Palestine Polytechnic University

Graduation Project

Design of Portable Three-Electrode Single-Channel ECG System

For Partially Fulfill the Requirements of Bachelor Degree at College of Applied Sciences

Work Team

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By the guidance of our supervisor, and by the acceptance of all members in the testing committee ,this project is delivered to Applied Electronic department of Applied Electronic And Applied Physics in the college of applied Sciences, to be as a partial fulfillments of the requirement of the department for the Bachelor degree in the Applied Electronic And Applied Physics .

Supervisor signature

Testing committee signature

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The head of department signature

Abstract

The topic of this thesis is to design a portable three-electrode signal channel electrocardiograph (ECG) recorder that has the capability to acquire the ECG waveform and display it on liquid crystal display (LCD) . The ECG system is supplied by two batteries of 9V each the system consists of AgCl electrodes, an instrumentation amplifier (INA) , and high and low pass filter to attenuate the noise . the filtered ECG signal is then amplified .

A microcontroller is used to sample the amplified signal and provide it to the LCD to monitor it .

This system can be carried by medics to monitor ECG signal for infected man in accident area, and can be used by doctors in areas where no electrical supply is found .

الملخص

قلب الإنسان هو اهم عضلة في جسم الإنسان , ومن اجل المحافظة عليها وابقاها في أفضل حالاتها نحن بحاجة لمراقبتها بشكل دوري ; إن أفضل طريقة لعمل ذلك هي بتصميم جهاز لمراقبة القلب بحيث يسمح لنا بنقله وتحريكه بحرية , ويكون سهل الحركة والاستعمال والقراءة , التصميم يجب أن يكون بسيط و خفيف : التصميم الذي قمنا بعمله قد أرضى كل الشروط السابقة بحيث قمنا بعمل جهاز لتخطيط القلب ويكون العرض على شاشة LCD بحيث يكون واضح وبوقت حقيقي.

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Chapter 1

Introduction

1.1 Introduction

The heart pumps blood to the lungs and to all the body's tissues by a sequence of highly organized contractions of its four chambers. For the heart to function properly, the four chambers must beat in an organized manner. This is governed by electrical impulse. A chamber of the heart contracts when an electrical impulse or signal moves across it, such a signal starts in a small bundle of highly specialized cells located in the right atrium — the sinoatrial node (SA node) . A discharge from this natural "pacemaker" causes the heart to beat.

The rhythmic behavior of the heart can be monitored and used as a diagnostic tool to detect heart abnormalities by acquiring the electrical activity on the body surface, across the heart, known as electrocardiograph (ECG). This electrical activity originates from the electrical activation of muscles in the heart, inherently indicating the approach of mechanical motion. The electric potentials generated by the heart appears throughout the body and can be measured across its surface.

The idea of the project design comes from that ECG can be used to assess if the patient has had a heart attack or evidence of previous heart attacks. An ECG can be used to monitor the effect of medicines used for coronary artery disease. These are the main issues that ECG helps doctors with heart attacks. By looking at the electrical patterns on an ECG, a doctor can determine the nature of an erratic heartbeat. There are many different types of abnormal heartbeats. Not all require treatment. In those that do, the ECG recording lets the doctor know which treatment is the best, for diagnosis of heart attack. It can also be determined whether the heart attack is old or recent. The possibility that there are narrowed arteries in the heart, which may lead to the heart attack in the future. Whether or not discomfort in the chest is being caused by the heart. Degeneration of the conduction system of the heart which can lead to dangerously slow heartbeats.

1.2 Project Goals

The idea of this project is to design a portable three channel ECG system that allow the patient to monitor his heart rhythm continually in any place at any time; which provide data to the doctor to use it in diagnosis in the future when needed.

1.3 Time schedule

The time plan views the stages of establishing the project with its components, divided into two semesters as shown in the following tables.

Table 1. 1: Time schedule of the first semester

Task\week	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15
Project Determination	▨	▨	▨												
Collect Information	▨	▨	▨	▨	▨	▨	▨	▨	▨	▨	▨	▨			
Basic Design								▨	▨	▨	▨	▨	▨		
Documentation									▨	▨	▨	▨	▨	▨	▨

Table 1. 2: Timing schedule of the second semester

Task\week	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15
Full Designing					▨	▨	▨	▨	▨						
Purchasing the Components							▨	▨	▨	▨	▨				
System Implementation										▨	▨	▨	▨		
ECG Signals Analysis													▨	▨	▨
Documentation						▨	▨	▨	▨	▨	▨	▨	▨	▨	▨

1.4 Total Cost

this section will be to explain the project cost as shown in table (1.3).

Table 1. 3: Total cost

Item	Cost [$]
Microcontroller	15
Resistors and Capacitors	10
Operation amplifiers	30
Instrumentation amplifier	35
LCD	110
Wilding board	10
Component Base	13
Electrode	12
Battery × 2 (9V)	10
Final Design	20
Total	265

Chapter 2

The Heart Anatomy, Physiology and ECG waveform

2.1 Introduction

"ECG" stands for electrocardiograph ; which is a measurement of the electrical signals that control the rhythm of the heartbeat. The signals that make the heart's muscle fibers contract come from the sinatorial node, which is the natural pacemaker of the heart. The ECG is a method of recording the electrical activity of the heart. Each heartbeat is caused by a section of the heart generating an electrical signal which then conducts through specialized pathways to all parts of the heart. These electrical signals also get transmitted through the chest to the skin where they can be recorded.

2.2 Anatomy and Physiology of the Heart

2.2.1 Heart Anatomy

The heart is a cardiac muscle that contract to pump the blood through all parts of the body, by the circulatory system. The heart is divided into four chambers shown in Figure(2.1).

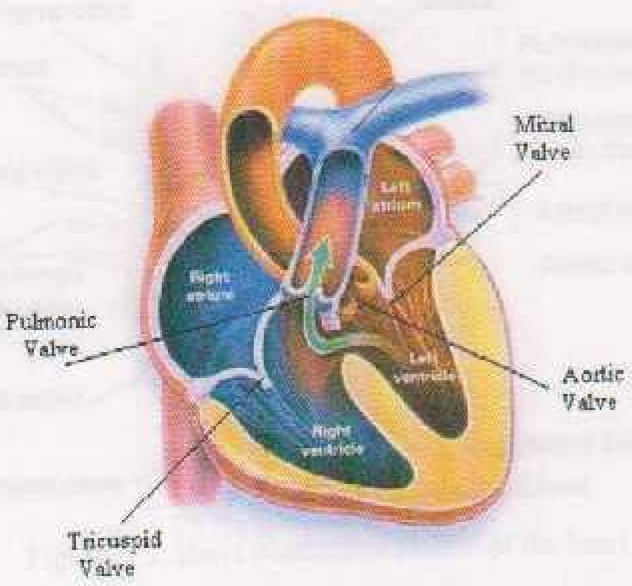


Figure 2. 1: The compartments of the heart

The upper two chambers, left and right atria are synchronized to act together. Similarly the two lower chambers, the ventricles, operate together. The right atrium receives the blood from the veins of the body and pumps it into the right ventricle. The right ventricle pumps the blood through the lungs, where it is oxygenated. The oxygen- enriched blood then enters the left atrium, from which is pumped into the left ventricle. The left ventricle pumps the blood into the arteries to circulate throughout the body.

2.2.2 The Heart Rate and Rhythm

The heart consist of several important parts that is shown in Figure (2.2) with the direction of blood flow, which has the blood come from different parts of the body into the lungs and back to the body as mentioned before [1] .

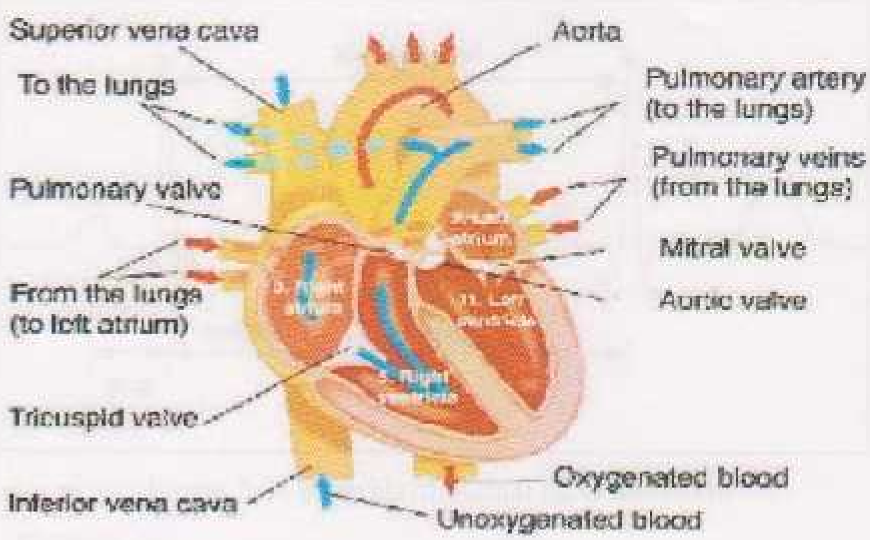


Figure 2. 2: Basic conduction system of the heart

Normally, the heart of an adult is depolarized 60 to 90 times per minute. A depolarization rate lower than this is called sinus bradycardia, while one that is higher is called sinus tachycardia. The heart rate of the normal newborn is much higher than that of an adult. The heart rate can be calculated by count the times of hart beat in one minute. This number is denoted BPM (Beats per minute).

2.3 The ECG-Waveform

By attaching electrodes to various parts of the body, a record of electrocardiogram wave can be obtained, an ECG signal can also be broken down into three main intervals: the P-R interval, the Q-T interval and the S-T interval, as shown in Figure (2.4).

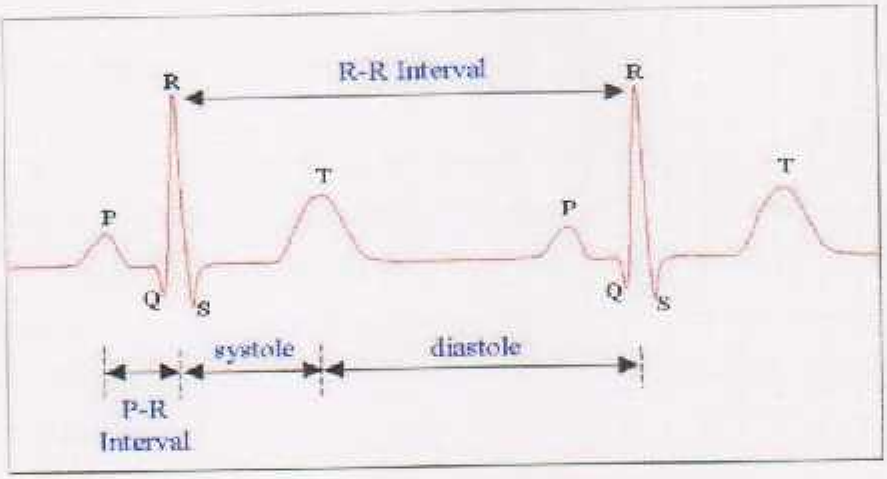


Figure 2. 3: Intervals of normal ECG waveform

Prominent parts of the ECG signal are the P wave, the QRS complex and the T wave, the P wave is caused by the spread of depolarization through the atria, and this is followed by atrial contraction, which causes a slight rise in the atrial pressure curve immediately after the P wave. Approximately after the onset of the P wave, the QRS waves appear as a result of depolarization of the ventricles, which initiates contraction of the ventricles and causes the ventricular pressure to begin rising, as also illustrated. Therefore, the QRS complex begins slightly before the onset of ventricular systole.

Finally, one observes the ventricular T wave in the electrocardiograph. This represents the state of depolarization of the ventricles at which time the ventricular fibers begin to relax. Therefore, the T wave occurs slightly prior to the end of ventricular contraction. Some ECG signal also contains small amplitude U wave following the T-wave, U waves are common in slow heart rates but a prominent U wave may reflect a heart abnormality [1].

Chapter 3
ECG Leads and Electrodes

3.1 Introduction

When we talk about ECG, we usually refer to 12-lead ECG. This ECG is most useful to doctors, but sometimes in emergencies or for monitoring, only 3-lead ECG is recorded. It basically depends what you want to know. If it is the only diagnosis of rhythm problems, then any lead will do. But when one wants to diagnose damage in the ventricular or atrial muscle or in the Purkinje conducting system, 12-lead ECG is the king.

An ECG is performed by placing 3 recording leads in certain specific locations on the body. They only record the heart's electrical activity. They do not produce any electricity on their own. The test does not really hurt and has no known side effects. It does not require any preparation (except for possibly shaving hair to get a better recording). The recording itself takes only a few seconds. Including the setup time and the time to disconnect the leads, the whole procedure takes about 5 minutes.

In an ECG procedure, the electrical impulses made while the heart is beating are recorded and usually printed out on a piece of paper. This is known as an electrocardiogram, and records any problems with the heart's rhythm, and the conduction of the electrical signal through the heart, which may be affected by underlying heart disease.

3.2 ECG Leads

The term "bipolar" means that the electrocardiogram is recorded from two electrodes located on different sides of the heart, in this case, on the limbs. Thus, a "lead" is not a single wire connecting from the body but a combination of two wires and their electrodes to make a complete circuit between the body and the ECG. As shown in Figure (3.2) .

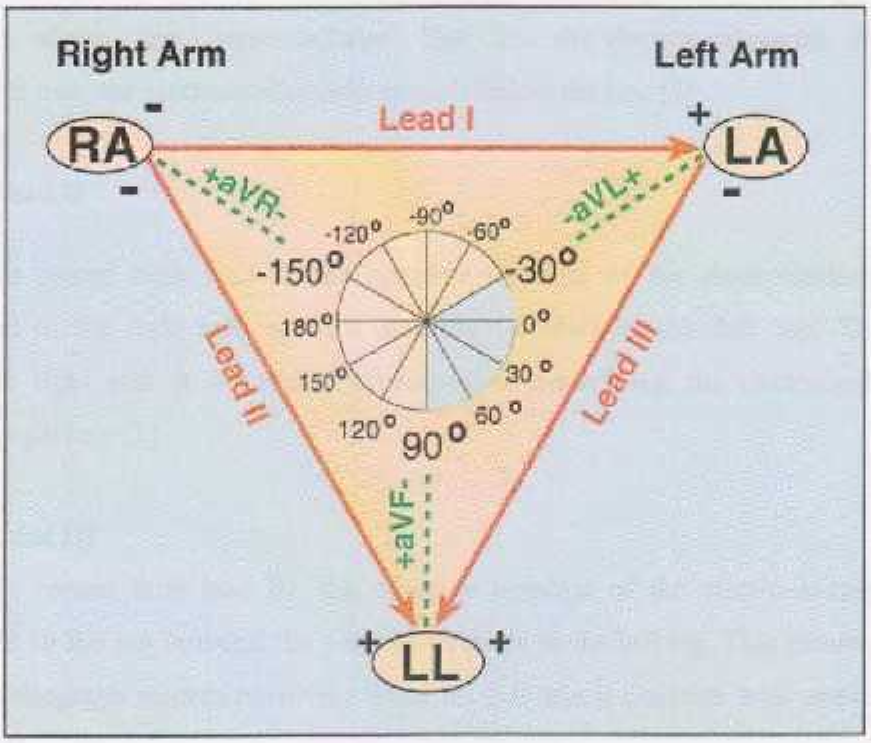


Figure 3. 1: Basic lead connection in ECG [3]

The triangle shown in the figure is called **Einthoven triangle** which is equilateral triangle used as a model of the standard limb leads used in electrocardiography.

The Einthoven's law says that the potential differences between the bipolar leads measured simultaneously will, at any given moment, have the value II=I−II [3] .

The three bipolar limb lead selections first introduced by Einthoven are as follows:

- **Lead I**

With recording limb lead I, the negative terminal of the electrocardiograph is connected to the right arm and the positive terminal to the left arm. Therefore, when the point where the right arm connects to the chest is electronegative with respect to the point where the left arm connects, the electrocardiograph records positively, that is, above the zero voltage line in the electrocardiogram. When the opposite is true, the electrocardiograph records below the line [3] .

- **Lead II**

To record limb lead II, the negative terminal of the electrocardiograph is connected to the right arm and the positive terminal to the left leg. Therefore, when the right arm is negative with respect to the left leg, the electrocardiograph records positively [3] .

- **Lead III**

To record limb lead III, the negative terminal of the electrocardiograph is connected to the left arm and the positive terminal to the left leg. This means that the electrocardiograph records positively when the left arm is negative with respect to the left leg [3] .

Einthoven also made the assumption that the heart is near to the center of the equilateral triangle, the apexes of which are the right and left shoulder and the crotch- The angle or region of the angle formed by the junction of two parts or members, such as two branches or legs. By assuming that the ECG potentials at the shoulders are essentially the same as the wrists and that the potentials at the crotch differ little from those at either ankle, he let the points of this triangle represent the electrode positions for the three limb leads.

The instantaneous voltage measured from any one of the three limb lead positions is approximately equal to the algebraic sum of the other two, or that the vector sum of the projections on all three lines is equal to zero [3] .

3.3 Electrodes

In order to measure and record potentials and , hence, currents in the body it is necessary to provide some interface between the body and the electronic measuring apparatus. Biopotential electrodes carry out this interface function of the electronic measuring circuit [2] .

3.3.1 The Electrode Interface

The passage of electric current from the body to an electrode can be understood by examining the electrolyte interface that schematically illustrated in Figure (3.2).

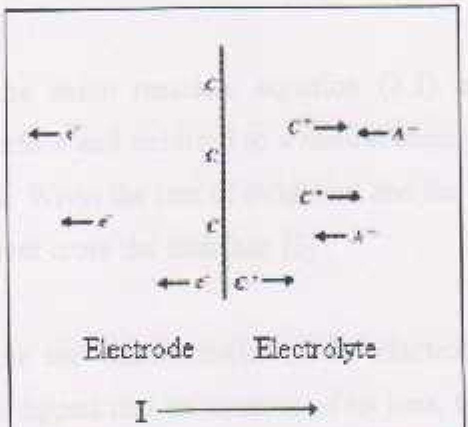


Figure 3. 2: Electrode-Electrolyte interface [2]

The electrolyte represents the body fluid containing ions , a net current that crosses the interface , passing from the electrode to the electrolyte consists of electrons moving in the direction opposite to that of the current in the electrode, cations (denoted by C+) moving in the same direction as the current, and anions (denoted by A-) moving in a direction opposite to that of the current in the electrolyte [2] .

At the interface between an electrode and an ionic solution, a redox (oxidation-reduction) reactions need to occur for a charge to be transferred between the electrode and the solution, these reactions can be represented in general by the following equations:

$$C \rightleftharpoons C^{n+} + ne^- \quad \textbf{(3.1)}$$

$$A^{m-} \rightleftharpoons A + me^- \quad \textbf{(3.2)}$$

Where n is the valence of cation material C, and m is the valence of anion material A, in the first equation (3.1) the electrode material C become oxidized to form a cation and one or more free electrons, the cation discharged into the electrolyte, the electrons remains as the charge carrier in the electrode, These ions are reduced when the process occurs in the reverse direction [2] .

In the case of the anion reaction, equation (3.2) an anion coming to the electrode-electrolyte interface and oxidized to a neutral atom, giving one or more free electrons to the electrode. When the rate of oxidation and the rate of reduction are not equal there are a net current cross the interface [2] .

To further explore the characteristics of the electrode-electrolyte interface ; when a piece of metal is dipped into an aqueous of its ions, these ions are cations and equal number of anions to maintain neutrality of charge, a local change in the concentration of the ions in solution near the metal surface is produced, This causes charge neutrality not to be maintained in this region, causing the electrolyte surrounding the metal to be at a different electrical potential from the rest of the solution, thus, a potential difference known as the half-cell potential is established between the metal and the bulk of the electrolyte. It is found that different characteristic potentials occur for different materials, these half-cell potentials can be important when using electrodes for low frequency or dc measurements [2] .

When two ionic solutions of different activity are separated by an ion-selective semi permeable membrane that allows one type of ion to pass freely through the membrane, It can be shown that an electric potential E will exist between the solutions on either side of the membrane, based upon the relative activity of the permeable ionsonn each of these solutions, this relationship is known as the Nernst equation (3.3) [2] .

$$E = -\frac{RT}{nF} \ln\left(\frac{a_1}{a_2}\right) \qquad \textbf{(3.3)}$$

Where a_1 and a_2 are the activities (for more information go to reference 2) of the ones on either side of the membrane, R is the universal gas constant, T is the absolute temperature, n is the valence of the ions, and F is the Faraday constant [2] .

3.3.2 Electrode Behavior and Circuit Models

The electric characteristics of bio-potential electrodes are generally nonlinear and a function of the current density at their surface, and in turn a nonlinear element is required for modeling electrode behavior, electrodes can be represented by an equivalent circuit of the form shown in Figure (3.3). In this circuit R_d account for the electrochemical processes taking place at the electrode-electrolyte interface and represent the leakage resistance across the double layer, C_d result from the distribution of ionic charge at the electrode-electrolyte interface that had been considered as a double layer of charge, R_s is the series resistance associated with interface effects and the resistance of the electrode materials themselves, the battery represents the half-cell potential described above [2] .

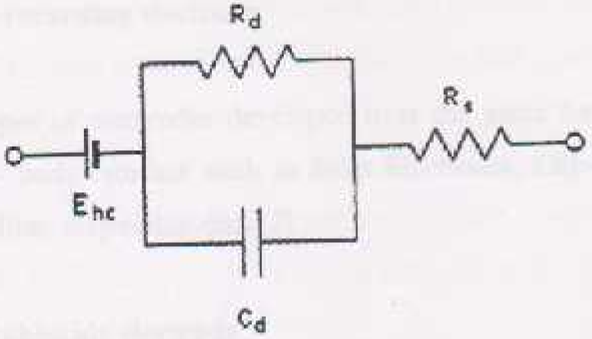


Figure 3. 3: The equivalent circuit for a bio-potential electrode [2]

It is seen that the impedance of this electrode will be frequency dependent, illustrated in Figure (3. 4). At low frequencies the impedance is dominated by the series combination of Rs and R_d, whereas at higher frequencies C_d bypasses the effect of R_d so that the impedance is now close to Rs, thus, by measuring the impedance of an electrode at high and low frequencies, it is possible to determine the component values for the equivalent circuit for that electrode [2].

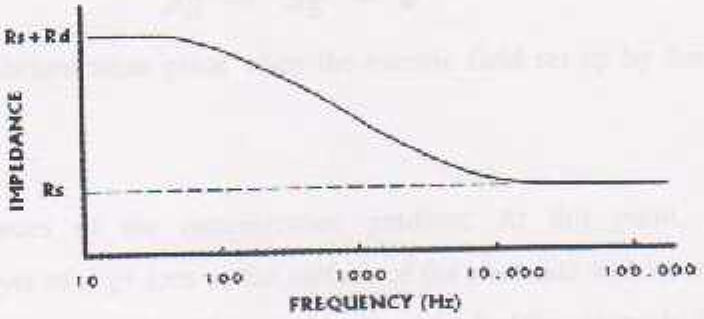


Figure 3. 4: Bio-potential electrode impedance as a function of frequency

3.3.3 Body surface recording electrode

Different types of electrodes developed over the years for recording various potentials on the body surface such as Snap Electrodes, Clip- Electrodes, Metal plate, Metal disk , film, disposable disk [2] .

3.3.4 Silver−silver chloride electrode

Electrodes for recording biopotentials are composed of a metal (usually silver for ECG Measurement), and a salt of the metal (usually silver chloride). In addition, some form of electrode paste or jelly is applied between the electrode (normally a flat silver disc) and the skin. The combination of the ionic electrode paste and the silver metal of the electrode forms a local solution of the metal in the past at the electrode−skin interface (also referred to as the electrode−tissue or electrode−electrolyte interface). Hence, some of the silver dissolves into solution producing Ag+ ions:

$$Ag \rightarrow Ag^{+} + e^{-}$$

Ionic equilibrium takes place when the electric field set up by the dissolving ions is balanced.

By the forces of the concentration gradient. At this point, there is a monomolecular layer of Ag+ ions at the surface of the electrode and a corresponding layer of Cl− ions adjacent to this. This combination is called the electrode double layer and there is a potential drop E across this layer, called the half−cell potential (0.8V in the case of the Ag−AgCl electrode). An ECG electrode is usually composed of a small metal plate surrounded by an adhesive pad, which is coated with conducting gel to help transmit the electrical signal [2] .

The wire that connects the ECG electrode to the ECG machine is clipped to the back of the electrode. Some electrodes are reusable, and other types are intended to be disposable after a single use.

Every ECG electrode placed on the body is attached by a wire to an ECG machine. The electricity that an electrode detects is transmitted via this wire to the machine, which translates the results into wavy lines that the machine then to the output stage [2] .

Chapter 4

System design

4.1 system

The ECG signal is very small signal (value) that need a lot of care in taking it , also it has a lot of noise signals; in this chapter we are going to describe the best way to eliminate the noise to have the desired signal, so there will be different stages to deal with these signals.

This chapter describes functionally and graphically, the system design and layout developed for this thesis. An overall block diagram of the system is shown in Figure (4.1).

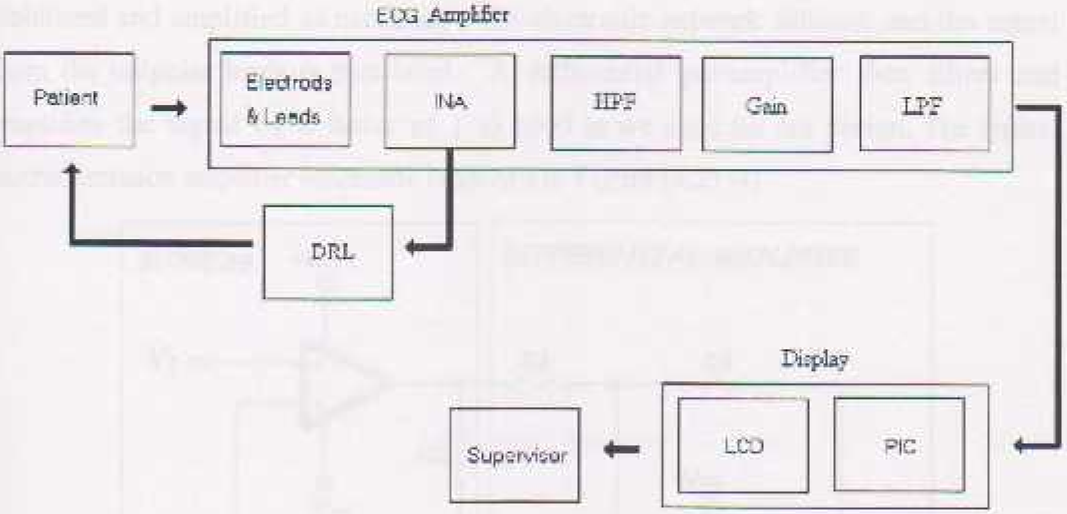


Figure 4. 1: Bock diagram of ECG System

The system is described in this chapter from electrode to liquid crystal display (LCD). Each major aspect of the design is then discussed in a separate . Those sections are ECG amplifier and the display . The ECG amplifier consists of instrumentation amplifier (INA) , a high pass filter (HPF) , low pass filter (LPF) , gain stage . The order of these stages is based on the consideration of amplify the ECG signal and attenuating the noise; the Display stage consist of Microcontroller and LCD.

4.2 ECG system component

4.2.1 ECG Amplifier

It consists of the following electronic components .

4.2.1.1 Instrumentation Amplifier (INA)

The instrumentation amplifier is useful when measuring relatively low level signals. During an ECG, the electrical signal from the body is transferred from the electrodes to the first section of the amplifier, the buffer amplifier. Here the signal is stabilized and amplified as necessary . An electronic network follows, and the signal from the unipolar leads is translated . A differential pre-amplifier then filters and amplifies the signal by a factor of 1 to 1000 as we need for our design. The typical instrumentation amplifier schematic is shown in Figure (4.2) [4] .

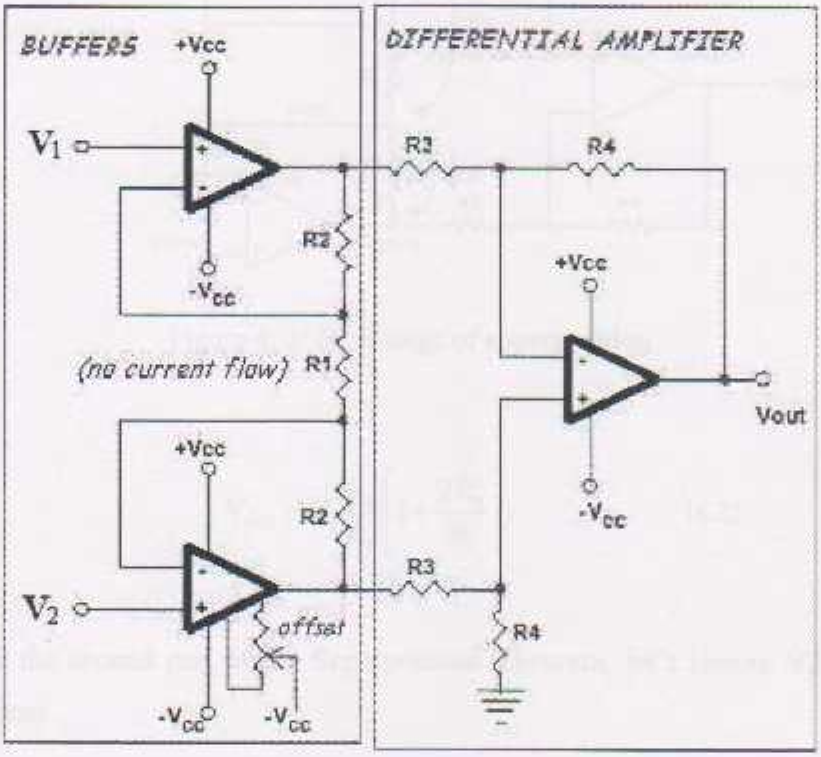


Figure 4. 2: Typical instrumentation amplifier schematic

It is well known that the instrumentation amplifier transfer function in Figure (4.2) is

$$V_{out} = (V_1 - V_2).\frac{R_4}{R_3}.\left(1 + \frac{2R_2}{R_1}\right), \qquad (4.1)$$

The proof of this transfer function starts with the Superposition Theorem.
Let's make V_2 zero by connecting U_2 input to ground, and calculate Vout1(equation. (4.2), see Figure (4.3).

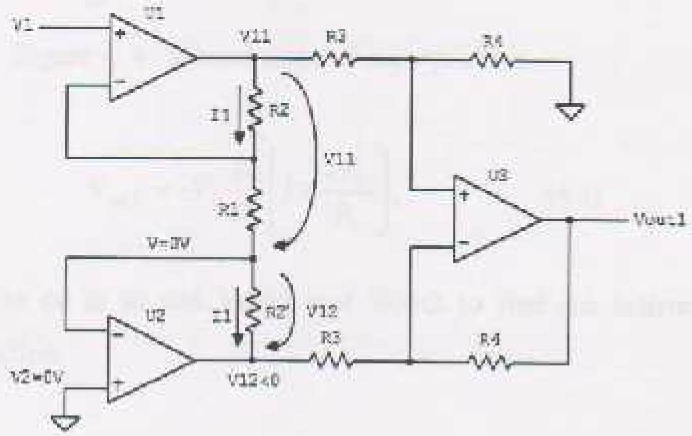


Figure 4. 3: First stage of superposition

$$V_{out1} = V_1\frac{R_4}{R_3}\left[1 + \frac{2R_2}{R_1}\right], \qquad (4.2)$$

For the second part of the Superposition Theorem, let's restore V2 and let's make V1 zero .

We well note the output voltage with Vout2(equation 4.3), see Figure (4.4).

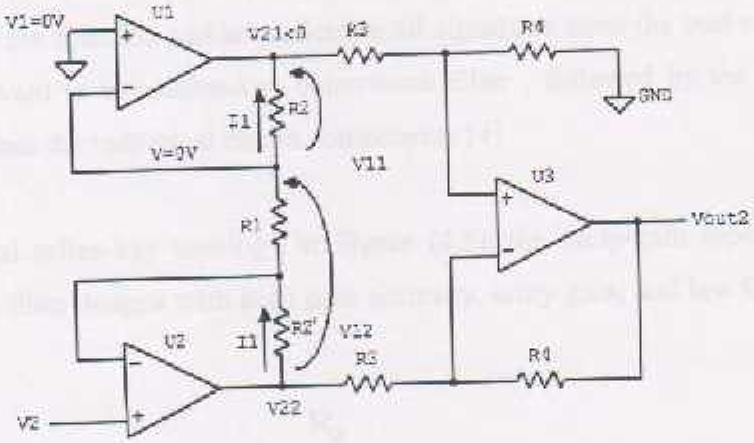


Figure 4. 4: Second stage of superposition

$$V_{out2} = -V_2 \frac{R_4}{R_3}\left[1+\frac{2R_2}{R_1}\right], \qquad (4.3)$$

All we need to do is to add Vout1 and Vout2 to find the instrumentation amplifier transfer function.

The result is given in equation (4.4).

$$V_{out} = V_{out1} + V_{out2} = (V_1 - V_2).\frac{R_4}{R_3}.\left[1+\frac{2R_2}{R_1}\right], \qquad (4.4)$$

Instead of using three op-amp. To build the INA , an AD620 is used in the project.

4.2.1.2 Sallen-Key High-Pass Filter (HPF)

A filter is a device that passes electric signals at certain frequencies or frequency ranges while preventing the passage of others.
In the design need for filtration and amplification all signals as same the best solution for this problem want to use sallen-key butterworth filter , followed by the design equations to calculate the individual circuit components [4] .

The general sallen-key topology in Figure (4.5); the unity-gain topology is usually applied in filter designs with high gain accuracy, unity gain, and low Qs (Q < 3) [4] .

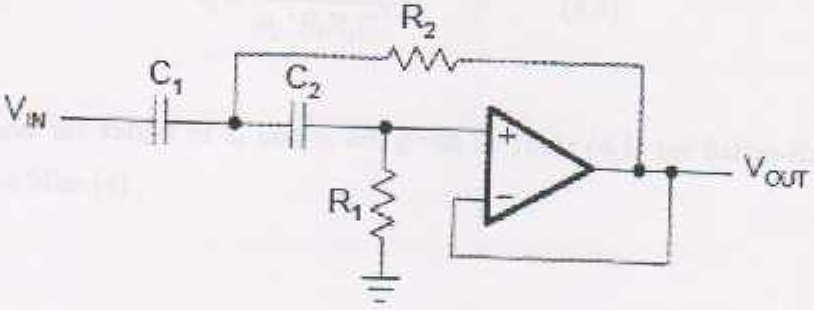


Figure 4. 5: Sallen-Key second-order high pass filter

The general transfer function of the circuit in Figure (4.5) is :

$$A_{(s)} = \frac{\alpha}{1+\frac{R_2(C_1+C_2)+R_1C_2(1-\alpha)}{\omega_C R_1R_2C_1C_2}\cdot\frac{1}{S}+\frac{1}{\omega_C{}^2R_1R_2C_1C_2}\cdot\frac{1}{S^2}},\ \alpha=1+\frac{R_4}{R_3},\ (4.5)$$

To simplify the circuit design, it is common to choose unity-gain (α = 1), and C1 = C2 = C.

The transfer function of the circuit in Figure (4.5) then simplifies to:

$$A_{(s)} = \frac{1}{1+\frac{2}{\omega_c R_1 C}\cdot\frac{1}{S}+\frac{1}{\omega_c^{\,2} R_1 R_2 C^2}\cdot\frac{1}{S^2}}, \qquad (4.6)$$

The coefficient comparison for this transfer function in equation (4.16) equal

$$a_1 = \frac{2}{\omega_c R_1 C}, \qquad (4.7)$$

$$b_1 = \frac{1}{\omega_c^{\,2} R_1 R_2 C^2}, \qquad (4.8)$$

Where the values of a_1 and b_1 are given in Table (4.1) for Sallen-Key Butterworth active filter [4] .

Table 4. 1 : Second-Order filter coefficients

Coefficient	Butterworth
a_1	1.4142
b_1	1

By let C , the resistor values for R1 and R2 are calculated through [4] .

$$R_1 = \frac{1}{\pi f_c C a_1}, \qquad (4.9)$$

$$R_2 = \frac{1}{4\pi f_c C b_1}, \qquad (4.10)$$

4.2.1.3 Sallen-Key Low-Pass Filter (LPF)

Low-pass filters use the same topology as the high-pass filters. The only difference is that the positions of the resistors and the capacitors have changed.

The general Sallen-Key topology in Figure (4.6); the unity-gain topology is usually applied in filter designs with high gain accuracy, unity gain, and low Qs(Q<3)[4].

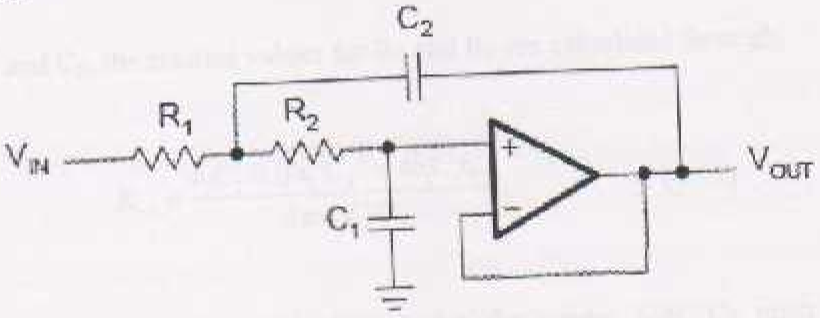


Figure 4. 6: Sallen-Key second-order low pass filter

The general transfer function of the circuit in Figure (4.6) is:

$$A_{(s)} = \frac{A_0}{1 + \omega_c\left[C_1(R_1 + R_2) + (1 - A_0)R_1C_2\right]s + \omega_c^2 R_1R_2C_1C_2s^2}, \qquad (4.11)$$

For the unity-gain circuit in Figure (4.6) (A0=1), the transfer function simplifies to:

$$A_{(s)} = \frac{1}{1 + \omega_c C_1\left[(R_1 + R_2)\right]s + \omega_c^2 R_1R_2C_1C_2s^2}, \qquad (4.12)$$

The coefficient comparison for this transfer function is:

$$a_1 = \omega_C C_1 \left(R_1 + R_2\right), \qquad (4.13)$$

$$b_1 = \omega_C^2 R_1 R_2 C_1 C_2, \qquad (4.14)$$

Where the values of a_1 and b_1 are given in Table (4.1) for Sallen-Key Butterworth active filter.

Given C_1 and C_2, the resistor values for R_1 and R_2 are calculated through:

$$R_{1,2} = \frac{a_1 C_2 \pm \sqrt{a_1^2 C_2^2 - 4 b_1 C_1 C_2}}{4\pi f_C C_1 C_2}, \qquad (4.15)$$

In order to obtain real values under the square root, C_2 must satisfy the following condition [4] .

$$C_2 \geq C_1 \frac{4b_1}{a_1^2}, \qquad (4.16)$$

4.2.1.4 Gain Stage

In this stage there will be an amplification of the signal to satisfy our requirements for the output device.

After attenuating the noise and DC voltage, an inverting amplifier shown in Figure (4.5) is chosen to amplify the pure ECG signal [4] .

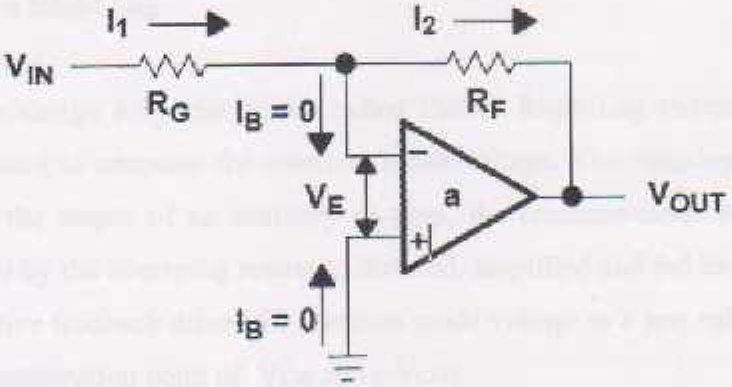


Figure 4. 7: Inverting amplifier

The non-inverting input of the inverting op-amp circuit is grounded. So the feedback keeps inverting the input of the op-amp at a virtual ground; The current flow in the input leads is assumed to be zero, hence the current flowing through RG equals the current flowing through RF. Using Kirchhoff's law, we write the equation (4.17); and the minus sign is inserted because this is the inverting input. Algebraic manipulation gives equation (4.18) [4] .

$$I_1 = \frac{V_{IN}}{R_G} = -I_2 = -\frac{V_{OUT}}{R_F}, \qquad (4.17)$$

$$\frac{V_{OUT}}{V_{IN}} = -\frac{R_F}{R_G}, \qquad (4.18)$$

The gain is only a function of the feedback and gain resistors, so the feedback has accomplished its function of making the gain independent of the op amp parameters [4] .

4.2.1.5 Driven Right Leg

In this design a special circuit called Driven Right-Leg circuit is shown in Figure (4.8) used to attenuate the common mode voltage. The right-leg electrodes is connected to the output of an auxiliary op-amp, the common-mode voltage on the body is sensed by the averaging resistors, inverted, amplified and fed back to the right leg, this negative feedback drives the common mode voltage to a low value, hence, the body act as a summation point of V_{CM} and ($-V_{CM}$).

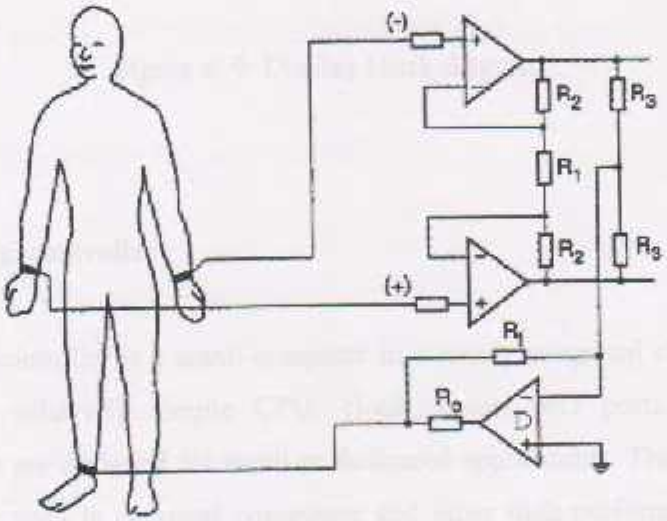


Figure 4. 8: Driven-right-leg circuit

4.2.2 Display

In order to get the graph of the ECG signal on the LCD, we need an interface to control and convert the ECG signal from its analog condition to a language that the LCD understand; hence we a suitable Microcontroller is required to control the flow of the data to the LCD from the ECG amplifier.The display unit is shown in Figure (4.9).

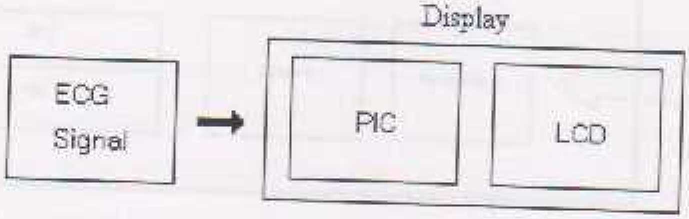


Figure 4. 9: Display block diagram

4.2.2.1 PIC Microcontroller

A microcontroller is a small computer in a single integrated circuit consisting internally of a relatively simple CPU, clock, timers, I/O ports, and memory. Microcontrollers are designed for small or dedicated applications. Thus, in contrast to microprocessors used in personal computers and other high-performance or general purpose applications [5] .

In this project the main function of the microcontroller is to convert from analog form to digital form and synchronize the operation of the display, this synchronization represented by providing the data to the LCD at the desired frequency; in our design we use PIC 16F877 [5] .

4.2.2.1.a Microcontroller Architecture Contents:

1. Microprocessor.
2. Memory.
3. I/O.

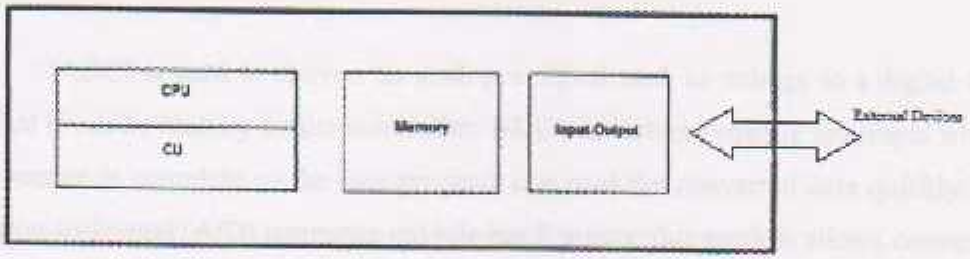


Figure 4. 10: The simplest microcontroller architecture

4.2.2.1.b Hardware feature of microcontroller:

1. Supply Voltage:

Microcontrollers 16F877 operate with the standard logic voltage of 5V that we get from the regulator which takes its power from the battery .

2. Clock:

The clock is provided by connecting the external crystal oscillator to the microcontroller.

3. Timers:

A timer is basically a counter which is driven either from an external clock pulse or from the internal oscillator of the micro controller. Most timers can be configured to generate an interrupt when they reach a certain count (usually when they overflow) [5] .

4. Analog-to-digital convertor:

(ADC) is used to convert an analogue signal such as voltage to a digital form so that it can be read by a microcontroller, ADC converters generate interrupts when a conversion is complete so the user program can read the converted data quickly. The Analog-to-Digital (A/D) converter module has 8 inputs; this module allows conversion of an analog input signal to a corresponding 10-bit digital number [5] .

4.2.2.2 LCD (Liquid Crystal Display)

LCD is a screen display technology developed in 1963 at the David Sarnoff Research Center in Princeton, NJ. LCDs are quite extraordinary.

The LCD is made of liquid crystals. The liquid crystals are the heart of the display and the display operation depends on the manipulation of the light mainly its polarization. The ambient light passes through the front polarizer, the front polarizer removes all of the light rays except the one polarized vertically. The polarizer works in the exact same way as in polarized sunglasses. Then the light goes through the liquid crystals and it gets 90° turn as shown in Figure (4.11). Then it goes through the rear polarizer gets reflected back and follows the same process and comes back in the front [6] .

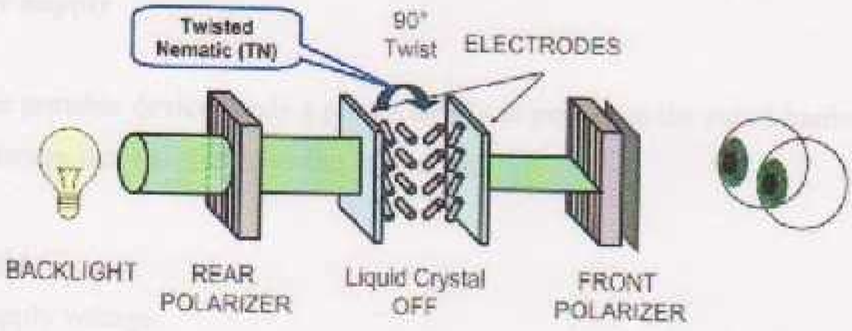


Figure 4. 11: The polarize of liqued crystal

The liquid crystal orientation can be changed by changing the charge applied to it. The charge is applied through some metal electrodes shaped like target image. For one pixel in graphics LCD it will be square. There is another kind of twist angle that called STN; in case of the STN (Super Twisted Nematic), liquid crystals give 270 degree twist to the light. This higher twist helps to achieve display operation with lower voltage. Which in effect provides faster switching speed.

For multiple pixels they are connected as a matrix and at each intersection of row and column introduced one pixel [6] .

4.2.2.2.a Processing in LCD interface

In order to show ECG signal on LCD screen there are several steps that must be traffic .

1. Analog to digital convert
2. Select the scale to view signal by use microcontroller .
3. Synchronous circuits by use crystal oscillator

4.3 Power Supply

The portable device needs a power supply to power up the entire hardware, so we need Battery that has the following properties:

1. Light weight .
2. Enough supply voltage .
3. Enough supply current .

From previous properties the 9V battery is enough to satisfy the hardware we use. And for various power supplies we use electronic regulators.

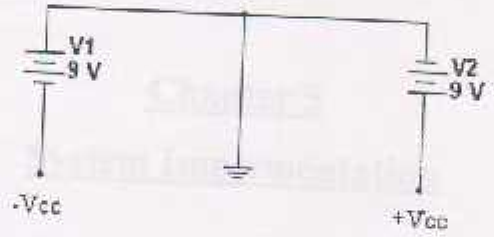


Figure 4. 12 : 9V power supply

Chapter 5

System Implementation

5.1 Introduction

In this chapter, the design process will be calculated and given in real values as discussed in Chapter 4.

5.2 Instrumentation Amplifier

This part is a first amplification stage of ECG signals . Amplification of INA limited by the input signal that contains common mode voltage V_{CM} up to 1.5V which comprises of 50Hz interference, and DC electrode offset potential of ± 500mV in addition to a small AC common mode voltage , V_{CM} ,has typically been a 0.05 to 1.5 mV; the signal represents the ECG signal. Figure (5.1) illustrates instrumentation amplifier input signal component.

Figure 5. 1: Instrumentation amplifier input signal component

Instrumentation amplifier supply voltage is ± 9V , to avoid saturating the output signal should not exceed ±7 V , common mode voltage in the worst case is a 500mV DC differential mode voltage and 1.5mg of AC common mode voltage .

$$V_{CM} = V_{CM\text{-}DC} + V_{CM\text{-}AC} \qquad (5.1)$$

$$V_{CM} = 500 \text{ mV} + 1.5\text{mV}_{(P\text{-}P)} \qquad (5.2)$$

Thus the DC differential mode voltage is a 300 times larger than the AC signal of interest,

At the same time, to convert the 1.5 mV AC into a representative signal that is of use, a total gain of 1000 or more is required, because the ECG signal is about 1mV to view it on the LCD it required a high gain amplifier . The solution for the above problem is performed in four steps :

1. Limit the gain of the instrumentation amplifier to avoid saturation.
2. Use the DRL circuit to attenuate the common mode voltage.
3. Implement a high pass filter in the next stage to remove the DC offset.
4. Apply a high gain.

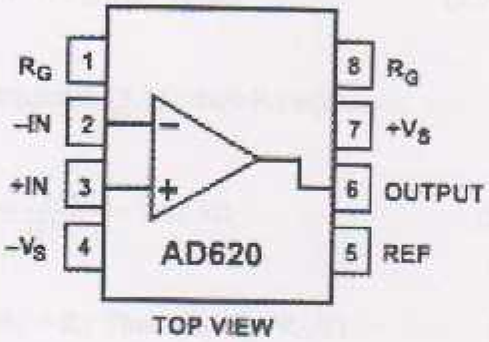


Figure 5. 2: Instrumentation amplifier pins layout

In the first step the gain of the instrumentation amplifier is limited to 13.35 (show later how this value calculates), thus, the maximum output voltage is 7 V, for instrumentation amplifier ADC620 the gain is determined by using equation (5.3) .

$$G = 1 + \frac{49.4K\Omega}{R_G} \qquad (5.3)$$

And the gain equal :

$$G = V_{outmax} / V_{in} \qquad (5.4)$$

The input voltage of INA is 1mV (ECG signal) and 1.5mV (AC common mode voltage) and 500mV (DC differential mode voltage).

$$V_{IN\ TOT} = 500mV + 1.5mV + 1mV \qquad (5.5)$$

Then to avoid saturation, the maximum output of AD620 is equal to

$$V_{out} = 9 - 2 = 7V \qquad (5.6)$$

$$G = 7/V_{IN\ TOT} = 13.93 \qquad (5.7)$$

To determine R_G from equation (5.1) then R_G equal

$$R_G = \frac{49.4K\Omega}{13.93-1} = 3.82\ k\Omega \qquad (5.8)$$

$$R_G = R_1 + R_2.\ \text{Thus}\ (R_1 = R_2 = R_G/2) \qquad (5.9)$$

But in this project used 4KΩ In order to avoid the use of a variable resistance . Then the gain of the INA will become A=13.35.

5.3 Driven Right-Leg Circuit:

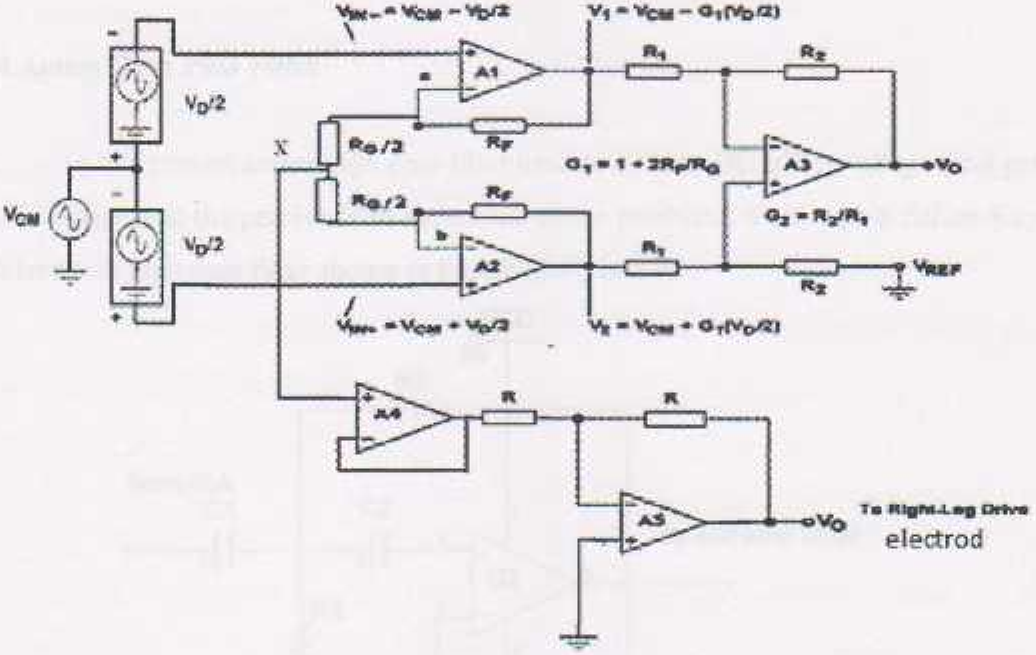


Figure 5. 3: Driven right-leg circuit for attenuating common-mode voltages

The input impedance of the inverting amplifier which is represented by the resistor R as shown in Figure (5.3), cause a loading effect on R_G, which draw current from node X, these resistors will alter and change the total R_G. As no current should flow from node X through the resistor R, a buffer (voltage follower) is used to block the current.

$$V_{IN(Buffer)} = \frac{V_{IN+}+V_{IN-}}{2} = V_{CM} \qquad (5.10)$$

Driven Right-Leg Circuit O/P

$$V_0 = -\frac{R}{R}V_{CM} = -V_{CM} \qquad (5.11)$$

5.4 Active High Pass Filter

In this project active high pass filter used to remove DC offset voltage and get an ECG signal at the precise level across the entire passband, a unity gain Sallen-Key Butterworth high pass filter shown in Figure (5.4) is used .

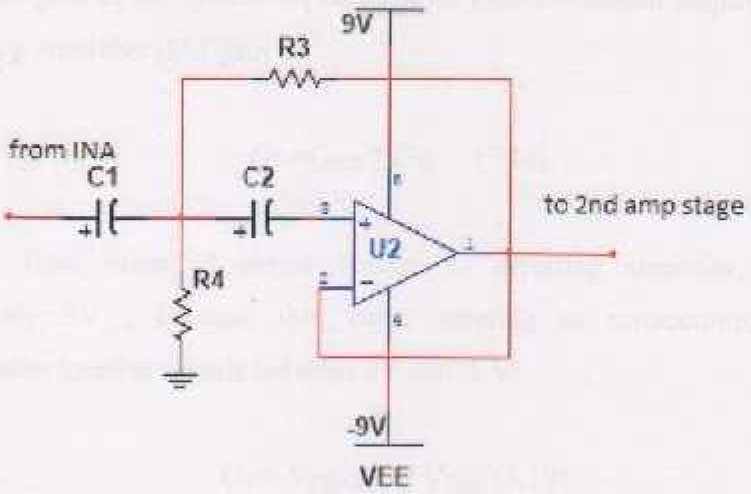


Figure 5. 4: Unity-Gain sallen-key high pass filter

For Butterworth high pass filter, a = 1.4142, b = 1 [2] .
For Butterworth high pass filter f_c =0.05Hz [1] .
Let C = 4.7μF the resistor values for R_1 and R_2 .

$$R_3 = \frac{1}{\pi * F_c * C * a} = 958.27\text{K}\Omega \qquad (5.16)$$

$$R_4 = \frac{a}{4 * \pi * F_c * C * b} = 497.12\text{K}\Omega \qquad (5.17)$$

5.5 Inverting Amplifier

After canceling DC offset voltage through a high pass filter, the ECG signal must be amplified by inverting amplifier shown in Figure (5.5) to be suitable for the next stages before sampling .

Total gain of the system G_T result from instrumentation amplifier (INA) gain and inverting amplifier (IA) gain :

$$G_T = G_{INS} * G_{IA} \quad (5.18)$$

The final value of output voltage to inverting amplifier, needs to be approximately 5V , because this value entering to microcontroller and the microcontroller handles signals between 0V and 5 V.

$$G_T = V_{OUTMAX} / V_{ECG} \quad (5.19)$$

$$G_T = 4V/1mV = 4000 \quad (5.20)$$

An inverting amplifier is used to provide this gain which is determined by Equation (5.19).

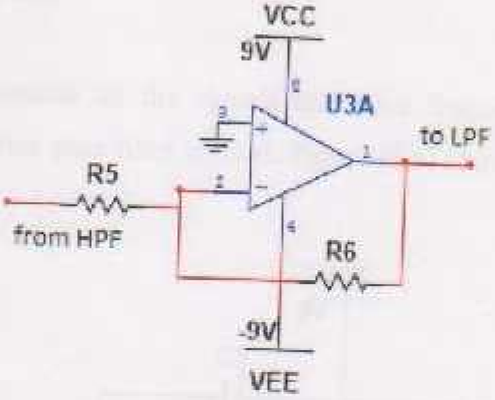


Figure 5. 5: Inverting amplifier circuit

$$G_{IA} = -R_6 / R_5 \qquad (5.21)$$

From Equation (5.19)

$$G_{IA} = G_T / G_{INA} = 4000/13.35 = 299.6$$

Let $R_5 = 1K\Omega$

From Equation (5.2)

$$R_6 = 299.6 * 1K\Omega = 299.6K\Omega$$

5.6 Active Low Pass Filter

In order to attenuate all the signals that have frequencies above 100Hz, a Sallen-Sallen-key butterwort low pass filter is used. Figure (5.6) shows a unity Gain Sallen-Key low pass filter.

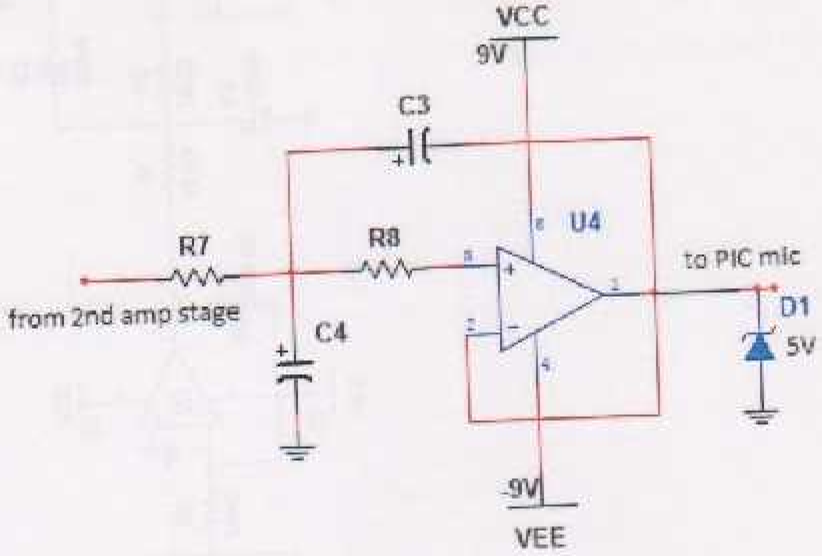


Figure 5. 6: Unity-Gain sallen-key low pass filter

Calculation:

Let R = 10kΩ then C_4 and C_3 equal

$$C_3 = \frac{\sqrt{2}}{2 * \pi * 10.22 * 10^3 * 100} = 220nf$$

C_4 =110nf

Use Zener diode to protect microcontroller from any error in gain (amplification of ECG signal).

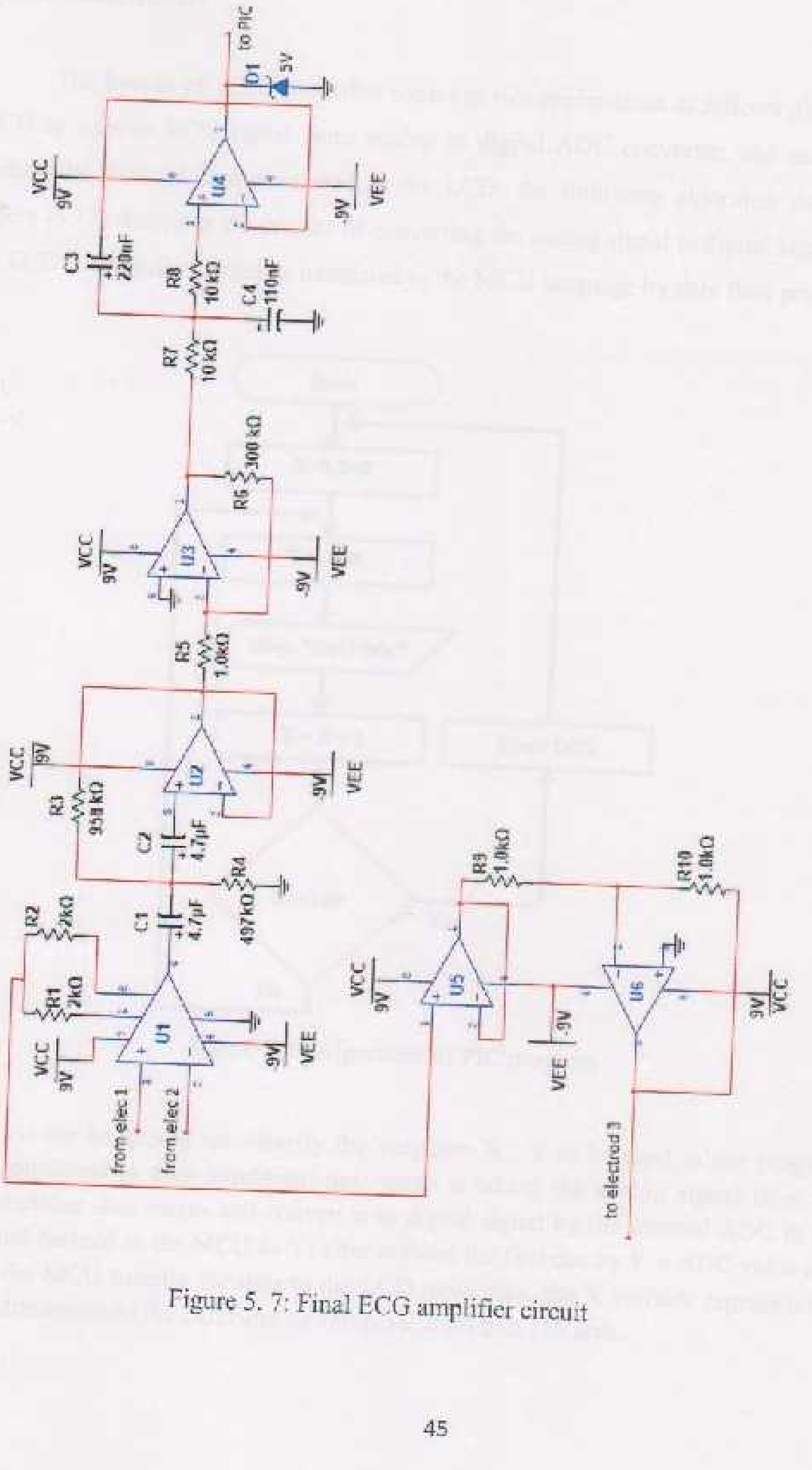


Figure 5. 7: Final ECG amplifier circuit

5.7 Microcontroller:

The benefit of microcontroller comes in two applications as follows ,first need MCU to convert ECG signal from analog to digital ADC converter, and second to control the flow of data processed to the LCD; the following algorithm shown in Figure (5.12) describes the presses of converting the analog signal to digital signal into the LCD continually , which is translated to the MCU language by data flow program.

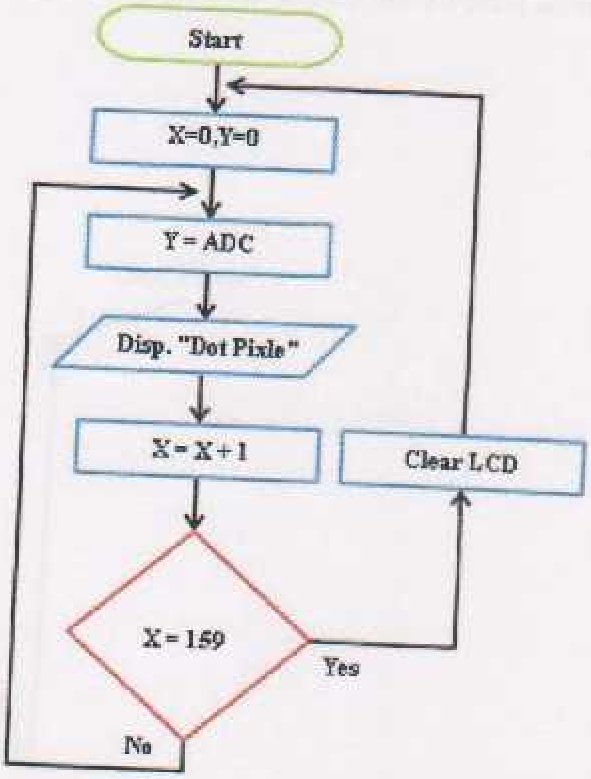


Figure 5. 8: Algorithm of PIC program

At the beginning we identify the variables X , Y to be used in our program which initialized to zero condition; next stage is taking the analog signal from the ECG amplifier -last stage- and convert it to digital signal by the internal ADC in the MCU and defined in the MCU as Y; after defined the first dot by Y = ADC value and X = 0, the MCU transfer the data to the LCD input pins, the X variable represent the X-axis dimension of the LCD and its 160 pixel from 0 to 159 dots.

5.8 LCD

In the project we use graphical LCD (DMF5001 NY-LY-AIE) with dimension of 160*128 dot ; this LCD can provide the ECG analog signal on the screen by connecting the MCU output ports to the input data port and control . For more information see Appendix-

In Figure (5.10) explains how the LCD display the dot-pixel according to time and voltage variation.

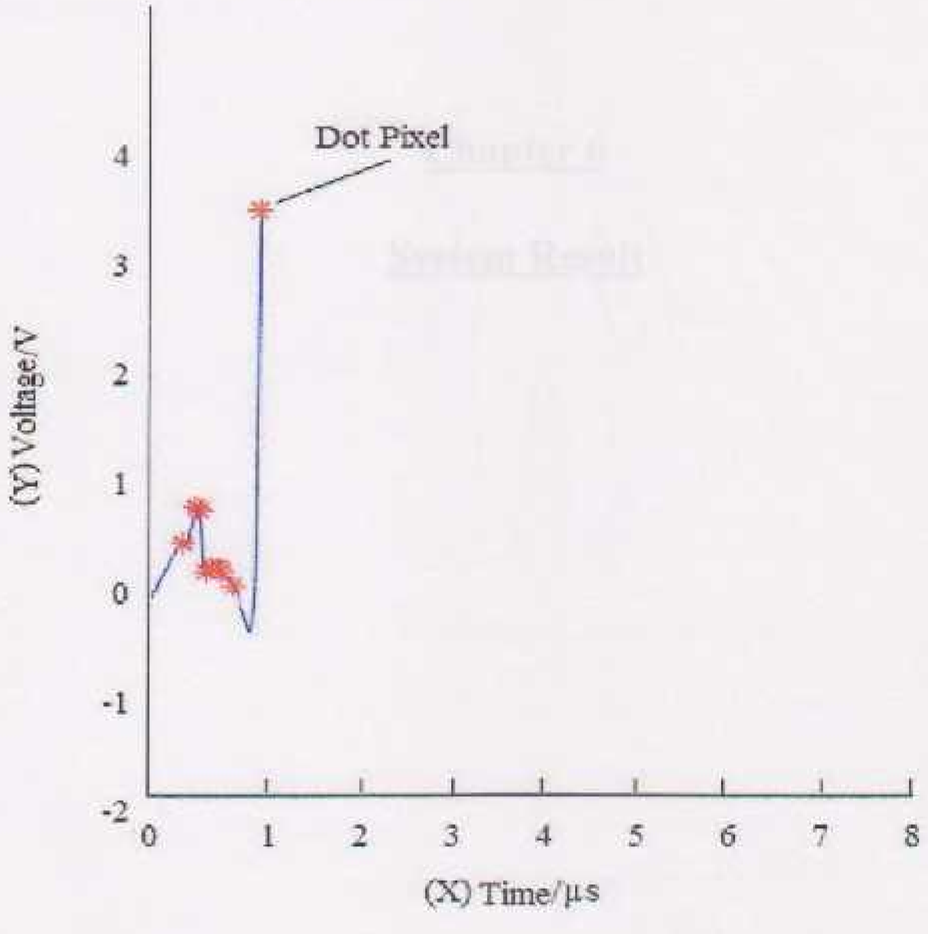


Figure 5. 9: Dot-Pixel according to time and voltage variation

Chapter 6

System Result

6.1 Introduction

This chapter explains and showing the practical implementation of the project , and this implementation started by implementing each individual subsystem. After completing this implementation, the individual subsystems are connected together to accomplish the project as one unit.

6.2 Electrode

In this system, the position of the three electrodes is fixed on the RA, LA and LL as shown in Figure (6.1).

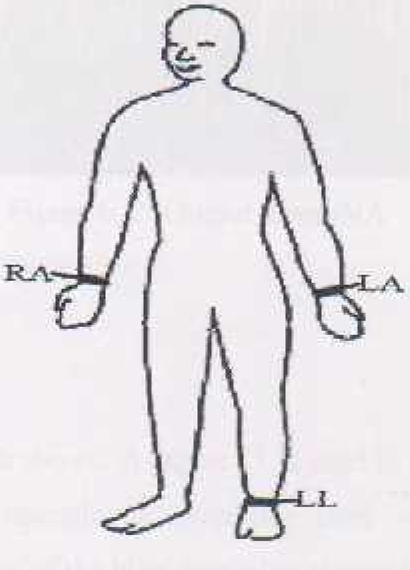


Figure 6. 1: Electrode position on human body

6.3 Instrumentation Amplifier:

Instrumentation Amplifier AD620 used for the amplification ECG signal .The output of the AD620 which shown in Figure (6.2) is the difference voltage of the two electrodes.

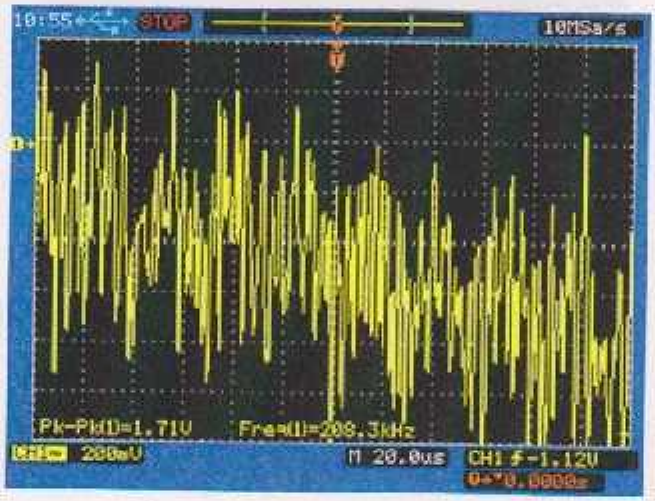


Figure 6. 2: Output from INA

6.4 High Pass Filter:

A high pass filter which shown in Figure (5.4) used to remove the DC offset comes from electrodes. TL082CN operational Amplifier used with passive component to implement this filter. The output of the high pass filter shown in Figure (6.3).

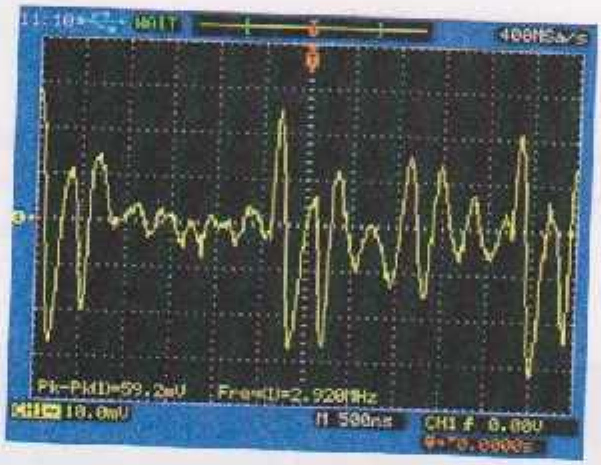


Figure 6. 3: Output from HPF

6.5 Inverting Amplifier:

Inverting amplifier shown in Figure (5.5) built by using TL082CN operational Amplifier, the output of the inverting amplifier is shown in Figure (6.4).

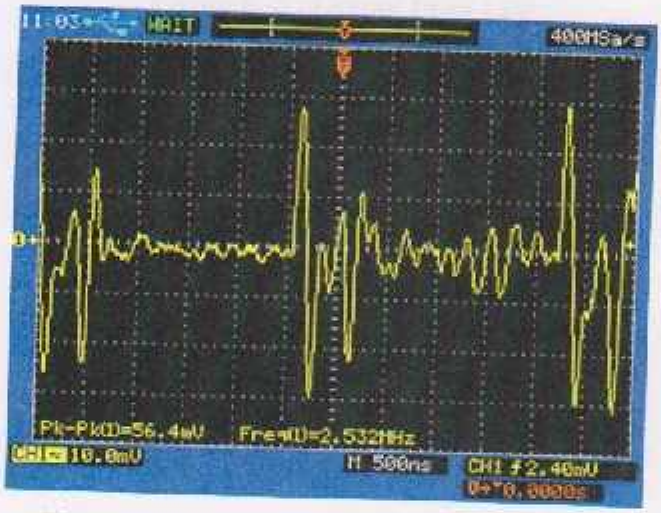


Figure 6. 4: Output from inverting amplifier

6.6 Low Pass Filter:

A low pass filter which shown in Figure (5.4) used to remove frequencies higher than 100Hz. TL082CN operational Amplifier used with passive component to implement this filter. The output of the low pass filter shown in Figure (6.5).

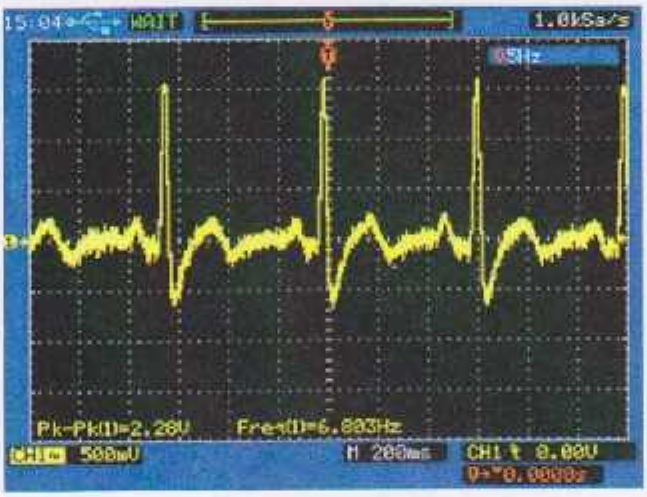


Figure 6. 5 : Output from LPF

Chapter 7

Future Work and Conclusion

7.1 Recommendations

In this project, the system has been designed to acquire and record three electrode single channel simultaneously, but these leads not enough to give the specialist with a total diagnostic, so additional three leads (a V_R, a V_L and a V_F) could be included in this system.

The channel where displayed on a graphical LCD which reinforces the portability of the system.

In order to acquire the ECG signals with optimal connection between the electrodes and the body, a lead fails detector circuit with conductivity indicator could be provided in this system.

7.2 Challenges

While designing the system, there are many challenges we're facing, such as:

- Not all the required component of the project are available in the Palestinian market; as a result some of the main components were purchased from Jordan.
- Some of the project components are expensive.
- Time response of the system which limits the sampling rate.

7.3 Conclusions

A single channel ECG system has been designed to record and display three electrode signal simultaneously, after designing, implementing and testing this system it is possible to:

- Connect the weak Leads voltages to the instrumentation amplifier .
- Record and display the single channel via three electrodes simultaneously.
- Attenuate the common mode voltages using driven right leg circuit connected to different locations on the body.

7.4 Future Work

Through technological progress and development the project can be developed as follows:

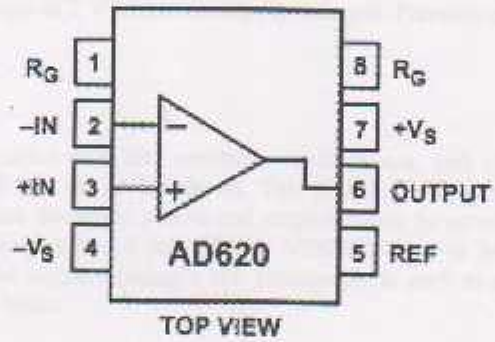
- The ECG signal could be sent by message to the doctor to be in touch with the patient continually and give diagnoses immediate.
- While the patient at home, sitting on the computer and to save power losses, the device could be connected to a computer through a medium to program on it .

References

- [1] John G. Webster, Encyclopedia of medical devices and instrumentation, Second Edition, Houghton Mifflin Company, Boston, MA. 2000
- [2] John G. Webster, Medical Instrumentation. Application and Design, Second Edition. Houghton Mifflin Company, Boston, MA. 1992.
- [3] Joseph D. Bronzino, Biopotintial amplifiers, The Biomedical Engineering Handbook, Second Edition. Boca Raton: CRC Press LLC, 1999.
- [4] Ron Mancifini, Op-am for everyone, Second edition, Texas Instruments Incorporated, 2009.
- [5] Han- Way Huang, PIC Microcontroller, An Introduction to Software and Hardware Interfacing, Cengage Learning, 2005.
- [6] American National Standard, Disposable ECG electrodes, Association for the Advancement of Medical Instrumentation (AAMI), 2010.
- [7] Amer Qureshi, PIC Microcontrollers For Absolute Beginners, Microtronics Pakistan, 2008.

Appendix A

AD620



GAIN SELECTION

The AD620's gain is resistor programmed by RG, or more precisely, by whatever impedance appears between Pins 1 and 8. The AD620 is designed to offer accurate gains using 0.1%–1% resistors. Table II shows required values of RG for various gains. Note that for G = 1, the RG pins are unconnected (RG = □). For

any arbitrary gain RG can be calculated by using the formula:

$$G = 1 + \frac{49.4K\Omega}{R_G}$$

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

Required Values of Gain Resistors

1% Std Table Value of R_G, Ω	Calculated Gain	0.1% Std Table Value of R_G, Ω	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

INPUT AND OUTPUT OFFSET VOLTAGE

The low errors of the AD620 are attributed to two sources, input and output errors. The output error is divided by G when referred to the input. In practice, the input errors dominate at high gains and the output errors dominate at low gains. The total VOS for a given gain is calculated as:

Total Error RTI = input error + (output error/G)

Total Error RTO = (input error ' G) + output error.

REFERENCE TERMINAL

The reference terminal potential defines the zero output voltage, and is especially useful when the load does not share a precise ground with the rest of the system. It provides a direct means of injecting a precise offset to the output, with an allowable range of 2 V within the supply voltages. Parasitic resistance should be kept to a minimum for optimum CMR.

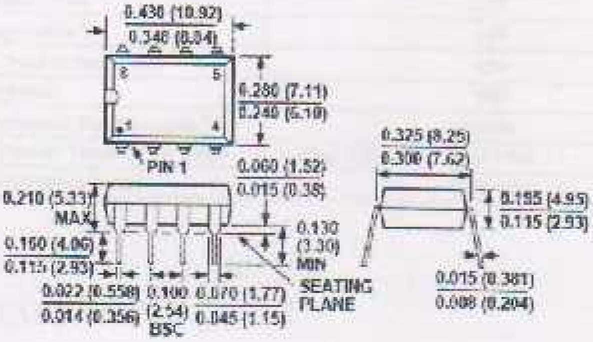
INPUT PROTECTION

The AD620 features 400 □□of series thin film resistance at its inputs, and will safely withstand input overloads of up to □15 V or □60 mA for several hours. This is true for all gains, and power on and off, which is particularly important since the signal source and amplifier may be powered separately. For longer time periods, the current should not exceed 6 mA (IIN □ VIN/400 □). For input overloads beyond the supplies, clamping the inputs to the supplies (using a low leakage diode such as an FD333) will reduce the required resistance, yielding lower noise.

AD620

OUTLINE DIMENSIONS

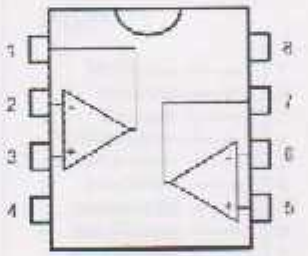
Dimensions shown in inches and (mm).



Appendix B

TL082

PIN CONNECTIONS (top view)

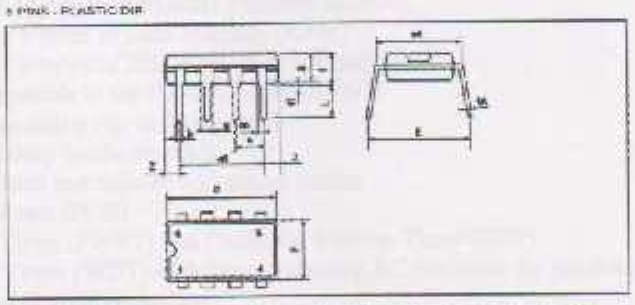


1 - Offset null 1
2 - Inverting input 1
3 - Non-inverting input 1
4 - V_{CC}^{-}
5 - Non-inverting input 2
6 - Inverting input 2
7 - Output 2
8 - V_{CC}^{+}

ABSOLUTE MAXIMUM RATINGS

Symbol	Parameter	TL082M, AM, BM	TL082I, AI, BI	TL082C, AC, BC	Unit
V_{CC}	Supply voltage - note [1]	±18			V
V_i	Input Voltage - note [2]	±15			V
V_{id}	Differential Input Voltage - note [3]	±30			V
P_{tot}	Power Dissipation	680			mW
	Output Short-circuit Duration - note [4]	Infinite			
T_{oper}	Operating Free-air Temperature Range	-55 to +125	-40 to +105	0 to +70	°C
T_{stg}	Storage Temperature Range	-65 to +150			°C

PACKAGE MECHANICAL DATA

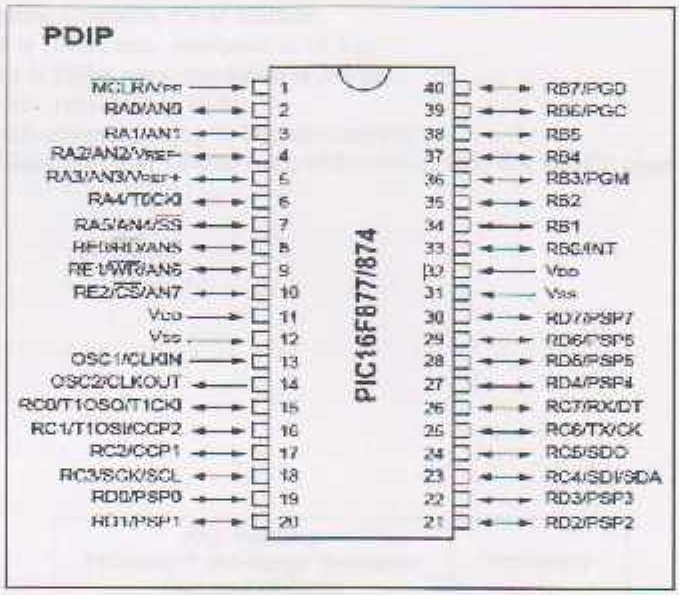


Dim.	Millimeters			Inches		
	Min.	Typ.	Max.	Min.	Typ.	Max.
A		3.32			0.131	
a1	0.51			0.020		
B	1.15		1.65	0.045		0.065
b	0.356		0.55	0.014		0.022
b1	0.204		0.304	0.008		0.012
D			10.92			0.430
E	7.95		9.75	0.313		0.384
e		2.54			0.100	
e3		7.62			0.300	
e4		7.62			0.300	
F			6.6			0.260
i			5.08			0.200
L	3.18		3.81	0.125		0.150
Z			1.52			0.060

Appendix C

PIC16F877 Microcontrollers

Pin Diagram



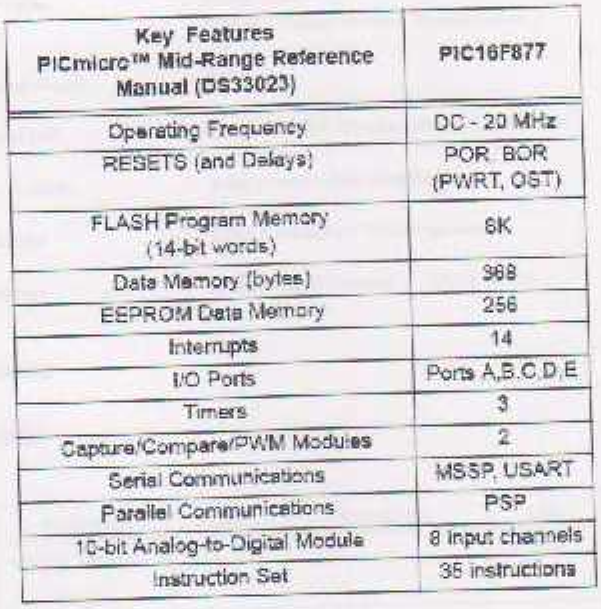
Microcontroller Core Features:

- High performance RISC CPU
- Only 35 single word instructions to learn
- All single cycle instructions except for program branches which are two cycle
- Operating speed: DC - 20 MHz clock input DC - 200 ns instruction cycle
- Up to 8K x 14 words of FLASH Program Memory,
  Up to 368 x 8 bytes of Data Memory (RAM)
  Up to 256 x 8 bytes of EEPROM Data Memory
- Pinout compatible to the PIC16C73B/74B/76/77
- Interrupt capability (up to 14 sources)
- Eight level deep hardware stack
- Direct, indirect and relative addressing modes
- Power-on Reset (POR)
- Power-up Timer (PWRT) and Oscillator Start-up Timer (OST)
- Watchdog Timer (WDT) with its own on-chip RC oscillator for reliable operation
- Programmable code protection
- Power saving SLEEP mode
- Selectable oscillator options
- Low power, high speed CMOS FLASH/EEPROM technology
- Fully static design
- Single 5V In-Circuit Serial Programming capability
- Processor read/write access to program memory
- Wide operating voltage range: 2.0V to 5.5V
- High Sink/Source Current: 25 mA

Peripheral Features:

- Timer0: 8-bit timer/counter with 8-bit pre-scalar
- Timer1: 16-bit timer/counter with pre-scalar, can be incremented during SLEEP via external crystal/clock
- Timer2: 8-bit timer/counter with 8-bit period register, pre-scalar and post-scalar
- Two Capture, Compare, PWM modules
  - Capture is 16-bit, max. resolution is 12.5 ns
  - Compare is 16-bit, max. resolution is 200 ns
  - PWM max. resolution is 10-bit
- 10-bit multi-channel Analog-to-Digital converter
- Parallel Slave Port (PSP) 8-bits wide, with external RD, WR and CS controls (40/44-pin only).

Key Features :



Key Features PICmicro™ Mid-Range Reference Manual (DS33023)	PIC16F877
Operating Frequency	DC - 20 MHz
RESETS (and Delays)	POR, BOR (PWRT, OST)
FLASH Program Memory (14-bit words)	8K
Data Memory (bytes)	368
EEPROM Data Memory	256
Interrupts	14
I/O Ports	Ports A,B,C,D,E
Timers	3
Capture/Compare/PWM Modules	2
Serial Communications	MSSP, USART
Parallel Communications	PSP
10-bit Analog-to-Digital Module	8 input channels
Instruction Set	35 instructions

Appendix D

DMF5001 NY-LY-AIE

General Specifications

Operating Temp : min. 0℃ ~ max. 50℃

Storage Temp. : min. -20℃ ~ max. 60℃

Dot Pixels : 160 (W) × 128 (H) dots

Dot Size : 0.54 (W) × 0.54 (H) mm

Dot Pitch : 0.58 (W) × 0.58 (H) mm

Viewing Area : 101.0 (W) × 82.0 (H) mm

Outline Dimensions : 129.0 (W) × 102.0 (H) × 19.2 max. (D) mm

Weight : 190g max.

LCD Type : NTD-7353
(STN / Yellow-mode / Transmissive)

Viewing Angle : 6:00

Control LSI : T6963C-0101 (Produced by TOSHIBA)

Data Transfer : 8-bit parallel data transfer

Backlight : LED Backlight / Yellow-green

Drawings : Dimensional Outline UE-34487A

Electrical Specifications

Absolute Maximum Ratings

Vss=0V

Parameter	Symbol	Conditions	Min.	Max	Units
Supply Voltage (Logic)	Vcc-Vss	–	-0.3	7.0	V
Supply Voltage (LCD Drive)	Vcc-VEE	–	-0.3	30.0	V
Input Voltage	V_I	–	-0.3	Vcc+0.3	V

DC Characteristics

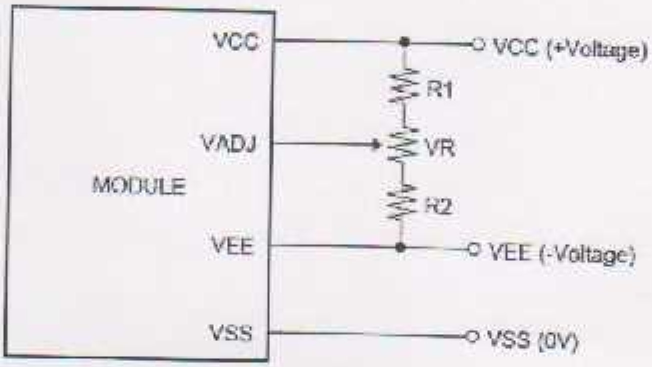
Ta=25℃, Vss=0V

Parameter	Symbol	Conditions	Min.	Typ.	Max.	Units
Supply Voltage (Logic)	$V_{CC}-V_{SS}$	—	4.5	—	5.5	V
Supply Voltage (LCD Drive)	$V_{CC}-V_{EE}$	—	23.0	—	26.0	V
	$V_{CC}-V_{ADJ}$	Shown in 3.1				V
High Level Input Voltage	V_{IH}	$V_{CC}=5.0V\pm10\%$	$V_{CC}-2.2$	—	V_{CC}	V
Low Level Input Voltage	V_{IL}	$V_{CC}=5.0V\pm10\%$	0	—	0.8	V
High Level Output Voltage	V_{OH}	$I_{OH}=-0.75mA$	$V_{CC}-0.3$	—	V_{CC}	V
Low Level Output Voltage	V_{OL}	$I_{OL}=1.75mA$	0	—	0.3	V
Supply Current	I_{CC}	$V_{CC}-V_{SS}=5.0V$	—	13.4	30.0	mA
	I_{EE}	$V_{CC}-V_{ADJ}=18.9V$	—	3.7	20.9	mA

I/O Terminal

No	Symbol	Level	Function
1	FG	—	Frame Ground
2	V_{SS}	—	Power Supply (0V, GND)
3	V_{CC}	—	Power Supply for Logic
4	V_{ADJ}	—	Voltage Level for LCD Contrast Adjustment
5	V_{EE}	—	Power Supply for LCD Drive
6	$\overline{WR}$	H / L	Write Signal L : Active
7	$\overline{RD}$	H / L	Read Signal L : Active
8	$\overline{CE}$	H / L	Chip Enable Signal L : Active
9	$C/\overline{D}$	H / L	Write Mode H : Command Write L : Data Write Read Mode H : Status Read L : Data Read
10	$\overline{HALT}$	H / L	Clock Operating Stop Signal L : Halt
11	$\overline{RESET}$	H / L	Reset Signal L : Reset
12	D0	H / L	Display Data
13	D1	H / L	Display Data
14	D2	H / L	Display Data
15	D3	H / L	Display Data
16	D4	H / L	Display Data
17	D5	H / L	Display Data
18	D6	H / L	Display Data
19	D7	H / L	Display Data
20	NC	—	Non-connection
21	LED A	—	LED Anode Terminal
22	LED K	—	LED Cathode Terminal

It is recommended to apply a potentiometer for the contrast adjust due to the tolerance of the driving voltage and its temperature dependence.



R1+R2+VR=10~20KΩ