# Bioinstrumentation Analysis: A Closer Look at the Automated External Defibrillator

Emily Beckman[1], Benjamin Bestmann[1], Everret Gallardo[1], Dalila Gonzalez Mejia[1], and Max Pendleton[1]

***Abstract* -- Sudden cardiac arrest, an electrical misfiring in the heart, is a leading cause of death in the United States. It can be treated effectively with an automated external defibrillator (AED). These devices are primarily used in emergency situations in non hospital settings. AEDs automatically read a patient's heart rhythm and resistance, and administer a strong current to the heart, or shock, effectively reinstating a normal heartbeat. In this paper, we designed and analyzed an AED circuit that detects electrical malfunctions via an electrocardiogram (ECG) and administers a shock to restart the myocardial pacemaker cells. We divided the circuit into 2 sections: an ECG, which measures the patient's heart rhythms, and a defibrillator, which administers a shock. These sections are connected by a black box which stands in for the computer which calculates whether a shock is necessary. Additionally, by implementing a biofeedback sensor, we successfully ensured that the shock is proportional to the patient's bodily impedance. The pseudocode for the Black Box is also included. Finally, we created a circuit to defibrillate the patient using a transformer. This design is monophasic and will successfully revive a patient; however, a biphasic design could result in better outcomes and decrease the chances of burning the patient.**

## I. INTRODUCTION

Sudden cardiac arrest (SCA) is a medical condition where the heart undergoes an electrical malfunction that can lead to fainting, lack of blood circulation, and even death. This is different from a heart attack, where blood flow is obstructed due to a blockage in the arteries. SCA is a leading cause of death in the United States, with about 320,000 out-of-hospital SCAs per year [1]. This statistic is particularly poignant because SCA is easily treatable with access to an AED. This bioinstrument's function is to deliver a shock to a person's chest in order to restart the heart's pacemaker cells, restoring normal electrical activity [2]. This paper will dive into our design and analysis of an AED circuit.

The first functional requirement of an AED is recognizing what a myocardial electrical malfunction looks like on an ECG. **Figure 1, part A** shows what a healthy ECG signal should look like. Directly below are the ECGs of two common heartbeat irregularities, or arrhythmias: ventricular fibrillation, an irregular and fast heartbeat; and ventricular tachycardia, a dangerously fast heartbeat [3].

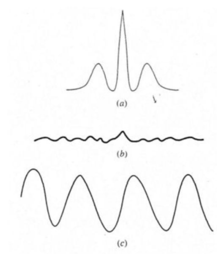

Fig. 1: The three ECG waves displated represent a) a healthy heart rhythm, b) ventricular fibrillation, and c) ventricular tachycardia [3].

After recognizing the heart's arrhythmia, the AED must be able to deliver up to approximately 1,400 Volts through the ECG leads to restart the heart [4]. The exact value of the Vout depends on the patient's bodily impedance. This consists of a variety of factors, including but not limited to: sweat, body fat percentage, age, and size. Finally, the AED should prompt the user to deliver a shock to the patient. This design uses an indicative LED light to signal the user that the AED is ready and the patient should be shocked.

In order to control for some of the variables that affect AED circuit execution, these assumptions were made:

1. The patient has no body hair, since AED kits include a razor.
2. Room temperature is roughly 25℃, and pressure is ~1atm.
3. The batteries being used are new, and deliver the nominal voltage.

Upon determining the broad functional requirements, the AED was dissected into 4 main stages. These four stages, which will be elaborated on in our methods, include: an instrumentational amplifier (IA), a low pass filter, a black box portion with the complimentary software, and a defibrillator.

All of the circuit values and decisions were made using our experience and knowledge from Dr. Gert Cauwenberghs' BENG 186B bioinstrumentation course. This includes, but is not limited to lecture 12, "Biopotential Amplifiers: Problems and Solutions," and homework 4, where which involved designing a 6-lead frontal ECG recording system [5]. In addition, we consulted some existing AED technical specifications and manuals, to better understand what was already out on the market [6][7].

## II. METHODS

### *A. Circuit Topology*

The initial circuit topology is based on the ECG biopotential signals via Einthoven's Triangle provided by Dr. Cauwenberghs. Additional stages are added for the AED to defibrillate the patient with the same electrodes used for the ECG measurement. The circuit topology is divided up into four stages. The first is an instrumentation amplifier for measurement of the ECG signal and the second stage, a low pass filter to remove high-frequency noise. The third stage, a computer to determine the impedance of the patient's abdomen, which can vary because of skin dryness, age, body mass index, and breast tissue. The fourth stage, a transformer with capacitor to accumulate charge to then switch to applying this built up voltage across the patient's abdomen. The circuit topology of the ECG and the defibrillator can be examined in figures one and two respectively.

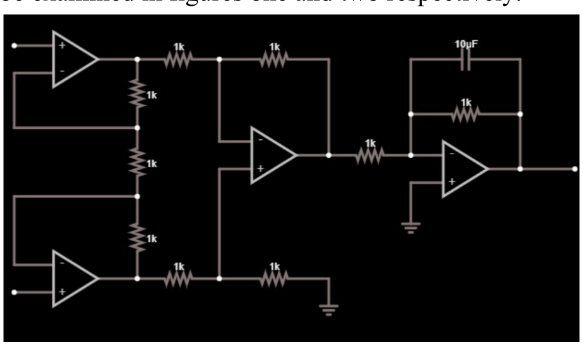


Fig. 2. Circuit Diagram of ECG Stage

This ECG leads into a black box. In other words, this box reads ECG data and determines whether or not to shock the patient based on the inputted heart rate data. This box then inputs a small voltage into the patient to determine the necessary voltage to shock said patient. This process will be further examined in the Results Section. This circuit configuration can be seen in Figure 2. The pseudocode used by the black box can be seen in entry 1 of the appendix. The defibrillation stage doubles as an impedance measurement stage, with a diode RC circuit connected to a switch which can switch to, and subsequently discharge current, across the 1k resistor (pictured) which is really the patient's abdominal impedance. The secondary loop is used to determine how high this resistance is, as it varies among different patients needing defibrillation. It is important to note the two switches are always in the opposite position of the other, that is to not destroy the ammeter used in the DC voltage impedance-measurement device when a large enough voltage is applied.

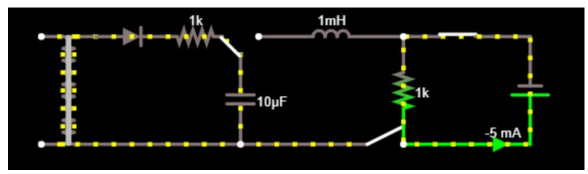


Fig. 3: Circuit Diagram of Defibrillator, set to read the resistance of a patient's chest by applying a small DC voltage.

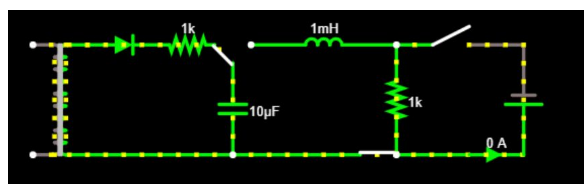


Fig. 4. Circuit Diagram of Defibrillator or an AED-set to input a voltage into the patient's chest

A full diagram of the circuit can be seen in entry 2 of the appendix.

## III. RESULTS

From our virtual circuit model, we were able to create a transfer function for our design instrumentation amplifier and low pass filter; these served as the basis of our calculations of the output current with a known voltage.

The characterization of the circuitry in this AED is done with voltage-in, voltage-out transfer functions, in the generalized form $H(j\omega) = \frac{V_{out}(j\omega)}{V_{in}(j\omega)}$. For the instrumentation amplifier, a transfer function was derived by a multiplication of the differential stage and the buffer stage resulting in the function independent of frequency [8]. Here, H is rearranged to still be in units of volts, not unitless.

$H(j\omega) = (1 + \frac{2R_2}{R_1})(\frac{R_4}{R_3})(V_2 - V_1)$

With $R_1 = 4 * 10^3 \Omega$ ,$R_2 = 18 * 10^3 \Omega$, $R_3 = 10 * 10^3 \Omega$, $R_4 = 100 * 10^3 \Omega$, $V_2$ and $V_1$ being the input voltages at the non-inverting and inverting nodes, respectively. For the low-pass filter of the ECG, a standard Zout - Zin relationship was used with

Zin =1 and Zout = 1/jωC | | R

With $C = 10 * 10^{-9}\ F$ , $R = 100 * 10^3 \Omega$

With $H(j\omega) = -\frac{V_{out}(j\omega)}{V_{in}(j\omega)}$, We are able to get a TF of $H(j\omega) = \frac{-R}{j\omega(0.001)+1}$

And with a cutoff frequency of $\omega_c = 1/(RC) = 1000\ rad/s$

and $f_c = 1/(2\pi RC) = 159.15\ Hz$

With these two pieces we were able to characterize the circuit and how it relates to the voltage being sent into the black box. Using this analysis we were able to isolate a relationship between source voltage (the small zap) and the patient's natural abdominal impedance:

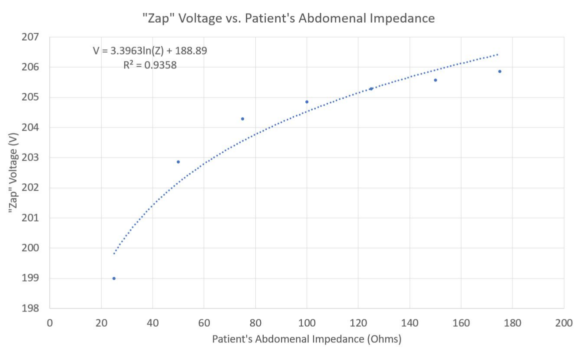


Fig. 5. Defibrillation Voltage as a Function of Abdominal Impedance

Here we see a logarithmic (ln(x)) relationship between the voltage and the abdominal resistance with an $R^2$ confidence value of 0.936. This curve suggests that the voltage increases quickly with a small increase in impedance but levels out afterwards with this figure suggesting that it approaches a plateau of ~207 - 210 Volts.

Now if we analyze the relationship between voltage and time, we look at the Bode plot derived from the circuit:

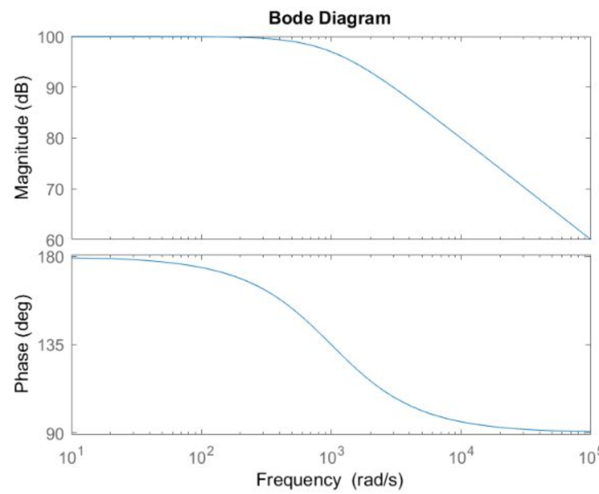


Fig. 6. Bode Plot of the ECG

Here we see a negative change of 20 db/Decade from $\sim 10^3 - 10^5 rad/s$.

This is important for informing the black box about rhymes in the heart. Without identifying the cutoff frequencies, we would be unable to read the heart accurately which would result in false positives for shocking causing further complications with the patient and a failure of a system.

The shocking stage mostly depends on the black box input to determine the shock type. This will be controlled by having the black box output a voltage to the transformer that will bring the shocker (i.e., across the capacitor) up to the desired voltage in 10 seconds. Since the applied voltage will be DC, there will be no oscillation of current, resulting in a monophasic shock.

## IV. CONCLUSIONS

The main purpose of this study was to determine the different ways in which bioinstrumentation can be applied to a real-life setting. We found that bioinstrumentation itself was somewhat limiting in analysing the functionality of an AED; thus, we examined the intersection of bioinstrumentation and bioinformatics.e Our design is able to charge and appropriate voltage across a capacitor and discharge fully within 10 seconds. This was achieved by having the appropriate RC time constant in the transformer stage of the circuit and an appropriate capacitance chosen with cost efficiency in mind. Because resistors are very inexpensive, many can be used without detriment to the overall cost of the device. With this in mind, the RC time constant was a function of the capacitance rather than resistance. Thus, our total analysis of the problem led us to conclude that this design is optimal in meeting all the functional requirements.

### *A. Model Advantages and Limitations*

The model proved itself to have interesting utility and advantages given the nature of the circuit topology. In the first stage, the ECG does not perform loading. All of the biopotentials do not get shorted at all through any external circuitry attached to the body. There is infinite input impedance to the first stage of the AED. Additionally, there is no loading of the low-pass filter.This was performed using topology analogous, but not identical to, Sallen-Key Topology to attenuate any amplifier ECG signals with a frequency higher than 1000 Hz. The black box allowed us to focus more on the circuitry of the design rather than the software. The defibrillation stage's advantages are grounded in its simplicity - it is simply a voltage transformer connected to an RC circuit. When appropriate (i.e. when the user presses the "shock" button) the accumulated voltage across the capacitor will discharge across the low resistance of the patient's abdomen, inducing a current large enough to defibrillate the heart back to a healthy ECG waveform.

There were a few areas of further research we identified, including algorithm improvements and bidirectional current flow. Although we included a pseudocode that controls whether or not a shock is administered it is important to acknowledge some shortcomings that exist in existing AEDs. Additionally, changing our design from monophasic to biphasic would reduce the probability of burning the patient, and decrease the time to charge [9].

The fields of computer science and bioinformatics are new and emerging, showing great promise and potential for advancing the medical field; however, we know that there is no one size fits all answer when it comes to human physiological patterns and thresholds. For example, a fever may go undiagnosed in person with a lower resting body temperature. Similarly, someone who runs marathons may be prescribed medication for their "dangerously" low resting heart rate, when in reality they do not need it. While we cannot account for every single variation in irregular heartbeat patterns, there is a certain amount of due diligence and research needed in tailoring our algorithm for different groups of people. This has the potential to improve health outcomes, and save lives.

Our design was monophasic, i.e., it was based on the monophasic waveform, where current flows in one direction from one electrode to another, stopping the heart momentarily, and allowing the basic sinus rhythm to be restored by a shock. In the future, our AED design would benefit from a biphasic design that sends current in two directions in phases; the first moves the current from one paddle to another like a monophasic while the second phase sees a current flow in the reverse direction.

One final way we can improve our design is by optimizing and experimenting with different ways to change the time to charge our shock delivering capacitor. For example, physically changing the distance between the capacitive plates or switching out our resistor for a potentiometer. In the future, it would be very interesting to take a look at all the different points of improvement within our design and experiment with them to find a combination that is cost effective and also delivers a high quality bioinstrument. If there is anything this course has taught us, it is that there are many different ways to accomplish the same circuit outcome. We look forward to optimizing this circuit design in the future, and learning more about the intersection between bioengineering and circuits and its applications to medical advancement.

## APPENDIX

$R_1$, $R_2$, $R_3$, $R_4$ are the resistances of the instrumentation amplifier. R is the negative-feedback resistance of the low-pass filter, with C being the capacitance. $H_1$ is the transfer function of the instrumentation amplifier, with H being the transfer function of the low pass filter, and the product of the two.

## ACKNOWLEDGEMENTS

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# Bioinstrumentation Analysis: A closer look at the Automated External Defibrillator: Supplemental Material

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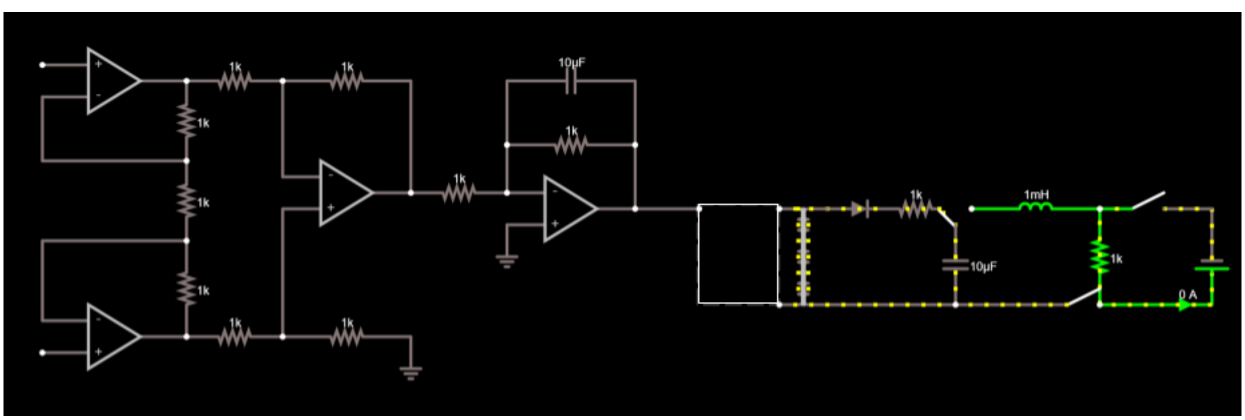
**1. Black Box Pseudocode:**

1 If only one signal then continue
2 Perform spectral analysis
3. Bandpass dominant, calculate period
4. If over 270bpm shock=true
5. Compare to database
6. If spectral analysis is within 5% of that of any ventricular fibrillation cont to 7, if not cont. to 8
7. If average amp>100microvolts shock=true
8. If spectral analysis is within 5% of that of any ventricular tachycardia cont
9. If adult and bpm>150 or child and bpm>200, shock=true
10. If no clear secondary frequency from the spectral analysis (if the waveform is exceedingly irregular)
11. If Shock = True, input small voltage
12. Measure current output
13. Calculate resistance V_Impedance=IR (where R is transfer function)
14. Calculate necessary voltage output Vout=3.963ln(R)+188.89
15. Turn on indicator light for user to shock the patient

**2. Circuit, All Four Stages (unlabeled "black box" flanked by low-pass filter and defibrillator).**



**3. The impedances/goals of adults vs children [10].**

| | | | Pediatric | | | AHA goal |
|---|---|---|---|---|---|---|
| **Rhythms** | **Adult** | **<1y** | **1y-8y** | **>8y** | **Records** | **Sens/Esp** |
| **Schockable** | | | | | | |
| **Coarse VF** | **374 (374)** | **3 (1)** | **18 (11)** | **37 (10)** | **200** | **90%** |
| **Rapid VT** | **200 (200)** | **8 (4)** | **39 (19)** | **19 (13)** | **50** | **75%** |
| **Non-shockable** | | | | | | |
| **NSR** | **292 (292)** | **14 (13)** | **312 (280)** | **214 (161)** | **100** | **99%** |
| **SVT** | **89 (89)** | **38 (29)** | **147 (103)** | **137 (104)** | **30** | **95%** |
| **Total** | **955 (820)** | **63 (39)** | **516 (357)** | **407 (216)** | **-** | **-** |