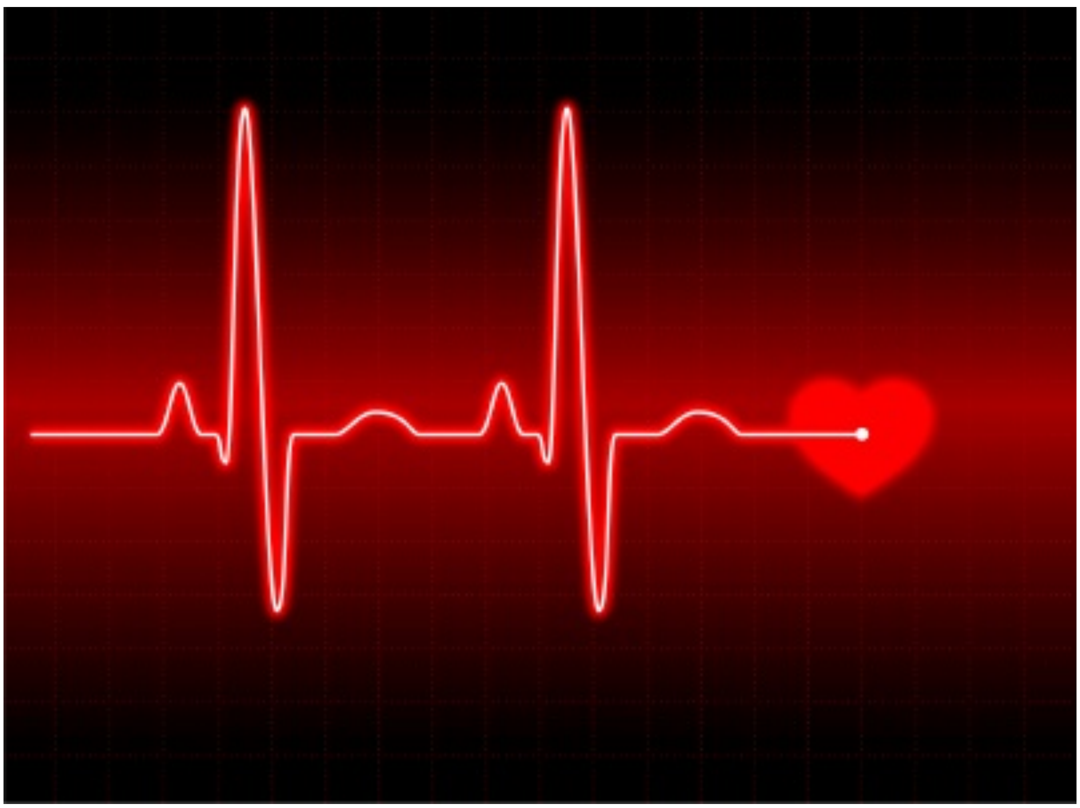
SIMPLE HEART RATE MONITOR FOR ANALOG ENTHUSIASTS



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6.101: Final Project Report
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Project Motivation

Heart Rate Monitors are quickly becoming ubiquitous in our daily lives. From FitBit's Flex Wristband to Garmin's Chest Strap to Philip's IntelliVue Bedside Monitor. In this age of information, the more we know about our health status, the better prepared we are to take good care of our selves, seek medical attention when necessary, and push our physical limits through vigorous exercise regimens.

Naturally, electronics play a big role in the implementation of most, if not all, heart rate monitors, along with sophisticated algorithms designed to analyze and display the pertinent details. Although heart rate monitoring is considered to be in the domain of digital electronics, an understanding of analog electronics is necessary to obtain the full functionality of any electrocardiogram (ECG) device.

Hence our decision to build a simple heart rate monitor for analog enthusiasts as a proof-of-concept application that explores the application of various technologies that have been taught in the analog electronics course (6.101) during the spring semester of 2014. We successfully implemented analog circuit solutions for designs that would otherwise be implemented using digital electronics given our design specifications. Moreover, we confirmed a good grasp of the concepts taught in the course, and built a useful product that demonstrates the relevance and practicality of analog electronics in ECG devices.

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Introduction

An electrocardiogram (ECG or EKG) is a simple, non-invasive way of measuring the heart's electrical conduction system by picking up electrical pulses generated by the polarization and depolarization of cardiac tissue and translating it into a waveform. An ECG shows:

- How fast a heart is beating
- Whether a heartbeat is steady or irregular
- The strength and timing of electrical signals as they pass through each part of a heart

ECGs are typically performed for diagnostic and research purposes such as detecting and studying heart problems, and investigating other disorders that can affect heart function.

Although the field of electrocardiography is already well established, we proposed and implemented a simple and cheap method of visually and audibly displaying the frequency of a heartbeat. In order to do so, we used a circuit design developed in laboratory exercise 5 of 6.101 to generate a heartbeat, even though we primarily relied on the ECG signal from the function generator for testing purposes. We then researched and realized circuits to convert a heartbeat into an optical signal, transmit the heartbeat along an optical fiber cable, detect and convert the heartbeat into an electrical signal, amplify the heartbeat signal, measure the pulse rate, and display the pulse rate using an analog meter, a speaker and a light emitting diode (LED).

Materials and Components

The materials listed below were used to construct and test the heartbeat monitor. While most items were already available in the MIT EECS Instructional Laboratories' stockroom, the fiber optic detector and receiver were ordered from Digi-Key, an electronic components distributor.

- ECG Printed Circuit Board (PCB) designed and built in lab 5 of 6.101
- Avago Technologies Discrete Fiber Optic Laser Diode Transmitter
- Avago Technologies Discrete Fiber Optic Photodiode Receiver
- Optical fiber
- Several LF356 Operational Amplifiers
- Several LF353 Operational Amplifiers
- Several LM311 Comparators
- Texas Instruments SN74LS86 active exclusive-OR gate
- Several 1% and 5% resistors of varying values

- Several 100K potentiometers
- Several 1N914 diodes
- Several capacitors of varying values
- Copper wire
- Analog meter
- Speaker
- Yellow Light Emitting Diode

Labor Division

Although we consider the project a team effort, we divided the work to streamline and accelerate the design and implementation of our circuit. Since we already had an ECG PCB from a lab experiment earlier on in the semester, our work primarily dealt with the signal transmission, detection, amplification, measurement and display. We designed our separate modules to be compatible with each other, so as to minimize complications during integration.

While Abigail focused on ECG signal transmission along the optical fiber, signal detection, and signal measurement and conversion to display it on an analog meter, Jelimo focused on signal conditioning and amplification, and signal demonstration using both a speaker and an LED.

We both contributed to the integration of the separate modules and jointly conducted testing of the complete system. Abigail implemented the circuit design in Eagle (PCB Schematic Design Software) and Jelimo verified her design to ensure no mistakes or misunderstandings. Both Abigail and Jelimo did the project presentation and demonstration, as well as participated in writing the final project report. More specifically, Abigail generated a final circuit schematic and wrote about the circuit aspects that she worked on, and Jelimo wrote about the circuit aspects that she worked on, and compiled the entire report.

Modular Design

It was essential that we take a modular design approach to enable easier integration and testing. Figure 1 below is a block diagram of the separate modules, followed by a more in-depth explanation of the design considerations within each module.

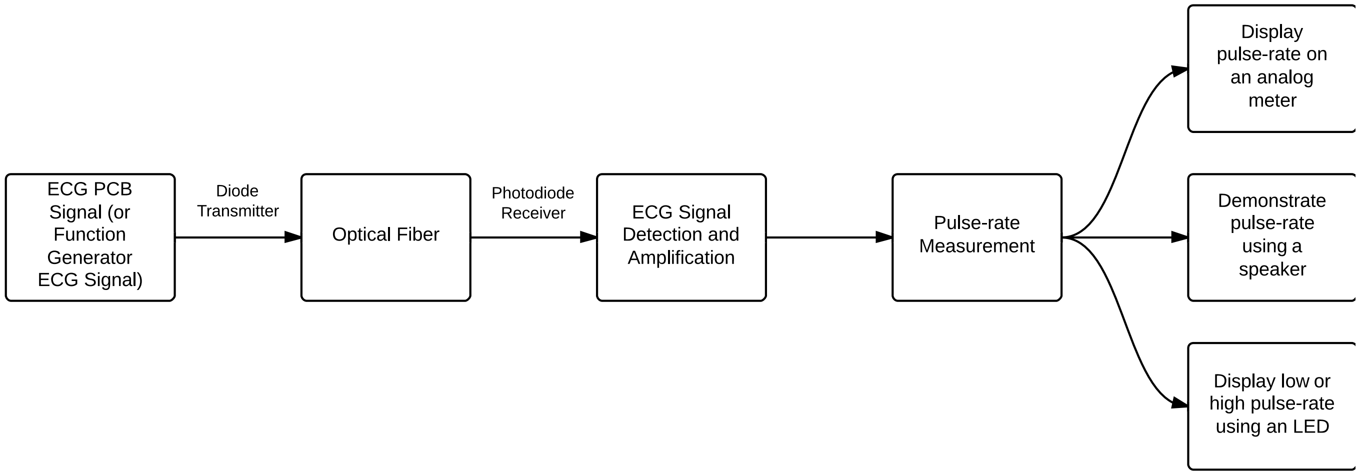

Figure 1: Block Diagram of the Circuit Implementation of our Heartbeat Monitor

a) **ECG Signal Generation**
While we primarily relied on the ECG signal from the function generator to test our circuit, we had the option of using the ECG PCB board from lab 5 of 6.101. We designed our circuit with the goal of being able to use either the function generator or PCB board to generate the ECG signal. The ECG signal (cardiac signal) is found under arbitrary signals when the arbitrary button of the HP-Agilent 33220 Model is pressed.

b) **ECG Signal Transmission**
As a consequence of both our interests in photonics, we decided to use optical fiber to transmit the ECG signal. Optical fiber not only has a very small attenuation factor, but also permits the transmission of other signals at different wavelengths. Although we intended to also transmit and play music in addition to displaying the pulse-rate, we were unable to do so due to time constraints.

To transmit the ECG signal, we modulated it at a higher frequency, converted it to an optical signal using the Avago Technologies Discrete Fiber Optic Laser Diode Transmitter, sent the optical signal along an optical fiber, detected it using the Avago Technologies Discrete Fiber Optic Photodiode Receiver, and demodulated it to obtain the pure ECG signal.

c) **ECG Signal Conditioning and Amplification**
In order to correct for any DC offset and signal attenuation introduced during signal transmission and reception, we implemented an adder and gain stage. To compensate for a DC offset, the ECG signal was integrated and compared to ground, and any difference was added at the

adder and gain stage. A Sallen-Key filter (a second-order low pass filter) was used to remove noise from the signal.

d) Pulse Rate Measurement
While the basic concept behind the pulse rate measurement was fairly straightforward, it proved challenging to implement. The idea was to convert a frequency (the heart rate) to a voltage, which would then control a voltage-controlled oscillator (VCO). We tried implementing several designs with the goal of minimizing the ripple as much as possible, and eventually settled for a current source and peak detector. (See the section pulse rate measurement under modular implementation and testing for further details on the different circuit configurations that were attempted.)

e) Visual Display of the Pulse Rate on an Analog Meter
Displaying the pulse-rate on an analog meter was simple once the frequency to voltage conversion was accomplished. We essentially focused on maximizing the voltage swing such that different heart rates could be displayed with finer accuracy, and finally settled on a voltage range of 7V to 13.5V, the upper limit of which was limited by our power rails. (We used a +15V and -15V power supply for all the active components in our circuit.)

f) Audible Display of the Pulse Rate Using a Speaker
Once the frequency was converted to a voltage, the voltage was used to control a voltage-controlled oscillator (VCO). The VCO basically produced either a square wave at a frequency that corresponds to the control voltage i.e. lower voltage = lower frequency, higher voltage = higher frequency. The oscillation would then be heard on a speaker that was powered by a Darlington pair. A Darlington pair was used to obtain a higher common emitter current gain than would be obtained using a single transistor.

g) Visual Display of the Pulse Rate Using an LED
The same voltage used to control the VCO was also compared to two voltage levels that we characterized as corresponding to extremely low (7.5V) or extremely high (12.5V) pulse-rates. The output from the two comparators was combined using an exclusive-OR (XOR) gate. At any given time, the combined output from the comparators would produce a high at the XOR output when the pulse rate is too low, low when the pulse rate is in the normal range (between 7.5V and 12.5V), or high when the pulse rate is too high. The XOR output was then amplified to power an LED that is on when the pulse rate is too low or too high, and off when the pulse rate is in the normal range.

Modular Implementation and Testing

a) ECG Signal Transmission (Abigail Rice)

The circuit schematic representing ECG signal transmission and detection is in figure 2 below.

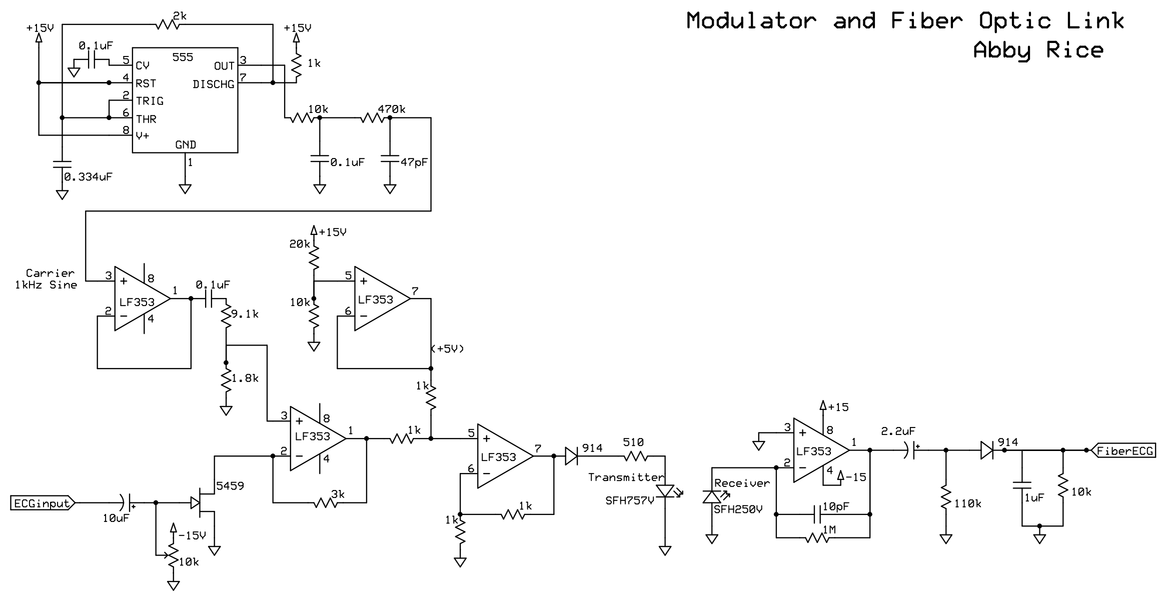

Figure 2: ECG Signal Transmission and Detection

Once I characterized the ECG PCB that we constructed in lab 5, I set the function generator parameters so that its cardiac signal matched that of the PCB as much as possible in case we switched from using the function generator to the PCB during testing and integration. The ECG signal used had an amplitude of 200 mV with 0 DC offset. I further tuned the function generator ECG signal to work in the frequency range of ~30 beats per minute (bpm) to ~120 bpm, which correspond to 0.5 Hz and 2 Hz respectively.

Amplitude Modulation (AM) was used to modulate the ECG signal with a higher frequency carrier. This was done by generating a carrier sine wave, multiplying the ECG and the carrier wave, and sending the modulated signal to the fiber optic transmitter.

To generate the carrier wave, I created a simple oscillator circuit using a 555 timer chip in an astable configuration as shown in figure 3 below. The 555 switches states every time the trigger drops below 2/3 of the supply rail, so by attaching it to a RC network with a specific time constant, it is easy to create a square wave at any frequency. A resistor of 2 kΩ and a capacitor of 0.334 μF were chosen, yielding a square wave at 1 kHz.

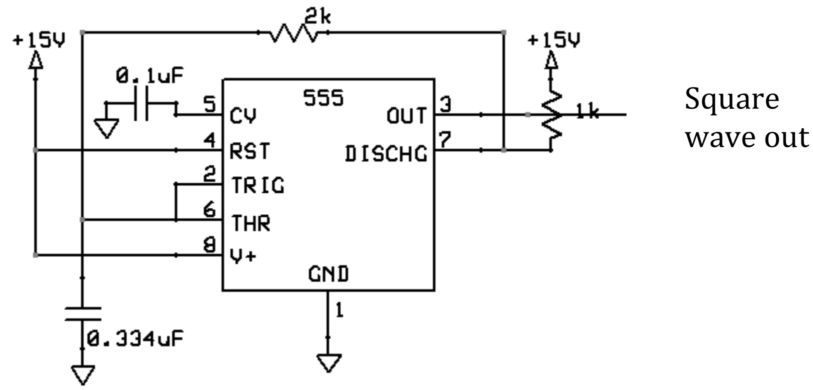

Figure 3: 555 Timer Chip in Astable Configuration

The square wave was then converted to a sine wave by passing it through low pass filters (LPFs) that are selected to allow the fundamental frequency through but not the harmonics that make a sine wave a square wave. Through experimentation, I settled on using two LPFs in series with slightly different time constants because they made the resulting sine wave much cleaner than when only one LPF was used. A resistor divider network was then added to reduce the sine wave amplitude to 200 mV.

I decided to use a JFET as a variable resistor inside an operational amplifier (op-amp) gain stage to modulate the signal. The op-amp multiplies the input signal by the gain, given by R2/R1. I chose R1 to be the variable resistor so that the output of the op-amp would be proportional to the product of the two input signals (the ECG and the 1 kHz carrier sine wave).

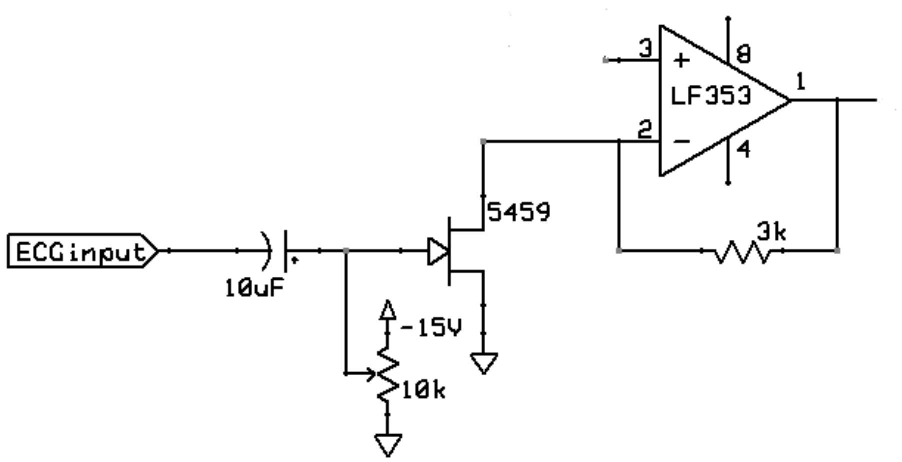

Figure 4: Modulator Circuit Using a JFET as a Variable Resistor

The JFET acts like a variable resistor whose resistance (slope of the I-V characteristic) varies with the gate-to-source voltage (the ECG signal) when biased in its linear region. JFETs must be biased slightly negative, so I added a biasing stage with a potentiometer to finely tune the bias point. I also needed to add a blocking capacitor so that the biasing voltage would not be fed back to the function generator.

After the signal was modulated, an LED in the Avago Technologies' Transmitter converted it to an optical signal and coupled it into the fiber. The brightness of the LED varies with current, not voltage, so I converted the voltage signal into a current signal. This was accomplished by increasing the DC bias by 5V, so that it was always above the threshold voltage of the LED (around 3V) and fed it through a series resistor. The current signal was then passed directly to the transmitting LED.

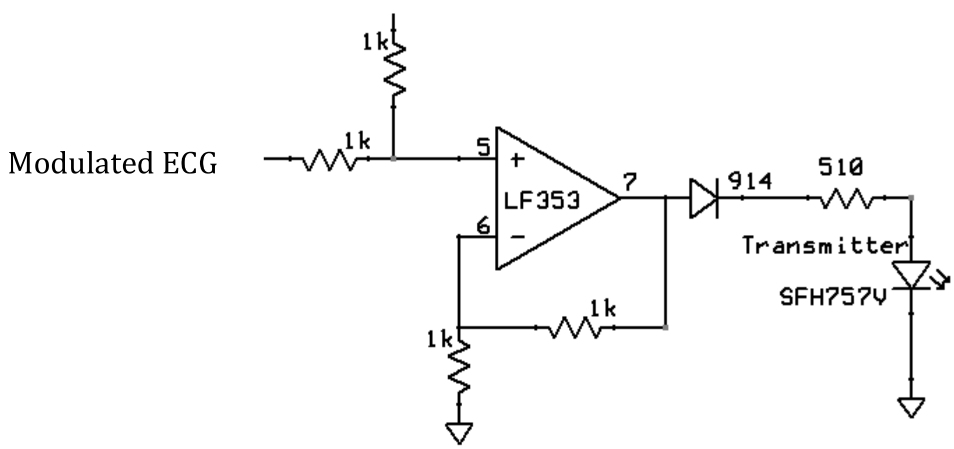

Figure 5: Sending the Signal to the Transmitting LED

At the other end of the optical fiber, the light hits the photodiode receiver, which converts it back to a current signal. This signal was detected using a transimpedance amplifier similar to the one built in lab 6 of 6.101. However, because the signal obtained was much stronger than the input signal in lab 6, there was a smaller gain and I had to add a DC blocking capacitor so that the demodulator would work correctly.

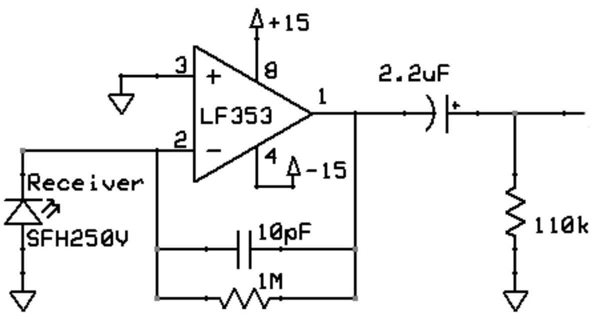

Figure 6: Photodiode Receiver and Transimpedance Amplifier

Signal demodulation was simpler than signal modulation. Since the amplitude was a direct representation of the signal, all I had to do was detect the envelope and remove the high frequency carrier. I used a diode in series with another LPF that had a time constant of 10 ms, thus removing the carrier wave as well as any noise that was higher than 100 Hz. This smoothed out and distorted the ECG signal a bit, but this was not detrimental for our application since we only needed the peak of each pulse to be distinct. After the signal was demodulated, it was conditioned and amplified at the next stage.

b) ECG Signal Conditioning and Amplification (Jelimo Maswan)

The circuit schematic representing ECG signal conditioning and amplifications is in figure below.

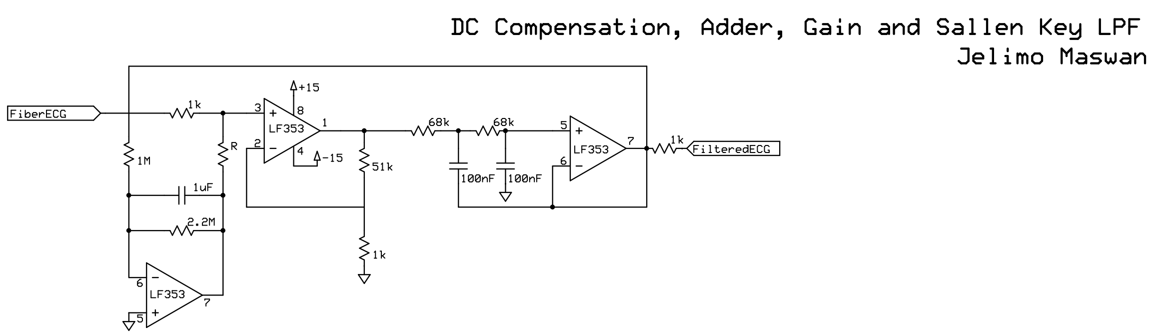

Figure 7: ECG Signal Conditioning and Amplification

This stage consisted of 3 major blocks: DC offset compensation, gain, and a Sallen-Key LPF. The implementation of this stage was borrowed from lab 5. To compensate for the DC offset voltage, the ECG signal is integrated and compared to ground, and any difference is added at the gain stage. An integration period of ~2 seconds was chosen to correspond to the upper frequency limit of our ECG signal i.e. 120 Hz.

A non-inverting op-amp configuration was used to add the DC offset compensation and get a gain of 51. Thereafter, the signal was filtered using a Sallen-Key LPF designed to have a cutoff frequency at 23 Hz to eliminate noise from the supply rails that is typically at around 60 Hz along with any other high frequency noise signals. Figure 7 above shows the capacitor and resistor values for DC offset compensation, the adder and gain stage, and the Sallen-Key LPF.

c) Pulse Rate Measurement (Abigail Rice)

The circuit schematic representing the pulse rate measurement technique is in figure below.

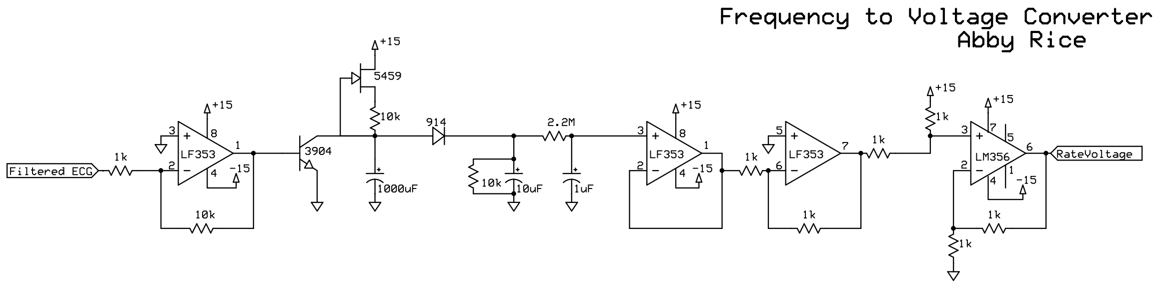

Figure 8: Pulse Rate Measurement

The frequency to voltage converter (FVC) module is responsible for measuring the heartbeat frequency. It turns the series of pulses into a DC voltage, which corresponds to the pulse rate. This voltage is displayed visually on an analog meter.

Originally, I thought this could be simply implemented with a diode and a large capacitor. The diode would force the signal to be positive (so as not to discharge the capacitor) and the capacitor would be charged at every pulse and discharge slowly between pulses. With rapid pulses the capacitor could not discharge fast enough, leading to a higher average voltage, but low rates meant a large drop in the voltage. We had used this circuit in lab 2 of 6.101 to turn an AC waveform into a DC voltage with a small ripple, which would not be visible on the analog meter. However, this does not work for heart rate frequencies (0.5 Hz – 2 Hz) because the ripple is a large percentage of the value. When the signal moves up and down only once every one or two seconds, it is very hard to read the average value on the analog meter. Adding an integrator after the conversion did not help much, because a long time constant was needed.

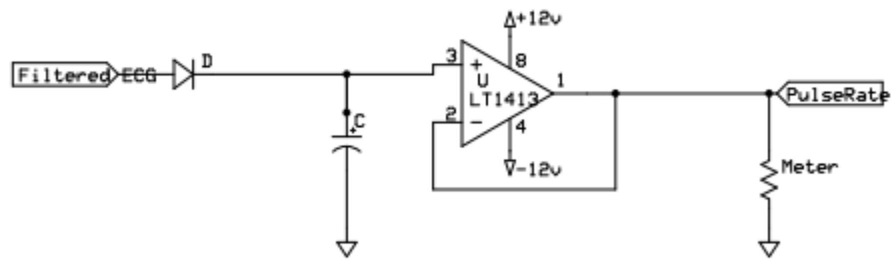

Figure 9: Proposed Method for FVC. Simple but Ineffective for Very Low Frequencies

After trying this, I attempted other strategies to solve the FVC problem. One of these was using a 555 timer as a one-shot pulse generator and configuring it such that it produces a PWM where the width of the pulse depends on the initial frequency. This was successful. However I had no means of translating a variable PWM into a DC voltage without using digital circuitry, which I wanted to avoid. Another idea was to use a LM331 chip (precision voltage to frequency converter) that has a well-documented application for frequency to voltage conversion. This would have been easy, but I wanted to find a way to do it with discrete components, even if this meant giving up some linearity, which was not crucial for the scope of our project.

Finally, I generated a sawtooth signal instead and created a DC voltage using a peak detector. First of all, I converted the ECG signal to a current source, which constantly charged a capacitor that would be shorted out at every pulse. The capacitor charges at a constant rate from a current source, so its voltage ramps up linearly. Shorting it to ground causes an immediate drop to zero at a varying point on its curve. A lower rate allowed the capacitor to charge more fully and have a higher peak voltage. This implementation was the opposite of the first scheme, but it produced a nice sawtooth waveform whose amplitude depended on the frequency.

I created this circuit using a 5459 JFET configured as a constant current source into a large capacitor (1000uF). I shorted the capacitor using a BJT switch connected to the incoming ECG signal. This created a clean sawtooth wave at the output that varied in amplitude between 7V and 13.5V. A sketch of the schematic is shown in figure 10 below.

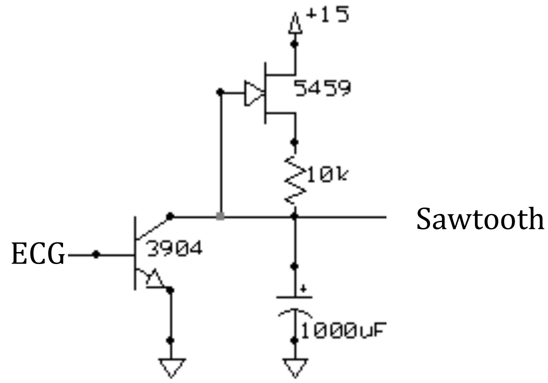

Figure 10: Sawtooth Wave Generation Using a JFET Current Source

Now that I had a variable sawtooth, I used a peak detector to capture and hold the peak value, effectively turning the sawtooth into a DC voltage at the peak level. The first method I tried was a resettable peak detector. A peak detector is just a diode and capacitor, as in the first scheme, so that when the input voltage is higher than the voltage on the capacitor the diode is on, and the capacitor charges to that higher voltage. But once it is equal, or goes lower, the diode turns off leaving the capacitor to store the value. This works great if all you want is the peak value, but for this application we needed to update the peak value to represent a constantly fluctuating frequency, and without being able to reset it we could only increase in frequency, never decrease. Therefore, a switch was placed in parallel with the capacitor to discharge it every cycle and allow it to collect a new peak value. Ideally, this reset should take place just before the peak value is reached so the capacitor can hold that value for the majority of the next period, with only a small dip down to zero when the reset happens.

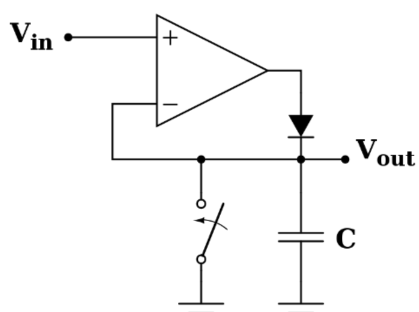

Figure 11: Resetting Peak Detector

There are two problems with the resetting peak detector circuit that led me to abandon it. First, I did not like the periodic pulses messing up the

DC voltage. Secondly, and more importantly, I needed to find a way to generate a signal that would reset the detector just *before* the ECG signal came in. Feeding the ECG directly to the reset switch was useless because we would be resetting the detector precisely when the peak was hitting, so we lost the signal and it only gave us another sawtooth at the output. Adding a small delay to a signal is easy in the digital realm, but difficult to implement in analog circuitry. I gave up this design for a simpler method of discharging that was a reasonable compromise between a low ripple voltage and being able to dynamically change the frequency up and down.

Instead of periodically resetting the capacitor voltage, I replaced the switch with a resistor so it would constantly discharge slowly. This has the advantage of being simple, effective, and the time constant is easy to tune to the desired frequency range. However, since the capacitor is constantly discharging, we have the same problem as the first case where the ripple voltage is large. Nevertheless, using a sawtooth instead of the pulses made the ripple more manageable, and I was able to find a time constant with an acceptable wiggle.

After the peak detector, I added a low pass filter to further reduce the ripple voltage. This filter had a long (few seconds) time constant, so it basically took the average voltage over 3-6 cycles. Thus a steady DC voltage was generated with a small ripple and a wide swing that varied with frequency from 7 volts to 13.5 volts. It took about five seconds to settle on a value after changing the frequency, but this was a perfectly acceptable trade-off because in real applications, heart rates do not change rapidly and it is reasonable to wait a few seconds before reading the voltage value.

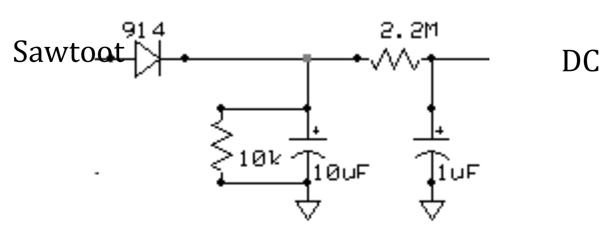

Figure 12: Peak detector and Low Pass filter

Having obtained a DC voltage, I made a few minor adjustments before transmitting it to the next module. Firstly, I buffered the signal so there is no current being drawn from the low pass filter. I then inverted the signal to correct for the inversion in the FVC (a low frequency corresponded to a high voltage because the capacitor was discharging every cycle). A DC offset was also added to put the DC voltage in a nice range for display on an analog meter and for the audio and LED modules.

DC

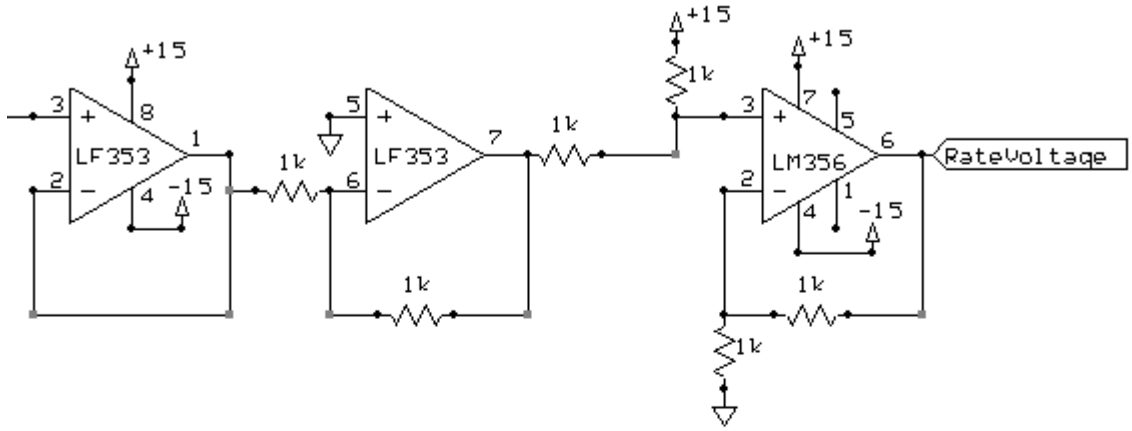

Figure 13: Voltage Buffer, Inverter and DC Offset

d) Audible Display of the Pulse Rate Using a Speaker (Jelimo Maswan)
The circuit schematic representing speaker module is in figure 14 below.

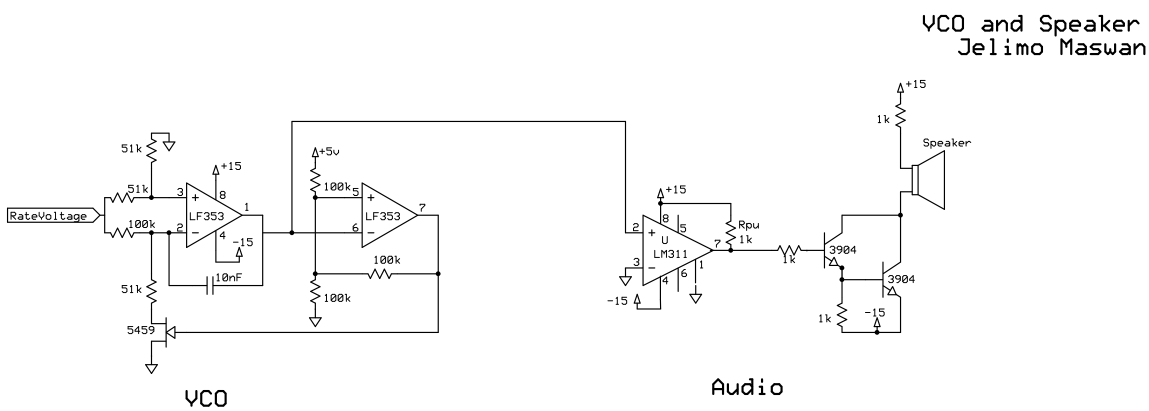

Figure 14: Audible Display of the Pulse Rate Using a Speaker

A VCO built from discrete components was used to convert the DC voltage from the FVC module to an oscillation. A VCO is basically an oscillator whose frequency is determined by a control voltage. The first op-amp is a voltage divider that puts the + input at half the control voltage. The op-amp attempts to keep its input at the same voltage, which requires a current flow across the 100k resistor to ensure that its voltage drop is half the control voltage.

When the MOSFET at the bottom is on, the current from the 100k goes through the MOSFET. Since the 51k resistor has the same voltage drop as the 100k but half the resistance, it must have twice as much current flowing through it. The additional current comes from the capacitor, charging it, so the first op-amp must provide a steadily rising output voltage to source this current. When the MOSFET at the bottom is off, the current from the 100k goes through the capacitor, discharging it, so a steadily falling output voltage is needed from the first op-amp. The switching of the MOSFET results in a triangle wave at the output of the first op-amp.

The second op-amp is a Schmitt trigger. It takes the triangle wave as an input. When the input voltage rises above the threshold of 3.33V, it outputs 5V and the threshold voltage falls to 1.67 V. When the input voltage falls below 1.67V, the output goes to 0V and the threshold moves back up. The output is a square wave. It is connected to the MOSFET, which causes the integrator to raise or lower its output voltage as needed. See figure 15 for a circuit schematic of the VCO.

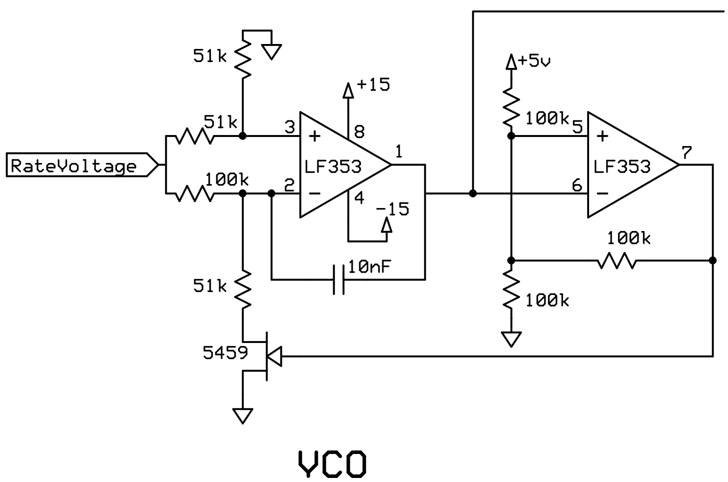

Figure 15: Voltage-Controlled Oscillator Circuit

Although the initial intention was to use the oscillation from the VCO to produce sound through a speaker, the output of the VCO had a negative DC offset that cannot be used to bias the BJTs in the Darlington pair configuration. Therefore, I used a comparator to produce an output voltage that swung between +15V and -15V, such that the BJT was on when the bias was positive, and off when the bias was negative, effectively producing an oscillation at the output of the Darlington pair which was at the same frequency as the oscillation at the input of the BJT. The Darlington pair was used to power a speaker that produced a tone at a frequency that corresponded to the control voltage at the input of the VCO, which was in turn proportional to the pulse rate. Although the relationship between the heart rate and the oscillation frequency was not linear, the tone produced still indicated the change in frequency as the pulse rate was raised or lowered. See figure 16 below for the Darlington pair configuration along with the speaker.

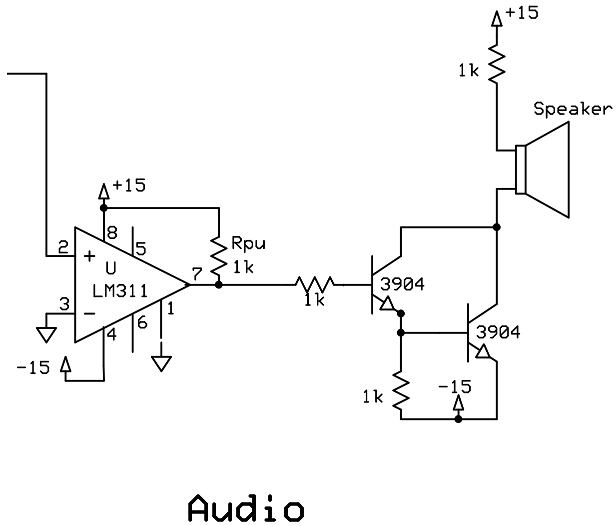

Figure 16: Darlington Pair and Speaker

e) Visual Display of the Pulse Rate Using an LED (Jelimo Maswan)

The circuit schematic representing the LED module is in figure below.

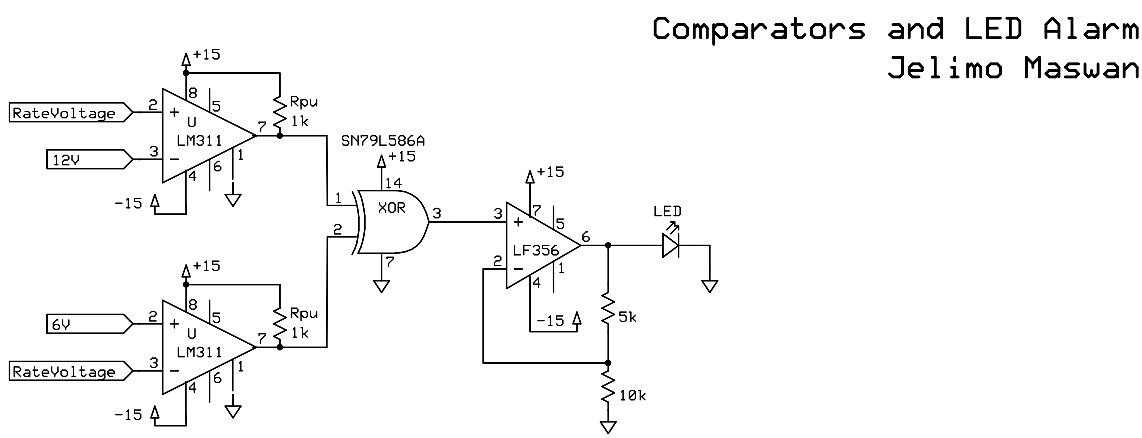

Figure 17: Visual Display of the Pulse Rate Using and LED

In order to indicate extremely low or extremely high pulse rates, two comparators and an exclusive-OR gate were used. A 100k potentiometer was used to obtain the reference voltages for the comparators, such that a voltage below 3V would indicate an extremely low pulse rate, and a voltage above 10V would indicate an extremely high pulse rate. The 3V and 10V values were chosen based on the range of voltage output levels from the frequency to voltage conversion stage. The outputs from the two comparators were then combined using an XOR gate, whose output was high only when the pulse rate was either too low or too high. The XOR gate output was then amplified before using it to power an LED. The LED was correspondingly only on when the pulse rate was either too low or too high, and off in the normal heartbeat frequency range.

Module Integration

Having designed our component values to be compatible with each module, integration proved to be less of a challenge than initially expected. Naturally, some design changes already mentioned above in the modular design section were made to accommodate variations in the output signal, along with the generous use of buffer stages between separate modules to isolate them.

The most difficult aspect of integration had to do with identifying the voltage output range from the frequency to voltage conversion range, such that the speaker and LED modules were compatible. Since I (Jelimo) decided to use potentiometers whenever possible to make adjusting the resistor values in the speaker and LED modules, it was fairly easy to accommodate a range of voltage ranges. Eventually, we were able to characterize the final implementation of the frequency to voltage conversion, and although the relationship is not linear, it was good enough and certainly usable for a low cost heart rate monitor. See figure below for a table of heart rate, frequency and voltage values as well as plot of the heart rate versus DC voltage levels.

Frequency	Heart rate	Voltage
0.5 Hz	30 bpm	7.2 V
0.7 Hz	42 bpm	7.8 V
1.0 Hz	60 bpm	9.4 V
1.3 Hz	78 bpm	10.8 V
1.5 Hz	90 bpm	11.4 V
1.8 Hz	108 bpm	12.2 V
2 .0 Hz	120 bpm	12.8 V
2.5 Hz	150 bpm	13.8 V

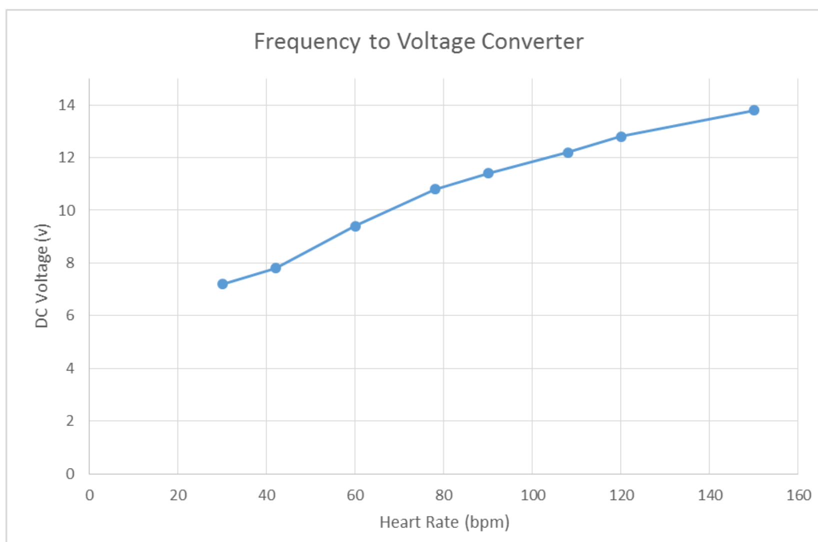

Figure 18: Frequency to Voltage Converter Measurements

Conclusion

While our simple heart rate monitor is not likely to ever be realized in practical medical applications because of the trend towards wireless transmission to avoid unnecessary cable connections (optical fiber in our case) to electronic medical devices, it is an excellent way to gain insight into the analog electronics behind the design of heart rate monitors.

Additionally, several of the concepts learned in 6.101 were used, enhancing our confidence in designing and implementing analog systems. Although we were unable to also transmit music along the same optical fiber at a different wavelength to demonstrate the capability of fiber optics to transmit different information using channels at different wavelengths, we were able to visually and audibly display the change in pulse rate.

Finally, the value of teamwork was clearly demonstrated, since we complemented each other's knowledge and assisted each other in researching possible circuit schematics to address some of the challenges we faced. Many thanks go out to the 6.101 staff for all the assistance and guidance provided throughout the semester, from teaching the basics of analog design through lectures, lab experiments, problem sets and an exam, to their availability at every part of the design, implementation, testing, debugging and demonstration of our final project. We are also grateful to Mary Caulfield for her advice and feedback on our presentation and writing assignments, which aided in streamlining our thought process and learning to effectively present our ideas. It was a pleasure to learn from both the 6.101 staff and other students in the course.