AHSANULLAH UNIVERSITY OF SCIENCE AND TECHNOLOGY

DEPARTMENT
OF
ELECTRICAL AND ELECTRONIC ENGINEERING

EEE 4238

Biomedical Instrumentation Lab

Table of Contents

Experiment 1 : To study the frequency response of electrode contact impedance.

Introduction

When applying electrical stimulation/current using surface electrodes, the body becomes part of an electric circuit as the current travels from one electrode to the other. The electrode-body surface interface offers a frequency dependent resistance, called contact impedance, which can be represented by an electrical equivalent circuit. Electrodes Figure 2: A simple equivalent circuit for a pair of electrodes applied to the skin Figure 3: Circuit connection for electrode contact potential measurement.

In the equivalent circuit shown right (figure 1):

Figure 1: Equivalent circuit for electrode contact impedance
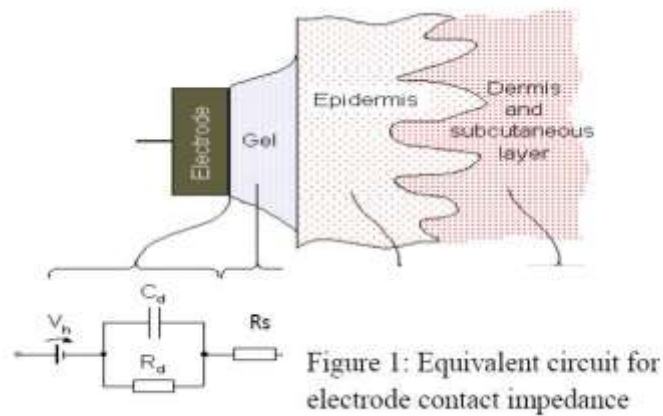
➢ Vh is the half-cell potential
➢ Rd and Cd makes up the impedance associated with the electrode-electrolyte interface.
➢ Rs series resistance associated with resistance of electrolyte and interface effect.

For a pair of electrodes applied to the skin and for alternating current the equivalent circuit can also be simplified as:
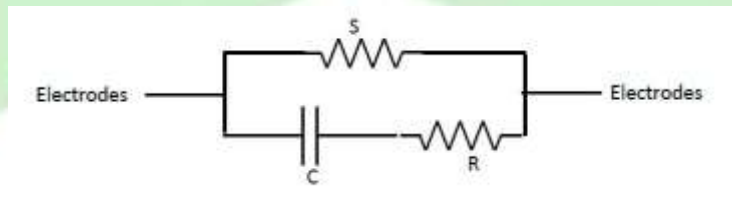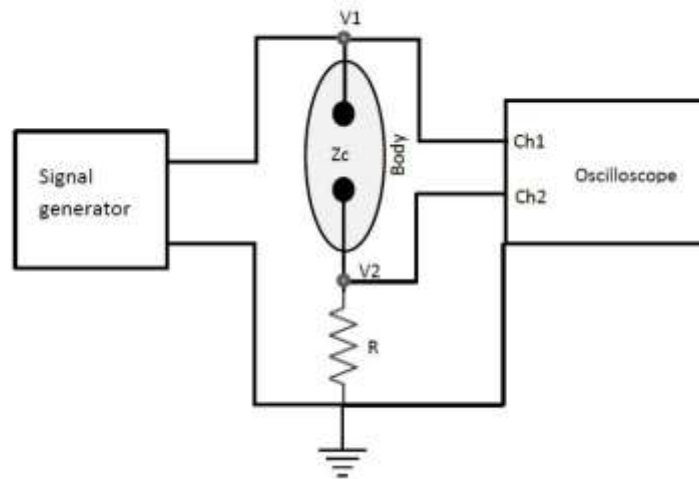

Equipment

Sine wave generator, Oscilloscope, Resistors, Surface electrodes, Jelly.

Methods

Apply a pair of electrodes to the forearm using jelly but no skin abrasion (unprepared skin).

1. Connect the circuit as shown in figure 3.

According to figure 3:

$$V_2 = \frac{V_1 * R}{Zc + R}$$

Or $$Zc = \frac{R(V_1 - V_2)}{V_2}$$

2. Measure the electrode impedance by applying a 1 V p-p sine wave at a range of frequencies from 10 Hz to 50 kHz (10, 100, 500, 1000, 5000, 10000, 50000, 100000, 500000, 1000000 Hz).

3. Repeat step 1-3 for another pair of electrodes on well prepared skin.

4. For each pair of electrodes also measure the any DC potential between the electrodes.

Analysis

1. Plot the electrode contact impedance against frequency both for unprepared and prepared skin on semi-log graph paper.

2. From the Impedance vs Frequency curves, determine the equivalent circuit for the electrodes on the assumption of a combination of two resistances and capacitor (figure 1). R can be determined from the high frequency part of the curve and S from the low frequency.

Frequency	Unprepared skin		Prepared Skin	
	V_2	Z_c	V_2	Z_c

Signature of the teacher:

Group No:

Experiment 2:   Introduction to Nerve Stimulator circuit

Objective:

Overall objective is to design a nerve stimulator and implement the circuit on breadboard. In the process the following circuits will be implemented 1. An astable multivibrator using 555 timer IC 2. A complimentary Darlington pair (Sziklai Pair) to amplify current from the astable multivibrator. 3. A step up transformer to get desired pulse voltage to stimulate nerve Theory: If a current of sufficient amplitude is passed between a pair of surface electrodes then muscles will contract. They contract because a stimulus is being introduced to the nerve fibers, which supply the muscles. If we use a short pulse of current, a graph can be drawn in which the current needed to cause stimulation is plotted against the duration of the current pulse. This graph (fig. 1(a)) shows that, unless an excessively high current is to be applied, a pulse lasting at least 50μs, and preferably as long as 2ms, is needed to stimulate the nerve fibers. The stimulation only occurs underneath one of the surface electrodes. The stimulation occurs underneath the electrode at which the pulse is seen as a negative pulse. We can explain this by considering what is actually happening to a nerve fiber when a current is applied. A nerve will be stimulated when the transmembrane potential is reversed by an externally applied current. If, where the current enters the nerve, the externally applied current flows from positive to negative, then the transmembrane potential will be increased, and where the current leaves the nerve, the transmembrane potential will be reduced. Stimulation will occur where the current leaves the nerve. If the positive electrode is called the anode and the negative the cathode, then stimulation starts underneath the cathode (fig. 1(b)).
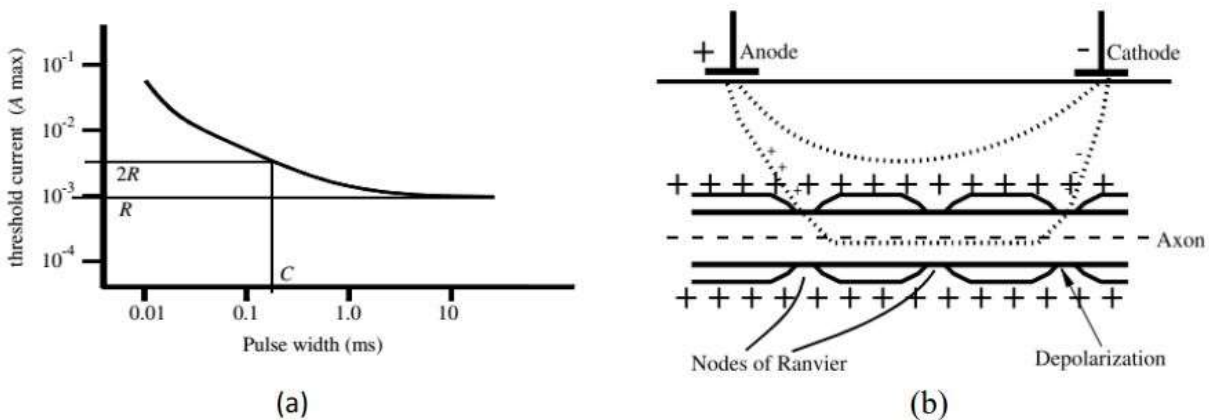

Fig. 1: (a) The current needed to cause muscle stimulation beneath a pair of surface electrodes on the arm is shown as a function of pulse width. (b) Stimulation occurs underneath the cathode, where the transmembrane potential is reduced.

An astable multivibrator is needed to produce such a pulse but will have very low current at output. Thus a complimentary Darlington pair is used in second stage to achieve high current

gain. Then the output is passed through a step up transformer to step up the pulse to desired voltage.

Equipment:

1. 555 timer IC
2. PNP and NPN transistor
3. Resistors
4. Capacitors
5. Diode
6. Step up transformer

Design & Assembly:

Assemble the circuit as shown in fig. 2. In the 555 timer IC, 8th pin and 1st pin of the IC are used to give power, Vcc and GND respectively. The 4th pin is RESET pin which is active low and is connected to Vcc to avoid accidental resets. 5th pin is the Control Voltage pin which is not used. So to avoid high frequency noises it is connected to a capacitor C' which has its other end connected to ground. Usually C' = 10nF. The Trigger (pin 2) and Threshold (pin 6) inputs are connected to the capacitor which determines the output of the timer. Discharge pin (pin 7) is connected to the resistor Rb such that the capacitor can discharge through Rb.
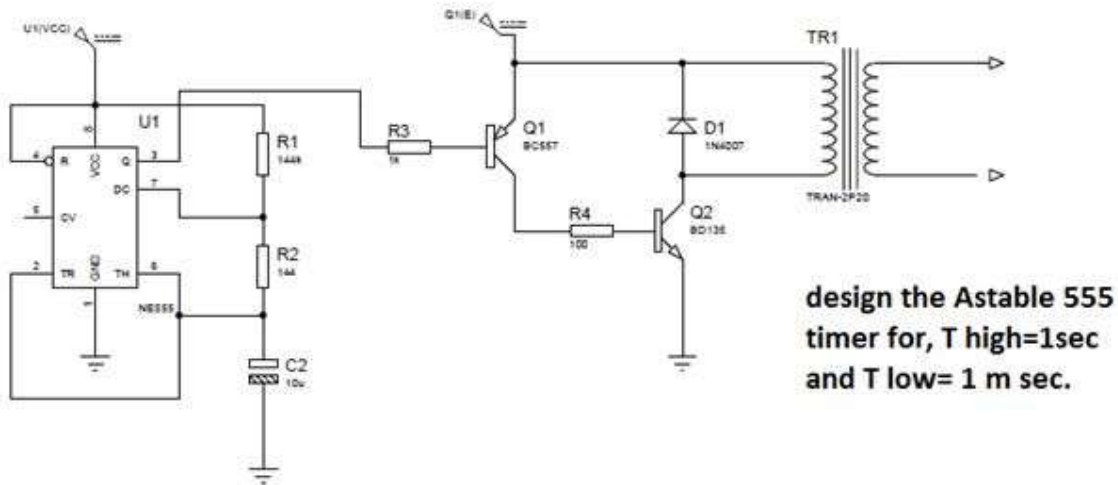

design the Astable 555 timer for, T high=1sec and T low= 1 m sec.

Fig.2: Circuit design of a nerve stimulator

Capacitor charges through Ra and Rb.

• Thigh = 0.693(Ra + Rb)C

Capacitor Discharges through Rb

- Tlow = 0.693RbC

- Output Frequency = 1/(Tlow + Thigh) = 1.44/((Ra + 2Rb) * C)

- Duty Cycle = Thigh/(Thigh + Tlow)

Calculate the values of Ra and Rb, according to your output waveform.

The output of this circuit is feed into the next section where complimentary Darlington pair provides high current gain. The step-up transformer then provides desired pulse voltage.

Results:

1. Calculate the Tlow and Thigh value according to practical values of Ra and Rb.

2. Find percentage of error for practical values of Tlow or Thigh with previous calculation.

3. Observe how much current gain is achieved after complimentary Darlington section.

4. Observe the pulse voltage after step up transformer section.

Discussion:

Discuss the problems, findings, shortcomings, outcome discrepancies of the experiment with theoretical explanation.

Experiment No 3: Introduction to Instrumentation Amplifier.

Introduction

An Instrumentation amplifier (Ins-amp) is basically a differential amplifier, but with high input impedance so that it can faithfully reproduce voltage signals from any source. The idea is that the voltage should be dropped across the input impedance of the amplifier. A differential amplifier is a basic circuit to pick up and amplify a small signal in presence of a relatively large background noise. However, it is only possible if the noise is the same at two electrical points, which carry the signal, with respect to a third point. That is the noise should be common to both the input terminals. To clarify, suppose a small signal, of the order of a few mV, appears between two electrical terminals. Now if an external noise with respect to a third terminal appears on both the above two terminals with equal magnitude and phase then it is possible to eliminate the noise. Instrumentation amplifiers are used in industry as well as in bioelectrical measurements.

Equipment List:

1. IC: LM324
2. Resistors: 1kΩ, 2.2kΩ, 10kΩ
3. A preset resistor, about 10kΩ to 20kΩ, preferably multi-turn type.

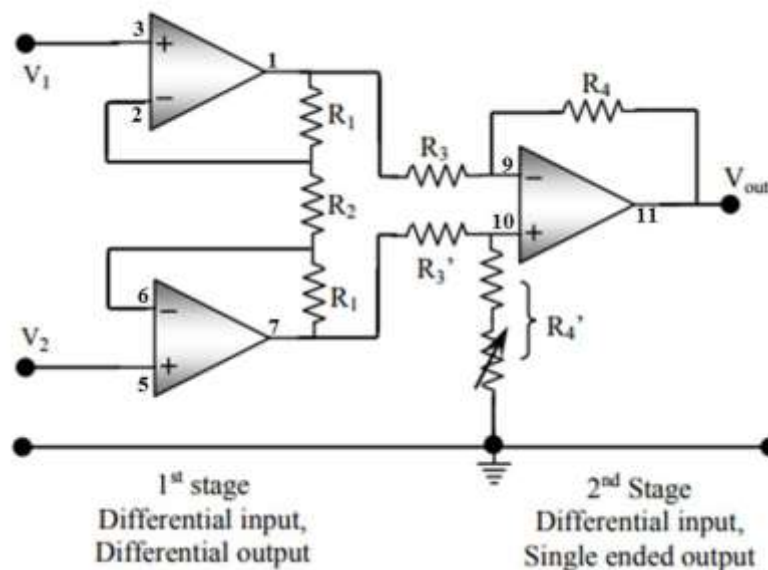**Theory of Instrumentation Amplifier**

Fig 1: Basic Bioelectric Amplifier

For a non-ideal instrumentation amplifier

$$\text{Vout} = A_d(V_2 - V_1) + \tfrac{1}{2}\,\text{Acm}(V_2 + V_1) \dots \dots (1)$$

Fig.1 shows the basic circuitry of an instrumentation amplifier consisting of 3 operational amplifiers (opamps). This is the most popular form of an instrumentation amplifier which has two inputs V_1 and V_2, and one output V_{out}, all referenced to ground or common (0V). Here we will perform the analyses based on ideal assumptions for the op-amps, and will also assume that A_{cm} is zero, initially. Here Vout ∞ $(V_2 - V_1)$, i.e., it is a differential amplifier where the output is proportional to the difference of two inputs. It has two stages as indicated in Fig.3. The first stage has a differential input and also has a differential output (double ended input, V_1 and V_2 and double ended output, V_1' and V_2'), while the 2nd stage has a differential input (double ended input, V_1' and V_2') and a single ended output V_{out}. In the circuit shown, Resistor R_4' has been shown to consist of a fixed resistor in series with a variable resistor. This is required to adjust the CMRR.

$$\text{CMRR} = (Ad\,/\,|Acm|) \dots \dots (2)$$

$$\text{CMRR in dB} = 20\log(Ad\,/\,|Acm|) \dots \dots (3)$$

Where Ad is the differential gain, Acm is the common-mode gain.

Analysis

In Fig.1, The first stage has both differential input and differential output given by,

$$V_2' - V_1' = A_1(V_2 - V_1), \dots \dots (4)$$

Where the Gain $\qquad\qquad A_1 = 1 + 2R_1/R_2 \dots \dots (5)$

The second stage has a differential input and a single ended output given by,

$$V_{out} = A_2(V_2' - V_1') \dots \dots (6)$$

where $A_2 = R_4\,/\,R_3$ and it is assumed that

$$R_4\,/\,R_3 = R_4'\,/\,R_3' \dots \dots (7)$$

The overall differential gain A_d is the product

$$A_d = A_1\,A_2 = (1 + 2R_2/R_1)\,(R_4\,/\,R_3) \dots \dots (8)$$

Introducing non-zero common mode gain

The common mode input voltage Vcm is given by the average of the two input voltages V_1 and V_2, i.e.,

$$Vcm = (V_1 + V_2)/2 \dots \dots (9)$$

If the common mode voltage gain Acm is not zero then we will have a contribution from Vcm as well, as given in Eq.1. In Eq.1, if the differential input voltage is zero, i.e.,

$$\text{if } V_2 - V_1 = 0, \text{ then } V_{out} = \text{AcmVcm} \dots \dots (10)$$

This gives a way of measuring Acm.

Design considerations for CMRR

CMRR should be as large as possible, and for an ECG amplifier, it should be > 60 dB, and for EEG it should be >100dB. In the 2nd stage of the amplifier two ratios, R_4 / R_3 and R_4' / R_3' , associated with the two inputs of the differential amplifier are assumed to be equal. This is essential to get a high value of CMRR. However, in practice it is almost impossible to have two exactly equal fixed resistors, so this condition is achieved manually, through experiment. Therefore, R_4' is configured as a variable resistor in series with a fixed resistor, and a maximum CMRR is achieved through experimental adjustment of the variable resistor, described later.

Adjustment for maximum CMRR and measurement of Acm
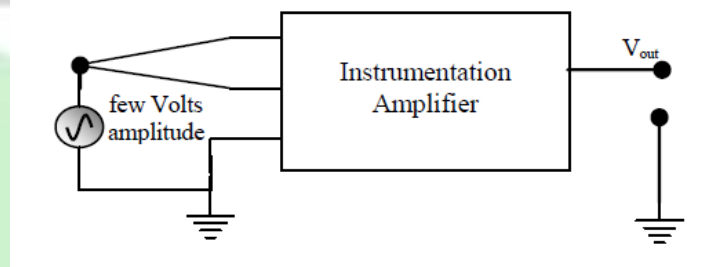

Fig 2: Connection for adjusting and measuring CMRR

The common mode rejection ratio has to be maximized by varying the preset resistor in R4' so that Eq.7 is satisfied. For this adjustment we need a common mode input, i.e., the same signal applied to both the inputs. The necessary arrangement is shown in Fig.2, with both inputs connected to the signal generator while the other end of the signal generator is connected to ground. Apply a few volts amplitude at 1 kHz and observe the output on an oscilloscope. Adjusting R4', minimize the output as far as possible. You may have to increase the signal input (but should not be more than power supply voltage to the circuit) and increase the input sensitivity of the oscilloscope to get a good adjustment since the output will be very small. Keeping this set-up, measure the output and the input voltage amplitudes (or peak to peak voltages) to obtain the common mode gain as

$$Acm = Vout / Vcm$$

Measure Acm at different frequencies, starting from the minimum that you have on the signal generator to the maximum. In this measurement check the input signal amplitude at each measurement frequency. Keeping this amplitude constant will make Vout proportional to the gain, and simplify calculations.

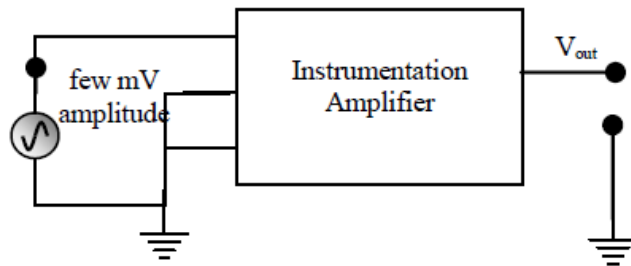## Measurement of differential gain with frequency

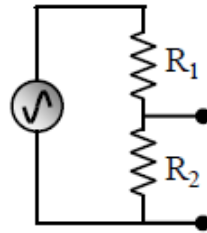**Fig 3:** Connection for measuring differential gain

Fig 4: Voltage divider for attenuating signal.
R$_2$ should be less than R$_1$.

To measure the differential gain, Ad, connect one of the differential input terminals (any one) to ground as shown in Fig.3 and apply about 10 mV (amplitude) signal to the other input terminal. If the signal generator cannot go down to 10 mV, make an attenuator yourself as indicated in Fig.4. In Fig.3 the instrumentation amplifier has been set up essentially as a single ended amplifier, and it gives the differential gain. Measure this differential gain over a range of frequencies as before (take the same frequency values as you had used for Acm above, this will make the CMRR plot in the next section easier). As above check the input signal amplitude at each measurement frequency.

Keeping this amplitude constant will make Vout proportional to the gain, and simplify calculations. Plot Ad as a function of frequency to see the frequency response of the amplifier.
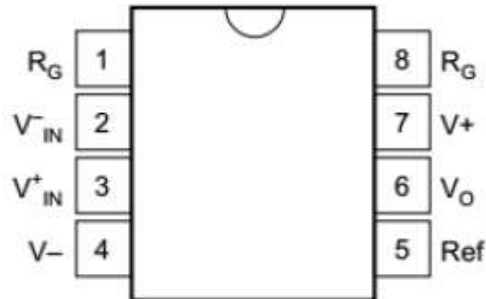
Determination of CMRR with frequency

From the above measurements obtain the CMRR in linear and dB scales at each measured frequency using Eq.2 and Eq.3. Plot both to see the frequency response of CMRR.
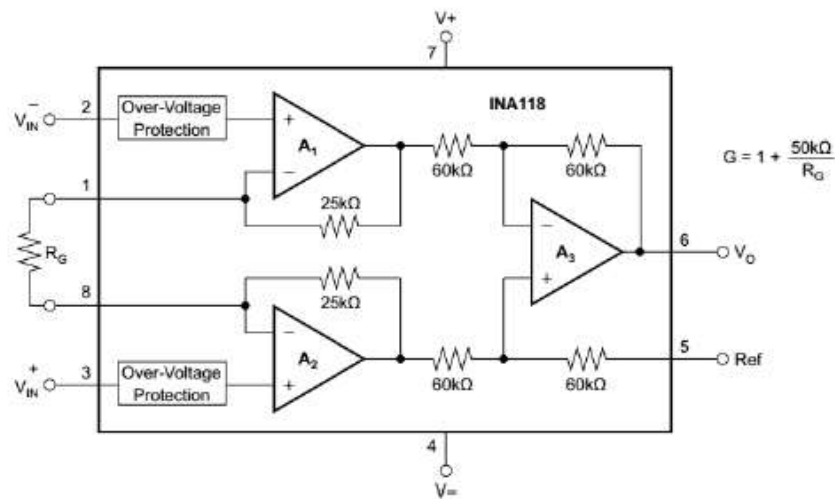
Report

The results of this experiment are the values of differential and common mode gains and the CMRR that you obtained at different frequencies. Present the results in graphical form in the main text and place the raw tables in an Appendix at the end. Although indicated above, to summarize, you need the following plots from your measured data.

1. A_d (linear scale) against frequency (log scale)
2. The above with gain in dB.
3. CMRR (in linear scale) with frequency (log scale)
4. The above with CMRR in dB.

Instrumentation Amplifer IC (INA118):

Pin layout of INA118

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

Internal circuit of INA118

6.1 Absolute Maximum Ratings

over operating free-air temperature range (unless otherwise noted) [1]

		MIN	MAX	UNIT
	Supply voltage		±18	V
	Analog input voltage		±40	V
	Output short-circuit (to ground)		Continuous	
	Operating temperature	−40	125	°C
	Junction temperature		150	°C
	Lead temperature (soldering, 10 s)		300	°C
T_{stg}	Storage temperature	−40	125	°C

Experiment No 4: Circuit design and breadboard implementation of an optical isolator.

Objective

Isolation amplifiers can be used to break ground loops, eliminate source ground connections, and provide isolation protection to patient and electronic equipment. In a biopotential amplifier, the main purpose of the isolation amplifier is the protection of the patient by eliminating the hazard of electric shock resulting from the interaction among patient, amplifier, and other electric devices in the patient's environment, specifically defibrillators and electrosurgical equipment. It also adds to the prevention of line frequency interferences.

Theory

An Optocoupler, also known as an Opto-isolator or Photo-coupler, are electronic components that interconnect two electrical circuits by means of an optical interface. The basic design of an optocoupler consists of an LED that produces infra-red light and a semiconductor photo-sensitive device that is used to detect this emitted infra-red light. Both the LED and photo-sensitive device are enclosed in a light-tight body or package with metal legs for the electrical connections as shown.

An optocoupler or opto-isolator consists of a light emitter, the LED and a light sensitive receiver which can be a single photo-diode, photo-transistor, photo-resistor, photo-SCR, or a photo-TRIAC and the basic operation of an optocoupler is very simple to understand.

Equipment List:

1. LM324 : 1 pc
2. PC817 : 1 pc
3. 500Ω resistance : 2 pcs
4. 1kΩ resistance :4 pcs
5. 2.2 kΩ resistance : 1pc

Photo-transistor Optocoupler
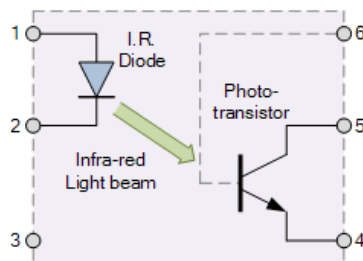
Assume a photo-transistor device as shown. Current from the source signal passes through the input LED which emits an infra-red light whose intensity is proportional to the electrical signal. This emitted light falls upon the base of the photo-transistor, causing it to switch-ON and

conduct in a similar way to a normal bipolar transistor. The base connection of the photo-transistor can be left open for maximum sensitivity or connected to ground via a suitable external resistor to control the switching sensitivity making it more stable.

When the current flowing through the LED is interrupted, the infra-red emitted light is cut-off, causing the photo-transistor to cease conducting. The photo-transistor can be used to switch current in the output circuit. The spectral response of the LED and the photo-sensitive device are closely matched being separated by a transparent medium such as glass, plastic or air. Since there is no direct electrical connection between the input and output of an optocoupler, electrical isolation up to 10kV is achieved.
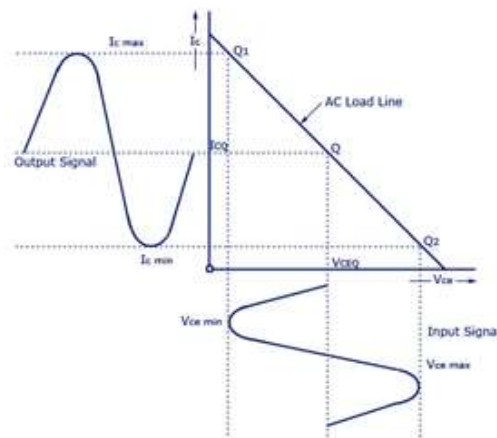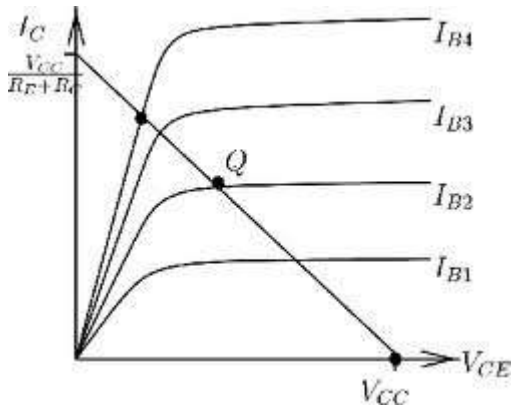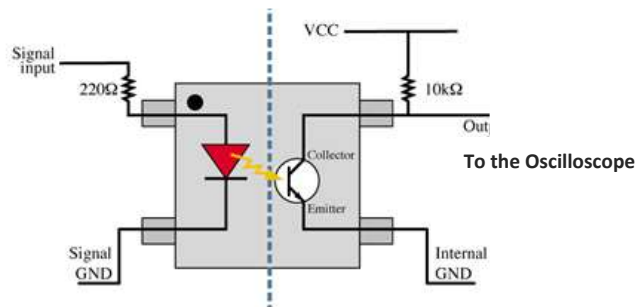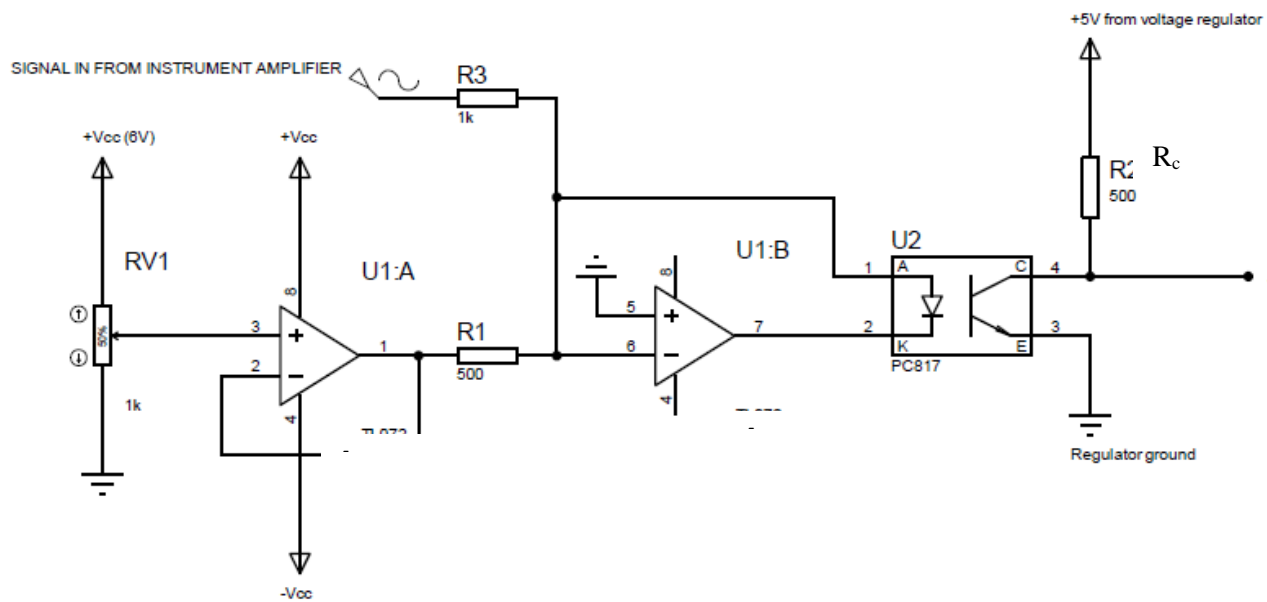
Operating the isolator in its Q point:

Figure: I_c vs V_{ce} curve for transistor

Experimental Setup

Data table

Collector Resistance Value	$V_{in}(V_{dc})$ Volt	V_0 (V_{ce}) Volt	I_d mA	I_c [$I_c=(V_{cc}-V_{ce})/R_c$] mA
	0			
	0.5			
	1			
	1.5			
	2			
	2.5			
	3			
$R_c= 500\Omega$	3.5			
	4			
	4.5			
	5			
	0			
	0.5			
	1			
	1.5			
	2			
	2.5			
	3			
$R_c= 1k\Omega$	3.5			
	4			
	4.5			
	5			
	0			
	0.5			
	1			
	1.5			
	2			
	2.5			
	3			
$R_c= 2.2k\Omega$	3.5			
	4			
	4.5			
	5			

Group No:

Signature of the Teacher

Report

1. Draw the I_c vs V_{ce} curve in a graph for different values of Collector resistance.
2. Draw the load line and find the Q point.
3. Discuss other methods of Isolation.
4. Mention some important reasons for Isolation.

**Experiment 5**: Introduction to Electrocardiograph (ECG).

Objective:
1. Design an amplifier for ECG analysis
2. Analyze the frequency response of the amplifier
3. Implement the amplifier and monitor ECG

Theory:
The Electrocardiography (ECG) is the process of visualizing the electrical activity of the heart over a period of time. These electrical activities result from the depolarization of the myocardium in a certain pattern. The electrical activities are monitored using surface electrodes over the body.

As we know, the heart has a certain rhythm over a course of time in which it contracts and expands, pumping blood out to the body. This periodic rhythm is caused by depolarization of some specific points on the myocardium, namely the Sino-Atrial Node (SA node), the Atrio-Ventricular Node (AV node) and the His-Purkunje Fibers. These depolarizations create electrical impulses. As the human body is a volume conductor, the effect of any electrical impulse inside the body can be detected from the surface.

Figure 01: Nodes in the myocardium
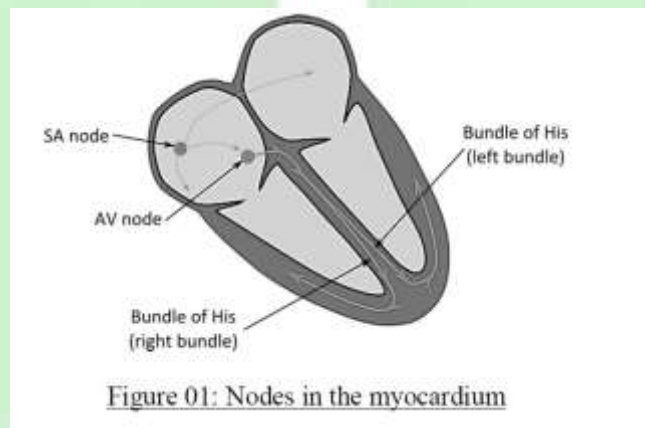
The electrical activity of the heart can be modelled as the movement of an electrical dipole. A dipole consists of two opposite charges. A cardiac vector is the line joining the two charges. Although the heart does not consist of two opposite charges, the electric field created due to the depolarization of the heart resembles that of an electric dipole. To describe a vector by its length in three dimensions, three directions are considered which are 60° apart. These directions form a triangle known as Einthoven's triangle. The direction from one point to another is known as a Lead, hence having three leads.
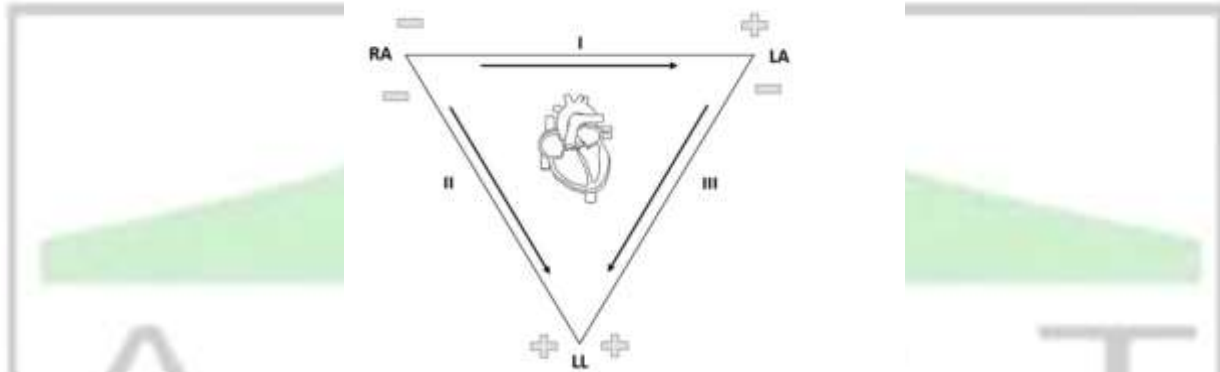
Figure 02: Einthoven's triangle and different Leads.

The body and the heart are three dimensional, so the electrical signals recorded from the skin will vary depending upon the position of the electrodes. For diagnostic purpose, some standardization of electrode positioning is needed. This is done by placing the electrodes on the limbs. The three points of the Einthoven's triangle are the three limbs where we put the electrodes. The three leads are thus considered as

Lead I : from Right Arm (-ve) to Left Arm (+ve)

Lead II : from Right Arm (-ve) to Left Leg (+ve)

Lead III : from Left Arm (-ve) to Left Leg (+ve)

As the lower half of the body can be considered equipotential, the bottom point in the Einthoven's triangle can be considered as any point in the lower thorax. For simplicity, we use the left leg for this point.

The ECG has an amplitude range of 0.5 mV – 3.0 mV and a frequency range of 1Hz to 50Hz. To effectively monitor the ECG, an amplifier is required with a gain of approximately 1000. For this, an Instrumentation amplifier with a high common mode rejection ratio is used. For patient safety, optical isolation is needed. In addition, a bandpass filter is also needed to suppress the contribution of the DC biasing and the 50Hz default supply line frequency. For proper filtering, the bandpass filter should have a lower cutoff of 0.1Hz and a higher cutoff of 500Hz (one decade more than the desirable frequency range on both sides). The cutoff frequency of the filter can be calculated using
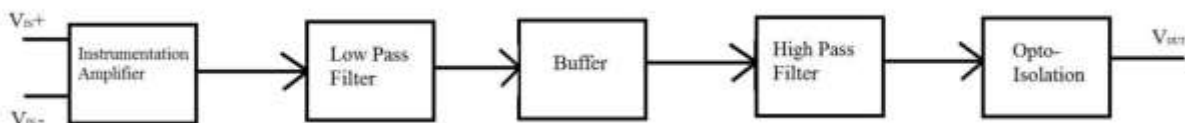
$$fc = \frac{1}{2\pi RC}$$

Figure 03: Block diagram for ECG amplifier.

Equipment: (Specify the model and quantities)
1. Op-Amps
2. Resistors
3. Variable resistors
4. DC power source
5. Signal Generator
6. Oscilloscope
7. Opto-isolator

Circuit Setup:

Set up the circuit given according to the block diagram given in Figure 03. For simplicity we will only observe Lead I, for which set the –ve input of the Instrumentation amplifier in the Right arm, the +ve input of the amplifier in the Left arm and connect the electrode to the Left leg to the ground. Before connecting the electrodes to the specimen body, make sure that your circuit is properly isolated to provide faulty current flow from the monitoring end. Design each block individually and test with generic signals before applying bio-signal.

Data Table:

Analyze the frequency response of the bandpass filter.

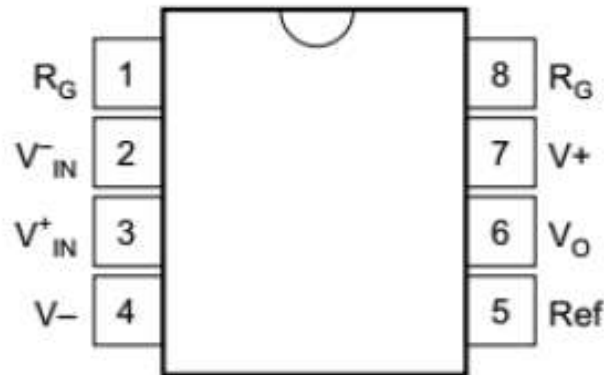| Frequency (Hz)	V_{IN} (mV)	V_{OUT} (V)	Gain

Results:

- Include the graphs for the frequency response of the filter.
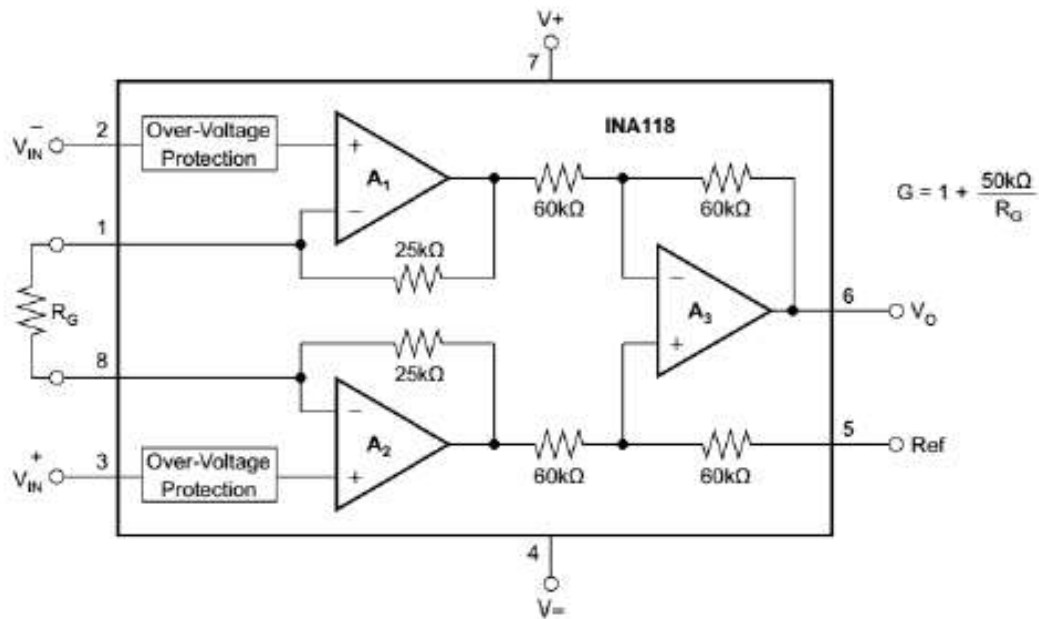- Include graphs for Lead I (take a snapshot of the oscilloscope with proper magnitude)

Discussion:

Discuss the problems, findings, shortcomings, outcome discrepancies of the experiment with theoretical explanation.

Instrumentation Amplifer IC (INA118):
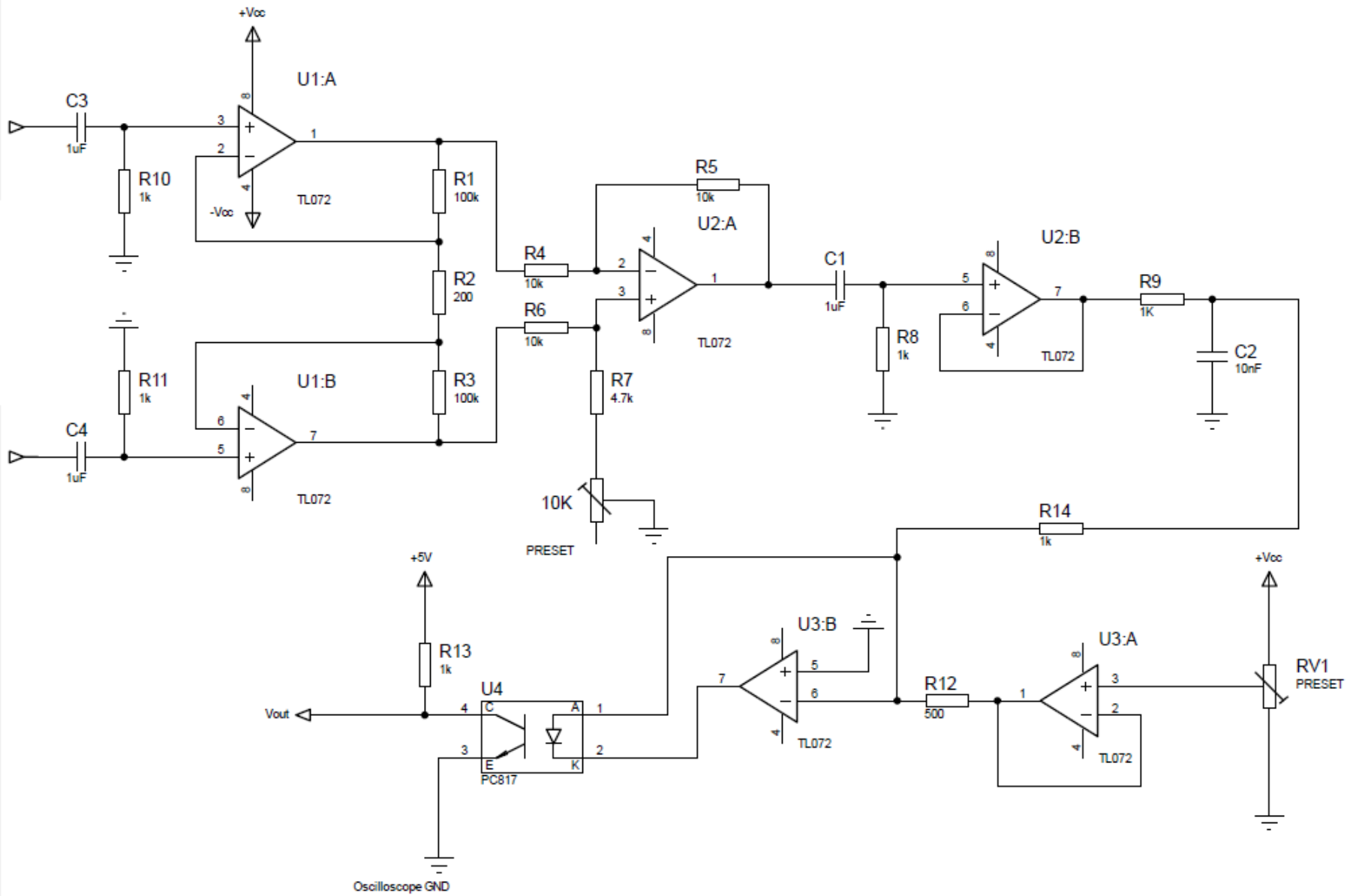

Pin layout of INA118

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

Internal circuit of INA118

6.1 Absolute Maximum Ratings

over operating free-air temperature range (unless otherwise noted) [1]

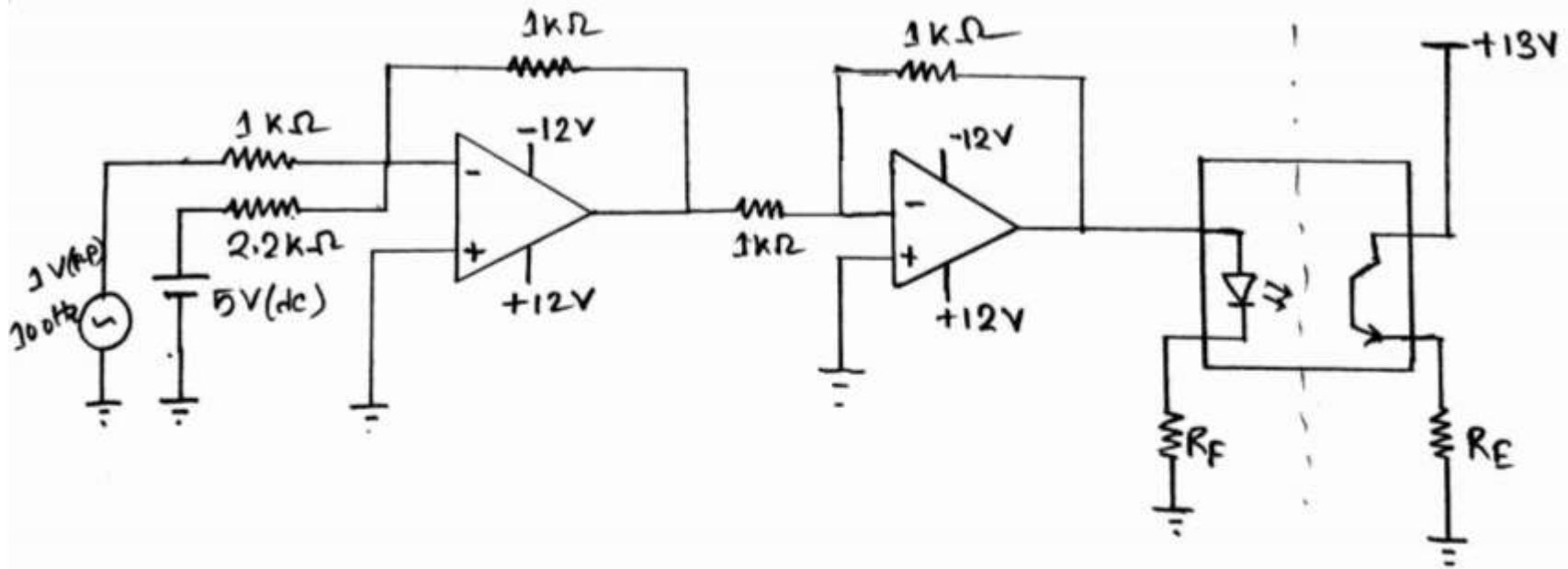|		MIN	MAX	UNIT
	Supply voltage		±18	V
	Analog input voltage		±40	V
	Output short-circuit (to ground)		Continuous	
	Operating temperature	−40	125	°C
	Junction temperature		150	°C
	Lead temperature (soldering, 10 s)		300	°C
T_{stg}	Storage temperature	−40	125	°C

Figure 02: Circuit diagram of the experiment

Experiment 6:   Introduction to Electromyogram (EMG)

Introduction:

EMG stands for Electro Myo Graphy – meaning visualisation of the electrical signals of muscles (Greek- Myos muscle). A mass of muscle that we feel at different places in our body consists of many muscle fibres. At rest each muscle fibre has a charge separation across its covering membrane (outer surface positive with respect to the inside of the fibre) giving rise to a polarised state. This is reversed or depolarised in segments of the fibre when an action is initiated, which after a relaxation period automatically reverts back to the resting polarised state. This contributes to an action potential which travels along the fibre. At any instant of time this charge separation contributes to an electric dipole vector. Similar dipole vectors are produced in all other muscle fibres in the group. A muscle is usually controlled by nerve signals (nerve action potentials) sent from the brain. When nerve action potentials reach the muscle, muscle action potentials are initiated through the neuro-muscular junctions which gives rise to EMG. Such an EMG is also called voluntary EMG since it is the result of a voluntary action of the person.
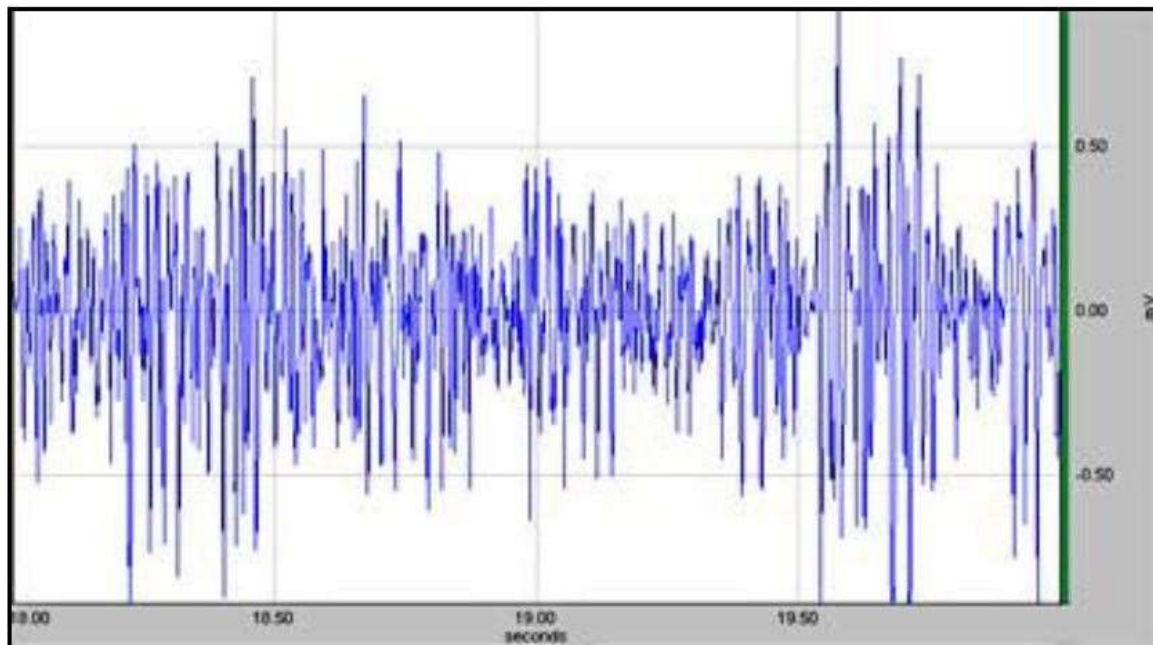

Fig.1: Typical EMG waveform on voluntary contraction

The body is a volume conductor, consisting of electrolytes, having both positive and negative free ions. Therefore, within the body the dipole vectors give rise to an external current which change with time. When a current flows through a resistor, a potential drop occurs across it. Similarly due to the external current distribution in the volume conductor, potential differences are produced between any two points. This potential difference can be picked up using two electrodes placed at two suitable points outside, which when plotted against time gives the well-known EMG as shown in Fig.1.

A metallic electrode connected to a mass of electrolyte such as the human body contributes to a dc contact potential of the order of a volt. Since both electrodes are usually of the same metal, the dc contact potentials have opposite directions in the circuit loop which should cancel out ideally. This cancellation is not total and some low frequency ac potential may be present due to differences in slow acting chemical reactions under the two electrodes. However, since EMG is an ac greater than 10Hz, the dc or the low frequency potentials may be eliminated using an appropriate high pass filter, which has indeed been done inside the EMG amplifier.
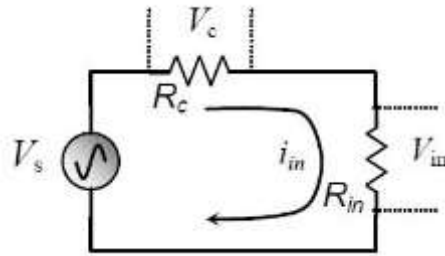

Fig.2: Voltage division between contact impedance and input impedance of the EMG amplifier

The contact impedance for skin surface electrodes is also important as discussed with the help of Fig.2. When connected to the input of an amplifier, the source voltage *vs* (EMG here) is dropped across both the contact impedance *Rc* and the input impedance *Rin* of the amplifier. We want Rc << Rin such that almost the whole of the source voltage is dropped across *Rin*. Besides, due to ground loop currents, some mains ac may be dropped across these contact impedances increasing the 50Hz noise as explained in the next section. Therefore, we have to take care that the contact impedances are minimum. This is very important for the 'Common' electrode which has a greater contribution to common mode 50Hz noise. Therefore, the common electrode is usually made of larger contact area compared to the active potential pick-up electrodes. Typically the contact impedances vary from about 5kΩ to about 50 kΩ at low frequencies depending on the metal type and the skin preparation. Human skin surface is composed of dead horny cells contributing to large impedance. Besides, there may be oil or grease increasing the impedance further. Therefore, before electrodes are applied, the skin should be rubbed well using alcohol, preferably with some fine abrasive to remove the dead cells, and conduction electrode gel should be applied to the electrodes before attaching to the skin.

Interference and noise, necessity of differential amplifier & filters

EMG is usually associated with a large noise voltage, mostly from the mains electrical lines at 50Hz (60 Hz in some countries). How this noise gets through is explained below.
The main source of noise in an EMG equipment is the 50Hz mains. Fig.3 shows how it gets into an amplifier. At the street poles, the transformer that supplies 220V to our premises, one of the secondary terminals is connected to the underground water layer, the ***Ground*** of our electrical

system. The underground water layer can be considered to be a good conductor of electricity and is the common Ground for all appliances and circuit connections. This ground connected secondary terminal is called the *Neutral* indicating that if we touch it while standing on the ground we would not get a shock since there is no electrical potential difference between the two points of touch. On the other hand if we touch the other secondary terminal, called the *Live*, while standing on the ground, we will get a shock since we will be in between a 220V ac potential.


Fig.3: Source of 50Hz noise
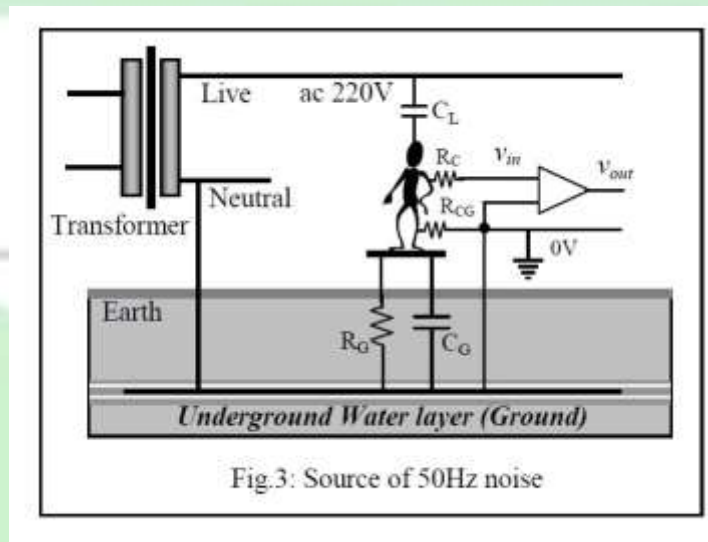
Our body has two electrical connections to the *Ground*, a very high resistance RG, which is the leakage path from our body to the Ground, and a capacitance CG in parallel. There is also a capacitance CL between our body and the *Live* terminal. In this electrical system, our body, full of electrolytes, acts as a conductor with low impedance. Now CL and CG (with RG in parallel to the latter) form a voltage divider and therefore without any other electrical connections, our body is at a substantial 50Hz potential (typically 20V) with respect to the *Ground*. The live wire is not only in the house wiring, it comes to equipment that we are handling, and therefore is quite close to us. Depending on the proximity of the live wire, we can have tens of volts of 50Hz ac on our body, which can be perceived when we touch the tip of an oscilloscope probe without touching the ground.

At our premises a third connection is brought out, side by side to the two mains ac lines, by placing a metallic plate or rod deep into the ground. The idea is that it will connect to the underground water layer (which is actually done so in large buildings through ground boring, but not so in smaller constructions), and this is the third connection in a 3-pin mains socket, called the *Earth* or *Ground* connection. If an electrode is attached to our body which is directly connected to the ground point, a contact resistance *RCG* at the electrode skin interface comes into play which is relatively small, and the 50Hz ac potential on our body drops to a low value (voltage dropped across *RCG*). If an EMG electrode is connected with a good skin preparation, *RCG* can be of the order of $10k\Omega$ and the 50Hz potential on the body, with respect to the ground, comes down to about 10 or 20 millivolts. This is low, but not so when we compare with the EMG signal which is only about 0.5mV in amplitude. So if we want to detect EMG signal using

an electrode at any point on our body (where the contact resistance has been shown to be *RC*) and connect it to the input of a normal single ended amplifier (having one input), the net input signal *vin* will consist of both the EMG signal (~0.5mV) and the 50Hz noise (~10mV) dropped across *RCG*. It is easy to see that the noise is about 20 times larger and it will be impossible to discern the EMG signal under such a situation. Besides, the frequency content of EMG picked up using skin surface electrodes is approximately between 10Hz and 3000 Hz, so filtering out such a large 50Hz signal will distort the EMG signal itself. Therefore normal amplifiers with one input port that are used in most normal applications, such as in radio and audio applications, cannot be used to amplify bioelectrical signals like EMG.

Fortunately, this noise voltage is almost the same all over the body since the body resistance is very small compared to series reactances and resistances in the 50Hz current loop. Therefore a differential amplifier may be used to eliminate this noise. This amplifier has two inputs with respect to a common terminal (0V reference), and the output is proportional to the difference of the two inputs, as shown in Fig.4. The two input terminals are connected to electrodes placed at two points on the muscle mass while a third electrode, placed at any other point on the body, is taken as the common terminal. Since the 50Hz noise has the same amplitude at both the input electrodes with respect to the common terminal as mentioned above, on subtraction this noise is eliminated while the difference signal, which is the EMG in this case, appears at the output, amplified through a differential voltage gain of *Ad*. The same 50Hz noise appearing at both the inputs is known as the *Common Mode* signal *(vcm)*. Ideally the common mode signal should get totally eliminated in a differential amplifier, and the relationship is expected to be,
RCG

Fig.4: Differential input of amplifier
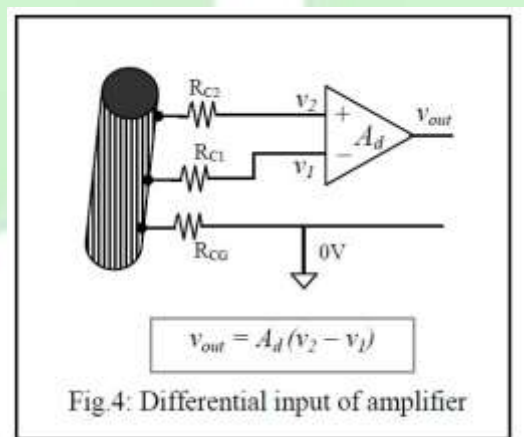
$$Vout = Ad\,(V2 - V1).\ \ldots\ \ldots\ (1)$$

Practically the common mode signal cannot be totally eliminated, a small non-zero common mode voltage gain *Acm* is present and the equation is modified as,

$$Vout = Ad\,(V2 - V1) + Acm\,Vcm\ \ldots\ \ldots\ (2)$$

We would want *Acm* to be as small as possible with respect to *Ad*. The ratio of *Ad* to *Acm* is defined as the Common Mode Rejection Ratio (CMRR), i.e.,

$$CMRR = Ad/Acm \ldots \ldots (3)$$

Obviously we would want CMRR to be as large as possible. Usually it is expressed in dB units given by,

$$CMRR = 20 \log Ad/Acm \ldots \ldots (4)$$

The 50Hz common mode noise needs to be reduced to about one hundredth of the maximum size of the signal, i.e., about 0.005mV for EMG. That is, a 10mV 50Hz noise has to be effectively reduced by 2000 times (or by 63dB). This you may achieve through better skin preparation, keeping a distance from mains connected equipment, keeping the lead wires as close as possible to minimise the area of the loop formed by these wires (to minimise emf generated due to magnetic fields created bu transformers, motors, etc.).

There is a safety aspect necessary with all hospital based electrical equipment. In hospitals there may be situations where catheters are connected directly to heart muscles. If a small current, of the order of 50μA at 50Hz passes through such a directly connected electrode, the heart may go into a state called defibrillation in which the rhythmic contraction is lost, resulting in loss of blood being pumped, causing eventual death. This is called 'microshock' in contrast to normal shock that we receive by touching a mains live cable by hand, where the dry skin provides a large resistance (about 30kΩ) limiting the current. Besides, the current distributes throughout the body having only a minute fraction passing through the heart itself. Thus in such a situation about 75mA (more than 1000 times the microshock value) is needed to cause defibrillation and death. Although most patients in hospitals do not have a catheter connected to their heart, we cannot take risk even if one patient out of a thousand have this dangerous situation. Therefore, all circuitry connected to the patient should be electrically isolated from the mains connected equipment such that not more than 10μA passes through the patient even in case the live 220V wire gets shorted to the body (ground) or to any other circuit part of the mains connected equipment.

There is another source of noise voltage. Since the EMG amplifier has to be connected to an oscilloscope or a computer which is driven by mains 50Hz, through capacitive coupling this 50Hz noise gets into the input of the amplifier.

The EMG amplifier supplied has been designed to tackle the above problems.

The Experiment, Methods and Measurements

Fig.5 shows the way to connect the EMG amplifier to a human subject on one side, and to a computer on the other through its USB port. The pad with two active electrodes is to be placed on the belly of a muscle, along its length, while the larger common electrode should be connected nearby at any point, preferably at a point at almost equal distances from the two active electrodes. It is better to keep the common electrode away from places with fatty layers underneath, because fat layers will increase the electrode impedance.

Prepare the skin well by rubbing with an anti-greasing agent like alcohol (liquid Savlon may do) with a mild abrasive pad, like a strong tissue paper. Let it dry. Now put a drop of electrode jel

onto the electrodes each and attach them to the hand muscle as shown using a Velcro strap supplied.
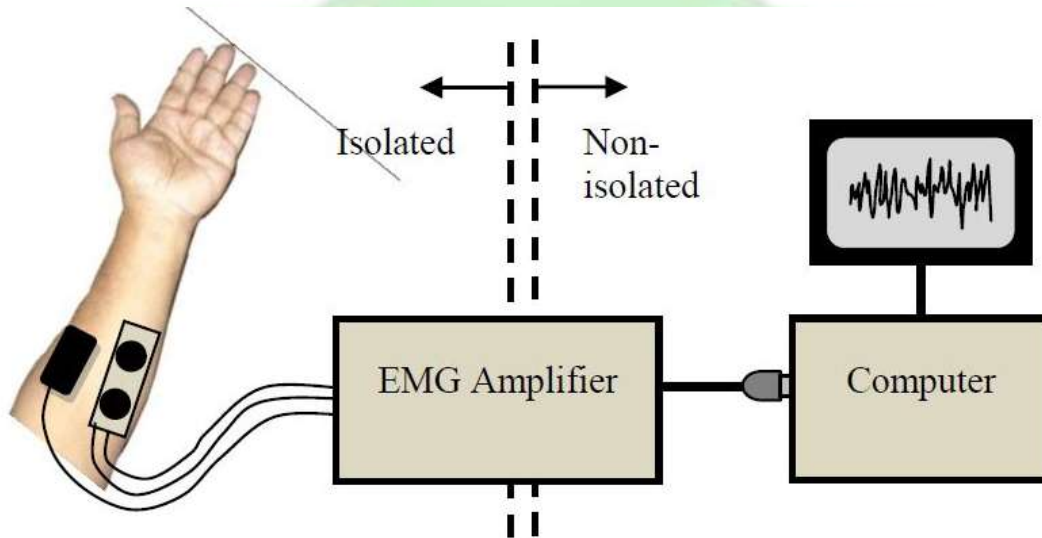

Fig.5: Connections for measuring EMG

Run the EMG programme on the computer and click on to the 'EMG' tab. Most instructions will be available on-screen as you go through the steps. You will see a display of several seconds duration in the graphical window.

With the hand muscles relaxed try to see that you get an almost flat response. You may get a periodical 50Hz noise. Try to move yourself away from the table or other equipment as much as possible, you will find a minimum of this noise at a certain arrangement.

Now grip your fingers into a strong fist. You may try holding a cylindrical object of about 3cm diameter. You will see the EMG signal on the screen. Get an estimate of the average maximum amplitude using a plastic scale placed on the screen. You do not need to be very accurate, an eye estimation will do for this laboratory experiment (you may do it accurately through software analysis of acquired data). Now reduce the strength of the grip and again measure the average maximum amplitude. Note the values. You may fix the electrodes on other muscles (getting information from an anatomy book) and see differences, if any. Now do the same with another subject, note the values and see the difference with yours. Of course here you do not have a measure of the gripping force, but instruments are available that can measure such forces.

Results and Presentations

Make a table with one column showing average maximum amplitude on light muscle contraction and another column showing that for strong contraction. Do this for several subjects and enter these in different rows.

You may similarly make another table for EMG from different muscles from the same subject.

Discussions

The discussion should be separated into three sections, on theory and assumptions, on methods and experiments, and on the results, but without any headings.

Under the first section you may include the following:

1. EMG obtained using surface electrodes gives a gross evaluation of a whole muscle mass. The signal picked up is a combination of action potentials of all the fibres in the contracted muscle, however, the ones nearer to the electrodes contribute more.

2. The signal may be analysed in different ways, such as cumulative integration of amplitude, FFT, etc., to get more information.

3. Since a large part of the EMG signal falls in the audio frequency range, the signal can be fed directly into an audio amplifier to hear the sound pattern.

Under the second section you may include the following:

1. Skin should be adequately prepared to get low values of contact impedances under the electrodes, otherwise 50Hz noise will increase.

2. Only a little amount of conducting gel should be used on the electrodes. If the gel spreads there is a chance that it will create a shorting path on the skin, thus reducing the EMG significantly. If the electrode is repositioned, the gel on the skin due to the previous placement should be wiped off with alcohol or savlon and dried.

3. If you get large 50Hz noise check the electrode contacts, mention which maneuvers helped reduce the noise – whether it was sitting away from a table on which other electrical equipment are placed, or whether it was due to moving yourself away from mains cable or wires that are from the mains connected 'non-isolated' side, or whether it was due to sitting on a wooden chair instead of a chair with a metallic frame, etc.

4. Try to create an electrical circuit model of the situation that contributed to increasing or decreasing noise.

Under the third section you may include the following:

1. How much the average amplitude varied with variation in contraction.

2. How much it varied for different persons.

3. How much it varied for different muscles.

You may further discuss:
1. EMG gives an insight to our body electricity.

2. Amplitude of EMG increases with increased contraction of muscle.

3. Amplitude of EMG will depend on muscle mass, a muscle with more mass may give larger EMG signals. A person with more muscle mass may give a higher EMG as well.

4. EMG signals can further be analysed to show average, cumulative integration, frequency spectrum, etc. to compare healthy muscles from ones with disorders.

5. Surface EMG may also be used to assess the strength of contraction. This may be useful to assess the recovery of stroke patients.

6. Surface EMG may also be used to control artificial prosthesis or robotic arms.

7. Surface EMG from muscles on the face may be used to control equipment, wheel chair, etc. by paralysed patients.

Why not try the following?
You may try doing some of the analyses mentioned above by developing appropriate software. You may make a gripping force measuring device using a spring and a way to measure the deformation   of   the   spring

Experiment 7: Heart rate measurement using Light Plethysmograph

Introduction

The Greek word Plethysmo relates to increase in something. Plethysmography in medical science is commonly used to measure the increases in volumes of the whole body or a part of the body typically due to pumping of blood during systole (higher blood pressure regime within the heart cycle). Due to pulsating movement of blood the volume of limbs vary a little with heart cycle. Plethysmography is used to measure this change. Of course with light plethysmography the volume may not be measured accurately, but the pulsating behaviour can be readily indicated. Red light is scattered by blood therefore if we look at a finger with a bright light placed at the opposite side the whole finger appears red. However, due to the pulsating blood this scattering intensity varies a little, which cannot be appreciated by looking at it, but if we use a light detector, we can see the change. The present experiment will allow us to measure these changes on an oscilloscope screen (or on a computer monitor if a computerised data acquisition system is used).
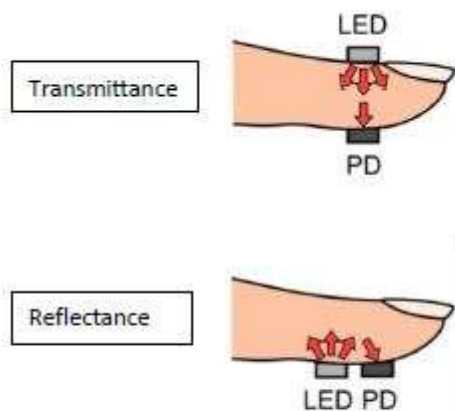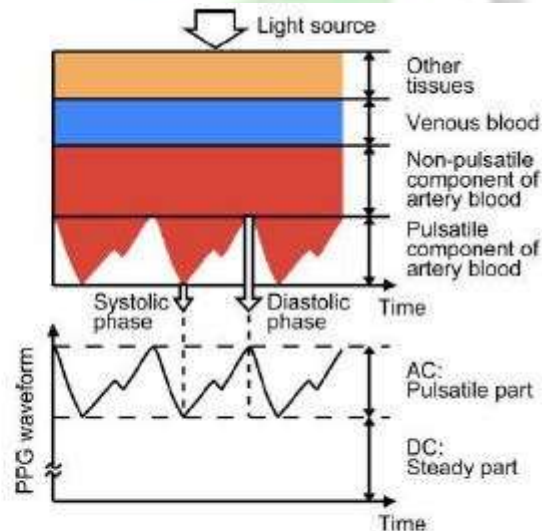

Fig.1: Light Plethysmography methods

Fig. 2: Light Plethysmograph waveform analysis

Methods and Measurements

This will require a light source and a detector with necessary electronics, and a storage oscilloscope to display the pulsating waveforms as shown in Fig.3. Alternatively, a PC based data acquisition system may be used to display the waveforms by writing the necessary software.
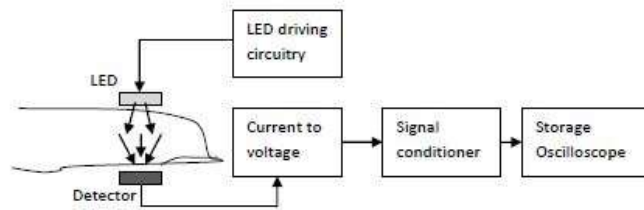
Fig.3 Block diagram of Light Plethysmograph

The LED needs a constant current supply in order to keep the light output constant. This can be done using a constant voltage power source driving the LED in series with a fixed resistor. However, with changes in temperature the forward voltage drop of the LED will change slightly resulting in a small change in the current as well. Therefore, a constant current source is preferred as shown in Fig.4. This uses a variable voltage regulator LM317 in a configuration that gives a constant current through the load (here the LED). The IC LM317 maintains a fixed voltage of about 1.2V between its output (OUT) and adjust (ADJ) terminals, which is almost independent of temperature. Therefore, the current I through R will be given by,

I = (VOUT – VADJ) / R

Taking 1.2V for the voltage difference and R = 50 Ω,

I = 24 mA

This current remains constant and flows through the LED. Current from the ADJ terminal is very small and negligible. Therefore, the total current through the LED is constant. The forward current through the LED should be kept at less than 30 mA to get a long working life, therefore this design of 24 mA is acceptable. However, 50 Ω is not a preferred value, the nearest is 47 Ω which would give about 6% higher current which is still acceptable.
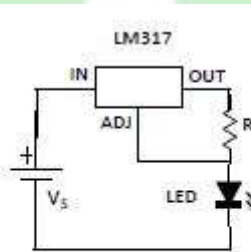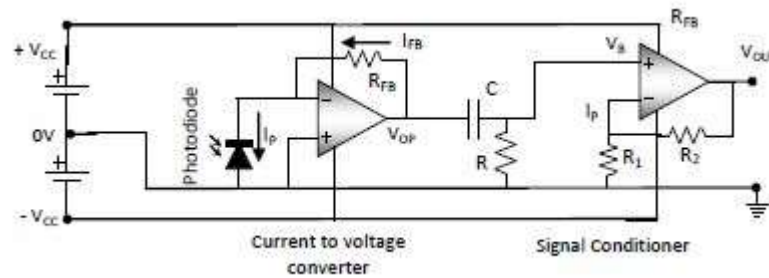

Fig.4: Constant current driving circuit for LED

The current to voltage converter and signal conditioner circuits are shown in Fig.5. A photodiode has been used as a detector here. With negative feedback the voltage at the inverting input equals

Current to voltage converter   Signal Conditioner

that at

the non-inverting input, which is 0V in this current to voltage converter circuit. Therefore, the photodiode has zero bias. The reverse photocurrent IP has to be supplied by the output terminal through feedback resistor RFB since no current can come from the input terminal of the op-amp. Therefore,

IP = IFB

And the output voltage VOP of the op-amp is given by,

VOP = IFB RFB = IP RFB

i.e., VOP ∞ IP

Depending on the magnitude of the photocurrent we have to choose RFB to get a reasonable output voltage. We may start with a value of about 1MΩ and then change it according to output obtained on actual measurement on a human finger.

The output is expected to have a large dc component and a small pulsating component. Since we are interested in this pulsating part of the signal we need to filter the dc out. This is the signal conditioner part of the circuit which is essentially a high pass filter. However, we need to take care that we do not cut out the frequency components that we are interested in. The pulsating blood has a period of about 1 sec, corresponding to a heart beat rate of 60 per minute. Therefore, the cut-off frequency of the high pass filter should be less than 1Hz, typically less than 0.5Hz. This is related to circuit parameters by,

0.5Hz = 1 / (2π RC)

or, RC = 0.32 sec, which is a time constant for the RC circuit. Here we have two unknowns but only one equation, which cannot be solved mathematically. From a design approach we use our experience to choose one of the unknowns and determine the other. If it turns out to be an impractical value then we adjust these to get an optimised combination.

For RC time constant, 0.32 sec is a large value. Therefore let us start with a large value of R, say 100kΩ. This gives C = 3.2 μF. This value of C is too large if we want to use non-polar capacitors. On the other hand if we use electrolytic capacitors, this is a practical value, but such capacitors have leakage problems, a large tolerance (±20%), and one needs to take care of the polarity in the circuit, which side will have a higher dc potential during operation. If we decide to go for non-polar capacitors then we may choose R = 2.2 MΩ and C = 0.15 μF (nearest preferred value). However, since the resistance has a high value, the output voltage will be reduced due to voltage divider action when we connect a load, which is going to be less than 2.2MΩ in most

probability. Therefore, we need an impedance buffer. This is usually achieved using an op-amp circuit with 100% negative feedback giving unity gain. However, an op-amp gives an opportunity to have some voltage gain with reduced feedback but still maintaining the necessary impedance buffer action. This we have done here through negative feedback using R1 and R2, the gain being (1 + R2/R1). The values of these resistors may be chosen during the experiment depending on the input requirements of the storage oscilloscope or of the PC based data acquisition system. We may choose 10 kΩ for both R1 and R2 giving a gain of 2, or choose R1 = 10 kΩ and R2 = 100kΩ for a gain of 11. If there is high frequency noise we may make a low pass filter by placing a suitable capacitor CP (not shown in figure) in parallel to R2. A cut-off frequency of between 10Hz and 20Hz should be good enough for this application (as it will also reduce 50Hz noise from the mains) and CP should be chosen accordingly.

To view the pulsating output, we have to set the horizontal time scale such that 3 or 4 heart cycles can be observed. This will need a total sweep of about 3 to 4 sec, i.e., 0.3sec to 0.4 sec per division (if there are 10 major divisions along the horizontal).

SEN-11574 Pulse Sensor Introduction (pulsesensor.com)

The SEN-11574 pulse sensor is mainly used for sensing heartbeat rate. Normally it is a very difficult task to measure the exact heartbeat rate, but this has become so much easy with the help of this pulse sensor amped. If we talk about heartbeat, then heart beat is a periodic signal that is produced by any software or hardware system for giving intimation to normal of working of any system. For measuring this periodic intimation signal, many sensors have been used currently in the market but here we shall only talk about SEN-11574 pulse sensor amped. This is basically plugged and play heartbeat sensor and can be used by athletes, game developers, and students in their hardware projects. It is easily available in the market or online shops.

A simple heartbeat pulse sensor is shown below:
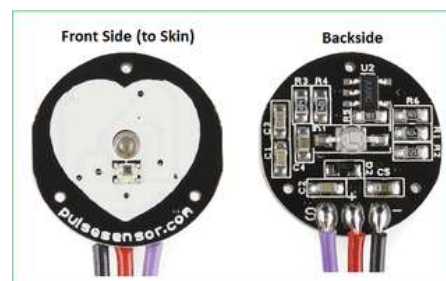
Pulse sensor front and back view

It has many applications in our daily life. Some common ones include using the sensor in exercise machines for heart beat measurement during workout and heart beat meter for local use at home.
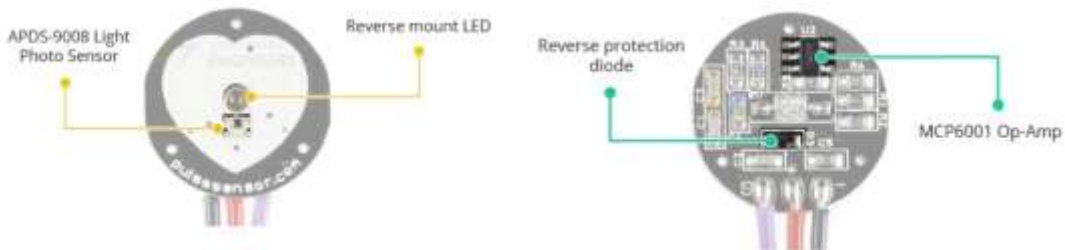
Pulse Sensor Circuit

The following figure shows the internal circuit diagram of a pulse sensor. It consists of optical heart beat sensor, an amplification circuit, and a noise cancellation circuit.

Internal block diagram of an SEN-11574 pulse sensor
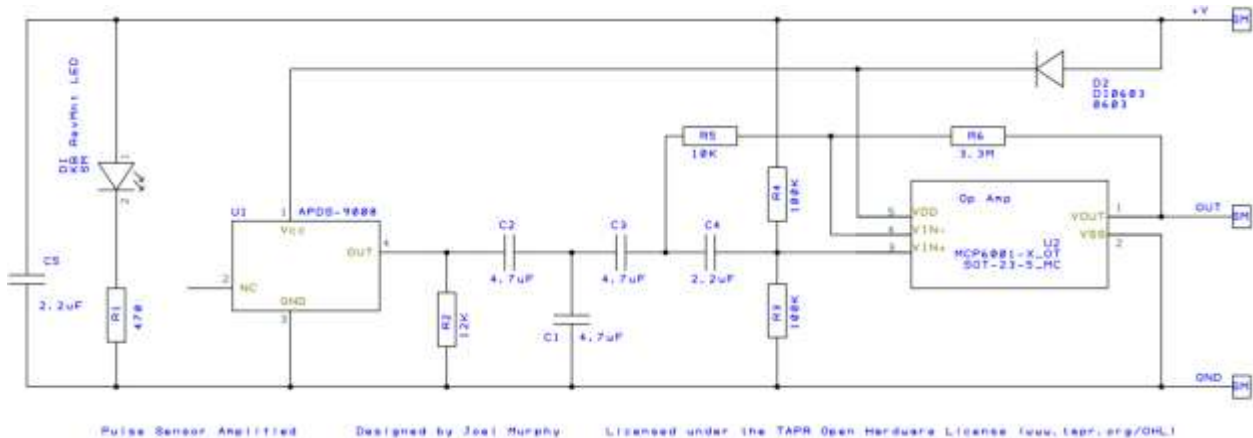
https://cdn.shopify.com/s/files/1/0100/6632/files/PulseSensorAmpd_-_Schematic.pdf?1862089645030619491

The front side of the pulse sensor, where the heart shape is drawn, consists of a central LED. This is a hole from where Kingbright's reverse installed LED glows and helps the sensor for detecting heartbeat rate. Just under the LED is a photodetector sensor. Besides this, there is another circuitry below the LED. This circuitry is called noise elimination circuitry. It is used for eliminating the noise to enhance the reading of the heart rate pulse sensor.
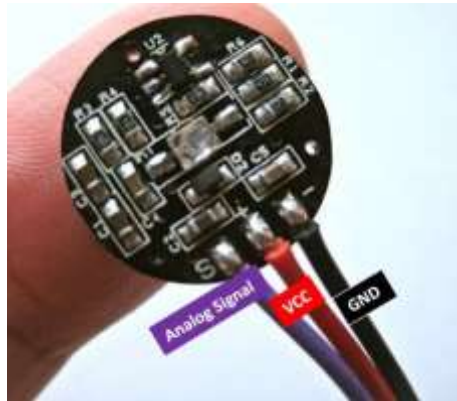
Pinout

This pulse sensor consists of three pins.

- First one is the **GND** pin which is used for supplying ground to this sensor and it is connected to source ground pin.
- The middle pin is the **VCC** pin which is used to supply power to the sensor. This sensor is powered on at almost 3.3V to 5V dc voltages.
- Similarly, the last one is the **A0** pin which is an analogue pin and it is used for receiving analogue signal. The sensor gives the output in the form of voltage. We need to measure this voltage waveform and extract heart beat rate out of this waveform.this sensor also consists of a central LED. This LED helps the sensor for detecting heartbeat rate. Beside this, there is another circuitry below LED and this circuitry is called noise elimination

circuitry. This circuitry is used for eliminating the noise which effects on the reading of the heart rate pulse sensor.

The pin configuration of this heart rate sensor is shown in the picture below:


SEN-11574 Pinout

Specifications

The table below shows some key specifications of the SEN-11574 sensor.

Operating Temperature	-40°C to +80°C
Input Voltage (VCC)	3V to 5.5V
Output Voltage	0.3V to VCC
Supply Current	3mA to 4mA
Dimensions (L x W)	15.8mm

How Pulse Sensor Works?

The working principle of this heartbeat rate sensor is very simple. If we talk about heartbeat rate, then heartbeat rate is the ratio of time between two consecutive heartbeats. Similarly, when the human blood is circulated in human body then this blood is squeezed in capillary tissues. As a result, the volume of capillary tissues is increased but this volume is decreased after each heartbeat. This change in volume of capillary tissues affects the LED light of heart rate pulse sensor, which transmits light after each heartbeat. This change in light is very small but this can be measured by connecting any controller with this pulse sensor. This means the LED light which has every pulse sensor helps for measuring pulse rate.
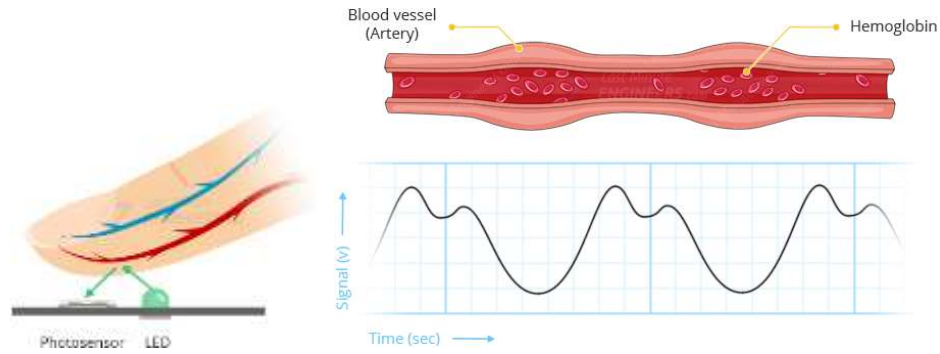
The working of this sensor could be checked by placing a human finger in front of this pulse sensor. When a finger is placed in front of this pulse sensor then the reflection of LED light is changed based on the volume of blood change inside capillary vessels. This means during the heartbeat the volume of blood in capillary vessels will be high and then will be low after each heartbeat. So, by changing this volume the LED light is changed. This change in of LED light measures the heartbeat rate of a finger. This phenomenon is known as "Photoplethysmogram."

Interfacing Pulse Sensor with Arduino
We will require the following components for this tutorial.

Required Components:
- Arduino board
- SEN-11574 pulse sensor
- Connecting Wires

As we know already that the pulse sensor has three pins that we have to connect with our Arduino UNO. These include the GND, VCC, and signal pin.

Follow the table and the schematic diagram below to successfully connect your circuit.

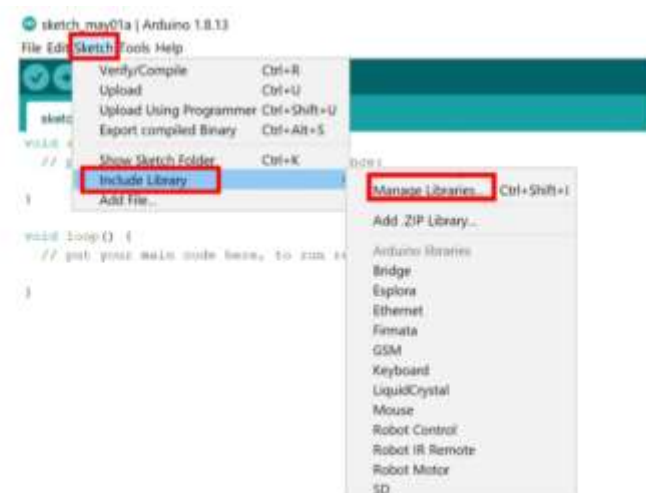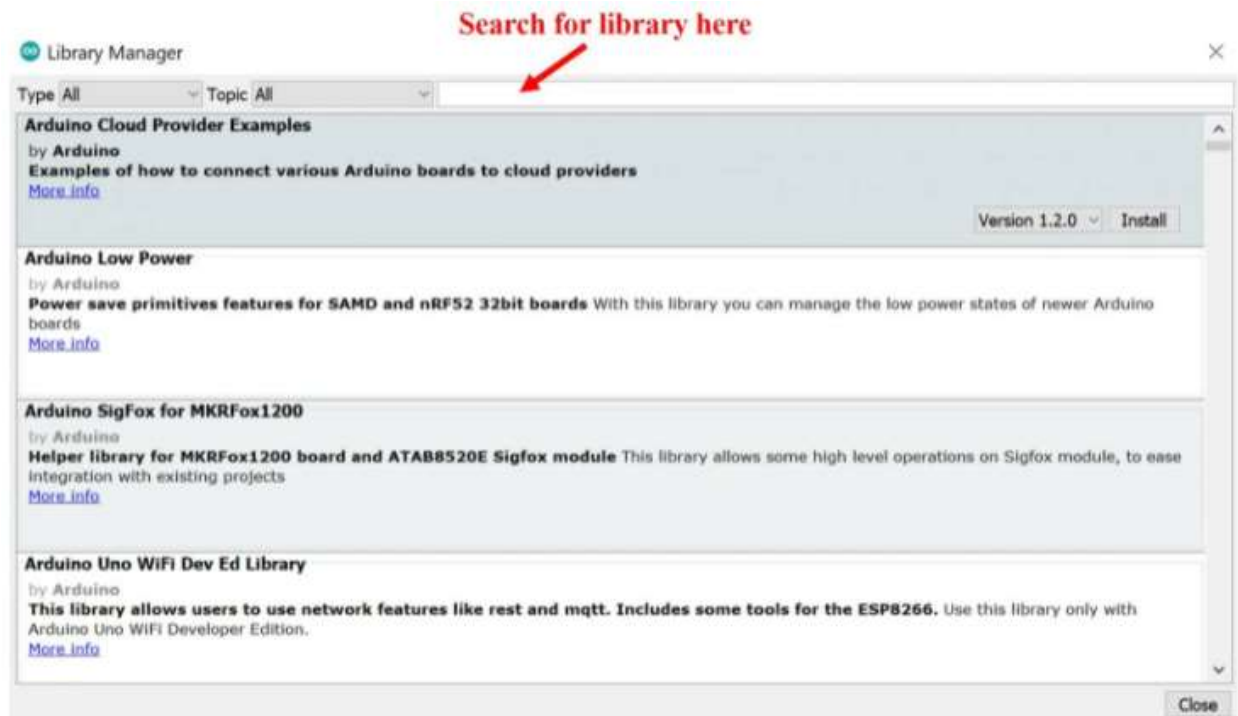| Arduino	Pulse Sensor
GND	GND
A0 (ANALOG IN)	A0
5V	VCC
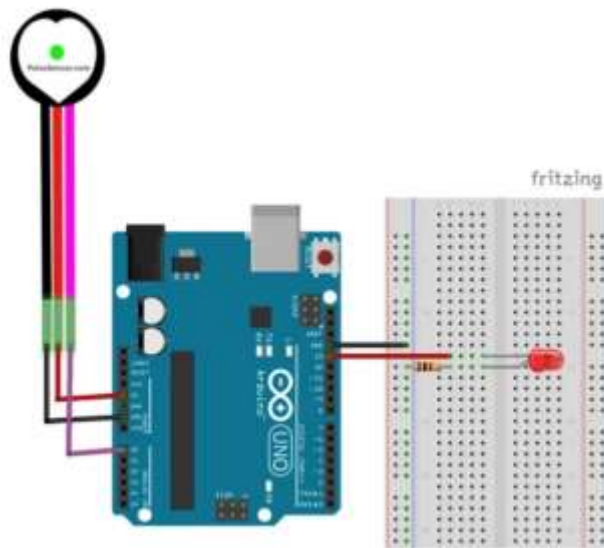
Arduino UNO with Pulse Sensor

As the pulse sensor requires an operating voltage in the range of 3.3-5V hence we will connect the VCC terminal of the sensor with 5V terminal of Arduino UNO. Both the grounds will be in common. Additionally, the analog signal pin of the sensor will be connected with A0 pin of the Arduino board.

Installing Pulse Sensor Arduino Library
We will use Arduino IDE to program our Arduino UNO. Thus, you should have the latest version of Arduino IDE. Open Arduino IDE and click on **Sketch > Library > Manage Libraries**

The following window will open up.

Type 'pulsesensor' in the search bar and press enter. Install the latest version of the library.

After installation of the library, restart your IDE.

Pulse Sensor Example Sketches for Arduino
In this section, we will show you how to use example sketches available in the Arduino IDE for PulseSensor Playground library that we just installed. We will look at different examples to effectively learn how to use the pulse sensor with our Arduino UNO.

LED blinking with hear beat using Pulse Sensor
Open your Arduino IDE and go to **File > Examples > PulseSensor Playground > GettingStartedProject**.
The following program code will open. This example sketch will blink the built-in LED of Arduino UNO in synchronization with your heart beat when you hold the sensor in between your fingers.

```
int const PULSE_SENSOR_PIN = 0;   // 'S' Signal pin connected to A0
```

```arduino
int Signal;            // Store incoming ADC data. Value can range from 0-1024
int Threshold = 550;   // Determine which Signal to "count as a beat" and which to ignore.

void setup() {
        pinMode(LED_BUILTIN,OUTPUT);  // Built-in LED will blink to your heartbeat
        Serial.begin(9600);         // Set comm speed for serial plotter window
}

void loop() {

        Signal = analogRead(PULSE_SENSOR_PIN); // Read the sensor value

        Serial.println(Signal);              // Send the signal value to serial plotter

        if(Signal > Threshold){             // If the signal is above threshold, turn on the LED
              digitalWrite(LED_BUILTIN,HIGH);
        } else {
              digitalWrite(LED_BUILTIN,LOW);    // Else turn off the LED

        }
        delay(10);

}
```

Code Explanation

The sketch is super simple. It begins with defining the pins used to connect the Pulse Sensor. Two variables are also defined; the Signal holds the incoming ADC data and the Threshold determines which signal to "count as a beat" and which to ignore.

```arduino
int const PULSE_SENSOR_PIN = 0;

int Signal;
int Threshold = 550;
```

In the setup, we define the Built-in LED pin (pin 13) as output and set up the serial monitor.

```arduino
void setup() {
        pinMode(LED_BUILTIN,OUTPUT);
        Serial.begin(9600);
}
```

In the loop, we read in the analog signal from the Pulse Sensor, and when the signal exceeds a threshold value, we turn the Built-in LED on.

```
void loop() {
        Signal = analogRead(PULSE_SENSOR_PIN); // Read the sensor value

        if(Signal > Threshold){             // If the signal is above threshold, turn on the LED
                digitalWrite(LED_BUILTIN,HIGH);
        } else {
                digitalWrite(LED_BUILTIN,LOW);     // Else turn off the LED
        }
        delay(10);
}
```

Demonstration

To see the demonstration of the above code, upload the code to Arduino. But, before uploading code, make sure to select the Arduino board from Tools > Board and also select the correct COM port to which the Arduino board is connected from Tools > Port.
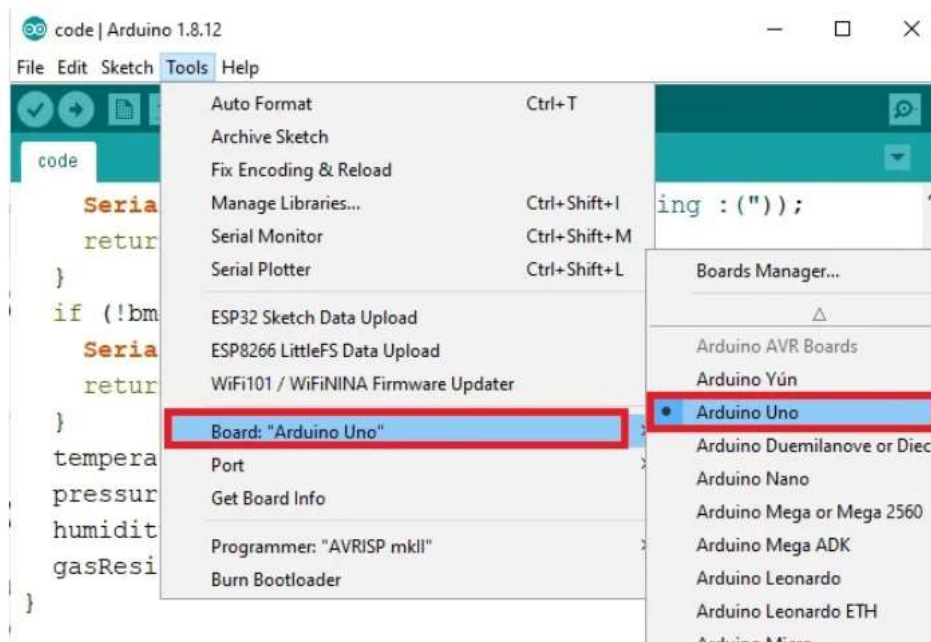
Once the code is uploaded to Arduino, hold the sensor in between your fingers. Do not hold it very tightly or very lightly. Make sure you apply normal pressure while holding the sensor. This will result in a better clean signal.

The built-in LED of Arduino UNO will start blinking with your heart beat. If for some reason the built-in LED is not blinking properly change the threshold value. The threshold value can take values from 0-1023.

Plotting Heart Beat Signal using Pulse Sensor

Now, we will plot the same signal retrieved from the pulse sensor through the Serial Plotter. We are using the same sketch as above. This time, however, we will show you a graph of the user's pulse.

Once the code is uploaded to Arduino, hold the sensor in between your fingers. Do not hold it very tightly or very lightly. Make sure you apply normal pressure while holding the sensor. This will result in a better clean signal.

Now go to **Tools > Serial Plotter** and set its baud rate to 9600. On the serial plotter you will be able to see the signal take various values within a certain range. After a while, the signal will stabilize and you will get a clearer heart beat signal. This is the signal that we obtained.
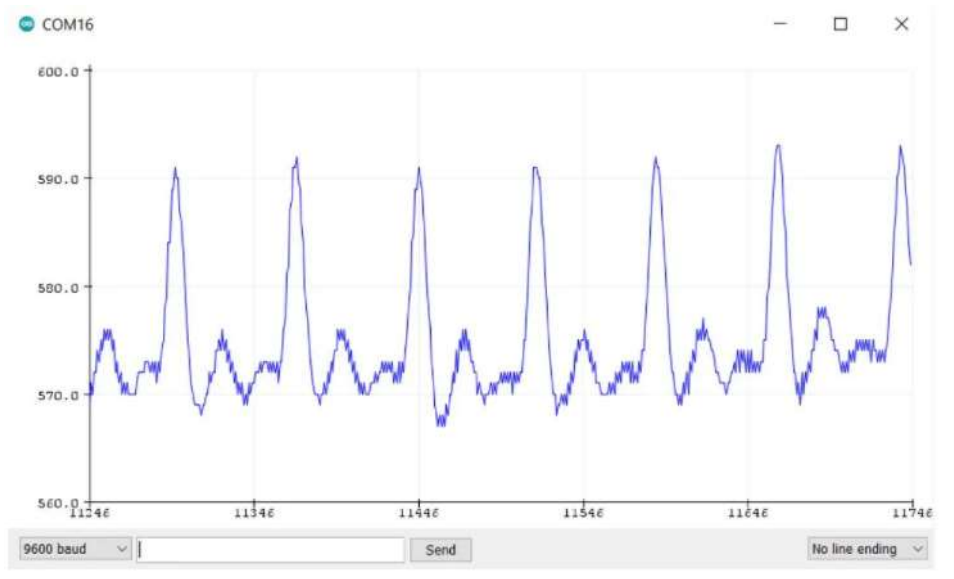
Serial Plotter

Monitoring BPM with Pulse Sensor and Arduino
Open your Arduino IDE and go to **File > Examples > PulseSensor Playground > Getting_BPM_to_Monitor.** The following program code will open. This example sketch will detect a heart beat and beats per minute (BPM) also known as heart rate. This will be displayed in the serial monitor.

Do not use this BPM monitor for medical purposes as it is not very accurate.

/* Getting_BPM_to_Monitor prints the BPM to the Serial Monitor, using the least lines of code and PulseSensor Library.
 * Tutorial Webpage: https://pulsesensor.com/pages/getting-advanced
 *
--------Use This Sketch To---
1) Displays user's live and changing BPM, Beats Per Minute, in Arduino's native Serial Monitor.
2) Print: "♥ A HeartBeat Happened !" when a beat is detected, live.
2) Learn about using a PulseSensor Library "Object".
4) Blinks LED on PIN 13 with user's Heartbeat.

```
-------------------------------------------------------------------*/

#define USE_ARDUINO_INTERRUPTS true    // Set-up low-level interrupts for most acurate
BPM math.
#include <PulseSensorPlayground.h>      // Includes the PulseSensorPlayground Library.

//  Variables
const int PulseWire = 0;        // PulseSensor PURPLE WIRE connected to ANALOG PIN 0
const int LED13 = 13;           // The on-board Arduino LED, close to PIN 13.
int Threshold = 550;            // Determine which Signal to "count as a beat" and which to ignore.
                        // Use the "Gettting Started Project" to fine-tune Threshold Value beyond
default setting.
                    // Otherwise leave the default "550" value.

PulseSensorPlayground pulseSensor;  // Creates an instance of the PulseSensorPlayground object
called "pulseSensor"


void setup() {

  Serial.begin(9600);          // For Serial Monitor

  // Configure the PulseSensor object, by assigning our variables to it.
  pulseSensor.analogInput(PulseWire);
  pulseSensor.blinkOnPulse(LED13);       //auto-magically blink Arduino's LED with heartbeat.
  pulseSensor.setThreshold(Threshold);

  // Double-check the "pulseSensor" object was created and "began" seeing a signal.
  if (pulseSensor.begin()) {
    Serial.println("We created a pulseSensor Object !");  //This prints one time at Arduino power-
up,  or on Arduino reset.
  }
}

void loop() {

 int myBPM = pulseSensor.getBeatsPerMinute();  // Calls function on our pulseSensor object
that returns BPM as an "int".
                        // "myBPM" hold this BPM value now.
```
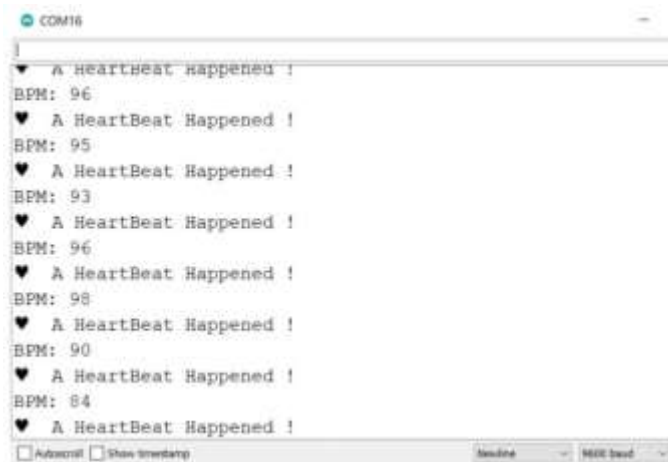
```
if (pulseSensor.sawStartOfBeat()) {          // Constantly test to see if "a beat happened".
 Serial.println("♥  A HeartBeat Happened ! "); // If test is "true", print a message "a heartbeat
happened".
 Serial.print("BPM: ");                   // Print phrase "BPM: "
 Serial.println(myBPM);                   // Print the value inside of myBPM.
}


 delay(20);              // considered best practice in a simple sketch.


}
```

To see the demonstration of the above code, upload the code to Arduino. But, before uploading code, make sure to select the Arduino board from Tools > Board and also select the correct COM port to which the Arduino board is connected from Tools > Port.

Once the code is uploaded to Arduino, hold the sensor in between your fingers. Do not hold it very tightly or very lightly. Make sure you apply normal pressure while holding the sensor. This will result in a better clean signal. Open your serial monitor and set the baud rate to 9600. After a few moments you will be able to see the heart rate.

Serial Monitor

Reference

1. https://pulsesensor.com

2. https://microcontrollerslab.com/pulse-sensor-arduino-tutorial

3. https://lastminuteengineers.com/pulse-sensor-arduino-tutorial

4. https://how2electronics.com/pulse-rate-bpm-monitor-arduino-pulse-sensor

**Experiment No. 08:   ECG Analysis with Matlab**

Loading and displaying an ECG from a file in your computer
The first step is to read the '.dat' file that contains the electrocardiogram, this can be done with the command 'load' like this:

ecg=load('E:\ECG Data\ecgpvc.dat');

This file contains 250 seconds of an ECG recording (it may vary for each file), since the sampling rate is 200 Hz for this dataset, so this ecgpvc.dat file contains 50,000 samples. We can now display the ecg in time with the command plot:

plot(ecg)
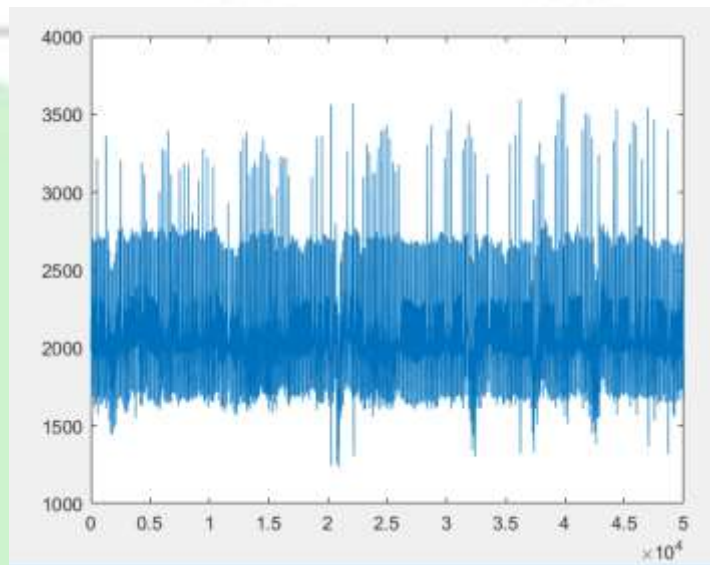
This data signal has been read from the website of Professor Rangaraj M. Rangayyan (http://people.ucalgary.ca/~ranga/).

Using 'urlread' we can access any data file from the internet that is available online

file = 'http://people.ucalgary.ca/~ranga/enel563/SIGNAL_DATA_FILES/ecgpvc.dat';
ecg_string = urlread(file);

Since the file was read from a website, it was assumed that it would be a string of characters, therefore we need to convert those characters to a number.

ecg2   = str2num(ecg_string);
plot(ecg2)

Zooming in
From the graph it is difficult to observe the detail of each heart cycle. We can zoom in manually with the magnifying glass of the figure, and also from the command line using the command axis,

axis ([3000 3800 1500 2800])
grid on
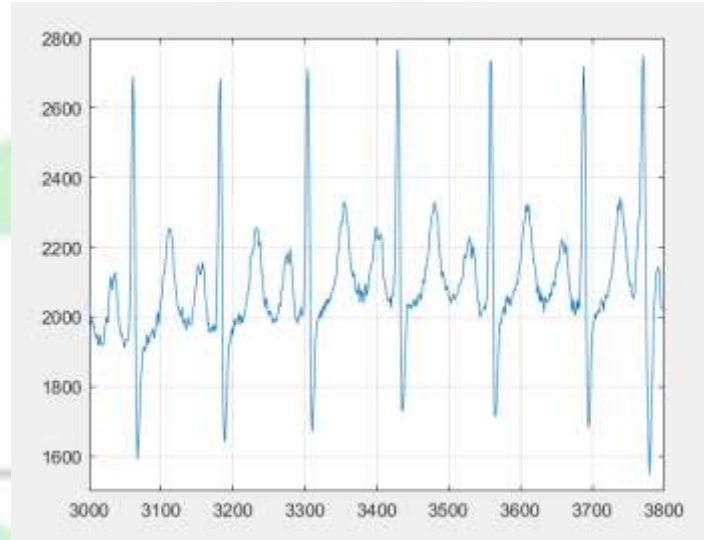
Detecting peaks of an ECG by thresholding
One interesting and simple exercise with this signal is to detect the peaks of each cycle. A simple attempt would be to detect regions of the signal that exceed a certain value. For instance, we can set a threshold (arbitrarily) at 2000, and generate a new signal from the comparison of the ecg against the threshold:

peaks_1 = ecg>2000;

At the same time, we can create a signal that corresponds to the time, this will be useful later on:

time = (1:numel(ecg))/200;

We can now compare the two signals, the original and the thresholded by plotting them together. Since the thresholded signal will have values between 0 and 1, we need to scale it in order to have the two signals in the same ranges. We can do that inside the plot command like this:
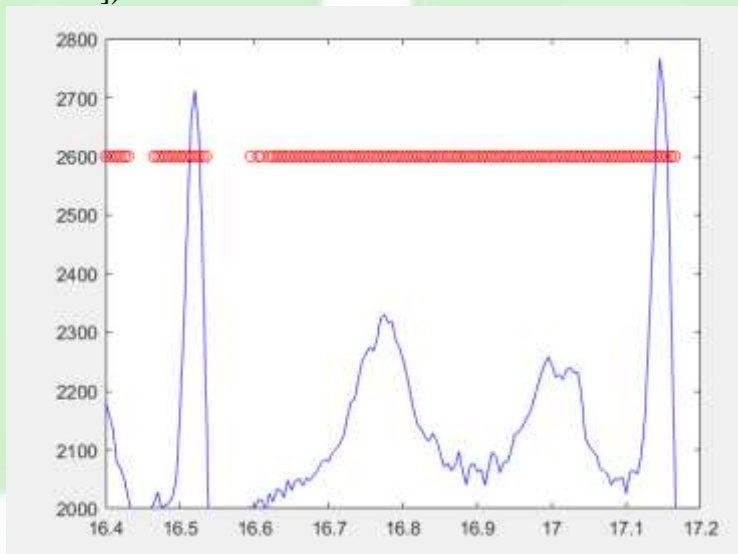
plot(time,ecg,'b',time,2600*(peaks_1),'ro')
axis ([15 20 1500 2800])
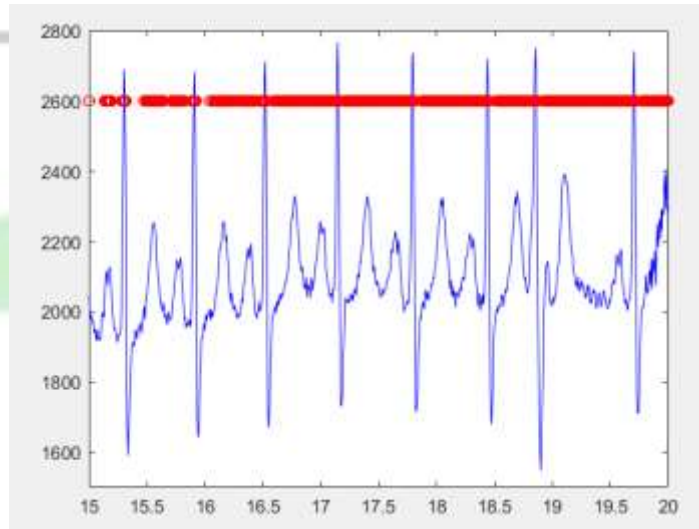
Size of the Peak

Notice that since we are now using time, the axis instruction has changed. If we look closely we can spot a problem:

axis ([16.4 17.2 2000 2800])

Detecting peaks with Matlab's 'findpeaks' function

The threshold is providing with a value for every single place where the signal exceed the given value, which is not exactly the same as detecting a peak. Fortunately Matlab has a function (always check to see if Matlab has a function for what you want, among thousands of pre-programmed functions, it is very likely that Matlab will simplify your work) to detect peaks called 'findpeaks', which is used like this:

[peaks_2,pos_peaks] = findpeaks(ecg);

We have now two vectors, one with the actual value of the peaks and a second with the location. Again, the best thing is to visualise the results:

plot(time,ecg,'b',pos_peaks/200,(peaks_2),'ro')
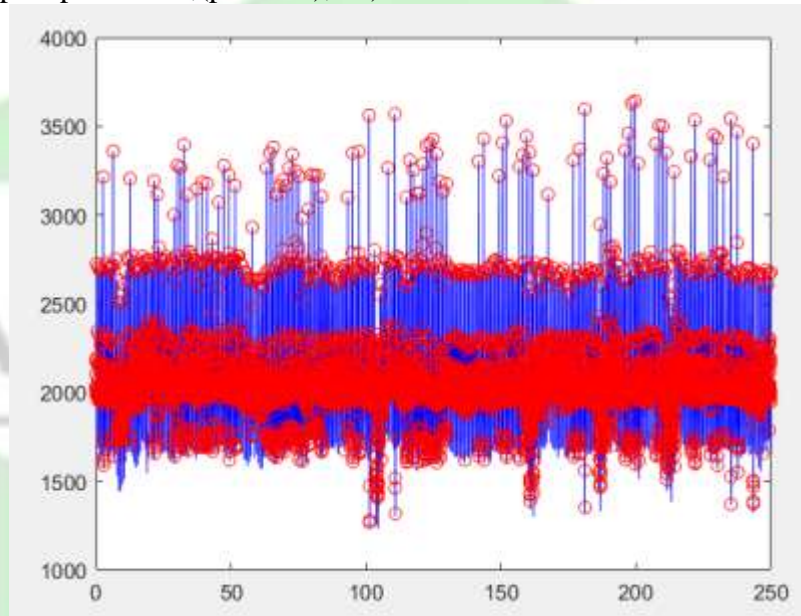
Notice two important things. First, since the positions are located from a vector, these have to be normalized with respect to time by dividing them by the sampling rate (200 Hz). Second, there seems to be an awful large number of "peaks" where there should not be any peaks, let's zoom in:

axis ([15 17 1500 2800])
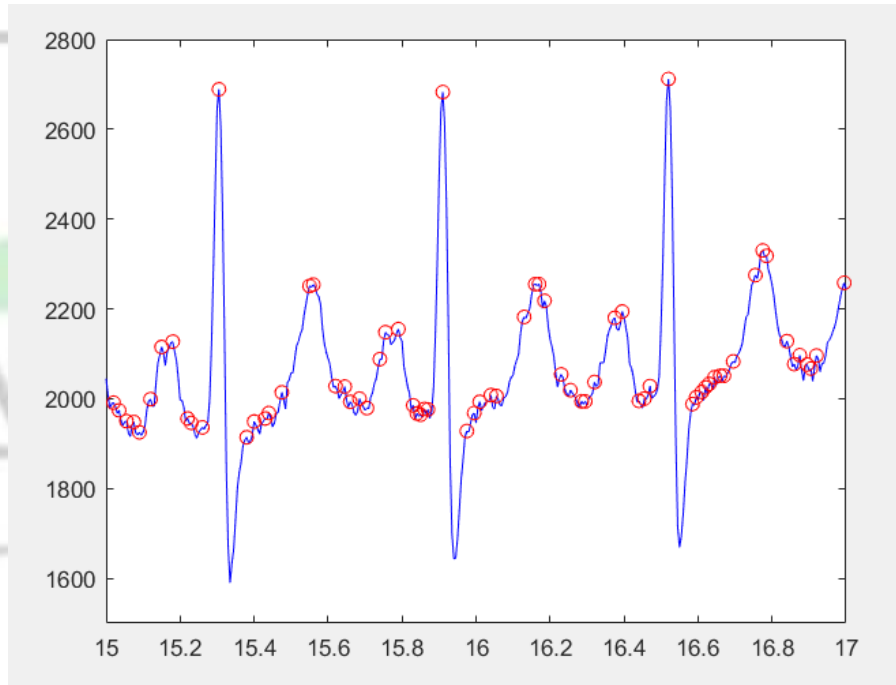
The finding of peaks is detecting ANY point which is higher than its neighbors, which is of course different of what we are trying to detect, which in turn leads to the question, which "peaks" are the ones we want?

Refining the peak detection

OK let's be specific, we want relatively big peaks (not those that are most probably due to noise) and the ones that are sharp and high, which should be closely related to the heart rate. That can be translated into rules like these: peaks that (a) are above a certain level (again 2400 can be a good threshold), (b) are not too close to each other. The second rule may not be immediately clear so let's take one at a time. First, let's discard all peaks that are below the threshold and display:

```
[peaks_2,pos_peaks] = findpeaks(ecg,'MINPEAKHEIGHT',2000);
plot(time,ecg,'b',pos_peaks/200,(peaks_2),'ro')
axis ([15 17 1500 3300])
```
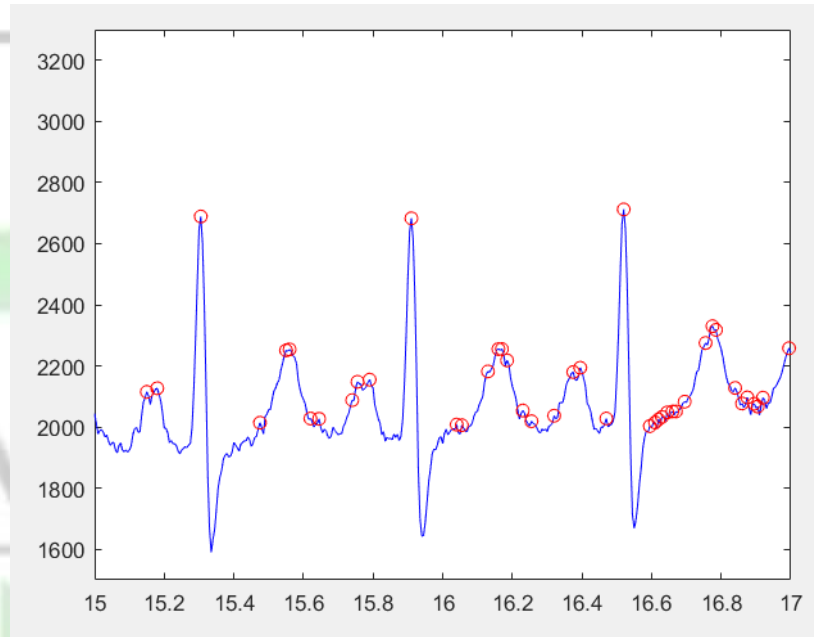
We can now see how the second rule will impact, as there are two positions detected within one single "peak". Let's repeat and find only those peaks that are at least 10 positions away from each other

```
[peaks_2,pos_peaks] = findpeaks(ecg,'MINPEAKDISTANCE',10,'MINPEAKHEIGHT',2000);
plot(time,ecg,'b',pos_peaks/200,(peaks_2),'ro')
axis ([15 17 1500 3300])
```
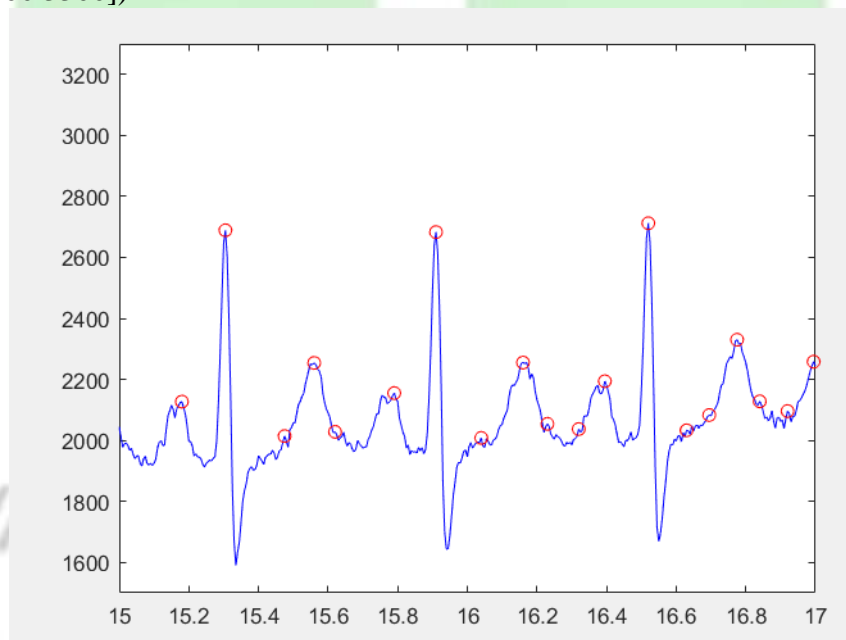

Inferring Heart Rate from the peaks
As a final step, we can display the ECG, and below it, the time differences between the peaks, that is, we can see the heart rate as inferred from these peaks. We can display two signals in a single figure with the command 'subplot' like this:

```
subplot(211)
plot(time,ecg,'b',pos_peaks/200,(peaks_2),'ro')
grid on
axis tight
ylabel ('ECG','fontsize',16)
subplot(212)
plot(time(pos_peaks(2:end)), 60*diff(pos_peaks/200));axis tight
grid on
axis tight
ylabel ('Heart Rate','fontsize',16)
xlabel ('Time (seconds)','fontsize',16)
x=60/diff(pos_peaks/200);
BPM=max(x)
```

Finally we have displayed the heart rate as extracted from the ECG.

Sample Code

You can create a matlab file for another dat file and run the code verify

```
ecg=load('H:\Sessional\EEE 4238 Biomedical\ECG Data\ecg3.dat');
figure (1)
plot(ecg)

figure(2)
plot(ecg)
axis([3000 3800 1500 3300])

peaks_1=ecg>2000;
figure(3)
plot(1:numel(ecg),ecg, 'b', 1:numel(ecg), 2000*(ecg>2000), 'ro');
axis([3000 3800 1500 3300])

time= (1:numel(ecg))/200;

figure (4)
[peaks_2, pos_peaks]=findpeaks(ecg);
```

```matlab
plot(time,ecg, 'b', pos_peaks/200, (peaks_2),'ro')
axis([15 19 1500 3300])


figure (5)
[peaks_2,pos_peaks] = findpeaks(ecg,'MINPEAKHEIGHT',2000);
plot(time,ecg, 'b', pos_peaks/200, (peaks_2),'ro')
axis([15 19 1500 3300])


figure (6)
[peaks_2,pos_peaks] = findpeaks(ecg,'MINPEAKDISTANCE',100,'MINPEAKHEIGHT',2000);
plot(time,ecg,'b',pos_peaks/200,(peaks_2),'ro')
axis ([15 19 1500 3300])


figure (7)
subplot(211)
plot(time,ecg,'b',pos_peaks/200,(peaks_2),'ro');
grid on
axis tight
ylabel('ECG', 'fontsize',16)


subplot (212)
plot(time(pos_peaks(2:end)), 60/(diff(pos_peaks/200)));
axis tight
x=60/diff(pos_peaks/200);
grid on
axis tight
ylabel ('Heart Rate','fontsize',16)
xlabel ('Time (seconds)','fontsize',16)
BPM=max(x)
```
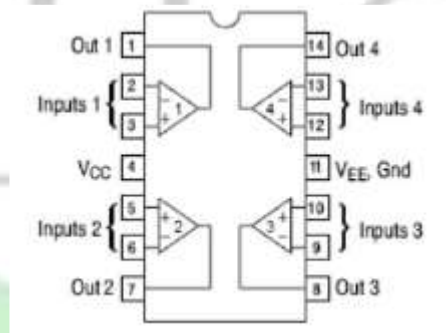
Appendix

Fig: Pin Diagram of LM324