Capacitive ECG Electrodes

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***Abstract*—In this paper, the team have gone over the design of a wireless ECG sensor network. Electrocardiography (ECG) is the process of recording the electrical activity of the heart over a period of time using electrodes placed on the skin. Using this wave, heart rate variability, beats per minute (bpm) can be detected. ECG is also vital in diagnosing cardiac arrhythmias such as ventricular tachycardia, atrial flutter, atrial fibrillation, sinus bradycardia, etc. In this project, the ECG signal is obtained using capacitive electrodes. Essentially, they are copper electrodes covered with an insulating material having a high dielectric to make them capacitive. The signal is operated on as it passes through the circuit. The circuit has amplification and filtering stages that output a finer ECG wave. This wave is further digitally filtered and displayed in MATLAB using a wireless Bluetooth module. The beats per minute and heart rate variability of the subject is calculated as well. This setup can be made portable with the use of a harness and a case to hold the circuit. Applications of this can be acquiring ECG waveform of burn victims whose skin is sensitive, and where wet or dry ECG electrodes cannot be used.**

Keywords—ECG; capacitive; heart rate variability; beats per minute; Bluetooth; MATLAB

I. BACKGROUND

The Electrocardiogram signal is generated due to polarizing and depolarizing of heart muscle during each heartbeat. Electrocardiography (ECG or EKG) is the process of recording this electrical activity of the heart over a period using electrodes placed on the skin. It is a very commonly performed cardiology test. An ECG records the heart's rhythm and activity on strip of paper or a line on the screen.

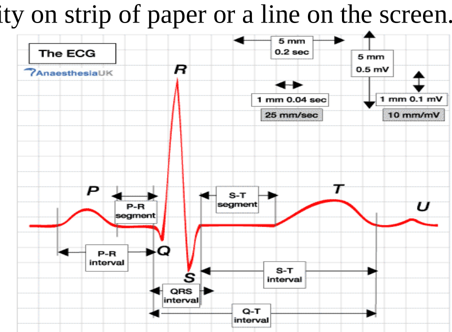

Figure 1: Basic ECG Signal [8]

An ECG waveform constitutes of the P wave, the QRS complex, the T wave and occasionally a U wave. The P wave and QRS complex represent atrial and ventricular depolarization, respectively. The T wave represents ventricular repolarization and the U wave represents Purkinje fiber repolarization. Based on the waveforms, cardiac arrhythmias and beats per minute of the heart can be calculated. The amplitude of signal varies form 0.1-5 mV and frequency range is 0.5-50 Hz. The table below summarizes this:

Event	Characteristics	Duration at 75 bpm (0.8 second cycle)
Atrial diastole Ventricular diastole	AV valves opened. Semilunar valves close. Ventricular filling.	0.4 seconds
Atrial systole Ventricular diastole	AV valves open. Semilunar valves closed. Ventricular filling.	0.1 seconds
Atrial diastole Ventricular systole	AV valves closed. Semilunar valves open. Blood pumped into aorta and pulmonary artery.	0.3 seconds

[9]

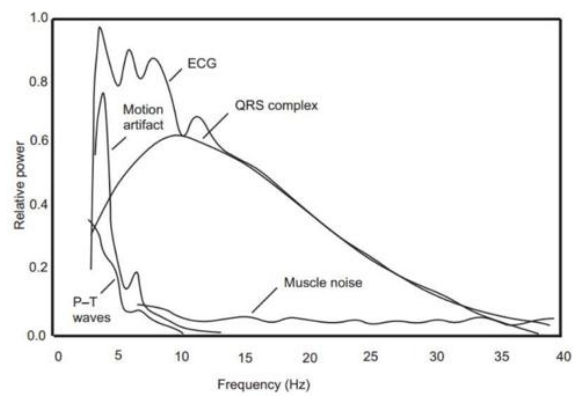

Figure 2: Relative Power Spectra of QRS Complex, P and T waves, muscle noise and motion artifacts based on an average of 150 beats [10]

The common frequencies of important components of the ECG are:

- Heart rate: 0.67 – 5 Hz (i.e. 40-300 bpm)
- P-wave: 0.67 – 5Hz
- QRS :10 -50 Hz
- T-wave: 1- 7Hz
- High frequency potential: 100-500 Hz

The traditional and the method which is most widely used by doctors to an get ECG signal is to use Ag/AgCl electrode with wet conductive gel. The basic principle used for this approach is to apply wet conductive electrolyte gel on the skin and place metal electrode on top of it, which helps electrode in detecting electrical activity under the skin.

The biggest advantage of using these electrodes is that they remain well-adhered to the skin which helps reduce the motion artifacts.

Apart from the above-mentioned conventional method, dry electrodes are also used, but are restricted to research applications and have yet to attain complete medical usage. They are designed with great precision to provide accurate high amplifier input impedance, bias current, etc. There are different types of materials used for dry electrodes, like a stiff metal, soft/flexible material, or fabric materials, to improve the electrode-skin contact and reduce the motion artifacts. [7]

II. Statement of Problem

Although traditional methods to measure ECG are simple and time tested, they have some major problems. The wet conductive gel method can cause skin irritation, and allergies, and it may dry up over the time, thus interfering with the signal transduction from body to electrodes. The method, of dry electrodes needs electrodes to be in contact with the body at all times but, without any gel it is difficult to hold electrodes in place and it will incur motion artifacts. [1] Apart from these issues, there were some major concerns related to the effect of noise on the signal. The friction of electrodes against the body also gave rise to artifacts in the received signals for dry electrodes and it was seen in a few low resistance wet contact electrodes as well. [6] Some more issues related to both these electrodes were interference with the AC power lines, perspiration effects (more common in dry electrodes), dry electrodes also suffered from charge sensitivity, etc.

The alternative design apart from the above mentioned is to use non-contact electrode. One topology for non-contact electrodes is to add a thin insulation layer to a dry-contact metal electrode [1]. Adding an insulation layer removes the half-cell voltage potential in electrode equivalent circuit and instead introduces a capacitance as a result of the air gap between the electrode and the body [1]. A non-contact electrode operates using the same principles as an insulated electrode, but it also must be able to couple through clothing which adds a parallel RC combination that must be considered in the electrode design [1]. Figure X outlines the basic types of electrodes described as well as shows the equivalent circuit models.

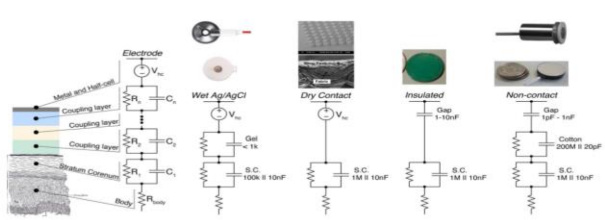

Figure 3: Common Types of Electrodes with Equivalent Circuits from [1].

The major challenges stopping the adoption of non-contact electrodes are outlined in [2] and are as follows: common mode electromagnetic interference from main power, the triboelectric effect from the additional clothing layer, motion artifacts introduced by motion related impedance changes, and electric noise.

All the above method needs electrodes to be in contact with body in some or the other way. (gel, dry contact etc). To avoid all the problems mentioned above when the electrode are intact to body the team propose to have capacitive measurement of ECG signal. This will help in reducing all the above problems. Also the team are transmitting the data wirelessly and processing it which will help in analyzing the signal better and also the wireless transmission will give more freedom to users.

III. Concept Development

The goal of our project was to build a prototype model using non-skin-contact capacitive electrodes for heart monitoring of first responders like fire fighters. The team found that this approach of non-skin-contact capacitive electrode is highly prone to noise from mains and even a small movement can introduce excessive EMG noise. To build the circuit which satisfies above goal the team need high precision when designing both the software and hardware portions of model. The team's capacitive electrode design gave reasonable results if the patient remained stationary, minimizing induced EMG noise.

To build the model based on a non-skin-contact capacitive topology the team needed to satisfy two basic criteria:

- There should not be direct contact between the body and the electrodes.
- A virtual capacitor must be formed between body and circuit to acquire ECG signal.

To fulfill first condition, the team covered our electrodes with nylon tape so that they were not in direct contact with the body. The second requirement was met in the way that the body acted as one plate of the capacitor, with the tape which behaved as a dielectric and the electrodes as second plate of capacitor, thus forming a virtual capacitor between body and circuit and enabling us to get the required signal.

The model that the team built broadly consist of four major part namely, 1) Electrodes, 2) Hardware unit, 3) Interfacing unit between Hardware and Software, 4) Software unit as shown in block diagram below.

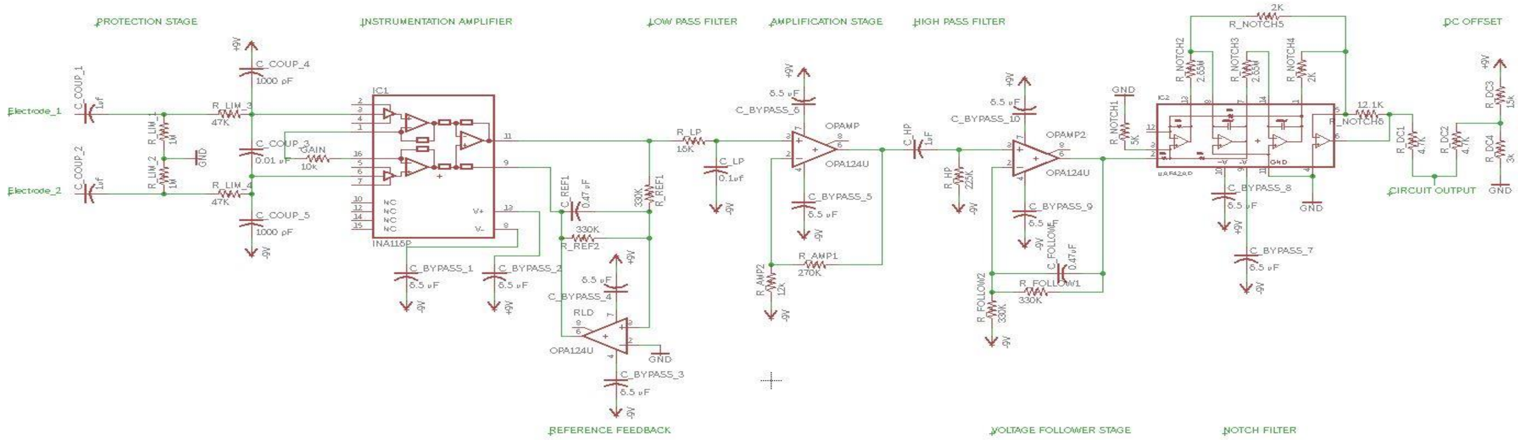

Figure 5: Circuit Schematic

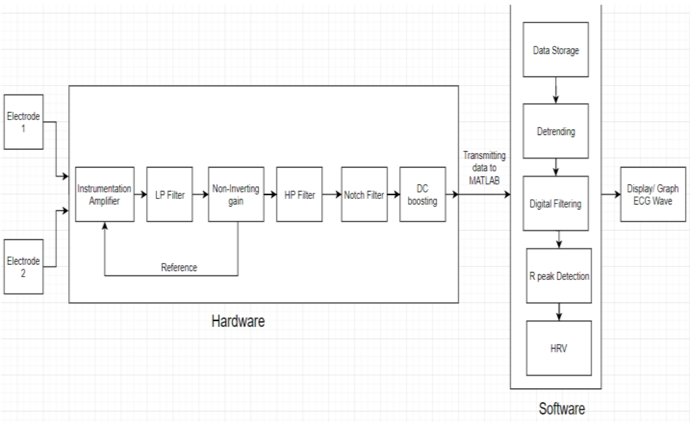

Figure 4: Block Diagram of System Architecture

The electrodes that the team used are the simple dual side copper PCB board cutouts. The shape of the electrode used was square as it will allow as to have more surface area and this will help in better acquisition of ECG signal from body.

The hardware unit which is one of the most vital part of model consist of instrumentation amplifier (INA) along with filtering and gain stages followed by DC boosting at end. The team went through number of papers, tried and implemented number of designs given in [1,8,9,11-13,16,19-21] and finally settled with the design given in [23]. The INA with high input impedance helps in getting signal into the circuit. The high pass and low pass filter together form bandpass filter with cutoff frequency 0.5 Hz-120Hz. The notch filter removes AC mains noise and DC booster shifts the signal by constant amount which helps in preserving the entire signal (negative part as well) and makes it easy to do software processing on signal.

On software side, once the signal is received in MATLAB the team store and detrend the signal to remove baseline drift. The the team do digital filtering of signal to make signal more clean and free from noise. The team then apply algorithm to detect R-peaks and Heart Rate Variability (HRV) and finally display the clean ECG signal.

IV. DESIGN

A. *Circuit*

The basic design of any ECG circuit has three stages:

1. Instrumentation Amplifier
2. Filtering
3. Amplification

An instrumentation amplifier measures the difference across two potentials and amplifies the difference between these two potentials. Placing electrodes some distance apart across the human body provides two reference voltages that the instrumentation amplifier can compare to generate an ECG signal. The output ECG waveform from an instrumentation amplifier can be susceptive to EMG noise from movement, common mode noise, mains power noise, electronic noise, and noise from other sources. To attenuate the noise from these sources, most ECG circuits are also equipped with filtering stages that usually consist of some bandpass region. The ECG waveform is on the scale of microvolts to millivolts and therefore the final stage is an amplification stage that boosts the voltage to a desirable level for analysis equipment.

In designing the circuit, the team first conducted a literature review of existing capacitive ECG circuits [1,8,9,11-13,16,19-21]. The team discovered that the basic circuit design of a capacitive and a gel ECG circuit is the same, the differences between the two devices comes primarily from the electrode design and special care in the instrumentation amplifier parameters. The team used a Cornell ECG circuit topology as a template but made improvements and adjustments such as a notch filter circuit, a DC offset circuit, and tuning the parameters of the design [23]. The team's design of the capacitive ECG circuit can be divided into 9 distance stages as follows:

1. Instrumentation Amplifier
2. Reference Feedback
3. Low Pass Filter
4. High Pass Filter
5. Amplification Stage
6. Protection Stage
7. Voltage Follower Stage
8. Notch Filter
9. DC Offset

Figure 5 shows these stages and their implemented parameters. The operation and design of the stages is described below.

1. Instrumentation Amplifier

For a capacitive electrode ECG device, the instrumentation amplifier must have a large input impedance. Therefore, the INA116 was chosen to fulfill the instrumentation amplifier stage. The INA116 provides a high input impedance, ultra-low input bias current, and provides shielding pins that can be used to reduce common mode noise. The INA116 is an all-inclusive instrumentation amplifier package so the only design for this stage was to set the gain of the stage. The gain of the INA116 is given by Equation 1. The resistor Rg was chosen to be 10k ohm, giving the instrumentation amplifier a gain of 51.

$$G = 1 + \frac{50k\Omega}{R_G} \tag{1}$$

2. Reference Feedback

The INA116 has a reference pin that can be used to provide a DC reference level. In ECG circuits, typically an RLD circuit is provided to this reference pin, however the team instead designed a protection stage so the RLD was not necessary. Instead, the reference pin is connected to a feedback loop that normalizes the output of the INA116 about a 0-voltage threshold.

3. Low Pass Filter

To minimize the high frequency noise a low pass filter was designed to operate about the ECG high passband frequency of 100Hz. Initially the team designed an active low pass filter using an operational amplifier, however it was determined that the performance between an active and passive filter has minimal difference. With the performances being comparable, the team chose to use a passive low pass filter design to save on power since the circuit is battery operated. Equation 2 describes the cutoff frequency for the low pass filter. The resistor and capacitor values were chosen as 15.9k ohm and 0.1u Farad respectively, given a designed cutoff frequency of 99.47Hz

$$Fc = \frac{1}{2\pi RC} \tag{2}$$

4. High Pass Filter

The high pass filter is primarily used to block DC noise from the circuit. The same design considerations were made with the high pass filter as described in the low pass filter section. Equation 2 also describes the cutoff frequency for the high pass filter. With the designed resistor and capacitor values of 225k ohm and 1u Farad, the cutoff frequency was designed to be 0.707Hz.

5. Amplification Stage

The INA116 was designed to provide a gain of 51, however this gain is not sufficient in setting the ECG peak to peak voltages to align nicely with a 0-5V analog to digital converter that the team chose to use. For a millivolt ADC signal, the team desired a total gain of around 1000 to push the signal into the volt range. To achieve this gain, the team designed a non-inverting amplifier. The gain of a noninverting amplifier is described by equation 3.

$$Av = 1 + \frac{R_2}{R_1} \tag{3}$$

The team chose R2 to be 270k ohm and R1 to be 12k ohm resulting in a gain for this stage to be 23.5. The total gain is the multiplication of the gain from the two stages or 1198.5. With a gain of ~1200 it was observed that from person to person the peak to peak voltage of the ECG signal was between 2V-4V which falls within the ADC range.

6. Protection Stage

When connecting any electronics to the human body, especially across the heart, it is of the ut-most importance to ensure the current that flows through the human body a microamp or below to avoid harming the patient. The protection stage circuit has two portions to help address the concern. The first is that the input stage has a current return path to ground through large resistors that will 1M ohm resistors in case any faults occur. The protection stage is also connected to the input of the INA116 which only accepts low input bias current and also has a high input impedance. Additionally, the protection stage has two large capacitors connected directly to the electrodes which apply AC coupling and block the DC offset of the electrode signals.

7. Voltage Follower Stage

The output of the amplification stage is passed to a voltage follower. The voltage follower is used to track the voltage levels of the ECG signal and amplifies the power of the signal. A voltage follower has a unity gain, so the voltage will not change, but the current will increase as a result of a low output impedance of the voltage follower op amp.

8. Notch Filter

Initial circuit tests showed a prevalence of 50-60Hz noise induced from mains. To attenuate this noise, a UAF42 was configured in a band-reject configuration as specified by the TI document “Design a 60Hz Notch Filter with the UAF42.” From the document the 3dB points are shown to be 54Hz and 66Hz with a -60dB attenuation around 60Hz.

9. DC Offset

The circuit was designed with two 9V batteries so that both positive and negative voltages could be used to power the operational amplifiers. An ECG signal swings both in the positive and negative regions, but the ADC that the team used can only accept voltages in the range of 0-5V. The team's measured and amplified average signal was around 3V peak-peak with around 2V above the reference and 1V below the reference. Therefore, a DC offset circuit was designed around 1.5V. The DC offset circuit works by combining the AC ECG signal with a constant DC voltage extracted from a voltage divider. Equation 4 shows the output voltage between the two resistors of the voltage divider circuit. With R1 = 15k ohm and R2 = 3k ohm and a voltage supply of 9V, the DC offset is calculated to be 1.5V.

$$Vdc = Vcc \times \frac{R_2}{R_1 + R_2} \tag{4}$$

B. Electrodes

To design capacitive electrodes, a virtual capacitor must be created with the human body. The coupling between the ECG signal and the circuit is directly related to the capacitance of the virtual capacitor. The capacitance of a capacitor is given by equation 5 where epsilon is the dielectric constant of the conductors, d is the separation between the two plates, and A is the surface area of the conductive plates. It is observed from equation 5 that for a large capacitance, a designer wants to minimize the spacing between the two conductors and maximize the dielectric constant and surface area. For a capacitive ECG circuit, the skin of the patient acts as one conductor and a metal plate acts as another conductor. An insulator is wrapped around the conductive plate and acts as the dielectric.

$$C = \frac{\varepsilon A}{d} \tag{5}$$

The electrode design was an iterative process for the team. Through research of dielectric materials, it was found that the best readily available materials with high dielectrics are glass and paper. The problem with using these two materials is glass is difficult to cut and there are safety concerns and paper is susceptible to being damaged by sweat and other natural elements. The next best commonly available dielectrics were different types of tape. The team tested with duct tape, electrical tape, and nylon tape. Nylon tape has the highest dielectric and resulted in the most coupling.

There is a tradeoff between maximizing the size of the metal conductor, patient comfort, and stability. As the conductor increases in size, the surface area increases and the capacitance increases, but mounting the electrode correctly also increases in difficulty. As the electrode gets larger it is more susceptive to the muscle movements of the patient which will introduce EMG noise to the circuit. After conducting a literature review the team found that the largest conductors used were 36 squared centimeters and the smallest was the size of a penny. Therefore, the team constructed a variety of electrodes as shown in Figure 6 and found a balance between conductivity and size with 25 squared cm square electrodes. The prototyping section will further explain the shapes and types of electrodes tested.

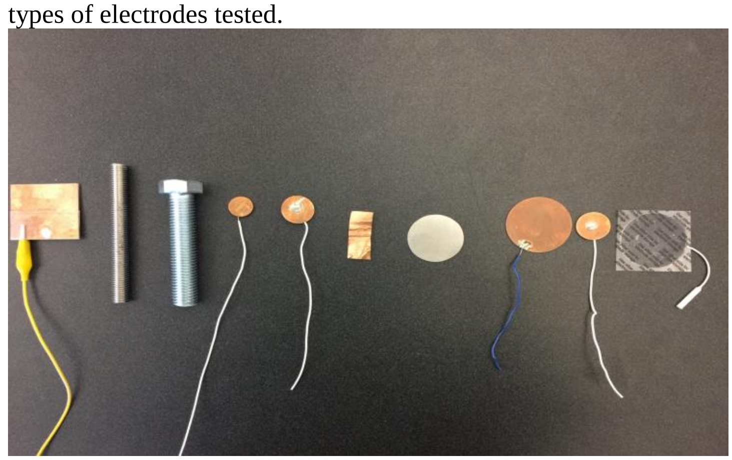

Figure 6: Sample of Electrodes Tested

C. Software Design

The control flow of the device's software is depicted in **Figure A.**

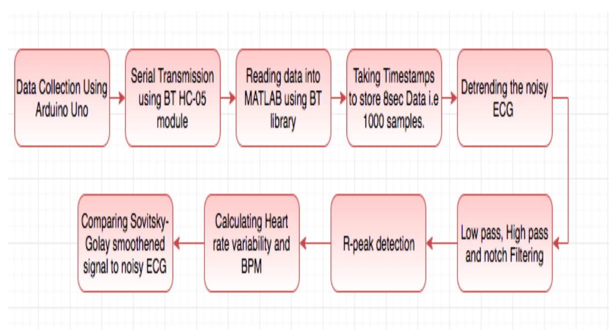

Figure 7: Control flow of the device's software.

The ECG signal is collected using an Arduino Uno microcontroller. Once digitized and converted to millivolts, the signal is passed to Matlab wirelessly, via Bluetooth serial, at a baud rate of 115200. In Matlab, the data is time-stamped using the system clock and stored in an array of predefined size corresponding to eight seconds of time.

Once digitized, converted to millivolts, and time-stamped, signal data is passed to Matlab wirelessly, via Bluetooth serial, at a baud rate of 115200. In Matlab, the serial data is stored in an array of predefined size, then split into voltages and associated time values. The time-stamping is done on the Arduino, pre-serialization, to avoid timescale corruption due to serial latency.

Once collected, an initial component of the signal always features transient discontinuities: sharp peaks that, if left in the signal, would interfere with signal

processing. To avoid this, the initial fraction of the signal containing these transient behaviors is removed from the data array.

First, the ECG signal is detrended to remove any baseline drift and the DC offset required by the Arduino's ADC. This is accomplished by fitting a low-order polynomial to the signal via linear regression, then subtracting this polynomial approximation from the actual signal. The low-order polynomial drift of the baseline is removed while preserving the higher-order shape of the signal [

After detrending, the signal is passed through a series of 2nd-order digital IIR filters using functions provided by Matlab, for the same purposes as described in the Design portion of Hardware:

- A low-pass filter with passband frequency 20 Hz and sampling rate 150 Hz, to remove high-frequency noise;
- A high-pass filter with passband frequency 0.5 Hz and sampling rate 20 Hz, to remove EMG and DC noise; and
- A Butterworth band-stop (notch) filter with half-power frequencies 59 and 61 Hz and sampling rate 1000 Hz, to remove AC interference.

The frequency responses of the above IIR filters are shown in **Figure B.**

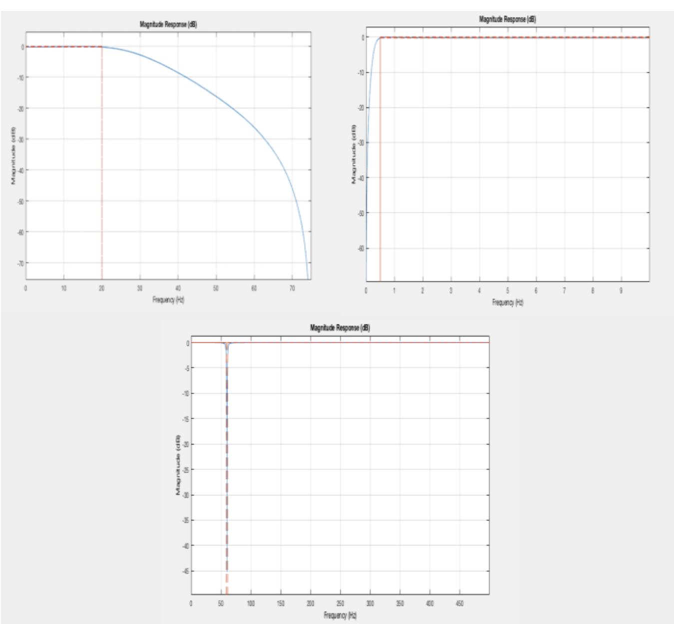

Figure 8: Frequency responses for, clockwise from top-left: the low-pass, high-pass, and band-stop (notch) IIR filters.

After filtration, the signal is searched for R-peaks. First, the highest-value sample in the signal is designated as the "highest" R-peak. All samples close enough to that value (i.e. only a certain amount lower) are classified as the other R-peaks. Any R-peaks closer than a minimum distance to a previously-detected peak are then removed as misclassifications.

After filtration, the signal is searched for R-peaks. This is accomplished by locating all peaks above a certain predefined threshold (150 mV) and a certain minimum distance from each other (0.65 seconds).

The number of R-peaks detected in the signal allows for the estimation of beats per minute (BPM), and the knowledge of their occurences in time allows for the calculation of heart-rate variability (HRV).

To calculate beats per minute, the R-peak count is divided by the time-length between the first and last detected peaks in seconds, to yield beats per second, and then multiplied by 60 seconds to yield beats per minute:

$$BPM = \frac{number\ of\ peaks\ in\ signal}{time\ between\ first\ and\ last\ peaks} \qquad (6)$$

To calculate heart-rate variability, the time-difference between peaks is calculated for each set of two peaks detected:

$$\Delta t_{ij} = t_j - t_i \qquad \text{for peaks } i, j \qquad (7)$$

These differences are then plotted over time.

ECG signals benefit from the use of smoothing filters, which use a moving average of the signal to remove noise without sacrificing the signal shape. To improve final visualization, a Savitzky-Golay smoothing filter, of order 4 and window size 21, is applied to the signal to reduce noise while preserving the overall signal shape.

D. *Harness Design*

The aim of this project is to make the device portable. This can be done by using a harness. For this the team are using a velcro strap on which the electrodes will be placed. A velcro will be used to strap it to the body. This will be diagonally placed across the person's body.

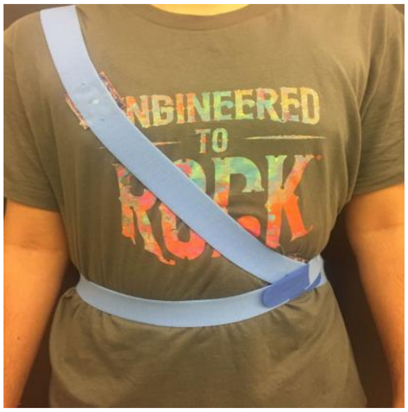

Figure 9: Harness Placement

Electrode placement is of critical importance because the ECG wave shape changes in the sense that the amplitude of R and T peaks varies as distance between electrodes changes. This happens as the lead from which the resultant wave is obtained varies. Optimum R-peaks are obtained when electrode placement is that of Lead III. The circuit and wireless module assembly will be enclosed in a case.

V. PROTOTYPING

A. Circuit and Electrodes

The circuit and electrode design were an iterative process for the team. The team constructed and tested the topologies in [1,8,9,11-13,16,19-21]. For every single topology tested no distinguishable ECG signal could be identified and the waveforms were always covered in noise. With multiple circuits built, the team believed that the problem must be the electrodes rather than the circuits as each circuit performed nearly identically. The team therefore built a variety of electrodes as shown in Figure 2 with multiple shapes, materials, and insulators. Copper, steel, zinc, and stainless steel were all tested as conductors. Duct tape, electrical tape, and nylon tape were tested as insulators. The team was still unable to acquire an ECG signal. The team then constructed a dry-contact electrode circuit and tried the steel bar and zinc bolts as conductors and was once again unable to get an ECG signal.

After consulting with an advisor, it was noted that the lab space was an extremely noisy environment. Moving the electrodes and circuits away from mains outlets and other electronics resulted in an ECG waveform becoming acquirable. Further testing the circuits in a home environment with much less noise resulted in even cleaner waveforms. The team concluded that since the circuit capacitively couples, it must be used in low noise environments. Once the team realized the circuits do work, but only in low noise environments, the team tested all the topologies in [X-Y] again with each type of electrode and insulator and found that the clearest ECG signal was acquired with 25 squared cm square copper electrodes, nylon tape, and the topology described in the design procedure section.

The electrodes positioning was changed to observe the impact of the signal.

The circuit was constructed on a breadboard as shown in Figure 10. To help reduce mains noise the through hole components were trimmed and the wires were shortened when possible. Inductive loops can further be reduced by using a printable circuit board with surface mount components.

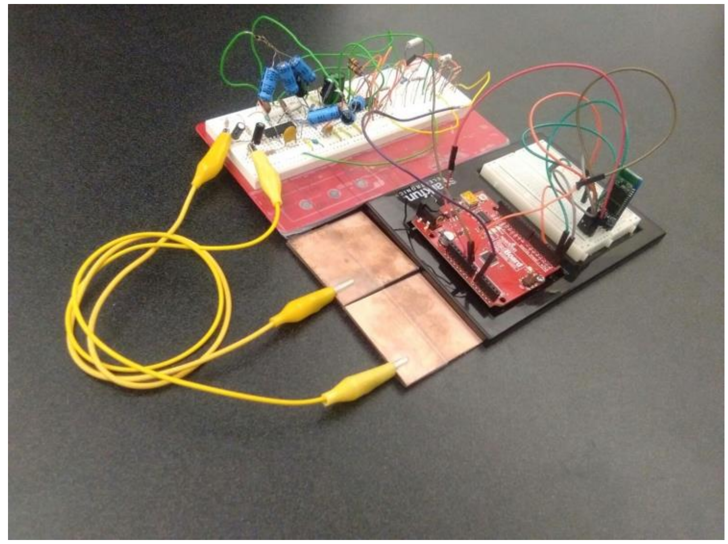

Figure 10: Picture of Circuit on Breadboard with Bluetooth Modules Connected

While constructing the circuits, the filters remained constant between all topologies. The filters were tested by using an Analog Discovery to generate the bode plots of the frequency response of the filters. Figure 11 displays the Bode plot for the high pass and low pass filters and Figure 12 shows the Bode plot for the Notch filter. The x-axis is frequencies swept, and the y axis is the voltage amplitude in dB relative to the input signal. The yellow line is the input signal and the blue line is the output signal relative to the input signal. It is observed that the low pass filter has a tested cutoff frequency of 99.75Hz and the high pass filter has a tested cutoff frequency of 0.705Hz with a 2dB attenuation thereafter. The designed cutoff frequencies were 99.47Hz and 0.707Hz respectively so the bandpass region is extremely close to the designed frequencies. The notch filter tested cutoff frequencies are 54.3Hz and 63.64Hz with a -20dB attenuation about 60Hz.

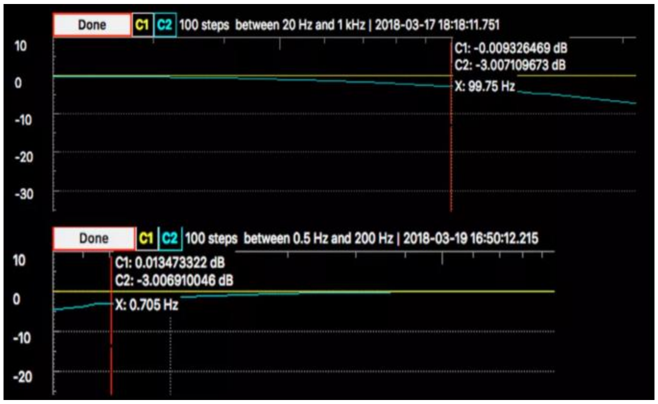

Figure 11: Low Pass Filter (top) & High Pass Filter (bottom) Bode Plots

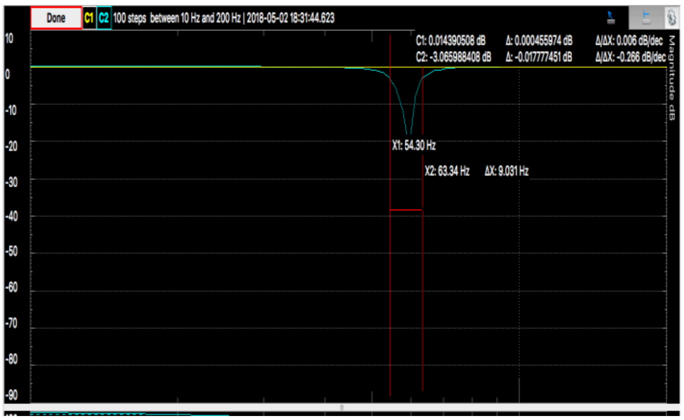

Figure 12: Notch Filter Bode Plots

B. *Software Simulation*

1. Signal Collection

Initial tests of signal collection and transmission via Bluetooth on cardiac signals simulated by function generators yielded sub-par results. Chiefly, ECG peaks suffered from significant sinusoidal windowing, sometimes leading to peaks being entirely removed from the recorded signal. This phenomenon is shown in Figure 13.

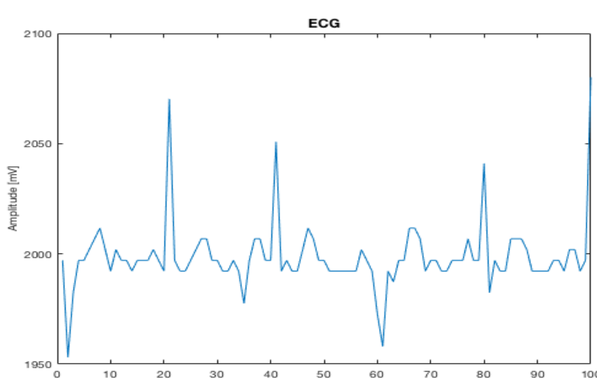

Figure 13: A recorded simulated ECG signal, featuring windowing of the ECG peaks. A third peak between samples 50 and 70 has been obscured entirely.

To avoid this, a delay was introduced to the Arduino's data-collection loop in the range recommended for ECG signals (in this case, 5 milliseconds). This helped prevent signal saturation in the serial transmission.

A variety of optimizations were implemented to improve Bluetooth serial transmission, including moving from a software serial library to the default serial library using the Arduino's TX and RX lines; scaling recorded values to mV to avoid sending floating-point values over serial; and increasing the serial baud rate from the default 9600 to 115200. On the MatLab side, the data array was changed from dynamically sized to of a predefined size, to improve execution time.

To prevent serial latency and buffer behavior from obfuscating the true timescale of the signal, time-stamping was added to the Arduino code, for pre-serialization time measurement.

2. Signal Processing

The signal processing pipeline was tested on simulated ECG signals created using the ECGSYN realistic ECG waveform generator [22]. ECG signals were generated with noise of predefined frequencies. Results of the signal processing pipeline on simulated ECG signals, including detrending, IIR filtering, and smoothing, are shown in Figure 14.

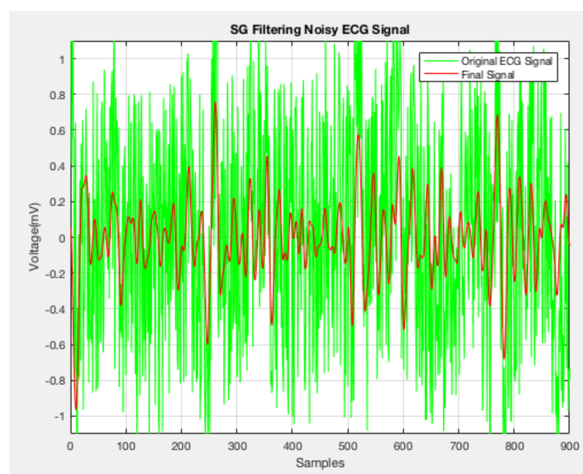

Figure 14: digitally simulated noisy ECG signal after detrending, filtration, and smoothing.

HRV calculation was performed for the simulated ECG signals. However, R-peak detection was not optimal, leading to missed peaks. Missing peaks dramatically increased the distance between neighboring peaks, creating spikes in the HRV plot. This phenomenon is shown in Figure 15.

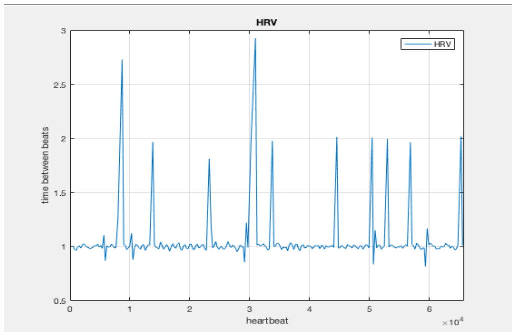

Figure 15: HRV plot from a signal with missing (undetected) R-peaks.

One R-peak thresholding was adjusted properly, the HRV plot improved. A more accurate HRV plot is shown in Figure 16, featuring a relatively constant one-second interval between beats for a simulated 1Hz ECG signal**.**

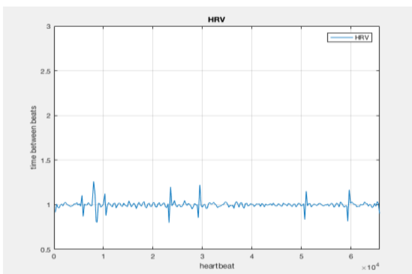

Figure 16: HRV plot from a signal with no missing R-peaks.

Of course, continuous-line graphs of HRV aren't physically intuitive. These were replaced with discrete points.

C. Zigbee

Zigbee communication is specially built for control and sensor networks on IEEE 802.15.4 standard. These operate at 868 MHz, 902-928MHz and 2.4 GHz frequencies. The date rate of 250 kbps is best suited for periodic as well as intermediate two way transmission of data between sensors and controllers.
Zigbee is low-cost and low-powered with 10-100 metres range. This is simpler than the short-range wireless sensor networks as Bluetooth and Wi-Fi.

Figure 17: A pair of Xbee radios (through-hole with the wire whip antenna type)

1. <u>Network Architecture Xbee:</u>

The basic network architecture of xbee is as shown in the figure below:

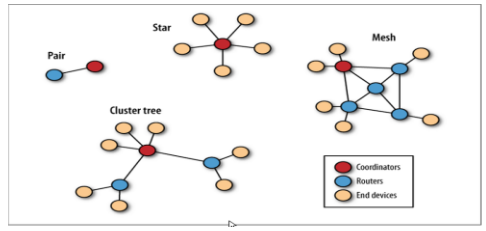

Figure 18: Types of Zigbee Network [26]

Zigbee structure consists of the following:

- Routers:A router is an FFD. A router is used in tree and mesh topologies to expand network coverage. The function of a router is to find the best route to the destination over which to transfer a message. A router performs all functions similar to a coordinator except the establishing of a network.
- Coordinators: A coordinator is an FFD and responsible for overall network management. Each network has exactly one coordinator.
- End device: An end device can be an RFD. An RFD operates within a limited set of the IEEE 802.15.4 MAC layer, enabling it to consume less power. The end device (child) can be connected to a router or coordinator (parent).[24]

2. <u>Xbee setup:</u>

The modem configuration tab in X-CTU software was opened after device selection and the firmware of the device was updated.

To set up the XBee to be a router, the following modifications were made:
PAN ID: 1234
JV channel verification: Enabled [1]
NI Node Identifier: Router
AP API Enable : API Enabled [1]

To set up the XBee to be a coordinator, the following modifications were made:
PAN ID: 1234
JV channel verification: Disabled [0]
NI Node Identifier: Coordinator
AP API Enable : API Enabled [1] [25]

In spite of all the advantages of Zigbees, the team were unable to use them in our project for the following reasons:

- Initially both Zigbees were configured and transmitting. While reconnecting the test circuit (xbee and a ECG circuit) in Matlab an error was given that transmitter was not sending data.

- For solving this issue, a default reset of the router was done. The router disappeared from XCTU after this and it was not to detected by the software.
- After contacting, Digi International, Xbee manufacturers, they came to a conclusion that this was a hardware issue which would take due time to troubleshoot.

This is why the team moved to our next best alternative, a bluetooth HC-05 module.

D. Bluetooth

The ECG signal is collected using an Arduino Uno microcontroller. The Uno is connected via TX and RX lines to an HC-05 Bluetooth module for serial Bluetooth transmission. The HC-05's state pin is used to determine when a successful connection to MATLAB has been achieved.

VI. RESULTS

A. Analog Circuit Results

The measured waveforms from the designed circuit are shown in Figures 19-22. Plot 19 and 20 show the output signal of the ECG in a low noise environment across the chest. The electrodes were positioned about 1 inch across vertically across the heart. This positioning was not used for the final signal because the R peaks were smaller in amplitude than the T waves due to the positioning of the electrodes. It can be observed that placing electrodes nearer to the heart help reduce noise, especially EMG from movement in the hands.

Figures 21 and 22 show the output signal from holding the electrodes in the hand. It can be observed that significant noise in induced from movement, but the location of the electrodes allows for defined R peaks to be detected. The ECG signal acquired from the hands was then passed to Matlab for digital signal processing.

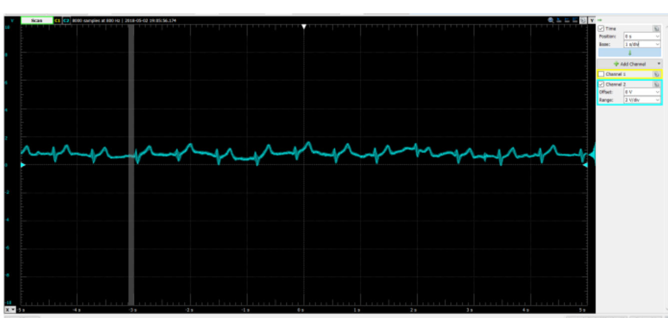

Figure 19: Output Waveform with Electrodes Across Chest

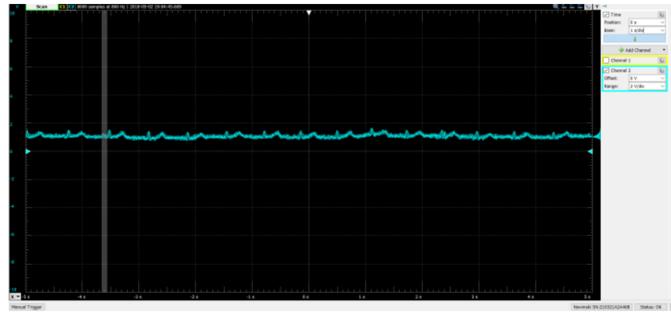

Figure 20: Output Waveform with Electrodes Across Chest + DC offset

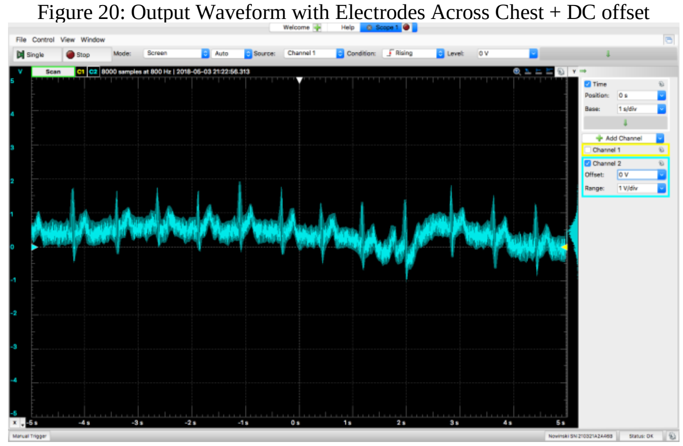

Figure 21: Output Waveform with Electrodes in Hands

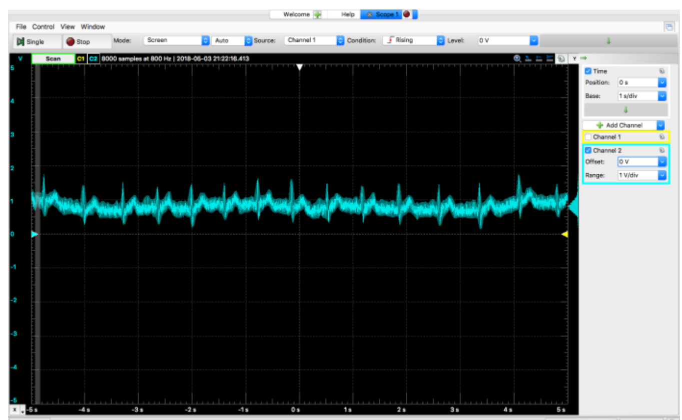

Figure 22: Output Waveform with Electrodes in Hands + DC offset

B. Digital Processing Results

In this section,all the graphs that were generated by the MATLAB code on an actual ECG signal are displayed. An arduino connection was made with the hardware circuit to collect the signals which is being recorded by the person placing the electrodes on his chest or hands and this collected data is being transmitted by the bluetooth HC-05 module to the computer.

STEP 1: Record original Noisy ECG signal using Bluetooth Configuration in MATLAB.

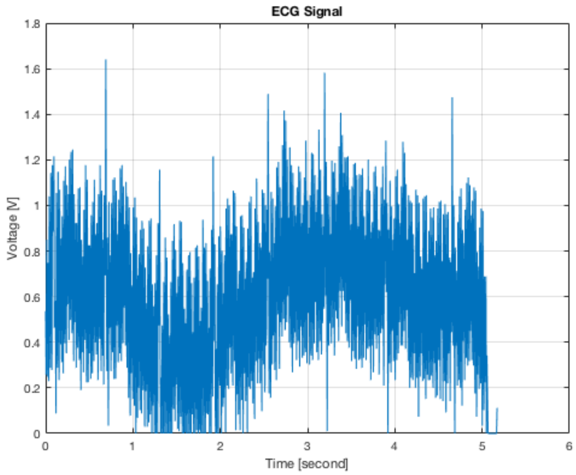

Figure 23: The Original ECG signal from Hardware Circuit

Note: Discontinuity in first second of recorded data. In many cases, this discontinuous peak was much larger than all of the actual R-peaks, leading to incorrect R-peak detection. Thus, the first second of transient data was removed for each recording.

STEP 2: Detrending the recorded signal to fix the problem of baseline wander using a simple polynomial fit method.

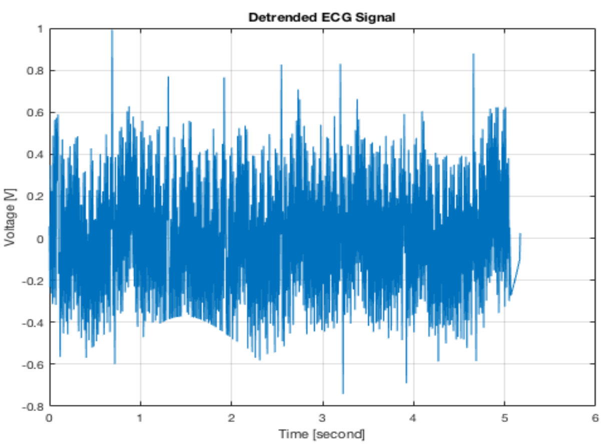

Figure 24: The signal after Detrending the original signal.

STEP 3: Using an IIR Low pass filter, removed all the high frequency noises from the signal which are found mostly above 100Hz. After LP filter the next step is to use an IIR High pass filter to remove the low frequency noises usually found below a value of 0.7 Hz. Also use a notch filter to eliminate the 60Hz Mains power Supply interference.

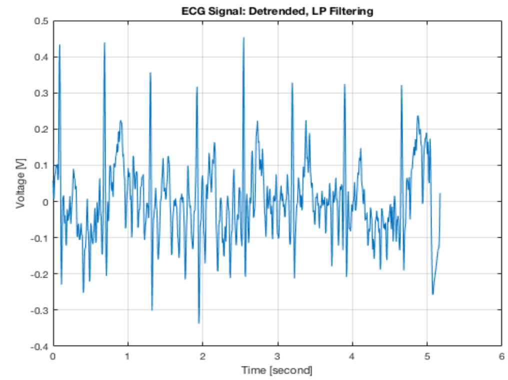

Figure 25: The signal after Detrending and LP filter

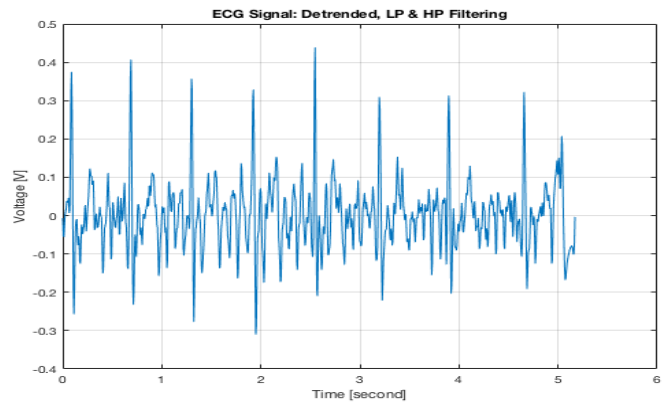

Figure 26: The signal after detrending, LP and HP filtering

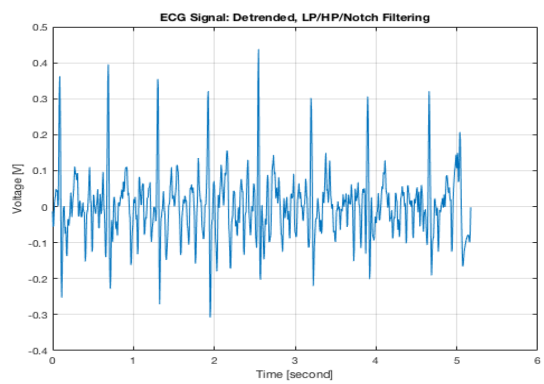

Figure 27: The signal after detrending, LP, HP and Notch filtering

STEP 4: After all the filtering is done, using the peak detection algorithm the R-peaks were obtained from the signal. It detected almost all the peaks correctly except for one since two of them appeared to be really close as seen

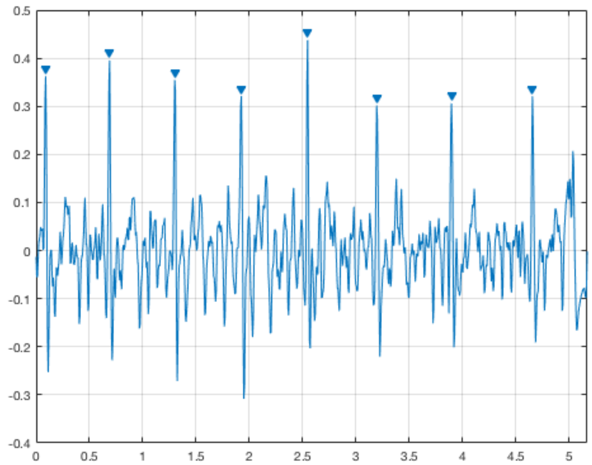

Figure 28: The R-peak detection

STEP 5: Once the count of the R- peaks is acquired , the calculation of beats per minute (BPM) is carried out along with the graph for the heart rate variability (HRV). The results are shown below.

```
bpm =

  105.0826
```

Figure 29: Calculated BPM

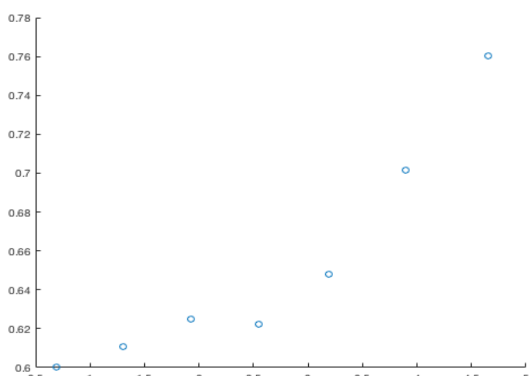

Figure 30: The HRV Plot

STEP 6: As an additional step the team decided to use a Savitsky-Golay filter to smoothen out the signal further and compared it to the original noisy signal.

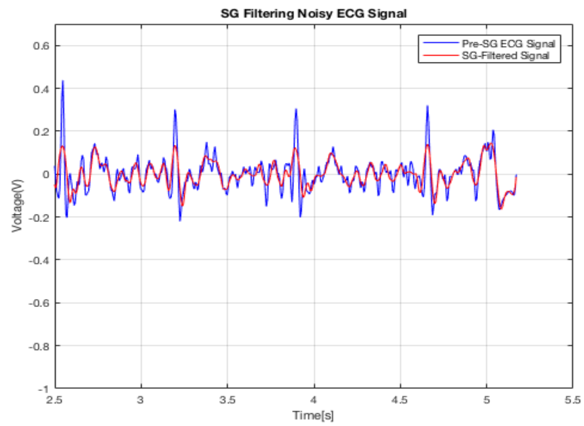

Figure 31: Post SG Filtering

VII. FUTURE SCOPE

The team was able to successfully construct a capacitive ECG circuit. The circuit is able to successfully couple with the human body to detect and record the ECG waveform. The circuit is extremely susceptible to noisy environments which can deteriorate the quality of the received ECG signal and even completely mask its presence. It is of the upmost importance to make sure no electronics are near or around the patient of the circuit. A number of future improvements can be made to further improve the quality of the circuit as follows:

<u>Printable Circuit Board:</u> Using a PCB with surface mount components will reduce the inductive paths and therefore coupling with noisy sources. A PCB will also reduce the size and complexity of trouble shooting. The entire circuit can be fit on a single side 25 square cm electrode so only one connection would need to be made connecting the second electrode to the circuit. The other side of the PCB could be used as the electrode with the insulator added further minimizing the size.

<u>Improved Shielding:</u> The INA116 has guard pins that can be connected to a constructed shield that will reduce common mode noise. Additionally, the circuit could be constructed within a Faraday Cage with only the Bluetooth portion and electrodes being outside of the cage. This design would complicate the aforementioned PCB design – the electrodes and circuit would have to be separated. The cables connected to the electrodes could also be shielded to reduce noise coupling.

<u>Q and S detection:</u> Currently the code can only detect R peaks. If the Q and S waves could also be detected the heart rate variability and beats per minute could have additional accuracy checks in case there is an error in peak detection.

<u>Filtering:</u> The software filters can be improved by adding a sliding frequency range for all of the filters, so the filters can be optimized in real time for a read signal.

Spin Coating Electrodes: The electrodes could be spin coated with a dielectric which would allow for a large dielectric to be coated with a minimum separation distance resulting in an increase of capacitance.

Harness Improvements: The harness is currently static. The harness can be improved by adding adjustable electrode positions and placement patterns.

The team realizes after completing this project that the main challenges with designing a capacitive electrode ECG device are noise considerations. Attenuating the noise in a capacitive ECG circuit increases design complexity and cost which is why this topology is rarely used in practice.

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Appendix

Code: https://github.com/lychrel/cap-electrodes