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Wireless Personal Health Monitoring System

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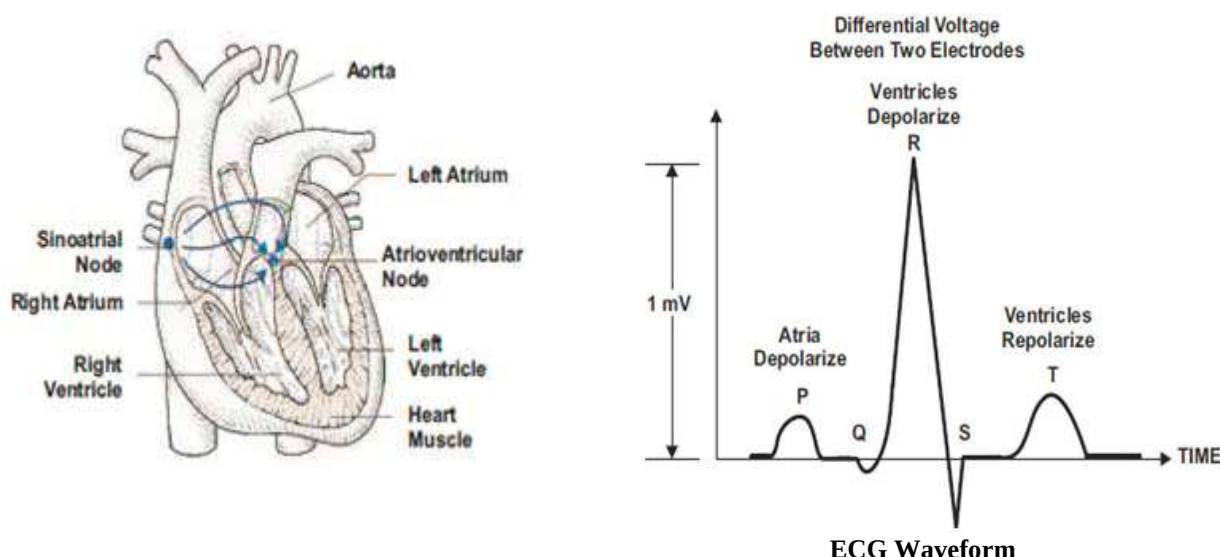
Abstract: The Human heart ECG is the most vital sign for analyzing an individual personal health. The cardiac activity monitoring systems plays a very important role in understanding such a vital sign. Developing a wireless version of such a monitoring system is our main intension. Our paper describes the wireless monitoring of cardiac signals through Smartphone using the devices dedicated for the signal acquisition and synthesis such as AD620. And an instrumentation amplifier designed around it to enhance its function .The Paper also explains the use of a suitable microcontroller to process the incoming signal to perform the signal transformation of an analog time varying signal to a digital equivalent transferable data. The same microcontroller is playing an important role in establishing wireless communication with the appropriate Smartphone.The Smartphone is the end user application where the observation on the ECG will be made to facilitate the understanding of the activity of the desired body vital sign.

I. INTRODUCTION

Mobile computing platforms are going to dominate the information communication technology (ICT) sector in this age of rapid technological innovations. Mobile device like smartphones and tablet computers continuously grow in processing power and become an integral part of daily life, even in developing countries in intensive care units, there are provisions for continuously monitoring patients. Their heart rates, temperatures, ECG etc. are continuously monitored. But in many cases, patients get well and come back to home from hospital. But the disease may return, they may get infected with a new Disease, there may be a sudden attack that may cause his death. So in that case, patients are released from hospital but still they are strongly advised to be under rest and observation for some period of time (from several days to several Months). In these cases, our system can be quite handy. The electrical impulses within the heart act as a source of voltage, which generates a current flow in the torso and corresponding potentials on the skin. The potential distribution can be modeled as if the heart were a time-varying electric dipole. If two leads are connected between two points on the body (forming a vector between them), electrical voltage observed between the two electrodes is measurable. An accurate indication of the frontal projection of the cardiac vector can be provided by three leads/electrodes, one connected at each of the three vertices of the Einthoven triangle. Generally, as many as twelve leads are used to monitor cardiac signals. The most prevalent and significant among these is Lead II for diagnosing rhythm problems. Signals from Lead II measure the variations in potential between the right arm and the left leg, with the electrode of the left arm acting as the ground.

II. ECG SIGNAL AND SOURCES OF NOISE IN ECG SIGNAL

ECG signal and sources of noise in an ECG signal are a part of a Cardiac activity and a very vital life indicating parameter. An electrocardiogram (ECG) is a graph showing electrical activity in the heart. It is the trace of voltage generated by the cardiac muscle during complete Systole and a Diastolic activity. It has become a medical standard to test the human heart for defects and diseases. The heart generates an electrochemical impulse that spreads out in the heart in such a fashion as to cause the cells to contract and relax in a timely order and, thus, give the heart a pumping characteristic. This sequence is initiated by a group of nerve cells called the Sino-Atrial (SA) node, resulting in a polarization and depolarization of the cells of the heart. This pulse travels from the SA node through the surrounding cells of the heart and then to the Atrio-Ventricular (AV) node. The AV node acts as a gate that allows the atria to finish contraction before allowing the pulse to move on to the ventricles. Each atrium pumps blood to a corresponding ventricle. The right atrium pumps blood to the right ventricle to provide blood to the lungs. The left ventricle, sourced by the left atrium, is the chamber that pumps blood throughout the body. An actual voltage potential of approximately 1mV develops between various body points.



ECG interpretation relies heavily on the QRS complex. It starts with a small downward deflection called a Q wave. It is narrow and small in amplitude. It is followed by first positive deflection of QRS complex called R wave. S wave is the first wave after the R wave that dips below the baseline. The end of the S wave occurs where the S wave begins to flatten out. This is called J point. The width of QRS complex often indicates the location of originating electrical impulse. Its duration is $<0.11\text{sec}$. The slope of qR segment determines the BW of the signal. Roughly it has duration of 0.04sec and the peak is approx. 1mV . Noise considerations also affect the BW of signal. The analog front end circuit for detection of ECG must be able to deal with extremely weak signals because raw ECG signal from electrodes ranges from 0.5 mV to 5.0 mV . It also has a dc component of up to $\pm 300\text{ mV}$ that results from the electrode skin contact plus a common-mode component of up to 1.5 V as a result of the potential between the electrodes and ground. The useful bandwidth of an ECG signal depends upon the particular application. It can range from 0.5 Hz to 50 Hz [1]. ECG signals may be corrupted by various kinds of noise [1]. The main sources of noise are:

- Power-line Interference: 50 Hz pickup and harmonics from the power mains
- Electrode contact noise: Variable contact between the electrode and the skin causes baseline drift.
- Motion Artifacts: Shifts in the baseline caused by changes in the electrode-skin impedance.
- Muscle Contraction: Electromyogram signals (EMG) are generated and mixed with the ECG signals.
- Respiration and Perspirations: Causes drift in the baseline
- Electromagnetic Interference: from other electronic

Devices, with the electrode wires serving as antennas, and noise coupled from other electronic devices, usually at high frequencies.

III. IMPLEMENTATION

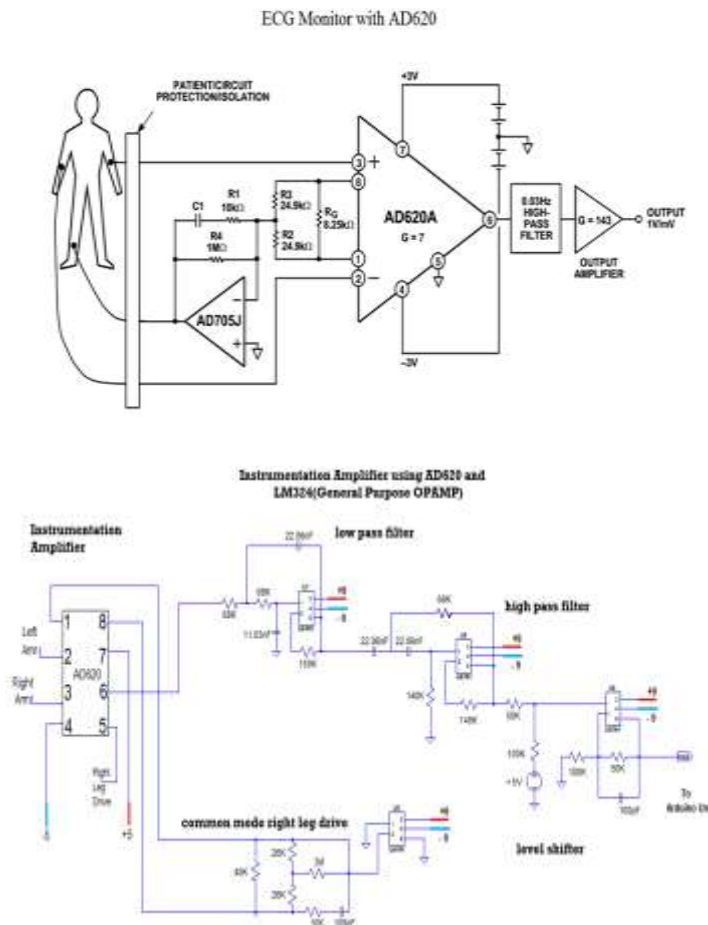
A. Low Cost Low Power Instrumentation Amplifier AD620

The AD620 is a low cost, high accuracy instrumentation amplifier that requires only one external resistor to set gains from 1 to 10,000. The AD620 features 8-lead SOIC and DIP packaging that is smaller than discrete designs and offers lower power (only 1.3mA max supply current), making it a good fit for battery powered, portable (or remote) applications. The AD620, with its high accuracy of 40 ppm maximum non-linearity, low offset voltage of $50\mu\text{V}$ max, and offset drift of $0.6\mu\text{V}/^\circ\text{C}$ max, is ideal for use in precision data acquisition systems, such as weigh scales and transducer interfaces, the low noise, low input bias current, and low power of the AD620 make it well suited for medical applications, such as ECG and noninvasive blood pressure monitors. The low input bias current of 1.0 nA max is made possible with the use of Super Beta processing in the input stage. The AD620 works well as a preamplifier due to its low input voltage noise of $9\text{ nV}/\sqrt{\text{Hz}}$ at 1 kHz , $0.28\mu\text{Vp-pin}$ the 0.1 Hz to 10 Hz band, and $0.1\text{ pA}/\sqrt{\text{Hz}}$ input current noise. Also, the AD620 is well suited for multiplexed applications with its settling time of $15\mu\text{s}$ to 0.01% , and its cost is low enough to enable designs with one in-amp per channel.

B. AD620 Medical ECG

The low current noise of the AD620 allows its use in ECG monitors where high source resistances of $1\text{ M}\Omega$ or higher are not uncommon. The AD620's low power, low supply voltage requirements, and space-saving 8-lead mini-DIP and SOIC package offerings make it an excellent choice for battery-powered data recorders. Furthermore, the low bias currents and low current noise, coupled with the low voltage noise of the AD620, improve the dynamic range for better performance. The value of capacitor C1 is chosen to maintain stability of

the right leg drive loop. Proper safeguards, such as isolation, must be added to this circuit to protect the patient from possible harm. The equivalent circuit of the above manufacturer specified functioning diagram using the same AD620 and the Band pass filters and the Voltage shifter along with the appropriate voltage supply to AD620 and integrated circuits is as shown below.



In addition to the instrumentation amplifier it contains:

- 1) A Low pass filter with cut off frequency=150 Hz.
- 2) A High pass filter with cut off frequency=0.05hz.
- 3) A Voltage level shifter.

B.1 Gain Selection

The AD620 gain is resistor-programmed by R_G , or more precisely, by whatever impedance appears between Pins 1 and 8. The AD620 is designed to offer accurate gains using 0.1% to 1% resistors. Note that for $G = 1$, the R_G pins are unconnected ($R_G = \infty$). For any arbitrary gain, R_G can be calculated by using the formula:

$$R_G = \frac{49.4k\Omega}{G-1} \quad (1)$$

To minimize gain error, avoid high parasitic resistance in series with R_G ; to minimize gain drift, R_G should have a low TC—less than 10 ppm/°C—for the best performance.

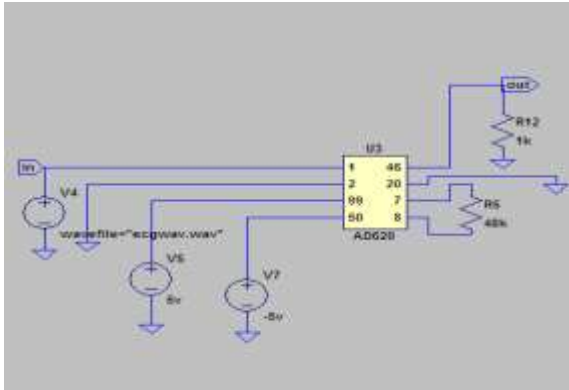
1% Std Table Value of $R_G(\Omega)$	Calculated Gain	0.1% Std Table Value of $R_G(\Omega)$	Calculated Gain
49.9 k	1.990	49.3 k	2.002
12.4 k	4.984	12.4 k	4.984
5.49 k	9.998	5.49 k	9.998
2.61 k	19.93	2.61 k	19.93
1.00 k	50.40	1.01 k	49.91
499	100.0	499	100.0
249	199.4	249	199.4
100	495.0	98.8	501.0
49.9	991.0	49.3	1,003.0

Rg for various gains

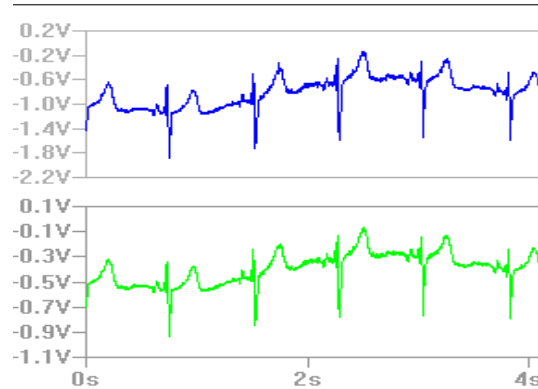
The design is simulated and tested using LTspice IV analog design environment.

Input stage: For simulation purpose the inputs to the instrumentation amplifier is given using a heartbeat wave file. The input is connected to non-inverting differential input and the inverting differential input is grounded. **Gain stage** The gain is chosen to be around $G=2$ to provide enough voltage range to make it suitable for passing to the subsequent stages. We have, Substituting for $G=2$, we get $R_g=48k$.

Output stage: According to the device characteristics, for the gain between 1-100 load resistances must be around 1-2kohm in order to prevent nonlinearity.



Circuit diagram AD620 Connections on Simulator



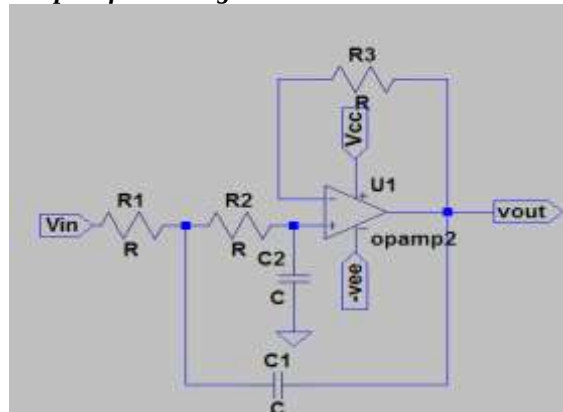
Simulation output (showing increase in gain)

Input signal amplitude(p-p)=730mv

Output signal amplitude(p-p)=1.45v

Gain=1.45v/730mv=1.98~2

B.2 Stage 2 –Second order Low pass filter Design.



The filter has a frequency response that falls off at the rate of 40db per decade above the upper cut off frequency. This steeper roll-off rate is achieved by using the $C2R2$ section together with feedback from the output via capacitor $C1$ to the junction of $R1$ and $R2$.

Circuit design:

Design starts with the selection of resistor $R1+R2$ small enough that no significant voltage drop occurs across it due to the op-amp input bias current. It is also desirable to keep $R1+R2$ as large as possible for maximum filter input impedance.

$$R1 + R2 = \frac{70mV}{Ib(max)} = \frac{70mV}{500nA} = 140k\Omega$$

$$R1 = R2 = 70k\Omega$$

$R3$ is made equal to $R1$ to equalize the resistance at the op-amp input terminals.

$$R3 = R1 + R2 = 140k\Omega$$

Capacitance $C2$ is calculated from

$$Xc1 = \sqrt{2} * R2 \text{ at } f_c$$

as $f_c=150hz$

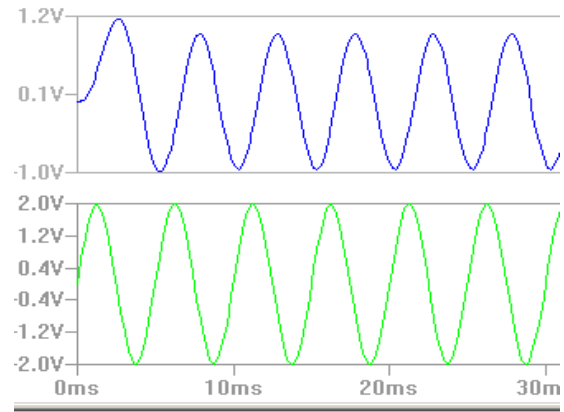
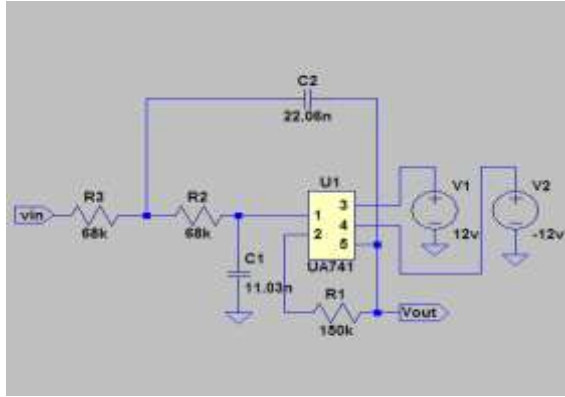
$$C2 = \frac{1}{2 * \Pi * f_c * \sqrt{2} * R2}$$

$$C2 = \frac{1}{2 * \Pi * 150 * \sqrt{2} * 68k\Omega}$$

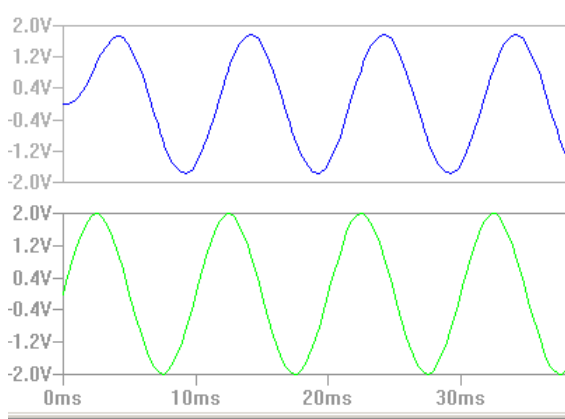
$$C2 = 11nf$$

$$C1 = 2C2$$

$$C1 = 2C2 = 22nf$$

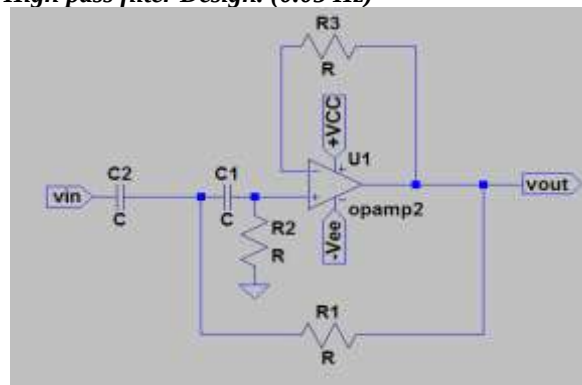


Simulation Output (For 200 Hz)



Simulation output (For 100 Hz)

B.3 Stage 3–Second order High pass filter Design. (0.05 Hz)



At high frequency, X_{c1} and X_{c2} are much smaller than $R1$ and $R2$, consequently, $C1$, $C2$, $R1$ and $R2$ have no significant effect on the circuit. The output voltage is then equal to the input, giving a voltage gain of 1.

Circuit design:

Design starts with the selection of resistor $R2$ small enough that no significant voltage drop occurs across it due to the op-amp input bias current. It is also desirable to keep $R2$ as large as possible for maximum filter input impedance.

$$R2 = \frac{70mV}{Ib(max)} = \frac{70mV}{500nA} = 140k\Omega$$

$$R1 = \frac{R2}{2} = 70k\Omega$$

R3 is made equal to R2 to equalize the resistance at the op-amp input terminals.

$$R3 = R2 = 140k\Omega$$

Capacitance C2 is calculated from

$$R2 = \sqrt{2} * X_{C1} \text{ At } f_c$$

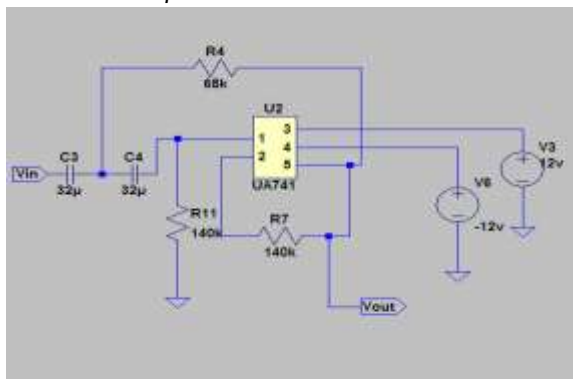
as $f_c = 0.05 \text{ Hz}$

$$C1 = \frac{1}{2 * \pi * f_c * \frac{R2}{\sqrt{2}}}$$

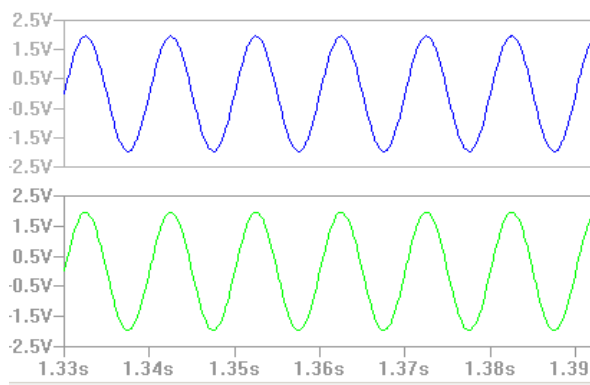
$$C1 = \frac{1}{2 * \pi * 0.05 * (\frac{68k\Omega}{\sqrt{2}})}$$

$$C1 = 32\mu f$$

$$C2 = C1 = 32\mu f$$



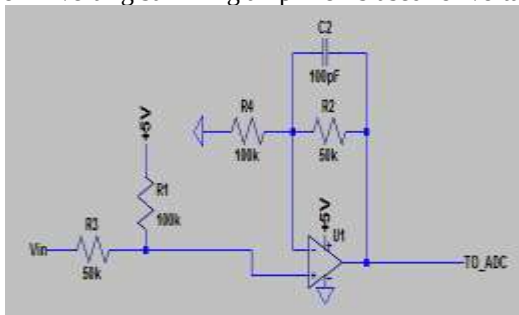
Circuit diagram second order high passes filter design. (0.05 Hz)



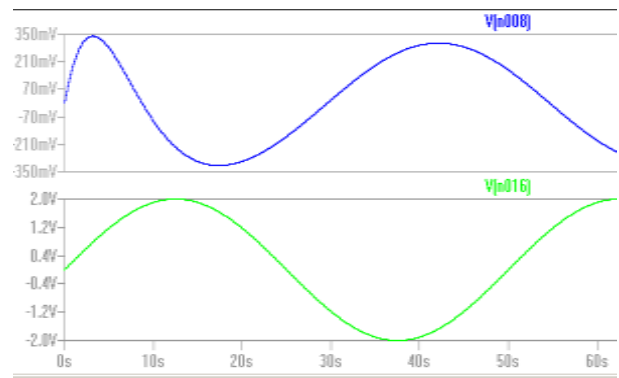
Simulation output (For 100hz frequency)

B.4. Stage 4– Voltage level shifter Design.

A non-inverting summing amplifier is used for voltage shifting.



Simulation of voltage shifter



Simulation output Simulation output (For 0.02 Hz frequency)



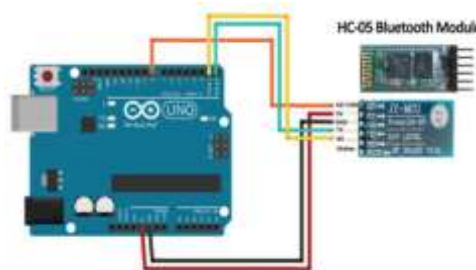
All Stage Waveform Simulation Result

IV. INTERFACING THE INSTRUMENTATION AMPLIFIER AND MICROCONTROLLER

The Arduino UNO microcontroller can accommodate up to 6 analog signals simultaneously, these analog pins on the board can be used to read data from external signal and can be processed continuously. The output from the Instrumentation amplifier is the “to ADC” terminal. This is the analogous data input to the microcontroller. An IDE (Integrated Development Environment) is employed to program the microcontroller. Then we can map the incoming signal of 0 to 5 volts to a value of 0 to 1023 level. Thus a voltage level of 5 mV is represented by an increase of one ADC level. Or we can say that the microcontroller provides a fine voltage resolution of 5mV per unit. The mapping function available within the IDE is used to scale up the lesser voltage signals or can even be used to scale down the incoming signal (but incoming signal is not to be more than 5Volts, as the microprocessor can only work on a max of 5Volts only). The analog pin can be from A0 to A5 any of one can be chosen as the input pin to the microprocessor. The output can be programmed as one of the 13 digital pins, and here we only use the digital pins 1 and 2 as receive and transmit pins for the Bluetooth module respectively. The USB from the computer being applied for the power supply to the Microcontroller is also used as the supply to the Bluetooth module, as the Bluetooth needs 3.3Vtg, this facility is being made use of from on board facility.

A. Using Serial Communication in Arduino Uno IDE

Used for communication between the Arduino board and computer or other devices. All Arduino boards have one serial port (UART or USART). Digital pins 0 and 1 are used to communicate as these pins are most recommended for the transmission and reception pins and also can be indicated using the on board Tx/Rx leds. These pins are not to be used as data pins for input or output. The IDE also allows the programmer to observe the serial port and at selectable baud rates and the serial port monitor window is a part of the IDE also used to send commands to external devices. The transmission and reception pins are to be connected to reception and transmission pins of the external device. The declarations are made in the Setup () Function for the same and is followed up once the Loop () function begins.



Connections to HC05 from Arduino UNO

The data is sent via Bluetooth and is received at the user end on smartphone.

V. CONCLUSION

The advancement in mobile and wireless health care solutions is contributing in different aspects of our lives that range from diagnosis to treatment of different diseases like cardiovascular disease. Android applications are also a part of diverse solutions that are offered as healthcare solutions together with a variety of health monitoring devices. This paper has covered mobile application development in the android Platform. The result of the project has achieved part of the objectives that were set at the beginning. However, the scope has been limited to establishing Bluetooth communication with the Smartphone device and receiving data. Appropriate algorithm for Heart BPM, body temperature needs to be developed as a part of future work. The communication mode can be elevated to WiFi level which can provide larger coverage area and more bandwidth with data security.

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