Electrocardiograms to Detect Atrial Fibrillation

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Abstract - Heart arrhythmia is a condition where the heart beats too fast, too slow, or irregularly, disrupting the normal rhythm of the heart. Electrocardiogram (ECG) is a diagnostic tool that measures the electrical activity of the heart and can identify various types of arrhythmias. ECG is a non-invasive test that involves placing electrodes on the chest, arms, and legs to detect the electrical impulses generated by the heart. The test can be performed in a doctor's office or hospital and is usually painless and quick. Our team focused on atrial fibrillation (Afib), which is a type of heart arrhythmia that occurs when the electrical signals in the heart's upper chambers (atria) become chaotic and disorganized. Instead of contracting normally, the atria quiver or flutter, which can lead to an irregular heartbeat. AFib is a common heart condition that affects millions of people worldwide, especially those over the age of 60. Some of the risk factors for AFib include high blood pressure, diabetes, obesity, heart disease, and a family history of the condition. We designed a 3 frontal lead electrocardiogram circuit with its primary function being to notify the user of AFib detection. The design consists of an instrumentation amplifier followed by a cascade of second order low-pass and high-pass filters forming a bandpass filter that is followed by a hysteretic comparator that signals whether an LED should or shouldn't light up in the case of Afib detection.

I. Introduction

Atrial fibrillation (AFib) is a common heart condition that affects millions of people worldwide. It occurs when the electrical signals in the heart's upper chambers (atria) become chaotic and disorganized, causing an irregular heartbeat. AFib can cause symptoms such as heart palpitations, shortness of breath, fatigue, dizziness, chest pain, and fainting. It can also lead to serious health complications such as stroke, heart failure, and other heart-related diseases. According to the Centers for Disease Control and Prevention (CDC), AFib affects an estimated 2.7-6.1 million people in the United States, and this number is expected to increase as the population ages [1]. AFib is more common in people over the age of 60, and the risk of developing the condition increases with age. Other risk factors for AFib include high blood pressure, diabetes, obesity, heart disease, and a family history of the condition.

The purpose of this bioinstrument is to notify the user of an atrial fibrillation detection, by utilizing a series of circuit components and using a comparator to signal an attached LED to light up if the condition is present. The goal is to successfully noninvasively detect Afib accurately using the filter as well as selected resistor values for the comparator. Before addressing the methods used to develop our circuit solution, it's important to understand the difference between a healthy heart reading from an electrocardiogram, versus a signal from a heart with AFib. Figure 1 depicts this difference visually [2].

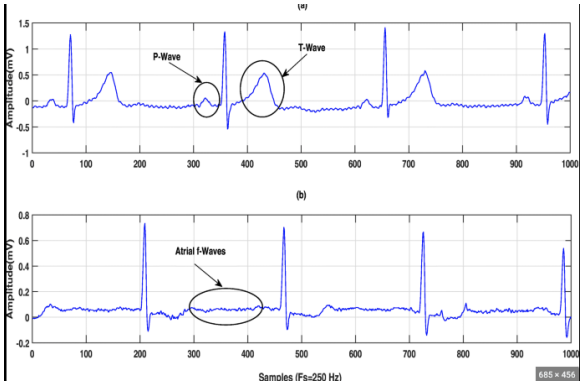

Figure 1: Demonstrates the difference between a normal ECG and the ECG of a patient with Afib.

II. Methods

A. Background

Throughout modern history one of the most effective devices for detection of heart arrhythmia has been an electrocardiogram(ECG). However, ECG struggles with being unable to distinguish between the different types of arrhythmia such as bradycardia and tachycardia, as well as even from a normal heartbeat when detecting atrial

fibrillation due to its irregular pattern. Afib is hard to diagnose as it is periodic and without a consistent pattern or wave form. As part of this project it can be chosen to optimize the accuracy of ECG readings for atrial fibrillation, which has been executed by designing a circuit composed of a band pass filter and specified resistor values for the comparator and using an LED as the indicator. An ECG records the heart's electrical activity by reading the depolarization current that travels through the heart muscle[3]. This is done via electrodes that attach to a patient's skin that are placed along the path this current travels. When a depolarization current travels towards an electrode, it is read as a positive deflection, and when the current travels away from the electrode, it is read as a negative deflection [3]. These readings create a graph that records voltage over time. This graph is known as the atrial depolarization loop which will have different properties depending on the status of the patient's heart. Conventional ECG's are 12 lead and are used to diagnose various heart abnormalities [3] however we are modifying the design in order to screen and detect Afib. The general outline for our device consists of a lead connected to an instrumentation amplifier, then to a band-pass filter, and finally to a comparator and LED. We are under the assumption that all op amps are ideal.

B. Components of Circuit Model

1. Measurements and Instrumentation Amplifier

Our ECG is powered by 3V batteries for all op-amps and reads the depolarization current via 3 leads that connect to the right arm, left arm, and left leg. There is a 4th connection which is on the right leg that leads to a driven right leg circuit [add picture]. The purpose of the driven right leg circuit is to reduce the interference that would lead to less accurate measurements from the leads connected to the arms and left leg[4]. The resistance values chosen reflect the constraint of there needing to be a -500 gain from the driven right leg and the fact that the leads can only draw 10μA.

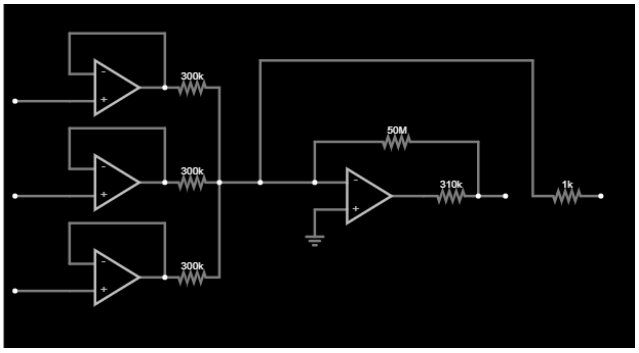

Figure 2: Right Arm, Left Arm, and Left Leg Leads with Driven Right Leg Circuit.

The leads are connected to an instrumentation amplifier which has the purpose of providing a high amount of gain in order to reduce the effect of high noise levels[5]. We need this because of how relatively low the signal we are looking for is. The chosen resistances on the instrumentation amplifier reflect the desire to have a differential gain of 2000.

$$CMRR = \frac{A_d}{A_{cm}}. \quad \text{Eq.1}$$

Because we are under the assumption that our operational amplifiers are ideal, we can say that our common mode gain is 0 which means that our common mode rejection ratio (CMRR) is infinity.

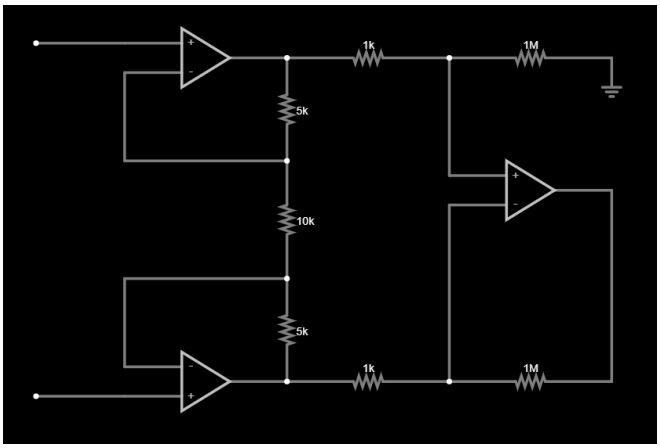

Figure 3: Instrumentation Amplifier with chosen resistor values for a gain of 2000.

2. Bandpass Filter

As mentioned before, following the output of our instrumentation amplifier, we have utilized a bandpass filtering stage made up of a 2nd-order, Sallen-Key low-pass filter that is then cascaded by a 2nd-order, Sallen-Key high-pass filter. Again, the use of the bandpass filter is to filter out frequencies that are outside of the range of atrial

fibrillation that we are looking for in our circuit design. The filters are shown in figures 4 and 5 below

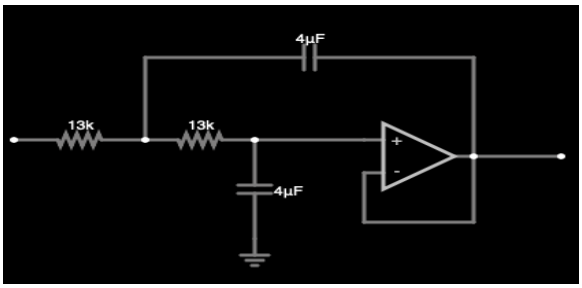

Figure 4: 2nd-order, Sallen-Key low-pass filter with calculated resistor and capacitor values for 2.92 Hz cut-off.

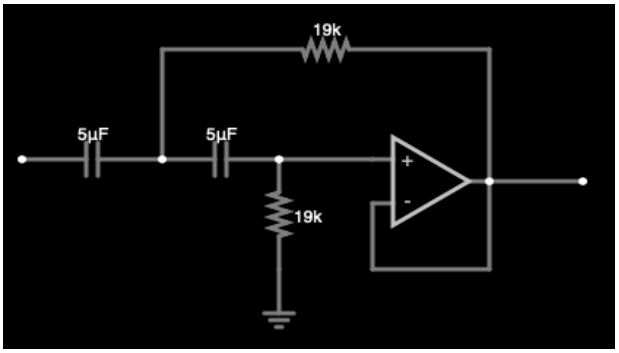

Figure 5: 2nd-order, Sallen-Key high-pass filter with calculated resistor and capacitor values for 1.67 Hz cut-off.

In order to decide the proper resistance and capacitance values for the two filters, a suggested heartbeat range for atrial fibrillation was used to determine the cut-off frequencies for both the low-pass and high-pass filter. While the typical human heart beats at a rate of 60 to 100 beats per minute (bpm), the range of heart beats associated with atrial fibrillation have been reported to be 100 to 175 bpm [7]. Converting this range into frequency in Hertz (Hz), we find that the range of frequencies that will be applied for our bandpass filter are 1.67-2.92 Hz. Using this frequency range, we apply 1.67 Hz to the high-pass filter and 2.92 Hz to the low-pass filter to find appropriate values for the resistors and capacitors. The frequency cut-off equation for the Sallen-Key high-pass and low-pass filter are the same and is shown below

$$f_c = \frac{1}{\sqrt{2\pi R_a R_b C_a C_b}}. \quad \text{Eq.2}$$

To simplify the calculation of the resistances and capacitances, we can set $R = R_a = R_b$ and $C = C_a = C_b$, which simplifies the equation to

$$f_c = \frac{1}{\sqrt{2\pi RC}}. \quad \text{Eq.3}$$

Using this equation we found the appropriate resistor and capacitor values in order to filter out frequencies not in the range of atrial fibrillation, shown figures 4 and 5.

3. Comparator & LED

Following the filtering stage of the design, the voltage output from the filtering stage now passes through to a hysteretic comparator, which will determine whether the voltage received is one correlating to atrial fibrillation in order to allow voltage to pass to the red LED and signal it to turn on. The comparator does this by comparing the input voltage of the comparator to chosen threshold voltage at the node ($V_{threshold}$) between the two resistors. Based on work in literature, we have decided to choose the voltage threshold to be 0.21 mV [6]. Using this value and the fact that the instrumentation amplifier provides a gain of 2000, resistor values for the comparator were chosen using the equation below

$$V_{threshold} = \frac{R_{feedback}}{R_{grounded}+R_{feedback}} V_s. \quad \text{Eq.4}$$

Using the equation, the feedback resistor needed to be 64.3kΩ and the grounded resistor 10kΩ. The figure below demonstrates the comparator and LED design following the filtering stage.

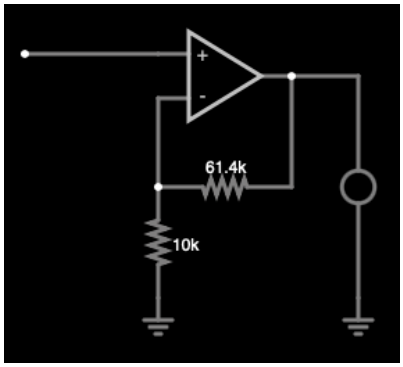

Figure 6: Hysteretic Comparator and LED.

C. Results

In order to check whether the circuit model is functioning correctly, we observed a Falstad simulation of our circuit. To test if our circuit would correctly light up the LED for Afib detection, we simulated the circuit at 1 Hertz frequency from the electrodes which is in the

normal range of heart beats and then we simulated at 2.3 Hertz which is in our specified atrial fibrillation range of frequencies. Figure 7 represents the simulation of our circuit at 1 Hertz frequency and we observed that the LED did not light up, which is what we would have expected from the circuit since that frequency is outside of the Afib range. Then in figure 8, which was our simulation at 2.3 Hertz, the LED did light up, signaling the detection of Afib, which is the result we had hoped for.

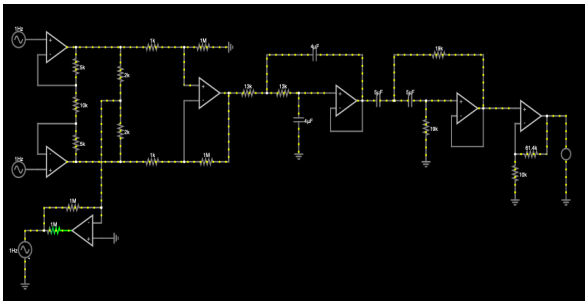

Figure 7: Circuit simulation at 1 Hertz, resulting in the LED to not light up.

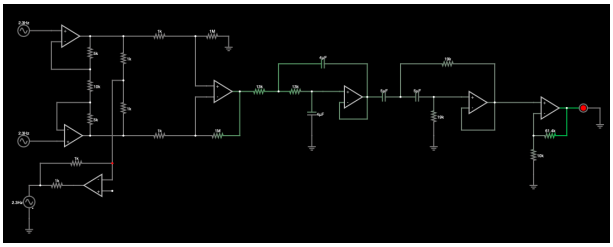

Figure 8: Circuit simulation at 2.3 Hertz, resulting in the LED to light up, signaling Afib detection.

III. Discussion

Utilizing Falstad software in the design of our ECG circuit allowed us to recognize and identify particular limitations, advantages and improvements within the model.

Certain limitations that we identified was specifically that the heart rate range of atrial fibrillation may overlap with other forms of tachycardia: sinus, supraventricular, and ventricular. Other limitations include the usage of active circuits which may lead to inaccuracy issues due to if the correct components or corresponding values are not chosen. Considerations which may also further limit the effectiveness of our system is that our circuit is designed to only detect one form of heart arrhythmia. We also determined that the usage of LEDs may lead to delayed response in the detection of Atrial fibrillation and generally speaking, there is also limited research regarding Afib detection using ECGs to gather and compare data significance.

Given that this ECG circuit model is a prototype design there can be significant improvements made to the design which include, but is not limited to, the identification of tachycardia range and comparing the ranges, which would help us distinguish the specified range for Afib to designate the bandpass filter and resistor values for the comparator in order to resolve the potential issue of overlapping signals in detection. Another improvement that could improve the success of our designed ECG circuit, would be to use different and multiple filters to detect other specified heart arrhythmia values: bradycardia and tachycardia.

Regardless of the limitations discussed, there were significant advantages such as the usage of silver-chloride electrodes onto patient's skin to enhance signal strength of the electrodes by reducing skin resistance. The addition of a band-pass filter allows us to target the ideal frequency range for detecting atrial fibrillation. This method is effective for noninvasive detection methods for detecting Atrial fibrillation accurately using the filter and selected resistor values for the comparator. Lastly, a significant advantage of our design is that it is cost-effective compared to other industry designs by disregarding the use of extra screening components.

ACKNOWLEDGEMENTS

Our group would like to personally thank Dr.Cauwenberghs for his guidance and material content provided to us throughout the quarter. A special thanks to the amazing TAs Samira, Vikrant, and Adyant on their advice in designing and modifying the ECG circuit along with and suggestions for this project and throughout the class

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