**Republic of Yemen**
**Ministry of Higher Education and Scientific Research**
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**Faculty of Engineering**

# Design of a Non-Invasive Kidney Function Monitoring Device Using Bioimpedance Technology

# تصميم جهاز لمراقبة وظائف الكلى باستخدام المعاوقة الحيوية

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# Abstract

Chronic Kidney Disease (CKD) is a progressive and widespread health condition that requires continuous monitoring to prevent complications and improve patient outcomes. Traditional diagnostic methods, such as blood tests and imaging, are often invasive, expensive, and inaccessible for routine use, particularly in remote or resource-limited areas. This project presents the design and development of a portable, non-invasive kidney function monitoring device based on bioimpedance technology. The system employs the AD5933 impedance converter chip, an Arduino Uno microcontroller, a 5-inch Nextion touchscreen display, and surface electrodes to measure tissue impedance correlated with extracellular fluid levels—an important indicator of kidney health. The device is powered by rechargeable lithium-ion batteries, making it suitable for mobile and at-home use.

The prototype was implemented in a compact, 3D-printed enclosure and tested on healthy volunteers. Results demonstrated a consistent impedance range aligned with physiological norms and showed sensitivity to hydration status. Although clinical validation is still required, the findings suggest that bioimpedance can be a promising tool for early detection and ongoing monitoring of kidney function.

This work lays the foundation for a low-cost, user-friendly device that can support both patients and healthcare providers in managing CKD. Future enhancements will focus on wireless communication, advanced signal processing, and improved measurement accuracy for broader clinical application.

# Dedication

To our fathers and mothers, who stood by us with unwavering support and gave us the strength to move forward, thank you for everything you've done, seen and unseen. To our siblings, who offered kind words and lifted our spirits when we needed it most—we are forever grateful

To our dedicated professors and academic advisors, whose guidance and encouragement pushed us to explore, think critically, and strive for excellence—thank you for shaping our minds and inspiring our ambitions

To our friends and colleagues, who walked this path beside us with shared laughter, late-night study sessions, and mutual support—this milestone would not have been as meaningful without your presence.

Finally, we dedicate this work to every dreamer who believes that persistence, knowledge, and faith can build a better tomorrow.

# Acknowledgment

Before and above all, we would like to record our endless thanks and praise to **Allah Almighty**, for granting us the strength, patience, and opportunity to complete this project.

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Our heartfelt thanks also go to all the faculty members of the Department of Biomedical Engineering for their dedication and efforts in providing us with the knowledge and tools we needed throughout our academic journey.Last but not least, we owe a great deal of gratitude and appreciation to our beloved families for their unwavering love, support, prayers, and encouragement. Without them, this achievement would not have been possible.

# Supervisor Certification

I certify that the preparation of this project entitled“BioImpedance-Based Kidney Function Monitoring Device”prepared by the students:Ali Saleh, Mohammed Ghazi, Abdullah Al-Dailami, Alaa Al-Kholani, and Omar Al-Zubairi,was carried out under my supervision at the Biomedical Engineering Departmentas partial fulfillment of the requirements for the Bachelor degree in Biomedical Engineering.

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**Project Title:** Bioimpedance-based Kidney Function Monitoring Device

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# List of Abbreviations

| | |
|---|---|
| BCA | Body Composition Assessment |
| CKD | Chronic Kidney Disease |
| DDS | Direct Digital Synthesis generator |
| DFT | Discrete Fourier Transform |
| ECF | Extra Cellular Fluid |
| ICF | Intra Cellular Fluid |
| LPF | Low Pass Filter |
| PGA | Programmable Gain Amplifier |
| TBC | Total Body Composition |
| TBW | Total Body Water |
| VCC | Voltage to Converter Current |
| BIS | Bioimpedance Spectroscopy |
| CT | Computed Tomography |
| CVC | Current to Voltage Converter |
| EBI | Electrical Bioimpedance |
| EC | Extra Cellular |
| MRI | Magnetic Resonance Imaging |

# *Chapter 1*

# Introduction

# Introduction

Chronic kidney disease (CKD) is a global health concern, with millions of people affected annually. The early detection and monitoring of kidney function are crucial for preventing the progression of CKD and improving patient outcomes. Current methods for assessing kidney health, such as blood tests and imaging, are often invasive, time-consuming, or inaccessible to many. Consequently, there is a growing demand for portable, non-invasive, and cost-effective solutions for kidney function monitoring.

Bioimpedance It is a non-invasive method used to measure the resistance of living tissues to electrical current. This technique relies on passing a low-intensity, high-frequency electrical current through the body, where the change in impedance (resistance) of the current is measured as it passes through different tissues such as fat, muscles, and body fluids.

The "Bioimpedance-based Kidney Function Monitoring Device" aims to bridge the gap between traditional diagnostic tools and the need for modern, user-friendly solutions. By integrating advanced bioimpedance technology with a compact and efficient design, this device seeks to provide reliable, real-time data that can
assist both clinicians and patients in monitoring kidney health. The development of such a device not only aligns with the latest advancements in medical technology but also addresses the pressing need for accessible healthcare innovations.
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## 1.1 Overview

This project aims to design and develop a portable device capable of monitoring kidney function using bioimpedance analysis. The system utilizes the AD5933 impedance converter chip, interfaced with an Arduino microcontroller, and a display module for real-time feedback. Surface electrodes are used to apply a small alternating current and measure the resulting voltage, allowing the system to estimate changes in tissue impedance, which may correlate with kidney performance.

By analyzing the bioimpedance spectrum, the device seeks to identify early signs of kidney dysfunction in a non-invasive and continuous manner. This prototype is particularly suited for  could potentially be adapted for at-home monitoring of patients at risk of renal complications.

## 1.2 Problem Statement

### 1.2.1 Delayed Diagnosis of Kidney Disease

Traditional methods for assessing kidney function, such as blood tests (e.g., serum creatinine levels) and imaging, often detect issues at advanced stages of kidney damage. This device enables earlier detection of kidney dysfunction through real-time monitoring, improving the chances of early intervention and treatment

### 1.2.2 Invasive and Time-Consuming Tests

Current diagnostic techniques for kidney
health often require invasive procedures, such as blood draws, which can be uncomfortable for patients. This device offers a non-invasive alternative, making it more patient-friendly.

### 1.2.3 Invasive and Time-Consuming Tests

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health often require invasive procedures, such as blood draws, which can be uncomfortable for patients. This device offers a non-invasive alternative, making it more patient-friendly.

### 1.2.4 Accessibility in Low-Resource Settings

Many rural and underdeveloped areas lack access to advanced diagnostic equipment and medical professionals. This portable and cost-effective device can be used in such settings to enhance kidney health monitoring, reducing healthcare disparities.

### 1.2.5 High Cost of Current Diagnostic Tools

Advanced kidney diagnostic tools, specialized operation. This project offers a more affordable solution for patients and healthcare providers. such as imaging machines
or biochemical analyzers, are expensive and require

### 1.2.6 Frequent Hospital Visits

Patients with kidney conditions often need to visit healthcare facilities repeatedly for tests, causing inconvenience and additional expenses. A portable bioimpedance device allows patients to monitor their kidney health at home, reducing the need for frequent hospital trips.

### 1.2.7 Overburdened Healthcare Systems

Healthcare facilities, particularly in developing countries, are often overwhelmed by the growing number of patients with chronic diseases. A bioimpedance-based device can ease this burden by enabling remote and decentralized monitoring.

### 1.2.8 Improved Decision-Making for Clinicians

The device provides clinicians with real-time and continuous data on kidney function, improving diagnostic accuracy and enabling better treatment decisions.

By addressing these challenges, the project has the potential to significantly improve patient outcomes, reduce healthcare costs, and contribute to advancements in medical technology.

## 1.3 Significance of the Project

1. Non-invasive: Bioimpedance technology provides a comfortable and safe method for measuring kidney function without the need for blood samples.
2. Continuous monitoring: Enables regular tracking of kidney function, helping doctors detect problems early and make accurate treatment decisions.
3. Cost reduction: The proposed device can serve as a cost-effective alternative to frequent laboratory tests or other traditional methods.
4. Improved quality of life: Assists patients,
5. Reduced pressure on healthcare facilities: By providing an instant solution for monitoring kidney function outside medical facilities.

## 1.4 Project Objectives

The primary objective of this project is to design and develop a bioimpedance-baseddevice capable of monitoring kidney function non-invasively.

### 1.4.1 Main Objective

Design a device based on Bioimpedance technology to measure and monitor kidney function in a non-invasive and accurate manner.

### 1.4.1 Sub-objectives

1. Develop a mechanism for measuring electrical resistance: Determine the relationship between tissue impedance and kidney function.
2. Design a compact and user-friendly device: Suitable for various age groups and effective in different environments.
3. Ensure accurate data analysis: Utilize software or algorithms to analyze measurements and extract precise information about kidney health.
4. Test the device: Validate its accuracy through comparative tests with traditional methods.
5. Reduce manufacturing costs: Develop the device at a low production cost to make it accessible to everyone.

## 1.5 project Scope and Limitations

### 1.5.1 Project Scope

This project aims to design and develop a prototype of a home-based, continuous kidney function monitoring device using bioimpedance analysis. The system integrates the AD5933 impedance converter chip with an Arduino microcontroller and displays real-time data on a screen. Surface electrodes are used to perform non-invasive measurements by injecting a low AC current and analyzing the resulting voltage to estimate body impedance. The device is specifically designed for use by chronic kidney disease (CKD) patients at home to support continuous health monitoring and early

detection of changes that may indicate deterioration in kidney function. The project focuses on usability, affordability, and non-invasive operation.

### 1.5.2 Limitations

The current version is a prototype and has not yet undergone clinical validation or regulatory approval for medical use.
Measurements may be affected by external factors such as movement, temperature, skin conditions, and electrode placement.
The system provides raw impedance data without full medical interpretation or integration with patient health records.
The AD5933 has a limited frequency range, which may limit sensitivity to subtle physiological changes.
While designed for home use, the prototype currently requires basic technical setup and supervision, which may not yet be suitable for all patients.

## 1.6 Project Methodology

A structured methodology was followed to design and implement a kidney function monitoring device using bioimpedance analysis. The development process was carried out through the following main stages:

### 1.6.1 Requirements Analysis

In this stage, the main objectives and functionalities of the device were defined. A thorough study of the electrical characteristics related to kidney function was conducted to determine how bioimpedance can be used to assess renal activity.

### 1.6.2 System Design

Appropriate electronic components were selected, including the AD5933 chip for signal generation and impedance measurement, an Arduino microcontroller for system control and data processing, a display screen for result visualization, and electrodes for signal application and acquisition.

### 1.6.3 Hardware Implementation

The components were assembled into a working circuit. Proper connections were ensured, and the device was physically constructed in accordance with the design specifications.

### 1.6.4 Software Development

The Arduino was programmed to control the AD5933, read impedance data, and process the measurements. The output was displayed on the screen in a user-friendly format.

### 1.6.5 Testing and Calibration

The system was tested using known resistive loads and biological tissue models to verify its accuracy. Calibration was performed to adjust the readings for more precise results.

### 1.6.6 Data Analysis

The obtained bioimpedance readings were analyzed to interpret their relevance to kidney function. The results were compared with standard physiological values to evaluate the performance of the system.

### 1.6.7 Documentation

All stages of the project, including circuit diagrams, code, data, and analysis, were thoroughly documented to ensure reproducibility and future development.

# Chapter 2

# Electrical Bioimpedance Measurements

# ٢. Electrical Bioimpedance Measurements

## 2.1 Introduction

Biological tissues are primarily composed of cells and fluids. For example, in the human body, the muscle tissue comprises long, tubular cells called myocytes, which contract to produce force. The cell is considered the basic structural and functional unit of all biological organisms, and it can exist as an independent unit of life. A cell is formed by a lipid bilayer membrane surrounding the cell, known as the cell membrane, which isolates the intracellular (IC) medium from the extracellular (EC) medium. The IC space contains the cell organelles, the cell nucleus, and other cell components. The EC space, which surrounds the cells, is divided into two major sub compartments the interstitial fluid and the blood plasma

Because of the presence of free ions, sodium (Na), potassium (K), chlorine (Cl), and protein ions that are contained in the IC and EC mediums, biological tissue can be considered to be an electrolyte that has electrical properties. Therefore, the IC and EC mediums are considered to be ionic conductors, and the lipid cell membrane is considered to behave as a capacitor because of its dielectric properties

Early studies of biological tissues and electricity contributed to the discovery and characterization of electrical properties in tissue (Fricke and Morse, 1925), [20]. Schwan 1957). H. Fricke presented the electrical equivalent model of blood cells, which is based on the assumption that the cells are suspended in the EC medium with electrical properties. This model, apart from being a precise model, has been fully acknowledged and extensively used with acceptable results. Fricke's representation model is shown in Figure 2.1, where R represents the resistance of the extracellular medium, R represents the resistance of the EC medium, and Cm represents the capacitance of the cell membrane. From Fricke's model in Figure 2.1, it is clear that the current will flow through different paths depending on the frequency. At low frequencies, the current will flow mainly through the EC medium. At higher frequencies, the modeled cell

membrane capacitor will act as a shunt that allows current to flow through the IC medium.

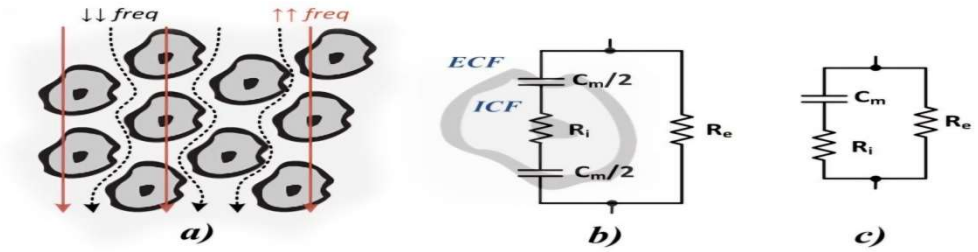


**Figure 2.1** Fricke's cell suspension electrical model and its circuit representation

Although Fricke's circuit model is a good model for a suspension of cells, it is less accurate for the more complex structures that constitute the human body. Other models, such as the Cole Model (Cole, 1940), [13] have been proposed and used, which shows a better agreement with the empirical data. Currently, the Cole Model is used in several EBI applications for tissue characterization such as Body Composition Analysis. The Cole Function and Model will be introduced in the following sections

The passive electrical properties of biological tissue exhibit a certain dependency on the frequency of the externally applied electrical field. The conductivity (6) and permittivity (c) are passive electrical properties of biological tissue that are expressed as a function of the frequency. The conductivity indicates how easily the free charges move through a medium and is related to the conductance. The permittivity is the measure of the resistance that is encountered when an electric field is formed in a dielectric medium. The permittivity is also expressed as the ability to permit the storage of electrical energy in a dielectric medium; this measurement is related to the conductance.

The impedance of the biological tissue varies with the frequency, and four specific dispersion windows have been identified: a, b, 8, and y (Schwan 1994, Schwan 1999) [41] , see Figure 2.2. The main contributions of these dispersions are explained as follows:

- α dispersion: This dispersion appears between 1 mHz and 1 kHz, and the mechanisms that contribute to this dispersion window are not fully clear (Schwan 1994).

- β dispersion: This dispersion is caused by the cellular structure of the tissue,which has poorly conducting membranes: the dispersion window is located between 1 kHz and 100 MHz
- δ dispersion: This dispersion appears between the B and Y dispersions, and itsinfluence is caused by amino acids and proteins in the frequency range of hundreds of MHz and a few GHz.y dispersion:
- γ This dispersion is caused by dipolar mechanisms in polar media such as salts, proteins, and water:this dispersion is found between 100 MHz and 100 GHz.

The A and Y dispersion windows are quite relevant for clinical applications because, in these dispersion windows, most changes in the electrical properties of human tissue occur, such as the accumulation of fluids or cell structure changes.

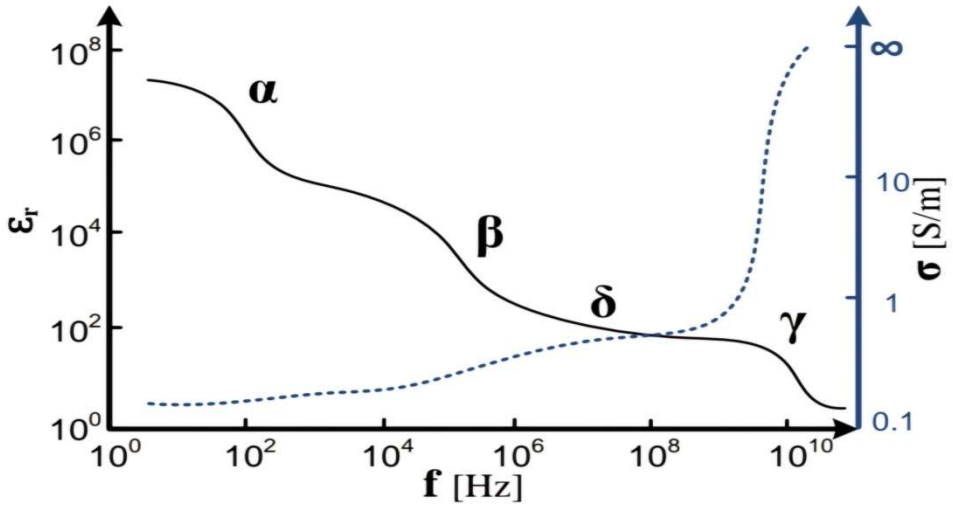


**Figure2.2** Frequency dependence of the dielectric constant ε (decreasing) and conductivity σ (increasing) of muscle tissue.Surce:(Schwan,1988), [40]

## 2.2 Electrical Bioimpedance measurements

The electrical impedance of a material is the opposition that the material offers to the flow of electrical charges through it. For materials that have a biological origin, the term Electrical Bioimpedance (EBI) is used. To characterise the electrical tissue properties, an external energy source is needed: in EBI measurements, the source of the energy is either an injected current or a voltage that is applied to the biological material. The resulting voltage or current is measured, and the impedance is obtained by applying Ohm's Law.

To characterise the electric properties over a frequency range, the most common method is to use one single frequency excitation signal at a time to extract the current and voltage signals and to
characterise the impedance over a frequency range. This procedure will be repeated a times when performing a frequency sweep (Pallàs-Areny and Webster 2001). Another proposed method uses an excitation signal that is formed by the summation of a signals at different frequencies, called multi-sine or chirp excitation. Compared to a single excitation signal (Bragos, Blanco-Enrich et al. 2001, Sanchez, Vandersteen et al. 2012), this approach has the advantage of accomplishing the impedance characterisation in a shorter amount of time compared with the frequency sweep methods.

### 2.2.1 Measurement electrode configuration

Depending on the EBI Application, there are two methods for obtaining the
impedance value: exciting with a controlled current or with a controlled voltage. Each method has its own advantages and disadvantages (Grimnes and Martinsen 2008), but the most common technique uses controlled current for excitation and measures the resulting voltage. The excitation current should be chosen to comply with the standard IEC-60601-1 for ensuring patient safety and electrical currents (International Electrotechnical Commission 2010), [24]. Throughout the remainder of this thesis, controlled current excitation is assumed.

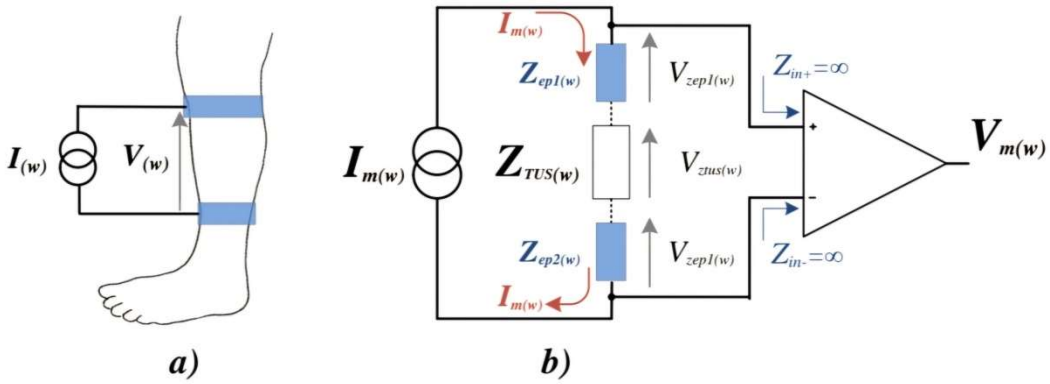


**Figure 2.3** Bipolar EBI measurement configuration (a) and equivalent measurement circuit(b)

$$Z_{(\omega)} = \frac{V_{(\omega)}}{I_{(\omega)}} \qquad (2.1)$$

A typical EBI measurement requires two, three or four contact points (Grimnes and Martinsen 2008) to measure the voltage and current values. The impedance is then obtained through Ohm's Law, see Equation (2.1), where w is the frequency in [rad/s]. (2.1)

In a 2-electrode or bipolar EBI measurement, two electrodes are used to inject the current, and the same electrodes are used to sense the resulting voltage. In Figure 2.3, an example of an EBI bipolar configuration measurement of the lower part of the leg is displayed (a), and the general bipolar configuration equivalent electrical circuit is electrode polarisation impedances, Ztus(ω) is the measured impedance, and the differential amplifier is used to obtain the voltage Vm(ω). The measurement impedance can be expressed in terms of its equivalent electric circuit equation displayed in Equation (2.1)

$$Z_{m\_2e(\omega)} = \frac{V_{m(\omega)}}{I_{m(\omega)}} = \frac{V_{zep1(\omega)} + V_{ztus(\omega)} + V_{zep2(\omega)}}{I_{m(\omega)}} \quad (2.2)$$

Because the electrical current flows through the sensing electrodes, the voltage drop across Zept(ω) and Zep2(ω) is included in the measured voltage together with the voltage over Ztus(ω). Assuming that the electrode polarisation impedances are approximately equal, the measured impedance equation Zm 20(ω) is shown in Equation (2.2).

### 2.2.2 The skin-electrode interface

Electrodes constitute an interface between the electronic currents in the measuring electronic instrumentation and the ionic currents that flow in the tissues. Typically, a non-invasive skin electrode is formed by a metal conductor, e.g. Silver, and an electrolytic gel, e.g., Silver-Chloride, which is applied to the skin surface, see Figure 2.5. The human skin is composed of three primary layers: the epidermis, the dermis and the hypodermis, also called the subcutaneous layer. The epidermis is the primary barrier between the outside world and the interior of the body, and it is primarily composed of dead cells that act as a dielectric membrane that is semi-permeable to ions. The dermis and subcutaneous layers are beneath the epidermis and containsome biological components that behave as ionic conductors, such as the hair follicles, sweat glands and blood vessels.

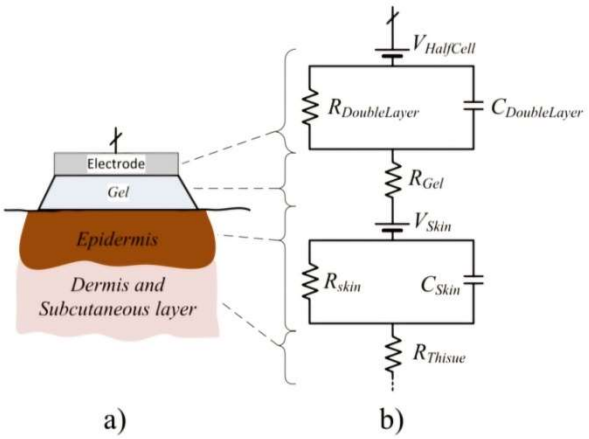


**Figure 2.5** The skin-electrode interconnection layers (a) and the electrical equivalent model (b).

The electrodes that are used for EBI measurements are typically non-invasive and they are placed on the skin surface. In Figure 2.5, the different electrode-skin layers (a) and the equivalent circuit model (b) are depicted. Each skin layer has an equivalent electrical model (Neuman 2009), [33]. The Cabin and Ratin represent the electrical behaviour of the epidermis, and the dermis and subcutaneous layer is represented by a resistor Rhine. Conventional skin electrodes are provided with an electrolyte gel layer that enables the transfer of ionic currents and that is modelled by the resistor Reel, and the interface electrode-electrolyte is represented by CDoubleLayer and Roublelaver, Figure 2.5.b The elements VHuif Cell and Vabis are generated by an accumulation of charges between the layers; this effect is known as the "Helmholtz double layer" effect.

Advances in textile materials and conductive yarns allow the development and validation of Textile Electrodes (Textrodes) for use with EBI measurements (Medrano, Beckmann et al. 2007, Beckmann, Neuhaus et al. 2010, Marquez, Seoane et al. 2013). Dry textrodes present high electrode-skin impedance because of the absence of electrolytic medium between the electrode and the skin: this effect is reduced sometime after the electrode has been applied and the skin starts sweating, which allows ions that are present in the sweat to function as an electrolytic interface. Factors such as the textile structure, the choice of textile-conductive. materials, and the skin hydration status could affect the textrode performance, and their influence must be taken into consideration when textrodes are used. The proliferation of the use of textrodes together

with their performance improvements enable a handful of novel and emerging applications, such as in the field of home healthcare and personal health systems.

As discussed in the previous section, the skin-electrode impedance polarization can play an important role in the estimation of the impedance Ztuaçu, especially for 2-and 3-electrode configurations. Measurement errors could occur due to several factors such as the use of non-ideal electronic instrumentation, the presence of high electrode polarisation impedance, stray capacitances or the presence other types of artefacts (Bogónez-Franco, Nescolarde et al. 2009, Buendía, Bogónez-Franco et al. 2012), [12]. Therefore, to minimise all of the possible errors, the selection of electrodes and the electronic designs must be considered in the development of EBI measuring systems.

## 2.3 EBI for body composition assessment

The use of EBI technology for the estimation of Body Composition Assessments (BCA) offers a quick, inexpensive and non-invasive measurement procedure compared with clinical methods such as Magnetic Resonance Imaging (MRI), Computed Tomography (CT) or Dual Energy X-Ray Absorptiometry (DEXA). These clinical methods are used as "Golden Standards" for BCA and they offer better accuracy and consistency compared to EBI methods: however they require expensive equipment that is usually located in clinies or hospitals. Traditional methods are invasive and require procedures such as blood samples, exposure to X-Ray radiations or the administration of contrast agents, which make these methods unsuitable for continuous monitoring applications.

The use of EBI spectroscopy measurements, Cole modelling and Hanai mixture theory has become one of the most common approaches for estimating BCA parameters (Van Loan, Withers et al. 1993, Matthie 2008). The following sections will introduce all of the factors that are related to the acquisition of BCA parameters from EBI spectroscopy measurements.

### 2.3.1 Cole modelling

In 1940. Kenneth S. Cole presented the empirical Cole function, see Equation (2.4) (Cole 1940), The Cole function is a complex non-linear function that models. and experimentally fits EBI measurements in the ß dispersion range, between several kHz and a few hundreds of MHz. The equation is defined by four parameters Ro Re a, rand the independent variable that represents the natural frequency.

$$Z(\omega) = R_\infty + \frac{R_0 + R_\infty}{1 + (j\omega\tau)^\alpha} \qquad (2.3)$$

In Equation (2.3), the parameter Ro represents the resistance at zero frequency, which contains information about the conductivity of the EC medium. R. represents, the resistance at a high frequency that includes information about the conductivity of both mediums, the EC and the IC: the parameter is the relaxation time constant, which is the inverse of the natural frequency 1/6, and a is a dimensionless parameter for which there is no clear explanation regarding its origin.The Fricke's Model and the Cole Function are quasi-compatible. The EC medium resistance Re is equal to the resistance at zero frequency Ro, as in Equation (2.4): at high frequency, the ionic currents flow through the EC and IC mediums. Therefore, the IC medium resistance R can be obtained as shown in Equation (2.5), and the membrane capacitance can be extracted using the Cole parameter r.

$$R_e = R_0 \qquad (2.4)$$

$$1/R_\infty = 1/R_e + 1/R_i \qquad (2.5)$$

To obtain the four Cole parameters, the measurement data are fitted to the Cole function using computational methods that will find the Cole parameters that approximate close as possible the experimental data. One implemented method is to fit the data to the semicircle in the impedance plane (Kun and Peura 1999). This method offers a good estimation of Ro, Ra and a, but the estimation of the parameter ris less accurate because

this method does not account for the frequency information of the EBI measurements. The Non Linear Least Square (NLLS) fitting method is also used extensively to fit the Cole model in the different planes, such as the magnitude plane or the imaginary plane, and it has shown good results compared with other fitting methods for the estimation of Cole parameters for their use in BCA applications (Buendia, Gil-Pita et al. 2011, Nordbotten, Tronstad et al. 2011).

## 2.6 Literature Review

Several studies have explored the use of bioimpedance for renal monitoring:
Piccoli et al. (2004) demonstrated that whole-body BIA can estimate fluid overload in hemodialysis patients with reasonable accuracy.
Moissl et al. (2013) developed a body composition monitor (BCM) based on BIA and showed its effectiveness in dialysis treatment planning.
Gonzalez-Correa et al. (2016) proposed a wearable impedance monitoring system for tracking hydration status, which showed potential for CKD applications.

Despite these advances, most

commercially available systems are expensive, require clinical supervision, or are not optimized for daily home use. Moreover, they may lack adaptability and open-source accessibility, limiting their integration into personalized healthcare models.

# Chapter 3
# Project Design

# 3. Project Design

This chapter introduces the general block diagram of the device and the materials and methods that are used for the implementation of portable EBI spectrometer devices. The first part will enumerate some of the system requirements that are considered for the development of the portable EBI prototype spectrometer. Subsequently, the novel System-on-Chip (SoC) Impedance Network Analyser AD5933 from Analog Devices Inc. (Analog Devices Inc. 2013) will , the Textrodes and traditional electrodes that are used for the validation of the system. will be described.

## 3.1 System Block Diagram

Block Diagram for Bioimpedance-based Kidney Function Monitoring Device

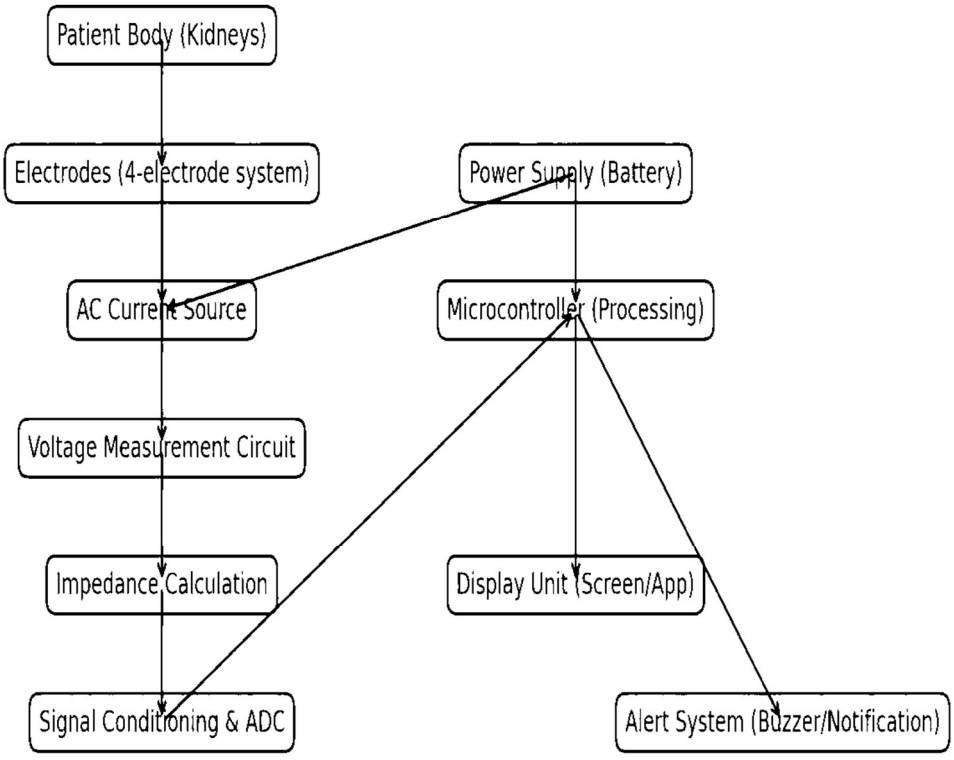


**Figure (3.1) Flowchart of the Kidney Function Monitoring Device Using Bioimpedance Technology**

## 3.2 Portable EBI spectrometer requirements

The requirements have been kept very general in terms of the specific EBI application; therefore factors such as usability requirements, wearability or system integration aspects, or critical factors for the implementation of a Personalized Healthcare Systems PHS are not considered in these first studies. The portable device will allow its use for different EBI applications, where the instrumentation can be customised to perform impedance measurements over different impedance ranges, e.g., in terms of the Total Body Composition.

This thesis discusses the development of EBI spectrometer
that can perform measurements using textile electrodes and functional garments with integrated textile electrodes. Therefore, some of the identified system requirements are the following:

- System accuracy; the implemented monitoring device should perform measurements that are equivalent to commercial EBI spectrometers.
- Patient electrical safety; the device must fulfil the requirements for patient electrical safety that are imposed by the standard IEC-60601-1 for medical electrical equipment (International Electrotechnical Commission 2010).
- Portability; the device should be small, lightweight, and battery operated,
- To enable accurate and flexible impedance measurements in a compact format, the Analog Devices AD5933 System-on-Chip (SoC) is used as a core component in the device. This integrated circuit is a high-precision impedance network analyzer capable of generating a frequency sweep and measuring the complex impedance of biological tissues.

- Figure 3.2 illustrates the block diagram of the AD5933 SoC, highlighting its key functional units: the programmable frequency generator, digital-to-analog converter (DAC), transimpedance amplifier (TIA), analog-to-digital converter (ADC), and digital signal processing (DSP) block. This configuration allows the SoC to generate sinusoidal signals, apply them to the tissue via electrodes, and compute the corresponding impedance magnitude and phase over a range of frequencies.

- The compact integration of these functions in a single chip makes the AD5933 particularly suitable for portable and wearable EBI applications.

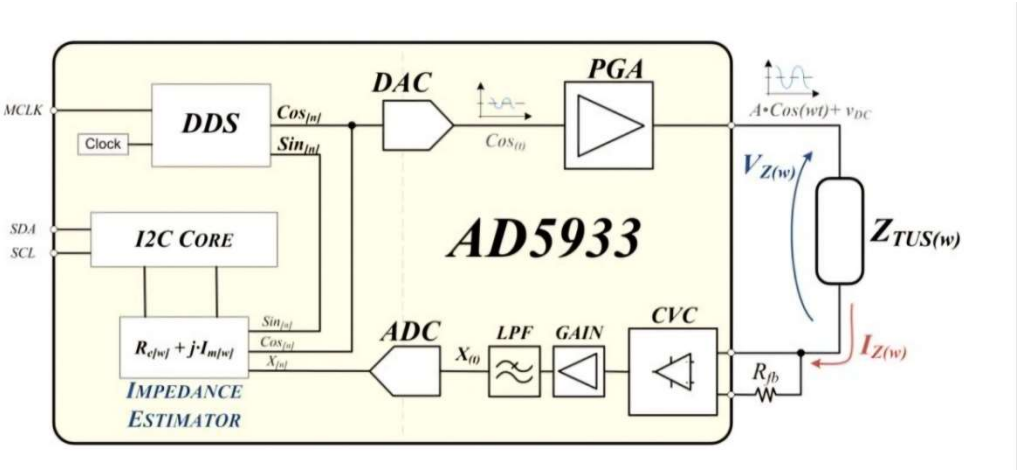


**Figure 3.2 SoC AD5933 Impedance Network Analyser Block diagram.**

The choice of electronic components for the implementation of the portable EBI monitor is based on the previously listed requirements; therefore factors such as power consumption, chip dimensions and printed electronic circuit technology are the key elements for the device implementation.

## 3.3 The system-on-chip impedance network analyser

The novel SoC Impedance Network Analyser AD5933 is, to date, the only SoC solution that incorporates all of the functional components that are necessary to characterise the impedance over a frequency range using a bipolar configuration. A typical measurement of the frequency range is from 1 kHz to 100 kHz: the impedance dynamic range is from 1k2 to 10M2, with a system accuracy of 0.5%, and it is provided with a serial I2C interface to control the internal functions and the measurement retrieval. The AD5933 can operate with a single power supply between 2.7v and 5.5v; it consumes only 33mW at 3.3v and has dimensions equal to 8x6x2 mm (Analog Devices Inc. 2013).

The main AD5933 functional blocks are depicted in the diagram in Figure 3.1. They are also listed below:

- The Stimulation Stage: This stage is formed by a Direct Digital Synthesizer (DDS) which generates the digital sine waves that could be configured for the specific impedance excitation frequency, denoted as ωn, a 12-bit Digital to Analog Converter (DAC) and a Programmable Gain Amplifier (PGA) which adapts the generated voltage that will be applied to the Impedance Ztus(ω).
- The Receiver Stage: This stage is formed by a Current to Voltage Converter (CVC), a Voltage Amplifier, an antialiasing Low Pass Filter (LPF) and a 12bit. Analog to Digital Converter (ADC). The receiver stage will adapt and convert the flowing current through Ztus(ω) into a voltage that will be fed to the following stage.
- The Impedance Estimation Stage: This stage uses 1024 sample points from the signals that are generated by the DDS module, cos (ωn.n) and sin(ωn.n), and the digitalised input signal x[n]to estimate the real and imaginary impedance values using the single Discrete Fourier Transform (DFT) method, see Equations (3.1) to (3.3).
- The chip control stage: This stage is formed by the 12C Core module that control all of the chip functions and stores all of the measurement results among other parameters; see the AD5933 datasheet for further details.

To characterise the impedance over a frequency range, the AD5933 uses a frequency sweep; thus, each time that a single frequency point is analysed, the sequence is repeated at a different frequency until the entire spectrum is obtained. For each frequency, the AD5933 will obtain 1024 samples of the output and input signals to estimate the real and imaginary component values performing a 1024-point. single frequency DFT analysis, which is also denoted as the Sine Correlation Method in the digital form previously introduced

$$X_{(w_n)} = \sum_{n=0}^{1023} \left( x_{|n|} * (cos[w_n \cdot n] - j * sin[w_n \cdot n]) \right) = Re_{(w_n)} - j \cdot Im_{(w_n)} \quad (3.1)$$

$$Re_{(w_n)} = \sum_{n=0}^{1023} x_{|n|} * \cos[w_n \cdot n] \quad (3.2)$$

$$Im_{(w_n)} = \sum_{n=0}^{1023} x_{|n|} * \sin[w_n \cdot n] \quad (3.3)$$

To obtain the impedance magnitude and phase values, the obtained real and imaginary components must be adjusted by a calibration factor; check the datasheet for the complete operation procedure (Analog Devices Inc. 2013). The calibration factor is obtained by performing a frequency sweep over a known impedance value. The manufacturer suggests that a middle single-point or two-point calibration method using a resistor could be used to characterize the system over a frequency range.

## 3.4 Traditional and textile electrodes

The portable EBI monitor was evaluated in human subjects by performing EBI measurements. Therefore, two types of electrodes were used: traditional gel Silver-silver chloride (Ag/AgCl) electrodes and custom-made strap garments that incorporate textrodes.

**Figure 3.3** Functional straps with incorporated textrodes for TRS measurements(left) and Ag/Agcl 3M traditional electrodes (right)

The 3M gel Ag/AgCl electrodes, shown in Figure 3.3 right side, belong to the category of non-polarisable electrodes, which allow the transfer of charges with minimum voltage generation at the skin-electrode. The electrodes are formed by a conductive and

sticky electrolytic gel that it is in contact with the skin, and an electrically conductive snap button which allows the connection with the electronic instrumentation.

The functional textrode straps (Marquez Ruiz 2013), which are shown in Figure 3.3 left side, were used to obtain total right side EBI measurements, and its design allows the correct placements of electrodes to avoid as much as possible any electrode placement variability. Each strap incorporates two separate inner layer of conductive fabric that are in contact with the skin and that are used as electrodes. The conductive fabric material is a knitted silver coated fabric P130+B from the manufacture Shieldex Technik-Tex, which is made of 78% polyamide and 22% elastomer and is coated with 99% conductive silver particles with an approximate resistivity lower than 2 2/sq. The electrode areas are provided with an inner layer of foam that applies pressure to the conductive fabric, to improve the skin-electrode contact area. The outer layer is made of blue synthetic wrap-knitted fabric. The straps are provided with Velcro fasteners and snap buttons on each textrode for the electronic interconnection.

## 3.5 Control Unit: Arduino Uno

In this project, the Arduino Uno board was used as the main control unit responsible for coordinating the different components of the device, such as the AD5933 measurement chip, the display screen, and the data logging interface. Arduino is an open-source electronics platform based on an easily programmable microcontroller using the Arduino IDE.

Figure 3.5 shows the Arduino Uno board used in the system. It serves as the central processing unit, responsible for handling data acquisition, communication, and control operations for the entire device.

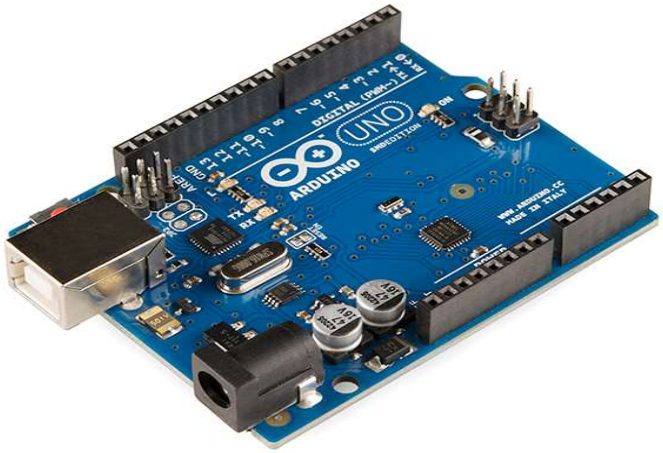


**Figure 3.5 Arduino Uno.**

### 3.5.1 Functions of the Arduino in the system:

- Sending control commands to the AD5933 chip via the I²C interface.
- Receiving data from the chip and performing preliminary processing.
- Sending the processed data to the display
- screen or logging system for result monitoring.
- Managing the workflow of the system and ensuring synchronization between measurement and display.

### 3.5.2 Specifications of Arduino Uno:

| Feature | Value |
|---|---|
| Microcontroller | ATmega328P |
| I/O Pins | 14 Digital, 6 Analog |
| Communication Interface | sI²C, UART, SPI |
| Operating Voltage | 5 V |
| Input Voltage (recommended) | 7 – 12 V |
| Flash Memory | 32 KB |

### 3.5.3 Advantages of using Arduino:

- Easy programming with a simplified language.
- Availability of libraries for interfacing with
- the AD5933 chip.Low cost and easy wiring.
- A large support community, which accelerates the development process.

## 3.6 Display Unit: Nextion Touchscreen

In this project, a Nextion touchscreen display was used to present measurement results in a clear and interactive manner. Nextion displays are intelligent human-machine interface (HMI) solutions that integrate a TFT display, touch panel, and onboard processor. They simplify GUI design by offloading graphical processing from the main microcontroller, allowing for a more responsive and efficient user interface.
Figure 3.6 shows the display unit used in the system.

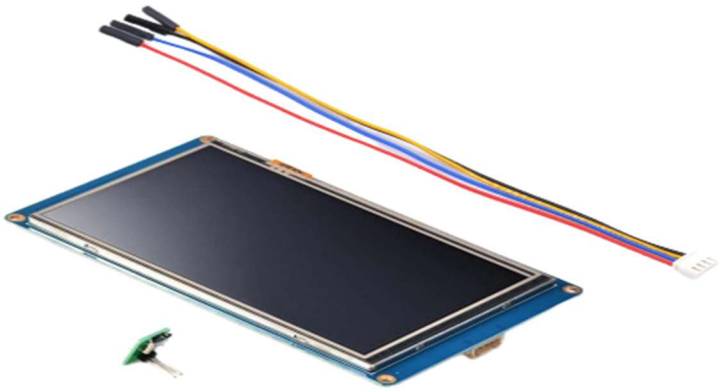

**Figure 3.6 Display Unit.**

### 3.6.1 Functions of the Nextion display in the system:

- Displaying impedance measurement results in real time.

- Providing a graphical user interface (GUI) that enhances user interaction.
- Receiving serial commands from the Arduino to update displayed values.
- Supporting user input through its touch functionality for future system enhancements.

### 3.6.2 Specifications of the Nextion display (model used in this project):

| Feature | Value |
|---|---|
| Display Type | TFT LCD with resistive touch |
| Screen Size | 5 inches |
| Resolution | 800 × 480 pixels |
| Communication Interface | UART (Serial) |
| Operating Voltage | 5 V |
| Flash Memory | 16 MB |

### 3.6.3 Advantages of using Nextion display:

- Simplifies GUI design using the Nextion Editor software.
- Reduces load on the main controller by handling graphics independently.
- Offers touch functionality for interactive applications.
- Easy integration with Arduino via serial communication.

## 3.7 Power Supply Unit: Dual 3.7V Lithium Batteries

The power supply for this project is based on two 3.7V lithium-ion batteries connected in series, providing a total output voltage of approximately 7.4V. This configuration ensures sufficient and stable power for the Arduino, AD5933 chip, and the display module. The batteries are rechargeable and integrated with a USB charging module, allowing easy recharging through a standard USB connection.

Figure 3.7 Dual 3.7V Lithium Batteries illustrates the configuration of the batteries used in the system

**Figure 3.7 Dual 3.7V Lithium Batteries**

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### 3.7.1 Functions of the power supply system:

- Providing a stable voltage source for the entire device.
- Enabling portable and wireless operation, especially for field or bedside applications.
- Supporting recharging via a micro-USB port without disassembling the device.
- Maintaining reliable operation over extended periods with high energy density.

### 3.7.2 Specifications of the power unit:

| Feature | Value |
|---|---|
| Battery Type | Lithium-ion (Rechargeable) |
| Nominal Voltage (each) | 3.7 V |
| Total Output Voltage | ~7.4 V (connected in series) |
| Charging Interface | Micro-USB |
| Charging Module | TP4056 or equivalent |
| Protection Circuit | Overcharge and overdischarge protection included |

### 3.7.3 Advantages of using lithium-ion batteries:

- High energy density and long battery life.
- Lightweight and compact design suitable for portable devices

- Rechargeable via standard USB ports.
- Stable voltage output for sensitive electronic components.

## 3.8 Design Considerations

**Several technical and medical factors were considered during the design process:**

- Electrical Safety:
- The current used is within safe biological limits to ensure no harm to the user.
- Measurement Accuracy:
- The AD5933 was selected for its high precision in impedance measurement.
- Portability:
- Rechargeable lithium battery enables use in various environments.
- Cost-effectiveness:
- Components were selected to balance performance and affordability.
- User Friendliness:
- A simple and intuitive display interface was implemented for ease of use.

## 3.9 Challenges and Solutions

**The project encountered various challenges, including:**

### Electrical Noise:

Addressed using shielded cables and software filtering techniques.

### Device Calibration:

Achieved through the use of known reference resistors prior to biological testing.

### Power Management:

Integrated the TP4056 charging module with built-in protection for safe battery operation

# Chapter 4
# Implementation and Testing

# 4. Implementation and Testing

This chapter details the practical implementation of the bioimpedance-based kidney monitoring device and the procedures followed to test its functionality. It outlines the hardware assembly, software deployment, and the experimental process used to validate the system's performance in real-life conditions.

## 4.1 Hardware Assembly

All selected components were integrated based on the finalized circuit design. The

### system includes:

1) Arduino Uno microcontroller
2) AD5933 impedance converter module
3) LCD display
4) 3.7V lithium-ion battery
5) TP4056 charging and protection module
6) LM7805 voltage regulator (3.3V/5V as needed)
7) Electrode cables with surface electrodes (placed on wrist and ankle)
8) Power switch and basic enclosure

The components were connected using jumper wires and soldered where necessary to ensure signal integrity and mechanical stability. The system was enclosed in a compact, portable case with external electrode ports and a display window

## 4.2 Software Deployment

The device firmware was developed usingthe Arduino IDE. The main functions of the program include:

1) I2C communication with AD5933
2) Frequency sweep setup and control
3) Real and imaginary data acquisition
4) Conversion to magnitude and impedance calculation
5) Displaying results on the LCD screen
6) Basic noise filtering and averaging
7) Calibration routines were included using known resistors ( 10kΩ) to correct for system offsets.

## 4.3 Calibration Procedure

To ensure measurement accuracy, the device was first calibrated using precision resistors. The process involved:

Connecting a known resistor across the electrode terminals Measuring the output impedance Calculating the gain factor and phase shift internally Applying these corrections to biological readings

This step was crucial in ensuring reliable comparison between subjects and repeated tests.

## 4.4 Testing on Human Subjects

The device was tested on several healthy volunteers from different age groups and genders to evaluate functionality and sensitivity. Each test followed this procedure:

The subject was seated comfortablyElectrodes were placed on the **right wrist and right ankle**The skin was cleaned with alcohol, and conductive gel was used The measurement was initiated via a physical buttonThe result (magnitude or impedance) was displayed on the LCD

## 4.5 Limitations Noted During Testing

Electrode contact quality significantly affected measurement consistency.

Movement artifacts led to fluctuations in readings during non-stationary conditions.

Two-electrode configuration introduces electrode-skin impedance, reducing absolute accuracy

## 4.6 Practical Implementation Images and Diagrams

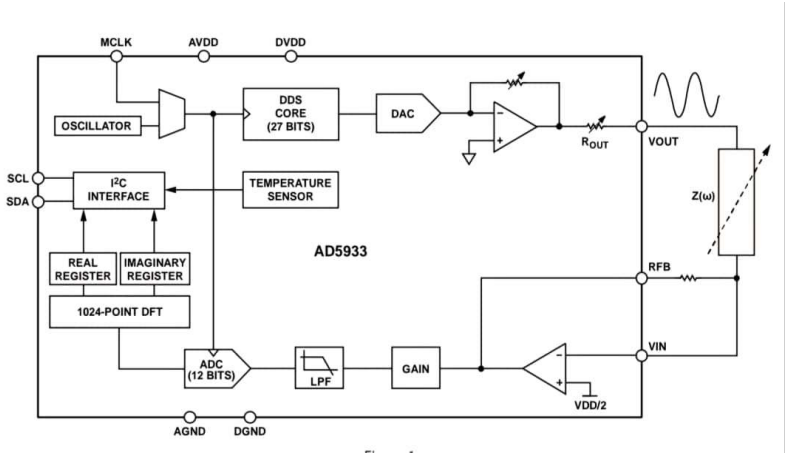


**Figure 4.1 Schematic diagram designed by supplier.**

This schematic integrates the AD5933 impedance converter core, passive filtering components, and an I2C interface for seamless microcontroller communication.

Figure 4.1 illustrates the basic schematic diagram of the system as designed by the supplier. It shows the interconnection of the main electronic components, such as the measurement unit, control unit, and display module. This diagram helps in understanding the overall structure of the device and facilitates its assembly and operation**.**

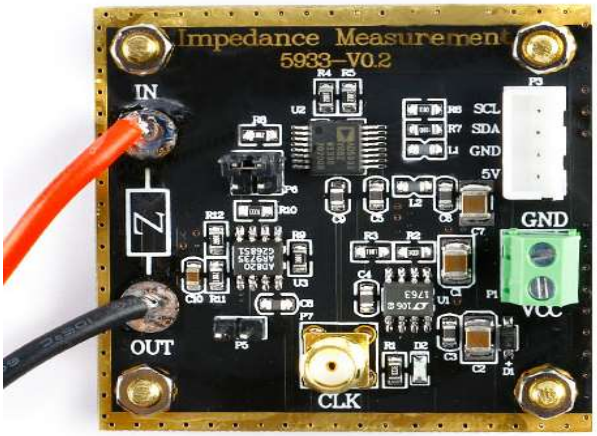


**Figure 4.2 The fabricated PCB of the impedance measurement system.**

Figure 4.2 shows the fabricated PCB of the impedance measurement system, which houses all these components in a compact and efficient design, ready for practical implementation.

Includes the AD5933 core, supporting circuitry, and a modular layout for easy integration with microcontroller units.

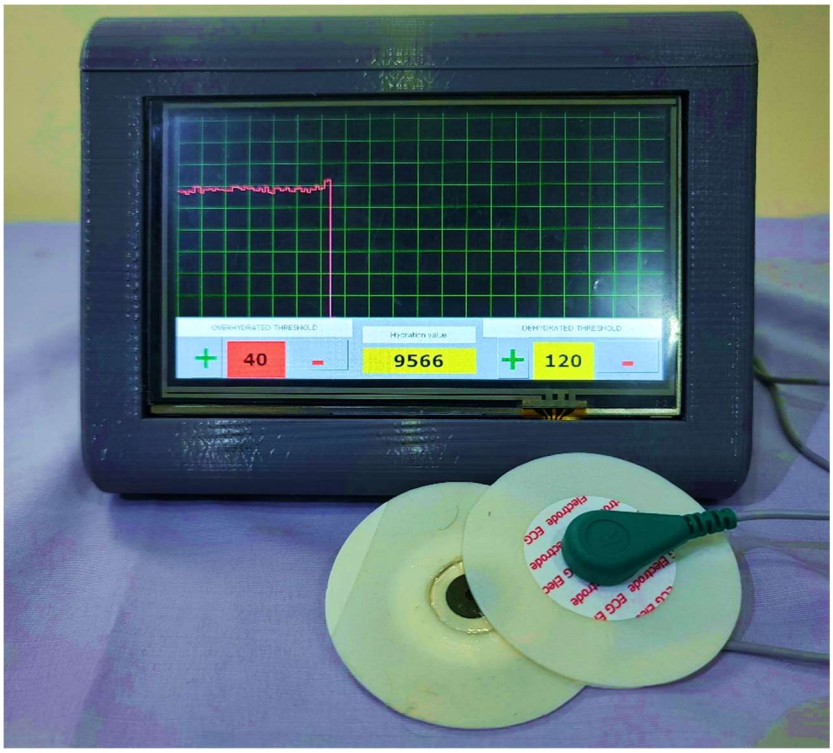


**Figure 4.3 Final assembled device – front view.**

The system is housed In a 3D-printed, pyramid-shaped enclosure, featuring a 5-inch capacitive Nextion touchscreen display (model NX8048T050_011) prominently mounted on its front surface. This enclosure not only provides mechanical protection and structural support for internal components but also enhances the device's ergonomic design and user interaction.
Figure 4.3 shows the final assembled device front view, highlighting the integration of the touchscreen and the sleek enclosure design.

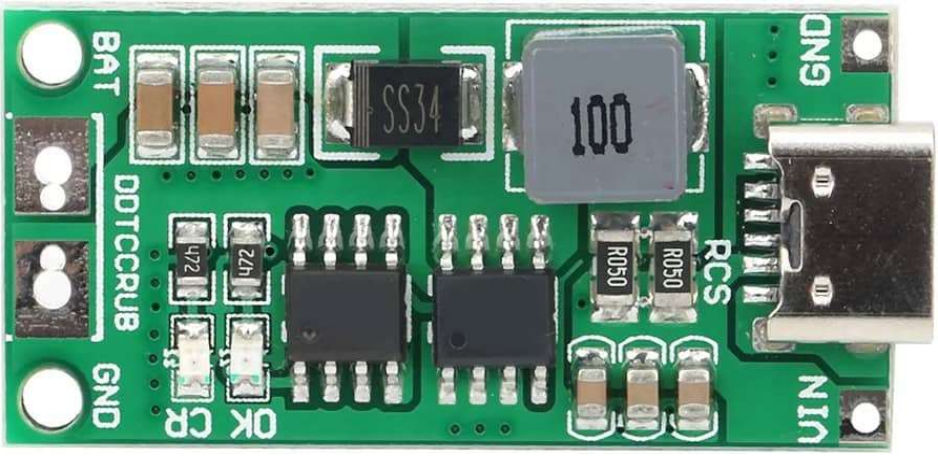


**Figure 4.4 Power supply module and charging system** .

The power supply system of this device is based on two 3.7V lithium-ion batteries connected in series, delivering a total voltage of approximately 7.4V. This configuration ensures stable and sufficient power for all components, including the Arduino board, AD5933 impedance converter, and Nextion touchscreen display. A dual-cell charger and boost module is employed to manage the charging process and provide a regulated 5V output, guaranteeing system stability and protection during operation. The integration of a USB charging interface allows convenient recharging via standard USB sources**.**

Figure 4.4 shows the power supply module and charging system, highlighting how the dual-cell charger and boost converter are integrated to ensure efficient energy management and system reliability.

# Chapter 5
# Results and Discussion

## 5.1 Project Results

The bioimpedance-based kidney function monitoring device prototype yielded promising results that support its potential clinical value. The key findings are as Follows:

1. Successful Measurement Implementation**:**

The device was able to measure tissue bioimpedance consistently across multiple individuals using a two-electrode configuration. Surface electrodes were applied to the wrist and ankle to measure impedance related to extracellular fluid levels, which are indicative of kidney function.

2. Impedance Ranges Across Age and Gender:

Healthy adult males (18–50 years): 40 – 120 Ω

Healthy adult females (18–50 years): 50 – 130 Ω

Children (5–17 years): 60 – 160 Ω

Men over 50: 60 – 150 Ω

Women over 50: 70 – 170 Ω

These results align with known physiological fluid distribution, with higher impedance generally observed in females and older subjects due to lower water content.

3. Repeatability and Reliability:

Measurements demonstrated good repeatability within ±5% when electrode placement was consistent.

The device was sensitive to hydration status, as a noticeable decrease in impedance was observed after fluid intake.

4. Practical Observations:

Stable results were obtained when the user remained still.

Poor skin contact or sweat introduced variations, emphasizing the importance

consistent electrode application

5. Technical Limitations Identified:

Movement artifacts and electrode contact quality significantly affected measurement accuracy.

The two-electrode setup introduced skin-electrode impedance, slightly compromising precision  n.

Despite these limitations, the system showed a strong correlation between measured impedance and known physiological indicators of kidney function. The results validate the potential of this portable, low-cost device for non-invasive monitoring, particularly in resource-limited or home settings**.**

# Chapter 6

# Conclusions and Recommendations

## 6.1 Conclusions

This graduation project successfully demonstrates the design and initial implementation of a non-invasive, bioimpedance-based kidney function monitoring device. By integrating the AD5933 impedance converter, Arduino Uno microcontroller, surface electrodes, and a user-friendly touchscreen interface, the prototype is capable of estimating extracellular fluid levels—an important indicator of kidney performance.

The results obtained from testing on healthy volunteers revealed consistent and physiologically relevant impedance values, supporting the potential of bioimpedance as a tool for early kidney dysfunction detection. Although the system is still in its prototype stage and lacks clinical validation, it provides a promising foundation for affordable, portable, and continuous kidney monitoring—especially in low-resource settings or for home use.

## 6.2 Recommendations

To enhance the performance, reliability, and clinical relevance of the device, the following recommendations are proposed for future work:

1. Validation Clinical

Conduct clinical trials comparing the device's output with standard kidney function indicators, such as serum creatinine and eGFR, to confirm diagnostic accuracy.

2. Four-Electrode Configuration

Upgrade the measurement setup from a two-electrode to a four-electrode (tetrapolar) configuration to minimize skin-electrode impedance effects and improve measurement precision.

3. Advanced Signal Processing

Implement digital filtering techniques (e.g., moving average, Butterworth, or wavelet filters) to reduce noise and motion artifacts during real-time measurements.

4. Analog Front-End Integration

Incorporate a dedicated analog front-end (AFE) for better signal conditioning, gain control, and impedance matching to enhance measurement stability and sensitivity.

5. Extended Frequency Range

Expand the operating frequency range to improve the distinction between intracellular and extracellular fluids, which will enhance the physiological insight of the data.

6. Enhanced User Interface

Develop a more intuitive graphical interface with real-time monitoring graphs, data logging, and alerts to improve usability for both clinicians and patients.

7. Wireless Connectivity

Add Bluetooth or Wi-Fi modules for remote data transmission and integration with mobile health platforms for continuous remote monitoring.

These improvements will help transform the prototype into a clinically viable device suitable for broader application in healthcare environments.

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