design, sensitivity analysis, and biomedical applications of Electrical Impedance Spectroscopy.

Keywords: Biological-Skin-Phantoms. Electrical-Impedance-Spectroscopy. Multilayer-Tissue-Models. Tissue-Characterization. Electrode-Configuration.

RESUMO

A Espectroscopia de Impedância Elétrica (EIS) tem se destacado como uma técnica não invasiva promissora para a caracterização de tecidos biológicos e a detecção precoce de alterações cutâneas. Entretanto, a estrutura heterogênea e multicamada da pele humana, associada à influência da geometria dos eletrodos sobre a distribuição da corrente elétrica, dificulta a avaliação da sensibilidade das medições realizadas na superfície em condições experimentais controladas. Nesse contexto, fantasmas biológicos multicamadas representam uma alternativa reprodutível para o estudo da resposta elétrica de estruturas semelhantes à pele. Esta dissertação apresenta o desenvolvimento, a calibração, a caracterização e a avaliação experimental de fantasmas biológicos multicamadas projetados para reproduzir as propriedades elétricas passivas da epiderme, derme e hipoderme. Foram estabelecidos procedimentos de calibração para determinar as combinações adequadas de gelatina, ágar, concentração de cloreto de sódio e temperatura necessárias para reproduzir as resistividades desejadas dos tecidos biológicos. Os resultados demonstraram que a gelatina permite reproduzir uma faixa mais ampla de resistividades, mantendo propriedades mecânicas adequadas, enquanto o ágar apresenta melhor desempenho para materiais de elevada condutividade elétrica. Também foi verificado que a temperatura e o conteúdo iônico intrínseco das diferentes marcas comerciais de gelatina influenciam significativamente as propriedades elétricas dos fantasmas, tornando indispensável a realização de um procedimento completo de calibração para cada formulação utilizada. Posteriormente, foram construídos três fantasmas multicamadas mantendo constantes as propriedades elétricas da epiderme e da hipoderme e modificando exclusivamente a resistividade da derme. As medições de Espectroscopia de Impedância Elétrica foram realizadas entre 200 Hz e 100 kHz utilizando um potenciostato EmStat4R operando em configuração tetrapolar. Foram avaliadas experimentalmente três configurações lineares e duas coplanares de eletrodos para investigar sua sensibilidade às variações de resistividade da derme. Os resultados experimentais demonstraram que a geometria dos eletrodos influencia significativamente os espectros de impedância e a sensibilidade às variações elétricas em camadas internas. As análises de Diferença Relativa (%RD) e Bland–Altman evidenciaram que cada configuração apresenta um perfil de sensibilidade distinto, destacando-se a configuração Linear Arrangement 1 como a mais sensível para discriminar as variações de resistividade propostas. A combinação das análises espectrais e estatísticas permitiu avaliar de forma sistemática o desempenho de cada configuração de eletrodos em condições experimentais controladas. A metodologia proposta fornece uma plataforma experimental reprodutível e de baixo custo para o desenvolvimento, validação e otimização de sistemas de bioimpedância destinados à avaliação não invasiva da pele. Os fantasmas desenvolvidos constituem uma ferramenta útil para futuros estudos envolvendo projeto de eletrodos, análise de sensibilidade e aplicações biomédicas da Espectroscopia de Impedância Elétrica.

Palavras-chave: Fantasmas-Biológicos-de-Pele. Espectroscopia-de-Impedância-Elétrica. Modelos-

de-Tecidos-Multicamadas. Caracterização-de-Tecidos. Configuração - de-Eletrodos.

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LIST OF ABBREVIATIONS AND ACRONYMS

AC	Alternating Current
AI	Artificial Intelligence
BCC	Basal Cell Carcinoma
EIS	Electrical Impedance Spectroscopy
EIT	Electrical Impedance Tomography
HF	High Frequency
LF	Low Frequency
MF	Medium Frequency
MF-EIT	Multi-Frequency Electrical Impedance Tomography
SCC	Squamous Cell Carcinoma
VHF	Very High Frequency
VLF	Very Low Frequency

LIST OF SYMBOLS

$\%$	Percent
$^{\circ}\text{C}$	Degree Celsius
Ω	Ohm
ε	Permittivity
ρ	Resistivity
σ	Conductivity
C	Coulomb
C_m	Membrane Capacitance
f_c	Characteristic Frequency
F	Farad
m	Meter
R_e	Extracellular Resistance
R_i	Intracellular Resistance
R_m	Membrane Resistance
s	Second
S	Siemens
V	Volt
X_c	Capacitive Reactance

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1 INTRODUCTION

The skin, the largest organ of the human body, performs multiple functions essential for homeostasis. Its primary role is to act as a protective barrier against external physical, chemical, and biological agents (SWANN, 2010)(ELIAS; STEINHOFF, 2013). To accomplish these functions, the skin is composed of three main layers: the epidermis, the dermis, and the subcutaneous cellular tissue. These layers also contain several specialized structures, including cutaneous appendages such as hair follicles, sweat glands, sebaceous glands, blood vessels, lymphatic vessels, nerves, and specialized neural structures (CALONJE, 2012). Due to its continuous exposure to harmful agents, many of which are associated with skin cancer (STERN, 2002), significant interest has emerged in the development of early diagnostic methods, since early detection has been shown to substantially reduce mortality rates (WATTS et al., 2021)(YOUUL et al., 2010)(GORDON et al., 2019). Consequently, considerable effort has been directed toward the development of minimally invasive tools for the early detection of skin pathologies and their integration into routine dermatological practice. These technologies can reduce the need for skin biopsies, decrease false-positive rates, and provide immediate diagnostic results (JUNG et al., 2021)(MCMULLAN, 2025).

Electrical Impedance Spectroscopy (EIS) is a non-invasive method with the potential to detect and confirm premalignant or malignant skin lesions, as well as early epigenetic alterations associated with the field of carcinogenesis (BRAUN et al., 2017) (GLADSTEIN et al., 2018) (LIEBICH et al., 2022). EIS measurements in healthy skin and in skin with malignant lesions, such as Basal Cell Carcinoma (BCC), Squamous Cell Carcinoma (SCC), and melanoma, are considered to differ due to changes in cell size, shape, and orientation, as well as alterations in the structure and functional state of cell membranes(BRAUN et al., 2017)(MOHR et al., 2013) . These changes alter the conductivity and capacitance of the tissue, allowing EIS to non-invasively detect differences between healthy skin and malignant lesions, which can be quantified based on the passive electrical properties of the tissues (BRAUN et al., 2017)(LIEBICH et al., 2022) (FAES et al., 1999).

Biological phantoms are artificial structures developed to reproduce the properties of living tissues (MIRBEIK-SABZEVAR; TAVASSOLIAN, 2019). In the medical field, they are essential for the evaluation, calibration, and analysis of medical devices, as well as for educational and research activities in academic settings (KAVITHA; REDDY, 2012). These models are commonly used to simulate biological tissues and may be fabricated in different physical forms, including liquid, solid, and semi-solid materials (MIRBEIK-SABZEVAR; TAVASSOLIAN, 2019). According to Kavitha and Reddy, phantoms are nonliving objects intentionally designed to emulate the properties of biological tissues, incorporating controlled variations capable of representing both healthy and pathological conditions (KAVITHA; REDDY, 2012).

Despite the availability of commercial phantoms, the growing need for application-specific models and the high associated costs have encouraged the development of low-cost

alternatives adapted to particular experimental requirements (MADSEN, 1978) (BURLEW, 1980)(KAVITHA; REDDY, 2012) . Consequently, phantoms intended to simulate different tissues and organs have been widely explored in several areas, including optical spectroscopy, biomedical imaging, dosimetry, and ballistic research, among others (POGUE; PATTERSON, 2006).

The application of EIS to skin analysis is directly associated with the heterogeneous and multilayered structure of the tissue. Each skin layer presents distinct passive electrical properties determined by its water content, cellular organization, extracellular matrix composition, and membrane characteristics. (MONTAGNA; KLIGMAN; CARLISLE, 1992) (MCKNEE, 1989) As a result, the electrical response measured at the skin surface corresponds to the combined contribution of the epidermis, dermis, and subcutaneous tissue, as well as other embedded structures such as blood vessels and glands (FAES et al., 1999). Variations in these structural and electrical characteristics can modify the impedance spectrum, enabling the identification of physiological and pathological tissue alterations (ROSENDAL, 1945) (SALTER, 1979).

The capability of EIS to detect these variations depends on the measurement configuration and on the distribution of the injected Alternating Electrical Current (AC) within the tissue (MARTINSEN; HEISKANEN, 2023). Parameters such as electrode geometry, inter-electrode configuration, electrode spacing, and measurement frequency influence the current penetration depth and the sensitivity of the technique to specific tissue regions (ABASI et al., 2022). Consequently, different electrode configurations may produce distinct responses when evaluating variations occurring at different depths within the skin (GRIMNES; MARTINSEN, 2006). Understanding these interactions is essential for improving the sensitivity and specificity of surface bioimpedance measurements in dermatological applications (MOHR et al., 2013).

However, direct experimental evaluation of these factors in human subjects presents several limitations, including biological variability, ethical considerations, and the difficulty of controlling tissue properties under reproducible conditions (ROMAN-GOMEZ; BERTEMES-FILHO, 2026) (FAES et al., 1999) (PRESTI; TEDESCO; SANTORO, 2023). In this context, biological phantoms represent an important alternative for the controlled investigation of bioimpedance measurements (MARCHAL; NADI; TOSSER, 1989) (CHEN et al., 2016). By reproducing the structural and passive electrical properties of biological tissues, phantoms enable systematic studies of current distribution, electrode sensitivity, and impedance variations under predefined experimental conditions (MARTINSEN; HEISKANEN, 2023).

This thesis begins with a historical overview of the study of electrical properties in biological media and the development of biological skin phantoms for experimental bioimpedance applications. Subsequently, the theoretical foundations of bioelectrical impedance in multilayer biological structures are presented, including the principles of Electrical Impedance Spectroscopy (EIS), tetrapolar measurement techniques, surface electrodes, and the main sources of measurement error. The electrical behavior of biological tissues is analyzed in terms of con-

ductivity, permittivity, frequency-dependent response, and dispersion phenomena. In addition, the distribution of electrical current in multilayer structures is discussed, emphasizing current penetration depth, electrical interaction between layers, sensitivity to internal variations, and the influence of electrode geometry and spacing on the measurement response and depth sensitivity.

The thesis also addresses the development and characterization of multilayer biological skin phantoms designed to reproduce the structural and passive electrical properties of the epidermis, dermis, and hypodermis. Different tissue-mimicking materials and fabrication protocols based on gelatine, agar, and saline solutions are presented, together with procedures for controlling layer thickness, structural stability, and reproducibility. Experimental characterization using EIS is subsequently described, including the detection of internal electrical variations and the evaluation of different electrode configurations. Finally, methods for bioimpedance signal processing, statistical analysis, and Artificial Intelligence (AI) are introduced for the extraction and classification of spectral features associated with phantom composition and internal structures. The obtained results are discussed considering the limitations of the experimental model and the potential application of bioimpedance-based techniques for non-invasive skin assessment.

1.1 MOTIVATION

Currently, clinical tools based on Electrical Impedance Spectroscopy (EIS) are available to support the early diagnosis of skin cancer, particularly melanoma. Systems such as Nevisense[®] have demonstrated consistent results in clinical studies reported in the literature (OLLMAR; GRANT, 2016). However, their application in research and laboratory environments remains limited due to the high cost of the disposable electrodes and specialized materials they require (ROMAN-GOMEZ; BERTEMES-FILHO, 2026). Furthermore, the proprietary nature of these systems restricts access to fundamental aspects such as electrode configurations, measurement protocols, and data processing procedures. This lack of access hinders the reproduction of results, limits the development of new methodologies, and reduces the opportunities for independent experimental validation. Likewise, it restricts their use as educational tools in academic training processes, where understanding each stage of the measurement system is essential.

In this context, the development of multilayer biological phantoms and EIS-based measurement methodologies provides an alternative for the controlled evaluation of electrode configurations, measurement protocols, and sensitivity to internal electrical variations. These models offer a reproducible platform in which the thickness and electrical properties of each layer can be systematically adjusted, enabling the investigation of the relationship between tissue structure and the measured spectroscopic response. This capability is particularly relevant for assessing the sensitivity of tetrapolar electrode configurations to localized changes occurring at different depths within the skin, as well as for improving the understanding of the mechanisms

governing current distribution in multilayer tissues.

Furthermore, the availability of low-cost and reproducible skin phantoms can facilitate the development and validation of novel instrumentation, reconstruction algorithms, and diagnostic methodologies without the ethical and practical constraints associated with in vivo experimentation. By providing a controlled environment for experimental studies, these models contribute to a deeper understanding of bioimpedance measurements in multilayer tissues and support the development of accessible technologies for non-invasive applications. Therefore, the construction of biological phantoms that accurately reproduce the passive electrical properties of the epidermis, dermis, and hypodermis represents an important step toward advancing skin bioimpedance research and expanding its clinical and technological applications.

1.2 OBJECTIVES

1.2.1 General Objective

Develop and validate multilayer biological phantoms with structural characteristics similar to human skin for surface electrical impedance spectroscopy studies aimed at detecting electrical variations in internal layers using different electrode configurations.

1.2.2 Specific Objectives

- a) Design and fabricate multilayer biological phantoms that reproduce the layer thickness and passive electrical properties of human skin using materials capable of simulating these characteristics in a controlled manner.
- b) Electrically characterize the developed phantoms through electrical impedance spectroscopy measurements over a wide frequency range, evaluating the stability and reproducibility of their electrical properties.
- c) Implement and compare different electrode configurations for tetrapolar surface bioimpedance measurements, analyzing their sensitivity to variations in resistivity and thickness of internal layers.
- d) Evaluate the capability of different electrode configurations to detect variations in layer thickness and passive electrical properties at different depths within multilayer phantom structures.
- e) Experimentally validate the relationship between structural and electrical variations in internal layers and the response obtained through surface electrical impedance spectroscopy.

1.3 RESEARCH CONTEXT

The study of electrical phenomena in biological systems constitutes one of the fundamental pillars underlying the development of electrical bioimpedance and numerous modern

techniques for tissue characterization (SÁNCHEZ-IGLESIAS; FERNÁNDEZ-LUCAS; TERUEL, 2012). Current understanding of the electrical properties of biological tissues is the result of a historical progression that integrates contributions from physics, physiology, electrochemistry, and electronic instrumentation. Since the late eighteenth century, researchers have observed that living organisms exhibit intrinsic electrical behavior, establishing the conceptual foundations of bioelectricity (MARTINSEN; HEISKANEN, 2023).

Electricity can be defined as the set of phenomena associated with the interaction of electric charges. Electric charge is a fundamental property of matter related to the interaction forces described by Coulombs law and may exist in two polarities: positive and negative. The presence of a charge distribution gives rise to an electric field, which is associated with an electric potential difference, or voltage, defined as the work required to move a charge from one point to another within that field. The motion of electric charges generates an electric current, whose flow depends on the presence of conductive materials. The opposition that a material presents to the flow of current is known as electrical resistance, whereas its ability to conduct current is referred to as electrical conductance. Together, these concepts provide the physical foundation necessary to understand the electrical behavior of biological tissues and their interaction with externally applied electrical currents.

The first experimental evidence related to bioelectricity emerged in the late eighteenth century through the work of Luigi Galvani. Around 1780, Galvani observed that muscular contractions occurred when contact was established between two different metals and the muscle of a frog, a phenomenon that he associated with the presence of an electric current (ONARAL; SUN; SCHWAN, 1984). According to another historical account, these observations occurred when an assistant accidentally touched the sciatic nerve of a frog with a metallic scalpel while an electrostatic machine was generating electrical discharges, causing involuntary contraction of the animals muscle (GRIMNES; MARTINSEN, 2008). These experiments marked the beginning of the study of so-called animal electricity and provided the first evidence suggesting that living organisms possess intrinsic electrical properties (LANDIS; DEWICK, 1929).

Galvanis findings attracted considerable scientific attention and initiated an intense debate regarding the origin of the electrical phenomena observed in biological tissues. While Galvani argued that electricity originated within living organisms, Alessandro Volta maintained that the current was generated by the contact between dissimilar metals. This controversy stimulated important advances in electrochemistry and ultimately led to the development of the voltaic pile in 1800, recognized as the first stable source of direct current. Voltas invention represented a major milestone in experimental electricity and provided essential tools for the subsequent development of bioelectricity research (MARTINSEN; HEISKANEN, 2023).

Between the late eighteenth century and throughout the nineteenth century, interest in the electrical properties of the human body increased considerably. In 1786, Bertholon published the treatise *L'électricité du Corps Humain dans l'État de Santé et de Maladie* (Electricity of the Human Body in Health and Disease), in which he described various electrical phenom-

ena associated with the human organism and discussed their relationship to health and disease. Later, in 1861, Meissner reported a series of observations on the electrical properties of the skin in the journal *Zeitschrift für rationelle Medicin* (LANDIS; DEWICK, 1929). Collectively, these studies reflected the growing scientific interest in understanding the electrical behavior of biological tissues and their potential association with different physiological and pathological states.

The development of bioelectricity was closely linked to advances in electronic instrumentation, as the detection of low-intensity bioelectric currents required devices with increasingly greater measurement capabilities. Among the most significant developments was the invention of the galvanometer in the early nineteenth century, which enabled the detection and quantification of small electrical currents. Subsequently, high-voltage generators, alternating-current systems, and new direct-current sources capable of providing more stable and controlled electrical signals were developed. These advances, together with Faradays contributions to the study of electrochemical phenomena, promoted the transition of bioelectricity from predominantly qualitative observations to quantitative and reproducible measurements.

The consolidation of electromagnetic theory through the work of Maxwell during the nineteenth century provided the theoretical framework necessary to describe electrical phenomena in physical and biological systems. During the 1880s, Oliver Heaviside introduced concepts such as *impedance*, *conductance*, *admittance*, *permittivity*, and *permeability*, as well as the term *permittance*, which later evolved into the concept of *susceptance* (GRIMNES; MARTINSEN, 2008). These concepts enabled the description of the electrical behavior of complex materials through parameters that account for both electrical conduction and energy storage processes.

In this context, the concept of electrical impedance extended the classical notion of resistance by incorporating frequency-dependent phenomena associated with capacitive and inductive processes. This formulation proved particularly relevant for the study of biological tissues, whose electrical behavior varies with frequency due to their heterogeneous structure and the presence of cell membranes. As a result, impedance became one of the fundamental parameters for the electrical characterization of biological systems (FRICKE, 1924) (PHILLIPPSON, 1921).

At the beginning of the twentieth century, the first systematic investigations aimed at electrically characterizing biological tissues through frequency-dependent analysis were initiated. In 1921, Phillipson performed one of the earliest measurements of tissue impedance as a function of frequency and observed that capacitance varied approximately as the inverse square root of the applied frequency (PHILLIPPSON, 1921). This finding demonstrated that biological tissues exhibit a complex and frequency-dependent electrical behavior, differing substantially from the purely resistive models employed at that time (COLE, 1928).

Subsequently, Fricke proposed in 1924 one of the first equivalent electrical models to represent the behavior of tissues and blood (FRICKE, 1924). In this model, cells were represented by a resistance associated with the intracellular fluid, a capacitance corresponding to the

cell membrane, and a resistance representing the extracellular medium. This approach enabled the interpretation of tissue electrical properties from a physiological and structural perspective, establishing the foundation for the equivalent circuit models later adopted in bioimpedance research.

The dielectric characterization of tissues led to the identification of dispersion phenomena associated with changes in conductivity and dielectric constant as the excitation frequency varies. Schwan and Ferris described dispersion as the phenomenon by which permittivity and conductivity transition from one approximately constant level to another as a function of the applied frequency (SCHWAN; FERRIS, 1968). These phenomena are associated with electrical relaxation mechanisms, which represent non-instantaneous tissue responses to an electrical stimulus. Such processes are characterized by specific time constants related to the structural and physiological properties of cells and tissues (COLE; COLE, 1941).

1.3.1 Electrical Bioimpedance

The historical development of bioelectricity established the conceptual and experimental foundations for the emergence of electrical bioimpedance as a biomedical analysis tool. As advances were made in physics, physiology, and electronic instrumentation, it became evident that biological tissues exhibit not only resistive behavior but also complex frequency-dependent electrical responses associated with their composition, structure, and cellular organization. These advances enabled the development of methods capable of non-invasively characterizing biological systems through their electrical properties (MARTINSEN; HEISKANEN, 2023).

In 1928, Kenneth Stewart Cole developed mathematical formulations to describe tissue impedance at zero and infinite frequencies and conducted studies on the permittivity and immittance of biological materials (COLE, 1928). These contributions provided a more rigorous framework for understanding the dielectric behavior of biological systems. Subsequently, Cole and Cole proposed an empirical equation capable of describing the complex permittivity observed in a wide range of biological tissues (COLE; COLE, 1941). The resulting ColeCole model remains one of the most widely used approaches in bioimpedance spectroscopy due to its ability to represent electrical relaxation and dispersion phenomena.

The contributions of Herman P. Schwan during the 1950s had a significant impact on the development of electrical biophysics. Schwan described frequency-domain measurement techniques employing ultra-high-frequency (UHF) signals and contributed substantially to the understanding of the dielectric properties of biological tissues and the mechanisms governing the interaction between electric fields and cellular structures (SCHWAN, 1955). His work helped establish spectroscopic methods as fundamental tools for the electrical characterization of biological tissues.

Beginning in the 1960s, direct clinical applications of electrical bioimpedance started to emerge. In 1962, Thomasset conducted pioneering studies using bioimpedance measurements as an index for estimating total body water through the use of subcutaneously inserted needle

electrodes (THOMASSET, 1962). Subsequently, Nyboer introduced the bioimpedance analysis technique employing four surface electrodes in 1970 (NYBOER, 1970). The tetrapolar configuration minimized the influence of electrodeskin contact impedance on electrical measurements. However, one of the limitations of surface electrodes was the high impedance of the skin, which required the use of relatively high currents to ensure stable current injection (BOULIER, 1990). Despite these limitations, the four-electrode configuration became one of the most widely adopted methods in bioimpedance analysis due to its simplicity, reproducibility, and ability to reduce errors associated with the electrode-tissue interface.

In this context, electrical bioimpedance emerged as a technique based on the interaction between electrical currents and biological tissues, enabling the assessment of dielectric and physiological properties through the analysis of the opposition that tissues present to current flow. Since then, the technique has evolved considerably and has led to numerous applications in biomedical engineering and health sciences, including tissue characterization, body composition assessment, nutrition, physiological monitoring, and the detection of pathological alterations in nephrology, oncology care, and the early diagnosis and confirmation of cancers of the cervix, breast, and skin. As a result, electrical bioimpedance has become an important tool for the study and functional analysis of biological systems (HALTER et al., 2008) (MIRANDA-ALATRISTE et al., 2022) (ZINK et al., 2015) (GONZÁLEZ-CORREA, 2018) (OLARTE-ECHEVERRI et al., 2010) (SVOBODA et al., 2019).

1.3.2 Electrical Properties of Biological Media

Biological tissues have the ability to conduct electric current. In general, this property is characterized by resistivity and permittivity; however, at relatively low frequencies, resistivity is the predominant parameter (PLONSEY; HEPPNER, 1967). Resistivity is defined as the real part of the complex impedance per unit length and cross-sectional area. In biological tissues, this parameter depends on the applied excitation frequency (GABRIEL; GABRIEL; CORTHOUT, 1996).

When an alternating current is applied during a bioimpedance measurement, it is necessary to consider the frequency-dependent behavior of cell membranes. At low and intermediate frequencies, the plasma membrane, which is primarily composed of a lipid bilayer, acts as a barrier to ionic conduction, limiting charge flow into the intracellular compartment. Consequently, the current flows predominantly through the extracellular space. At higher frequencies, the capacitive properties of the membrane reduce its opposition to current flow, allowing the electric field to penetrate more effectively into the interior of the cells (NOVELETTO et al., 2016) (ABE; NISHIZAWA, 2021b).

Under low- and intermediate-frequency conditions, electrical charges accumulate on both sides of the cell membrane, leading to polarization of the intracellular compartment. This phenomenon, known as electrical capacitance, describes the ability to store electrical charge and is quantified in farads (F). As a consequence, the opposition to ion movement associated

with cell membranes varies as a function of frequency (GONZALEZ-CORREA et al., 2003). However, it is important to consider that the cell membrane contains ion channels which, when open, reduce its charge storage capacity and therefore its opposition to current flow. Under these conditions, the capacitive reactance (X_c) decreases. When cell death occurs and the membranes lyse, capacitive reactance disappears (LIEBICH et al., 2021).

The set of bioimpedance measurements obtained at different frequencies constitutes the Electrical Bioimpedance Spectrum. This spectrum makes it possible to investigate various phenomena associated with the electrical properties of biological tissues, which can be analyzed through the concept of dispersion based on the dielectric relaxation theory of materials (ABASI et al., 2022).

In summary, biological tissues exhibit both resistive (R) and reactive (X_c) components. The resistance encountered by the electric current depends mainly on the volume, composition, and distribution of the intra and extracellular fluids, expressed in terms of ionic concentration, as well as on the structural arrangement of the cells and their functional state, which influences membrane permeability through the activation of ion channels (ÅBERG, 2004) (ALVERO et al., 2011). An example of this behavior is observed in inflammatory processes, where increased cellular metabolism and vascular permeability lead to a reduction in both the electrical resistance and reactance of the affected tissue (BROWN et al., 2005) (GONZÁLEZ-CORREA, 2018).

1.3.3 Biological Phantoms

Biological phantoms are artificial models designed to reproduce specific characteristics of biological tissues. Their use can be traced back to the late nineteenth century, when concerns regarding the harmful effects of ionizing radiation limited the availability of human volunteers for experiments involving X-ray exposure. As a result, phantoms emerged as a practical alternative for conducting controlled and reproducible studies without exposing individuals to unnecessary risks (DEWERD; KISSICK, 2014).

The design and composition of phantoms are closely related to their intended application. Early phantom models were primarily constructed from water and wax. Between the 1930s and 1970s, wood was widely employed as a phantom material, and during the 1960s, models incorporating sections of real human skeletal bone were proposed. As the use of phantoms became increasingly common, greater emphasis was placed on improving quality and reproducibility, motivating the adoption of alternative materials such as epoxy and polyurethane resins (DEWERD; KISSICK, 2014).

In the medical field, phantoms play an important role in the evaluation, analysis, and calibration of medical equipment, as well as in educational and research activities conducted in academic environments. They are widely used to simulate biological tissues and can serve as tissue substitutes in experimental studies. Depending on the intended application, phantoms may be manufactured in liquid, solid, or semisolid forms (MIRBEIK-SABZEVARI; TAVAS-SOLIAN, 2019). According to Kavitha and Reddy, phantoms are inanimate objects specifically

designed to reproduce the properties of biological tissues, incorporating variations that allow the representation of both normal and abnormal conditions (KAVITHA; REDDY, 2012).

Although commercial phantoms are available, the increasing demand for application-specific models and the associated costs have stimulated the development of low-cost alternatives tailored to particular experimental requirements. Consequently, phantoms that emulate different tissues and organs have been extensively investigated in a variety of fields, including optical spectroscopy, biomedical imaging, dosimetry, ballistic studies, and other biomedical applications (POGUE; PATTERSON, 2006).

Specific phantoms have been developed for applications involving therapeutic radiation, oncology, brachytherapy, X-ray imaging, computed tomography, mammography, ultrasonography, magnetic resonance imaging, and nuclear medicine (DEWERD; KISSICK, 2014). Phantoms have also been widely employed in studies involving Electrical Impedance Spectroscopy (EIS). Examples include the evaluation of chicken breast tissue as a basis for Multifrequency Electrical Impedance Tomography (MF-EIT) phantoms, the characterization of the electrical properties of vegetables for the development of MF-EIT phantoms (BERA; NAGARAJU; LUBINEAU, 2016), and the assessment of gelatin-based phantoms for the representation of human skin (PINTO; BERTEMES-FILHO; PATERNO, 2015).

Among the different phantom categories, liquid phantoms generally present the simplest manufacturing process, since their fabrication does not require complex preparation procedures (MIRBEIK-SABZEVAR; TAVASSOLIAN, 2019). One of the most widely adopted approaches is the use of gelatin-based phantoms to reproduce the dielectric properties of biological tissues. These phantoms are attractive because they can be produced at low cost using simple materials, primarily water and gelatin (MARCHAL; NADI; TOSSER, 1989). Furthermore, gelatin mixtures can be cast into molds, enabling the fabrication of structures with different geometries according to the tissue or organ being simulated, while maintaining relatively simple handling and preservation characteristics.

1.3.4 Biological Skin Phantoms

Biological phantoms provide anthropomorphic test domains with known structures and controlled dielectric properties that approximate the electrical characteristics of biological tissues. In the case of skin, these models constitute an important tool for the experimental evaluation of the interaction between electromagnetic fields and the human body, particularly in applications involving tissue characterization, ultrasound, microwave, and millimeter-wave technologies (MIRBEIK-SABZEVAR; TAVASSOLIAN, 2019).

From this perspective, the reproduction of the passive electrical properties of human skin in controlled laboratory environments through the use of phantoms expands the possibilities for experimental studies and educational activities at reduced cost (MARCHAL; NADI; TOSSER, 1989) (CHEN et al., 2016). Several methodologies have been proposed to reproduce the electromagnetic response of skin using multilayer biological phantoms fabricated from ma-

terials such as gelatin and agar (PINTO; BERTEMES-FILHO; PATERNO, 2015) (MARTINSEN; HEISKANEN, 2023) (KAVITHA; REDDY, 2012) (MADSEN, 1978) (BURLEW, 1980) (HANANI, 2016). These models enable the standardization of tissue-equivalent structures, facilitate the investigation of localized alterations such as granuloma formation, and provide a platform for the development of systems aimed at the early detection of pathological conditions (DUTRA, 2018).

In addition to their role as tissue substitutes, biological phantoms allow the controlled evaluation of the interaction between electromagnetic fields and multilayer structures, particularly in radiofrequency applications. To achieve this, it is essential to reproduce physical parameters such as electrical conductivity, thermal conductivity, density, and specific heat, as these properties determine the thermal and electromagnetic responses induced in biological tissues. Among these parameters, electrical conductivity plays a particularly important role in energy conversion processes, directly influencing the validity of preclinical cancer studies and the prediction of tissue behavior during skin treatments (MIRBEIK-SABZEVARI; TAVASSOLIAN, 2019).

Skin-equivalent phantoms have also proven to be valuable tools for the development and validation of technologies operating in the microwave and millimeter-wave frequency ranges, particularly in imaging applications and the detection of pathologies such as skin cancer. The dielectric characterization of these models over broad frequency ranges, together with the temporal stability of their electrical properties, enables the establishment of reproducible experimental conditions comparable to those observed in real biological tissues (MIRBEIK-SABZEVARI; TAVASSOLIAN, 2019). Furthermore, it has been demonstrated that the suitability of a phantom depends not only on the operating frequency range but also on its geometry, which should approximate the dimensions of the target tissue under the specific experimental conditions being considered (KRANOLD et al., 2020).

Based on these criteria, the construction of multilayer skin phantoms is generally guided by the structural and electrical information reported for the principal layers of human skin (FAES et al., 1999). In particular, these models seek to reproduce the epidermis, dermis, and hypodermis, which exhibit distinct geometrical characteristics and passive electrical properties. Each layer can be designed according to the ranges of thickness and resistivity reported in the literature, preserving the relative order and proportions observed in biological tissue. Nevertheless, for modeling and analytical purposes, phantoms are often considered homogeneous, isotropic, and semi-infinite media, thereby simplifying the representation of electromagnetic interactions within the tissue-equivalent structure.

2 BASIC ELEMENTS IN IMPEDANCE OF BIOLOGICAL MATERIALS

2.1 PRINCIPLES OF IMPEDANCE MEASUREMENTS

All materials, regardless of their physical state or classification, can be broadly categorized as conductors, insulators (dielectrics), or semiconductors. In insulating materials, such as rubber or glass, electrons remain strongly bound to specific atoms and therefore exhibit limited mobility. In contrast, metallic conductors contain one or more electrons per atom that are free to move throughout the material, enabling the transport of electrical charge (OHRING, 1995).

Electrical conductivity (σ) describes the ability of a material to conduct an electric current when subjected to an electric potential difference. It is defined as:

$$\sigma = n \cdot e \cdot \mu_n + p \cdot e \cdot \mu_p \quad (1)$$

where σ is the electrical conductivity ([S/m], S ($=\Omega^{-1}$) is given in siemens; n is the concentration of free electrons in the material [m^{-3}]; p is the concentration of free positive charge carriers, known as holes [m^{-3}]; e is the elementary charge [1.6022×10^{-19} C]; and μ_n and μ_p denote the mobilities of free electrons and holes, respectively [$\text{m}^2 \text{V}^{-1} \text{s}^{-1}$].

Electrical resistivity (ρ) describes the extent to which a material opposes the flow of electric current. As such, resistivity is the inverse of electrical conductivity and is expressed in $\Omega \cdot \text{m}$, according to:

$$\rho = \frac{1}{\sigma} = \frac{1}{n \cdot e \cdot \mu_n + p \cdot e \cdot \mu_p} \quad (2)$$

These properties depend both on the intrinsic characteristics of the material, such as the type of chemical bonding, crystal structure, phase composition, and the presence of structural defects, and on external factors including temperature, the frequency of the applied voltage, and the duration of exposure to the electric field, all of which influence the response of induced or permanent dipoles.

Materials exhibiting low electrical conductivity are referred to as insulators or dielectrics. The term *dielectric* was coined by William Whewell in response to a request from Michael Faraday, who during the 1830s experimentally observed that the capacitance of a capacitor changed when a material was introduced between its plates. In this context, Faraday introduced the concept of *specific inductive capacity*, defined as the ratio between the capacitance of an empty capacitor and that of a capacitor containing a dielectric material. This quantity is currently known as permittivity (ϵ) (BARNES; GREENEBAUM, 2006).

A dielectric can also be defined as a medium in which an electric field can be generated and maintained with little or no energy supplied from external sources. The energy required to establish such a field can be recovered partially or completely when the electric field is removed

(JOSSINET; SCHMITT, 1999). In an ideal dielectric, all the energy used to establish the electric field is recovered once the applied excitation is withdrawn. Consequently, a perfect dielectric would exhibit zero electrical conductivity (LIDE, 2006).

When an electric charge distribution exists, an electric field is generated, which can be understood as the interaction at a distance between electric charges. Associated with this field is an electric potential energy, or voltage, defined as the work required to move a charge from one point to another within the electric field. Voltage can also be interpreted as the driving force that promotes the movement of electric charges and is therefore often referred to as electromotive force. Its unit in the International System of Units is the volt (V).

The movement of electric charges gives rise to electric current, which can be defined as the amount of charge crossing a given cross-sectional area per unit time. Its unit is the ampere (A), and it is expressed as:

$$I = \frac{dQ}{dt} \quad (3)$$

where I is the electric current, Q is the electric charge, and t represents time.

For electric current to flow, a conductive medium is required. Such materials may oppose the flow of current through a property known as electrical resistance (R), whose unit is the ohm (Ω), or facilitate current flow through electrical conductance (G), whose unit is the siemens (S). These quantities are reciprocals of each other, and conductance can therefore be expressed as:

$$G = \frac{1}{R} \quad (4)$$

where G is the electrical conductance and R is the electrical resistance.

The electrical resistance of a material depends not only on its intrinsic properties but also on its geometry, particularly its length and cross-sectional area. The property inherent to a material that determines its ability to oppose the flow of electric current is known as electrical resistivity (ρ). Resistivity is related to the number of charge carriers, their mobility, and the electronic structure of the material (SHACKELFORD, 1995). Its SI unit is the ohm-meter ($\Omega \cdot m$).

Electrical resistance (R) can be expressed as a function of resistivity and the geometric factor of the material as:

$$R = \rho \frac{L}{A} \quad (5)$$

where ρ is the electrical resistivity, L is the length of the material, and A is the cross-sectional area through which the electric current flows.

When two conductive materials are separated by a vacuum or by an insulating material, also known as a dielectric, a capacitance (C) is established. Capacitance can be defined as the

ability to store electrical energy and corresponds to the ratio between the stored electric charge and the applied potential difference between the conductors:

$$C = \frac{Q}{V} \quad (6)$$

where Q represents the stored electric charge, V is the potential difference, and C is the capacitance expressed in farads (F).

A capacitor stores energy through the accumulation of electric charges on its plates. This phenomenon is possible because there is no conduction current through the dielectric material, meaning that no net charge transport occurs from one plate to the other through its interior.

When alternating current is applied, frequency-dependent phenomena such as capacitance and inductance become significant due to their association with energy storage mechanisms. Under these conditions, the opposition to current flow can no longer be described solely by resistance but must instead be represented by the electrical impedance (Z). Consequently, Ohm's law for alternating current can be written as:

$$Z = \frac{V}{I} \quad (7)$$

where Z is the electrical impedance (Ω), V is the applied voltage (V), and I is the electric current (A) Impedance can also be represented in polar form as:

$$Z = |Z| \angle \theta \quad (8)$$

where $|Z|$ is the impedance magnitude and θ is the phase angle between the voltage and current signals.

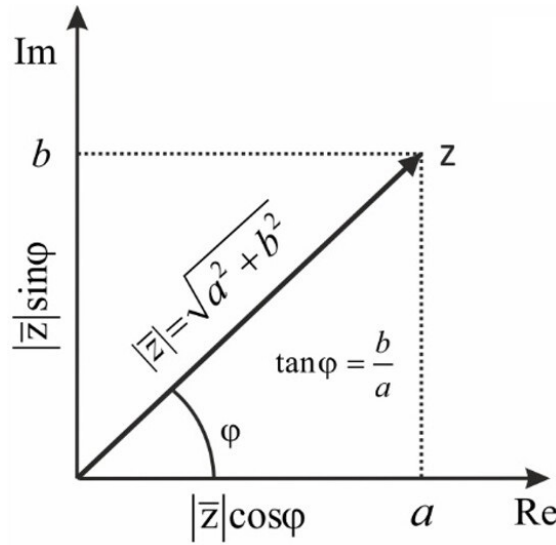
Alternatively, impedance can be expressed as a complex number:

$$Z = R + jX \quad (9)$$

where R denotes the resistance and X the reactance. Resistance is associated with dissipative processes within the material, whereas reactance represents energy storage phenomena of either capacitive or inductive nature.

In biological tissues, resistance (R) is primarily associated with ionic transport through the intra- and extracellular compartments. Reactance, on the other hand, is related to the accumulation of electric charges across cell membranes and is therefore predominantly capacitive in nature, commonly represented as capacitive reactance (X_c).

This complex representation allows the electrical behavior of tissues to be described in the complex plane, where the real axis corresponds to resistance and the imaginary axis corresponds to reactance. Furthermore, the impedance magnitude and phase angle can be obtained from the real and imaginary components according to:

Figure 1 – Representation of a complex number z .

Source: (LAZANAS; PRODRAMIDIS, 2023)

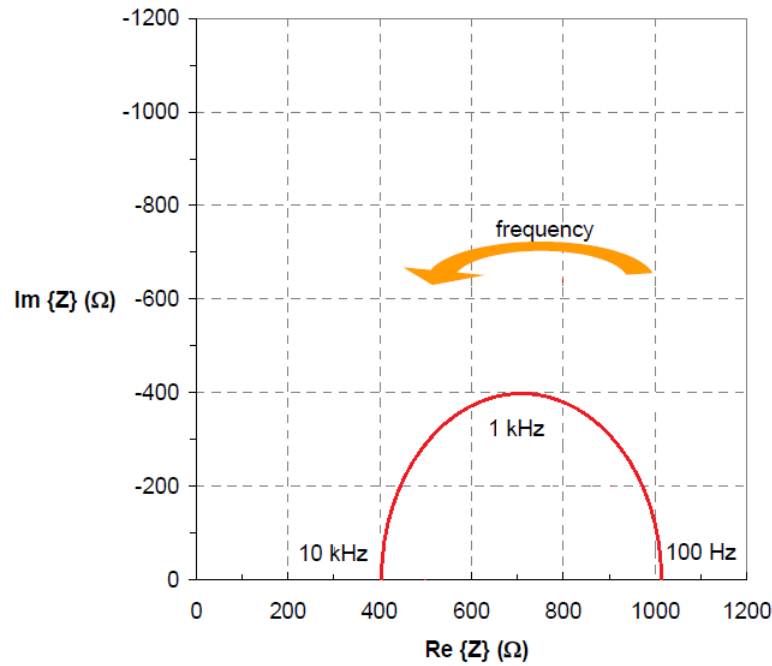
$$|Z| = \sqrt{R^2 + X^2} \quad (10)$$

$$\theta = \tan^{-1} \left(\frac{X}{R} \right) \quad (11)$$

where the magnitude is calculated using the Pythagorean theorem, while the phase angle is determined from the trigonometric relationship between reactance and resistance.

The imaginary part (with negative sign) of the impedance is plotted against the real part of the impedance for each measurement frequency. This representation, commonly known as the Nyquist plot, Wessel diagram, or Cole diagram, is widely used in electrochemistry and has been extensively adopted in bioimpedance studies. Its principal advantage lies in the fact that frequency-dependent electrical phenomena, known as dispersions, can be readily identified and visualized as characteristic arcs in the complex plane (IVORRA, 2002). Moreover, the shape, position, and dimensions of these arcs provide valuable information about the resistive and capacitive properties of the analyzed material. In biological tissues, these characteristics are directly related to physiological features such as extracellular and intracellular ionic conduction, cell membrane polarization, tissue composition, and structural organization (MARTINSEN; HEISKANEN, 2023). Consequently, the Cole diagram constitutes one of the most important graphical tools in bioimpedance analysis, enabling a comprehensive interpretation of the electrical behavior of biological systems over a broad frequency range.

Figure 2 – The Wessel diagram also called Cole diagram or Nyquist plot



Source: Adapted from (IVORRA, 2002)

2.1.1 Tetrapolar Measurements

It could be supposed that tissue impedance can be measured by a pair of electrodes attached to the surface of the sample under study. Both electrodes would be used to inject the current and to measure the voltage drop. However, the electrode–electrolyte interface impedances are in series with the sample impedance and, therefore, the measured impedance corresponds to the sum of the three impedances ($Z_x + Z_{e,1} + Z_{e,2}$). Unfortunately, these parasitic impedances are sufficiently large to disturb the measurements, especially at low frequencies. To overcome this limitation, an alternative measurement approach is commonly employed in which the current is injected through one pair of electrodes while the resulting voltage drop is sensed by a second pair. This configuration, known as the four-electrode or tetrapolar method, ideally minimizes the influence of the electrode–electrolyte interface impedance and has been used in bioimpedance measurements for more than a century (GEDDES, 1996; IVORRA, 2002).

Depending on the number of electrodes employed, different measurement topologies can be implemented. The most common configurations are the bipolar (two-electrode) and tetrapolar (four-electrode) methods. In bipolar measurements, the same pair of electrodes is used both to apply the excitation signal, either current or voltage, and to measure the resulting electrical response. In contrast, tetrapolar measurements use one pair of electrodes for current or voltage excitation and a separate pair for sensing the resulting voltage or current response, thereby reducing the influence of electrode interface effects on the measured impedance (BROWN; WILSON; BERTEMES-FILHO, 2000).

Nevertheless, surface electrodes introduce additional challenges associated with the rel-

atively high impedance of the skin-electrode interface. To ensure a stable current injection and minimize measurement errors, relatively high excitation currents and voltages have traditionally been required, particularly when the skin-electrode impedance reaches values on the order of several kilo-ohms per square centimeter (BOULIER, 1990).

Although the four-electrode configuration substantially reduces the influence of the electrode–electrolyte interface, it is not completely immune to measurement errors. Parasitic effects, such as stray capacitances between connecting wires, input impedances of the instrumentation, and residual contributions from the electrode–electrolyte impedances, may introduce inaccuracies, particularly at very low and very high frequencies (PALLÁS-ARENY; WEBSTER, 1993). As a consequence, the practical operating frequency range of the four-electrode technique is typically limited to frequencies spanning from a few hertz to several hundreds of megahertz.

Nevertheless, under specific conditions, two-electrode measurements can also provide satisfactory results. This occurs when the impedance associated with the electrode interfaces is significantly smaller than the impedance of the material or tissue under investigation. In such cases, the influence of the electrode impedance becomes negligible, especially when large-area electrodes (greater than 1 cm²) are employed and measurements are performed at frequencies above approximately 10 kHz (PALLÁS-ARENY; WEBSTER, 1993) (IVORRA, 2002).

2.1.2 Sources of Measurement Error

Although electrode polarization constitutes one of the most important sources of uncertainty in electrical impedance spectroscopy (EIS), several additional instrumental factors may also compromise measurement accuracy. These error sources originate from the electrical characteristics of the instrumentation, cables, connectors, and measurement circuitry, and their influence is strongly dependent on the excitation frequency. As a result, different types of errors tend to dominate in different frequency regions. To facilitate the discussion of these frequency-dependent effects, Table 1 presents the nomenclature commonly adopted for frequency bands as a function of the characteristic cutoff frequency (f_c) of the material under analysis.

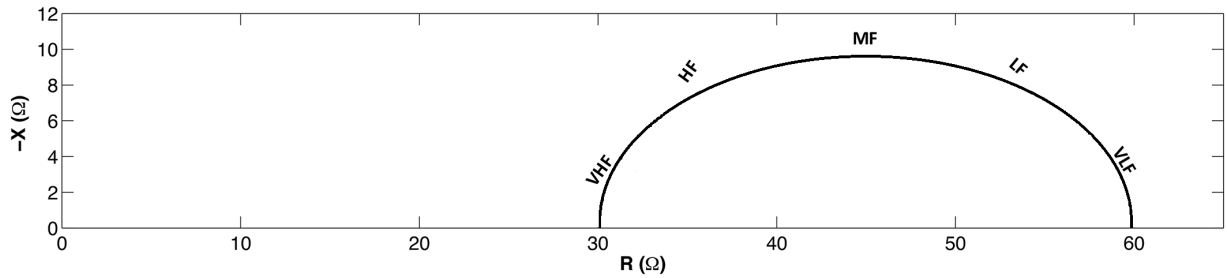
Table 1 – Nomenclature and frequency ranges used in Electrical Impedance Spectroscopy (EIS).

Abbreviation	Nomenclature	Frequency Range
VLF	Very Low Frequency	$f < \frac{f_c}{5}$
LF	Low Frequency	$\frac{f_c}{5} < f < \frac{f_c}{2}$
MF	Medium Frequency	$\frac{f_c}{2} < f < 2f_c$
HF	High Frequency	$2f_c < f < 5f_c$
VHF	Very High Frequency	$f > 5f_c$

Source: Adapted from Ayllón, Gil-Pita e Seoane (2016).

Ayllón et al. identified and classified six major types of measurement errors commonly

Figure 3 – Representation of Frequency Ranges in the Impedance Plane



Source: Prepared by the author based on (AYLLÓN; GIL-PITA; SEOANE, 2016).

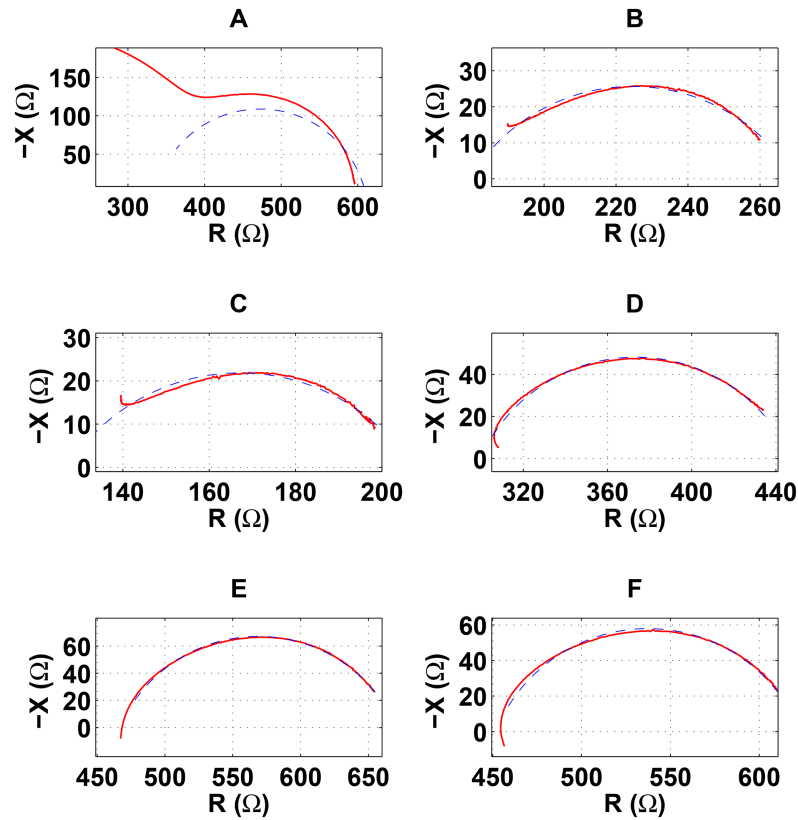
encountered in electrical impedance spectroscopy (AYLLÓN; GIL-PITA; SEOANE, 2016). The first category (Type A), often referred to as the *Hook effect*, is one of the most frequently observed artifacts. It is characterized by an increase in the magnitude of the reactance at medium and high frequencies (MF and HF), contrary to the decrease typically expected in EIS measurements (BUENDÍA, 2009).

Figure 3 presents the relationship between resistance, reactance, and frequency in the impedance plane. Error Types B and C are also characterized by an increase in the magnitude of the reactance; however, they differ in the frequency range in which they occur. Type B is predominantly observed in the very-high-frequency (VHF) region, whereas Type C appears within the high-frequency (HF) range. Type D is characterized by a reduction in the measured resistance at high frequencies. Finally, Error Types E and F are associated with an increase in reactance accompanied by a change in its sign, resulting in a transition from capacitive to inductive behavior. In Type E, this increase in reactance occurs simultaneously with a decrease in resistance, whereas in Type F, the resistance exhibits an increase within the VHF region (AYLLÓN; GIL-PITA; SEOANE, 2016).

The characteristic signatures of Error Types A, B, C, D, E, and F in the impedance plane (reactance versus resistance) are shown in Figure 4. The expected impedance response is depicted by a dashed line, while the distorted response associated with the measurement artifact is represented by a solid line. (AYLLÓN; SEOANE; GIL-PITA, 2009) (BUENDÍA, 2009)

Because measurement artifacts may significantly alter both the resistive and reactive components of the impedance spectrum, their identification is essential for ensuring data quality and preventing misinterpretation of bioimpedance measurements. The recognition of these characteristic error patterns also facilitates the diagnosis of problems related to electrodes, cabling, and instrumentation, thereby contributing to the reliability and reproducibility of impedance-based analyses (AYLLÓN; GIL-PITA; SEOANE, 2016).

Figure 4 – Instrumentation Errors Classified by Ayllón, Gil-Pita, and Seoane. (a) Type A error; (b) Type B error; (c) Type C error; (d) Type D error; (e) Type E error; and (f) Type F error.



Source: (AYLLÓN; GIL-PITA; SEOANE, 2016)

2.2 ELECTRICAL IMPEDANCE SPECTROSCOPY: BASIC CONCEPTS

Spectroscopy can be broadly defined as the study of how a material responds to an external energy stimulus as a function of frequency (BERTEMES-FILHO, 2018). Within this framework, Electrical Impedance Spectroscopy (EIS) is a measurement technique that investigates the interaction between an alternating electric field and a material over a range of excitation frequencies (BERTEMES-FILHO, 2002). In a typical EIS measurement, an alternating electrical signal is applied to the material under investigation through an electrode system, and the resulting voltage or current response is measured to determine its electrical impedance (MACDONALD; BARSOUKOV, 2005).

When applied to biological systems, this technique is commonly referred to as bio impedance spectroscopy. In this case, low-amplitude alternating currents are injected into biological tissues over a broad frequency range, and the resulting electrical response is analyzed. The measured impedance reflects the combined resistive and reactive contributions of the electrochemical and dielectric processes occurring within the tissue. Consequently, the impedance spectrum provides valuable information about tissue composition, cellular organization, membrane properties, and fluid distribution. Furthermore, the measured spectra can often be rep-

resented using equivalent electrical circuit models that describe the underlying physical and physiological mechanisms (NOURI et al., 2022; BARSOUKOV; MACDONALD, 2018).

The interpretation of bioimpedance spectra is closely related to the dielectric relaxation phenomena exhibited by biological tissues. By evaluating the impedance response over a wide frequency range, EIS enables the identification of frequency-dependent dispersion mechanisms associated with cellular membranes, extracellular and intracellular fluids, macromolecules, and water content. The resulting electrical bioimpedance spectrum therefore constitutes a powerful tool for characterizing the structural and functional properties of biological tissues (ABASI et al., 2022).

Several excitation strategies can be employed in electrical impedance spectroscopy. One approach consists of applying a voltage step and measuring the corresponding current response in the time domain. A second approach uses a broadband excitation signal, such as white noise, from which the impedance spectrum is extracted through spectral analysis. The third and most widely adopted approach involves the application of sinusoidal current or voltage signals at multiple frequencies, either sequentially or simultaneously, followed by the measurement of the corresponding electrical response (MACDONALD; BARSOUKOV, 2005).

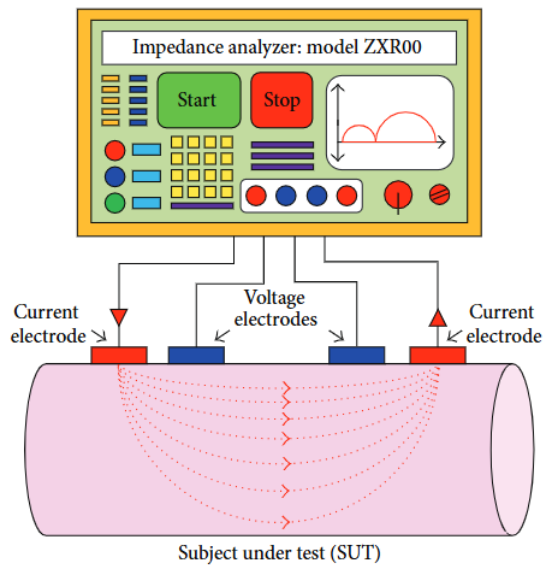
Due to its non-invasive nature and sensitivity to physiological and structural tissue properties, EIS has found numerous applications in medicine and biomedical research. Bioimpedance spectroscopy has been investigated as a diagnostic tool for the early detection of pathological changes in several organs and tissues, including the skin, cervix, esophagus, urinary bladder, and colon. In addition, Electrical Impedance Tomography (EIT), an imaging modality derived from bioimpedance principles, has been explored for applications such as breast cancer detection and functional tissue monitoring (GONZÁLEZ-CORREA, 2018).

2.2.1 Dispersion and Polarization Phenomena

The dielectric behavior of a material subjected to an alternating electric field is governed by polarization and relaxation phenomena. When an electric field is applied, free charges contribute to electrical conduction while bound charges undergo displacement or orientation, giving rise to polarization processes. Consequently, the electrical response of a material is determined by both its conductivity and permittivity, whose relative contributions depend on the excitation frequency (GABRIEL; GABRIEL; CORTHOUT, 1996) (BARSOUKOV; MACDONALD, 2018).

Polarization mechanisms do not respond instantaneously to changes in the applied electric field. Instead, each mechanism exhibits a characteristic relaxation time associated with the motion of charges, dipoles, or interfaces within the material. At low frequencies, polarization processes can follow the variations of the applied field, whereas at higher frequencies some mechanisms can no longer respond sufficiently fast. This progressive loss of polarization mechanisms produces frequency-dependent variations in conductivity and permittivity, a phenomenon known as dielectric dispersion (GABRIEL; GABRIEL; CORTHOUT, 1996) (MARTINSEN;

Figure 5 – EIS studies on biological materials with four-electrode technique using an impedance analyzer.



Source: (BERA, 2014)

HEISKANEN, 2023) (BARSOUKOV; MACDONALD, 2018).

Dielectric dispersion can be investigated through electrical impedance spectroscopy, in which alternating electrical signals are applied over a range of frequencies and the resulting impedance is measured. The acquired impedance spectrum reflects the combined conductive and dielectric contributions of the material and can be used to identify the dominant polarization and relaxation mechanisms present within the system (NOURI et al., 2022; BARSOUKOV; MACDONALD, 2018).

Based on the dominant polarization and relaxation mechanisms, the frequency-dependent electrical response of materials can be classified into different dispersion regions. In biological materials, these regions are traditionally referred to as the α , β , δ , and γ dispersions. This classification originated from the pioneering studies of Schwan and was later consolidated through extensive compilations of dielectric properties reported by Gabriel et al. (SCHWAN, 1957) (GABRIEL; GABRIEL; CORTHOUT, 1996). Each dispersion region corresponds to a characteristic relaxation process associated with specific polarization mechanisms (COLE; COLE, 1941) (MARKX; DAVEY, 1999) (MARTINSEN; HEISKANEN, 2023) (BOUNIK et al., 2022).

The α -dispersion occurs at the lowest frequencies and is primarily associated with ionic diffusion processes and electrode or interfacial polarization effects. The β -dispersion is dominated by interfacial polarization mechanisms, commonly described as Maxwell–Wagner polarization, which arise at interfaces separating regions with different electrical properties. At higher frequencies, the δ -dispersion is associated with the relaxation of bound water molecules and macromolecular structures, whereas the γ -dispersion is primarily related to the dipolar relaxation of free water molecules subjected to an alternating electric field (GABRIEL; GABRIEL; CORTHOUT, 1996) (MARKX; DAVEY, 1999) (BOUNIK et al., 2022).

2.3 ELECTRICAL IMPEDANCE IN BIOLOGICAL TISSUES

Biological tissues exhibit complex electrical behavior because they contain conductive fluids, dielectric structures, and interfaces with distinct electrical properties. As a result, both conduction and polarization phenomena contribute to the measured impedance response. The relative contribution of these mechanisms depends on the frequency of the applied electrical excitation and on the structural organization of the tissue.

2.3.1 Electrical Properties of Biological Tissues

Unlike metallic conductors, where charge transport occurs through the movement of free electrons, electrical conduction in biological tissues is predominantly ionic. The transport of charge is mediated by the displacement of ions dissolved in intracellular and extracellular fluids, including Na^+ , K^+ , Cl^- , Ca^{2+} , and other ionic species. Consequently, biological tissues behave as electrolyte conductors whose electrical response depends on both ionic concentration and mobility (GABRIEL; GABRIEL; CORTHOUT, 1996) (GRIMNES; MARTINSEN, 2008).

In addition to conductive properties, biological tissues also exhibit dielectric behavior. The ability of a tissue to store electrical energy in the presence of an electric field is described by its permittivity. Therefore, the electrical behavior of biological tissues is governed by the combined effects of conductivity and permittivity, both of which vary significantly with frequency (GABRIEL; GABRIEL; CORTHOUT, 1996).

Another important characteristic of biological tissues is anisotropy. In anisotropic tissues, the electrical properties depend on the direction of measurement. Skeletal muscle is one of the most representative examples, presenting significantly different conductivity and permittivity values along and across the muscle fibers. This directional dependence must be considered when interpreting bioimpedance measurements and developing tissue models (GABRIEL; GABRIEL; CORTHOUT, 1996) (FAES et al., 1999).

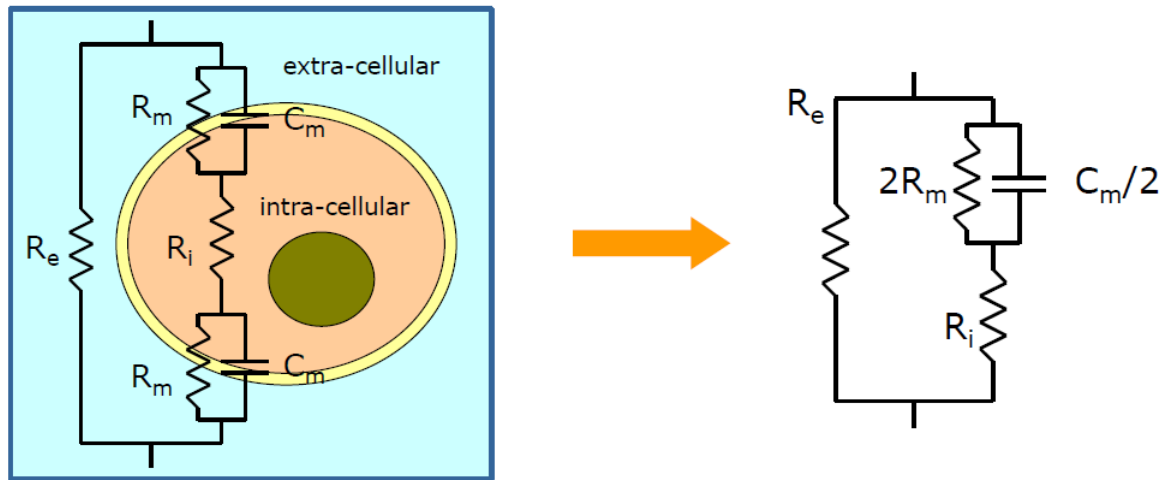
The electrical properties of biological tissues are strongly frequency dependent because different polarization mechanisms become dominant at different frequency ranges. Comprehensive databases of conductivity and permittivity values covering frequencies from 10 Hz to 100 GHz were compiled by Gabriel et al. and remain among the most widely used references for the dielectric characterization of biological tissues (GABRIEL; GABRIEL; CORTHOUT, 1996).

2.3.2 Electrical Representation of Biological Tissues

The electrical behavior of biological tissues can be understood from the electrical properties of their constituent cells. A cell consists of intracellular and extracellular conductive media separated by a lipid bilayer membrane. Since the membrane behaves as a dielectric layer between two conductive regions, it exhibits capacitive characteristics.

The passive component of the cell membrane is the bilayer lipid membrane, a structure approximately 7 nm thick that is permeable to lipids and water molecules but presents

Figure 6 – Equivalent electrical representation of a biological cell and its simplified circuit model.



Source: (IVORRA, 2002)

extremely low conductivity to ionic transport. As a result, the extracellular medium, the lipid bilayer membrane, and the intracellular medium form a conductor–dielectric–conductor structure that electrically behaves as a capacitor with a specific capacitance of approximately $1 \mu\text{F}/\text{cm}^2$ (IVORRA, 2002).

Embedded within the lipid bilayer are proteins, ion channels, pumps, and transport structures responsible for the active functions of the membrane. Ion channels provide selective pathways for ion transport across the membrane, while ion pumps actively transport ions using metabolic energy. These structures contribute significantly to the electrical behavior of cells and tissues by regulating ionic exchange between intracellular and extracellular compartments (IVORRA, 2002).

From an electrical perspective, the current injected into the extracellular medium may flow around the cell through the extracellular space, pass through membrane ion channels, or cross the membrane capacitance. Once inside the cell, the current flows through the intracellular medium before returning to the extracellular environment. This behavior can be represented by equivalent electrical circuits composed of extracellular resistance (R_e), intracellular resistance (R_i), membrane resistance (R_m), and membrane capacitance (C_m) (IVORRA, 2002).

2.3.3 Equivalent Circuit Models of Biological Tissues

Equivalent circuit models are widely used to describe the electrical behavior of biological tissues because they provide a direct relationship between measurable electrical parameters and underlying physiological structures. By representing tissues through combinations of resistive and capacitive elements, these models facilitate the interpretation of bioimpedance measurements and the identification of tissue-specific electrical properties.

In practice, however, the electrical response of biological tissues deviates from the be-

havior predicted by ideal circuit elements. Structural heterogeneity, distributed relaxation processes, surface roughness, and interfacial effects often produce non-ideal capacitive behavior. Consequently, the capacitive properties of biological tissues are frequently represented using a Constant Phase Element (CPE), whose impedance is given by:

$$Z_{CPE} = \frac{1}{Q(j\omega)^\alpha} \quad (12)$$

Where Q is a constant related to capacitance, ω is the angular frequency, and α is a dimensionless parameter ranging from 0 to 1. When $\alpha = 1$, the element behaves as an ideal capacitor, whereas $\alpha = 0$ corresponds to a purely resistive element.

When represented in the complex impedance plane, the Cole equation produces a depressed semicircular arc rather than the perfect semicircle predicted by the Debye model. This behavior reflects the non-ideal dielectric characteristics commonly observed in biological materials and is frequently represented through the use of a Constant Phase Element (CPE) (AYLLON; SEOANE; GIL-PITA, 2009) (MCADAMS; JOSSINET, 1996). The resulting graphical representation is commonly referred to as a Cole plot, although the terms Nyquist plot and Argand diagram are also found in the literature (MCADAMS; JOSSINET, 1996).

$$Z = R_\infty + \frac{R_0 - R_\infty}{1 + \left(j\frac{f}{f_c}\right)^\alpha} \quad (13)$$

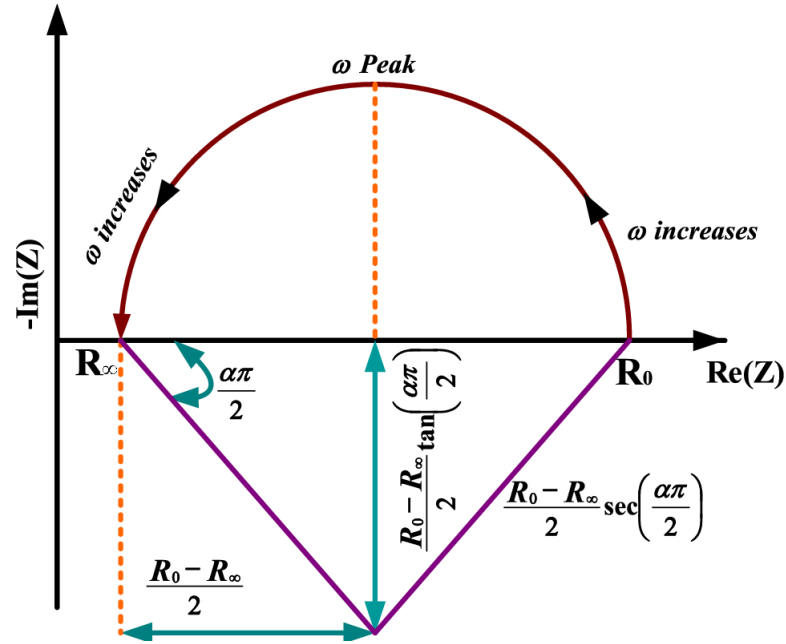
Where R_0 and R_∞ correspond to the resistances at zero and infinite frequency, respectively, f_c is the characteristic frequency, and α is a dispersion parameter ranging from 0 to 1.

When represented in the complex impedance plane, the Cole equation produces a depressed semicircular arc rather than the perfect semicircle predicted by the Debye model. This behavior reflects the non-ideal dielectric characteristics commonly observed in biological materials and is frequently represented through the use of a Constant Phase Element (CPE). (AYLLON; SEOANE; GIL-PITA, 2009) (MCADAMS; JOSSINET, 1996).

A CPE is an empirical circuit element commonly used to represent non-ideal capacitive behavior in electrochemical and biological systems. Unlike an ideal capacitor, whose phase angle remains constant at -90° , the CPE exhibits a frequency-independent phase angle between 0° and -90° , allowing the modeling of distributed relaxation processes, surface heterogeneity, porosity, and structural non-uniformities commonly observed in real materials (BARSOUKOV; MACDONALD, 2018) (MCADAMS; JOSSINET, 1996).

The Cole equation produces a depressed semicircular locus in the complex impedance plane, commonly referred to as a Cole plot. Unlike the ideal Debye response, which is represented by a perfect semicircle centered on the real axis, the Cole model introduces a dispersion parameter (α) that shifts the center of the arc below the real axis and accounts for the distributed

Figure 7 – Typical complex impedance locus of a biological tissue.



Source: Adapted from (ABDELATY et al., 2020)

relaxation behavior commonly observed in biological materials. The degree of depression of the semicircle is geometrically related to the angle $\alpha\pi/2$, while the radius and the vertical displacement of the circle center can be obtained from simple trigonometric relationships involving the low- and high-frequency resistances (R_0 and R_∞) (COLE; COLE, 1941) (FOSTER; SCHWAN, 1989) (MARTINSEN; HEISKANEN, 2023).

The maximum value of the imaginary component of impedance occurs at a characteristic angular frequency (ω_{peak}), which is associated with the dominant relaxation process of the material. These geometrical relationships provide a convenient interpretation of dispersive impedance spectra and facilitate the estimation of Cole parameters from experimental data. Although biological tissues frequently exhibit pronounced dielectric dispersion, materials with predominantly resistive behavior, such as saline or gelatin-based phantoms, generally present only a slight depression of the impedance arc and may approach the limiting case of an ideal resistor (MCADAMS; JOSSINET, 1996) (AYLLON; SEOANE; GIL-PITA, 2009) (ABDELATY et al., 2020).

2.4 ELECTRICAL CURRENT DISTRIBUTION IN MULTILAYER STRUCTURES

2.4.1 Current Penetration Depth

When an electrical current is injected into a conductive medium, its distribution is not uniform throughout the volume (PLONSEY; BARR, 2007). Instead, the current follows preferential pathways determined by the geometry of the electrode arrangement and by the electrical properties of the medium. In multilayer structures, the effective depth reached by the

current depends on factors such as electrode spacing, excitation frequency, and the conductivity of each layer. As a result, different portions of the structure contribute unequally to the measured impedance (BROWN; WILSON; BERTEMES-FILHO, 2000) (GRIMNES; MARTINSEN, 2008).

The injected current propagates through the medium forming current lines that represent the trajectories followed by the moving charge carriers. This current flow generates a spatial distribution of electrical potential between the current-injecting electrodes. The resulting voltage distribution can be represented by equipotential lines which are perpendicular to the current lines and connect points having the same electrical potential (PLONSEY; BARR, 2007). Together, current lines and equipotential lines provide a useful representation of the electric field distribution within the medium and constitute the basis for understanding current penetration and sensitivity in bioimpedance measurements (GESELOWITZ, 1971).

Analytical and numerical investigations performed by Bertemes-Filho demonstrated that the current density decreases progressively with depth in semi-infinite homogeneous conductors, indicating that only a fraction of the injected current effectively penetrates deeper regions of the medium. In layered structures, the presence of interfaces modifies the electric field distribution and alters the current pathways. Consequently, current penetration is not only governed by electrode spacing but also by boundary conditions imposed by adjacent layers and by their respective electrical conductivities. Furthermore, increasing the separation between electrodes tends to increase the effective penetration depth of the injected current, allowing deeper regions of the structure to contribute to the measured impedance (BERTEMES, 2006).

2.4.2 Electrical Interaction Between Layers

In layered media the measured impedance represents the combined electrical response of all layers involved in the current pathway. Variations in conductivity, permittivity, or thickness of one layer may alter the distribution of current and consequently affect the contribution of adjacent layers to the measurement. Therefore, the impedance obtained from a multilayer structure cannot generally be interpreted as the response of a single homogeneous material but rather as the result of electrical interactions among its constituent layers (BERTEMES-FILHO, 2018) (GRIMNES; MARTINSEN, 2008).

The response of an impedance measurement system is influenced by local changes occurring within the investigated volume. Modifications in the electrical properties or dimensions of an internal region may produce measurable variations in the recorded impedance, depending on the sensitivity of the electrode configuration to that specific location. Consequently, the ability to detect internal heterogeneities is directly related to the spatial distribution of the measurement sensitivity (BROWN; WILSON; BERTEMES-FILHO, 2000) (IVORRA, 2002).

However, the sensitivity profile can be modified through the selection of electrode geometry and measurement configuration. Understanding the balance between surface and deep sensitivity is essential when evaluating multilayer structures, since different layers may contribute

differently to the overall impedance response (BROWN; WILSON; BERTEMES-FILHO, 2000) (MARTINSEN; HEISKANEN, 2023).

In this work each layer is considered isotropic, homogeneous, and laterally semi-infinite with respect to the electrode dimensions and spacing. This assumption simplifies the analysis of current distribution and allows the effects of conductivity and thickness variations within multilayer structures to be investigated independently. The electrical interaction between adjacent layers also affects the spatial sensitivity distribution of the measurement system. Bertemes-Filho reported that sensitivity decreases substantially near insulating boundaries and that the contribution of deeper regions may become negligible under certain conductivity conditions. Conversely, when the superficial layer exhibits higher conductivity than the underlying layer, the sensitivity within deeper regions increases, producing a maximum sensitivity at a depth approximately proportional to the thickness of the superficial layer. These findings indicate that both layer thickness and conductivity ratios play a significant role in determining the detectability of internal variations in multilayer structures (BERTEMES, 2006).

3 ELECTRICAL BASICS ON SKIN AND TISSUE-MIMICKING PHANTOMS

3.1 SKIN STRUCTURE AND ELECTRICAL REPRESENTATION

3.1.1 Epidermis, Dermis and Hypodermis

The skin is the largest organ of the body (SWANN, 2010) and fulfills multiple functions of vital importance for homeostasis. It primarily acts as a physical protective barrier against external physical, chemical, and biological agents, which may be associated with the development of skin cancer. Furthermore, it performs functions in thermo regulation, immunological surveillance, and sensory transmission. It also prevents fluid loss to the outside and actively participates in the production of vitamin D (MULDER et al., 2016).

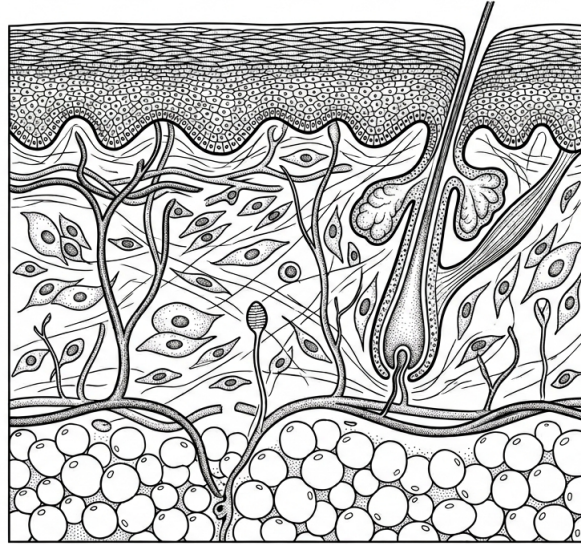
To achieve all these functions, it consists of three layers: the epidermis, the dermis, and the subcutaneous cellular tissue. Included within these are other important structures, such as cutaneous appendages, including hair follicles, sweat glands, sebaceous glands, blood vessels, lymphatic vessels, nerves, and specialized nervous structures (CALONJE et al., 2019).

The epidermis is the outermost layer, constituting a keratinized stratified squamous epithelium, which is composed of keratinocytes arranged in four layers (basal, spinous, granular, and corneum). The innermost layer of the epidermis corresponds to the basal membrane, whose cells are mitotically active, forcing the keratinocytes to move toward the surface through the epidermis. Subsequently, these cells flatten, lose their nucleus, and form the outer keratinized stratum corneum (KANG et al., 2019), consisting of dead cells lacking water and electrolytes. In addition to keratinocytes, the epidermis contains melanocytes, Langerhans cells, and Merkel cells, which perform different functions. Melanocytes, for their part, are essential in protecting against damage from ultraviolet radiation (FEATHER; RANDALL; WATERHOUSE, 2016) (KANG et al., 2022).

Below the epidermis lies the dermis, which provides mechanical and physiological support to the skin. Its thickness varies across anatomical regions, ranging from approximately 0.3 mm in the eyelid to 2.8 mm in the trunk (HWANG et al., 2016; WANG et al., 2022). The dermis is composed of connective tissue rich in collagen, elastic, and reticular fibers, and is organized into two layers: the papillary dermis, characterized by loosely arranged thin collagen fibers, and the reticular dermis, which contains thicker collagen bundles aligned parallel to the skin surface and extends toward the subcutaneous tissue. Fibroblasts represent the predominant cellular population and are responsible for synthesizing extracellular matrix components, including collagen and elastin, which provide elasticity and structural support (KANG et al., 2022; KANG et al., 2019).

Accurate knowledge of skin thickness is essential for determining electrode insertion depth and estimating the volume of tissue involved in bioimpedance measurements. In this context, Hwang et al. (HWANG et al., 2016) reported an average dermal thickness of approximately 0.9 mm in the forearm using high-frequency ultrasound in healthy subjects. Similarly,

Figure 8 – Schematic illustration of the normal structure of human skin.



Source: Adapted by the author based on (DALLEY; AGUR; MOORE, 2023) and (TORTORA; DERRICKSON, 2020)

Van Mulder et al. (MULDER et al., 2016) found a combined epidermisdermis thickness of 1.19 mm in the ventral forearm, with observable sex-related differences. In addition, the hypodermal layer in the forearm has been reported to vary between 1.06 and 1.9 mm (HWANG et al., 2016; WALLER; MAIBACH, 2005).

In the reticular dermis, there are also hair follicles and sweat glands whose function is to control body temperature, as well as to prevent microbial activity on the skin surface. At the dermal level, there are also blood vessels and sensory and autonomic nerve fibers. Sensory nerves reach free endings or specialized terminal organs that transmit signals of pressure, touch, and temperature, while autonomic nerves innervate the arrector pili muscles, blood vessels, and sweat glands (CALONJE et al., 2019). Finally, the subcutaneous cellular tissue constitutes the third layer of the skin and is largely composed of fatty connective tissue, with a reduced percentage of water (between 11 and 26%, depending on cell size) and electrolytes compared to other tissues such as muscle, whose water content varies between 76 and 77% (DIGIROLAMO; OWENS, 1976). This adipose layer functions as a thermal regulator and also protects the more superficial layers of the skin against bony prominences (KANG et al., 2019).

3.1.2 Equivalent Electrical Representation of Skin

The structural organization and electrical properties of the different skin layers summarized in Table 2 highlight the heterogeneous nature of cutaneous tissue. Each layer presents distinct thicknesses and resistivity values that influence the distribution and penetration of electrical currents during bioimpedance measurements. These electrical characteristics are particularly relevant in the context of skin cancer and the phenomenon of field cancerization, since neoplastic and premalignant processes are frequently associated with alterations in cellular or-

Table 2 – Thickness and resistivity of the different layers of the epidermis and dermis in the volar region of the forearm. Sources: ¹(TSAI et al., 2019), ²(ABE; NISHIZAWA, 2021a), ³(FAES et al., 1999), ⁴(SCHOTMAN et al., 2019)

Layer	Thickness (mm)	ρ @ 1 kHz ($\Omega \cdot m$)
Epidermis (ED)	0.075 ± 0.009^1	–
Stratum corneum (SC)	0.014 ± 0.004^1	–
Viable epidermis (VE)	0.061 ± 0.007^1	110^1
Dermis (DE)	1 ± 0.2^1	1.9^1
Total dermis and epidermis	1.2^1	$\sim 10^2$
Subcutaneous tissue / fat	13^2	$10^1 - 10^2$
Longitudinal muscle	2.2^2	2.4^3
Transverse muscle	8.3^2	6.8^3
Cortical bone	–	10^3
Extracellular fluid	0.5^2	0.67 ± 0.018^4

ganization, extracellular composition, and membrane integrity. Such changes can modify the electrical pathways within the tissue and, consequently, its measured impedance. Therefore, the electrical and geometrical parameters of healthy skin provide a fundamental reference framework for interpreting impedance variations associated with pathological processes and for supporting the development of bioimpedance-based approaches for the early detection and characterization of cutaneous lesions.

From a microscopic perspective, biological tissues are composed of collections of cells embedded within an extracellular matrix (BERTEMES-FILHO, 2018). Each cell is enclosed by a plasma membrane, a lipid bilayer formed by amphipathic phospholipids and glycolipids. In this structure, fatty acid chains (non-polar molecules) constitute the hydrophobic region, while alcohol or sugar groups (polar molecules) form the hydrophilic region. This arrangement promotes the self-organization of non-polar regions inward and polar regions outward, maintaining contact with intra- and extracellular fluids. As a result, the membrane behaves as a selective barrier that restricts the passage of ions and polar molecules. In addition, the plasma membrane is traversed by proteins responsible for biosensing, signaling, and transport (NELSON; COX, 2011).

According to cell theory, cells constitute the basic structural unit of living beings and are organized into specialized tissues to perform diverse physiological functions (RIBATTI, 2018). Human cells contain intracellular structures such as mitochondria, lysosomes, ribosomes, the Golgi apparatus, and the nucleus, all separated from the extracellular fluid by the plasma membrane. This membrane, due to its phospholipid composition, exhibits dielectric behavior and is surrounded by conductive media in both the intracellular and extracellular spaces. From an electrical perspective, the plasma membrane and the intra- and extracellular compartments form a system equivalent to a capacitor capable of storing electrical charge on both sides (ABASI et al., 2022).

When an electrical current is applied to a biological tissue using the bioimpedance technique, the frequency-dependent behavior of the membrane must be considered. At low and intermediate frequencies, the plasma membrane restricts ion movement into the intracellular space. At higher frequencies, however, this insulating behavior decreases, allowing ions to oscillate over short distances and increasing current penetration into the cells. Consequently, intracellular structures become polarized at low and intermediate frequencies, leading to charge storage across the membrane, a phenomenon known as electrical capacitance. Thus, the opposition to ionic movement associated with the cell membrane is frequency dependent (GONZALEZ-CORREA et al., 2003)(ABASI et al., 2022). In addition, ion channels can reduce membrane charge storage when open, decreasing capacitive reactance, while membrane disruption in cell death leads to the loss of capacitive effects (LIEBICH et al., 2022).

An important principle governing current propagation in biological tissues is that electrical current follows the pathways of lower impedance. Therefore, when current is injected through electrodes placed on the skin surface, it predominantly propagates through the dermis due to its relatively low resistivity. As electrode separation increases, deeper conductive tissues, particularly muscle, may also contribute to current transport, whereas the stratum corneum, subcutaneous fat, and cortical bone restrict current flow due to their higher resistivity values (ABE; NISHIZAWA, 2021b)(FAES et al., 1999).

Knowledge of skin thickness is important for determining both electrode insertion depth and the effective volume of tissue sampled during bioimpedance measurements.(HWANG et al., 2016) (BERTEMES, 2006) evaluated skin thickness using high-frequency ultrasound in healthy individuals, reporting an average dermal thickness in the forearm of approximately 0.9 mm. (MULDER et al., 2016) reported a combined epidermis and dermis thickness of 1.19 mm in the ventral forearm, with observed differences between men and women. Other studies indicate that the thickness of the hypodermis in the forearm ranges between 1.06 and 1.9 mm (HWANG et al., 2016)(WALLER; MAIBACH, 2005).

Various studies have described minimally invasive techniques for measuring skin bioimpedance these techniques use small electrodes that penetrate the stratum corneum, reducing its high resistivity and enabling more accurate measurements (ÅBERG, 2004). Aberg et al. described a probe with microelectrodes approximately 140 μm in length that penetrate the stratum corneum, enabling bioimpedance measurements for the diagnosis of cutaneous neoplasms. Subsequently, Glickman et al. developed the TranScan device based on this principle. A more recent implementation is the Nevisense 3.0 device from Scibase, which uses disposable gold-coated micro-pin electrodes that penetrate the skin (BRAUN et al., 2017). This system generates a score between 0 and 10, where values from 0 to 3 indicate benign lesions and values from 4 to 10 suggest cellular alterations compatible with malignant processes (LIEBICH et al., 2022). However, these devices are not widely available in clinical settings and have high costs, with each electrode unit reaching approximately 40 USD, which limits their use in experimental and research environments.

Electrical bioimpedance measurements differ between healthy skin and malignant lesions such as basal cell carcinoma, squamous cell carcinoma, or melanoma due to changes in cell size, shape, orientation, and membrane structure and function (BRAUN et al., 2017). Various studies have evaluated bioimpedance for the diagnosis of different neoplasms, including those of the thyroid, cervix, colon, prostate, oral cavity, melanoma, non-melanoma skin cancer, and Barrett's esophagus (GONZALEZ-CORREA et al., 2003) (SVOBODA et al., 2019) (ÅBERG, 2004). These approaches are based on the relationship between the passive electrical properties of biological tissues and their composition, as well as physiological factors such as total body water content, age, sex, race, physical condition, and metabolic status, all of which influence membrane permeability (ALVERO et al., 2011).

3.1.3 Skin Neoplasms with Field Cancerization

Skin cancer is the most frequent neoplasm in humans, and its incidence continues to rise worldwide (POZZOBON et al., 2018), mainly due to increased exposure to ultraviolet radiation, intrinsic factors such as personal and family history, lifestyle changes, and environmental influences (FEATHER; RANDALL; WATERHOUSE, 2016). According to the Global Cancer Observatory of the International Agency for Research on Cancer (IARC), more than 1.5 million new skin cancer cases are diagnosed worldwide each year, including approximately 330,000 melanoma cases, while the remainder corresponds to non-melanoma skin cancers (International Agency for Research on Cancer, 2025). Prognosis depends strongly on early diagnosis; therefore, timely detection and treatment of suspicious lesions remain essential in clinical practice. Within non-melanoma skin cancers, basal cell carcinoma (BCC) is the most frequent subtype, accounting for 75–80% of cases, followed by squamous cell carcinoma (SCC) with approximately 15%. Although melanoma represents only 3–5% of cases, it is responsible for about 75% of deaths related to cutaneous cancer (POZZOBON et al., 2018).

Beyond their epidemiological classification, skin cancers exhibit a wide range of morphological patterns that reflect their cellular origin and biological behavior. Non-melanoma tumors may present as localized nodular lesions or keratinized structures with relatively defined margins, whereas melanocytic malignancies often show irregular pigmentation, asymmetric borders, and heterogeneous surface structures associated with uncontrolled melanocyte proliferation. These morphological differences are clinically relevant for visual assessment and diagnostic screening, as they influence the spatial distribution of tissue alterations within the affected area. Representative examples of these morphological presentations are illustrated in Figure 10.

An important concept related to cancer diagnosis and early detection, not only in the skin but also in other tissues, and of particular relevance to electrical bioimpedance, is the phenomenon known as *field cancerization* or the *carcinogenic field effect*. This concept was first proposed by Slaughter and colleagues in 1953 based on histopathological observations of visually normal but histologically altered epithelium surrounding oropharyngeal squamous cell carcinomas. They hypothesized that multiple primary neoplasms can arise simultaneously and

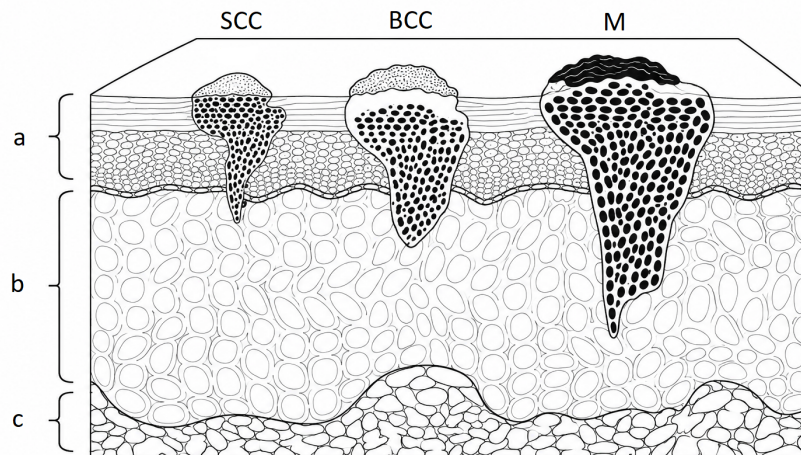


Figure 9 – Representative schematic illustrations of common skin cancer types: Squamous cell carcinoma (SCC), Basal cell carcinoma (BCC) and Melanoma; Epidermis (a), Dermis (b) and Hypodermis (c)

Source: Adapted by the author based on (DALLEY; AGUR; MOORE, 2023) and (TORTORA; DERRICKSON, 2020)

coexist with subclinical precursor lesions within a defined tissue field (SLAUGHTER; SOUTHWICK; SMEJKAL, 1953).

The field cancerization paradigm proposes that a region of epithelium becomes altered as a consequence of regional carcinogenic activity, leading to irreversible genetic changes in multiple clonal cell populations (CHRISTENSEN, 2018). Before the appearance of a clinically detectable lesion, a lineage of normal cells may acquire pro-tumorigenic genetic mutations or epimutations. These altered clones can subsequently expand and generate foci of cells predisposed to malignant transformation, while still appearing histologically normal or exhibiting only dysplastic changes (CURTIUS; WRIGHT; GRAHAM, 2017).

Paolo Dotto and colleagues (DOTTO, 2014) suggest that field cancerization contributes to the development of multifocal primary tumors and increases the probability of local recurrence, even after complete surgical excision of malignant lesions. More recently, the concept has evolved to encompass the presence of premalignant alterations in multiple tissue regions associated with the primary tumor (LIAO et al., 2019). Curtius and colleagues (CURTIUS; WRIGHT; GRAHAM, 2017) further describe field cancerization as a population of cells that has acquired phenotypic traits favoring malignant progression, including increased proliferative capacity, reduced cell death, and enhanced immune evasion.

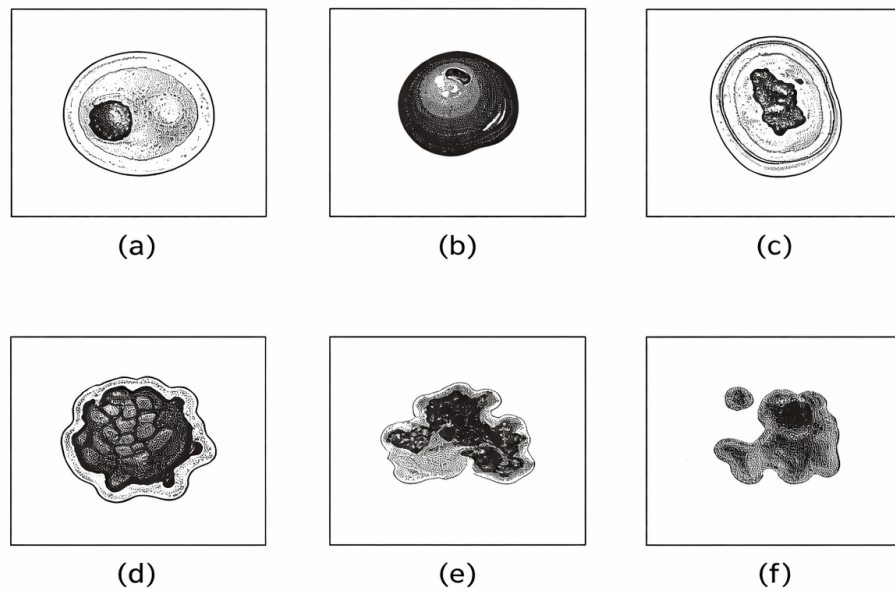


Figure 10 – Representative schematic illustrations of common skin cancer types: (a) Basal cell carcinoma (BCC); (b) Merkel cell carcinoma (MCC); (c) Squamous cell carcinoma (SCC); (d) Nodular melanoma; (e) Superficial spreading melanoma; (f) Lentigo maligna melanoma.

Source: Adapted by the author based on (DALLEY; AGUR; MOORE, 2023) and (TORTORA; DERRICKSON, 2020)

3.2 BIOLOGICAL SKIN PHANTOMS

3.2.1 Tissue-Mimicking Materials

Biological phantoms are objects designed to mimic the properties of living tissues. In the medical field, these devices play a fundamental role in the evaluation, analysis, and calibration of medical equipment, as well as in educational and research activities conducted in academic environments. Phantoms are widely used to imitate biological tissues and can be fabricated in different physical forms, including liquid, solid, and semi-solid configurations (MIRBEIK-SABZEVAR; TAVASSOLIAN, 2019). According to Kavitha and Reddy, phantoms are inanimate objects specifically designed to reproduce the properties of biological tissues, incorporating variations that allow the representation of both normal and abnormal conditions (KAVITHA; REDDY, 2012). Although commercial phantoms are available, the increasing demand for application specific models and the associated costs have stimulated the development of low-cost alternatives tailored to particular experimental requirements. In this context, phantoms that simulate various tissues and organs have been extensively investigated in several fields, including optical spectroscopy, biomedical imaging, dosimetry, and ballistic studies, among oth-

ers (POGUE; PATTERSON, 2006). Among the different types of phantoms, liquid phantoms present the simplest manufacturing process, as their fabrication generally does not require complex preparation procedures. One widely adopted approach involves the use of gelatin-based phantoms to simulate the dielectric properties of biological tissues.

These phantoms are particularly attractive because they can be produced at low cost using simple materials, mainly water and gelatin (MARCHAL; NADI; TOSSER, 1989). In addition, gelatin mixtures can be poured into molds, allowing the fabrication of structures with different shapes depending on the organ or tissue to be simulated, while their preservation and handling remain relatively simple. From this perspective, the replication of the passive electrical characteristics of human skin in a controlled laboratory environment through the use of phantoms provides new opportunities for experimental testing and educational applications at reduced cost (MARCHAL; NADI; TOSSER, 1989) (CHEN et al., 2016). Methodologies have been proposed to reproduce the electro magnetic response of human skin using multilayer biological phantoms composed of gelatin and agar. These models allow the standardization of tissue-mimicking structures, facilitate the study of anomalies such as granuloma formation, and support the development of systems aimed at the early detection of pathological conditions.

Beyond their role in tissue simulation, biological phantoms also enable a more accurate evaluation of the interaction between electromagnetic fields and multilayer structures, particularly in radio frequency applications. For these studies, the adequate representation of physical parameters such as electrical conductivity, thermal conductivity, density, and specific heat is essential to reproduce the thermal and electromagnetic responses induced in biological tissues. Among these parameters, electrical conductivity plays a particularly important role in the processes of energy conversion, directly influencing the validity of preclinical cancer studies and the prediction of tissue behavior during skin treatments.

Furthermore, skin-equivalent phantoms have proven to be valuable tools in the development and validation of technologies operating in the microwave and millimeter-wave frequency regimes, especially in imaging applications and in the detection of pathologies such as skin cancer. The dielectric characterization of these models over wide frequency ranges, together with the temporal stability of their electrical properties, enables reproducible experimental conditions comparable to those observed in real biological tissues. In addition, it has been demonstrated that the suitability of a phantom depends not only on the operating frequency range but also on its geometry, which should approximate the dimensions of the target tissue under specific experimental conditions (KRANOLD et al., 2020).

Based on these considerations, the construction of multilayer skin phantoms is commonly guided by the structural and electrical characteristics reported for the main layers of human skin. In particular, phantom models are designed to reproduce the epidermis, dermis, and hypodermis, which exhibit distinct thicknesses and passive electrical properties. Each layer can therefore be tailored according to the ranges of resistivity and geometrical dimensions described in the literature, preserving the relative organization observed in vivo. Nevertheless, for

modeling and analytical purposes, skin-equivalent phantoms are frequently approximated as homogeneous, isotropic, and semi-infinite media, simplifying the description of electromagnetic interactions within tissue-mimicking structures.

This work adopts a low-cost methodology for the fabrication of multilayer biological phantoms capable of reproducing the electromagnetic behavior of human skin. The proposed models emulate the geometrical and passive electrical properties of the epidermis, dermis, and subcutaneous tissue, providing an experimentally controlled environment for evaluating bioimpedance-based measurement techniques. Their fabrication relies on inexpensive and readily available materials, facilitating implementation and replication in different laboratories. In addition, these phantoms provide a reproducible platform for investigating changes in tissue electrical properties associated with localized abnormalities, while enabling the assessment of different electrode geometries, sensitivity distributions, and interrogation depths. Such characteristics contribute to the standardization of tissue-mimicking structures and support the development and validation of bioimpedance methodologies for the early detection and characterization of skin pathologies.

Considering these requirements, gelatin and saline solutions have been extensively employed as the basis for constructing biological phantoms because they are inexpensive, biocompatible, easy to process, and allow controlled adjustment of their electrical properties. The following section presents the fundamental characteristics of these materials and discusses their suitability for the fabrication of multilayer skin-equivalent phantoms.

3.2.2 Gelatine, Agar and Saline Based Phantoms

Gelatin is a homogeneous colloidal gel derived primarily from collagen present in animal tissues. This collagen may originate from various sources and exhibit different amino acid compositions and sequences, which enables its use in the production of gelatin (MADSEN, 1978) (BURLEW, 1980) (HANANI, 2016). As a consequence of this structural diversity, commercially available gelatins constitute heterogeneous mixtures of polypeptides (BOSCH; GIELENS, 2003).

Collagen is one of the most abundant fibrous proteins in animal organisms and performs multiple mechanical functions, particularly in mammals. In the skin, it constitutes a fundamental structural component of the dermis and contributes significantly to its mechanical strength and support properties. It is also part of the extracellular matrix of various tissues, including tendons, ligaments, bone, dentin, cartilage, and blood vessels.

The basic structural unit of collagen-rich tissues is the collagen fibril, a fiber whose thickness ranges approximately from 50 to several hundreds of nanometers. These fibrils are arranged through a hierarchical organization that results in complex tissue structures with different mechanical properties (FRATZL, 2008).

Due to its origin and physicochemical properties, gelatin has been widely used in the construction of hydrogel-based phantoms for the simulation of biological tissues. For decades,

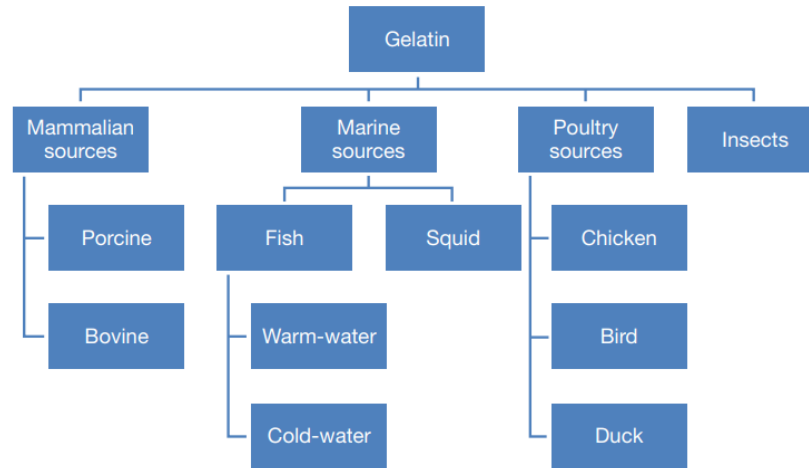


Figure 11 – Sources of Gelatin Production.

Source: (HANANI, 2016)

these materials have been employed in various biomedical applications (MADSEN, 1978), including ultrasound imaging studies (KAVITHA; REDDY, 2012), radiofrequency heating tests (YUAN, 2012), among others.

In general, hydrogels used in the fabrication of phantoms can be classified into two categories: physical gels, obtained through physical processes such as heating and cooling, and chemical gels, produced through chemical reactions that include polymerization processes. In the field of dermatological research and in the development of diagnostic methods based on electrical properties, gelatin-based phantoms have been widely used to simulate cutaneous tissues. This is due to their ease of preparation, low cost, and ability to reproduce mechanical and electrical properties comparable to those of the skin. Owing to these characteristics, such phantoms can be used for the evaluation and validation of measurement systems, as well as for studying the electrical behavior of the skin and possible alterations associated with pathological processes. Likewise, different research groups have characterized the acoustic and elastic properties of these phantoms in applications such as ultrasound elastography (KAVITHA; REDDY, 2012).

The fabrication of the cutaneous phantom utilizes a matrix of distilled water and 5% *NaCl*. To control the dielectric properties, the saline concentration is adjusted to calibrate the electrical conductivity, while the gelatin proportion remains constant to ensure mechanical stability. Since the molecular structure of the gelatin introduces a baseline ionic charge, maintaining a constant concentration ensures that the modulation of the final resistivity depends exclusively on the saline solution.

In this manner, the chemical formulation is established as a function of both the experimental conditions and the predefined dielectric parameters of cutaneous tissue, as reported by Faes et al. and Tsai et al. Specifically, the target resistivity values for emulating living tissue are approximately 100 Ωm for the epidermis, 10 Ωm for the dermis (FAES et al., 1999) (TSAI et al., 2019), and 38.5 Ωm for the hypodermis, given its adipose composition (KAVITHA; REDDY,

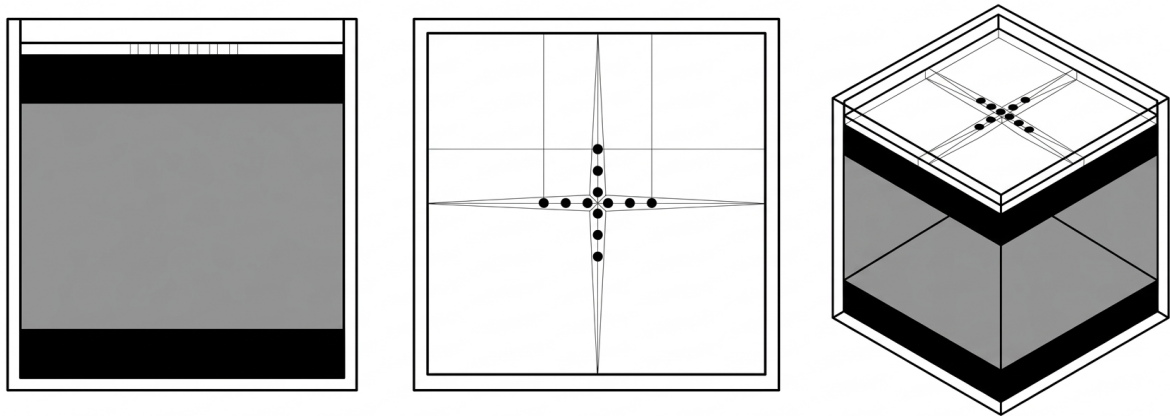


Figure 12 – Side, top, and isometric elevational views of the gelatin phantom. The visualization highlights the multilayered matrix and the corresponding perforated cover, prior to the final positioning and fixation of the measuring electrodes.

Source: The Author

2012). Since this study aims to evaluate whether surface impedance measurements accurately reflect variations in individual layer resistivities, this controlled synthesis allows for a precise parametric characterization of the phantom's electrical response.

Once the mixing parameters are established, the solution is cast into a mold to facilitate solidification and the formation of a stable matrix. The phantom is constructed using a multilayer architecture designed to emulate the primary anatomical strata of human skin. In this configuration, the layers are disposed with a thickness ratio of approximately 15% for the outer layers and 70% for the intermediate layer, thereby approximating the structural proportions of the human integumentary system.

Geometrically, the phantom is fabricated as a rectangular parallelepiped. Its lateral dimensions are specified to be at least three times the width of the inter-electrode array. This design ensures the approximation of a homogeneous, isotropic, and semi-infinite medium in the vicinity of the electrodes, effectively preventing the container boundaries from interfering with the electric field lines or the equipotential surfaces during impedance measurements.

3.3 FABRICATION OF MULTILAYER SKIN PHANTOMS

3.3.1 Phantom Preparation Protocols

The preparation of the gelatin-based solution was carried out without agar, using gelatin as the sole gelling agent. The saline solution was incorporated using a syringe to ensure precise control of ionic concentration. Initially, all components were weighed, and the analytical balance was tared to guarantee measurement accuracy. Distilled and deionized water was then added to the mixing container, followed by the gradual addition of pre-weighed gelatin under continuous stirring to prevent aggregation. The mixture was stirred until complete homogenization was achieved and no visible lumps remained.

Once a uniform solution was obtained, the saline solution was injected using a syringe to accurately control ionic content. Continuous stirring was maintained while the mixture was heated, progressively increasing the temperature until reaching approximately 80 °C. This step ensured complete dissolution and homogenization of the solution.

It is important to note that agar was not used in this formulation, since it presents higher intrinsic ionic conductivity than gelatin and contributes minimally to the mechanical stiffness required for the desired solid-like state. Consequently, larger quantities would be required to achieve comparable mechanical properties, which would undesirably increase the overall conductivity of the phantom and compromise its electrical similarity to biological tissue.

For calibration procedures, electrical resistivity measurements were recorded during the cooling ramp, with data acquisition performed every 5°C until the system reached 25°C. This range captures the transition from the liquid to the gel state, where the electrical properties become highly temperature dependent.

For phantom construction, the prepared solution was poured into an acrylic mold designed for multilayer assembly. Prior to casting, the internal volume of the mold was marked to define the proportional thickness of each anatomical layer, specifically 15% for the epidermis, 70% for the dermis, and 15% for the hypodermis. The solution was then carefully poured up to the predefined level for each layer, ensuring geometric consistency and reproducibility of the phantom structure.

Accurate control of layer thickness is ensured through precise mass measurement of all components using a high-precision analytical balance, enabling reproducible concentrations for each prepared solution. In addition, the total volume of each layer is controlled by filling the mold up to the predefined marking, ensuring a uniform and planar solidification surface. This procedure minimizes geometric variability and guarantees consistency in thickness across different phantom realizations.

Structural stability is achieved by allowing each layer to fully solidify before the subsequent layer is added. This sequential casting strategy prevents intermixing between layers and preserves the mechanical integrity of the multilayer structure. Furthermore, it ensures that each layer retains its intended electrical passive properties, since premature mixing or thermal disturbance could alter ionic distribution and, consequently, the effective conductivity of the system.

3.3.2 Reproducibility Considerations

Temperature is a critical factor in all experiments due to its direct influence on the electrical conductivity of materials. It is well established that temperature variations affect ionic mobility in electrolyte solutions, thereby modifying their effective conductivity. For this reason, all measurements in this work were performed under controlled thermal conditions, ideally maintaining temperature constant or with negligible variation during each phantom characterization.

The use of distilled and deionized water ensures the absence of uncontrolled ionic impurities that could introduce stochastic variations in the electrical properties of the phantom. During synthesis, the salinegelatin mixture is heated to 85°C to promote complete dissolution and homogenization of the polymeric matrix. However, since electrical conductivity exhibits a strong dependence on temperature, it must be treated as a temperature-dependent variable rather than a constant parameter.

To account for this behavior, a calibration procedure was implemented based on previous experimental datasets. In this step, conductivity was modeled using a polynomial regression approach constructed from controlled measurements obtained during a downward thermal sweep from 85°C to 30°C in 5°C increments. This range was selected because 30°C corresponds to the gelation threshold, below which increased viscosity and phase transition prevent proper probe insertion and compromise measurement repeatability.

By repeating this calibration procedure over ten independent synthesis cycles and systematically varying the NaCl concentration, a multivariable polynomial model was obtained. This model relates target electrical conductivity to both temperature and saline concentration, enabling estimation of the required NaCl fraction for phantom preparation, as shown in Eq. (14).

$$CS = -8.27 \times 10^{-4} T + 4.90 \times 10^{-2} \sigma + 1.65 \times 10^{-2} \quad (14)$$

Where CS represents the saline concentration (wt%), T is the operational temperature in degrees Celsius (°C), and σ denotes the desired electrical conductivity in Siemens per meter (S/m).

However, for the case study presented in Chapter 4, new calibrations were performed using lower gelatin concentrations in order to approximate the resistivity values reported in the literature. These adjustments allow the model to better match reference biological tissues while maintaining the reproducibility of the measurements.,,

4 CASE STUDY: ELECTRICAL CHARACTERIZATION OF MULTILAYER SKIN PHANTOMS

4.1 EXPERIMENTAL PROTOCOL

Three phantoms were developed, each composed of three stacked layers with known resistivities, assembled in a custom-made acrylic box. The layers follow a thickness ratio of 15% (outer layers) to 70% (middle layer), reflecting typical proportions of skin structure. The phantoms have the shape of right-rectangular parallelepipeds. They resemble homogeneous, isotropic, and semi-infinite spaces only in the dimensions adjacent to the electrode arrangement. The remaining dimension reflects the variation in layer thicknesses. Each layer was prepared by mixing distilled water to avoid conductive ions with sodium chloride at concentrations specified in Table 1 to achieve the desired conductivity. Gelatine were added at constant concentrations, with agar acting as a gelling and hardening agent (DUTRA, 2018). A cover with twelve circular holes is placed on the first layer, holding 1.6 mm diameter stainless steel electrodes arranged in a cross shape. The center-to-center electrode spacing is 5 mm, allowing a maximum interelectrode distance of 25 mm along each axis. The phantoms box edges measure 81 mm over three times this distance ensuring a semi-infinite space along the horizontal axes. This setup keeps the box walls far from the equipotential and electric field lines. The objective is to assess how the electrode spatial configuration affects impedance measurements of the multilayer phantom by varying each layers conductivity and the inter electrode distances. Measurements use four electrodes, two for current injection and two for voltage detection with configurations primarily in linear and coplanar arrangements.

Three linear electrode configurations vary the inter-electrode distances along one axis in 5 mm steps. In all cases, current electrodes are fixed at the ends, while voltage electrodes are positioned internally initially 5 mm and then 10 mm from the current electrodes. The third linear configuration uses only the four internal electrodes along the axis.

For coplanar arrangements, two configurations use two electrode axes. The first measurement places the four most central electrodes, with current electrodes on one side and voltage electrodes on the other. Subsequent measurements shift electrodes outward to the next external pairs, maintaining the same pattern. This results in three electrode configurations at increasing distances, varying proportions consistently across electrodes

4.1.1 Conductivity Temperature Calibration

The determination of the electrical conductivity of the phantoms initially required the calibration of the probe used with the LUTRON CD-4303 conductivity meter, given that the equipment directly provides conductance measurements in siemens (S), whereas conductivity must be expressed in units of siemens per meter (S/m). For this purpose, a 0.01 M NaCl standard solution at 25°C, with an expected conductivity of 0.141 S/m, was employed. From a measured conductance of 0.0011 S, a cell constant of 128.2 m^{-1} was determined. This value

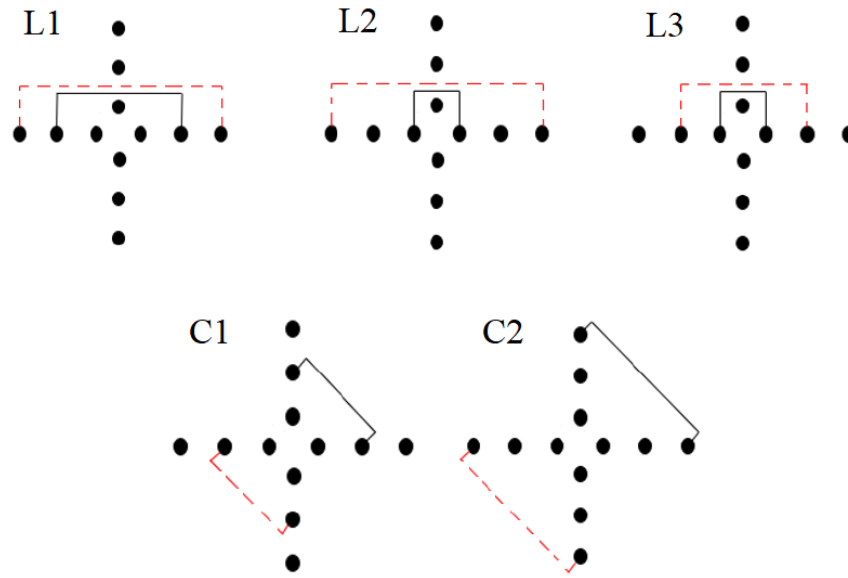


Figure 13 – The five different configurations for linear and coplanar interelectrode arrays. The dotted lines correspond to the current electrodes and the solid lines to the voltage electrodes respectively.

Source: The Author

was subsequently used to convert all conductance measurements into electrical conductivity.

During phantom preparation, the mixtures undergo substantial temperature variations, beginning with the use of hot water to facilitate the dissolution of the gelling agents, followed by cooling to room temperature and ultimately reaching the solidification stage. Since direct conductivity measurements at 10°C become impractical once the material has solidified, all formulations evaluated in this study were characterized by means of descending temperature ramps. Conductivity measurements were acquired from 60°C to 30°C at regular temperature intervals, allowing the relationship between temperature and electrical properties to be established and enabling the estimation of the expected resistivity at the measurement temperature of 10°C through regression models.

As an initial approach for selecting the base material for the phantoms, formulations prepared with agar and gelatin, the latter synthesized according to Dutra (DUTRA, 2018), were evaluated while maintaining fixed proportions of gelling agent and water. The conductivity curves obtained during the thermal ramps for both formulations are presented in Table 3. However, it was observed that the resulting resistivities were confined to a relatively narrow range and did not reproduce the higher resistivity values reported for certain skin layers. In particular, Faes (FAES et al., 1999) reports approximate resistivities of 100 $\Omega \cdot m$ for the epidermis, 10 $\Omega \cdot m$ for the dermis, and 40 $\Omega \cdot m$ for the hypodermis. These limitations were observed even in the absence of sodium chloride additions.

A second study was subsequently carried out using agar exclusively in an attempt to increase the resistivity while preserving suitable mechanical properties for multilayer fabrication.

Table 3 – Temperature-Dependent Electrical Properties for Phantom Test 1 (12.5 g Gelatin, 0.5 g Agar, 0.1 g NaCl, and 86.9 g Water).

T [°C]	Conductivity (S/m)	Resistivity ($\Omega\cdot\text{m}$)
65.0	0.7961	1.2561
62.5	0.7756	1.2893
60.0	0.7577	1.3198
57.5	0.7307	1.3685
55.0	0.7154	1.3979
52.5	0.6897	1.4499
50.0	0.6679	1.4972
47.5	0.6410	1.5601
45.0	0.6154	1.6251
42.5	0.5961	1.6775
40.0	0.5692	1.7568
37.5	0.5256	1.9025
35.0	0.5231	1.9118
32.5	0.5064	1.9748

A formulation containing 4 g of agar and 96 mL of water was subjected to the same cooling thermal ramp from 60°C to 30°C, and the results are presented in Table 4. Although resistivities ranging approximately from 19.7 $\Omega\cdot\text{m}$ to 28.9 $\Omega\cdot\text{m}$ were achieved, the material exhibited a viscous-liquid consistency even at low temperatures, preventing the fabrication of stable multi-layer structures and surface impedance measurements.

Table 4 – Temperature-Dependent Conductivity Measurements for the Agar Phantom Prepared with 4 g of Agar and 96 mL of Water.

T [°C]	Reading (S)	Reading (mS)	Conductivity (S/m)	Resistivity ($\Omega\cdot\text{m}$)
60.0	0.000395	0.395	0.0506	19.7476
57.5	0.000395	0.395	0.0506	19.7476
55.0	0.000368	0.368	0.0472	21.1965
52.5	0.000362	0.362	0.0464	21.5478
50.0	0.000352	0.352	0.0451	22.1600
47.5	0.000340	0.340	0.0436	22.9421
45.0	0.000331	0.331	0.0424	23.5659
42.5	0.000325	0.325	0.0417	24.0010
40.0	0.000308	0.308	0.0395	25.3257
37.7	0.000304	0.304	0.0390	25.6589
35.0	0.000294	0.294	0.0377	26.5317
32.5	0.000286	0.286	0.0367	27.2738
30.0	0.000270	0.270	0.0346	28.8900

Due to these limitations, gelatin-based phantoms were further investigated by varying the gelatin concentration. Additional formulations containing 8 g, 6 g, and 3 g of gelatin were prepared and characterized using the same temperature-ramp methodology to determine their

resistivity at 10°C and identify the maximum resistivity attainable while maintaining adequate mechanical stability. Subsequently, sodium chloride was incorporated in controlled amounts to broaden the range of achievable electrical properties. The resulting conductivity measurements enabled the adjustment of a multiple regression model relating temperature, salt concentration, and conductivity, allowing the estimation of both the amount of NaCl required to achieve a target conductivity and the expected resistivity of the phantom at 10°C. This procedure improved the reproducibility of phantom fabrication and provided a systematic methodology for tailoring the electrical properties of the multilayer skin phantoms used in subsequent experiments.

Table 5 – Temperature-Dependent Conductivity Measurements for the Gelatin Phantom Prepared with 8 g of Gelatin and 92 mL of Water.

T [°C]	Reading (S)	Reading (mS)	Conductivity (S/m)	Resistivity (Ωm)
60.0	0.001645	1.645	0.2109	4.7418
57.5	0.001488	1.488	0.1908	5.2421
55.0	0.001472	1.472	0.1887	5.2991
52.5	0.001409	1.409	0.1806	5.5361
50.0	0.001360	1.360	0.1744	5.7355
47.5	0.001318	1.318	0.1690	5.9183
45.0	0.001269	1.269	0.1627	6.1468
42.5	0.001224	1.224	0.1569	6.3728
40.0	0.001132	1.132	0.1451	6.8907
37.7	0.001108	1.108	0.1420	7.0400
35.0	0.001086	1.086	0.1392	7.1826
32.5	0.001048	1.048	0.1344	7.4430
30.0	0.000990	0.990	0.1269	7.8791

Table 6 – Temperature-Dependent Conductivity Measurements for the Gelatin Phantom Prepared with 6 g of Gelatin and 94 mL of Water.

T [°C]	Reading (S)	Reading (mS)	Conductivity (S/m)	Resistivity (Ωm)
60.0	0.001055	1.055	0.1353	7.3937
57.5	0.001021	1.021	0.1309	7.6399
55.0	0.000992	0.992	0.1272	7.8632
52.5	0.000965	0.965	0.1237	8.0832
50.0	0.000929	0.929	0.1191	8.3965
47.5	0.000903	0.903	0.1158	8.6382
45.0	0.000870	0.870	0.1115	8.9659
42.5	0.000843	0.843	0.1081	9.2530
40.0	0.000817	0.817	0.1047	9.5475
37.7	0.000797	0.797	0.1022	9.7871
35.0	0.000750	0.750	0.0962	10.4004
32.5	0.000725	0.725	0.0929	10.7591
30.0	0.000685	0.685	0.0878	11.3873

Table 7 – Temperature-Dependent Conductivity Measurements for the Gelatin Phantom Prepared with 3 g of Gelatin and 97 mL of Water.

T [°C]	Reading (S)	Reading (mS)	Conductivity (S/m)	Resistivity ($\Omega\cdot\text{m}$)
60.0	0.000585	0.585	0.0750	13.3339
57.5	0.000578	0.578	0.0741	13.4953
55.0	0.000562	0.562	0.0720	13.8796
52.5	0.000556	0.556	0.0713	14.0293
50.0	0.000528	0.528	0.0677	14.7733
47.5	0.000515	0.515	0.0660	15.1462
45.0	0.000495	0.495	0.0635	15.7582
42.5	0.000478	0.478	0.0613	16.3186
40.0	0.000461	0.461	0.0591	16.9204
37.7	0.000443	0.443	0.0568	17.6079
35.0	0.000425	0.425	0.0545	18.3537
32.5	0.000411	0.411	0.0527	18.9789
30.0	0.000391	0.391	0.0501	19.9496

The formulation containing 3 g of gelatin and 97 mL of water was selected as the epidermis-equivalent layer. Based on the temperature–resistivity calibration, the resistivity at the measurement temperature (10°C) was estimated using Eqs. (16)–(19), yielding an extrapolated value of approximately 23.84 Ωm . This value was adopted as the reference resistivity for the epidermis-equivalent phantom, while the target resistivities for the remaining layers were determined according to the relative ratios reported in the literature.

$$\rho(T) = -0.2227T + 26.0690 \quad (15)$$

$$\rho(10) = -0.2227(10) + 26.0690 = 23.84 \Omega\text{m} \quad (16)$$

$$\sigma(10) = \frac{1}{\rho(10)} = \frac{1}{23.84} = 0.0419 \text{ S/m} \quad (17)$$

$$\rho_{epi} = 23.84 \Omega\text{m} \quad (18)$$

Rather than reproducing the absolute resistivity values reported for human skin, the multilayer phantom was designed to preserve the relative electrical contrasts among the epidermis, dermis, and hypodermis described in the literature (FAES et al., 1999). Consequently, the target resistivities of the remaining layers were obtained by scaling the resistivity of the epidermis-equivalent formulation according to the relative ratios reported for biological skin tissues.

For the dermis:

$$\rho_{der} = \rho_{epi} \left(\frac{20}{100} \right) = 23.84 \left(\frac{20}{100} \right) = 4.77 \, \Omega\text{m} \quad (19)$$

$$\sigma_{der} = \frac{1}{\rho_{der}} = \frac{1}{4.77} = 0.2096 \, \text{S/m} \quad (20)$$

For the hypodermis:

$$\rho_{hyp} = \rho_{epi} \left(\frac{40}{100} \right) = 23.84 \left(\frac{40}{100} \right) = 9.54 \, \Omega\text{m} \quad (21)$$

$$\sigma_{hyp} = \frac{1}{\rho_{hyp}} = \frac{1}{9.54} = 0.1048 \, \text{S/m} \quad (22)$$

Table 8 – Target electrical properties adopted for the multilayer skin phantom based on relative resistivity scaling.

Layer	Relative Ratio	Resistivity (Ωm)	Conductivity (S/m)
Epidermis	1.00	23.84	0.0419
Dermis	0.20	4.77	0.2096
Hypodermis	0.40	9.54	0.1048

Once the target apparent resistivities had been defined, two additional gelatin formulations consisting of 3 g of gelatin and 97 mL of water were prepared by incorporating different amounts of sodium chloride (NaCl), namely 0.010 g and 0.020 g. Similar to the previous experiments, both formulations were subjected to a descending temperature ramp from 60°C to 30°C, while conductivity measurements were acquired at regular intervals. These data were used to establish an empirical relationship among temperature, NaCl concentration, and electrical conductivity, enabling the estimation of the amount of salt required to achieve a target resistivity at a specified temperature. The resulting conductivity profiles are presented in Tables 9 and 10.

To address this issue, an empirical multiple regression model was developed using gelatin formulations containing different sodium chloride concentrations and conductivity measurements acquired over a range of temperatures. Since the phantoms are fully solidified at the measurement temperature of 10°C, direct conductivity measurements at this stage are difficult to perform. Therefore, the model was employed to estimate both the amount of NaCl required to achieve a target electrical resistivity and the expected resistivity of the phantom at 10°C. The resulting relationship is given by:

$$\sigma_{epi} = \frac{1}{23.84} = 0.04195 \, \text{S/m}, \quad (23)$$

Table 9 – Simulated Temperature-Dependent Conductivity Profile for a Gelatin Phantom Prepared with 3 g of Gelatin, 97 mL of Water and 0.010 g NaCl.

T [°C]	Reading (S)	Reading (mS)	Conductivity (S/m)	Resistivity ($\Omega \cdot m$)
60.0	0.000817	0.817	0.1047	9.5475
57.5	0.000799	0.799	0.1024	9.7603
55.0	0.000784	0.784	0.1005	9.9487
52.5	0.000760	0.760	0.0974	10.2672
50.0	0.000741	0.741	0.0950	10.5263
47.5	0.000711	0.711	0.0911	10.9802
45.0	0.000698	0.698	0.0895	11.1732
42.5	0.000683	0.683	0.0876	11.4155
40.0	0.000674	0.674	0.0864	11.5741
37.7	0.000659	0.659	0.0845	11.8294
35.0	0.000646	0.646	0.0828	12.0804
32.5	0.000638	0.638	0.0818	12.2336
30.0	0.000625	0.625	0.0801	12.4844

Table 10 – Simulated Temperature-Dependent Conductivity Profile for a Gelatin Phantom Prepared with 3 g of Gelatin, 97 mL of Water and 0.020 g NaCl.

T [°C]	Reading (S)	Reading (mS)	Conductivity (S/m)	Resistivity ($\Omega \cdot m$)
60.0	0.001635	1.635	0.2096	4.7710
57.5	0.001602	1.602	0.2054	4.8687
55.0	0.001572	1.572	0.2015	4.9628
52.5	0.001523	1.523	0.1952	5.1229
50.0	0.001484	1.484	0.1902	5.2576
47.5	0.001424	1.424	0.1826	5.4765
45.0	0.001397	1.397	0.1791	5.5848
42.5	0.001368	1.368	0.1754	5.7018
40.0	0.001349	1.349	0.1730	5.7803
37.7	0.001319	1.319	0.1691	5.9134
35.0	0.001292	1.292	0.1656	6.0386
32.5	0.001276	1.276	0.1636	6.1112
30.0	0.001250	1.250	0.1602	6.2422

$$\sigma(T, m_{\text{NaCl}}) = 0.03908 + 2.87 \times 10^{-4}T + 8.3845 m_{\text{NaCl}}, \quad (24)$$

where σ is the electrical conductivity (S/m), T is the temperature (°C), and m_{NaCl} is the mass of sodium chloride added to the formulation (g).

Since all impedance measurements were carried out at 10°C, Eq. (24) simplifies to

$$\sigma(10, m_{\text{NaCl}}) = 0.04195 + 8.3845 m_{\text{NaCl}}, \quad (25)$$

The required amount of sodium chloride to obtain a target conductivity can then be estimated from

$$m_{\text{NaCl}} = \frac{\sigma_{\text{obj}} - 0.04195}{8.3845}, \quad (26)$$

or, equivalently, in terms of the desired resistivity,

$$m_{\text{NaCl}} = \frac{\frac{1}{\rho_{\text{obj}}} - 0.04195}{8.3845}. \quad (27)$$

Layer	Salt (g)	Resistivity ($\Omega \cdot m$)
Epidermis	–	23.84
Dermis	0.02	4.77
Hypodermis	0.0074	9.54

Table 11 – Salt composition in the different layers of the phantom

4.1.2 Electrical Resistivity Mimicking of Tissue Layers

The three proposed phantoms were designed to systematically investigate the effect of variations in the epidermal layer resistivity on the overall electrical response of the system.

Phantom 2 was constructed using the epidermal layer resistivity values reported in Table 11 at the surface. In Phantoms 1 and 3, the resistivities of layers 1 and 3 were kept constant, while the resistivity of layer 2 was modified in order to evaluate its effect on EIS measurements acquired from the surface using different electrode configurations.

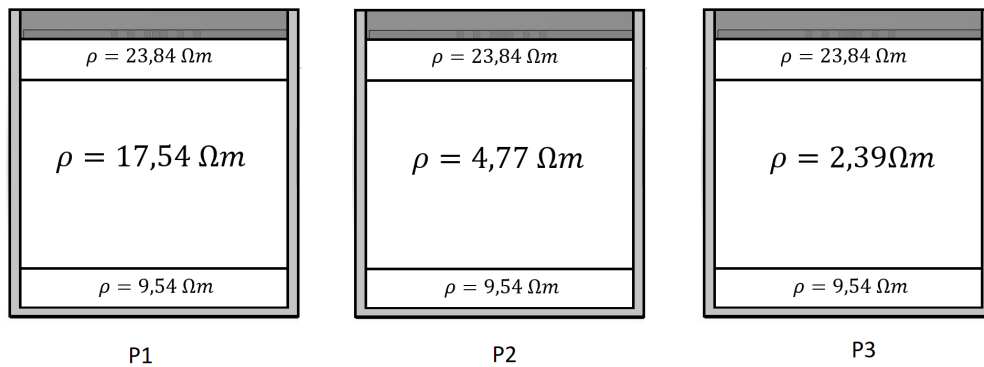


Figure 14 – Phantoms P1, P2, and P3 with the respective resistivity values of each layer.

Source: The Author

4.2 MEASUREMENT SYSTEM

Experimental data were acquired by means of Electrical Impedance Spectroscopy (EIS) using an EmStat4R potentiostat (PalmSens BV) operating in a four-electrode configuration. Two electrodes were used for current injection and two for voltage sensing, minimizing the influence of electrode–material contact impedance and improving the accuracy of the measured electrical properties.

Impedance measurements were performed over a frequency range from 200 Hz to 100 kHz. At each frequency, the impedance magnitude, phase angle, real impedance (Z'), and imaginary impedance (Z'') were automatically recorded.

The experimental protocol comprised three multilayer phantoms (P1, P2, and P3) evaluated using five electrode configurations, including three linear and two coplanar arrangements, resulting in a total of fifteen measurement sets. For each phantom–configuration combination, the impedance spectra were analyzed as a function of frequency using the impedance magnitude, phase angle, and the real and imaginary impedance components.

In addition, each phantom layer was individually characterized by EIS prior to the deposition of the subsequent layer. This procedure provided the characteristic electrical spectrum of each layer, serving as a reference for validating the electrical response of the complete multilayer phantom and assessing the contribution of each layer to the overall impedance response.

4.3 RESULTS

To evaluate the influence of dermis resistivity variations on the electrical response of the multilayer phantoms, the impedance spectra obtained for each electrode configuration were initially analyzed. The impedance magnitude and the real part of the impedance were compared as a function of frequency for the three proposed phantoms (P1, P2, and P3) to identify the changes produced exclusively by modifications in the dermal layer.

The epidermis and hypodermis resistivities were kept constant in all three phantoms at 23.84 Ωm and 9.54 Ωm , respectively. The only parameter modified was the dermis resistivity, which was set to 17.54 Ωm for Phantom P1, 4.77 Ωm for Phantom P2, and 2.39 Ωm for Phantom P3. Phantom P2 was adopted as the reference because its dermis resistivity corresponds to the value reported for human skin by Faes et al. (FAES et al., 1999), whereas Phantoms P1 and P3 represent conditions with higher and lower dermis resistivity, respectively.

Figures 16–20 present the impedance magnitude ($|Z|$) and real impedance (Z') spectra obtained using the three linear and two coplanar electrode configurations. For each configuration, the responses of Phantoms P1, P2, and P3 are presented over the frequency range from 200 Hz to 100 kHz, enabling a direct comparison of the sensitivity of each electrode configuration to the variations introduced in the dermis resistivity.

As observed in Figures 16 and 17, the Coplanar Arrangement 1 and Coplanar Arrangement 2 electrode configurations exhibited higher impedance magnitude and real impedance

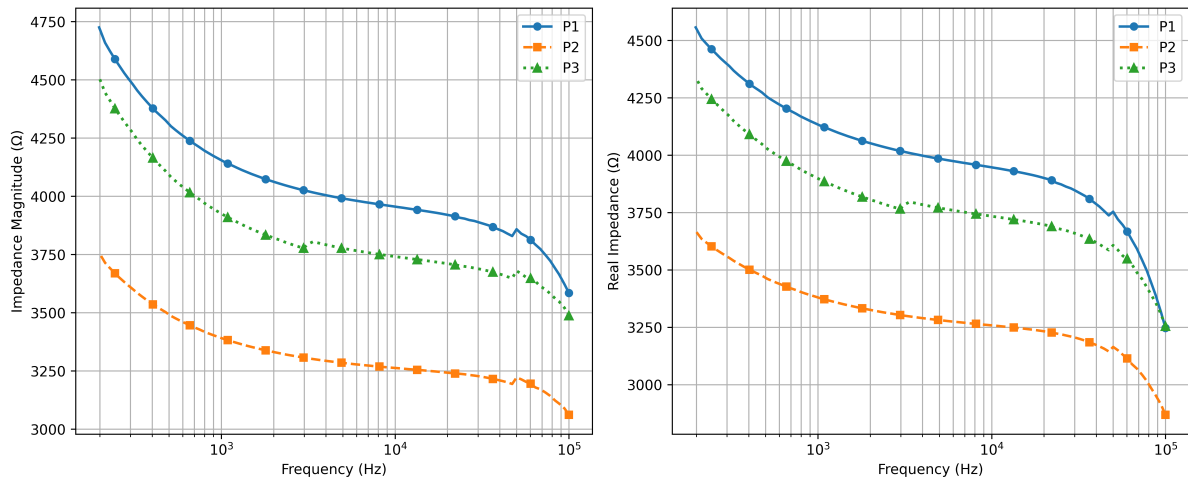


Figure 15 – Impedance magnitude ($|Z|$) and real impedance (Z') spectra of Phantoms P1, P2, and P3 obtained using the Coplanar Arrangement 1 electrode configuration over the frequency range from 200 Hz to 100 kHz. The figure illustrates the effect of epidermal resistivity variations on the measured electrical response.

Source: The Author

values for Phantoms P1 and P3 than for Phantom P2 over most of the analyzed frequency range. These differences remained nearly constant throughout the spectrum, with Phantom P1 consistently exhibiting slightly higher values than Phantom P3. However, the responses of the three phantoms gradually converged as the frequency increased, becoming more similar at frequencies close to 100 kHz.

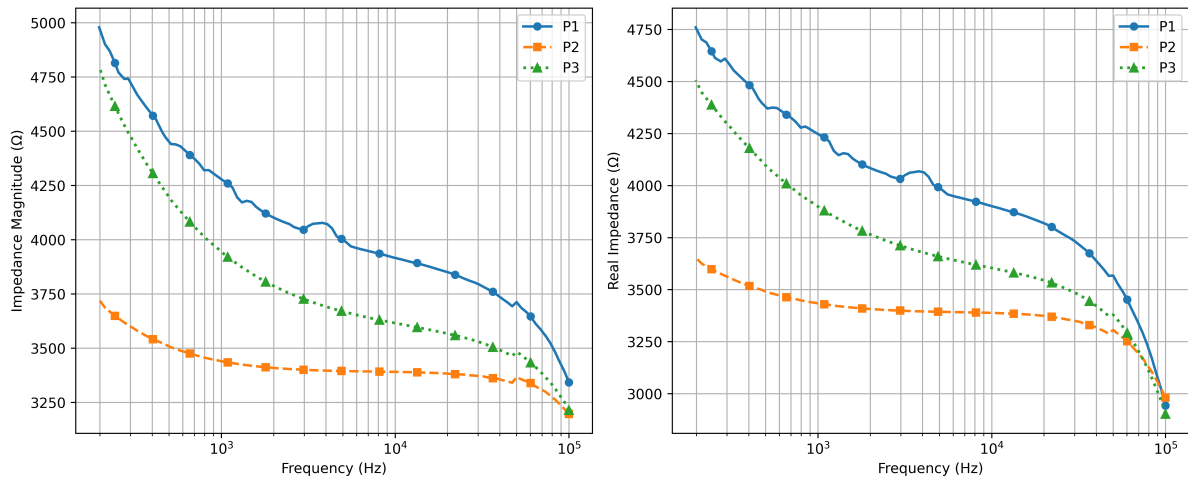


Figure 16 – Impedance magnitude ($|Z|$) and real impedance (Z') spectra of Phantoms P1, P2, and P3 obtained using the Coplanar Arrangement 2 electrode configuration over the frequency range from 200 Hz to 100 kHz. The figure illustrates the effect of epidermal resistivity variations on the measured electrical response.

Source: The Author

For the Linear Arrangement 1 electrode configuration, a nearly constant difference of less than 1000 Ω was observed between Phantoms P1 and P2 over the entire frequency range

for both the impedance magnitude and the real impedance. In contrast, Phantom P3 exhibited a markedly different behavior, with impedance magnitude and real impedance values approximately five to ten times higher than those of Phantoms P1 and P2 at low frequencies. As the frequency increased, both parameters decreased progressively, converging toward the responses of P1 and P2 at approximately 100 kHz.

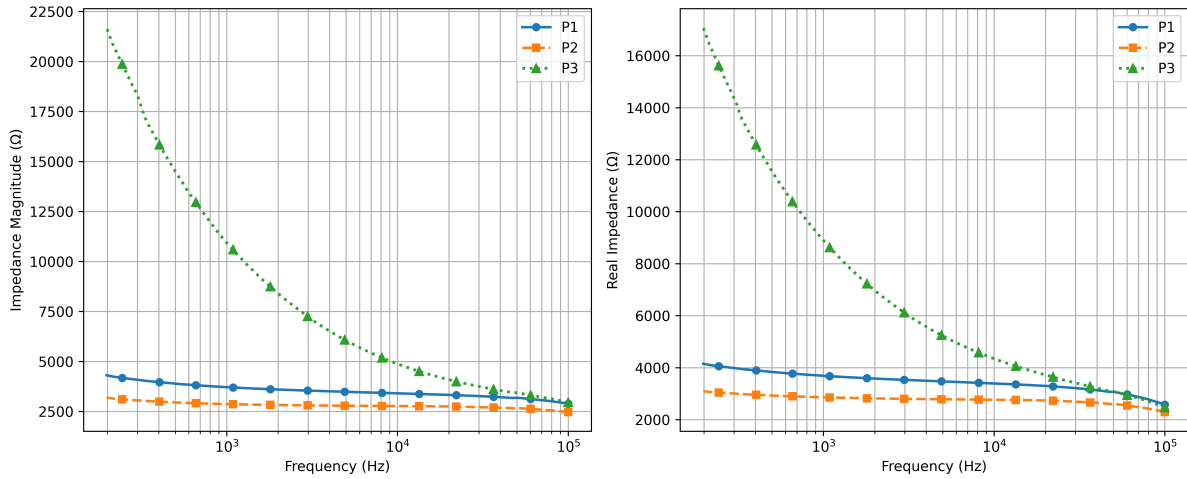


Figure 17 – Impedance magnitude ($|Z|$) and real impedance (Z') spectra of Phantoms P1, P2, and P3 obtained using the Linear Arrangement 1 electrode configuration over the frequency range from 200 Hz to 100 kHz. The figure illustrates the effect of epidermal resistivity variations on the measured electrical response.

Source: The Author

For the Linear Arrangement 2 electrode configuration, higher impedance magnitude and real impedance values were observed for Phantom P1 at the lower end of the frequency spectrum (200 Hz). In contrast, Phantoms P2 and P3 exhibited similar responses, maintaining an approximately constant separation throughout most of the analyzed frequency range despite the gradual decrease in impedance with increasing frequency. As the frequency approached the upper limit of the investigated spectrum, the responses of all three phantoms converged rapidly toward lower impedance values.

For the Linear Arrangement 3 electrode configuration, a nearly constant separation between the responses of the three phantoms was observed over the entire analyzed frequency range. Both the impedance magnitude and the real impedance exhibited a gradual decrease from 200 Hz to 100 kHz, with the three spectra following approximately parallel trends and maintaining a similar difference throughout the spectrum.

Since the epidermis resistivity was identical in all three phantoms, these results suggest that the Linear Arrangement 3 configuration is predominantly sensitive to variations in the electrical properties of the deeper layers. The nearly uniform separation between the spectra indicates that changes in dermis resistivity produce a proportional variation in the measured impedance over the investigated frequency range.

To complement the analysis of the impedance spectra, a statistical evaluation was sub-

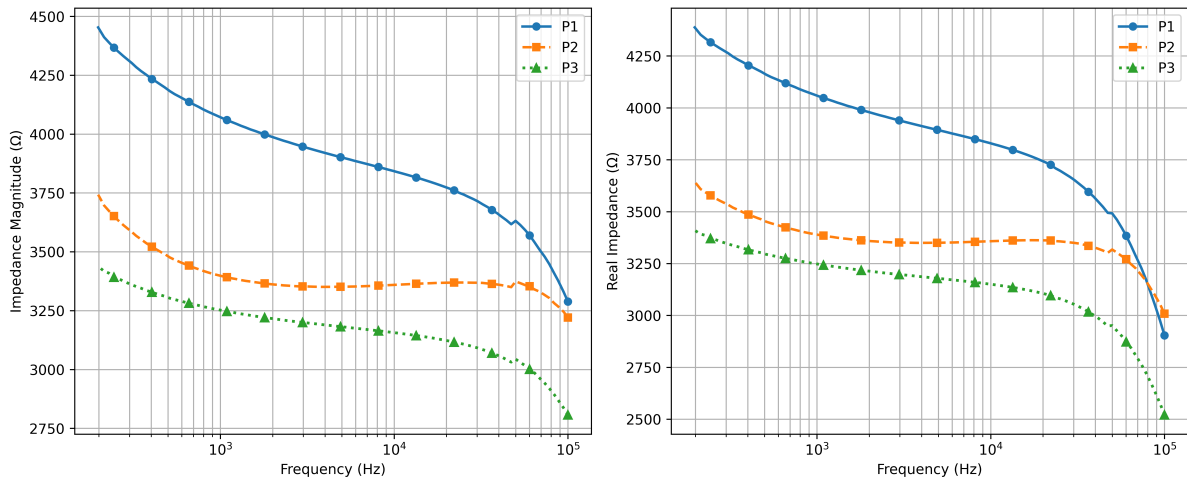


Figure 18 – Impedance magnitude ($|Z|$) and real impedance (Z') spectra of Phantoms P1, P2, and P3 obtained using the Linear Arrangement 2 electrode configuration over the frequency range from 200 Hz to 100 kHz. The figure illustrates the effect of epidermal resistivity variations on the measured electrical response.

Source: The Author

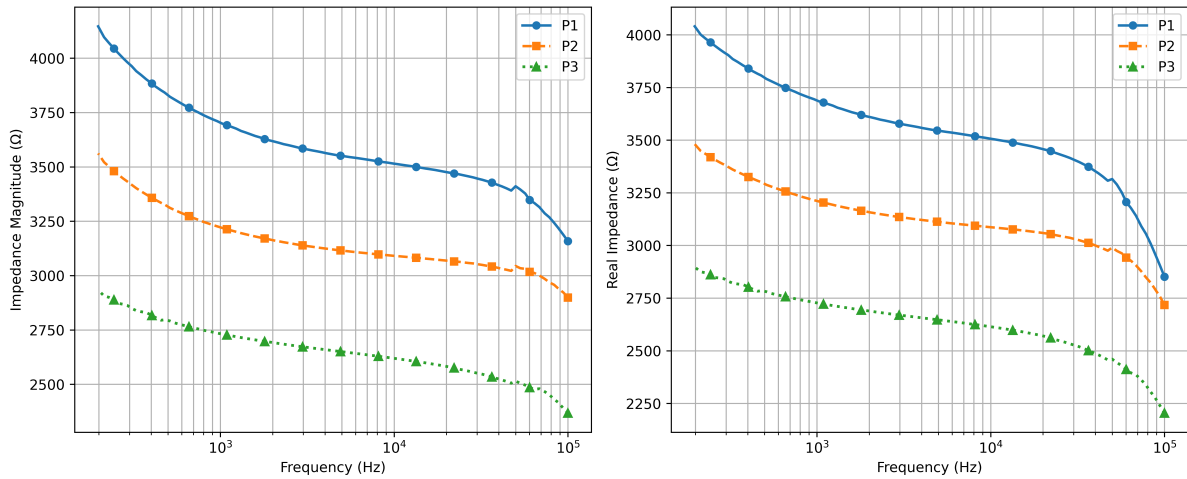


Figure 19 – Impedance magnitude ($|Z|$) and real impedance (Z') spectra of Phantoms P1, P2, and P3 obtained using the Linear Arrangement 3 electrode configuration over the frequency range from 200 Hz to 100 kHz. The figure illustrates the effect of epidermal resistivity variations on the measured electrical response.

Source: The Author

sequently performed using the relative difference (%RD) and Bland–Altman analysis. These methods allowed an objective quantification of the differences between the reference phantom (P2) and Phantoms P1 and P3, as well as an assessment of the agreement between the impedance measurements obtained with each electrode configuration (TRONSTAD; PRIPP, 2014).

4.3.1 Statistical Analysis

Figures 21–25 present the relative difference (%RD) and Bland–Altman plots for the five electrode configurations evaluated. The relative difference was calculated from the impedance magnitude by comparing the reference phantom (P2) with Phantoms P1 and P3, whereas the Bland–Altman analysis was used to assess the agreement between these comparisons over the frequency range from 200 Hz to 100 kHz (TRONSTAD; PRIPP, 2014).

The relative difference plots show the percentage variation in impedance magnitude as a function of frequency for the P1–P2 and P3–P2 comparisons. This analysis identifies the frequency intervals in which changes in dermis resistivity produced the largest deviations from the reference phantom, as well as the regions where the electrical response remained comparatively stable.

The Bland–Altman plots present the differences between paired impedance measurements as a function of their mean values, allowing the bias and limits of agreement to be evaluated for each electrode configuration. Together, these statistical analyses provide a quantitative assessment of the sensitivity of each electrode arrangement to the variations in dermis resistivity introduced in the multilayer phantoms (TRONSTAD; PRIPP, 2014).

It is important to emphasize that, in the context of the present study, the objective of the relative difference (%RD) analysis is not to minimize the relative difference, but to evaluate the sensitivity of each electrode configuration to changes in dermis resistivity. Consequently, a %RD value close to zero indicates that the electrode configuration is relatively insensitive to the modifications introduced in the inner layer. Conversely, a high and reproducible %RD indicates that the configuration is capable of detecting changes in dermis resistivity from surface impedance measurements. Likewise, a nearly constant %RD over the analyzed frequency range is also a desirable characteristic, as it reflects a stable and predictable response of the electrode configuration to variations in the electrical properties of the dermis, thereby facilitating the identification and quantification of subsurface changes.

Unlike the relative difference analysis, which quantifies the sensitivity of each electrode configuration to changes in dermis resistivity, the Bland–Altman analysis evaluates the consistency of these differences over the entire frequency range. In the present study, the objective is not to obtain a bias close to zero, but rather to determine whether the differences between the reference phantom (P2) and Phantoms P1 and P3 are systematic and reproducible. Therefore, a bias consistently different from zero, together with a compact distribution of the data and relatively narrow limits of agreement, indicates that the electrode configuration reliably detects changes in dermis resistivity from surface impedance measurements.

For the Coplanar Arrangement 1 electrode configuration, the relative difference (%RD) between Phantoms P1 and P2 decreased from approximately 22% at 200 Hz to 15% at 100 kHz. Similarly, the comparison between Phantoms P3 and P2 exhibited a gradual decrease from approximately 18% to 13% over the same frequency range. The smooth reduction in %RD with

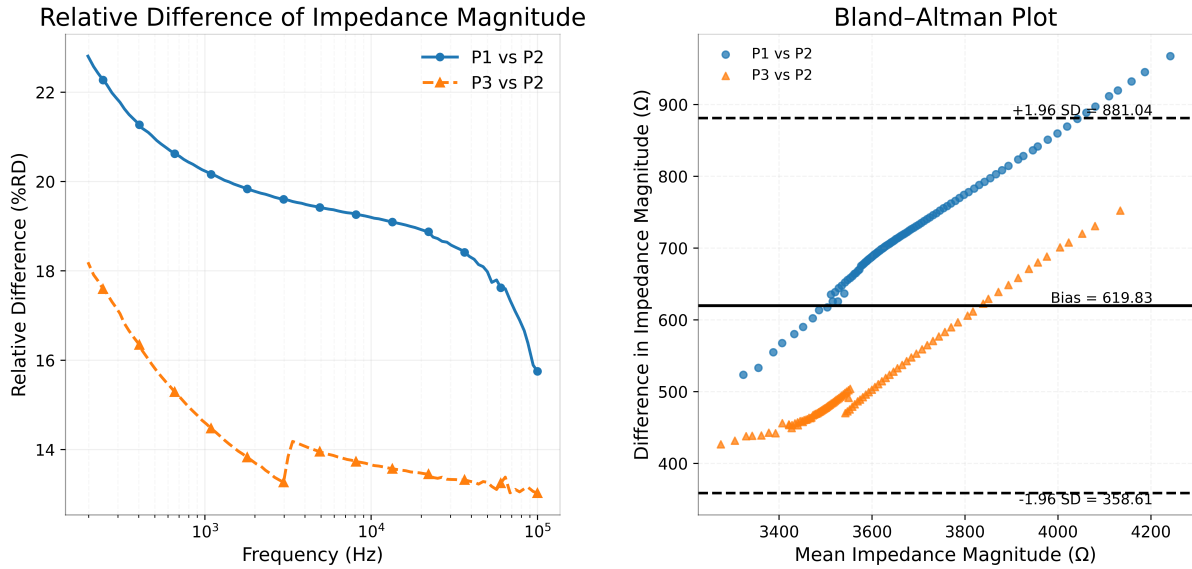


Figure 20 – Relative difference (%RD) and Bland–Altman analyses for the Coplanar Arrangement 1 electrode configuration. The left panel shows the relative difference in impedance magnitude for the P1–P2 and P3–P2 comparisons, while the right panel presents the corresponding Bland–Altman plots.

Source: The Author

increasing frequency indicates that the impedance responses of the three phantoms became progressively more similar at higher frequencies while maintaining a consistent separation throughout the spectrum.

The corresponding Bland–Altman plot shows two well-defined ascending clusters associated with the P1–P2 and P3–P2 comparisons, with differences ranging from approximately 400 Ω to more than 1000 Ω . The calculated bias of 619.83 Ω indicates a systematic difference between the reference phantom (P2) and the modified phantoms. Furthermore, the compact distribution of the data around each trend suggests that these differences remain consistent over the analyzed frequency range, demonstrating that the Coplanar Arrangement 1 configuration is capable of reliably distinguishing variations in dermis resistivity.

For the Coplanar Arrangement 2 electrode configuration, a behavior similar to that observed for Coplanar Arrangement 1 was obtained. The Bland–Altman analysis yielded a bias of 511.89 Ω , indicating a systematic difference between the reference phantom (P2) and the modified phantoms. The data points formed two closely spaced ascending clusters, with differences ranging from approximately 0 Ω to 1200 Ω , demonstrating a consistent trend over the analyzed frequency range.

The relative difference (%RD) also exhibited a gradual decrease with increasing frequency for both the P1–P2 and P3–P2 comparisons. As the frequency approached the upper limit of the analyzed spectrum, the relative differences tended to decrease toward zero, indicating that the electrical responses of the three phantoms became progressively more similar above 100 kHz.

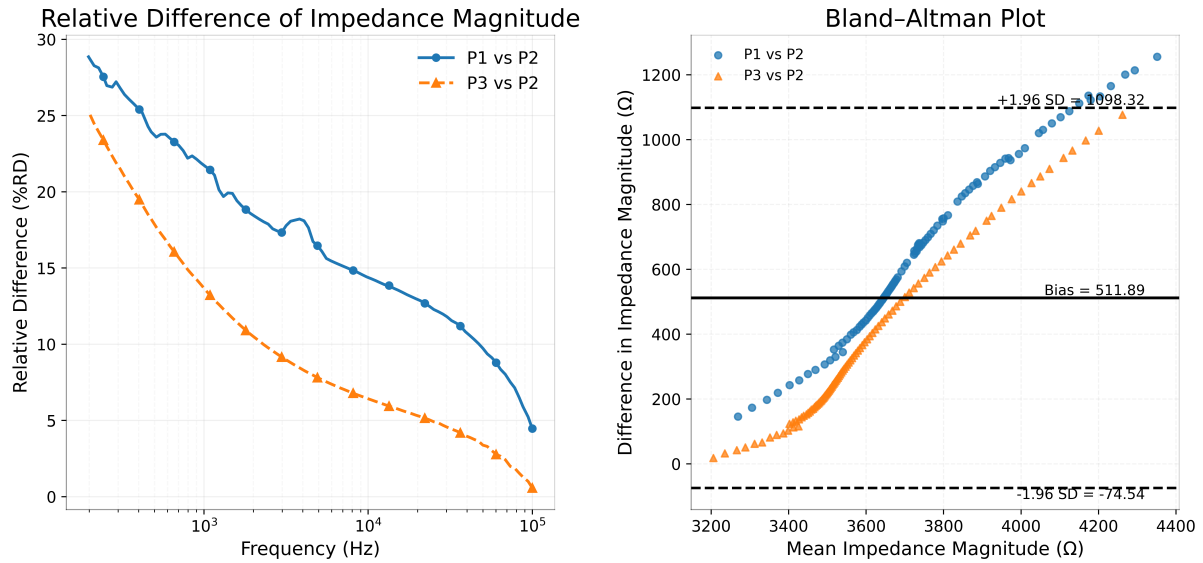


Figure 21 – Relative difference (%RD) and Bland–Altman analyses for the Coplanar Arrangement 2 electrode configuration. The left panel shows the relative difference in impedance magnitude for the P1–P2 and P3–P2 comparisons, while the right panel presents the corresponding Bland–Altman plots.

Source: The Author

For the Linear Arrangement 1 electrode configuration, the %RD between Phantoms P1 and P2 remained relatively low and nearly constant, decreasing slightly from approximately 30% to 20% over the analyzed frequency range. In contrast, the P3–P2 comparison exhibited a pronounced descending trend, with the %RD decreasing from approximately 140% at 200 Hz to values close to 20% at 100 kHz, where it converged toward the P1–P2 comparison.

The corresponding Bland–Altman plot clearly distinguishes the two phantom comparisons. The P1–P2 data points form a compact cluster below the bias, concentrated at lower mean impedance values and characterized by relatively small differences. In contrast, the P3–P2 comparison follows a marked ascending trend, with the differences increasing from approximately 0 Ω to values exceeding 15 kΩ. The calculated bias of 3062.96 Ω, together with the wide separation between the two clusters, indicates that the Linear Arrangement 1 configuration is highly sensitive to the modifications introduced in the dermis resistivity, particularly for Phantom P3.

For the Linear Arrangement 2 electrode configuration, the %RD between Phantoms P1 and P2 remained relatively low, decreasing from approximately 17% to 2% over the analyzed frequency range. An even smaller relative difference was observed for the P3–P2 comparison, varying between approximately 8% and 14%. These results indicate that the Linear Arrangement 2 configuration exhibits a limited sensitivity to the variations in dermis resistivity, particularly for the P3–P2 comparison.

The corresponding Bland–Altman plot shows a clear separation between the two phantom comparisons. The P3–P2 data are predominantly distributed below the zero-difference line and, consequently, below the calculated bias of 148.28 Ω, reflecting the lower impedance of

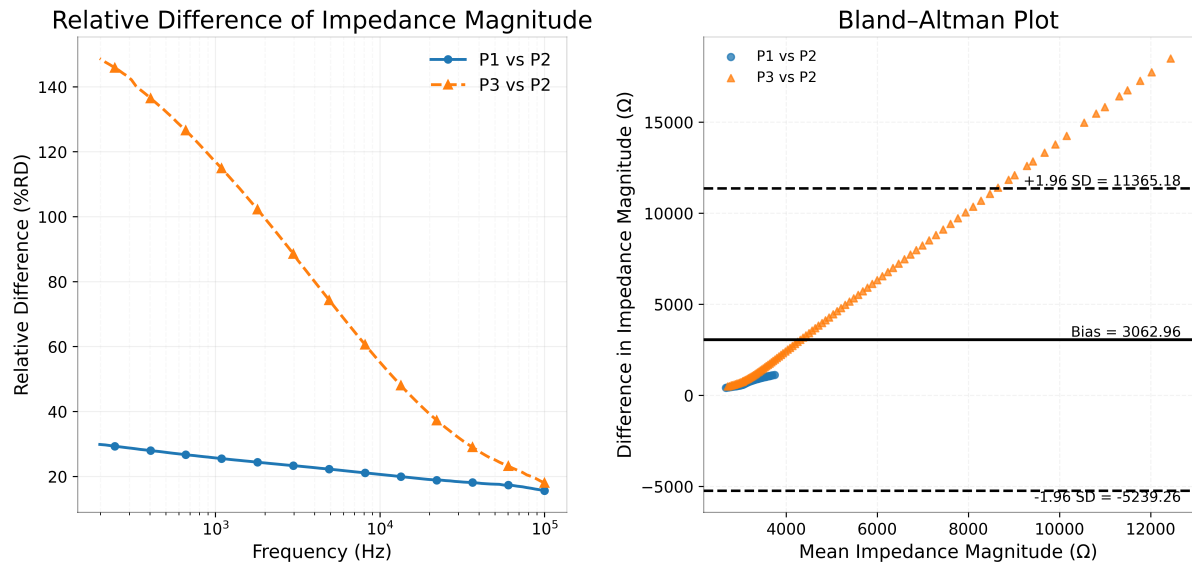


Figure 22 – Relative difference (%RD) and Bland–Altman analyses for the Linear Arrangement 1 electrode configuration. The left panel shows the relative difference in impedance magnitude for the P1–P2 and P3–P2 comparisons, while the right panel presents the corresponding Bland–Altman plots.

Source: The Author

Phantom P3 resulting from its higher dermis conductivity. In contrast, the P1–P2 comparison is located almost entirely above the bias and is concentrated over mean impedance values ranging from approximately 3200 Ω to 4000 Ω . The distinct distribution of both clusters indicates that the Linear Arrangement 2 configuration consistently differentiates between increases and

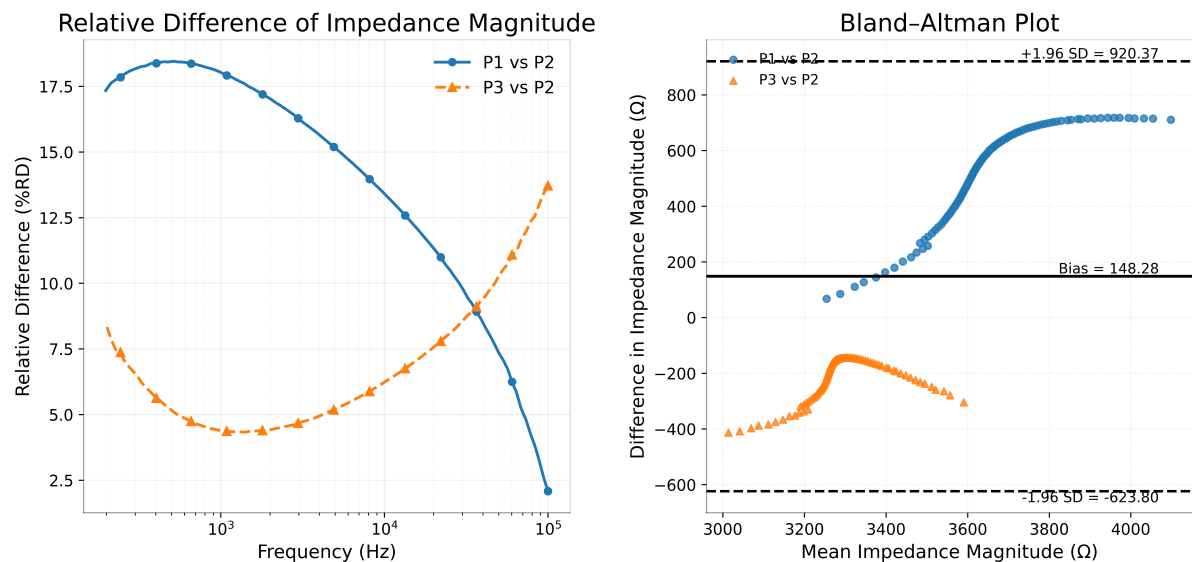


Figure 23 – Relative difference (%RD) and Bland–Altman analyses for the Linear Arrangement 2 electrode configuration. The left panel shows the relative difference in impedance magnitude for the P1–P2 and P3–P2 comparisons, while the right panel presents the corresponding Bland–Altman plots.

Source: The Author

decreases in dermis resistivity with respect to the reference phantom.

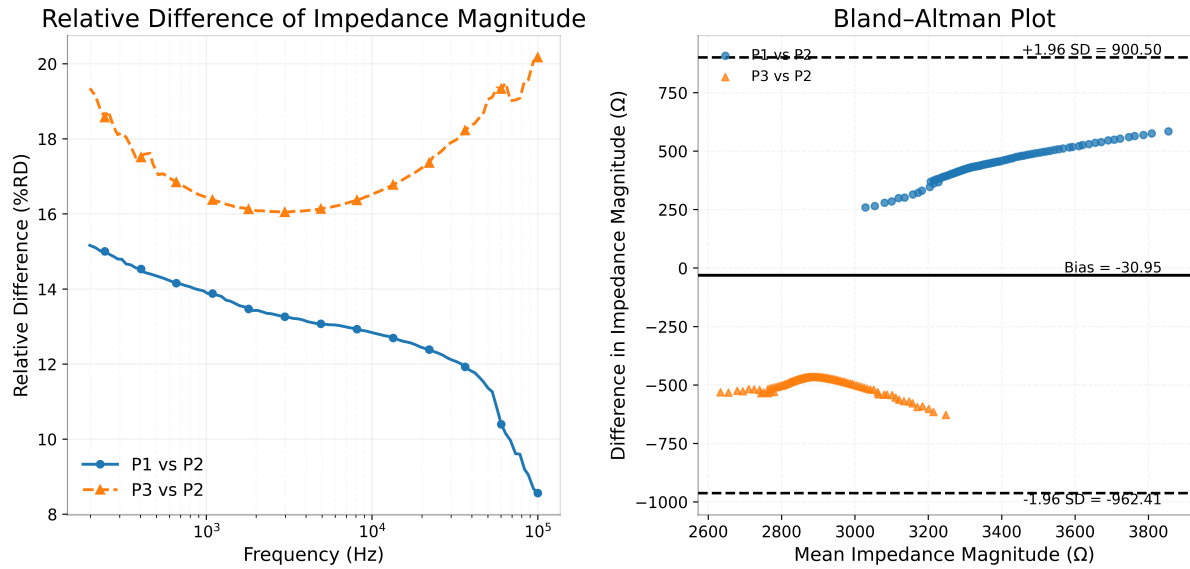


Figure 24 – Relative difference (%RD) and Bland–Altman analyses for the Linear Arrangement 3 electrode configuration. The left panel shows the relative difference in impedance magnitude for the P1–P2 and P3–P2 comparisons, while the right panel presents the corresponding Bland–Altman plots.

Source: The Author

For the Linear Arrangement 3 electrode configuration, the relative difference remained below 20% throughout the analyzed frequency range. The P1–P2 comparison exhibited a gradual decrease, tending toward zero at higher frequencies, whereas the P3–P2 comparison showed the opposite behavior, with the relative difference increasing above approximately 10 kHz. This divergence between the two comparisons at higher frequencies indicates that the Linear Arrangement 3 configuration is capable of distinguishing changes in dermis resistivity, particularly in the upper portion of the analyzed spectrum.

The corresponding Bland–Altman plot yielded a bias of -30.95Ω . Although the overall bias was close to zero, this result reflects the compensation between the positive differences observed for the P1–P2 comparison and the negative differences associated with the P3–P2 comparison, rather than the absence of systematic differences. The P1–P2 data form a nearly constant positive cluster, with differences ranging from approximately 250Ω to 600Ω , whereas the P3–P2 comparison remains consistently below the zero-difference line, with negative differences spanning approximately 150Ω . The clear separation between the two clusters demonstrates that the Linear Arrangement 3 configuration consistently discriminates between increases and decreases in dermis resistivity relative to the reference phantom.

5 DISCUSSION, CONCLUSIONS, AND FUTURE WORK

5.1 DISCUSSION AND CONCLUSIONS

The calibration procedure demonstrated that agar and gelatin exhibit markedly different electrical behaviors. Agar-based phantoms present a high ionic conductivity, making them suitable for reproducing low-resistivity tissues. In contrast, gelatin provides a considerably wider resistivity range, particularly for high-resistivity tissues, while maintaining adequate mechanical stability even at relatively low gelatin concentrations. This characteristic makes gelatin a more versatile material for reproducing the passive electrical properties of multilayer biological tissues.

It was also observed that the intrinsic ionic content varies among commercial gelatin brands, leading to differences in their electrical conductivity even when identical preparation protocols are employed. Consequently, the same gelatin brand should be used throughout an experimental study, and the complete electrical calibration procedure should be repeated whenever a different commercial formulation is selected. Likewise, temperature calibration proved to be an essential step in phantom fabrication, since the electrical resistivity of gelatin-based materials strongly depends on temperature. The proposed calibration methodology provides the appropriate combinations of NaCl concentration, gelatin concentration, and operating temperature required to reproduce the target resistivity values of each biological layer with high repeatability.

The main objective of this work was to evaluate the sensitivity of different tetrapolar electrode configurations to detect variations in dermis resistivity using electrical impedance spectroscopy performed from the phantom surface. Three multilayer phantoms were therefore developed by maintaining constant the electrical properties of the epidermis and hypodermis while modifying only the resistivity of the dermis. This approach isolated the contribution of a single layer to the overall electrical response, enabling an objective comparison of the sensitivity of the proposed electrode configurations.

The experimental results demonstrated that the electrode geometry significantly influences the measured impedance spectra. Although all configurations were capable of distinguishing the three phantoms, the magnitude and frequency dependence of the observed differences varied substantially. Some electrode arrangements exhibited nearly constant differences over the entire frequency range, whereas others showed pronounced frequency-dependent behavior. These findings indicate that the current distribution generated by each tetrapolar configuration modifies the effective sensing depth and, consequently, the relative contribution of each layer to the measured impedance.

The coplanar configurations exhibited a relatively uniform response throughout the analyzed spectrum, characterized by moderate relative differences and a gradual convergence of the impedance spectra as frequency increased. This behavior suggests that these configurations provide stable measurements while presenting a moderate sensitivity to changes in dermis re-

sistivity. In contrast, the linear configurations displayed markedly different performances. In particular, Linear Arrangement 1 produced the largest separation between the phantoms, with relative differences exceeding 100% for the P3–P2 comparison at low frequencies and a clear separation of the corresponding Bland–Altman clusters. These results indicate that this configuration offers the highest sensitivity for detecting changes in dermis resistivity from surface measurements.

Linear Arrangement 2 showed considerably smaller relative differences while maintaining well-defined and reproducible Bland–Altman distributions. Although the magnitude of the differences was reduced, the measurements remained highly consistent throughout the investigated frequency range. Linear Arrangement 3 exhibited a distinct behavior, with nearly constant differences over most of the spectrum followed by an increasing divergence above approximately 10 kHz. This frequency-dependent response suggests that the sensitivity of this configuration is strongly influenced by the excitation frequency, indicating that specific spectral regions may be more suitable for detecting electrical changes occurring in deeper tissue layers.

The combined use of %RD and Bland–Altman analysis provided complementary information regarding the performance of each electrode configuration. While the %RD quantified the sensitivity of each arrangement to variations in dermis resistivity, the Bland–Altman analysis demonstrated whether these differences remained systematic and reproducible throughout the analyzed frequency range. In the context of the present work, a large %RD should not be interpreted as an experimental error, but rather as an indication of increased sensitivity to changes occurring in the dermis. Similarly, Bland–Altman plots characterized by compact point distributions and well-defined bias values indicate that the observed differences are systematic rather than resulting from random experimental variability.

From a physical perspective, the observed differences can be explained by the current density distribution generated by each tetrapolar electrode geometry. The spacing and arrangement of the electrodes determine the sensing volume, thereby controlling the relative contribution of the epidermis, dermis, and hypodermis to the measured impedance. Consequently, relatively small modifications in electrode geometry produce substantial differences in sensitivity to subsurface electrical properties. This interpretation is consistent with previous numerical and experimental studies on electrical impedance sensitivity, which have demonstrated the strong dependence of current pathways on electrode configuration.

Overall, the proposed multilayer phantoms provide a reliable experimental platform for evaluating electrode configurations intended for superficial bioimpedance applications. The methodology enables selective modification of the electrical properties of individual layers while preserving the remaining structure, thereby providing a controlled environment for sensitivity assessment. These findings constitute an important step toward the development and optimization of electrical impedance spectroscopy systems for the non-invasive characterization of superficial tissues and the detection of localized changes within the skin.

5.1.1 Articles

- *Consistency Measurements of Skin Impedance using Different Electrode Types*. This article was accepted and presented at the 2026 IEEE International Symposium on Medical Measurements and Applications (MeMeA 2026), Montevideo, Uruguay. It reports in vivo experiments performed on human skin using different electrode types, together with a robust statistical analysis to evaluate measurement repeatability. Additionally, comparisons were conducted among three distinct measurement sites within the same individual and three different inter electrode configurations to assess the influence of electrode arrangement on impedance consistency. DOI: <https://doi.org/10.1109/MeMeA69746.2026.11537313>.
- *A Preliminary Study on a Device to Test Skin Bioelectrical Impedance Characteristics Related to Acupuncture Theory*. This article was accepted and presented at the 2026 IEEE International Symposium on Medical Measurements and Applications (MeMeA 2026), Montevideo, Uruguay. The study was conducted in collaboration with two research groups to validate a novel device for measuring skin bioelectrical impedance. The primary contributions of the author focused on the statistical design of the experimental protocol and the analysis of the acquired data, supporting the assessment of the device's performance and measurement reliability. DOI: <https://doi.org/10.1109/MeMeA69746.2026.11537345>.
- *Biological Skin Phantom for Electrical Impedance Spectroscopy*. This manuscript was submitted to the XXX Congresso Brasileiro de Engenharia Biomédica (CBEB 2026) and is currently awaiting an editorial decision. The study is based on the case studies developed in Chapters 4 and 5 of this thesis, focusing on the design, fabrication, and electrical characterization of multilayer skin phantoms intended to reproduce the passive electrical properties of human skin for electrical impedance spectroscopy experiments.

5.2 FUTURE WORK

Future work will focus on strengthening the physical and electrical representation of the developed phantoms through numerical simulations, optimization of the materials employed in their fabrication, and the experimental validation of new measurement configurations. In particular, finite element simulations using COMSOL Multiphysics are proposed to model the distribution of the electric field within multilayer structures, allowing a more detailed analysis of the contribution of each layer to the overall impedance response. These simulations will also facilitate the validation of different electrode configurations, sensitivity studies, and the optimization of experimental protocols under varying geometrical and electrical conditions.

Another aspect of interest concerns the incorporation of polymer blends into phantom formulations to expand the achievable range of resistivities and improve the fine control of

their electrical and mechanical properties. The combination of different polymeric matrices could enable a more precise adjustment of the material response, reproducing not only the average passive electrical properties of biological tissues but also conductivity gradients and heterogeneous distributions present in human skin.

Additionally, the inclusion of insulating spheres or spherical inclusions capable of emulating different capacitance values is proposed. These structures could reproduce localized electrical heterogeneities associated with the phenomenon of field cancerization, enabling the study of lesions with varying sizes, depths, developmental stages, and spatial distributions within the multilayer medium. Such an approach may contribute to a better understanding of the sensitivity of electrical bioimpedance measurements to premalignant alterations and diffuse tissue changes occurring in regions surrounding cutaneous neoplasms.

Finally, future investigations should include *in vivo* studies involving healthy volunteers while considering physiological variables such as sex, age, hydration status, skin phototype, and anatomical measurement site. The acquisition and analysis of these data could provide a framework for parameterizing the electrical characteristics of different skin types and physiological conditions, contributing to the establishment of reference values and promoting the development of bioimpedance-based diagnostic tools for the non-invasive assessment of cutaneous tissues.

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APPENDIX A – CONSISTENCY MEASUREMENTS OF SKIN EIS

This appendix contains a copy of the paper entitled *Consistency Measurements of Skin Impedance Using Different Electrode Types*, presented at the 2026 IEEE International Symposium on Medical Measurements and Applications (MeMeA 2026), Montevideo, Uruguay. .

Consistency Measurements of Skin Impedance using Different Electrode Types

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Abstract—Electrical Impedance Spectroscopy (EIS) has emerged as a promising non-invasive technique for skin characterization and early detection of dermatological pathologies. However, the reliability and repeatability of EIS measurements are strongly influenced by the electrode–skin interface, particularly by electrode material and geometry. This work presents an experimental evaluation of the variability and repeatability of skin impedance measurements acquired using three different electrode types. Impedance spectra were recorded at three distinct locations on the biceps brachii of a healthy male subject under carefully controlled environmental conditions. For each electrode and measurement site, ten consecutive measurements were performed over a frequency range from 10 Hz to 100 kHz using identical acquisition protocols. Statistical analyses were conducted to assess intra-point repeatability, site-level consistency and inter-electrode variability, including tests for normality, homoscedasticity and non-parametric comparisons. The results showed weak statistically significant differences in the impedance modulus among the electrode configurations, resulting in a significance level of 0.05. These findings highlight the impact of electrode characteristics on measurement dispersion and support the importance of standardized electrode selection in EIS-based skin assessment to improve measurement comparability and reproducibility across experimental setups and future clinical or research-oriented EIS applications.

Index Terms—Electrical Impedance Spectroscopy, Bioimpedance Measurements, Electrode Materials, Electrode–Skin Contact, Measurement Variability, Medical Instrumentation

I. INTRODUCTION

The skin is the largest organ of the human body and performs multiple functions that are essential for homeostasis. It primarily acts as a physical protective barrier against external physical, chemical, and biological agents [1][2]. It is constantly exposed to harmful agents, many of which are associated with skin cancer [3]; moreover, several studies have shown that its early detection significantly reduces mortality [4][5][6]. This has generated significant interest in the development of minimally invasive tools that assist in the early detection of skin pathologies and can be used in routine dermatological practice. These tools substantially reduce the need for skin biopsies, minimize the risk of false positives, and enable the immediate delivery of test results[7][8]. Electrical Impedance Spectroscopy (EIS) is a non-invasive method with the potential to detect and confirm premalignant or malignant

skin lesions, as well as early epigenetic alterations associated with the field of carcinogenesis [9][10][11]. EIS measurements in healthy skin and in skin with malignant lesions, such as Basal Cell Carcinoma (BCC), Squamous Cell Carcinoma (SCC), and melanoma, are considered to differ due to changes in cell size, shape, and orientation, as well as alterations in the structure and functional state of cell membranes[9]. These changes alter the conductivity and capacitance of the tissue, allowing EIS to non-invasively detect differences between healthy skin and malignant lesions, which can be quantified based on the passive electrical properties of the tissues [9][11][12]. Interindividual variability in the electrical properties of the skin, influenced by factors such as age, sex, epidermal thickness, and hydration level, limits the widespread implementation of Electrical Impedance Spectroscopy (EIS), as it complicates the standardization of protocols and the interpretation of results [12][13][14].

The electrical properties of the skin vary between individuals and across different anatomical regions, influencing the consistency and comparability of bioimpedance measurements under similar experimental conditions [12][14]. Factors such as hydration level, skin temperature, and local blood perfusion may also introduce variability in the measured impedance. Likewise, the electrode material, geometry, and surface characteristics play a crucial role in determining the accuracy and precision of the acquired data, as they directly affect the electrode–skin interface and the resulting contact impedance [15].

The variability of impedance readings was evaluated at three specific points on the biceps of 1 male participant under controlled environmental conditions. Three different types of electrodes—each with distinct material composition and surface area—were used, and measurements were obtained at each of the three designated locations. For every measurement point, ten consecutive readings were recorded to quantify the dispersion, repeatability, and temporal stability of the impedance values. The collected data were processed to identify potential deviations associated with electrode type and measurement site. The experimental protocol was designed and executed following methodological recommendations widely adopted in bioimpedance research [16].

The analysis was performed in a hierarchical manner.

First, the correlation among the ten consecutive readings at each measurement point was evaluated to assess intra-point consistency. Finally, the average impedance across the three points on the biceps was calculated to evaluate site-level consistency for each individual. All analyses were conducted within the frequency band of interest, where the electrical properties of skin layers are most relevant for EIS applications. Additionally, statistical tools were applied to assess the normality and homoscedasticity of the measurements, ensuring the validity of subsequent comparative analyses[16][17][18]. Were computed to evaluate the measurement consistency across repetitions, electrode types, and measurement points. This analytical approach enables the identification of the effects of electrode position, contact quality, and biological variability on the recorded impedance values. The results help establish a quantitative criterion for assessing measurement reliability and reproducibility, thereby supporting the future clinical implementation of Electrical Impedance Spectroscopy (EIS) in skin evaluation.

The main objective is to quantitatively assess the repeatability and reliability of the skin by using EIS measurements. Also, the impact of both electrode characteristics and anatomical measurement site variability were investigated.

II. METHODOLOGY

A. Measurements Procedure

A single healthy male subject was evaluated at three specific points (P1, P2, and P3) along the transverse axis of the biceps brachii of the dominant arm. The subject was the author of this study. Measurements were performed with the elbow flexed at 90° and the wrist in a neutral position. Laboratory temperature was maintained at approximately 22 °C throughout all experiments in order to minimize thermal variability. The biceps region was selected due to its high reproducibility in impedance measurements [19].

An excitation amplitude of 0.1 V was applied to avoid electroporation of cell membranes, according to the standard IEC 60601-1. This low voltage level confines the current path between the electrodes, directing most of the electric field through the highly conductive dermal layer [20].

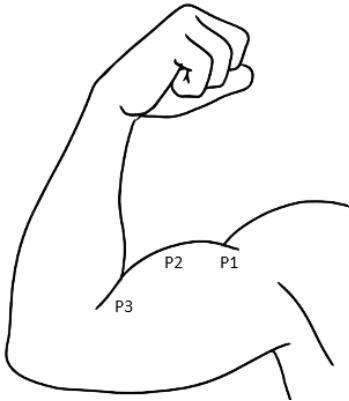


Fig. 1. Measuring points sites 1/2/3 along the central axis of the biceps.

Impedance spectra were acquired using the EmStat4R potentiostat in two-electrode mode, within a frequency range of 10 Hz to 100 kHz. Each point was measured ten times (R1–R10) with three different type of electrode (E1–E3), resulting in 90 readings per frequency (1 subject $\times$ 3 points $\times$ 10 readings $\times$ 3 electrode types).

S1 is a Descart ECG Hypoallergenic Neonatal Electrode 2258 3M composed of acrylic gel with KCl and a stainless-steel pin with ABS polymer counter-pin coated in Ag/AgCl. S2 is an adhesive bioelectrode for electro-stimulation (TENS). Finally, S3 is a commercial precordial suction electrodes featuring Ag/AgCl-plated contact surfaces and universal pear-shaped cardioclips. In order to reduce measurement variability, the skin was gently cleaned with an isotonic non-alcoholic solution before electrode placement. All recordings were performed by the same operator for minimizing procedural bias.

B. Data Processing

Both real (R) and imaginary (X_c) part of the impedance (Z) were measured and sort by participant, point, reading, electrode type, and frequency. Each dataset was preprocessed using Python (v3.11) to ensure consistent frequency sampling, to remove incomplete spectra, and to correct potential outliers exceeding ± 2 standard deviations from the group mean. Python scripts were also used to automate the dataset generation and to convert impedance readings into matrices for offline statistical evaluation.

Data analysis was performed in two complementary stages using Python and R (v4.3.2). First, in Python, Nyquist and Bode plots were generated for each electrode type and subject in order to visualize the dispersion of impedance spectra across frequencies. The plots allows the identification of patterns associated with measurement stability, electrode contact, and frequency-dependent deviations. The dispersion of readings was quantified by calculating intra- and inter-point variance for each frequency.

The statistical processing focused on comparative analysis between electrodes, measurement sites, and individuals. Descriptive statistics (mean, standard deviation, and coefficient of variation) were computed for each frequency and electrode configuration. Dispersion patterns were evaluated through box-plots and variance profiles, enabling a visual and quantitative assessment of consistency across electrodes.

Impedance modulus and phase angle were normalized to minimize inter-subject baseline differences. The impedance matrices (frequency $\times$ measurement) were structured for multilevel comparison at three hierarchical levels, such as:

- Intra-point variability to exam the dispersion between consecutive readings (R1–R10);
- Intra-subject variability to analyze the consistency across points (P1–P3); and
- Inter-electrode variability to compare the impedance parameters among electrode types (S1–S3).

Figure 3 shows Nyquist plots for the impedance spectra at measurement points P1, P2, and P3, with mean resistance values of 1957.2 for P1, 1919.6 for P2, and 1849.1 for P3. The differences are statistically significant. Due to non-normality and heteroscedasticity of the reactance data (Shapiro–Wilk p -values of 5.38×10^{-8} for P1, 9.71×10^{-25} for P2, and

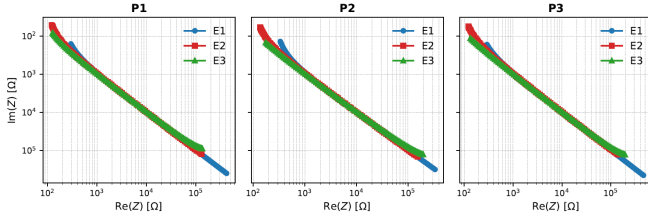


Fig. 3. Nyquist plots of the mean impedance spectra, separated by measurement point (P1–P3).

1.06×10^{-13} for P3), the Kruskal–Wallis test was applied, yielding $H = 21.56$ with a p -value of 2.08×10^{-5} . Post hoc pairwise comparisons were performed using Bonferroni or Tamhane T2 tests according to data properties, confirming statistically significant differences among the measurement points.

A comparative statistical analysis across the three measurement points was performed for the impedance modulus Z after averaging the ten repeated measurements at each frequency. The Shapiro–Wilk test indicated deviations from normality in several frequencies (global $p < 0.01$), and the Levene test revealed unequal variances across points ($p = 0.032$). Consequently, the Kruskal–Wallis test was applied ($H = 18.74$, $p = 8.6 \times 10^{-5}$), indicating differences among measurement sites. Post hoc pairwise comparisons were conducted according to the data properties: Tukey test for normal and homoscedastic data, Bonferroni test for non-normal but homoscedastic data, and Tamhane T2 test for non-normal and heteroscedastic data. Significant differences were found between P1 and P3 ($p = 1.2 \times 10^{-4}$) and between P2 and P3 ($p = 3.7 \times 10^{-3}$), while no difference was observed between P1 and P2 ($p = 0.081$). The trend in impedance magnitude was $P1 > P2 > P3$.

C. Comparison Between Electrode Types

Across electrode types, S2 showed a more stable statistical behavior than S1 and S3. For the impedance magnitude (Z), S2 presented 88 normal cases (88%) and 11 non-normal cases (12%), with Shapiro–Wilk p -values ranging from 0.12 to 0.95 and Levene test $p = 0.08$, indicating homoscedasticity. S1 showed 75 normal cases (75%) and 24 non-normal cases (24%), with Shapiro–Wilk p -values between 0.03 and 0.91 and Levene test $p = 0.04$. Electrode S3 exhibited 70 normal cases (70%) and 29 non-normal cases (29%), with Shapiro–Wilk p -values ranging from 0.001 to 0.87 and Levene test $p = 0.02$, indicating deviations from normality and unequal variances compared with S1 and S2.

The same trend was observed in the resistive component (R), with S2 achieving 82 normal cases (82%, Shapiro–Wilk $p = 0.07$ –0.92, Levene $p = 0.10$), S1 52 normal cases (52%, Shapiro–Wilk $p = 0.02$ –0.89, Levene $p = 0.03$), and S3 51 normal cases (51%, Shapiro–Wilk $p = 0.001$ –0.85, Levene $p = 0.01$). For the reactive component (X_c), all electrodes demonstrated high normality, with S2 at 91 normal cases

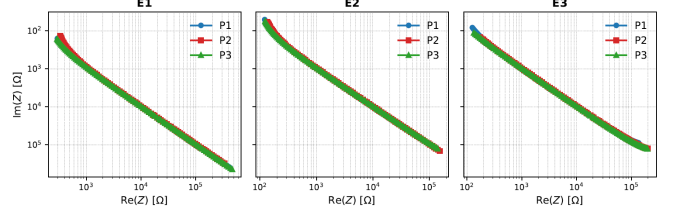


Fig. 4. Nyquist plots of the mean impedance spectra, separated by measurement sensor (E1–E3).

TABLE II
STABILITY SUMMARY ACROSS ELECTRODE TYPES

Electrode	Variable	Normality (%)	Shapiro–Wilk p	Levene p
S1	Z	75	0.03–0.91	0.04
	R	52	0.02–0.89	0.03
	X_c	89	0.05–0.95	–
S2	Z	88	0.12–0.95	0.08
	R	82	0.07–0.92	0.10
	X_c	91	0.14–0.97	–
S3	Z	70	0.001–0.87	0.02
	R	51	0.001–0.85	0.01
	X_c	85	0.02–0.92	–

(91%, Shapiro–Wilk $p = 0.14$ –0.97), S1 at 89 (89%, Shapiro–Wilk $p = 0.05$ –0.95), and S3 at 85 (85%, Shapiro–Wilk $p = 0.02$ –0.92).

Overall, ANOVA applied to normal and homoscedastic data (S2) showed significant differences across points ($F = 5.21$, $p = 0.009$), Kruskal–Wallis for non-normal data (S3) yielded $H = 14.36$, $p = 0.001$, and post hoc comparisons using Tukey, Bonferroni, and Tamhane T2 tests confirmed a hierarchical stability ranking of $S2 > S1 > S3$.

IV. DISCUSSIONS AND CONCLUSION

The analysis of impedance measurements across P_1 , P_2 , and P_3 using three electrode types (S_1 , S_2 , S_3) shows that electrode design directly affects data distribution. For Z , P_3 presented the highest proportion of non-normal cases (33.5%), indicating higher sensitivity to contact variability. S_2 (TENS adhesive) exhibited the highest stability, with 88% of cases following a normal distribution and homoscedasticity ($p = 0.08$). S_3 (suction electrode) showed the largest deviation from normality (29 non-normal cases, 54.8%), consistent with its perimeter-only metallic contact and suction coupling that introduce stochastic noise. S_1 (disposable ECG) showed intermediate performance (75% normality).

Despite these differences, the trend $P_1 > P_2 > P_3$ remained consistent across all electrodes. Statistical analysis followed the workflow dictated by data properties. These tests confirmed that while S_2 provides the most stable measurements, anatomical location remains the main determinant of impedance magnitude.

The resistive (R) and reactive (X_c) components followed the same trends as Z . Non-normality was highest at P_3 , while S_2 consistently showed the lowest variability. Differences between points were statistically significant ($p < 0.01$) even when absolute changes were small, supporting the use of averaged repeated measurements for spectral analysis. S_2

provides the highest statistical stability and reproducibility. S_3 introduces the most variability due to its perimeter contact geometry. P_3 is the least stable measurement point, exhibiting the highest frequency of non-normal data. Violations of parametric assumptions in 11–30% of cases justify a framework using Shapiro-Wilk and Levene tests to select between ANOVA and Kruskal-Wallis, with Tamhane T2 for post hoc analysis. S_2 is preferred for high-precision spectroscopy; S_3 requires non-parametric treatment or additional filtering due to inherent noise.

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APPENDIX B – STUDY ON A DEVICE TO TEST SKIN BIOELECTRICAL IMPEDANCE

This appendix contains a copy of the paper entitled *A Preliminary Study on a Device to Test Skin Bioelectrical Impedance Characteristics Related to Acupuncture Theory*, presented at the 2026 IEEE International Symposium on Medical Measurements and Applications (MeMeA 2026), Montevideo, Uruguay. .

A Preliminary Study on a Device to Test Skin Bioelectrical Impedance Characteristics Related to Acupuncture Theory

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Abstract— Although still controversial, acupuncture, a technique of traditional Chinese medicine, is currently being studied from several scientific perspectives. Electrical phenomena are one of them, and electroacupuncture—particularly electroacupuncture according to Voll (EAV), developed in the mid-twentieth century—has attracted growing interest. Some authors consider EAV a potential link between Eastern and Western medical approaches. More recently, acupuncture points and meridians have also been explored using concepts from electrical bioimpedance spectroscopy (EBIS). This article presents an initial technical validation of a homemade device, named Binatron, designed to derive both EAV and EBIS parameters from a single resistance measurement using square pulses and Fourier transformation. Measurements were conducted on a convenience sample of 61 chronically ill patients and 6 healthy individuals. Statistical comparisons ($\alpha = 0.05$) were performed on four acupuncture meridians (lung, large intestine, stomach, and spleen), both within and between groups. Analyses included comparisons between maximal EAV values (EAVmax) and modeled resistance at zero frequency (R_0), as well as EBIS-derived resistance (R) and reactance (X_c) at discrete frequencies between 2.9 and 111.3 Hz. Among all comparisons, only reactance (X_c) showed a statistically significant difference between groups (Mann–Whitney $U = 53.9$, $p = 0.028$). For R_0 , the median in patients was 607.48Ω (95% CI: 417.26 – 687.38Ω), while the mean in healthy individuals was 560.45Ω (95% CI: 445.92 – 674.98Ω). These results represent a preliminary evaluation of the Binatron device.

Keywords— Acupuncture, electroacupuncture, electroacupuncture according to Voll (EAV), chronic diseases, electrical impedance, electrical bioimpedance spectroscopy (EBIS)

I. INTRODUCTION

The characterization of acupuncture points (APs) and meridians, in terms of their distinctive bioelectrical properties.

remains a controversial topic in biomedical engineering. Historically, early research suggested that APs exhibit distinctly low electrical resistance compared to surrounding tissue. Recent studies suggest that this localized skin conductance may be mediated by nitric oxide (NO) signaling, the release of neuropeptides such as substance P, and a higher density of gap junctions [1]. Based on those early hypotheses, electroacupuncture according to Voll (EAV) was developed in the mid-20th century as a direct current (DC) or low-frequency measurement technique, that relates skin resistance to physiological conditions [2]. Around the same time, a similar approach was also developed by the Japanese physician Yoshio Nakatani [3], a technique known as Ryodoraku acupuncture (RA). EAV has been proposed, by some authors, as a possible link or bridge between Eastern and Western medicine [4] and the addition of EBIS could further contribute to this goal.

However, the biophysical validity of EAV and RA, as well as the reproducibility of APs electrical profiles are subject to considerable scientific debate. Some rigorous empirical studies have failed to reproduce the low impedance of APs. For example, Kramer et al., 2009 [5] found that, although momentary differences can be observed, the long-term reproducibility of skin electrical resistance at primary points (PPs) is very low, making it a poor diagnostic method. Similarly, independent studies on EAV have not found impedance differences between PPs and non-meridian points, suggesting that the observed variability may be due to methodological errors rather than actual physiology. Factors affecting the readings, especially for direct current (DC) measurements, are: electrode size, pressure applied, detection of acupoints, environment, skin moisture and instrumental limitations [3,6].

To overcome the limitations of single-frequency and DC resistance measurements, modern tissue characterization has shifted toward electrical bioimpedance spectroscopy (EBIS).

Unlike conventional EAV, which only provides arbitrary units susceptible to contact artifacts, EBIS maps the passive electrical response of tissues across a frequency spectrum. This allows for the extraction of inherent biophysical parameters, such as extracellular resistance (R_0) and membrane reactance (X_c), through Cole-Cole modeling [7]. The precision of EBIS is currently being applied to objective localized diagnostics, including the bioimpedance measurement of localized joint injuries [8].

This article presents preliminary data on the technical validation of a custom-designed prototype of a device, named "Binatron" by one of the authors [9], which eventually could provide a bridge to overcome the methodological gap between classical EAV theory and orthodox EBIS analysis. Square EAV impulses are applied and the passive response is converted both into EAV variables, as well as into EBIS parameters (resistance, R and capacitance, X_c , both in ohms), via Fourier transformation, which, in turn, are modeled to the Cole-Cole model and geometrical parameters [7]. In other words, both approaches are derived from the same measurement. With the proposed device, the authors seek to explore a more precise biophysical relationship between arbitrary Voll units and the modeled impedance, under controlled environmental and instrumentation conditions. In this preliminary report, maximal EAV (EAV_{max}) values are compared to EBIS R_0 values, as well as readings in four different acupuncture meridians (lung, large intestine, stomach and spleen or LU, LI, ST and SP, respectively), as well as between chronic ill patients ($n=61$) and a small group of healthy individuals ($n = 6$).

II. MATERIAL AND METHODS

A. Study Design and Population

A preliminary, cross-sectional, observational study was conducted, using a convenience sample of 61 chronic ill patients (38 female, 23 male) diagnosed with osteoarthritis (CIPG), and, for an initial comparison, a small sample of 6 healthy individuals (HIG), although of a different age group, due to the difficulty of recruiting old healthy individuals (**Table 1**). Pacemakers or metallic implants that could distort the signal were considered as exclusion criteria. The study protocol was approved by the University Ethics Committee, and all participants provided written informed consent, prior to enrollment. After being clinically examined by a certified acupuncturist, patients were assigned to four different groups, based on their symptoms, targeting specific acupuncture channels: LU ($n=15$), LI ($n=14$), ST ($n=18$), or SP ($n=14$).

TABLE I. MEAN AGE (YEARS) AND SD OF THE SAMPLE OF VOLUNTEERS PARTICIPATING IN THE STUDY.

Gender	Chronically ill patients (CIPG)			Healthy individuals (HIG)		
	n	Age (y)		n	Age (y)	
		Mean	SD		Mean	SD
Female	38	65.6	11.6	3	22.0	4.0
Male	23	59.6	11.3	3	21.7	1.2
Total	61	63.3	11.7	6	21.8	2.6

B. Environmental and homeostatic Controls

All subjects fasted for a minimum of 2 hours before the examination, and rested for 5 minutes in a supine position before baseline recordings. Ambient temperature was controlled and maintained constant between 18°C and 21°C. Pressure applied to the probe was also controlled, via an active probe integrated to a load sensor, ensuring a contact pressure, with forces maintained between 0.3 to 0.5 N. The target skin was prepared with 75% isopropyl alcohol.

C. Prototype of the Binatron

Data was collected with a custom prototype capable of simultaneously capturing conventional EAV readings and converting the applied voltage and measured current into electrical bioimpedance spectrometry (EBIS) data, via Fourier transform. The system complies with IEC 60601, via galvanically isolated USB communication. Its core hardware components include: a) a central measurement unit, b) two brass-bronze electrodes (a cylindrical sink and a pencil-like source), and c) a 5V (≥ 2000 mAh) power supply. The device's technical specifications are: a) maximum applied voltage of 1.2 V; b) current range of 0–12 μ A, with a resolution of 5.8 nA; c) EAV resolution of 0.0244 arbitrary units; d) impedance range of 5–500 k Ω ; e) bandwidth of 0.025–120 Hz, and f) programmable pulse duration from 100 ms to 20 s in 10 ms steps.

D. Measurement Procedure and Data Processing

Measurements were adjusted according to EAV standards [10]. The cylindrical electrode was held in the patient's right hand, while the operator applied the active probe to the designated contralateral hand acupuncture point, or to the ipsilateral point in the foot, depending on the meridian being measured. Three consecutive readings were taken from each subject and averaged. The passive electrical response was treated with a Fourier transform to determine resistance (R) and capacitive reactance (X_c) at 8 discrete frequencies (2.9, 4.9, 10.1, 30.4, 50.6, 70.8, 91.1 and 111.3 Hz). Finally, the impedance parameterization was carried out by applying the three-point (3P) modeling proposed by González-Correa 2019 [7] and González-Correa et al., 2022 [11].

E. Statistics

Statistical analyses were performed to evaluate whether the electrical parameters obtained with the proposed device differed across meridians and between the two groups considered in the study: CIPG and HIG. The electrical parameters evaluated included: EAV_{max} , resistance at 0 Hz (R_0), and averaged modeled resistance (R) and reactance values (X_c), at 8 frequencies in the ∞ dispersion. Normality of the distributions was assessed using the Shapiro–Wilk test. Statistical comparisons were selected according to distribution characteristics and sample size. For comparisons among four meridian groups, the following tests were applied: one-way analysis of variance (ANOVA), when normality assumptions were met, and Kruskal–Wallis test when distributions deviated from normality. For comparisons between two populations (patients vs healthy individuals): Welch's t-test was used for normally distributed variables, Mann–Whitney U test was applied for non-normally distributed variables. For correlations, the Spearman test was used. The statistical significance threshold was $\alpha = 0.05$. and all analyses were performed using Python.

III. RESULTS

When comparing $EAV_{\max}$ and R_0 for the four meridians (LU, LI, ST and SP), in each group, no statistically significant differences were found. For the CIPG and the HIG groups, p -values of 0.197 and 0.702 were obtained, respectively, for $EAV_{\max}$, both by using the Kruskal-Wallis test. For R_0 , p -values were 0.142 for the CIPG (also Kruskal-Wallis) and 0.547 for the HIG (one-way ANOVA). When comparing $EAV_{\max}$ and R_0 between the CIPG and the HIG, p -values obtained with a Mann-Whitney U test were 0.409 and 0.624, respectively. Finally, when comparing the average values of R and X_c (8 frequencies) pertaining to both groups of participants, p -values obtained were 0.295 (Welch t -test) and 0.028 (Mann-Whitney U test), respectively.

In the correlation between $EAV_{\max}$ and R_0 , trend lines with the highest R^2 were of a potential nature, as shown in **Figures 1a** and **1d**, which show an R^2 of 0.9568 for the CIPG and of 0.9941 for the HIG. When the data is logarithmically transformed (**Figures 1b** and **1e**), an excellent negative linear correlation is obtained, with an R^2 of 0.9622, for the CIPG and 0.9915 for the HIG. In **Figures 1c** and **1f**, Nyquist plots are also presented, for the averaged data of all readings taken from the CIPG and the HIG, respectively. In both cases, the bioelectrical impedance data fit well for most of the frequencies, with only a big deviation at the highest frequency (111.3 Hz) for the CIPG and two minor deviations for the two lower ones (2.9 and 4.9 Hz), in the HIG.

Normality testing using the Shapiro-Wilk test showed that $EAV_{\max}$ values and R_0 values in the patient group were not normally distributed ($p < 0.001$), whereas R_0 values in the healthy group were compatible with normality ($p = 0.326$). Accordingly, medians with bootstrap 95% confidence intervals were calculated for non-normal variables, and means with parametric confidence intervals were calculated for normally distributed variables. The median $EAV_{\max}$ value was 12.9 (95% CI: 10.7–16.0) in patients and 12.0 (95% CI: 10.8–19.1) in healthy individuals. For R_0 , the patient median was 607.48 Ω (95% CI: 417.26–687.38 Ω), while the mean R_0 in healthy individuals was 560.45 Ω (95% CI: 445.92–674.98 Ω). A strong inverse correlation was observed between logarithmically transformed maximum EAV values and resistance at 0 Hz (R_0) in the patient cohort (Pearson $r = -0.981$, $p < 10^{-42}$, $n = 61$). A similar association was confirmed using Spearman correlation ($\rho = -0.984$, $p < 10^{-44}$). Linear regression in log-log space yielded the relationship $\log(R_0) = 9.028 - 1.061 \cdot \log(\text{Max EAV})$, indicating an approximately reciprocal power-law relationship between the two electrical parameters.

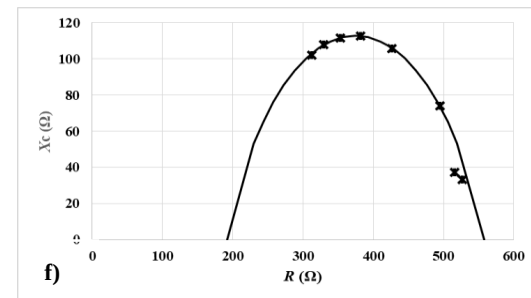
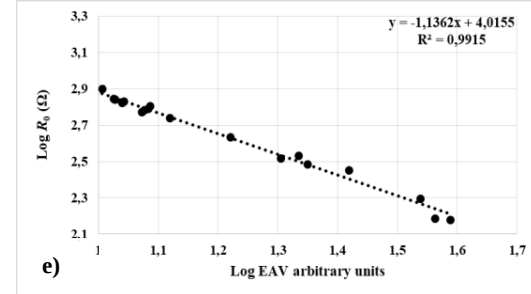
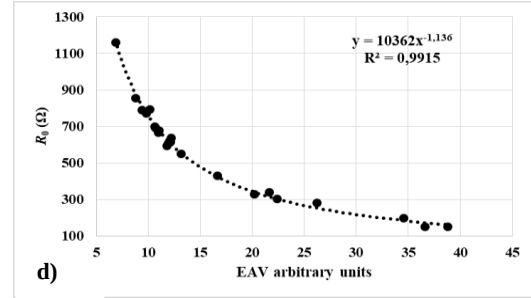
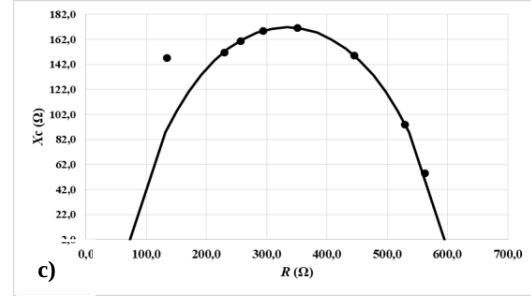
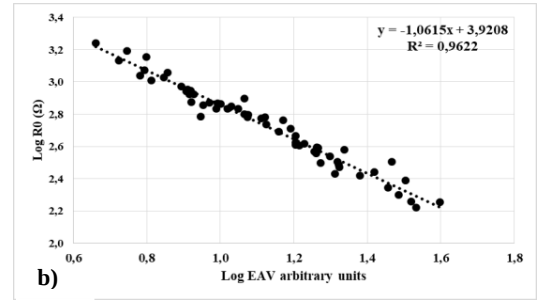
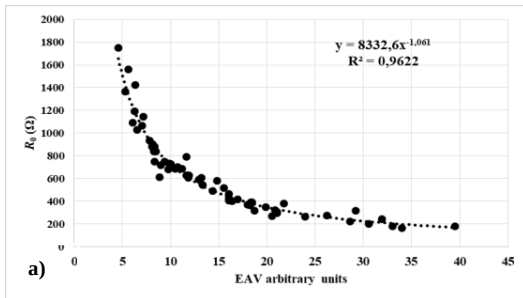


Fig. 1. Trend lines (a and d) for the raw data and correlation lines (b and e) for the logarithmically transformed values, for the pooled data of the 61 patients (a and b), and for the small sample of 6 healthy individuals (d and e). Nyquist plot for the average of all readings in both groups are also presented (c for the CIPG and f, for the HIG).

IV. DISCUSSION

Across the electrical parameters evaluated, most variables - including $EAV_{\max}$ values, R_0 , and modeled R - showed no statistically significant differences, either among meridians or between populations. However, reactance (X_c) demonstrated a significant difference between patients with chronic diseases and healthy individuals. Since electrical reactance is associated with membrane capacitance and tissue structural properties, this finding suggests that reactance may represent the most sensitive EBIS-derived parameter for detecting physiological differences between the two populations. These results support the feasibility of combining EAV measurements with bioimpedance spectroscopy modeling within a single device, while highlighting reactance as a potentially valuable biomarker for future diagnostic applications. The study shows some limitations, that will be dealt with in future studies: a) the convenience samples recruited for this preliminary study are small; b) the ages of the two groups (ill patients and healthy individuals) are quite different, due to the difficulty of recruiting old people in healthy conditions; c) the EAV readings obtained with the Binatron have to be compared with those obtained with a reliable Dermatran; d) there are many other parameters, different to those considered in this preliminary study, to be analyzed, such as the indicator drop, for EAV, and R_{∞} , phase angle, impedance ratio and the exponent α , for EBIS.

V. CONCLUSIONS

As a preliminary technical validation, this study focused on testing the Binatron device's capability to bridge EAV readings and EBIS parameters. EAV is widely used in many different medical situations, even in the treatment of obesity [12]. A search in PubMed on 202603/04 with the expression "Electroacupuncture"[Mesh], gave a total of 5,885 results, the majority of them in this century, where the number of publications on the matter has been steadily increasing. Future clinical studies with healthy cohorts are necessary to define diagnostic reference values.

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APPENDIX C – BIOLOGICAL SKIN PHANTOM FOR ELECTRICAL IMPEDANCE SPECTROSCOPY

This appendix contains a copy of the manuscript entitled *Biological Skin Phantom for Electrical Impedance Spectroscopy*. This manuscript was submitted to the XXX Congresso Brasileiro de Engenharia Biomédica (CBEB 2026) and was subsequently accepted for presentation. The study is based on the case studies developed in Chapters 4 and 5 of this thesis, focusing on the design, fabrication, and electrical characterization of multilayer skin phantoms intended to reproduce the passive electrical properties of human skin for electrical impedance spectroscopy experiments.

Tetrapolar Electrode Geometries for Skin Phantoms Using Electrical Impedance Spectroscopy

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Abstract—This work presents the fabrication and electrical characterization of multilayer gelatin-based phantoms designed to reproduce the passive electrical properties of the epidermis, dermis, and hypodermis, and to evaluate the sensitivity of tetrapolar electrode geometry to subsurface impedance changes. A temperature-resistivity calibration was first performed, comparing agar- and gelatin-based formulations; agar exhibited excessive intrinsic conductivity and a limited resistivity range, whereas gelatin provided both adequate mechanical stability and the target resistivity values needed for multilayer fabrication. Using an empirical regression relating temperature, NaCl concentration, and conductivity, three phantoms (P1, P2, P3) were built with identical epidermis and hypodermis resistivities but distinct dermis resistivities, allowing isolated evaluation of subsurface sensitivity. Electrical Impedance Spectroscopy (200 Hz–100 kHz) was performed on each phantom using five tetrapolar electrode configurations (three linear, two coplanar), and the resulting spectra were analyzed with Relative Difference (RD) and Bland–Altman statistics. Coplanar arrangements showed moderate, stable sensitivity to the middle layer variations, with impedance differences that gradually converged at higher frequencies. Among the linear configurations, Linear Arrangement 1 was the most sensitive, with relative differences exceeding 100% at low frequencies for the phantom with the greatest resistivity change, while Linear Arrangements 2 and 3 produced smaller but more consistent responses across the spectrum. Bland–Altman analysis confirmed that the observed differences were systematic rather than random, verifying that electrode geometry governs current distribution and sensing depth. The results demonstrate the potential of the proposed phantoms as a controlled experimental platform for assessing electrode configurations and electrical variations in subsurface skin layers, with educational applications using biological phantoms.

Index Terms—Multilayer measurement, customized skin phantoms, EIS, electrode geometries

I. INTRODUCTION

The skin, the largest organ of the human body, acts as a protective barrier against physical, chemical, and biological agents while maintaining essential homeostatic functions [1] [2]. It is composed of three main layers, namely the epidermis, dermis, and subcutaneous tissue, each exhibiting distinct structural and electrical characteristics [3]. Owing to its continuous exposure to harmful agents associated with skin cancer [3], considerable efforts have focused on developing minimally invasive techniques for early diagnosis, reducing unnecessary

biopsies and improving clinical decision-making [4] [5] [6] [7] [8].

Electrical Impedance Spectroscopy (EIS) has emerged as a promising non-invasive technique for differentiating healthy and malignant skin tissues by exploiting changes in their passive electrical properties [9] [10] [11]. Variations in cell morphology, membrane integrity, and tissue composition modify tissue conductivity and capacitance. These alterations produce measurable changes in the impedance spectrum and have been associated with lesions such as Basal Cell Carcinoma (BCC), Squamous Cell Carcinoma (SCC), and melanoma. [9] [12] [11] [13]

The multilayered structure of the skin strongly influences the measured impedance response. Since each layer presents different passive electrical properties, the measured spectrum results from the combined contribution of the epidermis, dermis, hypodermis, and embedded anatomical structures [14] [15] [13]. Moreover, the sensitivity of EIS depends on the distribution of the injected alternating current which is governed by electrode geometry, inter-electrode spacing, measurement configuration, and excitation frequency. [16] [17] [18] [12].

Direct investigation of these factors in vivo is limited by biological variability, ethical constraints, and the difficulty of reproducing controlled experimental conditions [19] [13] [20]. Consequently, biological phantoms have become an important alternative for evaluating bioimpedance systems under reproducible conditions [21] [22] [23] [24]. By reproducing the passive electrical properties of biological tissues, multilayer phantoms enable systematic investigations of current distribution, electrode sensitivity, and impedance variations while providing a low-cost platform for instrumentation development and experimental validation [16].

Commercial EIS systems, such as Nevisense®, have demonstrated promising clinical performance for melanoma detection [25]. However, their high cost and proprietary electrode configurations limit their use in research and educational environments, restricting experimental reproducibility and the development of new measurement methodologies [19]. Therefore, low cost multilayer skin phantoms can provide a controlled and accessible tool for teaching EIS, testing electrode configurations, and studying electrical variations in skin layers. These models may support the development of measurement

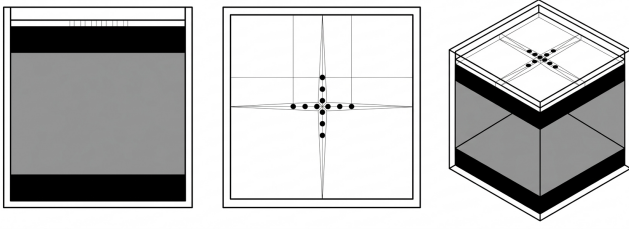


Fig. 1. Side, top, and isometric elevational views of the gelatin phantom. The visualization highlights the multilayered matrix and the corresponding perforated cover, prior to the final positioning and fixation of the measuring electrodes.

methods that may contribute to future diagnostic applications, including cancer detection.

This work investigates the sensitivity of a three layered gelatin skin phantom to variations in tetrapolar electrode geometry using Electrical Impedance Spectroscopy. The study focuses on controlled changes in dermis resistivity and does not evaluate cancer detection.

II. MATERIALS AND METHODS

A. Creation of the Phantoms

Three multilayer skin phantoms were fabricated using three layers representing the epidermis, dermis, and hypodermis. The layers has thicknesses of 12.15 mm, 56.70 mm, and 12.15 mm, respectively, corresponding to 15%, 70%, and 15% of the 81 mm phantom depth. Each layer was prepared using distilled water and commercial gelatin with documented physicochemical specifications, while controlled amounts of sodium chloride (NaCl) were added to obtain the target electrical conductivity of each skin mimicking layer [26], [27]. The gelatin label indicates a pH of 5.2 in a 1% solution, moisture below 12%, ash below 2%, nitrogen below 15%, and a particle size of 30 mesh. The electrical properties were determined experimentally through the calibration procedure described below.

The phantom geometry was designed to approximate a homogeneous, isotropic, and semi-infinite medium only in the region adjacent to the electrode array where the electric field is established during the measurements. Each phantom was prepared in a cubic acrylic mold with internal dimensions of 81x81x8 mm and stored for 8 hours before measurements. A cover with twelve circular holes held 1.6 mm diameter polished stainless steel electrodes (AISI 308L) in a cross arrangement. The center to center spacing was 5 mm, with a maximum interelectrode distance of 25 mm. The cover weighs 40 g, resulting in a contact pressure of approximately 16 kPa.

B. Conductivity Temperature Calibration

Because the electrical properties of gelatin-based materials are temperature-dependent, controlled cooling ramps were performed prior to phantom fabrication to characterize the relationship between temperature, NaCl concentration, and electrical conductivity. One gelatin-agar and three gelatin-only

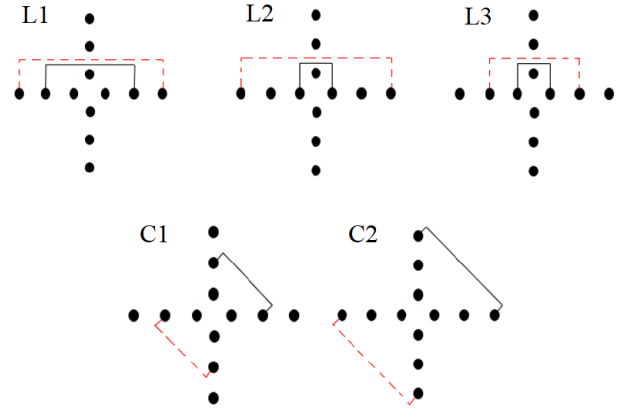


Fig. 2. The five different configurations for linear and coplanar interelectrode arrays. The dotted lines correspond to the current electrodes and the solid lines to the voltage electrodes respectively.

formulations were evaluated for their temperature-dependent electrical behavior and mechanical stability. The resulting calibration curves were used to estimate resistivity at the target measurement temperature (10°C) and to guide formulation selection for multilayer phantom fabrication [27].

TABLE I
TEMPERATURE-DEPENDENT RESISTIVITY OF THE EVALUATED PHANTOM FORMULATIONS.

$T(^{\circ}\text{C})$	A($\Omega\cdot\text{m}$)	B($\Omega\cdot\text{m}$)	C($\Omega\cdot\text{m}$)	D($\Omega\cdot\text{m}$)
60.0	19.75	4.74	7.39	13.33
57.5	19.75	5.24	7.64	13.50
55.0	21.20	5.30	7.86	13.88
52.5	21.55	5.54	8.08	14.03
50.0	22.16	5.74	8.40	14.77
47.5	22.94	5.92	8.64	15.15
45.0	23.57	6.15	8.97	15.76
42.5	24.00	6.37	9.25	16.32
40.0	25.33	6.89	9.55	16.92
37.7	25.66	7.04	9.79	17.61
35.0	26.53	7.18	10.40	18.35
32.5	27.27	7.44	10.76	18.98
30.0	28.89	7.88	11.39	19.95

A: 4 g agar and 96 mL water; B: 8 g gelatin and 92 mL water; C: 6 g gelatin and 94 mL water; D: 3 g gelatin and 97 mL water. Resistivity values are expressed in $\Omega\cdot\text{m}$.

The formulation containing 3 g of gelatin and 97 mL of water was selected as the epidermis-equivalent layer. Based on the temperature-resistivity calibration, the resistivity at the measurement temperature (10°C) was estimated using Eqs. (1)–(3), yielding an extrapolated value of approximately 23.84 $\Omega\cdot\text{m}$. This value was adopted as the reference resistivity for the epidermis-equivalent phantom, while the target resistivities for the remaining layers were determined according to the relative ratios reported in the literature.

$$\rho(T) = -0.2227T + 26.0690 \quad (1)$$

$$\rho(10) = -0.2227(10) + 26.0690 = 23.84 \text{ } \Omega\text{m} \quad (2)$$

$$\rho_{epi} = 23.84 \text{ } \Omega\text{m} \quad (3)$$

Rather than reproducing the absolute resistivity values reported for human skin, the multilayer phantom was designed to preserve the relative electrical contrasts among the epidermis, dermis, and hypodermis described in the literature [28], [29], [13]. Consequently, the target resistivities of the remaining layers were obtained by scaling the resistivity of the epidermis-equivalent formulation according to the relative ratios reported for biological skin tissues.

For the middle layer (dermis):

$$\rho_{der} = \rho_{epi} \left(\frac{20}{100} \right) = 23.84 \left(\frac{20}{100} \right) = 4.77 \text{ } \Omega\text{m} \quad (4)$$

For the bottom layer (hypodermis):

$$\rho_{hyp} = \rho_{epi} \left(\frac{40}{100} \right) = 23.84 \left(\frac{40}{100} \right) = 9.54 \text{ } \Omega\text{m} \quad (5)$$

TABLE II
TARGET ELECTRICAL PROPERTIES ADOPTED FOR THE MULTILAYER SKIN PHANTOM BASED ON RELATIVE RESISTIVITY SCALING.

Layer	Relative Ratio	Resistivity (Ωm)	Conductivity (S/m)
Epidermis	1.00	23.84	0.0419
Dermis	0.20	4.77	0.2096
Hypodermis	0.40	9.54	0.1048

Once the target apparent resistivities had been defined, two additional gelatin formulations consisting of 3 g of gelatin and 97 mL of water were prepared by incorporating different amounts of sodium chloride (NaCl), namely 0.010 g and 0.020 g. Similar to the previous experiments, both formulations were subjected to a descending temperature ramp from 60°C to 30°C, while conductivity measurements were acquired at regular intervals. These data were used to establish an empirical relationship among temperature, NaCl concentration, and electrical conductivity, enabling the estimation of the amount of salt required to achieve a target resistivity at a specified temperature. The resulting conductivity profiles are presented in Table II.

Controlled additions of NaCl were subsequently used to adjust the conductivity of each layer, resulting in target resistivities of 23.84 Ωm , 4.77 Ωm , and 9.54 Ωm for the top, middle, and bottom layers, respectively. To address this issue, an empirical multiple regression model was developed using gelatin formulations containing different sodium chloride concentrations and conductivity measurements acquired over a range of temperatures. Since the phantoms are fully solidified at the measurement temperature of 10°C, direct conductivity measurements at this stage are difficult to perform. Therefore, the model was employed to estimate both the amount of

TABLE III
TEMPERATURE-DEPENDENT ELECTRICAL PROPERTIES OF GELATIN PHANTOMS WITH DIFFERENT NaCl CONCENTRATIONS.

T ($^{\circ}\text{C}$)	A		B	
	σ (S/m)	ρ ($\Omega\cdot\text{m}$)	σ (S/m)	ρ ($\Omega\cdot\text{m}$)
60.0	0.1047	9.55	0.2096	4.77
57.5	0.1024	9.76	0.2054	4.87
55.0	0.1005	9.95	0.2015	4.96
52.5	0.0974	10.27	0.1952	5.12
50.0	0.0950	10.53	0.1902	5.26
47.5	0.0911	10.98	0.1826	5.48
45.0	0.0895	11.17	0.1791	5.58
42.5	0.0876	11.42	0.1754	5.70
40.0	0.0864	11.57	0.1730	5.78
37.7	0.0845	11.83	0.1691	5.91
35.0	0.0828	12.08	0.1656	6.04
32.5	0.0818	12.23	0.1636	6.11
30.0	0.0801	12.48	0.1602	6.24

A: Gelatin phantom prepared with 3 g gelatin, 97 mL distilled water, and 0.010 g NaCl. B: Gelatin phantom prepared with 3 g gelatin, 97 mL distilled water, and 0.020 g NaCl.

NaCl required to achieve a target electrical resistivity and the expected resistivity of the phantom at 10°C. The resulting relationship is given by:

$$\sigma(T, m_{\text{NaCl}}) = 0.03908 + 2.87 \times 10^{-4} T + 8.3845 m_{\text{NaCl}}, \quad (6)$$

where σ is the electrical conductivity (S/m), T is the temperature ($^{\circ}\text{C}$), and m_{NaCl} is the mass of sodium chloride added to the formulation (g).

Since all impedance measurements were carried out at 10°C, Eq. (6) simplifies to

$$m_{\text{NaCl}} = \frac{\frac{1}{\rho_{\text{obj}}} - 0.04195}{8.3845}. \quad (7)$$

C. Measurements and Data Collection

Three different phantoms (P1, P2, and P3) were independently constructed to investigate the influence of the middle layer resistivity on the measured impedance. The top and bottom layers were kept unchanged across all phantoms, while only the middle layer resistivity was varied to represent different electrical conditions.

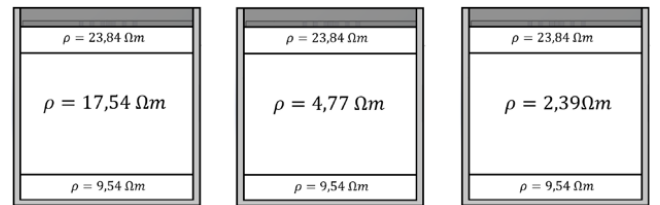


Fig. 3. Phantoms P1, P2, and P3 with the respective resistivity values of each layer.

EIS measurements were done by using the EmStat4R potentiostat (PalmSens BV) operating in a tetrapolar configuration.

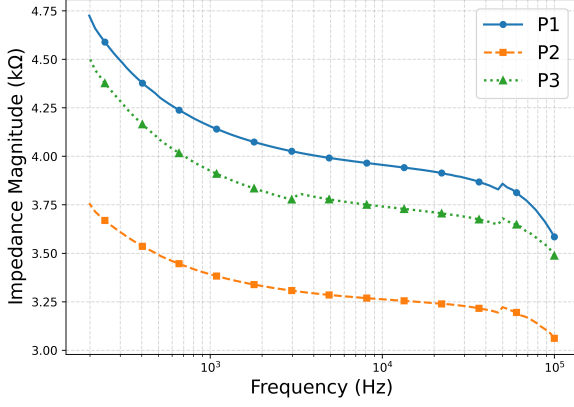


Fig. 4. Impedance magnitude ($|Z|$) spectra of Phantoms P1, P2, and P3 obtained using the Coplanar Arrangement 1 electrode configuration over the frequency range from 200 Hz to 100 kHz. The figure illustrates the influence of variations in the resistivity of the middle layer on the surface impedance response of the multilayer phantom.

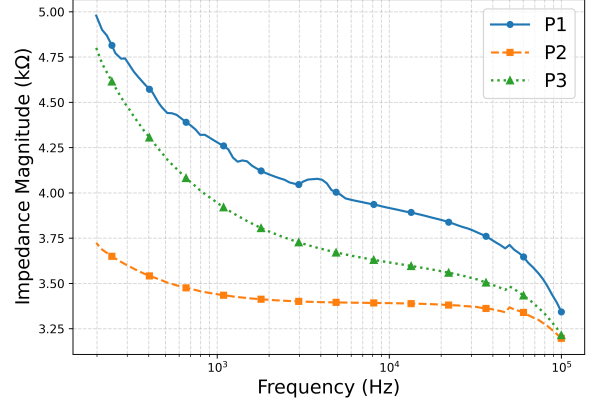


Fig. 5. Impedance magnitude ($|Z|$) spectra of Phantoms P1, P2, and P3 obtained using the Coplanar Arrangement 2 electrode configuration over the frequency range from 200 Hz to 100 kHz. The figure illustrates the influence of variations in the resistivity of the middle layer on the surface impedance response of the multilayer phantom.

Impedance magnitude and phase were measured from 0.2 to 100 kHz, recording. Each phantom was measured using the five electrode configurations, resulting in fifteen experimental datasets. In addition, each layer was individually characterized before depositing the subsequent layer to validate its electrical properties.

The sensitivity of each electrode configuration to middle layer resistivity variations was evaluated by comparing the impedance spectra of the three phantoms. Relative Difference (%RD) and Bland–Altman analyses were employed to quantify the differences between the reference phantom (P2) and the modified phantoms (P1 and P3), following the methodology described in [30].

III. RESULTS

Figures 4–8 present the impedance magnitude ($|Z|$) and real impedance (Z') spectra obtained using the three linear and two coplanar electrode configurations. For each configuration, the responses of Phantoms P1, P2, and P3 are presented over the frequency range from 200 Hz to 100 kHz, enabling a direct comparison of the sensitivity of each electrode configuration to the variations introduced in the middle layer resistivity.

As observed in Figures 4 and 5, the Coplanar Arrangement 1 and Coplanar Arrangement 2 electrode configurations exhibited higher impedance magnitude and real impedance values for Phantoms P1 and P3 than for Phantom P2 over most of the analyzed frequency range. These differences remained nearly constant throughout the spectrum, with Phantom P1 consistently exhibiting slightly higher values than Phantom P3. However, the responses of the three phantoms gradually converged as the frequency increased, becoming more similar at frequencies above 100 kHz.

For the Linear Arrangement 1 electrode configuration, a nearly constant difference of less than 1000 Ω was observed between Phantoms P1 and P2 over the entire frequency range for both the impedance magnitude and the real impedance.

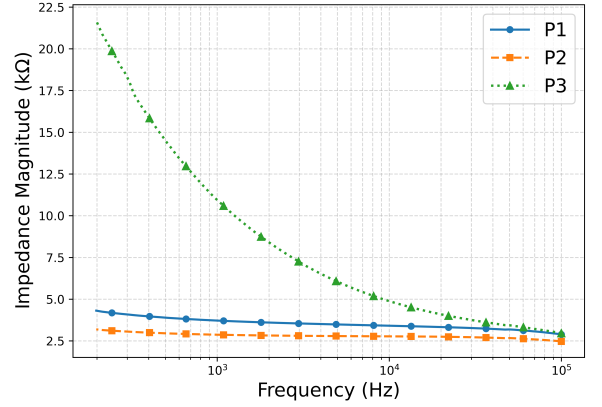


Fig. 6. Impedance magnitude ($|Z|$) spectra of Phantoms P1, P2, and P3 obtained using the Linear Arrangement 1 electrode configuration over the frequency range from 200 Hz to 100 kHz. The figure illustrates the influence of variations in the resistivity of the middle layer on the surface impedance response of the multilayer phantom.

In contrast, Phantom P3 exhibited a markedly different behavior, with impedance magnitude and real impedance values approximately five to ten times higher than those of Phantoms P1 and P2 at low frequencies. As the frequency increased, both parameters decreased progressively, converging toward the responses of P1 and P2 at approximately 100 kHz.

For the Linear Arrangement 2 electrode configuration, higher impedance magnitude and real impedance values were observed for Phantom P1 at the lower end of the frequency spectrum (200 Hz). In contrast, Phantoms P2 and P3 exhibited similar responses, maintaining an approximately constant separation throughout most of the analyzed frequency range despite the gradual decrease in impedance with increasing frequency. As the frequency approached the upper limit of the investigated spectrum, the responses of all three phantoms converged rapidly toward lower impedance values.

For the Linear Arrangement 3 electrode configuration, a

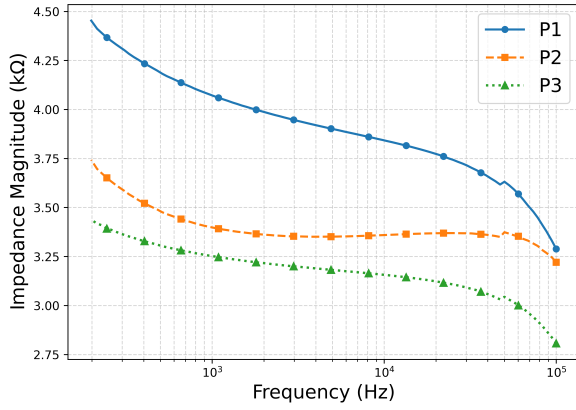


Fig. 7. Impedance magnitude ($|Z|$) spectra of Phantoms P1, P2, and P3 obtained using the Linear Arrangement 2 electrode configuration over the frequency range from 200 Hz to 100 kHz. The figure illustrates the influence of variations in the resistivity of the middle layer on the surface impedance response of the multilayer phantom.

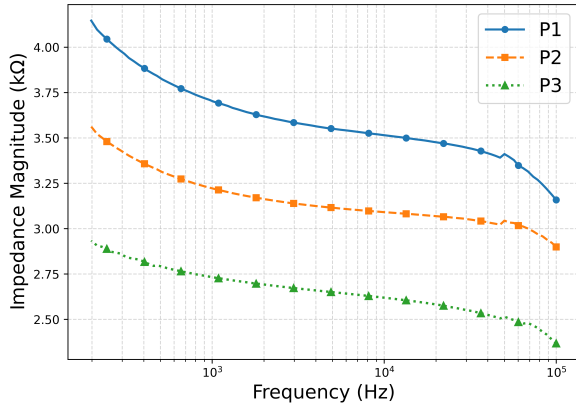


Fig. 8. Impedance magnitude ($|Z|$) spectra of Phantoms P1, P2, and P3 obtained using the Linear Arrangement 3 electrode configuration over the frequency range from 200 Hz to 100 kHz. The figure illustrates the influence of variations in the resistivity of the middle layer on the surface impedance response of the multilayer phantom.

nearly constant separation between the responses of the three phantoms was observed over the entire analyzed frequency range. Both the impedance magnitude and the real impedance exhibited a gradual decrease from 200 Hz to 100 kHz, with the three spectra following approximately parallel trends and maintaining a similar difference throughout the spectrum.

Since the top layer resistivity was identical in all three phantoms, these results suggest that the Linear Arrangement 3 configuration is predominantly sensitive to variations in the electrical properties of the deeper layers. The nearly uniform separation between the spectra indicates that changes in middle layer resistivity produce a proportional variation in the measured impedance over the investigated frequency range.

To complement the analysis of the impedance spectra, a statistical evaluation was subsequently performed using the relative difference (%RD) and Bland–Altman analysis. These methods allowed an objective quantification of the differences

between the reference phantom (P2) and Phantoms P1 and P3, as well as an assessment of the agreement between the impedance measurements obtained with each electrode configuration [30].

A. Statistical Analysis

Figures 9–13 present the Relative Difference (%RD) and Bland–Altman plots for the five electrode configurations evaluated. %RD was calculated using P2 as the reference according to:

$$\%RD = \frac{|Z_i - Z_{P2}|}{Z_{P2}} \times 100, \quad (8)$$

where Z_i represents the impedance magnitude of P1 or P3 at each frequency.

The Bland–Altman analysis was used to characterize the distribution and bias of both reference (P2) and modified resistivity phantoms. The absolute impedance difference was calculated as $\Delta Z (= |Z_i - Z_{P2}|)$ in order to avoid interpretation based only on the reference magnitude.

Values above 100% indicate that the absolute impedance difference exceeded the P2 reference value. Because the calculation uses the reference impedance as the denominator, then small reference values produce large %RD values. Therefore, %RD is better interpreted together with the absolute impedance difference.

The Relative Difference plots show the percentage variation in impedance magnitude as a function of frequency for the P1–P2 and P3–P2 comparisons. This analysis identifies the frequency intervals in which changes in middle layer resistivity produced the largest deviations from the reference phantom, as well as the regions where the electrical response remained comparatively stable.

The Bland–Altman plots present the impedance differences as a function of the mean values, allowing the bias and distribution of the differences evaluated for each electrode configuration. Together, these analyses quantify the sensitivity of each electrode arrangement to changes in the middle layer resistivity in the multilayer phantoms [30].

It is important to emphasize that, in the context of the present study, the objective of the %RD analysis is not to minimize the relative difference, but to evaluate the sensitivity of each electrode configuration to changes middle layer resistivity. Consequently, a %RD value close to zero indicates that the electrode configuration is relatively insensitive to the modifications introduced in the inner layer. Conversely, a high and consistent %RD indicates that the configuration is capable of detecting changes in resistivity of the middle layer from surface impedance measurements. Likewise, a stable %RD over the analyzed frequency range indicates a consistent response of the electrode configuration to variations in the electrical properties of the middle layer.

Unlike the relative difference analysis, which quantifies the sensitivity of each electrode configuration to changes in resistivity of the middle layer. The Bland–Altman analysis evaluates the distribution and consistency of these differences over the analyzed frequency range. It is better to emphasize

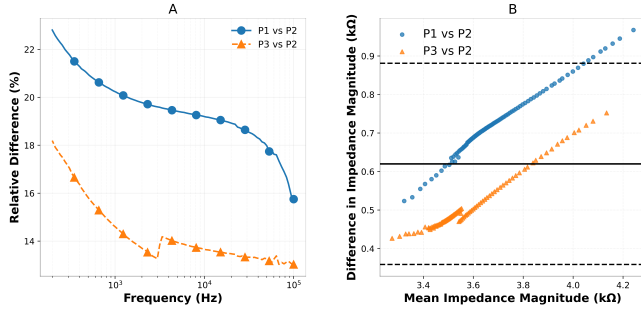


Fig. 9. (A) %RD in impedance magnitude for the P1–P2 and P3–P2 comparisons obtained with the Coplanar Arrangement 1 electrode configuration. (B) Bland–Altman analysis of the corresponding measurements, with a mean bias of 619.83 Ω and limits of agreement defined as ± 1.96 SD.

that the objective here was not intended to obtain a bias close to zero, since the phantoms were intentionally prepared with different resistivities for the middle layers. Therefore, a consistent bias with a compact distribution of the differences indicates a stable response to the introduced changes in resistivity of the middle layers.

For the Coplanar Arrangement 1 electrode configuration, the Relative Difference (%RD) between Phantoms P1 and P2 decreased from approximately 22% at 200 Hz to 15% at 100 kHz. Similarly, the comparison between Phantoms P3 and P2 exhibited a gradual decrease from approximately 18% to 13% over the same frequency range. The smooth reduction in %RD with increasing frequency indicates that the impedance responses of the three phantoms became progressively more similar at higher frequencies while maintaining a consistent separation throughout the spectrum.

The corresponding Bland–Altman plot shows two well-defined ascending clusters associated with the P1–P2 and P3–P2 comparisons, with differences ranging from approximately 400 Ω to more than 1000 Ω . The calculated bias of 619.83 Ω indicates a systematic difference between the reference phantom (P2) and the modified phantoms. Furthermore, the compact distribution of the data around each trend suggests that these differences remain consistent over the analyzed frequency range, demonstrating that the Coplanar Arrangement 1 configuration is capable of reliably distinguishing variations in middle layer resistivity.

For the Coplanar Arrangement 2 electrode configuration, a behavior similar to that observed for Coplanar Arrangement 1 was obtained. The Bland–Altman analysis yielded a bias of 511.89 Ω , indicating a systematic difference between the reference phantom (P2) and the modified phantoms. The data points formed two closely spaced ascending clusters, with differences ranging from approximately 0 Ω to 1200 Ω , demonstrating a consistent trend over the analyzed frequency range.

The %RD also exhibited a gradual decrease with increasing frequency for both the P1–P2 and P3–P2 comparisons. As the frequency approached the upper limit of the analyzed spectrum, the relative differences tended to decrease toward

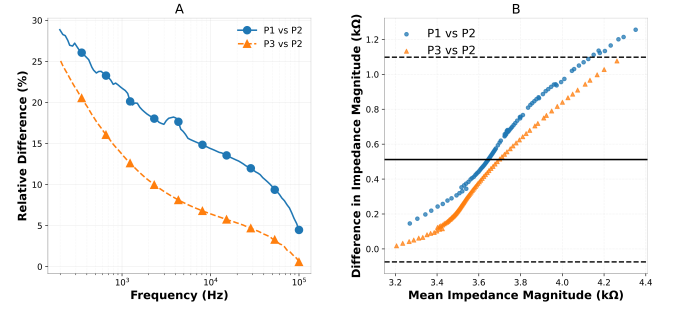


Fig. 10. (A) %RD in impedance magnitude for the P1–P2 and P3–P2 comparisons obtained with the Coplanar Arrangement 2 electrode configuration. (B) Bland–Altman analysis of the corresponding measurements, with a mean bias of 511.89 Ω and limits of agreement defined as ± 1.96 SD.

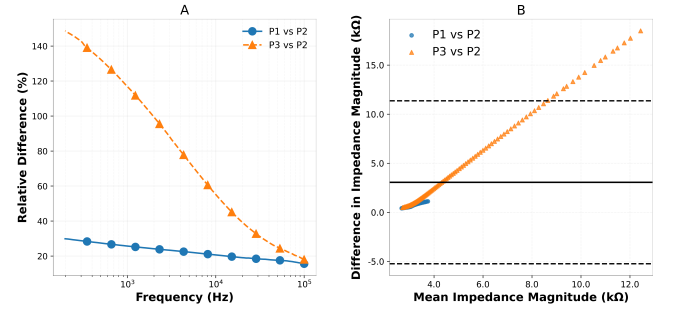


Fig. 11. (A) %RD in impedance magnitude for the P1–P2 and P3–P2 comparisons obtained with the Linear Arrangement 1. (B) Bland–Altman analysis of the corresponding measurements, with a mean bias of 3062.96 Ω and limits of agreement defined as ± 1.96 SD.

zero, indicating that the electrical responses of the three phantoms became progressively more similar above 100 kHz.

For the Linear Arrangement 1 electrode configuration, the %RD between Phantoms P1 and P2 remained relatively low and nearly constant, decreasing slightly from approximately 30% to 20% over the analyzed frequency range. In contrast, the P3–P2 comparison exhibited a pronounced descending trend, with the %RD decreasing from approximately 140% at 200 Hz to values close to 20% at 100 kHz, where it converged toward the P1–P2 comparison.

The corresponding Bland–Altman plot clearly distinguishes the two phantom comparisons. The P1–P2 data points form a compact cluster below the bias, concentrated at lower mean impedance values and characterized by relatively small differences. In contrast, the P3–P2 comparison follows a marked ascending trend, with the differences increasing from approximately 0 Ω to values exceeding 15 kΩ. The calculated bias of 3062.96 Ω , together with the wide separation between the two clusters, indicates that the Linear Arrangement 1 configuration is highly sensitive to the modifications introduced in the middle layer resistivity, particularly for Phantom P3.

For the Linear Arrangement 2 electrode configuration, the %RD between Phantoms P1 and P2 remained relatively low, decreasing from approximately 17% to 2% over the analyzed frequency range. An even smaller relative difference was

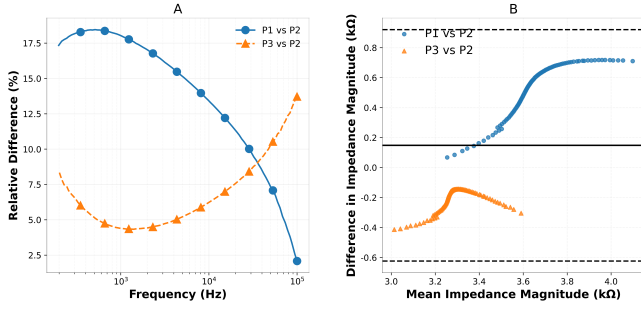


Fig. 12. (A) %RD in impedance magnitude for the P1–P2 and P3–P2 comparisons obtained with the Linear Arrangement 2. (B) Bland–Altman analysis of the corresponding measurements, with a mean bias of 148.28 Ω and limits of agreement defined as ± 1.96 SD.

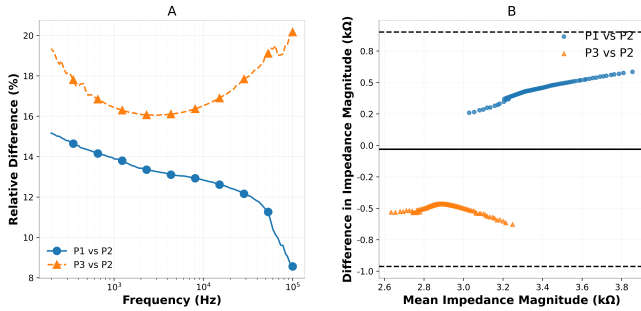


Fig. 13. (A) %RD in impedance magnitude for the P1–P2 and P3–P2 comparisons obtained with the Linear Arrangement 3. (B) Bland–Altman analysis of the corresponding measurements, with a mean bias of -30.95 Ω and limits of agreement defined as ± 1.96 SD.

observed for the P3–P2 comparison, varying between approximately 8% and 14%. These results indicate that the Linear Arrangement 2 configuration exhibits a limited sensitivity to the variations in middle layer resistivity, particularly for the P3–P2 comparison.

The corresponding Bland–Altman plot shows a clear separation between the two phantom comparisons. The P3–P2 data are predominantly distributed below the zero-difference line and, consequently, below the calculated bias of 148.28 Ω , reflecting the lower impedance of Phantom P3 resulting from its higher middle layer conductivity. In contrast, the P1–P2 comparison is located almost entirely above the bias and is concentrated over mean impedance values ranging from approximately 3200 Ω to 4000 Ω . The distinct distribution of both clusters indicates that the Linear Arrangement 2 configuration consistently differentiates between increases and decreases in the middle layer resistivity with respect to the reference phantom.

For the Linear Arrangement 3 electrode configuration, the relative difference remained below 20% throughout the analyzed frequency range. The P1–P2 comparison exhibited a gradual decrease, tending toward zero at higher frequencies, whereas the P3–P2 comparison showed the opposite behavior, with the relative difference increasing above approximately 10 kHz. This divergence between the two comparisons at

higher frequencies indicates that the Linear Arrangement 3 configuration is capable of distinguishing changes in middle layer resistivity, particularly in the upper portion of the analyzed spectrum.

The corresponding Bland–Altman plot yielded a bias of -30.95 Ω . Although the overall bias was close to zero, this result reflects the compensation between the positive differences observed for the P1–P2 comparison and the negative differences associated with the P3–P2 comparison, rather than the absence of systematic differences. The P1–P2 data form a nearly constant positive cluster, with differences ranging from approximately 250 Ω to 600 Ω , whereas the P3–P2 comparison remains consistently below the zero-difference line, with negative differences spanning approximately 150 Ω . The clear separation between the two clusters demonstrates that the Linear Arrangement 3 configuration consistently discriminates between increases and decreases in middle layer resistivity relative to the reference phantom.

IV. DISCUSSION AND CONCLUSIONS

The calibration procedure revealed substantial differences between agar and gelatin based formulations. Agar exhibited high intrinsic conductivity, which limited the attainable resistivity range, particularly for epidermis-mimicking layers. Increasing the agar concentration improved mechanical strength but further increased conductivity. In contrast, gelatin based formulations simultaneously provided adequate mechanical stability and the target resistivity range, making them suitable for multilayer phantom fabrication.

The experimental results demonstrated that electrode geometry strongly affects the impedance response of multilayer structures. Although all configurations distinguished the three phantoms, the magnitude and frequency dependence of the measured differences varied considerably. These variations indicate that the electrode arrangement modifies the current distribution and consequently the contribution of each layer to the measured impedance.

The coplanar configurations exhibited moderate sensitivity and progressive convergence of the impedance spectra with increasing frequency. In contrast, the linear arrangements showed markedly different performances. Linear Arrangement 1 produced the highest sensitivity to middle layer resistivity variations, exhibiting relative differences exceeding 100% at low frequencies. Linear Arrangements 2 and 3 displayed lower sensitivities but more stable responses across the analyzed spectrum.

The combined use of relative difference (%RD) and Bland–Altman analysis provided complementary information regarding the performance of each electrode configuration. The %RD quantified the sensitivity to changes in middle layer resistivity, whereas Bland–Altman analysis demonstrated that the observed differences were systematic and reproducible rather than arising from random variability.

The use of tissue mimicking phantoms limits direct extrapolation to biological tissue. Electrode skin contact, tissue heterogeneity, hydration, temperature fluctuations, and motion

were not evaluated. Complementary in vivo measurements using different electrode types were reported separately in [19]. The present study evaluated a defined set of electrode configurations without numerical simulations or geometry optimization.

From a physical perspective, the observed behavior can be attributed to the current density distribution imposed by each tetrapolar geometry. Small modifications in electrode spacing and arrangement produced substantial changes in sensing depth and sensitivity to subsurface electrical properties.

Overall, the proposed multilayer phantoms provided a controlled platform for assessing electrode configurations and variations in the electrical properties of subsurface mimicking skin layers. The methodology enables controlled modifications of individual layers for superficial bioimpedance measurements.

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APPENDIX D – EXPERIMENTAL IMPEDANCE MEASUREMENTS

This appendix contains the complete experimental impedance measurements acquired using the EmStat4R potentiostat. The data correspond to the three multilayer phantoms (P1, P2, and P3) evaluated with the five electrode configurations over the frequency range from 200 Hz to 100 kHz. Each table reports the frequency, phase angle, impedance magnitude, and the real and imaginary components of the measured impedance.

Table 12 – EIS measurements for Phantom P1 using the Linear Arrangement 1 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
100000,00	26,99	2887,71	2573,19	1310,55
94339,62	25,65	2925,77	2637,47	1266,44
87719,30	24,30	2961,85	2699,39	1218,96
83333,34	23,30	2987,55	2743,94	1181,62
78125,00	22,20	3020,69	2796,74	1141,40
73529,42	21,19	3047,66	2841,64	1101,49
68493,15	20,00	3073,42	2888,01	1051,33
64102,56	18,99	3102,11	2933,32	1009,31
60240,96	18,09	3124,65	2970,22	970,16
56818,18	17,32	3145,20	3002,57	936,40
53191,49	16,40	3169,66	3040,63	895,17
50000,00	15,65	3184,78	3066,64	859,39
47169,81	14,81	3176,59	3071,13	811,72
44247,79	14,12	3192,26	3095,79	778,85
41666,67	13,49	3205,79	3117,30	748,03
39062,50	12,85	3220,77	3140,14	716,15
36496,35	12,22	3235,21	3161,93	684,72
34482,76	11,70	3245,12	3177,67	658,21
32258,06	11,18	3256,53	3194,77	631,19
30303,03	10,69	3267,56	3210,85	606,10
28571,43	10,26	3277,59	3225,19	583,74
26737,97	9,80	3287,63	3239,63	559,79
25125,63	9,38	3297,07	3252,96	537,50
23584,91	9,00	3306,16	3265,45	517,22
22123,89	8,30	3314,52	3279,81	478,41
20833,33	8,00	3322,34	3289,99	462,51
19531,25	7,70	3330,53	3300,50	446,20
18382,35	7,43	3338,25	3310,19	431,91
17241,38	7,17	3346,48	3320,31	417,73
16181,23	6,93	3353,86	3329,34	404,82
15197,57	6,71	3361,31	3338,29	392,70
14285,71	6,50	3368,39	3346,71	381,52
13404,83	6,31	3375,56	3355,08	371,21
12594,46	6,14	3382,48	3363,11	361,50
11820,33	5,97	3389,64	3371,24	352,71
11111,11	5,83	3396,36	3378,82	344,74
10438,41	5,69	3403,12	3386,36	337,31
9803,92	5,56	3409,82	3393,77	330,54
9208,10	5,45	3416,67	3401,23	324,45
8650,52	5,35	3423,39	3408,48	319,14
8116,88	5,26	3430,14	3415,71	314,34
7621,95	5,17	3436,91	3422,90	309,98
7163,32	5,10	3443,62	3429,97	306,30
6729,48	5,04	3450,45	3437,10	303,25
6313,13	4,99	3457,45	3444,33	300,96
5931,20	4,95	3464,47	3451,55	298,96
5567,93	4,92	3471,73	3458,96	297,47

Table 13 – EIS measurements for Phantom P1 using the Linear Arrangement 1 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
5230,13	4,89	3478,64	3465,97	296,60
4911,59	4,88	3486,33	3473,72	296,28
4616,81	4,87	3492,69	3480,09	296,40
4332,76	4,87	3499,63	3487,00	297,08
4071,66	4,88	3507,09	3494,39	298,11
3822,63	4,89	3514,70	3501,88	299,87
3589,38	4,92	3522,16	3509,19	301,91
3371,54	4,95	3529,31	3516,16	304,37
3166,56	4,99	3537,36	3523,96	307,56
2974,42	5,03	3544,49	3530,84	310,74
2793,30	5,08	3551,93	3537,97	314,64
2463,05	5,22	3568,25	3553,48	324,39
2172,97	5,37	3585,34	3569,58	335,79
2040,82	5,46	3594,19	3577,88	342,06
1916,44	5,57	3603,44	3586,45	349,49
1799,86	5,66	3613,17	3595,55	356,38
1690,33	5,78	3622,74	3604,33	364,86
1490,76	6,03	3642,91	3622,76	382,68
1400,17	6,17	3653,75	3632,61	392,47
1314,75	6,31	3664,07	3641,85	402,88
1234,87	6,46	3675,37	3652,00	413,81
1159,82	6,63	3687,31	3662,65	425,70
1089,32	6,80	3699,20	3673,17	438,09
1022,91	6,98	3710,90	3683,40	451,01
960,61	7,16	3724,63	3695,55	464,55
902,20	7,37	3738,10	3707,25	479,22
847,31	7,57	3751,37	3718,69	494,08
795,80	7,79	3765,65	3730,92	510,23
747,38	8,03	3782,03	3744,99	528,02
659,20	8,50	3812,81	3770,90	563,77
619,12	8,77	3829,62	3784,89	583,65
581,40	9,04	3846,77	3798,96	604,60
546,03	9,31	3864,07	3813,17	625,11
512,82	9,61	3882,37	3827,90	648,05
477,27	9,96	3908,68	3849,79	675,92
456,52	10,19	3923,42	3861,55	694,03
420,00	10,63	3950,64	3882,87	728,63
403,85	10,83	3963,56	3892,93	744,93
375,00	11,24	3990,54	3913,96	777,99
350,00	11,65	4015,56	3932,90	810,59
328,13	12,03	4042,36	3953,64	842,27
308,82	12,38	4071,06	3976,45	872,61
291,67	12,76	4095,41	3994,34	904,23
276,32	13,10	4119,43	4012,27	933,49
256,10	13,62	4155,48	4038,69	978,26
244,19	13,95	4178,17	4054,89	1007,44
228,26	14,45	4214,66	4081,31	1051,82
214,29	14,93	4249,43	4106,05	1094,52
198,11	15,52	4310,53	4153,32	1153,50

Table 14 – EIS measurements for Phantom P1 using the Linear Arrangement 2 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
100000,00	27,99	3288,34	2903,62	1543,45
94339,62	26,73	3330,26	2974,47	1497,73
87719,30	25,27	3378,34	3054,98	1442,32
83333,34	24,21	3408,73	3108,90	1397,92
78125,00	22,95	3447,71	3174,77	1344,46
73529,42	21,83	3478,93	3229,35	1293,91
68493,15	20,66	3509,94	3284,18	1238,49
64102,56	19,55	3541,48	3337,26	1185,23
60240,96	18,58	3569,50	3383,48	1137,28
56818,18	17,70	3591,81	3421,69	1092,31
53191,49	16,70	3613,80	3461,32	1038,64
50000,00	15,95	3631,34	3491,61	997,63
47169,81	14,98	3617,04	3494,20	934,64
44247,79	14,21	3634,29	3523,13	891,98
41666,67	13,53	3648,33	3547,06	853,62
39062,50	12,84	3664,18	3572,60	814,10
36496,35	12,15	3677,40	3595,01	774,06
34482,76	11,59	3689,87	3614,62	741,40
32258,06	10,98	3701,93	3634,16	705,13
30303,03	10,46	3714,16	3652,42	674,38
28571,43	9,99	3723,78	3667,29	646,14
26737,97	9,48	3733,44	3682,42	615,10
25125,63	9,04	3743,05	3696,60	587,84
23584,91	8,59	3751,00	3708,91	560,30
22123,89	7,84	3760,67	3725,50	513,15
20833,33	7,50	3768,40	3736,14	492,06
19531,25	7,17	3775,90	3746,37	471,33
18382,35	6,87	3782,94	3755,77	452,65
17241,38	6,58	3790,05	3765,11	434,04
16181,23	6,30	3796,84	3773,89	416,84
15197,57	6,05	3803,26	3782,07	400,92
14285,71	5,82	3809,55	3789,93	386,11
13404,83	5,59	3815,66	3797,49	371,97
12594,46	5,39	3821,56	3804,67	358,83
11820,33	5,19	3827,40	3811,68	346,51
11111,11	5,02	3833,06	3818,37	335,28
10438,41	4,85	3838,41	3824,64	324,85
9803,92	4,70	3844,10	3831,18	314,94
9208,10	4,56	3849,32	3837,14	305,99
8650,52	4,43	3854,81	3843,30	297,66
8116,88	4,31	3860,17	3849,28	289,79
7621,95	4,20	3865,43	3855,06	282,87
7163,32	4,10	3870,69	3860,80	276,42
6729,48	4,01	3875,82	3866,35	270,81
6313,13	3,92	3881,12	3872,03	265,54
5931,20	3,85	3886,14	3877,37	260,90
5567,93	3,79	3891,80	3883,30	257,03

Table 15 – EIS measurements for Phantom P1 using the Linear Arrangement 2 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
5230,13	3,72	3897,00	3888,77	253,17
4911,59	3,68	3902,11	3894,07	250,40
4616,81	3,63	3907,47	3899,61	247,64
4332,76	3,60	3912,98	3905,25	245,75
4071,66	3,57	3918,39	3910,77	244,22
3822,63	3,55	3923,94	3916,41	242,97
3589,38	3,54	3929,57	3922,09	242,34
3371,54	3,53	3935,30	3927,85	242,06
3166,56	3,52	3941,03	3933,59	242,12
2974,42	3,52	3947,21	3939,74	242,68
2793,30	3,54	3953,17	3945,64	243,78
2623,30	3,55	3959,28	3951,69	245,09
2463,05	3,57	3965,82	3958,12	246,92
2172,97	3,63	3978,30	3970,32	251,87
2040,82	3,67	3985,08	3976,93	254,74
1916,44	3,71	3991,75	3983,40	258,08
1799,86	3,75	3998,47	3989,89	261,84
1690,33	3,81	4005,66	3996,81	266,12
1587,30	3,87	4012,98	4003,83	270,92
1400,17	4,01	4027,78	4017,92	281,65
1314,75	4,09	4035,20	4024,92	287,87
1234,87	4,17	4043,67	4032,96	294,08
1159,82	4,26	4051,59	4040,38	301,23
1089,32	4,35	4059,48	4047,76	308,25
960,61	4,56	4077,37	4064,46	324,23
902,20	4,67	4086,30	4072,73	332,71
847,31	4,79	4095,43	4081,13	341,96
795,80	4,92	4105,04	4089,93	351,97
747,38	5,06	4116,08	4100,06	362,81
701,85	5,20	4126,04	4109,08	373,77
659,20	5,34	4136,91	4118,94	385,21
619,12	5,49	4147,99	4128,95	397,04
581,40	5,66	4159,00	4138,72	410,22
512,82	6,00	4182,13	4159,19	437,48
477,27	6,22	4199,00	4174,29	454,86
456,52	6,36	4208,44	4182,51	466,44
420,00	6,64	4226,48	4198,17	488,39
403,85	6,76	4233,70	4204,28	498,22
375,00	7,02	4251,82	4219,94	519,68
350,00	7,27	4267,49	4233,15	540,23
328,13	7,51	4284,03	4247,25	560,12
308,82	7,74	4302,24	4263,08	579,13
291,67	7,98	4317,28	4275,49	599,28
276,32	8,21	4331,63	4287,27	618,28
256,10	8,54	4353,04	4304,80	646,28
244,19	8,77	4366,84	4315,77	665,86
228,26	9,11	4389,65	4334,31	694,83
214,29	9,42	4411,69	4352,22	721,96
198,11	9,84	4453,18	4387,62	761,31

Table 16 – EIS measurements for Phantom P1 using the Linear Arrangement 3 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
100000,00	27,99	3288,34	2903,62	1543,45
94339,62	26,73	3330,26	2974,47	1497,73
87719,30	25,27	3378,34	3054,98	1442,32
83333,34	24,21	3408,73	3108,90	1397,92
78125,00	22,95	3447,71	3174,77	1344,46
73529,42	21,83	3478,93	3229,35	1293,91
68493,15	20,66	3509,94	3284,18	1238,49
64102,56	19,55	3541,48	3337,26	1185,23
60240,96	18,58	3569,50	3383,48	1137,28
56818,18	17,70	3591,81	3421,69	1092,31
53191,49	16,70	3613,80	3461,32	1038,64
50000,00	15,95	3631,34	3491,61	997,63
47169,81	14,98	3617,04	3494,20	934,64
44247,79	14,21	3634,29	3523,13	891,98
41666,67	13,53	3648,33	3547,06	853,62
39062,50	12,84	3664,18	3572,60	814,10
36496,35	12,15	3677,40	3595,01	774,06
34482,76	11,59	3689,87	3614,62	741,40
32258,06	10,98	3701,93	3634,16	705,13
30303,03	10,46	3714,16	3652,42	674,38
28571,43	9,99	3723,78	3667,29	646,14
26737,97	9,48	3733,44	3682,42	615,10
25125,63	9,04	3743,05	3696,60	587,84
23584,91	8,59	3751,00	3708,91	560,30
22123,89	7,84	3760,67	3725,50	513,15
20833,33	7,50	3768,40	3736,14	492,06
19531,25	7,17	3775,90	3746,37	471,33
18382,35	6,87	3782,94	3755,77	452,65
17241,38	6,58	3790,05	3765,11	434,04
16181,23	6,30	3796,84	3773,89	416,84
15197,57	6,05	3803,26	3782,07	400,92
14285,71	5,82	3809,55	3789,93	386,11
13404,83	5,59	3815,66	3797,49	371,97
12594,46	5,39	3821,56	3804,67	358,83
11820,33	5,19	3827,40	3811,68	346,51
11111,11	5,02	3833,06	3818,37	335,28
10438,41	4,85	3838,41	3824,64	324,85
9803,92	4,70	3844,10	3831,18	314,94
9208,10	4,56	3849,32	3837,14	305,99
8650,52	4,43	3854,81	3843,30	297,66
8116,88	4,31	3860,17	3849,28	289,79
7621,95	4,20	3865,43	3855,06	282,87
7163,32	4,10	3870,69	3860,80	276,42
6729,48	4,01	3875,82	3866,35	270,81
6313,13	3,92	3881,12	3872,03	265,54
5931,20	3,85	3886,14	3877,37	260,90
5567,93	3,79	3891,80	3883,30	257,03

Table 17 – EIS measurements for Phantom P1 using the Linear Arrangement 3 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
5230,13	3,72	3897,00	3888,77	253,17
4911,59	3,68	3902,11	3894,07	250,40
4616,81	3,63	3907,47	3899,61	247,64
4332,76	3,60	3912,98	3905,25	245,75
4071,66	3,57	3918,39	3910,77	244,22
3822,63	3,55	3923,94	3916,41	242,97
3589,38	3,54	3929,57	3922,09	242,34
3371,54	3,53	3935,30	3927,85	242,06
3166,56	3,52	3941,03	3933,59	242,12
2974,42	3,52	3947,21	3939,74	242,68
2793,30	3,54	3953,17	3945,64	243,78
2623,30	3,55	3959,28	3951,69	245,09
2463,05	3,57	3965,82	3958,12	246,92
2313,74	3,60	3971,69	3963,87	249,05
2172,97	3,63	3978,30	3970,32	251,87
1916,44	3,71	3991,75	3983,40	258,08
1799,86	3,75	3998,47	3989,89	261,84
1690,33	3,81	4005,66	3996,81	266,12
1490,76	3,94	4019,70	4010,21	276,01
1400,17	4,01	4027,78	4017,92	281,65
1314,75	4,09	4035,20	4024,92	287,87
1159,82	4,26	4051,59	4040,38	301,23
1089,32	4,35	4059,48	4047,76	308,25
960,61	4,56	4077,37	4064,46	324,23
902,20	4,67	4086,30	4072,73	332,71
847,31	4,79	4095,43	4081,13	341,96
795,80	4,92	4105,04	4089,93	351,97
747,38	5,06	4116,08	4100,06	362,81
701,85	5,20	4126,04	4109,08	373,77
659,20	5,34	4136,91	4118,94	385,21
619,12	5,49	4147,99	4128,95	397,04
581,40	5,66	4159,00	4138,72	410,22
546,03	5,82	4169,57	4148,05	423,09
512,82	6,00	4182,13	4159,19	437,48
477,27	6,22	4199,00	4174,29	454,86
456,52	6,36	4208,44	4182,51	466,44
420,00	6,64	4226,48	4198,17	488,39
403,85	6,76	4233,70	4204,28	498,22
375,00	7,02	4251,82	4219,94	519,68
350,00	7,27	4267,49	4233,15	540,23
328,13	7,51	4284,03	4247,25	560,12
308,82	7,74	4302,24	4263,08	579,13
291,67	7,98	4317,28	4275,49	599,28
276,32	8,21	4331,63	4287,27	618,28
256,10	8,54	4353,04	4304,80	646,28
244,19	8,77	4366,84	4315,77	665,86
228,26	9,11	4389,65	4334,31	694,83
214,29	9,42	4411,69	4352,22	721,96
198,11	9,84	4453,18	4387,62	761,31

Table 18 – EIS measurements for Phantom P1 using the Coplanar Arrangement 1 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
100000,00	25,09	3584,91	3246,72	1519,99
94339,62	23,77	3622,18	3314,92	1459,97
87719,30	22,30	3665,17	3391,11	1390,62
83333,34	21,31	3690,42	3438,15	1341,01
78125,00	20,07	3722,73	3496,58	1277,76
73529,42	19,05	3746,75	3541,45	1223,21
68493,15	17,83	3773,56	3592,28	1155,55
64102,56	16,83	3792,73	3630,34	1097,94
60240,96	15,85	3812,54	3667,50	1041,59
56818,18	15,14	3828,32	3695,50	999,66
53191,49	14,27	3839,56	3721,04	946,64
50000,00	13,37	3858,60	3753,97	892,46
47169,81	12,56	3829,31	3737,70	832,59
44247,79	11,86	3839,85	3757,81	789,49
41666,67	11,26	3850,42	3776,31	751,81
39062,50	10,61	3858,71	3792,75	710,45
36496,35	9,98	3867,95	3809,47	670,05
34482,76	9,49	3875,33	3822,26	639,15
32258,06	8,99	3882,06	3834,33	606,83
30303,03	8,49	3888,15	3845,53	574,09
28571,43	8,08	3894,08	3855,40	547,45
26737,97	7,65	3897,97	3863,29	518,81
25125,63	7,26	3904,01	3872,70	493,42
23584,91	6,91	3907,75	3879,39	469,93
22123,89	6,18	3914,00	3891,23	421,60
20833,33	5,90	3917,74	3896,97	402,93
19531,25	5,62	3921,67	3902,82	384,04
18382,35	5,37	3925,45	3908,20	367,65
17241,38	5,13	3929,06	3913,30	351,60
16181,23	4,91	3932,31	3917,86	336,77
15197,57	4,71	3935,57	3922,27	323,25
14285,71	4,53	3938,61	3926,31	311,03
13404,83	4,35	3941,66	3930,29	299,20
12594,46	4,20	3944,60	3934,00	288,98
11820,33	4,05	3947,62	3937,75	279,05
11111,11	3,92	3950,73	3941,47	270,35
10438,41	3,81	3953,48	3944,75	262,52
9803,92	3,70	3956,53	3948,28	255,43
9208,10	3,61	3959,42	3951,58	249,04
8650,52	3,52	3962,38	3954,89	243,46
8116,88	3,45	3965,20	3958,03	238,45
7621,95	3,38	3968,54	3961,62	234,32
7163,32	3,33	3971,55	3964,83	230,89
6729,48	3,29	3974,57	3968,02	228,03
6313,13	3,25	3978,02	3971,61	225,71
5931,20	3,23	3981,29	3974,99	224,01
5567,93	3,21	3984,59	3978,35	222,94

Table 19 – EIS measurements for Phantom P1 using the Coplanar Arrangement 1 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
5230,13	3,20	3988,21	3982,00	222,38
4911,59	3,20	3991,85	3985,63	222,77
4616,81	3,20	3995,40	3989,16	223,31
4332,76	3,22	3999,39	3993,07	224,66
4071,66	3,24	4003,30	3996,90	226,38
3822,63	3,27	4007,68	4001,14	228,89
3589,38	3,31	4011,92	4005,24	231,55
3371,54	3,36	4016,41	4009,52	235,10
2974,42	3,47	4026,28	4018,90	243,60
2623,30	3,61	4036,50	4028,49	254,22
2463,05	3,70	4042,10	4033,70	260,52
2313,74	3,78	4047,86	4039,05	266,99
2172,97	3,88	4053,99	4044,70	274,33
2040,82	3,99	4060,29	4050,46	282,34
1916,44	4,10	4066,79	4056,40	290,54
1799,86	4,22	4073,48	4062,44	299,77
1690,33	4,35	4080,80	4069,04	309,58
1587,30	4,48	4088,37	4075,87	319,46
1490,76	4,63	4095,92	4082,54	330,87
1400,17	4,78	4104,27	4089,97	342,35
1234,87	5,12	4121,93	4105,46	368,08
1159,82	5,30	4131,49	4113,83	381,55
1089,32	5,48	4140,66	4121,71	395,78
1022,91	5,68	4150,70	4130,29	411,14
960,61	5,90	4161,88	4139,85	427,69
902,20	6,11	4172,88	4149,16	444,24
795,80	6,58	4196,85	4169,22	480,80
747,38	6,83	4211,03	4181,15	500,77
701,85	7,10	4224,38	4191,98	522,20
659,20	7,37	4238,17	4203,14	543,81
619,12	7,66	4252,88	4214,92	566,96
581,40	7,95	4268,04	4226,97	590,66
546,03	8,26	4284,07	4239,59	615,71
512,82	8,59	4300,88	4252,67	642,18
477,27	8,98	4326,19	4273,14	675,46
456,52	9,23	4339,51	4283,34	695,98
420,00	9,72	4364,15	4301,45	737,14
403,85	9,96	4377,21	4311,29	756,80
375,00	10,41	4403,37	4330,83	795,98
350,00	10,87	4428,78	4349,32	835,18
328,13	11,30	4454,41	4368,04	872,93
308,82	11,68	4481,71	4388,96	907,05
291,67	12,10	4505,55	4405,40	944,67
276,32	12,52	4529,16	4421,42	982,01
256,10	13,11	4565,62	4446,68	1035,34
244,19	13,50	4588,86	4462,06	1071,27
228,26	14,06	4624,33	4485,82	1123,33
214,29	14,61	4659,58	4508,82	1175,71
198,11	15,33	4725,19	4557,03	1249,37

Table 20 – EIS measurements for Phantom P1 using the Coplanar Arrangement 2 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
100000,00	28,30	3342,59	2943,16	1584,51
94339,62	26,97	3392,19	3023,26	1538,46
87719,30	25,46	3443,04	3108,63	1480,17
83333,34	24,35	3481,81	3172,10	1435,54
78125,00	23,16	3524,41	3240,39	1386,14
73529,42	22,02	3556,20	3296,74	1333,42
68493,15	20,83	3588,21	3353,73	1275,82
64102,56	19,68	3614,40	3403,33	1217,07
60240,96	18,78	3646,50	3452,32	1174,10
56818,18	17,88	3667,13	3490,03	1125,85
53191,49	16,89	3686,02	3527,01	1070,93
50000,00	16,05	3712,19	3567,55	1026,11
47169,81	15,09	3693,28	3565,89	961,64
44247,79	14,32	3711,45	3596,14	917,96
41666,67	13,63	3726,00	3621,12	877,83
39062,50	12,94	3742,89	3647,91	837,85
36496,35	12,23	3760,06	3674,70	796,67
34482,76	11,67	3770,52	3692,59	762,61
32258,06	11,08	3782,64	3712,08	727,20
30303,03	10,53	3794,24	3730,32	693,50
28571,43	10,07	3803,35	3744,73	665,14
26737,97	9,56	3811,23	3758,33	632,83
25125,63	9,11	3819,85	3771,65	604,90
23584,91	8,70	3829,44	3785,42	578,93
22123,89	7,94	3838,18	3801,38	530,18
20833,33	7,61	3846,20	3812,35	509,17
19531,25	7,28	3853,12	3822,08	488,09
18382,35	6,99	3859,38	3830,71	469,56
17241,38	6,70	3865,96	3839,54	451,25
16181,23	6,45	3873,05	3848,52	435,26
15197,57	6,22	3879,37	3856,52	420,43
14285,71	6,01	3885,23	3863,88	406,72
13404,83	5,80	3892,54	3872,59	393,64
12594,46	5,61	3895,61	3876,93	381,08
11820,33	5,43	3901,40	3883,87	369,50
11111,11	5,28	3907,15	3890,58	359,53
10438,41	5,14	3912,53	3896,80	350,52
9803,92	5,01	3917,78	3902,80	342,27
9208,10	4,90	3923,73	3909,40	334,95
8650,52	4,80	3930,13	3916,35	328,80
8116,88	4,71	3935,82	3922,52	323,28
7621,95	4,63	3940,81	3927,93	318,41
7163,32	4,57	3946,06	3933,50	314,55
6729,48	4,52	3951,53	3939,24	311,52
6313,13	4,48	3956,89	3944,81	309,01
5931,20	4,45	3962,64	3950,70	307,38
5567,93	4,43	3969,32	3957,48	306,42

Table 21 – EIS measurements for Phantom P1 using the Coplanar Arrangement 2 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
5230,13	4,45	3988,37	3976,38	309,12
4911,59	4,43	4004,08	3992,11	309,34
4616,81	4,43	4015,13	4003,15	309,85
4332,76	4,13	4052,78	4042,27	291,75
4071,66	3,79	4072,05	4063,14	269,25
3822,63	3,87	4077,53	4068,25	275,00
3589,38	4,21	4075,55	4064,58	298,86
3371,54	4,42	4073,39	4061,27	314,03
2974,42	4,76	4045,91	4031,95	335,76
2793,30	4,84	4051,13	4036,69	341,75
2623,30	4,92	4057,99	4043,01	348,35
2463,05	5,03	4072,27	4056,59	356,99
2313,74	5,14	4080,01	4063,62	365,35
2172,97	5,25	4088,96	4071,81	374,10
2040,82	5,37	4098,72	4080,75	383,33
1916,44	5,50	4109,07	4090,16	393,74
1799,86	5,63	4121,32	4101,42	404,59
1690,33	5,78	4135,09	4114,10	416,11
1587,30	5,92	4150,99	4128,88	427,90
1490,76	5,99	4174,09	4151,32	435,33
1400,17	6,14	4179,00	4155,02	447,03
1314,75	6,42	4171,99	4145,82	466,51
1159,82	6,57	4240,30	4212,42	485,42
1089,32	6,64	4259,55	4231,00	492,36
1022,91	6,85	4273,26	4242,74	509,81
960,61	7,07	4288,84	4256,19	528,20
902,20	7,28	4304,86	4270,13	545,73
847,31	7,53	4320,59	4283,32	566,28
795,80	8,02	4320,21	4277,97	602,67
747,38	8,28	4350,27	4304,89	626,69
701,85	8,41	4372,12	4325,14	639,22
659,20	8,67	4390,32	4340,20	661,48
619,12	8,98	4410,53	4356,52	688,09
581,40	9,30	4430,49	4372,28	715,79
512,82	10,28	4440,99	4369,74	792,35
477,27	10,64	4472,21	4395,31	825,81
456,52	10,88	4496,71	4415,94	848,46
420,00	11,18	4555,84	4469,41	883,20
403,85	11,38	4571,79	4481,85	902,38
350,00	12,32	4636,70	4529,87	989,57
328,13	12,79	4668,32	4552,45	1033,64
308,82	13,13	4706,28	4583,22	1069,22
291,67	13,57	4741,92	4609,55	1112,59
276,32	14,21	4741,14	4596,11	1163,70
256,10	14,80	4770,83	4612,50	1218,88
244,19	15,27	4814,57	4644,67	1267,73
228,26	15,72	4869,14	4687,09	1318,98
214,29	16,37	4900,33	4701,78	1380,77
198,11	17,06	4978,79	4759,80	1460,39

Table 22 – EIS measurements for Phantom P2 using the Linear Arrangement 1 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
100000,00	21,92	2470,12	2291,57	922,06
94339,62	20,87	2495,91	2332,11	889,28
87719,30	19,73	2520,09	2372,18	850,66
83333,34	18,85	2536,34	2400,25	819,64
78125,00	17,91	2558,73	2434,69	787,00
73529,42	17,05	2575,10	2461,87	755,21
68493,15	16,13	2592,07	2490,03	720,16
64102,56	15,22	2610,20	2518,60	685,41
60240,96	14,43	2626,37	2543,50	654,53
56818,18	13,72	2643,63	2568,22	626,91
53191,49	13,02	2657,64	2589,30	598,81
50000,00	12,33	2669,13	2607,56	570,00
47169,81	11,56	2661,13	2607,16	533,21
44247,79	10,91	2671,91	2623,61	505,70
41666,67	10,36	2680,83	2637,11	482,19
39062,50	9,78	2689,16	2650,10	456,70
36496,35	9,25	2697,15	2662,08	433,54
34482,76	8,77	2704,42	2672,78	412,45
32258,06	8,26	2711,77	2683,63	389,66
30303,03	7,81	2717,77	2692,54	369,44
28571,43	7,41	2723,52	2700,77	351,27
26737,97	7,00	2727,30	2706,95	332,57
25125,63	6,61	2732,80	2714,63	314,60
23584,91	6,26	2736,53	2720,24	298,21
22123,89	5,53	2743,55	2730,81	264,16
20833,33	5,27	2746,21	2734,60	252,30
19531,25	4,96	2750,16	2739,87	237,77
18382,35	4,71	2752,85	2743,54	226,21
17241,38	4,48	2755,46	2747,04	215,28
16181,23	4,26	2757,57	2749,93	205,06
15197,57	4,08	2759,54	2752,55	196,30
14285,71	3,90	2761,50	2755,11	187,73
13404,83	3,73	2763,09	2757,22	179,95
12594,46	3,58	2764,78	2759,37	172,86
11820,33	3,45	2766,31	2761,31	166,24
11111,11	3,31	2768,05	2763,42	160,04
10438,41	3,20	2769,49	2765,16	154,72
9803,92	3,09	2771,06	2767,03	149,43
9208,10	2,99	2772,54	2768,76	144,65
8650,52	2,90	2773,97	2770,42	140,37
8116,88	2,82	2775,46	2772,09	136,75
7621,95	2,76	2776,85	2773,63	133,68
7163,32	2,71	2778,43	2775,33	131,24
6729,48	2,66	2779,76	2776,76	129,13
6313,13	2,62	2781,34	2778,43	127,24
5931,20	2,59	2782,83	2779,98	125,74
5567,93	2,57	2784,41	2781,60	125,09

Table 23 – EIS measurements for Phantom P2 using the Linear Arrangement 1 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
5230,13	2,56	2786,02	2783,25	124,28
4911,59	2,55	2787,89	2785,13	124,02
4616,81	2,55	2789,46	2786,70	123,88
4332,76	2,56	2791,53	2788,75	124,47
4071,66	2,57	2793,20	2790,38	125,39
3822,63	2,59	2795,16	2792,30	126,50
3589,38	2,62	2797,45	2794,53	127,87
3371,54	2,66	2799,31	2796,30	129,90
3166,56	2,71	2801,79	2798,67	132,37
2974,42	2,75	2803,78	2800,55	134,55
2623,30	2,87	2809,23	2805,71	140,61
2313,74	3,02	2814,51	2810,60	148,33
2172,97	3,11	2817,83	2813,68	152,88
1916,44	3,30	2824,35	2819,66	162,68
1799,86	3,41	2827,69	2822,69	168,16
1690,33	3,53	2831,14	2825,78	174,15
1490,76	3,77	2839,15	2833,00	186,77
1400,17	3,91	2843,41	2836,79	193,98
1314,75	4,06	2847,80	2840,66	201,61
1234,87	4,21	2852,56	2844,87	209,41
1159,82	4,37	2857,37	2849,07	217,61
1089,32	4,54	2862,43	2853,46	226,43
1022,91	4,72	2867,34	2857,61	236,05
960,61	4,92	2873,64	2863,06	246,38
902,20	5,11	2879,30	2867,87	256,32
847,31	5,32	2885,53	2873,10	267,56
795,80	5,54	2892,13	2878,63	279,08
747,38	5,77	2899,66	2885,00	291,30
701,85	6,02	2907,10	2891,06	305,02
659,20	6,27	2914,77	2897,32	318,41
619,12	6,53	2922,73	2903,78	332,25
581,40	6,81	2931,19	2910,54	347,37
546,03	7,10	2939,74	2917,23	363,12
512,82	7,40	2948,57	2924,02	379,71
477,27	7,77	2962,27	2935,10	400,33
456,52	7,99	2969,70	2940,90	412,57
420,00	8,45	2984,55	2952,14	438,64
403,85	8,68	2991,25	2956,96	451,63
375,00	9,12	3005,62	2967,63	476,39
350,00	9,56	3019,61	2977,68	501,49
328,13	9,98	3033,64	2987,77	525,53
308,82	10,34	3049,82	3000,34	547,15
291,67	10,74	3063,70	3010,08	570,67
276,32	11,13	3076,89	3019,00	594,04
256,10	11,71	3097,16	3032,68	628,70
244,19	12,10	3110,52	3041,38	652,20
228,26	12,65	3130,80	3054,80	685,61
214,29	13,18	3150,50	3067,52	718,30
198,11	13,85	3190,92	3098,21	763,59

Table 24 – EIS measurements for Phantom P2 using the Linear Arrangement 2 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
100000,00	20,93	3220,62	3008,11	1150,53
94339,62	19,79	3245,54	3053,81	1099,00
87719,30	18,44	3268,04	3100,19	1033,88
83333,34	17,52	3281,77	3129,45	988,20
78125,00	16,47	3303,49	3167,91	936,70
73529,42	15,56	3316,31	3194,73	889,75
68493,15	14,42	3330,95	3225,99	829,59
64102,56	13,55	3340,17	3247,22	782,50
60240,96	12,72	3353,16	3270,85	738,41
56818,18	11,94	3358,22	3285,52	694,96
53191,49	11,16	3366,68	3302,97	651,82
50000,00	10,49	3373,85	3317,48	614,17
47169,81	9,71	3349,68	3301,64	565,24
44247,79	9,07	3354,53	3312,54	529,09
41666,67	8,52	3357,93	3320,84	497,69
39062,50	7,96	3361,15	3328,74	465,64
36496,35	7,44	3363,19	3334,86	435,57
34482,76	6,97	3365,26	3340,41	408,25
32258,06	6,51	3367,71	3346,03	381,55
30303,03	6,10	3368,49	3349,43	357,80
28571,43	5,74	3368,56	3351,67	336,95
26737,97	5,36	3368,95	3354,24	314,46
25125,63	5,01	3369,40	3356,54	294,17
23584,91	4,68	3369,33	3358,10	274,79
22123,89	4,02	3369,00	3360,72	236,06
20833,33	3,77	3368,81	3361,50	221,78
19531,25	3,53	3368,37	3361,98	207,43
18382,35	3,32	3367,87	3362,22	195,06
17241,38	3,11	3367,37	3362,40	182,82
16181,23	2,92	3366,42	3362,04	171,71
15197,57	2,75	3365,77	3361,90	161,47
14285,71	2,60	3364,88	3361,43	152,39
13404,83	2,45	3363,92	3360,86	143,58
12594,46	2,31	3362,95	3360,22	135,56
11820,33	2,19	3361,93	3359,48	128,43
11111,11	2,08	3360,97	3358,75	122,05
10438,41	1,98	3360,07	3358,06	116,29
9803,92	1,89	3359,00	3357,17	110,88
9208,10	1,82	3358,10	3356,41	106,44
8650,52	1,75	3357,19	3355,63	102,28
8116,88	1,69	3356,17	3354,72	98,84
7621,95	1,64	3355,35	3353,98	95,93
7163,32	1,60	3354,49	3353,19	93,48
6729,48	1,56	3353,81	3352,56	91,59
6313,13	1,54	3352,99	3351,78	90,21
5931,20	1,53	3352,37	3351,18	89,37
5567,93	1,52	3352,04	3350,86	88,78

Table 25 – EIS measurements for Phantom P2 using the Linear Arrangement 2 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
5230,13	1,52	3351,49	3350,31	88,85
4911,59	1,53	3351,21	3350,02	89,30
4616,81	1,54	3351,18	3349,97	90,29
4332,76	1,57	3350,92	3349,67	91,55
4071,66	1,59	3350,88	3349,59	93,17
3822,63	1,63	3350,77	3349,41	95,34
3589,38	1,68	3351,23	3349,79	97,97
3371,54	1,72	3351,81	3350,29	100,81
3166,56	1,78	3352,54	3350,92	104,22
2974,42	1,84	3352,90	3351,17	107,81
2793,30	1,92	3353,28	3351,40	112,18
2623,30	1,99	3354,53	3352,51	116,51
2313,74	2,17	3357,15	3354,75	126,90
2172,97	2,27	3359,23	3356,60	132,86
2040,82	2,38	3360,82	3357,93	139,41
1916,44	2,49	3362,87	3359,70	145,92
1799,86	2,61	3365,37	3361,88	153,20
1690,33	2,74	3367,89	3364,05	160,94
1587,30	2,87	3370,63	3366,40	168,87
1400,17	3,17	3376,24	3371,09	186,50
1314,75	3,32	3379,75	3374,07	195,92
1234,87	3,49	3383,94	3377,65	206,27
1159,82	3,67	3387,40	3380,46	216,73
1089,32	3,85	3391,91	3384,26	227,77
1022,91	4,05	3396,38	3387,91	239,65
960,61	4,26	3401,33	3391,94	252,54
902,20	4,46	3406,50	3396,18	264,90
847,31	4,68	3412,45	3401,09	278,23
747,38	5,15	3426,22	3412,39	307,51
701,85	5,40	3432,98	3417,72	323,33
659,20	5,66	3441,18	3424,39	339,50
619,12	5,94	3448,94	3430,43	356,83
581,40	6,22	3457,36	3437,01	374,57
512,82	6,82	3475,62	3451,02	412,78
477,27	7,19	3490,67	3463,23	436,82
456,52	7,42	3498,07	3468,76	451,87
420,00	7,89	3513,86	3480,58	482,45
403,85	8,11	3521,22	3485,96	497,05
375,00	8,55	3536,42	3497,08	526,03
350,00	8,98	3552,74	3509,17	554,68
328,13	9,40	3567,58	3519,63	582,92
308,82	9,76	3584,71	3532,82	607,71
291,67	10,16	3599,44	3543,04	634,73
276,32	10,54	3613,61	3552,67	660,83
256,10	11,11	3635,99	3567,83	700,70
244,19	11,50	3651,49	3578,24	727,75
228,26	12,04	3674,06	3593,29	766,17
214,29	12,56	3696,81	3608,35	803,86
198,11	13,20	3742,73	3643,77	854,94

Table 26 – EIS measurements for Phantom P2 using the Linear Arrangement 3 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
100000,00	20,39	2898,99	2717,38	1009,96
94339,62	19,37	2921,50	2756,14	968,95
87719,30	18,11	2941,41	2795,74	914,18
83333,34	17,35	2957,43	2822,83	882,07
78125,00	16,35	2969,09	2848,98	835,93
73529,42	15,44	2984,83	2877,15	794,50
68493,15	14,43	3000,43	2905,74	747,81
64102,56	13,58	3011,76	2927,58	707,09
60240,96	12,81	3016,95	2941,81	669,14
56818,18	12,20	3030,88	2962,48	640,30
53191,49	11,38	3033,88	2974,20	598,81
50000,00	10,85	3044,12	2989,72	572,92
47169,81	10,07	3021,76	2975,20	528,44
44247,79	9,50	3026,95	2985,44	499,58
41666,67	9,01	3031,30	2993,87	474,87
39062,50	8,50	3036,64	3003,32	448,64
36496,35	8,00	3041,95	3012,34	423,39
34482,76	7,61	3044,61	3017,77	403,40
32258,06	7,17	3048,87	3025,05	380,32
30303,03	6,82	3052,26	3030,64	362,64
28571,43	6,47	3055,75	3036,28	344,41
26737,97	6,12	3057,95	3040,51	326,05
25125,63	5,82	3060,46	3044,66	310,57
23584,91	5,54	3063,36	3049,07	295,52
22123,89	4,89	3065,15	3053,98	261,40
20833,33	4,67	3067,37	3057,17	249,84
19531,25	4,46	3069,64	3060,37	238,47
18382,35	4,27	3071,65	3063,13	228,60
17241,38	4,09	3073,80	3065,99	219,09
16181,23	3,91	3075,99	3068,83	209,78
15197,57	3,76	3077,92	3071,30	201,78
14285,71	3,61	3080,09	3073,96	194,13
13404,83	3,48	3081,88	3076,18	187,27
12594,46	3,36	3083,76	3078,44	180,93
11820,33	3,25	3085,80	3080,82	175,19
11111,11	3,16	3087,60	3082,91	170,14
10438,41	3,07	3089,57	3085,14	165,39
9803,92	2,99	3091,38	3087,18	161,10
9208,10	2,92	3093,52	3089,50	157,59
8650,52	2,85	3095,45	3091,61	154,17
8116,88	2,80	3097,47	3093,76	151,49
7621,95	2,75	3099,59	3096,01	148,88
7163,32	2,71	3101,82	3098,35	146,88
6729,48	2,68	3103,95	3100,55	145,30
6313,13	2,66	3106,14	3102,79	144,11
5931,20	2,64	3108,39	3105,10	143,02
5567,93	2,63	3110,77	3107,50	142,53

Table 27 – EIS measurements for Phantom P2 using the Linear Arrangement 3 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
5230,13	2,62	3113,38	3110,12	142,50
4911,59	2,62	3115,83	3112,57	142,49
4616,81	2,63	3118,32	3115,03	143,24
4332,76	2,64	3121,12	3117,80	143,95
4071,66	2,66	3123,84	3120,48	145,05
3822,63	2,69	3126,65	3123,21	146,75
3589,38	2,72	3129,51	3125,98	148,57
3371,54	2,76	3132,71	3129,08	150,77
3166,56	2,80	3135,66	3131,91	153,28
2974,42	2,85	3138,93	3135,05	155,93
2793,30	2,90	3142,25	3138,23	158,91
2623,30	2,96	3146,07	3141,87	162,47
2463,05	3,02	3149,76	3145,38	166,17
2313,74	3,09	3153,42	3148,82	170,23
2172,97	3,17	3157,60	3152,76	174,76
2040,82	3,25	3161,52	3156,42	179,46
1916,44	3,34	3166,27	3160,88	184,64
1799,86	3,43	3170,37	3164,68	189,90
1690,33	3,53	3175,14	3169,11	195,68
1587,30	3,64	3179,83	3173,41	201,94
1490,76	3,76	3184,78	3177,94	208,60
1400,17	3,87	3189,92	3182,64	215,42
1314,75	4,00	3195,31	3187,52	223,02
1234,87	4,14	3201,34	3193,00	230,95
1159,82	4,27	3207,29	3198,37	238,99
1089,32	4,43	3213,28	3203,70	247,95
1022,91	4,58	3219,45	3209,19	256,87
960,61	4,75	3226,32	3215,25	266,97
902,20	4,91	3233,15	3221,29	276,69
847,31	5,10	3240,54	3227,73	287,83
747,38	5,48	3256,78	3241,92	310,80
701,85	5,68	3264,69	3248,63	323,38
659,20	5,90	3273,62	3256,29	336,44
619,12	6,12	3282,51	3263,78	350,23
581,40	6,36	3291,87	3271,62	364,56
512,82	6,86	3311,42	3287,72	395,43
477,27	7,16	3326,99	3301,02	414,92
456,52	7,36	3334,43	3306,96	427,12
420,00	7,75	3350,08	3319,49	451,70
403,85	7,93	3357,26	3325,16	463,14
350,00	8,67	3387,62	3348,94	510,50
328,13	9,01	3402,08	3360,11	532,71
308,82	9,31	3417,71	3372,66	553,09
291,67	9,65	3432,72	3384,17	575,27
276,32	9,96	3446,11	3394,14	596,24
244,19	10,75	3480,10	3418,99	649,35
228,26	11,23	3502,06	3435,04	681,87
214,29	11,66	3521,51	3448,85	711,70
198,11	12,25	3562,16	3481,01	756,04

Table 28 – EIS measurements for Phantom P2 using the Coplanar Arrangement 1 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
100000,00	20,45	3061,45	2868,48	1069,72
94339,62	19,45	3089,04	2912,75	1028,62
87719,30	18,22	3110,25	2954,39	972,25
83333,34	17,34	3122,84	2980,88	930,87
78125,00	16,34	3142,64	3015,72	884,11
73529,42	15,47	3156,69	3042,30	842,07
68493,15	14,53	3171,30	3069,83	795,80
64102,56	13,64	3179,22	3089,55	749,77
60240,96	12,93	3195,08	3114,05	715,01
56818,18	12,18	3202,73	3130,58	675,97
53191,49	11,42	3213,89	3150,23	636,53
50000,00	10,83	3222,00	3164,57	605,61
47169,81	10,07	3193,97	3144,80	558,32
44247,79	9,50	3201,21	3157,34	528,18
41666,67	8,98	3206,22	3166,90	500,61
39062,50	8,47	3211,11	3176,05	473,20
36496,35	7,97	3215,74	3184,66	445,99
34482,76	7,56	3219,89	3191,88	423,77
32258,06	7,13	3223,86	3198,90	400,42
30303,03	6,77	3227,33	3204,85	380,25
28571,43	6,43	3230,08	3209,73	361,98
26737,97	6,07	3232,70	3214,57	341,90
25125,63	5,76	3235,66	3219,31	324,90
23584,91	5,47	3237,68	3222,96	308,36
22123,89	4,83	3238,93	3227,44	272,59
20833,33	4,61	3241,22	3230,75	260,27
19531,25	4,39	3243,32	3233,83	248,00
18382,35	4,19	3245,35	3236,67	237,26
17241,38	4,01	3247,48	3239,54	226,92
16181,23	3,83	3249,21	3241,94	217,17
15197,57	3,67	3251,11	3244,43	208,32
14285,71	3,53	3252,90	3246,73	200,36
13404,83	3,40	3254,67	3248,95	192,81
12594,46	3,28	3256,34	3251,03	186,04
11820,33	3,16	3258,01	3253,05	179,85
11111,11	3,07	3259,81	3255,14	174,41
10438,41	2,98	3261,63	3257,24	169,36
9803,92	2,89	3263,27	3259,12	164,58
9208,10	2,82	3264,88	3260,92	160,72
8650,52	2,76	3266,69	3262,91	157,15
8116,88	2,70	3268,65	3265,02	154,04
7621,95	2,65	3270,37	3266,86	151,44
7163,32	2,62	3272,48	3269,07	149,41
6729,48	2,58	3274,38	3271,05	147,65
6313,13	2,56	3276,47	3273,19	146,42
5931,20	2,54	3278,31	3275,09	145,26
5567,93	2,53	3280,58	3277,38	144,93

Table 29 – EIS measurements for Phantom P2 using the Coplanar Arrangement 1 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
5230,13	2,53	3282,93	3279,74	144,67
4911,59	2,53	3285,25	3282,04	145,03
4616,81	2,54	3287,46	3284,24	145,55
4332,76	2,55	3290,12	3286,86	146,63
4071,66	2,57	3292,62	3289,29	147,90
3822,63	2,60	3295,22	3291,82	149,67
3589,38	2,63	3298,13	3294,65	151,50
3371,54	2,68	3301,07	3297,48	154,11
3166,56	2,72	3304,03	3300,31	156,72
2793,30	2,83	3310,76	3306,74	163,29
2623,30	2,89	3314,36	3310,16	166,83
2463,05	2,95	3317,79	3313,38	171,00
2313,74	3,03	3321,47	3316,82	175,62
2172,97	3,11	3325,72	3320,83	180,31
2040,82	3,19	3329,84	3324,67	185,50
1916,44	3,29	3334,11	3328,63	191,09
1799,86	3,39	3338,48	3332,65	197,18
1690,33	3,49	3343,17	3336,97	203,62
1587,30	3,60	3348,15	3341,54	210,37
1490,76	3,72	3353,12	3346,05	217,71
1400,17	3,85	3358,38	3350,81	225,33
1314,75	3,98	3364,06	3355,96	233,36
1234,87	4,11	3369,86	3361,18	241,80
1159,82	4,26	3375,93	3366,59	250,99
1089,32	4,41	3382,30	3372,27	260,25
1022,91	4,58	3388,56	3377,74	270,52
960,61	4,75	3396,11	3384,44	281,26
902,20	4,93	3403,08	3390,51	292,21
847,31	5,11	3410,66	3397,11	303,72
747,38	5,51	3428,26	3412,42	329,22
701,85	5,72	3436,56	3419,45	342,51
619,12	6,17	3455,48	3435,45	371,52
581,40	6,42	3465,15	3443,41	387,55
546,03	6,67	3475,59	3452,06	403,71
512,82	6,93	3486,39	3460,91	420,77
456,52	7,45	3511,31	3481,67	455,27
420,00	7,85	3527,95	3494,85	482,10
403,85	8,05	3535,63	3500,80	495,02
375,00	8,42	3552,54	3514,26	520,10
350,00	8,79	3569,10	3527,19	545,32
328,13	9,15	3585,11	3539,51	569,96
308,82	9,46	3601,76	3552,80	591,85
291,67	9,80	3616,99	3564,23	615,59
276,32	10,13	3632,09	3575,46	638,86
256,10	10,61	3654,14	3591,61	673,12
244,19	10,94	3669,22	3602,50	696,57
228,26	11,40	3692,12	3619,29	729,76
214,29	11,85	3714,54	3635,34	762,96
198,11	12,45	3757,91	3669,50	810,34

Table 30 – EIS measurements for Phantom P2 using the Coplanar Arrangement 2 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
100000,00	21,17	3196,71	2981,00	1154,40
94339,62	20,00	3219,27	3025,05	1101,25
87719,30	18,74	3245,41	3073,37	1042,65
83333,34	17,86	3262,99	3105,78	1000,60
78125,00	16,82	3281,42	3141,02	949,59
73529,42	15,88	3298,77	3172,85	902,72
68493,15	14,82	3311,34	3201,14	847,17
64102,56	13,89	3324,69	3227,49	798,05
60240,96	13,14	3339,63	3252,21	759,12
56818,18	12,43	3347,73	3269,25	720,61
53191,49	11,56	3356,26	3288,18	672,55
50000,00	10,94	3367,70	3306,45	639,35
47169,81	10,18	3340,64	3288,09	590,20
44247,79	9,57	3347,44	3300,88	556,35
41666,67	9,02	3352,10	3310,68	525,34
39062,50	8,48	3358,51	3321,78	495,35
36496,35	7,94	3361,51	3329,29	464,26
34482,76	7,53	3364,58	3335,54	441,11
32258,06	7,04	3369,79	3344,37	413,17
30303,03	6,62	3371,75	3349,24	388,99
28571,43	6,27	3373,72	3353,51	368,71
26737,97	5,91	3375,11	3357,19	347,33
25125,63	5,57	3378,08	3362,13	327,88
23584,91	5,23	3379,19	3365,13	307,91
22123,89	4,57	3380,46	3369,70	269,43
20833,33	4,32	3381,76	3372,13	255,01
19531,25	4,08	3383,39	3374,83	240,57
18382,35	3,86	3384,47	3376,79	227,80
17241,38	3,65	3385,57	3378,72	215,28
16181,23	3,45	3386,60	3380,47	203,67
15197,57	3,27	3387,23	3381,72	193,04
14285,71	3,10	3388,00	3383,04	183,10
13404,83	2,94	3388,54	3384,08	173,79
12594,46	2,79	3389,16	3385,14	165,03
11820,33	2,66	3389,67	3386,02	157,25
11111,11	2,53	3390,11	3386,79	149,92
10438,41	2,42	3390,54	3387,51	143,21
9803,92	2,32	3391,08	3388,31	137,02
9208,10	2,22	3391,31	3388,76	131,47
8650,52	2,14	3391,72	3389,36	126,38
8116,88	2,06	3392,06	3389,88	121,76
7621,95	1,99	3392,33	3390,29	117,60
7163,32	1,92	3392,66	3390,75	113,90
6729,48	1,87	3393,08	3391,28	110,77
6313,13	1,82	3393,27	3391,55	107,96
5931,20	1,78	3393,69	3392,05	105,46
5567,93	1,75	3394,13	3392,55	103,73

Table 31 – EIS measurements for Phantom P2 using the Coplanar Arrangement 2 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
5230,13	1,72	3394,56	3393,03	102,07
4911,59	1,71	3395,01	3393,50	101,12
4616,81	1,69	3395,26	3393,77	100,34
4332,76	1,68	3395,92	3394,46	99,83
4071,66	1,69	3396,46	3394,98	100,04
3822,63	1,70	3396,97	3395,48	100,60
3589,38	1,71	3397,80	3396,29	101,36
3371,54	1,73	3398,57	3397,02	102,50
3166,56	1,75	3399,58	3397,99	103,95
2974,42	1,79	3400,75	3399,10	106,05
2623,30	1,87	3402,72	3400,90	111,20
2463,05	1,92	3404,02	3402,10	114,14
2313,74	1,98	3405,38	3403,35	117,77
2172,97	2,05	3406,85	3404,68	121,69
2040,82	2,12	3408,51	3406,19	125,99
1916,44	2,20	3410,41	3407,90	130,76
1799,86	2,28	3412,20	3409,50	135,58
1690,33	2,37	3414,46	3411,54	141,23
1587,30	2,47	3416,54	3413,37	147,18
1490,76	2,58	3418,96	3415,50	153,66
1400,17	2,68	3421,85	3418,10	160,15
1234,87	2,94	3427,66	3423,17	175,53
1159,82	3,08	3431,47	3426,53	184,17
1089,32	3,21	3434,78	3429,38	192,59
1022,91	3,36	3438,45	3432,53	201,71
960,61	3,53	3443,09	3436,56	211,92
847,31	3,87	3452,03	3444,17	232,91
795,80	4,06	3457,00	3448,34	244,53
747,38	4,25	3463,58	3454,06	256,63
701,85	4,46	3468,77	3458,25	270,01
619,12	4,90	3482,45	3469,75	297,22
581,40	5,13	3489,02	3475,04	311,99
546,03	5,38	3496,44	3481,06	327,62
512,82	5,63	3504,23	3487,32	343,85
477,27	5,95	3516,62	3497,69	364,40
456,52	6,15	3523,28	3503,00	377,42
420,00	6,54	3535,99	3513,01	402,49
403,85	6,72	3541,64	3517,31	414,47
375,00	7,09	3554,24	3527,03	438,97
350,00	7,47	3567,60	3537,36	463,53
328,13	7,81	3580,47	3547,25	486,60
308,82	8,12	3594,19	3558,20	507,40
291,67	8,47	3606,08	3566,73	531,29
276,32	8,79	3618,47	3575,97	552,94
256,10	9,27	3637,46	3589,99	585,77
244,19	9,60	3649,24	3598,13	608,61
228,26	10,05	3668,49	3612,15	640,49
214,29	10,49	3686,67	3625,01	671,45
198,11	11,06	3723,51	3654,33	714,44

Table 32 – EIS measurements for Phantom P3 using the Linear Arrangement 1 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
100000,00	33,58	2959,80	2465,91	1636,98
94339,62	32,64	3008,26	2533,22	1622,48
87719,30	31,63	3064,12	2608,97	1606,91
83333,34	30,91	3099,30	2659,08	1592,17
78125,00	30,15	3142,19	2716,97	1578,43
73529,42	29,39	3186,99	2776,92	1563,85
68493,15	28,71	3229,96	2832,84	1551,67
64102,56	28,01	3273,06	2889,77	1536,94
60240,96	27,52	3316,75	2941,47	1532,50
56818,18	27,08	3355,84	2988,04	1527,51
53191,49	26,58	3396,09	3037,18	1519,53
50000,00	26,19	3435,73	3083,06	1516,23
47169,81	25,71	3442,67	3101,93	1493,31
44247,79	25,43	3483,62	3146,17	1495,74
41666,67	25,17	3523,22	3188,81	1498,20
39062,50	24,99	3566,99	3232,97	1507,10
36496,35	24,81	3612,56	3279,06	1516,02
34482,76	24,72	3652,17	3317,59	1527,07
32258,06	24,58	3698,91	3363,64	1538,78
30303,03	24,57	3745,31	3406,23	1557,23
28571,43	24,54	3791,06	3448,68	1574,40
26737,97	24,56	3841,55	3493,95	1596,82
25125,63	24,52	3891,77	3540,84	1615,03
23584,91	24,64	3943,48	3584,37	1644,18
22123,89	24,38	4000,82	3644,20	1651,17
20833,33	24,47	4054,54	3690,41	1679,33
19531,25	24,62	4114,00	3740,07	1713,73
18382,35	24,79	4171,94	3787,52	1749,22
17241,38	24,96	4235,24	3839,80	1786,95
16181,23	25,21	4300,86	3891,32	1831,69
15197,57	25,37	4367,38	3946,22	1871,19
14285,71	25,51	4434,91	4002,49	1910,10
13404,83	25,87	4509,17	4057,15	1967,78
12594,46	26,09	4583,12	4116,02	2015,77
11820,33	26,33	4660,94	4177,28	2067,54
11111,11	26,61	4739,92	4237,75	2123,28
10438,41	26,91	4822,98	4300,61	2183,08
9803,92	27,17	4908,58	4367,10	2241,10
9208,10	27,47	4997,66	4434,06	2305,58
8650,52	27,75	5089,79	4504,31	2370,04
8116,88	28,08	5187,73	4577,18	2441,71
7621,95	28,35	5286,83	4652,54	2510,87
7163,32	28,65	5389,01	4728,99	2584,20
6729,48	28,94	5494,80	4808,42	2659,30
6313,13	29,22	5606,04	4892,57	2736,88
5931,20	29,50	5718,58	4977,43	2815,56

Table 33 – EIS measurements for Phantom P3 using the Linear Arrangement 1 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
5567,93	29,81	5837,23	5065,03	2901,50
5230,13	30,08	5958,14	5155,70	2986,33
4911,59	30,36	6083,65	5249,40	3074,84
4616,81	30,64	6211,34	5344,36	3165,20
4332,76	30,91	6346,92	5445,64	3260,12
3822,63	31,38	6624,50	5655,29	3449,89
3589,38	31,68	6774,04	5764,60	3557,66
3371,54	31,93	6924,44	5876,59	3662,46
3166,56	32,15	7078,42	5992,85	3766,92
2974,42	32,56	7259,69	6118,91	3906,66
2793,30	32,82	7430,40	6244,22	4027,47
2623,30	33,04	7601,95	6372,87	4144,43
2463,05	33,26	7779,17	6505,20	4265,91
2313,74	33,47	7960,74	6640,34	4390,82
2172,97	33,68	8148,56	6780,65	4519,05
2040,82	33,86	8340,09	6925,51	4646,97
1916,44	34,05	8538,49	7074,22	4781,35
1799,86	34,24	8742,85	7227,39	4919,57
1690,33	34,37	8946,95	7384,68	5051,18
1587,30	34,51	9160,71	7548,55	5190,18
1490,76	34,71	9385,81	7715,42	5344,70
1400,17	34,92	9618,22	7886,73	5505,42
1314,75	34,98	9842,58	8064,27	5643,04
1234,87	35,12	10085,43	8249,36	5802,06
1159,82	35,26	10333,37	8437,12	5966,03
1089,32	35,46	10592,31	8627,94	6144,56
1022,91	35,53	10846,11	8826,65	6303,04
960,61	35,57	11107,96	9035,17	6461,62
847,31	36,00	11690,87	9458,29	6871,48
795,80	36,20	11996,11	9680,74	7084,49
747,38	36,38	12314,45	9914,81	7303,57
701,85	36,55	12633,02	10148,10	7523,92
619,12	36,74	13281,41	10642,94	7945,03
581,40	36,82	13618,89	10902,14	8161,96
512,82	37,14	14350,16	11439,99	8663,36
477,27	37,24	14799,17	11782,27	8955,09
456,52	37,25	15059,05	11986,96	9115,25
420,00	37,39	15583,60	12380,75	9463,92
403,85	37,44	15839,72	12576,29	9629,84
375,00	37,55	16333,45	12949,80	9954,11
350,00	37,58	16794,29	13309,31	10242,58
328,13	37,80	17281,09	13654,98	10591,40
308,82	38,04	18026,25	14196,76	11108,46
291,67	38,30	18537,71	14548,46	11488,65
276,32	38,16	18909,40	14867,89	11683,81
256,10	38,28	19520,97	15323,51	12093,72
244,19	38,17	19865,99	15618,73	12276,51
228,26	38,09	20404,55	16060,18	12586,36
214,29	37,99	20890,64	16464,70	12858,17
198,11	38,06	21681,81	17071,98	13365,93

Table 34 – EIS measurements for Phantom P3 using the Linear Arrangement 2 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
100000,00	26,10	2807,20	2520,91	1235,07
94339,62	24,81	2837,00	2575,07	1190,63
87719,30	23,31	2870,50	2636,18	1135,93
83333,34	22,32	2893,65	2676,88	1098,87
78125,00	21,07	2919,48	2724,27	1049,63
73529,42	19,98	2941,75	2764,64	1005,33
68493,15	18,82	2964,21	2805,76	956,16
64102,56	17,72	2985,92	2844,18	909,04
60240,96	16,81	3000,93	2872,74	867,69
56818,18	15,92	3016,07	2900,37	827,38
53191,49	15,03	3030,03	2926,34	785,88
50000,00	14,20	3044,07	2951,04	746,82
47169,81	13,35	3031,34	2949,45	699,83
44247,79	12,61	3042,02	2968,70	663,87
41666,67	11,95	3052,43	2986,23	632,29
39062,50	11,30	3061,24	3001,89	599,87
36496,35	10,64	3070,01	3017,25	566,71
34482,76	10,11	3078,06	3030,28	540,22
32258,06	9,56	3084,46	3041,66	512,03
30303,03	9,05	3091,13	3052,69	486,00
28571,43	8,58	3096,51	3061,82	462,15
26737,97	8,11	3102,52	3071,50	437,61
25125,63	7,69	3107,54	3079,57	415,97
23584,91	7,30	3111,67	3086,46	395,30
22123,89	6,54	3116,21	3095,91	355,17
20833,33	6,23	3120,39	3101,95	338,70
19531,25	5,91	3124,56	3107,95	321,74
18382,35	5,63	3128,01	3112,93	306,76
17241,38	5,35	3131,71	3118,04	292,24
16181,23	5,09	3135,09	3122,72	278,20
15197,57	4,86	3138,23	3126,97	265,61
14285,71	4,63	3141,21	3130,95	253,74
13404,83	4,42	3144,24	3134,89	242,33
12594,46	4,22	3147,03	3138,48	231,85
11820,33	4,04	3149,64	3141,82	221,83
11111,11	3,87	3152,24	3145,07	212,61
10438,41	3,71	3154,77	3148,15	204,29
9803,92	3,56	3157,16	3151,06	196,16
9208,10	3,43	3159,60	3153,95	188,79
8650,52	3,30	3161,98	3156,74	181,95
8116,88	3,18	3164,37	3159,50	175,49
7621,95	3,07	3166,60	3162,06	169,53
7163,32	2,97	3168,81	3164,55	164,26
6729,48	2,88	3170,99	3166,99	159,21
6313,13	2,79	3173,28	3169,51	154,55
5931,20	2,72	3175,26	3171,69	150,41
5567,93	2,65	3177,53	3174,14	146,74

Table 35 – EIS measurements for Phantom P3 using the Linear Arrangement 2 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
5230,13	2,58	3179,89	3176,66	143,26
4911,59	2,53	3181,84	3178,74	140,31
4616,81	2,48	3184,20	3181,22	137,71
4332,76	2,44	3186,42	3183,54	135,56
4071,66	2,40	3188,60	3185,81	133,42
3822,63	2,37	3190,84	3188,10	132,04
3589,38	2,34	3193,33	3190,67	130,48
3371,54	2,33	3195,45	3192,82	129,66
3166,56	2,31	3197,66	3195,07	128,76
2974,42	2,30	3199,80	3197,23	128,26
2793,30	2,29	3202,17	3199,61	127,93
2623,30	2,30	3204,55	3201,98	128,33
2463,05	2,30	3206,90	3204,31	128,92
2313,74	2,31	3209,53	3206,93	129,12
2172,97	2,33	3212,26	3209,61	130,55
2040,82	2,35	3215,18	3212,48	131,72
1916,44	2,37	3217,71	3214,94	133,28
1690,33	2,44	3223,56	3220,63	137,33
1587,30	2,48	3226,53	3223,51	139,54
1490,76	2,53	3229,73	3226,59	142,49
1400,17	2,58	3233,17	3229,89	145,54
1314,75	2,63	3236,21	3232,81	148,48
1234,87	2,69	3239,74	3236,17	152,10
1159,82	2,76	3243,37	3239,62	156,00
1089,32	2,83	3246,81	3242,85	160,38
1022,91	2,91	3250,82	3246,63	164,91
960,61	2,98	3254,50	3250,09	169,36
902,20	3,08	3258,38	3253,69	174,79
847,31	3,16	3263,06	3258,08	180,15
795,80	3,27	3266,99	3261,68	186,08
747,38	3,37	3272,23	3266,58	192,33
701,85	3,49	3276,74	3270,65	199,75
659,20	3,60	3281,70	3275,23	206,11
581,40	3,86	3291,79	3284,33	221,46
546,03	3,99	3297,20	3289,20	229,52
477,27	4,32	3311,20	3301,81	249,19
456,52	4,43	3315,49	3305,58	256,11
420,00	4,66	3323,68	3312,71	269,80
403,85	4,77	3328,49	3316,98	276,65
375,00	4,99	3335,79	3323,13	290,33
328,13	5,41	3352,29	3337,37	315,92
308,82	5,60	3359,98	3343,95	327,90
291,67	5,78	3367,82	3350,68	339,33
276,32	5,99	3375,79	3357,38	352,06
256,10	6,26	3387,59	3367,42	369,15
244,19	6,46	3391,95	3370,45	381,36
228,26	6,75	3409,41	3385,75	401,01
214,29	7,02	3417,30	3391,72	417,37
198,11	7,34	3438,15	3409,96	439,39

Table 36 – EIS measurements for Phantom P3 using the Linear Arrangement 3 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
100000,00	21,41	2367,86	2204,48	864,32
94339,62	20,44	2388,69	2238,26	834,29
87719,30	19,33	2417,09	2280,85	800,04
83333,34	18,59	2431,00	2304,19	774,89
78125,00	17,57	2451,85	2337,44	740,24
73529,42	16,72	2466,04	2361,78	709,50
68493,15	15,74	2479,50	2386,50	672,69
64102,56	14,77	2476,92	2395,05	631,55
60240,96	13,92	2485,12	2412,17	597,69
56818,18	13,36	2495,95	2428,38	576,83
53191,49	12,60	2504,63	2444,29	546,46
50000,00	11,95	2514,72	2460,19	520,87
47169,81	11,28	2505,08	2456,69	490,00
44247,79	10,70	2513,11	2469,43	466,52
41666,67	10,18	2520,23	2480,54	445,52
39062,50	9,66	2526,58	2490,73	424,10
36496,35	9,17	2533,85	2501,50	403,63
34482,76	8,76	2539,41	2509,77	386,85
32258,06	8,31	2546,03	2519,33	367,76
30303,03	7,90	2550,53	2526,29	350,74
28571,43	7,57	2555,14	2532,88	336,56
26737,97	7,18	2560,64	2540,56	320,00
25125,63	6,86	2565,27	2546,92	306,27
23584,91	6,56	2569,39	2552,56	293,63
22123,89	5,89	2575,48	2561,90	264,12
20833,33	5,64	2579,49	2567,01	253,42
19531,25	5,40	2583,65	2572,20	242,94
18382,35	5,18	2587,41	2576,85	233,54
17241,38	4,96	2591,00	2581,29	224,12
16181,23	4,76	2594,74	2585,79	215,44
15197,57	4,58	2598,29	2590,01	207,27
14285,71	4,40	2601,78	2594,11	199,68
13404,83	4,24	2604,89	2597,77	192,48
12594,46	4,09	2608,09	2601,46	185,81
11820,33	3,94	2611,21	2605,03	179,53
11111,11	3,81	2614,29	2608,51	173,80
10438,41	3,69	2617,15	2611,72	168,45
9803,92	3,57	2620,27	2615,17	163,34
9208,10	3,47	2623,12	2618,32	158,68
8650,52	3,37	2626,11	2621,57	154,35
8116,88	3,28	2628,86	2624,56	150,27
7621,95	3,19	2631,69	2627,60	146,67
7163,32	3,12	2634,39	2630,49	143,36
6729,48	3,05	2637,18	2633,44	140,35
6313,13	2,99	2639,93	2636,34	137,80
5931,20	2,93	2642,49	2639,03	135,18
5567,93	2,88	2645,29	2641,93	133,13

Table 37 – EIS measurements for Phantom P3 using the Linear Arrangement 3 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
5230,13	2,84	2647,91	2644,66	131,21
4911,59	2,80	2650,64	2647,47	129,60
4616,81	2,77	2653,27	2650,16	128,33
4332,76	2,74	2655,88	2652,84	127,17
4071,66	2,72	2658,70	2655,70	126,36
3822,63	2,71	2661,34	2658,38	125,62
3589,38	2,70	2664,14	2661,19	125,50
3371,54	2,69	2666,78	2663,84	125,37
3166,56	2,69	2669,67	2666,72	125,39
2974,42	2,70	2672,66	2669,70	125,68
2793,30	2,70	2675,30	2672,32	126,18
2623,30	2,72	2678,35	2675,34	127,03
2463,05	2,74	2681,42	2678,35	128,32
2313,74	2,77	2684,43	2681,30	129,62
2172,97	2,80	2687,53	2684,32	131,19
2040,82	2,83	2690,96	2687,68	132,92
1916,44	2,86	2694,28	2690,92	134,66
1799,86	2,91	2697,22	2693,73	137,13
1690,33	2,96	2700,52	2696,92	139,50
1490,76	3,08	2707,83	2703,91	145,61
1314,75	3,22	2714,89	2710,61	152,39
1234,87	3,29	2718,75	2714,26	156,27
1159,82	3,37	2722,80	2718,09	160,22
1089,32	3,46	2726,92	2721,95	164,61
1022,91	3,56	2730,85	2725,59	169,46
902,20	3,76	2739,96	2734,05	179,83
847,31	3,88	2744,70	2738,42	185,53
795,80	3,99	2748,78	2742,11	191,35
747,38	4,12	2754,46	2747,36	197,67
659,20	4,39	2765,01	2756,89	211,84
619,12	4,54	2770,49	2761,80	219,20
581,40	4,70	2776,04	2766,71	227,44
546,03	4,86	2782,23	2772,23	235,64
512,82	5,11	2791,39	2780,29	248,69
477,27	5,18	2796,70	2785,27	252,56
456,52	5,32	2794,64	2782,60	259,08
420,00	5,61	2808,53	2795,09	274,47
403,85	5,71	2816,76	2802,80	280,13
375,00	6,02	2829,97	2814,39	296,58
350,00	6,25	2833,78	2816,94	308,47
328,13	6,47	2839,04	2820,95	320,03
308,82	6,67	2848,90	2829,64	330,69
291,67	6,91	2862,65	2841,86	344,40
276,32	7,20	2867,87	2845,23	359,67
256,10	7,55	2872,98	2848,10	377,26
244,19	7,85	2888,71	2861,65	394,46
228,26	8,22	2900,08	2870,29	414,59
214,29	8,42	2906,15	2874,80	425,68
198,11	9,02	2933,81	2897,53	459,95

Table 38 – EIS measurements for Phantom P3 using the Coplanar Arrangement 1 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
100000,00	21,07	3488,00	3254,80	1253,98
94339,62	20,09	3520,69	3306,52	1209,20
87719,30	18,81	3548,25	3358,74	1144,09
83333,34	17,97	3561,35	3387,56	1098,92
78125,00	17,04	3581,45	3424,23	1049,49
73529,42	16,07	3599,49	3458,88	996,24
68493,15	15,09	3613,35	3488,75	940,71
64102,56	14,20	3635,17	3524,06	891,90
60240,96	13,39	3648,45	3549,27	844,93
56818,18	12,79	3651,90	3561,32	808,31
53191,49	11,96	3667,11	3587,55	759,74
50000,00	11,39	3679,80	3607,39	726,42
47169,81	10,55	3648,39	3586,77	667,70
44247,79	9,99	3654,57	3599,17	633,95
41666,67	9,45	3661,95	3612,24	601,39
39062,50	8,95	3668,25	3623,59	570,67
36496,35	8,44	3674,72	3634,96	539,15
34482,76	8,05	3679,13	3642,90	515,11
32258,06	7,59	3684,18	3651,87	486,88
30303,03	7,23	3688,22	3658,93	463,91
28571,43	6,88	3691,57	3664,98	442,26
26737,97	6,52	3696,24	3672,34	419,67
25125,63	6,21	3698,73	3677,04	400,02
23584,91	5,92	3701,71	3681,93	382,11
22123,89	5,29	3705,91	3690,14	341,54
20833,33	5,07	3708,97	3694,43	328,01
19531,25	4,85	3712,06	3698,76	313,94
18382,35	4,67	3714,72	3702,42	302,15
17241,38	4,48	3717,60	3706,23	290,49
16181,23	4,32	3720,54	3709,97	280,24
15197,57	4,17	3723,18	3713,32	270,81
14285,71	4,04	3725,80	3716,56	262,22
13404,83	3,91	3728,55	3719,85	254,54
12594,46	3,80	3731,12	3722,90	247,49
11820,33	3,71	3733,88	3726,07	241,30
11111,11	3,62	3736,61	3729,15	235,95
10438,41	3,55	3739,11	3731,95	231,34
9803,92	3,48	3741,93	3735,03	227,17
9208,10	3,43	3744,86	3738,15	224,10
8650,52	3,38	3747,64	3741,10	221,22
8116,88	3,35	3750,56	3744,15	219,14
7621,95	3,33	3753,63	3747,31	217,81
7163,32	3,31	3756,72	3750,45	216,93
6729,48	3,30	3759,99	3753,75	216,53
6313,13	3,31	3763,36	3757,10	217,00
5931,20	3,32	3766,78	3760,48	217,84
5567,93	3,33	3770,29	3763,91	219,22

Table 39 – EIS measurements for Phantom P3 using the Coplanar Arrangement 1 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
5230,13	3,36	3773,94	3767,44	221,44
4911,59	3,40	3778,00	3771,35	223,94
4616,81	3,45	3781,81	3774,97	227,31
4332,76	3,50	3786,20	3779,13	231,34
4071,66	3,56	3790,57	3783,25	235,51
3589,38	3,71	3800,13	3792,16	245,93
3371,54	3,80	3804,93	3796,57	252,12
3166,56	4,15	3795,41	3785,45	274,84
2974,42	4,46	3777,53	3766,11	293,57
2793,30	4,53	3783,26	3771,45	298,77
2623,30	4,61	3789,58	3777,34	304,35
2463,05	4,70	3796,87	3784,12	310,96
2313,74	4,79	3803,72	3790,42	317,78
2172,97	4,91	3812,06	3798,06	326,46
1916,44	5,10	3826,48	3811,36	339,87
1799,86	5,20	3834,36	3818,58	347,43
1587,30	5,44	3850,80	3833,44	365,20
1490,76	5,58	3859,66	3841,38	375,18
1400,17	5,73	3868,71	3849,40	386,13
1314,75	5,88	3878,57	3858,15	397,41
1234,87	6,05	3888,89	3867,26	409,62
1159,82	6,22	3899,42	3876,47	422,47
1089,32	6,41	3910,05	3885,64	436,28
1022,91	6,60	3921,11	3895,10	450,89
902,20	7,03	3945,52	3915,83	483,15
847,31	7,26	3958,18	3926,48	500,01
795,80	7,50	3971,58	3937,60	518,39
747,38	7,74	3987,09	3950,74	537,12
701,85	8,00	4001,36	3962,37	557,23
659,20	8,28	4016,41	3974,56	578,27
619,12	8,55	4032,34	3987,53	599,48
581,40	8,85	4048,13	3999,97	622,58
546,03	9,14	4065,14	4013,47	646,06
512,82	9,46	4082,91	4027,34	671,34
477,27	9,85	4108,63	4048,10	702,67
456,52	10,09	4122,98	4059,25	722,15
420,00	10,59	4150,46	4079,79	762,63
403,85	10,82	4164,93	4090,87	781,92
375,00	11,28	4191,48	4110,55	819,65
350,00	11,70	4217,59	4129,89	855,61
328,13	12,14	4243,57	4148,65	892,50
308,82	12,53	4272,72	4170,88	927,30
291,67	12,93	4296,92	4188,05	961,14
276,32	13,28	4320,20	4204,68	992,36
256,10	13,81	4355,15	4229,32	1039,33
244,19	14,13	4377,01	4244,53	1068,71
228,26	14,66	4412,19	4268,62	1116,37
214,29	15,15	4445,23	4290,77	1161,64
198,11	15,81	4509,96	4339,30	1228,90

Table 40 – EIS measurements for Phantom P3 using the Coplanar Arrangement 2 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
100000,00	25,47	3214,95	2902,41	1382,72
94339,62	24,16	3251,68	2966,82	1330,95
87719,30	22,71	3287,58	3032,70	1269,21
83333,34	21,80	3313,43	3076,38	1230,75
78125,00	20,65	3343,30	3128,58	1178,85
73529,42	19,53	3364,88	3171,35	1124,70
68493,15	18,37	3392,63	3219,78	1069,07
64102,56	17,43	3413,95	3257,13	1022,82
60240,96	16,52	3433,54	3291,80	976,35
56818,18	15,65	3449,95	3322,08	930,54
53191,49	14,81	3468,64	3353,42	886,60
50000,00	14,06	3483,46	3379,04	846,48
47169,81	13,20	3464,18	3372,70	790,86
44247,79	12,51	3475,58	3393,05	752,89
41666,67	11,91	3484,86	3409,82	719,32
39062,50	11,27	3495,54	3428,08	683,43
36496,35	10,67	3505,30	3444,68	649,05
34482,76	10,18	3512,81	3457,56	620,58
32258,06	9,65	3522,47	3472,66	590,27
30303,03	9,19	3528,92	3483,60	563,77
28571,43	8,76	3535,42	3494,19	538,38
26737,97	8,34	3542,54	3505,09	513,77
25125,63	7,96	3548,13	3513,95	491,29
23584,91	7,61	3553,90	3522,59	470,69
22123,89	6,92	3559,15	3533,21	428,89
20833,33	6,65	3564,09	3540,15	412,46
19531,25	6,38	3569,18	3547,10	396,40
18382,35	6,14	3573,99	3553,46	382,55
17241,38	5,91	3578,52	3559,47	368,72
16181,23	5,71	3583,13	3565,36	356,44
15197,57	5,52	3587,58	3570,92	345,34
14285,71	5,35	3591,82	3576,15	335,14
13404,83	5,20	3596,06	3581,26	325,91
12594,46	5,06	3600,23	3586,19	317,62
11820,33	4,94	3604,69	3591,32	310,16
11111,11	4,83	3608,63	3595,82	303,75
10438,41	4,74	3612,87	3600,54	298,24
9803,92	4,65	3617,18	3605,25	293,52
9208,10	4,58	3621,37	3609,78	289,47
8650,52	4,53	3625,66	3614,34	286,22
8116,88	4,48	3630,27	3619,16	283,75
7621,95	4,45	3634,86	3623,91	281,97
7163,32	4,43	3639,55	3628,68	281,04
6729,48	4,42	3644,26	3633,42	280,87
6313,13	4,42	3649,23	3638,37	281,34
5931,20	4,43	3654,34	3643,42	282,33
5567,93	4,45	3659,78	3648,74	284,05

Table 41 – EIS measurements for Phantom P3 using the Coplanar Arrangement 2 electrode configuration.

Freq (Hz)	Phase angle (°)	Impedance magnitude (Ω)	Real part (Ω)	Imaginary part (Ω)
5230,13	4,49	3665,37	3654,14	286,74
4911,59	4,53	3671,03	3659,57	289,88
4616,81	4,58	3677,03	3665,29	293,63
4332,76	4,64	3683,31	3671,22	298,19
4071,66	4,71	3689,67	3677,19	303,29
3822,63	4,79	3696,63	3683,70	308,89
3589,38	4,88	3703,70	3690,25	315,32
3371,54	4,99	3711,19	3697,15	322,58
3166,56	5,10	3718,96	3704,25	330,48
2974,42	5,21	3727,12	3711,72	338,43
2793,30	5,34	3735,49	3719,28	347,69
2623,30	5,48	3744,36	3727,26	357,47
2463,05	5,62	3753,27	3735,21	367,83
2313,74	5,78	3762,71	3743,60	378,73
2172,97	5,94	3773,31	3753,05	390,54
2040,82	6,12	3783,56	3762,02	403,16
1916,44	6,29	3794,42	3771,58	415,74
1690,33	6,69	3818,37	3792,37	444,80
1587,30	6,90	3831,16	3803,43	460,09
1490,76	7,12	3844,46	3814,83	476,46
1400,17	7,34	3859,08	3827,42	493,33
1314,75	7,58	3872,92	3839,09	510,78
1234,87	7,83	3888,57	3852,32	529,74
1159,82	8,08	3903,90	3865,14	548,79
1089,32	8,35	3920,80	3879,22	569,52
1022,91	8,63	3938,01	3893,41	591,01
902,20	9,21	3975,55	3924,26	636,53
847,31	9,51	3993,92	3939,01	659,97
795,80	9,83	4013,33	3954,42	685,11
747,38	10,16	4037,11	3973,84	711,95
701,85	10,51	4058,83	3990,79	740,07
619,12	11,22	4106,61	4028,16	798,83
581,40	11,58	4131,96	4047,83	829,56
546,03	11,96	4157,07	4066,85	861,41
512,82	12,37	4183,58	4086,48	896,11
477,27	12,83	4220,97	4115,59	937,27
456,52	13,11	4242,04	4131,45	962,27
403,85	13,92	4306,28	4179,81	1035,99
375,00	14,48	4344,07	4206,16	1085,92
350,00	15,02	4383,99	4234,30	1135,81
328,13	15,52	4420,83	4259,66	1182,80
308,82	15,98	4459,94	4287,57	1227,95
291,67	16,48	4492,76	4308,12	1274,74
276,32	16,94	4528,15	4331,58	1319,70
256,10	17,63	4581,28	4366,06	1387,67
244,19	18,08	4615,66	4387,81	1432,28
228,26	18,73	4665,80	4418,74	1498,14
214,29	19,35	4714,35	4448,16	1561,70
198,11	20,16	4800,24	4506,08	1654,56

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Eu, Pedro Bertemes-Filho, professor do curso de Mestrado em Engenharia Elétrica do Programa de Pós-Graduação em Engenharia Elétrica da Udesc, declaro que esta é a versão final aprovada pela comissão julgadora da dissertação intitulada: **“BIOLOGICAL SKIN PHANTOM FOR ELECTRICAL IMPEDANCE SPECTROSCOPY: MODELING AND VALIDATION”** de autoria do acadêmico Elkin Roman Gomez.

Joinville, - Setembro de 2026.

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Prof. Pedro Bertemes-Filho, PhD