Design of Blood Pressure Monitor Indicating Hypertension

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ABSTRACT–THE MAIN PREDICTOR OF CARDIOVASCULAR DISEASE IS HYPERTENSION IN THE HUMAN HEART[1]. IF LEFT UNTREATED, HYPERTENSION CAN LEAD TO SIGNIFICANT HEALTH PROBLEMS BOTH PHYSICALLY AND MENTALLY. BLOOD PRESSURE MONITORS CAN INDICATE THE SEVERITY OF THE CONDITION AND CAN HELP PATIENTS DECIDE WHETHER THEY ARE FACING A LIFE-THREATENING SITUATION. WHILE CURRENT BLOOD PRESSURE MONITORS GIVE READINGS PERIODICALLY, ONE ASPECT THAT THE DESIGN ADDRESSES IS ONGOING MONITORING OF THE PATIENT SUCH AS DURING INTENSIVE CARE. THE DESIGN OF OUR BLOOD PRESSURE MONITOR IS TO CONTINUOUSLY MONITOR THE HEART RATE AND DETERMINE IF THE PATIENT'S BLOOD PRESSURE GOES ABOVE SYSTOLIC READINGS SIGNIFYING POSSIBLE HYPERTENSION [1]. THIS IS ACHIEVED BY MONITORING THE BLOOD PRESSURE WITH A MICRO-TIPPED MANOMETER, CONNECTED TO A PRESSURE TRANSDUCER AND BANDPASS FILTER, TO INDICATE THE PATIENT'S CURRENT CONDITION WITH THE HELP OF THREE SEPARATE COMPARATORS AND LEDS. LIMITATIONS WERE FOUND DURING OUR DESIGN PROCESS, AND THE ONLY MEASUREMENT OBTAINED BY THE CIRCUIT WAS SYSTOLIC PRESSURE. REGRETTABLY, A NON-INVASIVE TECHNIQUE WAS NOT CONSIDERED AND COULD BE A POTENTIAL FUTURE IMPROVEMENT IN THE DESIGN TO HELP WITH EASE OF USE. THE MONITOR IS DESIGNED TO BE AN EARLY SIGN OF WARNING AGAINST ILLNESSES SUCH AS CARDIOVASCULAR DISEASE, KIDNEY DISEASE.BLOOD VESSEL DAMAGE, AND OTHER BODILY DISORDERS.

I. INTRODUCTION

Hypertension is a medical condition where a person experiences abnormally high blood pressure. A healthy person will have a systolic pressure under 120 mmHg and diastolic pressure under 80 mmHg. The first stage of hypertension is reached when systolic pressure reaches 130-139 mmHg or diastolic pressure reaches 80-89 mmHg. The second stage of hypertension is reached when systolic pressure reaches 140-180 mmHg or diastolic pressure reaches 90-120 mmHg. The third stage of hypertension, also known as hypertensive crisis, is reached when systolic pressure exceeds 180 mmHg or diastolic pressure exceeds 120 mmHg [1].

According to the Centers for Disease Control and Prevention (CDC), approximately 47% of the adult population in the United States have hypertension. Moreover, the CDC estimates that 20% of adults do not know that they have hypertension [2]. If left untreated, hypertension causes damage to the arteries by exposing arterial walls to constant high blood pressures, which increases the risk of aneurysms. The heart also works harder to maintain these high blood pressures, which increases the risk of heart disease and in the worst case scenario, heart failure. Lastly, since arteries supply blood to vital organs throughout the body, hypertension can indirectly impair the functions of these organs. For example, abnormally high blood pressure in the brain increases the risk of stroke or dementia [2][3].

Given the high percentage of adults who have hypertension in the United States, as well as the concerning percentage of adults who do not know they have hypertension, it is imperative that healthcare workers have access to a tool that can quickly and accurately determine if a patient has hypertension. The blood pressure monitor proposed in this paper was designed to fill this need. The blood pressure monitor first measures the systolic pressure of a patient via a micro-tipped manometer. The signal measured by the pressure transducer is then filtered using a bandpass filter to remove noise outside of the typical heart rate frequency range. Lastly, if the signal indicates a systolic pressure within one of the three stages of hypertension, the monitor will display the stage of hypertension the patient's readings are in with the use of three LED lights. This blood pressure monitor will give healthcare workers the means to effectively diagnose patients with hypertension, which gives affected patients the opportunity to begin treatment to avoid further health complications.

II. MONITOR DESIGN

A. CATHETER TUBE

The circuit can be divided into four general components. The first component is the electrical equivalent circuit for the micro-tipped manometer catheter tube system. The catheter tube allows the pressure transducer to measure a continuous time waveform of the blood pressure [4]. Without the tube, only discrete signals of the pressure would be measured, similar to that of the pressure cuff in a sphygmomanometer. However, developing a continuous time signal allows the measurement of heart rate and continuous blood pressure. In order to develop the catheter tube design, the electrical equivalent for the fluid-diaphragm system must be acknowledged.

There are four factors of the catheter tube diaphragm system that require consideration in the equivalent circuit: fluid inertia, fluid resistance, compliance of the diaphragm, and compliance of the bubbles.

Fluid inertia and fluid resistance are reliant on the fluid that pushes into the diaphragm when pressure levels change in the blood. To quantify these phenomenon as circuit components, the following formulas were used to develop impedances:

$$L = \rho \frac{l}{\pi r^2} \textbf{ [Eqn.1]}$$

$$R = \frac{8\eta l}{\pi r^4} \textbf{ [Eqn.2]}$$

L represents the inertia of the fluid and is related to the length of the catheter tube l, radius of the catheter tube r, and density of the fluid ρ. R represents the resistance of the fluid flow in the tube and is related to the viscosity of the fluid η, length of tube l, and radius of tube r. The radius and length of the tube were determined to be 2mm and 40cm respectively by model constraints. The fluid of choice for the diaphragm movement was water r as the fluid transducer which gives the values of $\rho = 997\frac{kg}{m^3}$ and $\eta = 1.0016 Pas$ [5]. From this the component values can be derived and are displayed in Table 1.

To account for compliance, the effect of the bubbles in this monitor system was assumed to be minimal. This makes the compliance of the system only reliant on the diaphragm. Below is the equation for calculating the capacitance of diaphragm compliance:

$$C_d = \frac{\Delta V}{\Delta P} = \frac{1}{E_d} \textbf{ [Eqn.3]}$$

where ΔV represents the change in volume of the tubing and ΔP represents the change in pressure resulting from the blood. E_d is the Young's modulus in relation to the elasticity of the diaphragm. Note that E_d is different from the Young's Modulus as it is characterized as a response to volume change. By constraining the Young's Modulus for the diaphragm to 30 MPa, the value of E_d was found using the following equation by relating strain to the change in volume:

$$E_d = \frac{E}{3V} \textbf{ [Eqn.4]}$$

where V is the spherical volume of the catheter. Below are the final values of the impedances in the fluid-diaphragm system:

Equivalent	Impedance	Value
Inertia of Fluid	L	$31.8MH$
Resistance to Fluid Flow	R	$63.8M\Omega$
Compliance of Diaphragm	C	$3.35pF$

Table 1: Values derived from the constraints of the design for the electrical equivalent catheter-tube. These values were calculated using physical constraints and applying them to Equations 1-4.

An important note for this portion of the circuit is its practical validity. The values for the impedances may seem improbable, but the purpose of this portion of the circuit is to produce time continuous waveforms based on the physical constraints. Therefore, the electrical equivalent values may not make sense as they are all theoretical ways to apply physical phenomenon to waveform generating. While these values are unrealistic, the constraints themselves are not, which allows this design to still be functional. In the circuit simulator, the input voltage was simulated to be AC as it represents the changing pressure coming into the catheter. with the continuous wave generated from this component, the pressure transducer can generate a blood pressure signal.

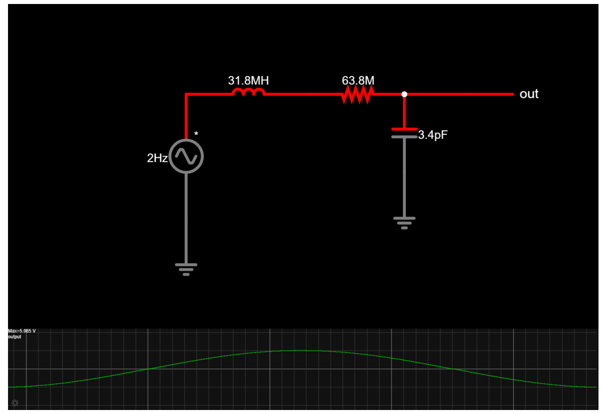

Figure 1: The electrical equivalent of the Catheter Tube system. The output of this circuit is connected directly to the pressure transducer.

B. PRESSURE TRANSDUCER

In order to quantify the change in pressure from the catheter tube, a pressure transducer is needed. An active wheatstone bridge is used to convert the pressure into readings with one gauge resistor with a gauge factor of $G = 100$. The sensitivity of the circuit to pressure changes must remain small for it to read values from 0-200 mmHg across an op-amp powered by a 5V battery. Therefore, the target sensitivity of the pressure transducer is constrained to 22.5mV/mmHg. The value of the transducer components are found in Table 2.

Component	value
Feedback resistor	$20.16k\Omega$
Wheatstone Resistors	$1k\Omega$

Table 2: Values for the wheatstone bridge derived from the sensitivity constraint, gauge factor, and the range of blood pressure 0-200mmHg.

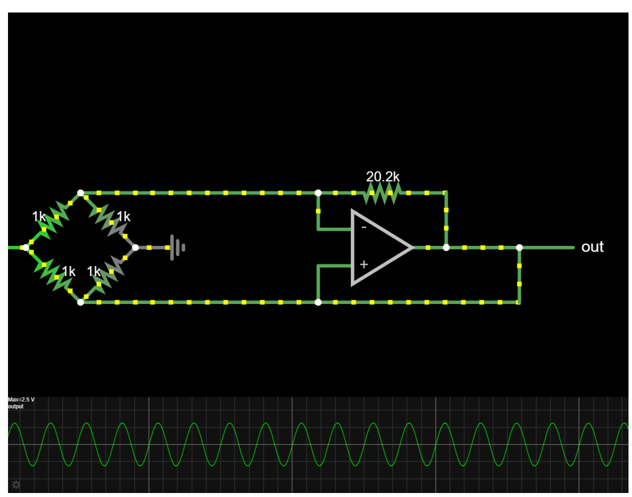

Figure 2: Diagram of the Wheatstone bridge that takes input from the catheter tube along with its scope. The output of the

op-amp connects to the bandpass filter and the input is the voltage from the catheter tube.

Figure 2 depicts the response of voltage change in the output. The variability in the voltage is in response to the strain gauge resistor changing. While the variability of the strain resistor can't be modeled in the circuit simulator, modifying the voltage coming in at the input acts as the variability of the resistor causing voltage change in a real wheatstone bridge. The response of voltage to pressure change in figure 3 further validates the wheatstone bridge design and shows a linear relationship as expected.

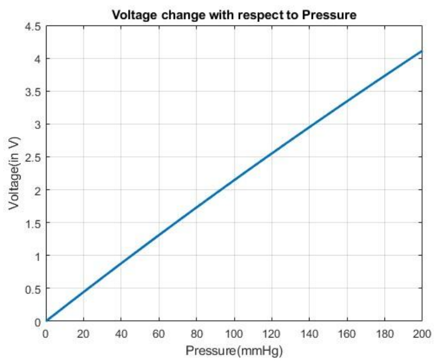

Figure 3: Graph of Voltage response in the wheatstone bridge in response to the pressure change from blood pressure.

C. BANDPASS FILTER

In order to filter out the noise from the pressure transducer, it is necessary to analyze the rate at which systolic and diastolic pressures occur. Blood pressure spikes as a response to the heart contracting, so the rate at which systolic/diastolic occurs is the same as the heart rate[6]. This allows cut-off frequencies to be made in order to filter out noise. These frequencies were found to be 0.5 Hz and 2.5 Hz based on the target heart rate range of 30 beats per minute to 150 beats per minute [7]. The conversion equation can be seen below:

$$F_c = \frac{HR}{t_{pp}} \textbf{ [Eqn. 5]}$$

where HR is the minimum or maximum heart rate in beats per second and t_{pp} is the time interval between systolic pressure peaks.

An active inverting bandpass filter powered by a 5V battery was used to limit the frequency range between these cutoffs. Knowing the cutoff frequencies allows the components for the filter to be determined by the following equations:

$$f_c = \frac{1}{2\pi CR} \textbf{ [Eqn.6]}$$

where C and R are the respective capacitance and resistance correlating to the particular cutoff frequency. The following table shows the capacitor and resistor values found by Equation 6.

Component name	value
low-cutoff resistor R_1	$31.8k\Omega$
low-cutoff capacitor C_1	$10\mu F$
high-cutoff resistor R_2	$6.37k\Omega$
high cutoff capacitor C_2	$10\mu F$

Table 3: Values for the component values of the bandpass filter, where the 1 indicates the lower threshold and the 2 represents the upper.

$$Gain = -\frac{R_2}{R_1} \textbf{ [Eqn.7]}$$

This filter's gain is shown in equation 7. Which gives a total gain of the bandpass as -0.2, indicating that the voltage is inverted through the bandpass filter as expected.

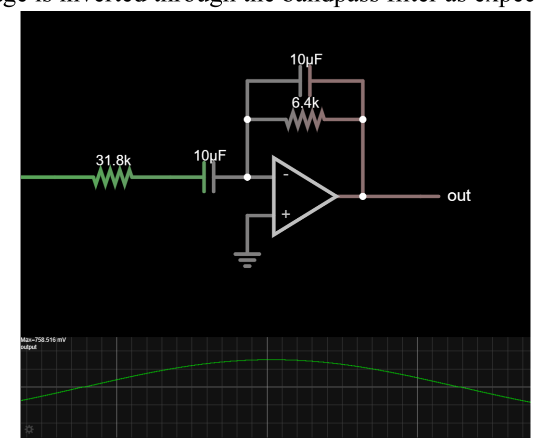

Figure 4: Diagram of the bandpass filter and scope. The output leads into the comparator system. Noise is inputted to determine the effectiveness of the bandpass filter.Noise is on the right graph and filter is on the left.

Figure 4 shows how effective the bandpass filter is at limiting noise outside the frequency thresholds.The signal was tested within the frequency range allotted and indicates that a smoother signal can be extracted with the use of this component. Noise is attributed to the possible excess movement of fluid flow in the catheter, excess movement of the diaphragm itself, or possible blood back flow in the catheter. Using the bandpass filter in the same frequency range as the heart rate allows the signal to effectively drop noise correlating to small vibrations in the diaphragm from smaller blood movement and fluid movement. From this point continuous heart rate and blood pressure measurements can be made.

C. COMPARATORS AND LED SYSTEM

Hypertension stages are characterized by their range of systolic and diastolic pressures. Because both the systolic and diastolic rise in hypertension, the systolic pressure is the peak in the signal intervals. Hypertension stages can then be quantified on the peak of the signal coming out from the bandpass filter for each peak-to-peak reading. Due to prior knowledge of the hypertensive systolic pressures, it is possible to quantify the voltages at each stage. By running the input pressure through the entire circuit's transfer functions, the voltages for the minimum thresholds can be derived below:

Hypertensive stage	Voltage
Stage 1	$-\ 0.5505V$
Stage 2	$-\ 0.5904V$
Stage 3 (crisis)	$-\ 0.7466V$

Table 4: Values corresponding for the minimum thresholds for each classification of hypertension. These thresholds are based on the systolic pressure.

Three inverting Hysteretic comparators are used at the output to compare the voltage output to the threshold stage voltages. To find the threshold voltage for the reference node in the comparator system that satisfy the hypertensive stages, the following equation is used:

$$V_{ref} = \frac{R_{s1}}{R_{s1}+R_{s2}} * V \textbf{ [Eqn.8]}$$

where V is the 5V battery connected to the comparator and R_{s1}/R_{s2} are the resistances of the voltage divider at the voltage reference node. The comparator resistor values were found using the stage voltage values as V_{ref} using Equation 7. The purpose of using an inverting Hysteretic comparator was to invert the voltage from the previous inversion from the bandpass filter and to minimize the chance of noise activating the LED lights.

Component	Resistance value
Stage 1	$3.752k\Omega$
Stage 2	$3.952k\Omega$
Stage 3 (crisis)	$4.733k\Omega$

Table 5: Component values of voltage division in the comparator feedback loop. Note each resistor here is connected to a $1k\Omega$ resistor R_{s1}.

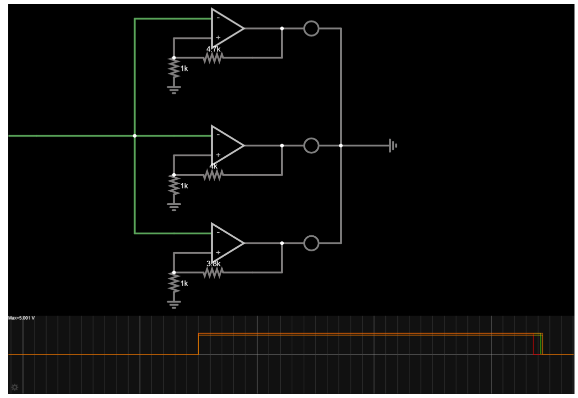

Figure 5: Diagram of the comparator system connected to the LEDs and the Scope graph for the signal for each comparator. From bottom to top the comparators compare Stage 1, Stage 2, and Crisis.

The comparators were able to successfully activate when the threshold voltage is met as indicated by the slight difference between the intervals of the square waves starting and stopping. For this entire design to function properly, the pressure catheter tube pressure system would have to be calibrated to ensure that voltages are zeroed out at a blood pressure reading of zero.

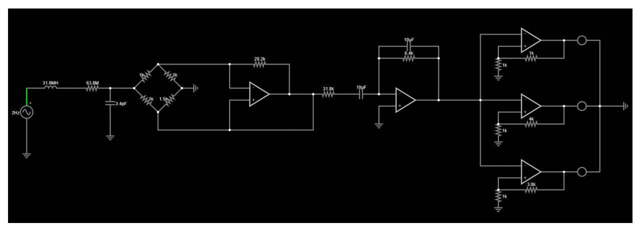

Figure 6: The complete circuit for the blood pressure monitor. From left to right is the catheter equivalent, wheatstone bridge transducer, bandpass filter, and comparators/LEDS.

III. LIMITATIONS

The primary limitation of this blood pressure monitor is that it only measures systolic blood pressure. Under the current design, a patient with normal systolic pressure but abnormally high diastolic pressure will be considered healthy. In this sense, the blood pressure monitor would be inaccurate in diagnosing other blood pressure related disorders directly.

The blood pressure monitor also has limitations in terms of practical use. Since this monitor uses an invasive approach to measure blood pressure, via a micro-tipped manometer, a trained healthcare worker is needed to set up the blood pressure monitor – the average person may injure others or themselves if they attempt to insert the catheter without appropriate training. Moreover, since this is an invasive technique, the experience of having their blood pressure taken will likely result in a discomfort for the patient, as the catheter needs to stay in their body throughout the entire blood pressure reading.

IV. FUTURE IMPROVEMENTS

Adding a component to measure diastolic pressure would significantly increase the accuracy of the blood pressure monitor, as both systolic and diastolic pressures are used to determine rarer forms of hypertension and other significant heart issues. This would eliminate the primary limitation of the current design, where a patient with normal systolic pressure but abnormally high diastolic pressure will be considered healthy.

Taking the continuous blood pressure readings with non-invasive techniques, such as using a time continuous cuff instead of a catheter, will help make the monitor more patient-friendly. This would mean that the patient can have their blood pressure readings without the need of a professional healthcare worker.

Adding additional comparators to measure hypotension, or abnormally low blood pressure, would also increase the scope of the monitor. This would be a relatively simple addition, as the current design already measures systolic pressure and filters the resulting biosignal. Only the voltage threshold for hypotension needs to be calculated to extend the scope of this monitor.

V. CONCLUSION

The goal of this design is to measure heart rate and blood pressure continuously when used on the patient and use LEDs to indicate the high blood pressure stage determined by the voltage threshold within the comparators.

A catheter and micro-tipped manometer is used to measure the biosignal from the patient. The blood pressure is then valued as a voltage by a wheatstone bridge. Then, the signal is filtered to the heart rate frequency and three comparators connected to 3 LEDs to show what state of hypertension the patient is in. The limitations are that there is no measurement of diastolic pressure or hypotension. Implications from the study could arise from future testing of the accuracy of the design. Future improvements would be to add more comparators to allow for detection of hypotension and to come up with avenues in how to detect blood pressure as the current method is too invasive for anyone. This design could be used for further studies into continuous heart rate and blood pressure monitoring with emphasis on determining heart disorders.

VI. REFERENCES

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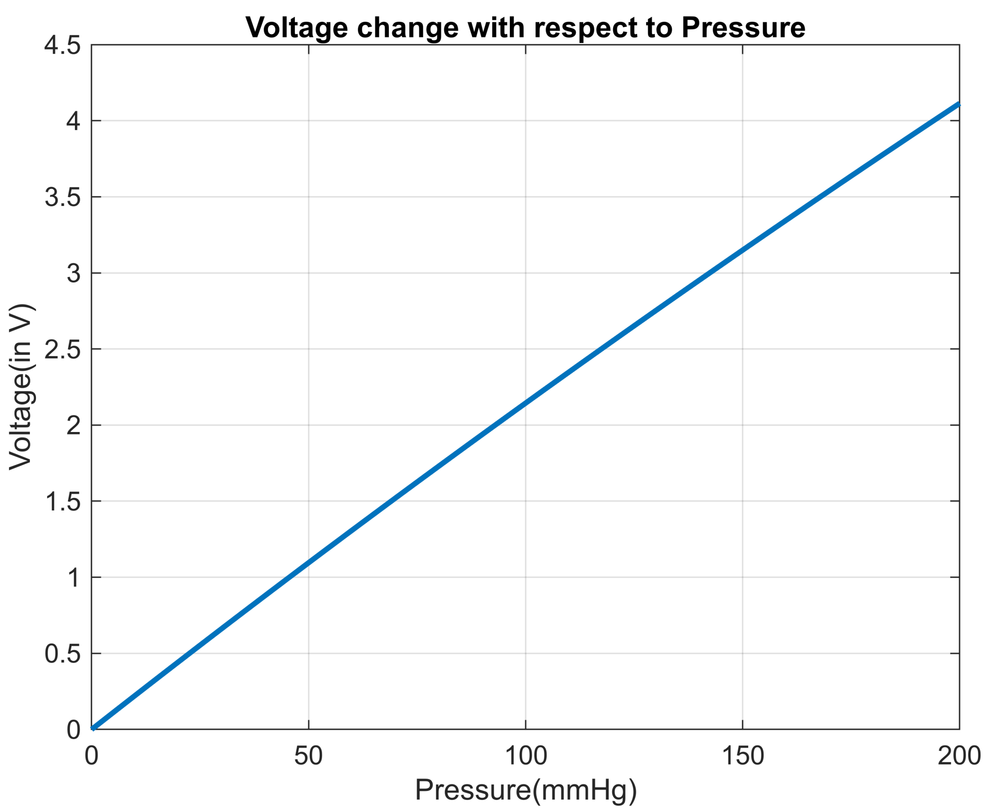
APPENDIX-BENG 186B PROJECT CODE

Wheatstone Bridge

```
% hehe
Rg = 1000;
Rf = 20160;
Vs = 5;
E = 30*10^6;
conv = 133.322;
G = 100;
Gain_WB = @(p) 0.5*Vs*((Rf*(G*p*conv/E))/(Rg*(1+G*p*conv/E)));
pp = linspace(0,200,200);
WB_V = zeros(200,1);
for i = 1:length(pp)
    WB_V(i) = Gain_WB(pp(i));
end
figure();
    plot(pp,WB_V,'LineWidth',2);
    title('Voltage change with respect to Pressure')
    xlabel('Pressure(mmHg)')
    ylabel('Voltage(in V)')
    grid on
```



Bandpass Filter

```
% h
fc = [0.5, 2.5]; % cutoff frequencies
C = 10*10^-6; % Capacitor value for i and f
r = @(frequency) 1/(2*pi*frequency*C); % function to find the Resistance
R = zeros(2);
for i = 1:length(fc)
    R(i) = r(fc(i));
end
disp(R)
```

```
   1.0e+04 *

    3.1831         0
    0.6366         0
```

```
Gain_BP = -R(2)/R(1) % Gain from bandpass filter
```

```
Gain_BP = -0.2000
```

Comparators

```
% Finding the voltages at the output of the bandpass filter from the lower
% threshold of each stage of hypertension
V_stage1 = Gain_WB(130)*Gain_BP
```

```
V_stage1 = -0.5505
```

```

V_stage2 = Gain_WB(140)*Gain_BP
```

```
V_stage2 = -0.5904
```

```

V_Crisis = Gain_WB(180)*Gain_BP
```

```
V_Crisis = -0.7466
```