Development of Electrocardiograph Amplifier ECE 3813 Final Project

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***Abstract*—Electrocardiography (ECG) is an important practice in modern medicine, as it provides for diagnosis for many heart-related diseases. We developed an electrocardiograph amplifier to measure the electrical activity of the heart over a period of time for the final project of ECE 3813, Microelectronics. The amplification circuit is meant to be connected to the body in three locations and amplify and filter the electrical signal generated from the heart's electrophysiologic pattern of depolarization and repolarization during heartbeats. The circuit amplifies the signal from the body to yield a measurable voltage for electrocardiogram construction and analysis. The device produces an output from about -3 to 9 Volts, enough for simple data acquisition. In order to achieve amplification, the device relies on op-amps for active filtering and gain, including an instrumentation amplifier. The overall device design consists of four stages: instrumentation amplification, high-pass filtering, low-pass filtering, and amplification. Each stage is realized through the use of operational amplifiers and filtering is achieved through the implementation of Butterworth type filters. The device relies on dual 9V batteries to achieve amplification. Both theoretical analysis and experimental analysis were thoroughly investigated for the circuit. Initially, theoretical analysis included the mathematical implementation of the system and determination of values to suit the system's design. The design was then implemented in Multisim for computer simulation to validate the system and tune for more practical components. Finally, the system was attached to a team member which generated a valid electrocardiogram.**

***Index Terms*—Electrocardiograph, ECG, Amplifier.**

I. Introduction

The objective of the project was to design and implement a multistage ECG amplifier that can capture electric signals from the human body and output cardiac activity in the form of an electrocardiograph. The system was be composed of four different stages that first amplified the signal which the heart generates, filtered the desired signal on the frequency range that the cardiac contraction produces, and amplify the remaining signal for visualization on a oscilloscope or other data acquisition methodologies. The method will implemented an instrumentation amplifier, both a low-pass and high-pass filter designed to exclude noise from the signal received, and an non-inverting amplifier to output a signal large enough for measurement. Other equipment utilized included a breadboard, a 9V battery, wires, passive components, and software for analysis.

Final project developed for the University of Oklahoma Microelectronics class, ECE 3813, a part of the Gallogly College of Engineering and the Electrical and Computer Engineering Department

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II. Background

Electrocardiography's (ECG) first appearance was from an experiment recorded by Carlo Matteuci of a frog's cardiac contraction in 1842. This experiment also demonstrated that cardiac contractions are accompanied by electric current. In the following 5 decades, many scientists continued to create instrumentation to precisely record cardiac contraction. The most successful ones, in the early 1870's, were from French physicist Gabriel Lippmann and his invention of a capillary electrometer, used to record electrical current from cardiac contractions. However, the person credited for inventing the ECG is Waller Einthoven. In 1924 he received the Noble Prize for his string-galvanometer 3 leads ECG machine invention of 1901. Although Einthover's invention is not the ECG of today, his 3 leads were sufficient to diagnose arrhythmia. The next ECG advancement was done by Wilson in 1931 which described the unipolar lead concept which later he uses to record myocardial infarction precordial leads, which could not be diagnosed until then. The 12-lead electrocardiogram known of today was the result of many sequential discoveries with the final advancement made in 1942 by Emanuel Goldberger's final refinement of Wilson's original unipolar limb leads[3]. This led to the American Heart Association standardization of the 12-lead ECG's practices in 1951[4].

III. Methods

A. Circuit Design

Since the human cardiac activity produces signals on the order of micro-volts and a source resistance in the order of mega-ohm range, the recording of cardiac contractions first requires a quality amplified to achieve common mode rejection. The amplifier of choice is the analog device INA 126, which is tuned to reach a gain of 10 V/V. However, together with cardiac activity, other electric signals from different sources (i.e. muscle contraction) are also generated by the body and captured by the instrumentation amplifier INA 126, requiring the implementation of filters for attenuation. To exclude the noise from the signal, a high-pass and low-pass filter were implemented on the system to attenuate signals under 0.5 Hz and above 150 Hz. The devices of choice are two second-order Butterworth filters. After the signal is filtered the last optional step is to amplify the remaining signal and generate a frequency response. The overall design stages implementation are shown on Figure 1, which generates a clear signal of the cardiac activity.

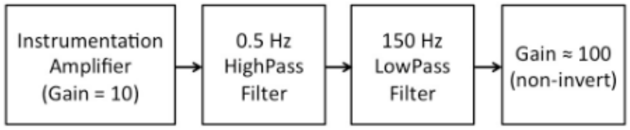

Fig. 1. ECG Amplifier Block Diagram [6]

1) High-Pass Filter: To attenuate signals below 0.5 Hz a second order high-pass filter, Figure 1, of gain $A = 1$ is implemented to the circuit

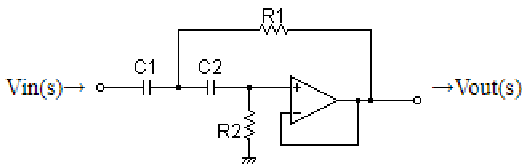

Fig. 2. High-Pass Butterworth Filter Design [5]

From Figure 1, we can derive the transfer function:

$$G(s) = \frac{s^2}{s^2 + s(\frac{1}{R2C1} + \frac{1}{R2C2}) + \frac{1}{R1C1R2C2}} \tag{1}$$

The high-pass cutoff frequency is set to 0.5 Hz, giving the relation:

$$\frac{1}{\sqrt{R1R2C1C2}} = 2\pi * 0.5 \tag{2}$$

Using equation (2), the resistors and capacitors are selected to be $R1 = 22K\Omega$, $R2 = 47K\Omega$, $C1 = C2 = 10\mu F$. After inputting these values in the transfer function, we obtain:

$$G(s) = \frac{s^2}{s^2 + 4.26s + 9.67} \tag{3}$$

The frequency cutoff occurs when the magnitude of the transfer function of equation (3) is at -3.04 dB. After plotting the magnitude and phase of equation (3), we confirm that for -3.04 dB, the frequency is as expected, which is 0.5 Hz, as shown on Figure 2 Bode Plot.

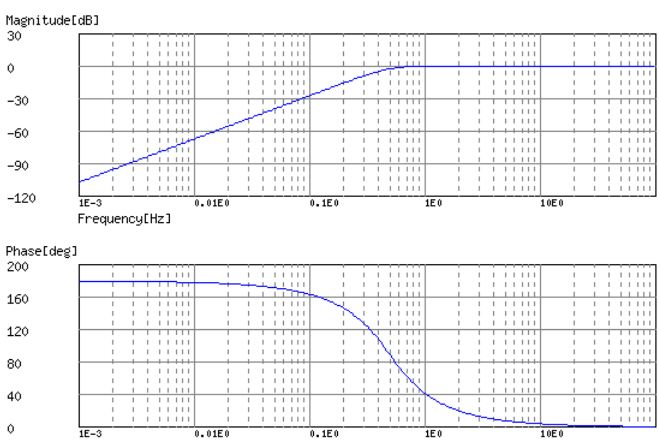

Fig. 3. Bode Plot of High-Pass Filter [5]

2) Low-Pass Filter: Attenuates signals higher than 150 Hz, joined together with the high-pass filter, the signal reading ranges from 0.5 Hz to 150 Hz. The low-pass filter is chosen to be of second order Butterworth type, Figure 3, with a gain of $A = 1$.

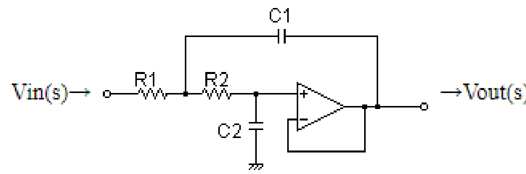

Fig. 4. Low-Pass Butterworth Filter Design[5]

To find the response signal of the circuit of Figure 3, we derive to the transfer function:

$$G(s) = \frac{\frac{1}{R1C1R2C2}}{s^2 + s(\frac{1}{R2C1} + \frac{1}{R1C1}) + \frac{1}{R1C1R2C2}} \tag{4}$$

For a cutoff frequency of 150 Hz we get:

$$\frac{1}{\sqrt{R1R2C1C2}} = 2\pi * 150 \tag{5}$$

By selecting $R1 = 20K\Omega$, $R2 = 12K\Omega$, $C1 = 0.1\mu F$ and $C1 = 0.047\mu F$ and populating the transfer function of equation (4), we find:

$$G(s) = \frac{0.89 * 10^6}{s^2 + 1.33 * 10^3 s + 0.89 * 10^6} \tag{6}$$

Similarly to the frequency cutoff of the high-pass filter transfer function of equation (3), the low-pass frequency cutoff also needs to occur at -3.04 dB. By plotting equation (6) on Figure 4 the frequency of 150 Hz also indicates a magnitude of -3.04 dB.

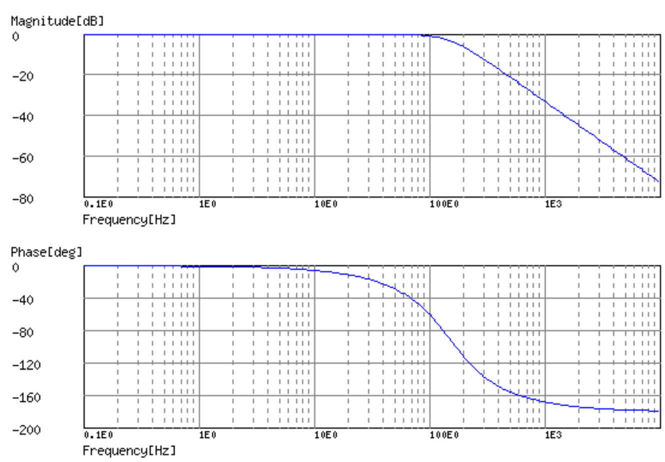

Fig. 5. Bode Plot of Low-Pass Filter

3) Non-inverting amplifier: The operational amplifier is perhaps the most common electronic component and one of the most important. It is used for the last stage of the circuit as a buffer to the signal that passes the filters, figure (6). The signal is amplified in this stage for the objective of being displayed.

The gain of interest is A=100 for this stage, which together with the other components of the system adds to approximately 1000 overall gain of the circuit.

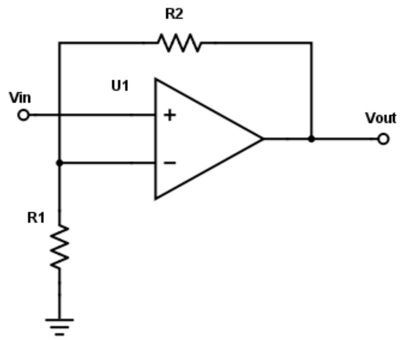

Fig. 6. Non-inverting Amplifier Diagram

The non-inverting amplifier of Figure (6) have a gain of:

$$A = 1 + \frac{R2}{R1} \tag{7}$$

For a gain of 100 the resistors ratio are $\frac{R2}{R1} = 99$. Thus, by selecting standard resistors to be $R1 = 1K\Omega$ and $R2 = 100K\Omega$, the desired ratio is approximately reached.

B. Circuit Simulation

Following the design of the circuit, the system was implemented in Multisim for simulation. This simulation allowed for validation of the aforementioned design as well as the fine tuning of different resistor and capacitor values in order to match practical values which could be obtained easily. Each stage was individually realized in Multisim and finally concatenated into the single large design to analyze overall response of the filtering components. The components which were analyzed were the high-pass filter, low-pass filter, and the non-inverting amplifier. Although many of the frequency plots seem similar, this is due to the response of the Butterworth filter and how it is illustrated on a logarithmic scale.

1) High-Pass Filter: The high-pass filter yielded a sharp frequency response, as expected from the Butterworth type filter. Figure 7 illustrates the simulated response of the second-order Butterworth high-pass filter. Our implementation relies on a slight change to the resistor values and outputs a frequency response with cutoff frequency at about 1.2Hz. This was not seen as an issue, as it still effectively mitigates DC bias entering the signal.

2) Low-Pass Filter: The low-pass filter was implemented without issue. Since practical values were used in analysis, no further change was necessary. This allowed the simulation to yield a 149 Hz cutoff frequency, nearly the exact same as the desired cutoff which was modeled above. The desired cutoff frequency was achieved with practical resistor and capacitor values which remained unchanged from the analysis above.

3) Overall System Response: After simulating and analyzing each filter's individual response in the cascade, the non-inverting gain was added. This yielded a maximum gain of about 16 dB, which boosted the final signal to an appropriate

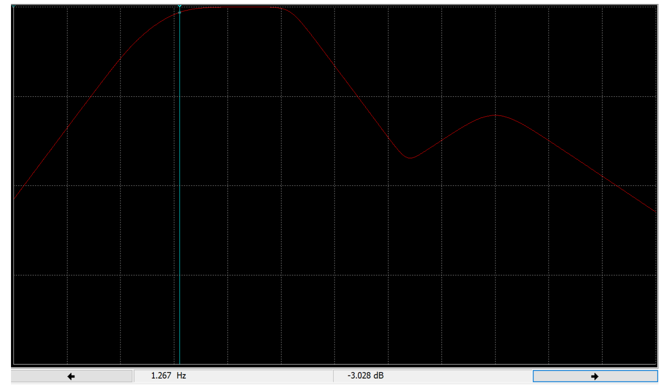

Fig. 7. Simulated frequency response for high-pass filter (logarithmic)

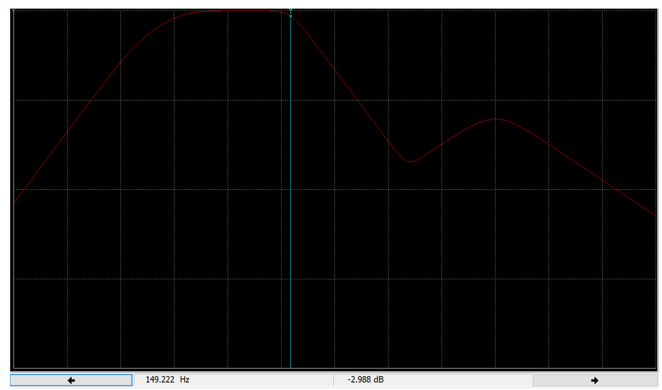

Fig. 8. Simulated frequency response for low-pass filter cascaded after high-pass filter (logarithmic)

voltage for data acquisition. The high-pass and low-pass cutoffs remained at appropriate locations and the overall system simulation yielded the band-pass response which was required to filter the ECG frequencies and amplify to a reasonable level.

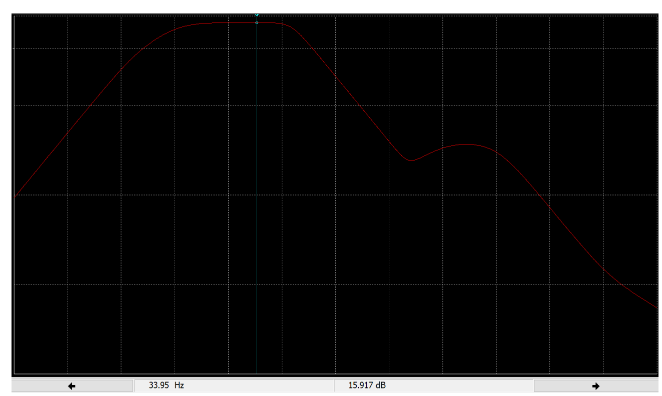

Fig. 9. Simulated frequency response for high-pass, low-pass filter including non-inverting gain (logarithmic)

C. Final Design

The final design after simulation included slight modifications to the high-pass filter resistors, although little changed beyond that. The entire filtering system exhibited a frequency response as seen in Figure 9. The maximum simulated gain is approximately 16 dB. The final design and actual implementation of the device relied on the following active components:

- L7905CV Negative Voltage Regulator
- L7805CV Positive Voltage Regulator
- INA126P Instrumentation Amplifier
- (2) TLC2272AC Operational Amplifiers

The construction of the device is as follows: the device is powered by two 9V batteries wired in series. The middle terminal at the intersection of one battery's positive terminal and the others negative terminal is used as reference (ground). The series of batteries then produces +9V and -9V. These power 'rails' are inputted into the positive voltage regulator and negative voltage regulator, respectively, in order to produce +5V and -5V rails for use with the op-amps, as their maximum voltage was exceeded with the original 9V batteries. The signal is first fed into the positive and negative terminals of the instrumentation amplifier. The instrumentation amplifier relies on a $16k\Omega$ resistor in order to provide a 10 V/V gain on the input signal. This is important to provide a signal which is big enough for our system to even filter. The signal leaving the instrumentation amplifier enters the second-order Butterworth high-pass filter with resistor and capacitor values given in Figure 10. This filter will block all DC bias in the input. The output is then cascaded into the second-order low-pass Butterworth filter with resistors and capacitors given in Figure 10, which will block all other frequencies which are not of interest greater than 150Hz. Each capacitor used in the design and the construction of the system was a ceramic capacitor. Preference was given to the most accurate resistors for implementation. Finally, the signal is passed into a non-inverting amplifier with $100k\omega$ and $1k\omega$ resistors to achieve the final amplification of the filtered signal.

Lastly, each active component features a capacitor between the power rails to provide a slight buffer and filter any unwanted noise entering the system. The experimental results reflected the success of the addition of the capacitors for each of the active components.

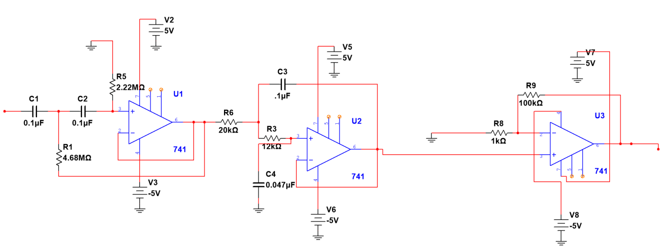

Fig. 10. Overall implementation of the device's filtering and non-inverting amplification

IV. Results

Ultimately, the design and construction of a rather simple ECG amplifier yielded an effective circuit which successfully produced an electrocardiogram when attached to a team member. The electrocardiogram which was obtained from the system is depicted in Figure 11. This device was able to filter almost all unwanted noise and generate an accurate electrocardiogram with visible portions of each cycle in the heartbeat's cycle. The accuracy of results are also in part due to the medical pads which were used to attach the amplifier to the person. The connection between the device and the person whose heart it is measuring is crucial to the operation of the device. In addition to the success of the filters based upon their frequency response, the capacitors added to the power rails of each op-amp mitigated much of the noise which would have arisen simply from the power of the active filters.

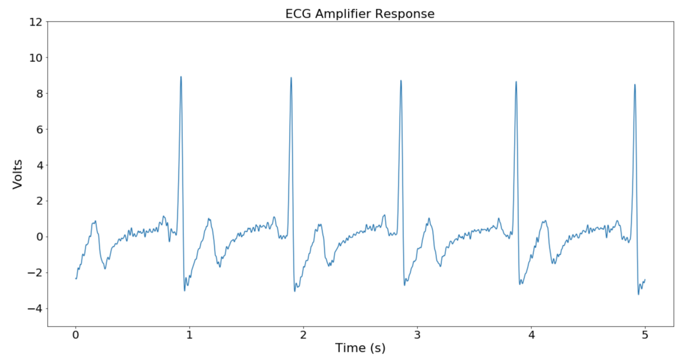

Fig. 11. Electrocardiogram obtained from device

V. Conclusion

The project was able to successfully create a device from simple components to develop a means of generating an accurate electrocardiogram. The skills and knowledge obtained in ECE 3813 allowed our team to first design a theoretical device, simulate its implementation, and finally construct a working system. For such a ubiquitous device in the medical industry, and surely an expensive one, there was great success in our team's accomplishment of developing our own ECG system with minimal cost. In the future, further budget applications may exist for similar technologies. In small at-home units, there are likely similar components to our simple design. The project proved the feasibility of creating a valuable device with simple, inexpensive components to achieve accurate results.

References

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