ECE 445
Fall 2025
Design Document

Heart Restart

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1 Introduction

1.1 Problem

Research has shown that defibrillators delivering a single shock achieve relatively low survival rates of only 13.3%. In contrast, Double Sequential External Defibrillators (DSED)—which deliver two rapid consecutive shocks—significantly improve survival rates to 30.4%. However, DSED currently requires two separate defibrillators, making it impractical in real-world emergencies, particularly in ambulances where only one defibrillator is typically available. Furthermore, the precise millisecond-level timing required between each shock further increases the difficulty of coordinating two devices. In addition, most commercially available defibrillators lack impedance measurement capabilities, preventing them from automatically adjusting shock strength and timing based on patient-specific characteristics such as body type and chest impedance. Those that do include such sensing capabilities tend to be larger, more complex, and significantly more expensive, which limits their accessibility in emergency medical settings.

1.2 Solution

Our project aims to design a compact, intelligent defibrillation system capable of delivering Double Sequential External Defibrillation (DSED) using only a single defibrillator unit. By integrating impedance sensing and automated shock timing control, our design eliminates the need for multiple defibrillators and ensures that both shocks are delivered within the optimal millisecond window. This innovation not only enhances practicality in emergency scenarios but also improves patient outcomes by optimizing shock delivery based on individual impedance measurements. Compared to existing systems, our approach offers a smaller, cost-effective, and adaptive solution that bridges the gap between clinical efficacy and real-world deployability.

1.3 Visual AID

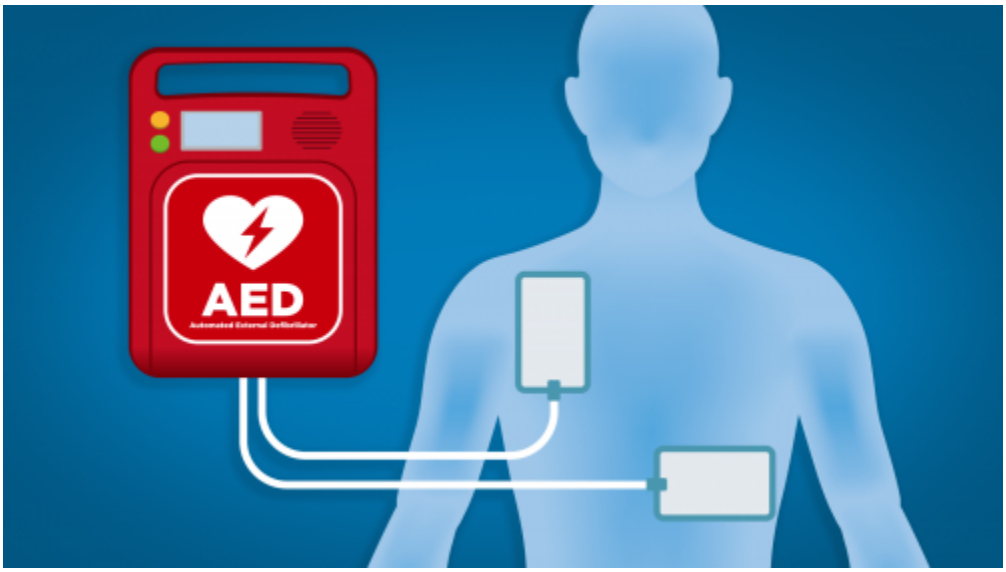

Figure 1: Modern publicly accessible AED

Modern and easily accessible AEDs today are single-shock devices; however, as described in the problem, this could result in a typical survival rate of only 18%. Not to mention the cost of a typical AED, which at the cheapest would run an organization upwards of $ 1,400. For this project, we will continue with a similar design seen in Figure 1.

Our design will be a little different from what a typical AED has; instead of two leads, we will have four total. Two will be dedicated to monitoring heart rate, and the other will be for impedance sensing. This is seen in Figure 2 on the placements of these leads.

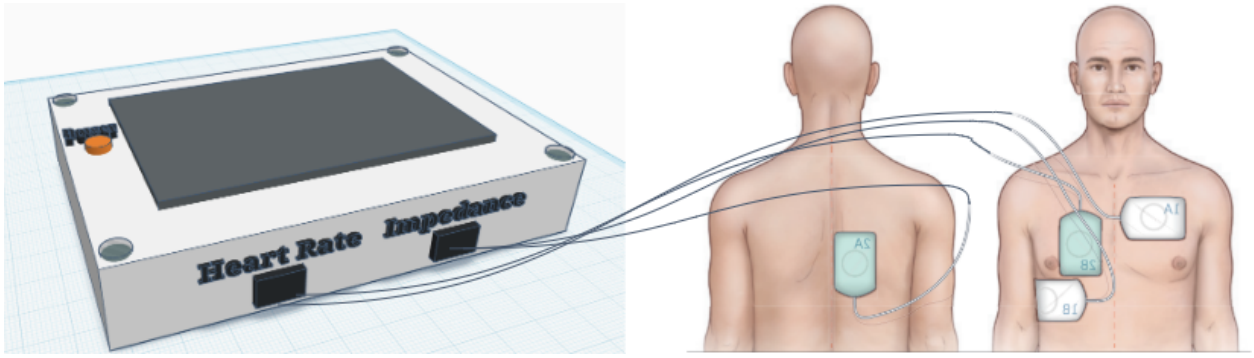

Figure 2: Leads and Positions

For testing, we are going to use 3.5mm snap connectors, which are compatible with the 3m electrode pads that were given by the medical students, Figure 3.

Figure 3: 3M electrode and snap connector

As one may notice, everything will be one system, thus nothing will be connected via Bluetooth, wifi, or anything RF related.

1.4 High-Level Requirements

To consider our project successful, our design must fulfill the following criteria as expected by our medical student mentors:

1. Our design of the ECG must be able to detect the heart rate. Amplify that heart rate for the microcontroller to sense and have an accuracy of up to 2.5%. This data will potentially be saved to a USB stick for post-analysis and clinical diagnosis.
2. Design an impedance sensor that can accurately measure impedance to within 5%. This will be used to determine a certain shock value from the set of capacitor banks.
3. Display the data on a screen on the device itself. This is for ease of use and a more intuitive way to diagnose a patient on the heart rate and impedance.

2 Design

2.1 Physical Design

As seen in Figure 2, our physical design will be something similar to that, where there will be two outputs with two leads each, one for impedance and one for monitoring heart rate. With that combined with the screen, motherboard, and power unit, our system will effectively have five total parts. As described already, the impedance sensor will take in measurements to accurately determine the amount of charge to shock an individual. The ECG sensor will take the measurements and show the heart rate of the individual to determine whether a shock is needed. The microcontroller will analyze the data sent from the impedance and ECG to be sent to the display and potentially sent to a USB stick. The display will show the user the impedance and the heart rate in real time. Finally, the power unit will supply the necessary power for the other subsystems.

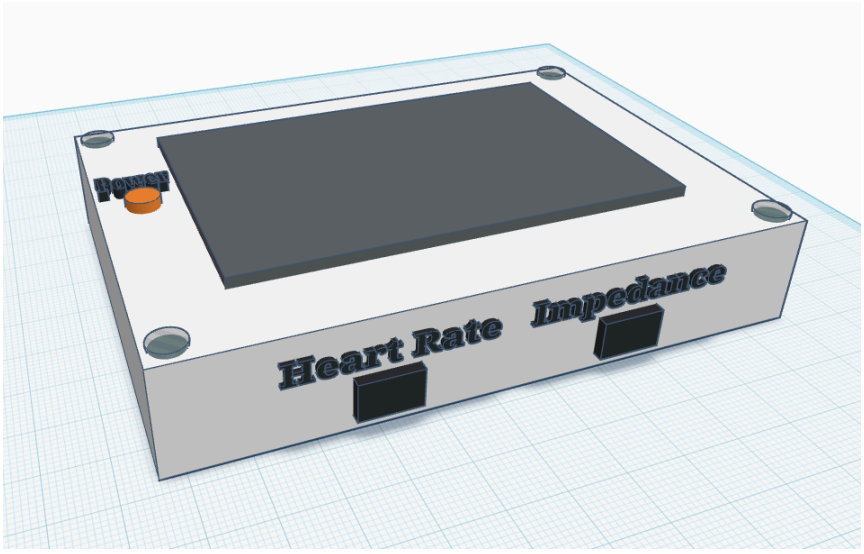

Figure 4: Physical Diagram

2.2 Block Diagram

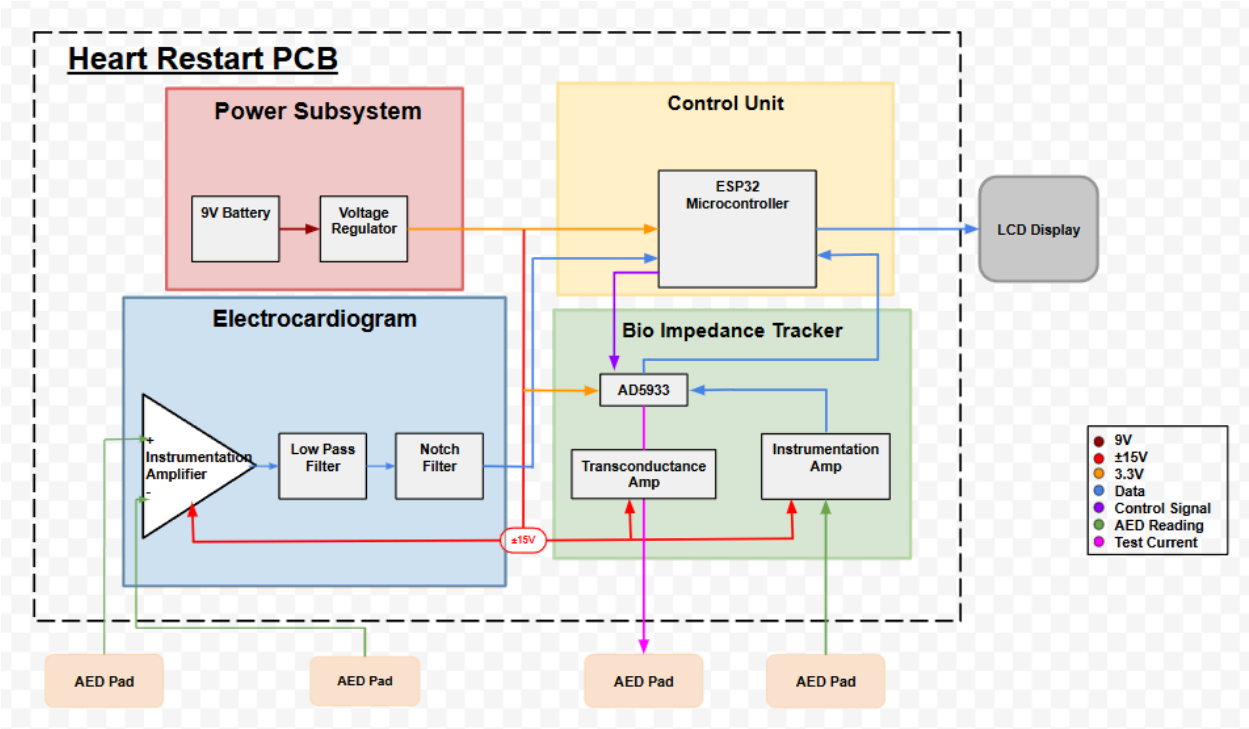

Figure 5: Block Diagram

2.3 Functional Overview & Block Diagram Requirements

2.3.1 ECG Subsystem.

The ECG sensor is responsible for detecting and amplifying the electrical signals that are found when the heart beats. This is done by using three subcomponents. The instrumentation amplifier is used to amplify the signal with respect to the voltage found in the body. This is done via three op amps and various resistors as seen in Figure 6.

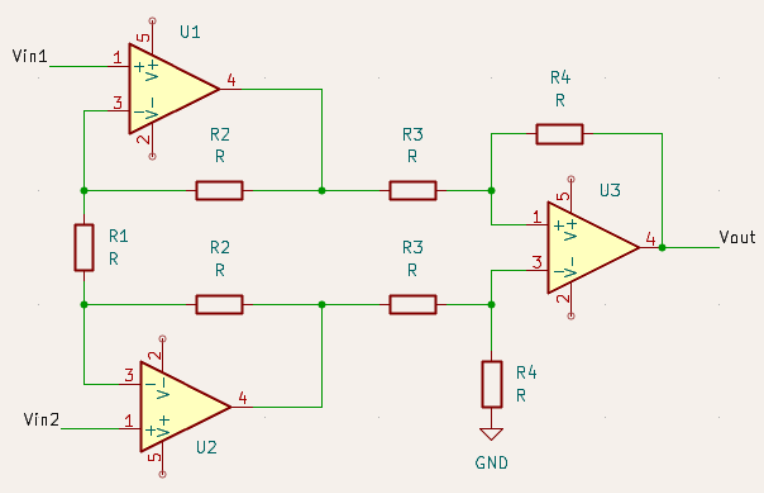

Figure 6: Instrumentation Amp Schematic

The instrumental amplifier will do most of the amplification with a rough gain of 1000. This is calculated via the equation:

$$gain = \frac{Vout}{Vin2 - Vin1} = (1 + \frac{2*R2}{R1})(\frac{R4}{R3})$$

The second part will be the notch filter. The notch filter will effectively remove the 60 Hz AC frequency that naturally gets picked up by the human body. The accuracy of this filter will depend on the Q factor. The Q factor is defined as the ratio of the center frequency f_0 to the bandwidth Δf, or $Q = f_0 / \Delta f$. In this case, we will be using a value of 15. Which will then be applied to the following equations. We chose 15 as it deepens the reduction on the impact from 60 Hz whilst keeping the rest of the frequencies open to amplification. The notch filter we decided to use is a modified multiple-feedback band-pass filter as seen in Figure 7.

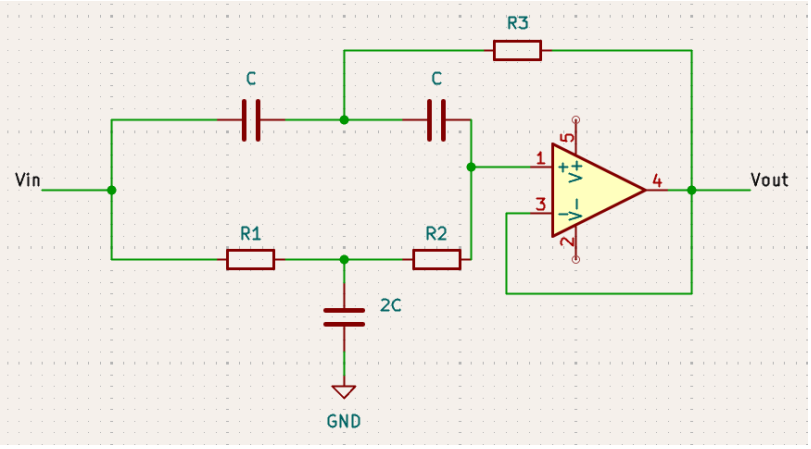

Figure 7: Notch Filter Schematic

We used these formulas and chose 0.56 uF capacitors with a Q factor of 15 to calculate resistor sizes.

$$R_1 = \frac{1}{2Q\omega_0 C} \quad R_2 = \frac{2Q}{\omega_0 C} \quad R_3 = \frac{R_1 R_2}{R_1 + R_2} \quad Q = \frac{\omega_0}{\beta} \quad \beta = \omega_{c2} - \omega_{c1}$$

However, this layout is going to change as we will switch to a dual-op-amp twin-T notch filter, which will allow us to tune our Q factor with greater ease.

The third and final part of the ECG is the low-pass filter. This is currently set at a 150 Hz cutoff frequency and utilizes a Sallen-Key Butterworth design, as shown in Figure 8. We will explore the possibility of using a Chebyshev low-pass filter to achieve a more defined frequency drop and minimize noise as much as possible. This cutoff frequency can continue to be pushed further as we develop more understanding of reading ECG.

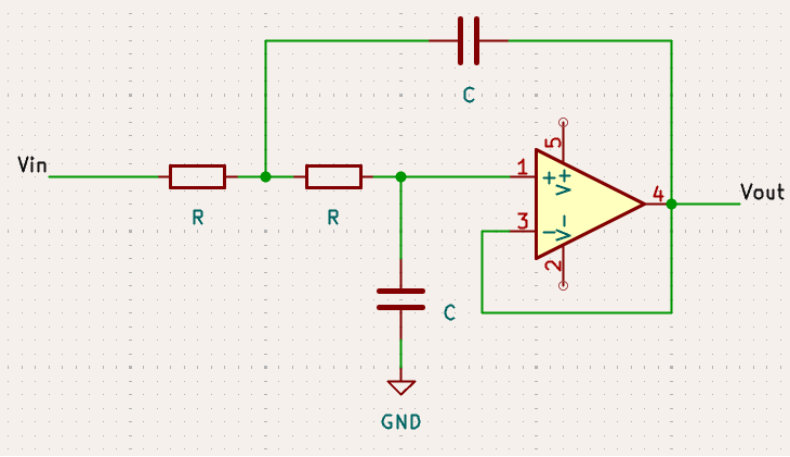

Figure 8: Low-pass filter Schematic

Requirement	Specific Component	Verifications
The amplifier must amplify the signal roughly by 1000 times	Instrumentation Amplifier	• Connect an oscilloscope to the input and the output of the amplifier to ensure amplification
The filter should cut out frequencies around 60 Hz, depending on the Q factor.	Notch Filter	• Insert a parasitic signal of 60 Hz at the input of the notch filter; the output should not see the parasitic signal • Send a small range of frequencies to determine what frequencies are attenuated to match calculations
Frequencies outside of 150 Hz must be attenuated to nearly zero.	Low-pass Filter	• Insert a parasitic signal greater than 150 Hz at the input of the low-pass filter; the output should not see the parasitic signal

2.3.2 Bio Impedance Subsystem

The impedance sensor is responsible for sending small amounts of current into the human body and reading the voltage on the other end of the node. This will determine the impedance of the human body. To start this process, we will use an AD5933, which is a 12-bit impedance converter. There will be three stages to measuring the impedance, starting with the rebias, transconductance amplification, and the instrumentation amplification stages, respectively.

The rebias stage is to center the output voltage signal in the middle of the supply voltage range. This is done by setting a capacitor in series with two equal-value resistors, which are set up in a simple voltage divider as seen in Figure 9. This gets fed into an op-amp to be set up as a voltage follower, making sure that the resistors in this voltage divider do not interfere with the next stage. This stage also acts as a high-pass filter with the cutoff frequency set at 100 Hz.

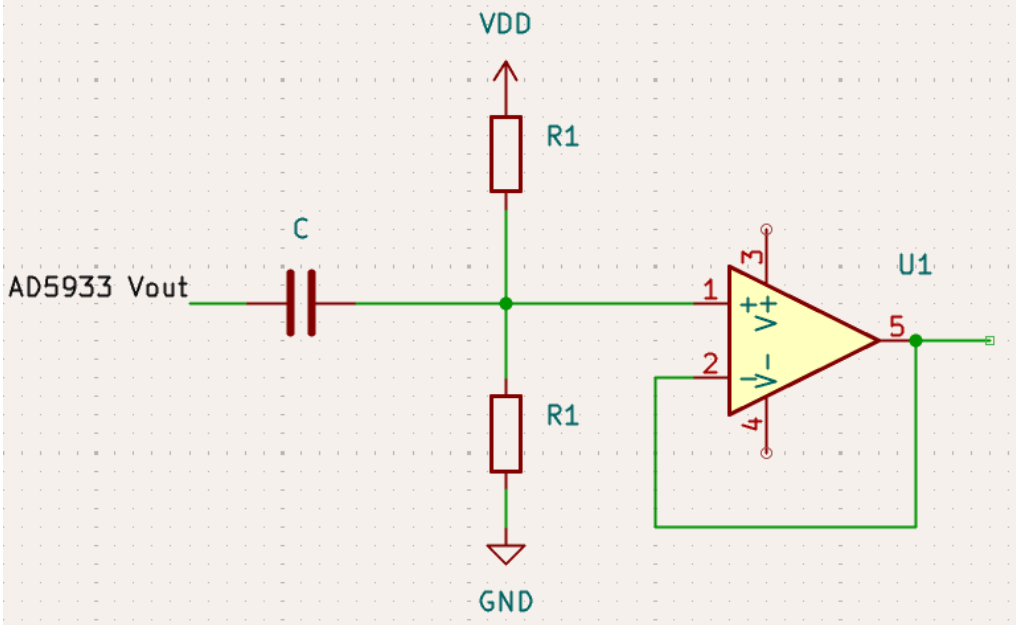

Figure 9: Rebias Stage Schematic

The next stage is the transconductance amplification stage. This stage is where the current sensing resistor, protection resistor, and the trans-conductance amplifier reside. The current sensing resistor makes sure that the current is within safe levels to send through the body. This value is on the order of greater than 100k ohms. This is determined by the voltage from the

rebias stage and the current that is planned to pass through the body. For testing purposes, we have placed a potentiometer to observe and tune the current.

The trans-conductance amplifier is set up where the current coming from the current sensing resistor must flow through the body, as the protection resistor is very big, making the setup into a voltage-controlled current source. The amplifier will also have a reference voltage that will maintain the current through the body. The protection resistor is on the scale of a couple of megaohms, in our case 2 Mohms, as seen in Figure 10.

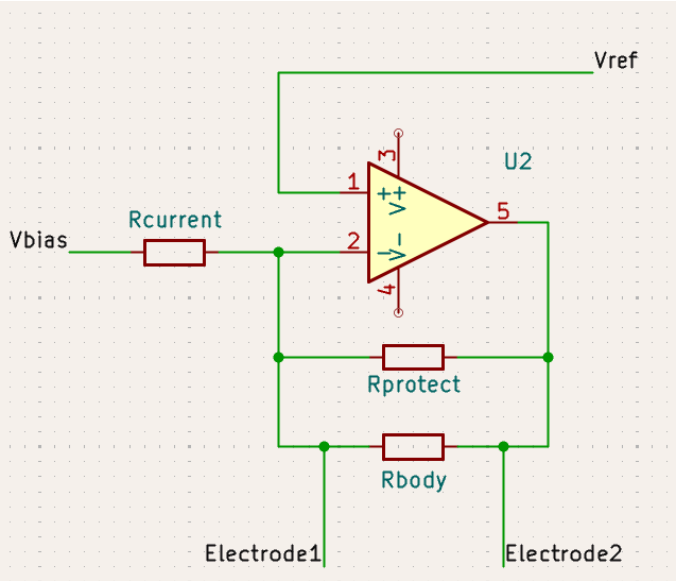

Figure 10: Trans-conductance amplifier Stage

The instrumentation amplifier is the final stage, measuring the voltage across the body. Due to the voltage oscillating around 0, an input voltage Vref is needed, similar to the rebiasing stage. After the amplifier are the two resistors, Rin and RFB. This will provide the gain ratio of A = -RFB / Rin. However, more calibration and calculations will be needed as the AD5933 has its own gain factor that will need to be accounted for. We have decided to go with potentiometers for Rin and RFB with a maximum resistance of 10k and 50k ohms, respectively, as seen in Figure 11. This will allow us to tune the gain, however deemed necessary for the AD5933 to read a clean value. The instrumentation amplifier we went with is the INA333, as it is low-cost and has relatively good noise reduction.

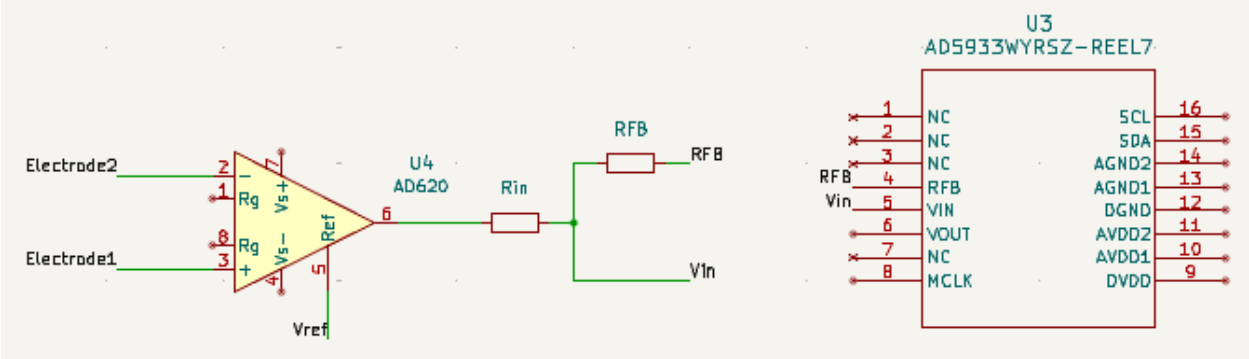

Figure 11: Instrumentation Amplifier Stage

Requirement	Specific Component	Verifications
The rebiasing stage should set the starting voltage at VDD/2 whilst filtering out lower frequencies	Rebiasing stage	• Connect an oscilloscope to measure the voltage before and after the op amp • Send a parasitic signal of less than 100 Hz to determine if high high-pass filter from the capacitor attenuates the frequencies.
The transconductance amplifier should be able to control the current going through a certain resistance	Transconductance	• Connect a voltmeter to probe for voltage and current across a dummy resistor in place of R_{body}. • Test points over $R_{current}$ to enforce current limits.
An instrumentation amplifier should amplify the voltage by the calculated gain.	Instrumentation	• Connect an oscilloscope to RFB and Vref to determine gain • Connect a multimeter to determine the resistance of RFB and Rin

2.3.3 Control Subsystem

The control subsystem, illustrated below in Figure 12, is responsible for receiving signals from the ECG and impedance subsystems, as well as from a start/stop push button. The ESP32-S3-WROOM-1U-N16R8 microcontroller processes these inputs and displays the calculated heart rate, impedance values, and the analog ECG waveform on a liquid crystal display (LCD).

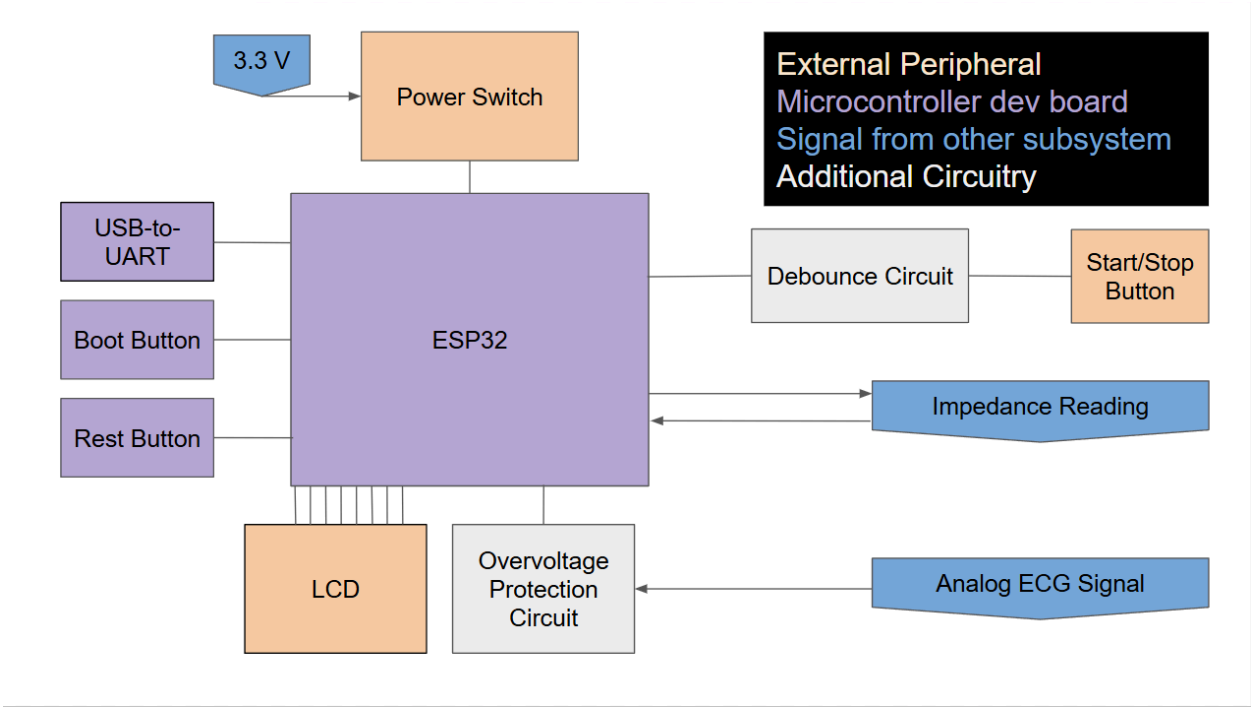

Figure 12: Block Diagram of Control Subsystem

The control subsystem is built around the ESP32-S3-DEVKITC development board. This platform was selected due to its availability for in-person development and testing and its high operating frequency. The board provides USB connectivity and programming interfaces, eliminating the need for a fully custom microcontroller PCB integration. The standard module supplied with the ESP32-S3-DEVKITC has been replaced with a higher-memory variant of the ESP32-S3-WROOM-1U to accommodate video output and data buffering requirements. The ESP32-S3 offers sufficient GPIO pins to support all necessary peripherals, including analog inputs, SPI communication lines, and control signals, making it well-suited for the subsystem's hardware demands.

The selected LCD screen features a 320x240 pixel resolution and is targeted to refresh at 1 Hz. The display communicates over the SPI protocol, which enables high-speed data transfer and straightforward integration with the ESP32's built-in SPI peripheral. The implementation of the SPI connection and display rendering is based on the Adafruit GFX and Adafruit LCD driver libraries, which provide well-documented functions for initializing the display, transmitting pixel data, and drawing graphical elements. This approach reduces development complexity and ensures consistent performance.

A common issue when interfacing mechanical buttons with digital systems is "button bounce," where mechanical contacts briefly oscillate after being pressed or released, resulting in multiple unintended button presses. To address this, a Schmitt trigger inverter–based debounce circuit was implemented as seen in Figure 13. This approach provides clean, stable transitions and minimizes the risk of false triggering due to contact bounce. When designing the RC network for the debounce circuit, it was important to select resistor and capacitor values that provided an appropriate time constant for the button's mechanical characteristics. An RC time constant of approximately 10 ms was chosen, as this duration effectively filters out the transient bounces without introducing noticeable input lag for the user. Determining the RC time constant was straightforward, since the relationship $\tau = RC$ allows the use of common component values such as a 1 µF capacitor and a 10 kΩ resistor to achieve the desired delay.

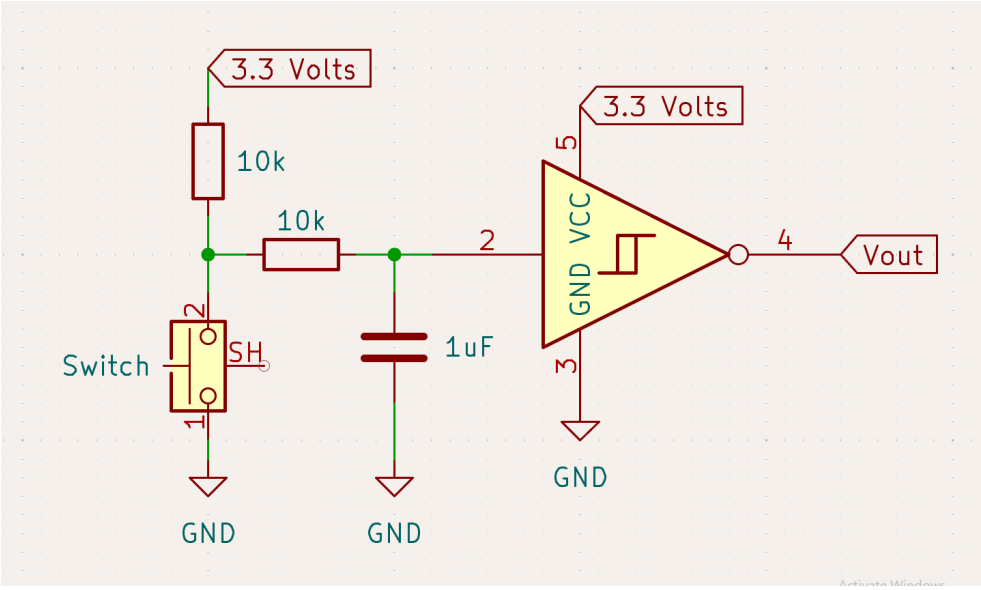

Figure 13: Debounce Circuit

The ECG subsystem is expected to output a voltage in the 1–2 V range. However, in the event of an unexpected voltage surge, it is critical to prevent the analog input from exceeding the ESP32's maximum rated voltage of 3.3 V. To achieve this, a Zener diode clamp circuit was incorporated, as shown in Figure 14. The Zener diode limits the voltage to approximately 3.3 V, protecting the microcontroller's GPIO. Additionally, a 700 Ω series resistor is included to limit the input current to approximately 2.5 mA in the event of a 5 V surge, ensuring the circuit remains within the operating limits of the ESP32 IC.

The resistor value was determined using Ohm's law:

$$R = \frac{V_{IN} - V_Z}{I} = \frac{5V - 3.3V}{2.5mA} \approx 700\Omega$$

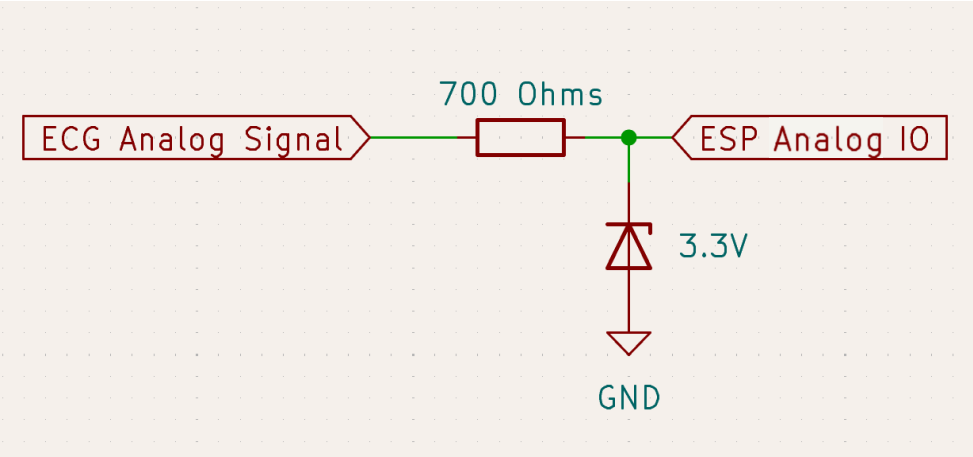

Figure 14: Overvoltage protection circuit

The control subsystem firmware follows a structured event-driven flow, as illustrated in Figure 15. Upon power-up, the system initializes the ECG, impedance, LCD display, and start/stop button interfaces. After initialization, the firmware enters a waiting state, monitoring the start button. If the power switch is turned off, the program will stop as the microcontroller no longer has power. When the start button is pressed, a 1-second peripheral timer is configured to generate periodic interrupts. These interrupts drive the sampling and processing loop. During each timer interval, the firmware performs an analog read of the ECG signal and stores the data in memory. Once the timer interrupt occurs, the collected ECG data is processed to calculate the heart rate, the impedance is read from the external IC, and the results are displayed on the LCD. This cycle

repeats continuously until the stop button is pressed, at which point the data is saved to USB storage before returning to the waiting state.

The firmware will be developed using the ESP-IDF, the official development framework for the ESP32. It provides direct access to hardware features for timing, interrupt handling, and control of the ECG, impedance, and display subsystems.

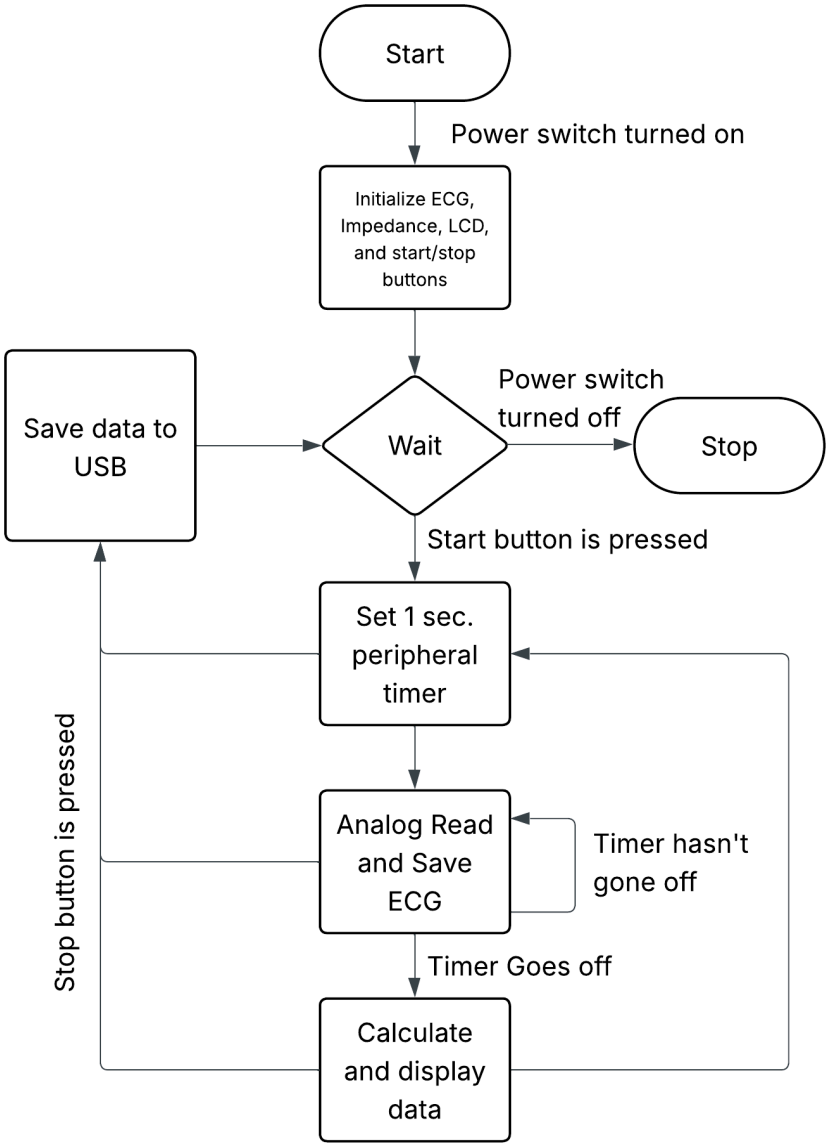

Figure 15: Embedded Code Flowchart

Requirements	Verifications
The start/stop push button must produce clean digital signals without multiple unintended triggers (debouncing).	• Connect an oscilloscope to the button's output after the Schmitt trigger debounce circuit. • Press and release the button multiple times and verify single clean transitions without bounce artifacts.
The analog input from the ECG subsystem must not exceed 3.3 V to protect the ESP32 GPIO.	• Apply controlled overvoltage (e.g., 5 V) to the ECG input and measure voltage after the Zener clamp using an oscilloscope. • Confirm voltage is limited to ~3.3 V. • Verify current through the series resistor remains below 2.5 mA.
Must accurately display heart rate, impedance, and ECG waveform on the LCD at a refresh rate of 1 Hz.	• Connect known test signals to ECG and impedance inputs. • Observe LCD visually to verify data refresh at 1 Hz. • Cross-check displayed values against known inputs.
The control subsystem firmware must follow the event-driven flow shown in Figure X, ensuring proper initialization, waiting for user input, periodic sampling and processing, data display, and safe shutdown when the stop button is pressed.	• Review the source code to confirm the structure matches the flowchart • Observe system behavior in real time: the system should remain idle until the start button is pressed, then enter the sampling loop, update the display periodically, and return to the waiting state after the stop is pressed.

2.3.4 Power Subsystem:

The power subsystem, illustrated below in Figure 16, is responsible for supplying power to each component within our device. We chose to work with a 9-Volt battery as that is the industry standard. We utilize two converters to step down the voltage to 3.3V and 5V for the microcontroller and Amplifier ICs. In our block diagram, we have 15V noted. This is only in our first iteration PCB; these IC's have been replaced with ones that require 5V instead. To achieve our desired output voltages, we use an MP2338 DC-DC Converter IC. This allows us to both step down or step up our voltage source to our desired output voltage. This allows the system to work even when there is a slight voltage drop as the battery naturally discharges. We chose this power converter as it is 90% Efficient when in the range of our current output. The ratio between the two is what determines the output voltage. The capacitors we will choose for this unit will be ceramic for the best performance. The power inductors we will use are the MPL-AL6060-6R8 6.8µH. The resistor and indicator values were chosen based on this equation from the datasheet.

$$R1 = \frac{V_{OUT} - V_{REF}}{V_{REF}} \times R2 \qquad L = \frac{V_{OUT}}{f_{sw} \times \Delta I_L} \times \left(1 - \frac{V_{OUT}}{V_{IN}}\right)$$

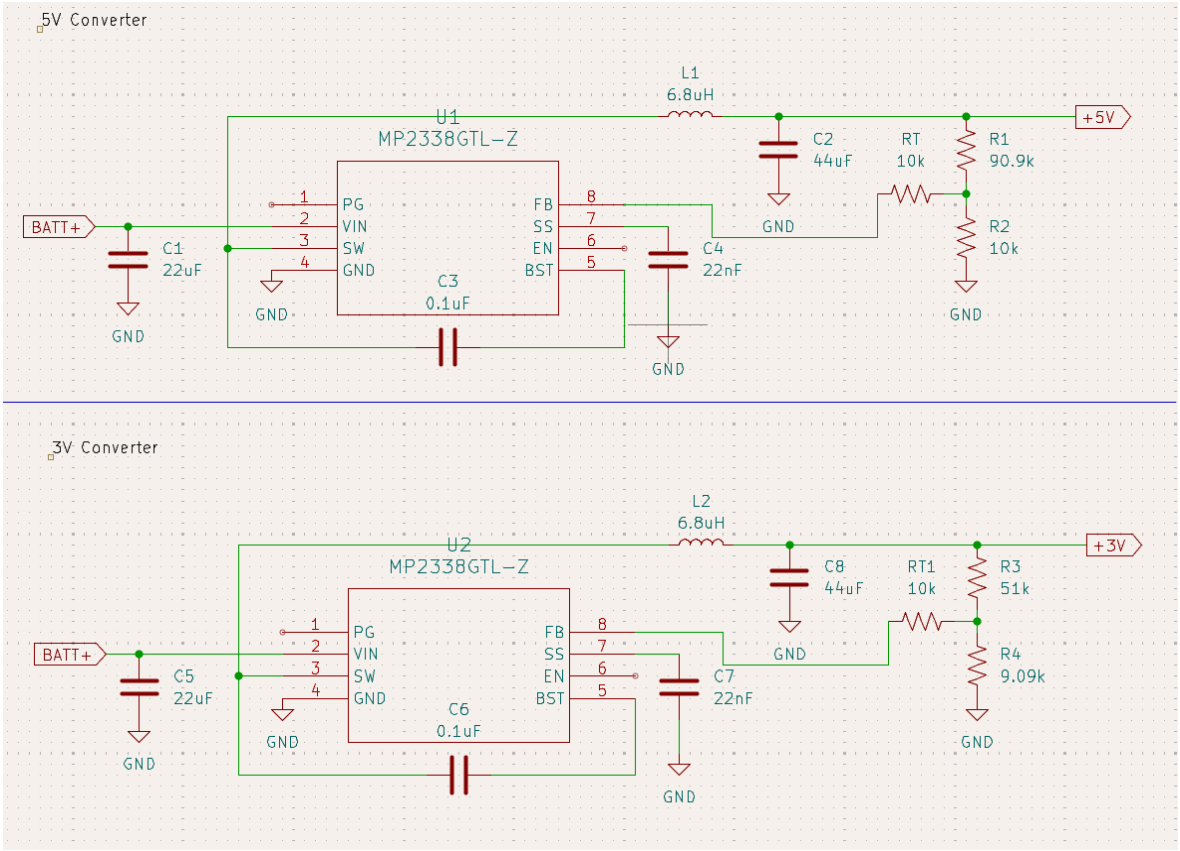

Figure 16: Power Subunit Schematic

Requirement	Verifications
Supply power to all subsystems at rated currents and voltages.	The supply power should successfully step down the 9V battery voltage to 3.3V and 6V, respectively.
At maximum power draw, supply voltages should not drop.	The power supply for 3.3V should never diverge more than 0.2V when running under maximum load, meaning all other subunits are being powered. The 5V power supply should not diverge more than 0.5V.

2.4 Tolerance Analysis

Our calculations did not include the tolerances on the components that were chosen. However, most components were within 1% of their true value. The impedance and noise considerations were also not accounted for within the op amps. This could contrast with our expected results. The AD5933 also has certain impedance tolerances, along with output current tolerances that will be accounted for once testing of the chip begins.

If we don't take into account the worst-case scenarios for our power draw, we could end up pulling more current from our battery, LDO, or buck/boost converter, leading to damaged parts. It is our responsibility to make sure that our current draw, specifically for the impedance module, is within the allowable amplitude and frequency in compliance with the American National Standard for Safe Current Limits for Electromedical Apparatus.

3 Cost and Schedule

3.1 Cost Analysis

The total cost for parts, as shown below in Figure X, after shipping is $55.00. Assuming we order a second and third PCB board, our total cost for electronics would be around $100, considering most elements are reusable between iterations. We can estimate that each of us will work on this project about 60 hours during the course of this semester, working at $40 an hour. ($40/hour) x 2.5 x 60* 3 This would yield a total of 18000$ in labor cost. This yields a total cost estimate of 18100$

Value	Reference	Manufacture Part Number	Qty	Price	Link
2 Mohm Resistor	Rprotect	CPF0805B2M0E	1	$0.17	link
16nF Cap	C_bias	ECW-F6163HL	2	$1.56	link
0.56uF Cap	C1,C2,C3,C6	ECW-FD2W564J4	4	$0.66	link
0.067uF Cap	C4	ECH-U1C683GX5	1	$0.80	link
0.01uF Cap	C5	ECH-U1C103GX5	1	$0.52	link
Audio Jack	J1,J2	54-00174	2	$0.62	link
2kohm Resistor	R1,R4,R13,R14	ERA-3ARW202V	4	$0.40	link

1kohm Resistor	R2,R3,R5,R19,R20	ERA-3AEB102V	5	$0.10	link
100kohm Resistor	R6,R7,R17,R18	RC0402FR-07100KL	2	$0.10	link
150ohm Resistor	R8,R10	RNMF14FTC150R	2	$0.09	link
135kohm Resistor	R9	RT0603BRD07135KL	1	$0.13	link
12.1kohm Resistor	R11	TNPW040212K1BEED	1	$0.21	link
137kohm Resistor	R12	ERA-8AEB1373V	1	$0.22	link
500ohm Resistor	R15,R16	PLT0603Z5000AST5	2	$0.10	link
200kohm Pot	Rcurrent	TC33X-2-204E	1	$0.23	link
50kohm pot	Rfb	TC33X-2-503E	1	$0.23	link
10kohm pot	Rin	TC33X-2-103E	1	$0.23	link
OpAmp	Opamps	TL051CP	9	$0.46	link
AD5933	Impedance IC	AD5933	1	$20.41	link
INA333AIDGKR	Instrumentation AMP	INA333AIDGKR	1	$0.10	link

Figure X: Cost Table

3.2 Schedule

Week	Task	Person
	Order PCB 1	Brian
	Order Parts	Ethan
October 5-12	Research on Microcontroller and LCD	Will
	Assembling the initial PCB	Will
	Retuning Notch and lowpass filter designs	Brian
	Design Power Unit step down to 3.3 and 5V	Ethan
	Testing our PCB 1	Everyone
	Layout for PCB 2	Brian
October 13- 19	**Order PCB 2**	Everyone
	Prepare Breadboard demo	Everyone
	Coding the Microcontroller	Everyone
October 20 - 26	Testing on PCB2	Everyone
	Breadboard demo 2	Everyone
	PCB 3 design	Everyone
October 27- November 2	R&D bonus subsystem (Shock)	Everyone
	PCB 3 Final Order	Everyone

November 3- 9

	Iterate on PCB 3	Everyone
November 10 - 17	PCB 4 Order	Everyone
November 18 - 23	Start preparing for the mock Demo	Everyone
November 24	Enjoy Fall break and prepare for the Final Demo	Everyone
December 1st	Final Demo and presentation	Everyone

4 Ethics and Safety

Potential Safety Concerns

The primary safety concern is injecting current for impedance measurement. Secondary concerns include electrical hazards from the battery and DC–DC circuitry, as well as potential software errors in the microcontroller module. We mitigate these risks with careful circuit design, isolation, overcurrent protection, and thorough verification of software functionality before testing.

Ethical and Safety Issues

The primary safety concern in our project is the measurement of patient impedance, which requires injecting small currents into the body at different frequencies. While these currents are extremely low, we recognize the potential risk and are working with medical professionals to ensure they remain within clinically safe ranges. Potential misuses of our project include exceeding safe current limits during impedance measurements and assuming that the prototype can function as a fully operational AED capable of delivering shocks.

Safety and Regulatory Standards

Our project will comply with all relevant safety and regulatory standards throughout the design and development process. Because the planned design will involve passing a small electric current through human subjects to measure impedance, we will adhere to the American National Standard Safe Current Limits for Electromedical Apparatus, which specifies both the maximum permissible current levels to ensure human safety and the standardized testing procedures used to verify compliance. We will also follow the ECE 445 course battery safety standard, as the final device is intended to be portable and battery-powered. All team members have completed the university's laboratory safety training, and we will strictly observe the requirement that a minimum of two people be present in the laboratory at all times during testing or design activities.

Ethical Standards

The IEEE Code of Ethics requires prioritizing public safety, health, and welfare, and promptly disclosing any factors that may pose risks. Our team will act lawfully, avoid conflicts of interest, reject bribery, provide and accept honest feedback, correct errors, make accurate claims, and give proper credit to others. We will maintain technical competence and only undertake tasks for which we are qualified or disclose any limitations.

Avoiding Ethical Breaches

To avoid ethical breaches, we will prioritize safety and honesty. We will only use clinically safe currents, clearly state that the device is a prototype and not a real AED, and thoroughly test all hardware and software before use. We will follow state and federal regulations, industry standards, and campus policies, and keep clear documentation of all procedures.

4.1 Risk Analysis

The primary safety concern for this project arises from the requirement to inject a small electrical current into the human body to measure impedance. Although the current levels involved are extremely low, any device that interfaces electrically with a person must be designed and tested to ensure that it poses no risk of harm under normal operation or fault conditions.

Before performing any human testing, the impedance module will first be validated using electronic components with known impedances, such as capacitors and inductors. These components will be connected in place of a human subject, and the module's impedance measurements will be compared against values obtained using a calibrated ohmmeter. This will ensure that our measured impedance values are within our tolerances and that the injected current is within safe levels for human testing.

When moving to human testing, the module will follow the American National Standard for Safe Current Limits for Electromedical Apparatus. This standard defines the maximum allowable electrical currents that can be safely applied to or flow through a human body when using medical or bioinstrumentation equipment. In compliance with this standard, all electrode connection points will be clearly labeled, and the current injection circuitry will be fully enclosed

and galvanically isolated from other electronics. Series current-limiting resistors will ensure that injected current remains below 100 µA r.m.s. under normal conditions and 500 µA r.m.s. under single-fault conditions. Properly shielded electrode cables will be used to minimize stray currents, and the injected signal frequency and amplitude are chosen to remain within the standard’s frequency-dependent safe current limits.

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