Using Photoplethysmography and Electrocardiography to Monitor Vitals for Patients with Arrhythmia and Hypertension

Cole Wener
Department of Bioengineering
University of California, San Diego

Mohak Patel
Department of Bioengineering
University of California, San Diego

John-David Russo
Department of Electrical and Computer Engineering
University of California, San Diego

Zaira Adrianwala
Department of Bioengineering
University of California, San Diego

***Abstract*— Monitoring vitals of patients with diagnosed arrhythmia, hypertension, and other cardiovascular diseases is very important. This is because it can help the care provider insights into the health condition of the patients and prescribe the appropriate treatments needed. Currently there are devices in use by the healthcare industry that are not very mobile, and need a separate technician to operate the devices. Along with this, there are no devices that will measure vitals such as heart rate, heart rate variability, ECG, and estimate blood pressure all through one device. Oftentimes, the care provider has to use separate devices to do this such as a PPG/Pulse Oximeter, ECG machine, and a Sphygmomanometer. Therefore, the main goal of this project is to measure all these vitals from one device. The set up primarily consists of a 3 lead ECG device, a transmissivity dependent PPG sensor cuff, a DAQ, and a computer with MATLAB computing capabilities. The project design includes receiving LED light signals from the PPG sensor cuff, voltage biosignals from 3 leads placed on the body for ECG, and finally processing the signals from both devices with the help of a MATLAB algorithm to determine vitals of interest.**

Introduction

Monitoring vitals signals such as heart rate (HR), heart rate variability (HRV), and blood pressure (BP) is crucial for the effective management and treatments of patients with cardiovascular diseases such as Arrhythmia, hypertension, and other related disorders. These conditions pose significant challenges to both patients and healthcare providers due to their potential to cause life-threatening complications if not managed appropriately. Thus, the development of biomedical devices capable of accurately measuring these vital signs is imperative to enhance the quality of care and improve patient outcomes. Arrhythmia is characterized by irregular heart rhythms which can lead to serious complications such a stroke, heart failure, and sudden cardiac arrest [1]. Lifewise, hypertension is a condition characterized by elevated blood pressure level, and is a major risk factor for cardiovascular events such as heart attacks and strokes [2]. Patients with these conditions require regular monitoring of their vital signs to detect any abnormalities or changes promptly, enabling timely interventions and adjustments to their treatment plans.

The advancements in biomedical technology, particularly in electrocardiography (ECG) and photoplethysmography (PPG) have revolutionized healthcare by offering non-invasive and continuous monitoring solutions [11]. ECG technology records the electrical activity of the heart, allowing for the detection of abnormalities in heart rhythm and rate [3]. PPG technology, on the other hand, measures changes in blood volume in peripheral blood vessels, providing valuable insights into parameters such as heart rate and blood pressure [4]. Combining these technologies in a single device can offer us a promising approach to monitoring multiple vital signs simultaneously, enhancing diagnostic capabilities,

and enabling personalized treatment strategies for patients with cardiovascular diseases. [5, 7-9]

In this paper, we present the circuit and algorithm design of a novel biomedical device capable of measuring vitals signs such as heart rate, HRV, and estimating blood pressure using ECG and PPG technology. By leveraging the capabilities of these advanced technologies, our device aims to address the critical need for accurate and continuous monitoring of patients with Arrhythmia, Hypertension, and other cardiovascular disorders, ultimately improving their overall quality of life and clinical outcomes.

II. Physiology

The purpose of this system is to monitor patients with heart arrhythmias and hypertension. A thorough understanding of the indicator locations and measurement methods of heart rate (HR), heart rate variability (HRV) and blood pressure (BP) is essential prior to system development.

A. Heart Rate (HR) and Heart Rate Variability (HRV)

Heart rate is measured through the pulse rate, which can be found by measuring volume variations in the blood. This is done using the AC signal that comes from shining LED light through the finger into the tissue. Heart rate is calculated as the number of contractions per minute, and corresponds to variation in blood vessel volume.

$$HR = \frac{60}{mean(R_{pp})} \quad (1)$$

Equation (1) above shows how heart rate is calculated by dividing 1 minute (60 seconds) by the mean of the heart rate contractions, shown here by mean (R_{pp}). R_{pp} is the time between successive R peaks [6].

Subsequently, heart rate variability can also be found using the number of contractions per minute. Unlike the heart rate calculation, however, heart rate variability does not take into account the mean of the contractions.

$$HRV = \frac{60}{R_{pp}} \quad (2)$$

Equation (2) shows how similar to heart rate, heart rate variability is derived from dividing 1 minute (60 seconds) by time between successive R peaks, shown in Eq. (2) by R_{pp}.

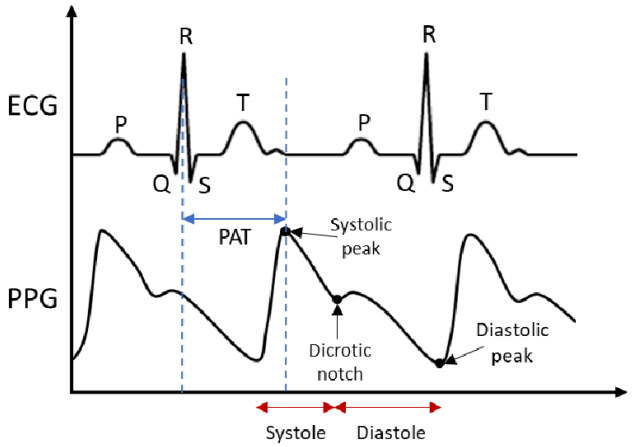

Figure 1: Graphical representation of the relationship between ECG and PPG known as PAT [7].

B. Blood Pressure (BP)

Blood pressure can be estimated using the calculated heart rate, as well as pulse arrival time (PAT). Pulse arrival time is the time difference between the arrival of a pulse wave compared to the R peak. This is a reflection of the amount of time it takes for the pulse wave generated by the heart's contraction to propagate to the point of measurement. A graphical representation of PAT can be seen in Figure 1 above.

$$PAT = t_{PPG\ peak} - t_{ECG\ peak} \quad (3)$$

Equation (3) shows this pulse arrival time difference, with $t_{PPG\ peak}$ being the arrival pulse wave on the PPG, and $t_{ECG\ peak}$ being the R peak as seen on the ECG.

This PAT value can now be used in conjunction with the heart rate to find both systolic and diastolic blood pressure. Systolic blood pressure is the maximum blood pressure during contraction, while diastolic is the minimum blood pressure just prior to the next contraction.

$$t = \frac{60000}{HR - PAT} \quad (4)$$

Equation (4) describes the time interval (in ms) used to measure systolic and diastolic blood pressure. Heart rate (HR) and pulse arrival time (PAT) can be found using Eqs. (1) and (3). Since heart rate is measured in beats per minute, 60000 is used to convert from minutes to milliseconds. This gives a value for the time duration between heartbeats [5].

$$BP_{sys} = 184.3 * 1.329 * HR + 0.0848 * t \quad (5)$$

$$BP_{dia} = 55.96 *.02912 * HR + 0.02302 * t \quad (6)$$

Equations (5) and (6) describe using the found value of t to solve for both systolic and diastolic blood pressure [5]. These can be solved using programming languages such as MATLAB or Python.

III. Circuit Design

The full circuit design is based on transmission photoplethysmography, in addition to a 3 lead electrocardiography device as both are needed for blood pressure estimation as seen in Figure 2 below.

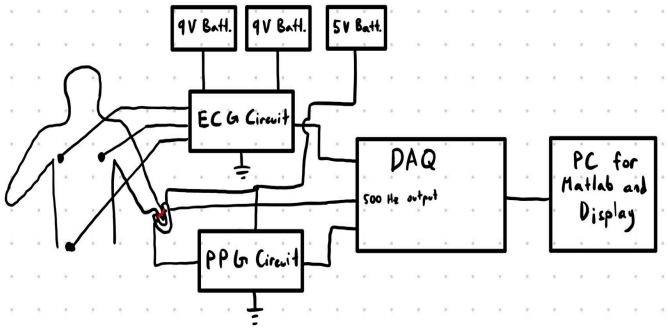

Figure 2: General Circuit/Apparatus Diagram.

Two 9 Volt Batteries are needed to power the ECG Circuit, and one 5 Volt Power source for the PPG Circuit. The detectors for each circuit, electrodes in ECG and a photodiode in PPG, go through their respective circuits for filtering and amplification, etc. before their signals are picked up by the DAQ through a BNC cable. The DAQ also outputs signals to the LEDs to control their frequency. A PC then uses the signals picked up by the DAQ for further signal processing and computation using MATLab to calculate and display our final values for blood pressure, blood oxygen, heart rate variability, and heart rate as well as the ECG waveform.

A. PPG

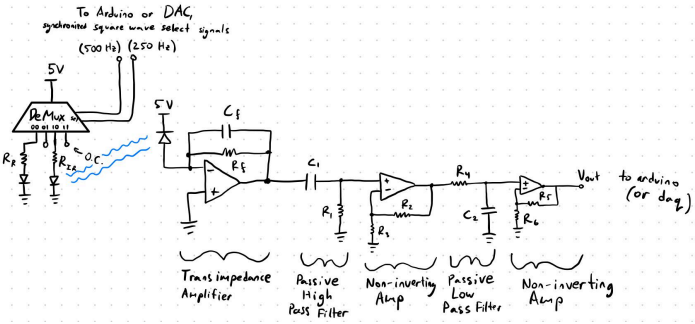

Figure 3: PPG Circuit

Figure 3 [6] shows the circuit designed for the PPG portion of the system. We opted to implement transmission photoplethysmography as opposed to reflective photoplethysmography in our design for better accuracy (meaning measuring the amount of light absorbed that passes through tissue from one side to the other, as opposed to light that is reflected off of tissue and measured from the same side that it is emitted). This decision informed our decision to choose a red and an infrared LED as our two light sources because infrared light is good at passing through tissue.

The LED part of the circuit uses a 1x4 DeMultiplexer to pulse one diode, then off, then pulse the other diode, then off again, with a frequency of 500 Hz. The receiving end of the circuit is designed to have a gain of 3720, which is approximately the 3700 typical of PPG's (Langereis). The photodiode is reverse-biased with a positive voltage at the cathode. Also, Vout leads to our DAQ via a DNC Cable. The first section consists of a transimpedance amplifier. The transimpedance amplifier converts the current produced by the photodiode operating in a photoconductive mode to a voltage for the rest of the circuit. The second section of the circuit is a passive high pass filter with a cutoff frequency of 0.4 Hz. The third section is a non-inverting amplifier with a gain of 60. The fourth section is a passive low pass filter with a cutoff frequency of 20 Hz to filter out any high frequency noise. The final portion is another non-inverting amplifier with a gain of 62.

a. DeMultiplexer

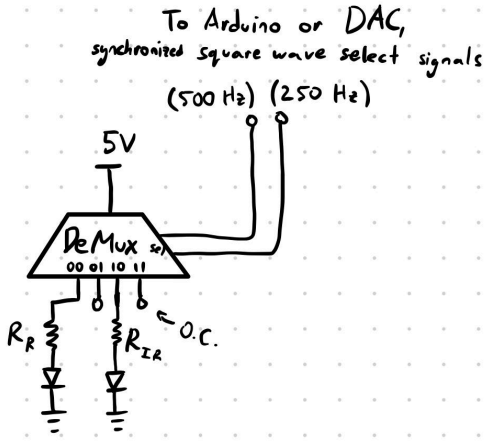

Figure 4: Demultiplexer Diode Circuit

Figure 4 shows the design for the circuit to pulse our LEDs. Using a DeMultiplexer with one input and 2 select lines to switch between 4 outputs, we alternate between each LED and off at 500 Hz. This is accomplished with two synchronized binary square wave signals from the DAQ, with the least significant bit at 500 Hz and most significant bit at 250 Hz. They are synchronized so that together they start at 00, they count through 0-3 in binary to select the index of the output line. This results in an on-off cycle, in which each LED individually has a duty cycle of 25%, which helps us minimize the power consumption and time that we are using each LED such that we prevent them from overheating.

b. Transimpedance amplifier

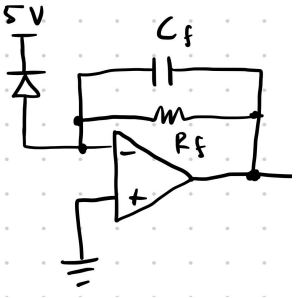

Figure 5: Transimpedance Inverting Amplifier

A transimpedance amplifier is used to convert our current signal from the photodiode to a resistance. Using the math below, we solved for what we would like the resistance to be.

$$Z_{eq} = C_f \| R_f = \frac{1}{\frac{1}{R_f} + jwC_f} = \frac{R_f}{1 + jwR_fC_f} \quad (7)$$

$i_1 = i_2 \qquad V_1 = 0$ (virtual ground)

$$i_2 = (V_{out} - 0) \div \left(\frac{R_f}{1 + jwR_fC_f}\right)$$

$$i_2 = V_{out}\left(\frac{1 + jwR_fC_f}{R_f}\right)$$

Approximation: $i_c \approx 0$ because

$$i_c = \frac{virtual\ ground}{\frac{1}{jwC_i}} \approx 0$$

$$I = i_c + i_1 \rightarrow i_1 = I - i_c \approx I \rightarrow i_1 \approx I(jw)$$

$$I(jw) = V_{out}(jw) * \left(\frac{1 + jwR_fC_f}{R_f}\right) \quad (8)$$

$$H(jw) = \frac{V_{out}}{I_{in}} \approx \frac{R_f}{1 + jwR_fC_f} \quad (9)$$

One pole at $\frac{1}{R_fC_f}$ so we want R_fC_f to equal $\frac{1}{800}$ so the desired signal of approximately 500 Hz is let through

$$R_f = 1k\Omega\ ,\ C_f = 1.25\mu F \quad (10)$$

c. High Pass Filter and Non-inverting Amplifier

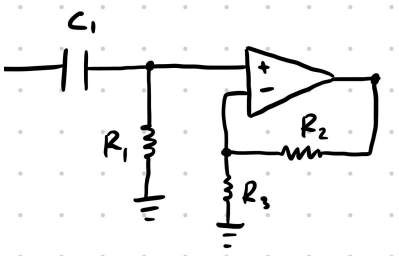

Figure 6: High Pass Filter and Non-Inverting Amplifier

Our High Pass Filter is used because we want to isolate the jumps in voltage that the transimpedance amplifier produces due to the turning on and off of the diodes, and filter out the rest. We then amplify the signal because it is so small. The following is our math:

Input signal (f) $\approx 1Hz$ target f_c for high pass = 0.4 Hz

$$f_c = \frac{1}{2\pi R_1C_1} = 0.4Hz \quad (11)$$

$$C_1 = \frac{1}{2\pi\,(0.4\,Hz)\,R_1} \qquad \text{set } R_1 = 10\Omega$$

$$C_1 = \frac{1}{2\pi\,(0.4\,Hz)\,10\Omega} = 0.040\,F \qquad (12)$$

Target Gain = 60

$$\text{Gain} = \frac{A_f(\frac{f}{f_c})}{\sqrt{1+(\frac{f}{f_c})^2}} = 60 \qquad A_f = \frac{V_{out}}{V_{in}} = 1 + \frac{R_2}{R_3}$$

$$\frac{R_2}{R_3} = \frac{60\sqrt{1+(\frac{f}{f_c})^2}}{\frac{f}{f_c}} - 1 = 63.62$$

$$R_2 = 63.62\,R_3 \qquad \text{set } R_3 = 10\Omega$$

$$R_2 = 636.2\Omega \qquad (13)$$

d. Low Pass Filter and Non-inverting Amplifier

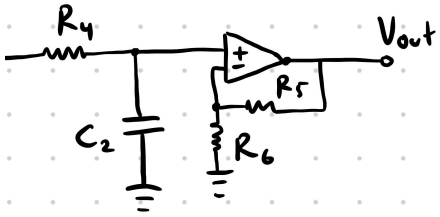

Figure 7: Low Pass Filter and Non-Inverting Amplifier

Our low pass filter is used to filter out high frequency noise from our environment and possible movement by the patient. The math below was used to solve for what we would like our circuit values to be:

Input signal (f) $\approx 1Hz$ target f_c for low pass = 20 Hz

$$f_c = \frac{1}{2\pi R_4 C_2} = 20Hz \qquad (14)$$

$$C_1 = \frac{1}{2\pi\,(20\,Hz)\,R_4} \qquad \text{set } R_4 = 10\Omega$$

$$C_1 = \frac{1}{2\pi\,(20\,Hz)\,10\Omega} = 0.796\,mF \qquad (15)$$

Target Gain = 62

$$\text{Gain} = \frac{A_f}{\sqrt{1+(\frac{f}{f_c})^2}} = 62 \qquad A_f = \frac{V_{out}}{V_{in}} = 1 + \frac{R_5}{R_6}$$

$$\frac{R_5}{R_6} = 62\sqrt{1 + (\frac{f}{f_c})^2} - 1 = 61.08$$

$$R_5 = 61.08\,R_6 \qquad \text{set } R_6 = 10\Omega$$

$$R_5 = 610.8\Omega$$

B. ECG

The ECG component of the device is designed specifically for a 3 lead ECG circuit, to reduce the complexity of the device, and to make it more user friendly. The Figure 4 below is a circuit design sourced from a lab directions document provided to the UCSD students for the course Bioengineering 152; commonly referred to as BENG 152. [12]

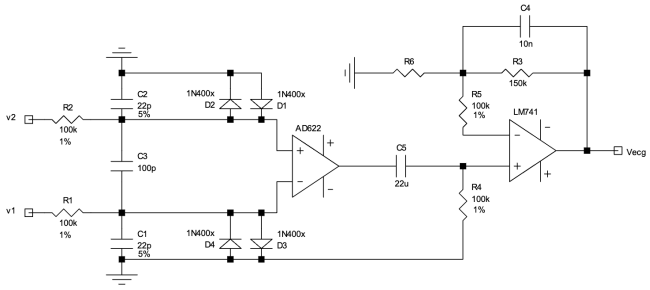

Figure 8 : ECG circuit diagram without necessary chips.

However, this circuit can be simplified further by using an AD622 and a LM741 operational amplifier (Figure 5). The AD622 is a low cost instrumentation amplifier that can be used in a variety of applications, including but not limited to transducer interface, low cost thermocouple amplifier, industrial process controls, and for low cost data acquisition.

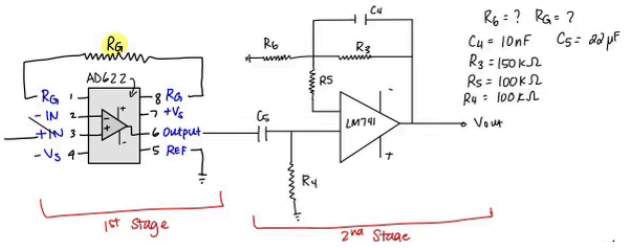

Figure 9: Simplified circuit diagram with the use of AD622(left) and LM741(right).

The ECG circuit employs AD622 and LM741 Op Amps. The AD622 simplifies the circuit, with the gain determined by Rg. R_1-C_1 and R_2-C_2 form low-pass filters. R_4-C_5 creates a high-pass filter with a time constant of 2.2s and a cutoff of 0.07 Hz. R_3-C_4 forms low-pass filters with a time constant of 1.5 ms and a cutoff of 100 Hz. The right stage, post-AD622, acts as a bandpass filter (0.07-100Hz), ideal for removing 60Hz noise. The circuit is powered by two

9V batteries, and input signals coming from chest and abdomen leads.

The designed circuit has a total gain of 1000, as it is ideal for ECG signals. As seen in the previous Figure 5, the values for R_6 and R_g were the only ones needed to be calculated to accommodate the chosen constraints. R_6 is related to the second stage of the circuit connected with the LM741, and Rg is the resistor that needs to be put across the AD622 in the first stage of the circuit to get the desired output and gain. In order to calculate these resistor values, we had chosen a gain for the second stage of the circuit; G_2 to be 30. From this, one will be able to solve for the gain associated with the first stage of the circuit as such.

$$G_{total} = G_1 \times G_2$$

$$1000 = G_1 \times 30$$

$$G_1 = 33.33 \quad (19)$$

This can now help determine the desired R_6 and R_g values. From doing some literature review in the same lab document provided in the course, the following formulas were obtained.

$$G_2 = 1 + \frac{R_3}{R_6} \text{ and } G_1 = 1 + \frac{50.5k\Omega}{R_g} \quad (16)$$

Therefore, plugging in the information that we already know such as for G_1, G_2, and R_3 we can solve for R_6 and R_g as such:

$$30 = 1 + \frac{150k\Omega}{R_6} \Rightarrow R_6 = 5.17 \times 10^3\Omega. \quad (17)$$

$$33.33 = 1 + \frac{50.5k\Omega}{R_g} \Rightarrow R_g = 1.5610^3\Omega \quad (18)$$

Three leads were positioned on the body: one on the lower abdomen (serving as the ground), one on the right chest (negative), and one on the left chest (positive). In the circuit, the left chest lead connects to pin 3, the right chest lead to pin 2, and the abdomen lead to pin 5 of the AD622 chip. The analog signal, represented as Vout, feeds into a data acquisition (DAQ) system on a PC. MATLAB then processes the voltage signals to generate the ECG chart, performing necessary analysis and calculations.

We were able to validate this circuit simply by constructing the designed circuit on a breadboard, and attaching it to an oscilloscope using a BNC cable. The output signal on the oscilloscope can be seen in Figure 6 found below.

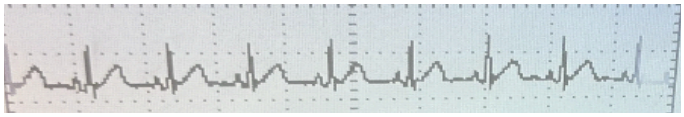

Figure 10 : The output voltage signal from the ECG circuit on the oscilloscope for validation.

IV. MATLAB Algorithm

As mentioned earlier, the analog circuit is to be connected to a computer device for the final stage of data acquisition and for signal processing, as well as for required calculations. To achieve this, both the circuits can be constructed on a breadboard as designed above with appropriate components including ones necessary for single input collection. And then by connecting the breadboard to a DAQ device connected with the computer using a BNC cable. The algorithm to be used for achieving the purpose is written on MATLAB for reliability and lowering complexity.

A. Plotting Voltage Output from ECG and PPG

To start this process, a code is required to produce a graphical representation of the ECG and PPG voltage signals. Along with this, a code appropriate for acquiring the data set to go along with the graph is also required for further calculations such as heart rate, heart rate variability, systolic, and diastolic blood pressure. The code that can do this can be seen in Figure 7 located below.

```
d = daqlist;
d(1, :)
d{1, "DeviceInfo"}
dq = daq("ni")
ch = addinput(dq,"Dev1", "ai0","Voltage")
data = read(dq, seconds(10));
plot(data.Time, data.Dev1_ai0);
ylabel("Voltage (V)");
dq.Rate = 400;
[data, startTime] = read(dq, seconds(10));
plot(data.Time, data.cDev1_ai0);
ylabel("Voltage (V)");
```

Figure 11: MATLAB algorithm for generating a data set of time position and associated voltage output readings and to plot a graph for the desired amount of time.

The code retrieves available data acquisition devices, creates a 'dq' object for National Instruments hardware, and adds an analog input channel ("ai0") from "Dev1" to measure voltage. It then reads voltage data for 1 second, plotting it against time. The acquisition rate is set to 5000 samples per second for a 2-second data read, with the new voltage data plotted. After execution, compiled code displays ECG graph time and voltage readings in MATLAB's workspace. Similar code can process PPG signals, utilizing voltage readings for plotting in MATLAB. This section of the code has been validated while in BENG 152 lab as the code takes in ECG signals and outputs a graph which can be seen in Figure 8 .

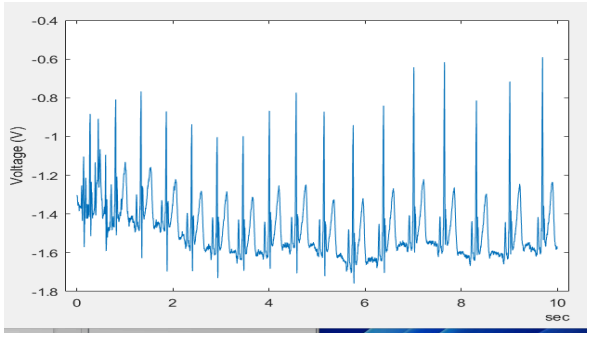

Figure 12: The output graph obtained from ECG signals.

B. Peak Detection

Now moving on to the code used for peak detection for ECG waveform as well as for the PPG waveform. The code can be seen below in Figure that contains the code for the ECG waveform. A very similar code can be seen below in Figure for the PPG waveform. For peak detection in ECG signals, the process begins with iterating through each data point of the ECG signal within a loop. Utilizing if and conditional statements, peak criteria are established to check if a data point is greater than the one preceding and succeeding it, ensuring it surpasses a predefined threshold. If the criteria are met, the voltage amplitude and time position are stored in an array labeled "val" and "pos" respectively. Subsequently, in the plot, peaks are identified with red asterisks. Peak detection for PPG signals follows a similar loop iteration as the ECG peak detection section. It applies the same criteria for peak counting. Likewise, voltage values and time positions of detected peaks are stored in arrays labeled "val" and "pos1" respectively. Similar to the ECG code, peaks in the PPG code are plotted, with the peaks indicated by green asterisks. The code that was constructed to perform peak detection for the ECG signal and the PPG signal acquired from the sensors can be seen in Figure 9 and Figure 10. [10]

```
%% peak detection of ECG
j=1;
n=length(y);
for i=2:n-1
    if y(i)> y(i-1) && y(i)>= y(i+1) && y(i)> 0.45*max(y)
       val(j)= y(i);
       pos(j)=i;
       j=j+1;
     end
end
ecg_peaks=j-1;
ecg_pos=pos./1000;
plot(pos,val,'*r');
title('ECG peak');
```

Figure 13: Peak detection code for ECG.

```
%% peak detection of PPG
m=1;
n=length(z);
for i=2:n-1
    if z(i)> z(i-1) && z(i)>= z(i+1) && z(i)> 0.45*max(z)
       val(m)= z(i);
       pos1(m)=i;
       m=m+1;
     end
end
ppg_peaks=m-1;
ppg_pos=pos1./1000;
ppg_val=val;
plot(pos1,val,'*g');
title('ECG & PPG signal');
legend('ECG signal','PPG signal');
```

Figure 14: Peak detection code for PPG very similar to that for ECG.

C. HR, HRV, PRV, PTT

With peak detection for both ECG and PPG signals that stores the peak amplitude and locations in separate arrays, a code for calculating the heart rate, heart rate variability, pulse rate variability, and PTT can be written. Heart rate provides the number of heartbeats per minute, reflecting the basic cardiac

activity. Heart rate variability (HRV) measures the variation in time intervals between consecutive heartbeats, offering insights into autonomic nervous system function and overall cardiovascular health. Pulse rate variability (PRV) evaluates changes in pulse intervals, correlating with sympathetic and parasympathetic nervous system activity. Pulse transit time (PTT) measures the time it takes for a pulse wave to travel between two points, indicating arterial stiffness, blood pressure changes, and vascular health.

To calculate the heart rate from the ECG signal, the time intervals between successive R peaks, detected by the code in the previous slide (RR Interval), are utilized within a loop. These intervals are saved in a new array, 'e', containing j elements. The heart rate (HR) is determined by dividing 60 seconds by the mean RR interval (in seconds). Additionally, heart rate variability (HRV) can be estimated using the formula 60 divided by 'e', considering each RR interval data point.The corresponding code for this can be seen in Figure 11. [10]

```
j=1;
for i=1:ecg_peaks-1
    e(j)= ecg_pos(i+1)-ecg_pos(i);% gives RR interval
    j=j+1;

end
hr=60./mean(e); % 60/ mean of RR interval

hrv= (60./e); % 60/ each RR interval
figure,stairs(hrv);
title('HRV');
xlabel('samples');
ylabel('hrv');
```

Figure 15: MATLAB code for calculating heart rate from the data with RR interval and it's mean. The output should be in beats per minute (BPM) as the code calculates it for a 60 second time interval.

Pulse Rate Variability (PRV) refers to the fluctuation in heart rate, similar to HRV but derived from PPG signals. This code closely resembles the ECG HRV code, calculating time intervals between successive PPG signal peaks by computing differences between consecutive peaks in the 'pos1' array. The computation of pulse rate and PRV mirrors that of heart rate and HRV, depicted in the image on the top right. Both HRV and PRV metrics are displayed, as shown in the algorithm in the top right and bottom right images. Pulse Transit Time (PTT) is determined using time interval arrays from both ECG and PPG signals, subtracting ECG peak time positions from PPG peak time positions, as executed in the previous slide's code. Subsequently, a graph illustrating each PTT can be generated. The code for carrying this out is in Figure 12. [10]

```
%% PRV
k=1;
for i=1:ppg_peaks-1
    f(k)= ppg_pos(i+1)-ppg_pos(i);
    k=k+1;
end
pr=60./mean(f);
prv= 60./f;
figure,stairs(prv);
title('PRV');
xlabel('samples');
ylabel('prv');
%% PTT
ptt=(ppg_pos-ecg_pos);
figure,stairs(ptt);
title('PTT');
xlabel('ptt');
ylabel('time');
```

Figure 16:MATLAB code for calculating PRV and PTT. These can give feedback on the patient's heart rhythm and insights on Arrhythmia.

D. Calculating PAT

PAT is essential to calculate from the data that we have collected and processed through the previous code. Pulse Arrival Time (PAT) refers to the duration it takes for a pulse wave to travel between two arterial sites, often measured from the R wave of an ECG to the onset of a corresponding pulse wave in a peripheral artery. It is used to assess arterial stiffness, blood pressure changes, and overall vascular health. As mentioned earlier, the estimation formulas for blood pressure can be found under the Physiology section formula (5) and (6). To calculate the blood pressure, we will need a mean value of PAT as can be seen in the same formulas. PAT can be found from the locations of the peaks in terms of time as shown in formula (3) in the Physiology section. Below is Figure 13 with the code that will help calculate PAT. [10]

```
% Calculate Pulse Arrival Time (PAT)
pat = zeros(1, min(ecg_peaks, ppg_peaks) - 1);
for i = 1:min(ecg_peaks, ppg_peaks) - 1
    pat(i) = ppg_pos(i) - ecg_pos(i);
end

% Calculate mean PAT
mean_pat = mean(pat);
```

Figure 17: Code used to calculate PAT from the acquired ECG and PPG waveforms. Along with this, finding the mean of PAT.

To calculate Pulse Arrival Time (PAT), an array named 'pat' is initialized with zeros, its length equal to the minimum number of detected peaks in both the ECG and PPG signals minus one. Through a for loop iterating over the indices of the minimum number of detected peaks between the two signals, the code calculates PAT for each corresponding pair of peaks. This is done by subtracting the ECG peak position from the PPG peak position, representing the time difference between the arrival of the pulse wave at the PPG sensor and the corresponding R peak in the ECG signal. Subsequently, the mean PAT for each peak in the ECG and PPG can be calculated. [10]

E. Estimating Blood Pressure

Now that the code has been used to calculate values needed for estimating the Systolic and Diastolic Blood Pressure using the formulas (5) (6) found from literature review, a code can be constructed that uses the formula and the calculated values to estimate the blood pressure. In the end, when the code is executed, a value for both systolic and diastolic blood pressures. This can be seen in the code down below in Figure 14.

$$BP_{sys} = 184.3 * 1.329 * HR + 0.0848 * t \quad (5)$$

$$BP_{dia} = 55.96 *.02912 * HR + 0.02302 * t \quad (6)$$

```
% Estimate Systolic Blood Pressure (SBP)
sbp = 184.3 - 1.329 * hr + 0.0848 * ((6000 / hr) - mean_pat);

% Estimate Diastolic Blood Pressure (DBP)
dbp = 55.96 - 0.02912 * hr + 0.02302 * ((6000 / hr) - mean_pat);

% Display the estimated SBP and DBP
fprintf('Estimated Systolic Blood Pressure (SBP): %.2f mmHg\n', sbp);
fprintf('Estimated Diastolic Blood Pressure (DBP): %.2f mmHg\n', dbp);
```

Figure 18: The code is analogous to Formulas (5) and (6) under the physiology section (the same formulas are displayed again before this figure). These will be used to estimate blood pressure.

When this whole algorithm is put together, it can be used to process the data collected through the sensors used by the physical circuits for ECG and PPG. Executing this code will also help print the calculated heart rate, heart rate variability, other quality indexes, and an estimated blood pressure.

V. Results

In order to validate the circuit design, and the algorithm written for peak detection, and analysis we will need real sensors or signals that resonate similarly to the biopotential signals that are sensed using ECG and PPG sensors. However, not having the resources needed to do this, the PPG circuit design was attempted to be validated using the Falstad simulation software available online. On the other hand, the circuit design for the ECG component of the device was validated during the BENG 152 lab session as the circuit was built on a breadboard, and appropriate 3 leads were used for generating the input signal coming from the sinus rhythm.

A. Overall Circuit Design for The PPG Component of The Device

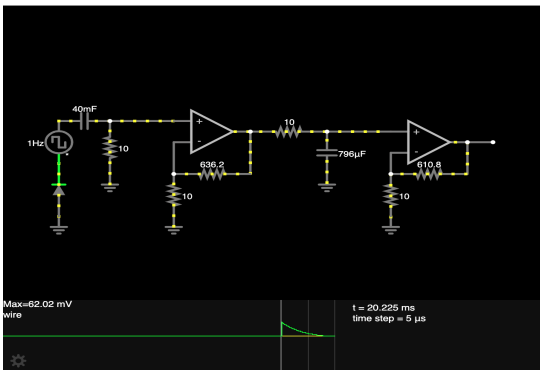

Figure 19: Circuit simulation done for the PPG circuit.

The simulation in Figure 15 above, shows the PPG circuitry. The square wave is used to mimic the signal of heartbeat pulses, as there are no other waveforms or collected data available for this. And the output can be seen on the graph at the bottom. The circuit as seen above was created and used for the simulation because the software does not contain the MCP6002 chip that the main circuit design intended to use for the PPG circuitry. The output signal as seen plotted in the bottom of the circuit figure, is collected by using a square signal as mentioned, which is why it is not sure if that is the signal behavior that was expected from the input and by passing it through the circuit. This suggests that further validation is needed for the PPG part of the circuit to make sure that the circuit works as intended.

B. Transimpedance Amplifier

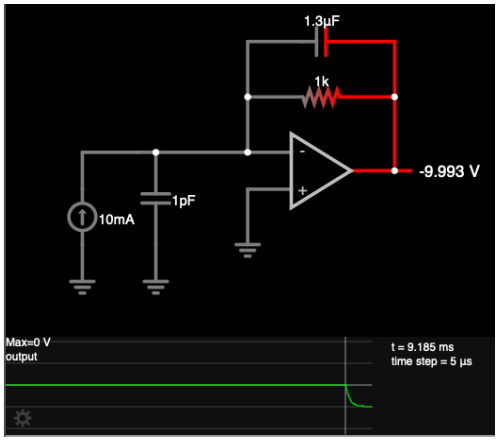

Figure 20: Transimpedance Amplifier

We approximated the alternating current signal from the photodiode as a current source in parallel with a capacitance. The actual approximation would be for an ac current source in a square wave, but since there is no ac current source in Falstad, we have simply taken the very beginning of the simulation to be representative of how the circuit will respond to a wave front. We correctly see that it jumps to a voltage value peak, which is what we want. By flashing the lights, we will cause the current from the photodiode to jump to different values based on the amount of light hitting the diode, which is sensitive to both wavelengths of light being shown.

These jumps will be correctly converted into jumps and peaks in voltage by the transimpedance amplifier, with noise filtered out later in the circuit.

Further validation should be done with software that is able to create an ac current source, and additionally, a square wave ac current source is not a perfect model for the type of data that would be collected, especially due to ambient light. Addressing these issues would better confirm that our transimpedance amplifier works as intended.

C. Overall Circuit Design for The ECG Component of The Device

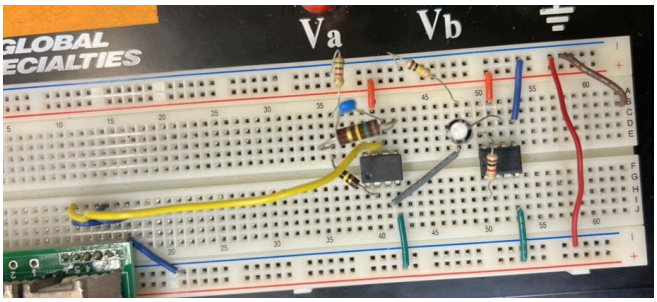

Figure 21: ECG circuit constructed on a breadboard as designed in Figure 5.

On the contrary, the ECG circuit has been validated by physically constructing it on a breadboard and using electrode leads to acquire sinus rhythm signal as the input signal. The circuit built using the AD622 and LM741 amplifier chips can be seen above in Figure 16. And the signal that the circuit outputs can be seen in Figure 6 displayed earlier. The output signal suggested that the circuit was working as intended as the gain was set to G=1000, and it is evident from the oscilloscope readings that the output signal was in Volts, compared to the input signals from the electrode leads being in mV. The signal seen in the oscilloscope output also has minimal noise due to the filters placed in the circuitry to help ensure only desired signal frequency is collected and displayed. Due to this, there is a clear PQRST waveform visible in the signal being outputted.

D. Overall Algorithm for Signal Display and Analysis

Lastly, the complete algorithm has not been validated because there was no biosignal to be displayed or analyzed. However, the MATLAB code that was used to display collected signals from the analog circuit through data acquisition using the DAQ was validated during the BENG 152 lab session while validating the analog circuit design for the ECG component of the device. The code collected analog signal data and displayed it digitally through MATLAB for 10 seconds, and the resulting voltage reading chart for ECG can be seen in Figure 8 under the MATLAB Algorithm section above.

VI. Conclusion

In conclusion, the development of a medical device circuit integrating ECG and PPG capabilities into a single user-friendly device represents a significant advancement in cardiovascular monitoring technology. This innovation streamlines the process for patients with arrhythmia, hypertension, and other cardiovascular conditions, eliminating the need for multiple devices and enhancing convenience. The device offers comprehensive functionality, enabling the calculation of heart rate, heart rate variability, quality statistics, and estimated blood pressure

through a unified interface. This consolidation simplifies monitoring procedures and promotes patient compliance with vital sign tracking protocols.

While the current design demonstrates remarkable utility, several avenues for future improvement have been identified. Addressing the limitations associated with PPG accuracy, such as implementing compact fluorescent lumens to enhance peak interval accuracy while minimizing power consumption, presents a promising pathway for refinement. Additionally, incorporating a shroud to mitigate ambient light interference and exploring wearable integration options, such as incorporating the system into a watch or wristband, can further enhance the device's usability and versatility.

Furthermore, the potential integration of machine learning algorithms to estimate blood pressure without relying on ECG signals holds promise for increasing portability and user-friendliness. Despite the challenges posed by factors like skin tone, age, sweat, and activity level affecting PPG accuracy, advancements in algorithmic approaches can mitigate these limitations and enhance the device's reliability across diverse user demographics. Addressing power consumption concerns, perhaps by exploring alternative power sources or optimizing circuit design, would enhance the device's portability and reduce operational costs. By overcoming these challenges, the device can achieve greater accessibility and effectiveness in clinical and home monitoring settings.

In summary, while acknowledging the current design's strengths, this report underscores the importance of ongoing research and development efforts to refine and optimize the medical device circuit design. By embracing innovation and addressing identified areas for improvement, we can continue to advance patient care and empower individuals to monitor their cardiovascular health with greater ease and accuracy.

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