Palestine Polytechnic University

College of Engineering & Technology

Electrical & Computer Engineering Department

Introduction to Graduation project

Design of a Three Electrodes , Three Leads ECG System

Project 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 department of electrical and computer engineering in the college of engineering and technology , to be as a partial fulfillment of the requirement of the department for the degree of B.sc .

Supervisor signature

Testing committee signature

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The head of department signature

جامعة بوليتكنك فلسطين

الخليل – فلسطين

كلية الهندسة والتكنولوجيا

دائرة الهندسة الكهربائية والحاسوب

Design of a Three Electrodes ,Three Leads ECG System

فريق المشروع

إيهاب أبو الرُّب محمد القيسي نضال زماعرة

بناء على نظام كلية الهندسة والتكنولوجيا وإشراف ومتابعة المشرف المباشر على المشروع وموافقة أعضاء اللجنة المناقشة , تم تقديم هذا العمل إلى دائرة الهندسة الكهربائية والحاسوب . وذلك للوفاء بمتطلبات درجة البكالوريوس في هندسة الأجهزة الطبية .

توقيع المشرف

توقيع اللجنة المناقشة

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توقيع رئيس الدائرة

الإهداء

إلى الذي علمنا بأننا وكل ما نملك من علم وبناء هو لفلسطين ,,, إلى من علمنا الشهادة مع وقف التنفيذ ,,, إلى رئيس دولتنا الأول وقائدها دوماً ,,, إلى روح الشهيد ياسر عرفات " أبو عمار " ,,,

إلى آبائنا ,,, خير منشئين ومربين ,,, وخير من يرتقبوننا في كل لحظة نصنع بها نجاحاً ,,, فإلى آمالهم وثقتهم ,,, وإلى أسمائهم التي نحملها بكل فخر ,,,

إلى أمهاتنا ,,, وشغف ما زال في القلب ينفطر ,,, إلى كل صباح كنتم به تبدؤونه بالدعاء لنا ,,, وإلى كل مساء كنتم به تحلمون بنجاحنا ,,, إلى ملائكة الصبر والتفاؤل والأمل ,,,

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نقـــــدم لكم جميعا باسمكم عمـــــلنا البسيط هذا

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نضال زماعرة

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Abstract

The overall objective of this project is to provide a new ECG system Design based on a modified single channel ECG system with multiplexer circuit to record and display the three ECG leads (I, II and III) simultaneously by three Electrodes to reduce complexity associated with traditional multi-channel ECG system.

The suggested system record the three leads by switching the three electrodes on single instrumentation amplifier , in Each stage two electrodes are active and the remaining electrode act is the reference .

ملخص المشروع

تقوم فكرة هذا المشروع بتزويد نظام جديد يستند على التخلص من المشاكل التقنية الموجودة في النظام القديم لجهاز راسم الإشارة الكهربائية للقلب والتي تتمحور حول وجود نظام متعدد القنوات والمشاكل الناتجة عنه ومن أهمها وجود عدة مكونات والسيئات الناتجة عنها والتكلفة العالية جدا لهذه المكونات .

المشروع يقدم حلا يتم من خلال إلغاء نظام متعدد القنوات باستبداله بنظام وحيد القناة يتم تجميع الإشارات الثلاث التي نهدف قياسها ويمكنا المشروع من إظهار الإشارات الثلاث على وحدة العرض مما يخدم في عملية التشخيص ودقتها.

Contents

Chapter Five : System Design

Chapter Six : System Implementation & Results

Chapter seven : Future Work and Conclusion

List of Figures

List of Tables

CHAPTER

1

INTRODUCTION

1.1 Overview:

The Electrocardiograph signal is one of the most important bio-signal that used in diagnosis for the most important organ in the body, from old ages, its known what is the important of the hearts so many techniques used for heart diagnosis.

Traditional ECG's system developed from single channel which record one lead at a time to multi-channel ECG systems which record multiple leads simultaneously, all leads recording performed through standard 10 electrodes.

This project will provide a method for reducing traditional ECG's drawbacks by recording multiple leads simultaneously through a three electrodes multiplexed over single channel ECG system.

1.2 Project Importance:

The importance of the project comes from its wide benefits for providing a low cost and simple system design with minimum number of electrodes; all these benefits reinforced the ECG portability.

1.4 Literature Review:

The study of this project was performed independent of any other projects the current ECG's system measure and record three leads (I, II and III) via three channel ECG system and four electrode.

No study and projects founds in the field of measuring and recording the three leads via single channel ECG system and three electrodes.

1.5 Project Cost:

Table 1.1: Hardware Cost

Main Component	Cost
Amplifier (Instrumentation Amplifier, Op-Amp's)	20$
Display device (LCD)	100$
Microcontroller	21$
Electronics Elements	5$
Electrodes	10$
Total	**156 $**

1.6 Scheduling Table:

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

Table 1.2: Timing schedule of the First Semester

Task\week	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15
Collect information															
Basic Design															
Specification Design															
Documentation															
advanced features															

Table 1.3: 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															

CHAPTER

2

PHYSIOLOGY

Introduction

A primary objective of this introductory chapter is to introduce the reader to some of the basic of any electrical behavior of the heart; also this chapter is going to illustrate the anatomy and physiology of the heart.

2.1 Anatomy and function of the heart:

From *figure **2.1*** we can see the heart serves as a four-chambered pump for the circulating system. The upper two chambers are called the atria, and the lower two chambers are called the ventricles. Its main pumping function is supplied by the ventricles. The atria are thin-walled, low-pressure pumps, merely antechambers to store blood during the time the ventricles are pumping. The resting or filling phase of the heart cycle is referred as diastole. Whereas the contractile or pumping phase is called systole.

The smooth, rhythmic contraction of the atria and ventricles has an underlying electrical precursor in the form of a well-coordinated series of electrical events that takes place within the heart. These electrical activation patterns in the wall of the atria and ventricles are initiated by a coordinated series of events within the specialized conduction system of the heart.

In relation to the heart as a whole, the specialized conduction system is very small and constitutes only a minute portion of the total mass of the heart. The wall of the left ventricle is 2.5 to 3.0 times as thick as the right ventricular wall. Considering the heart as a bioelectric source, the source strength at each instant can be expected to be directly related to the active muscle mass at that moment.

2.2 Electrical behavior of the heart:

From cellular physiology we recall that bio-potentials are produced as a consequence of chemical activity of excitable or irritable cells. Excitable cells are components of the neural, muscular, glandular as well as many plant tissues. More specifically, bio-potentials are generated as a consequence of ionic concentration difference of electrolytes (mainly Na^+, K^+, Cl^- ions) across the cellular membrane of excitable cells. Ionic differences are maintained by membrane permeability properties and active transport mechanisms across the cellular membrane (ionic Pumps). [1]

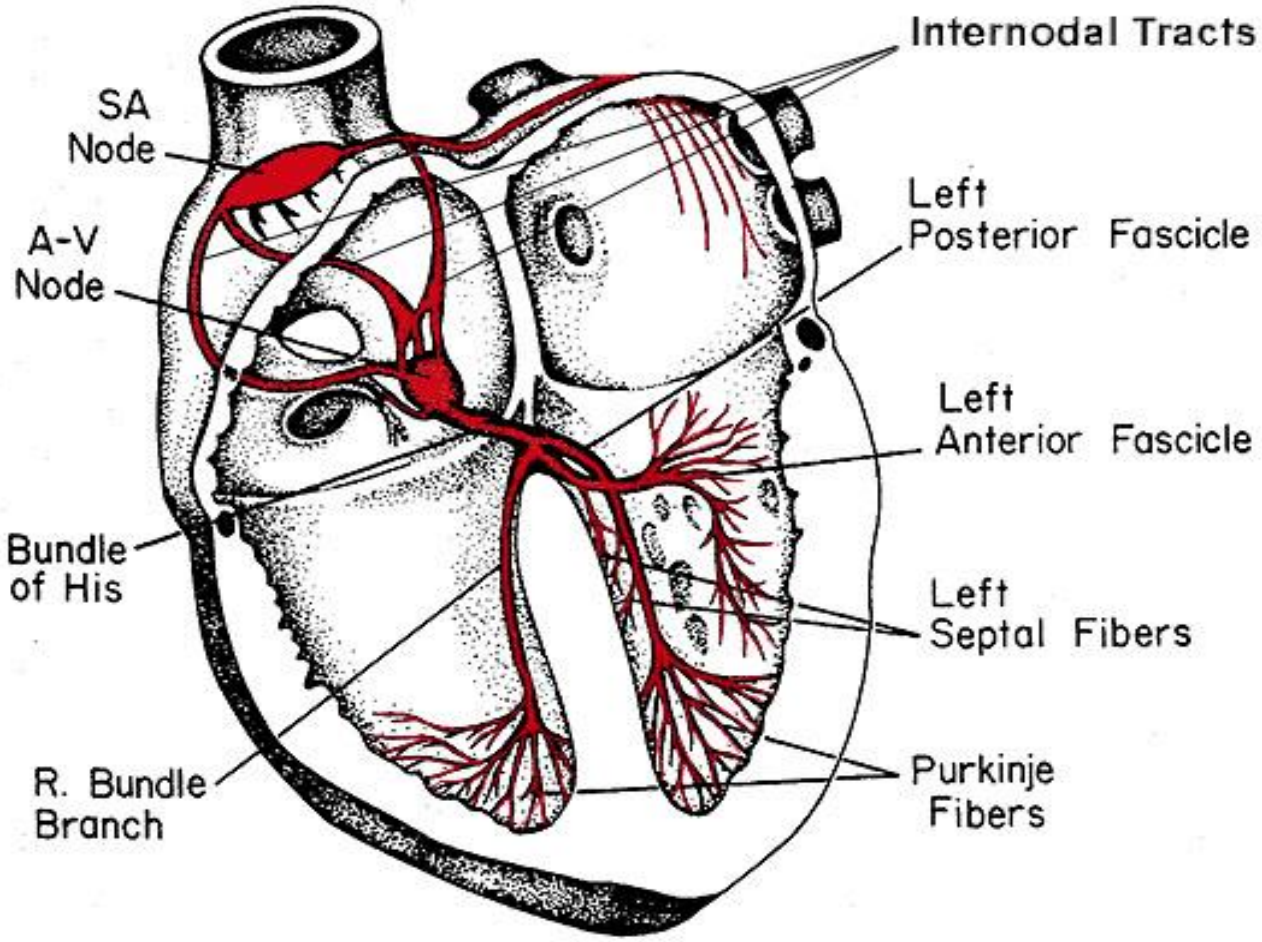

Figure 2.1: Anatomy of the heart, showing the Distribution of specialized conductive tissues in the atria and the ventricles. [1]

The cellular membrane is a semipermeable lipid bilayer that separates the extracellular and intracellular fluids having different ionic concentrations. As a consequence of semipermeability and differences in concentration of ions, electrochemical gradients are set up across the membrane. Ionic transfer across the membrane by diffusion and active transport mechanisms results in the generation of a voltage difference (membrane potential), which is negative inside. The resting membrane potential is mainly established by the efflux of K^+ ions due to diffusion and is balanced by the consequent inward electric field due to charge displacement. This equilibrium voltage can be estimated by the Nernst equation, which results from application of electric field theory and diffusion theory. [1]

If we consider the effects of the three main ions, potassium (K^+), sodium (Na^+), and chloride (Cl^-), the Goldman– Hodgkin and Katz equation can be used to calculate the resting membrane potential.

The smallest sources of bioelectric signals (bio-sources) are single excitable cells. These cells exhibit a quiescent or resting membrane potential across the cellular membrane of several

[mV] (-90 mV with several hundred milliseconds in duration for ventricular myocytes). When adequately stimulated, the transmembrane potential in excitable cells becomes positive inside with respect to outside (depolarization) and action potentials are generated (at the peak of the action potential in ventricular myocytes, the membrane potential reaches about +20 mV).

Action potentials as shown in *figure **2.2*** are produced by sudden permeability changes of cellular membrane to ions, primarily sodium and potassium ions. Action potentials are all-or-none monophasic waves of depolarization that travel unattenuated with a constant amplitude and speed along the cellular membrane.

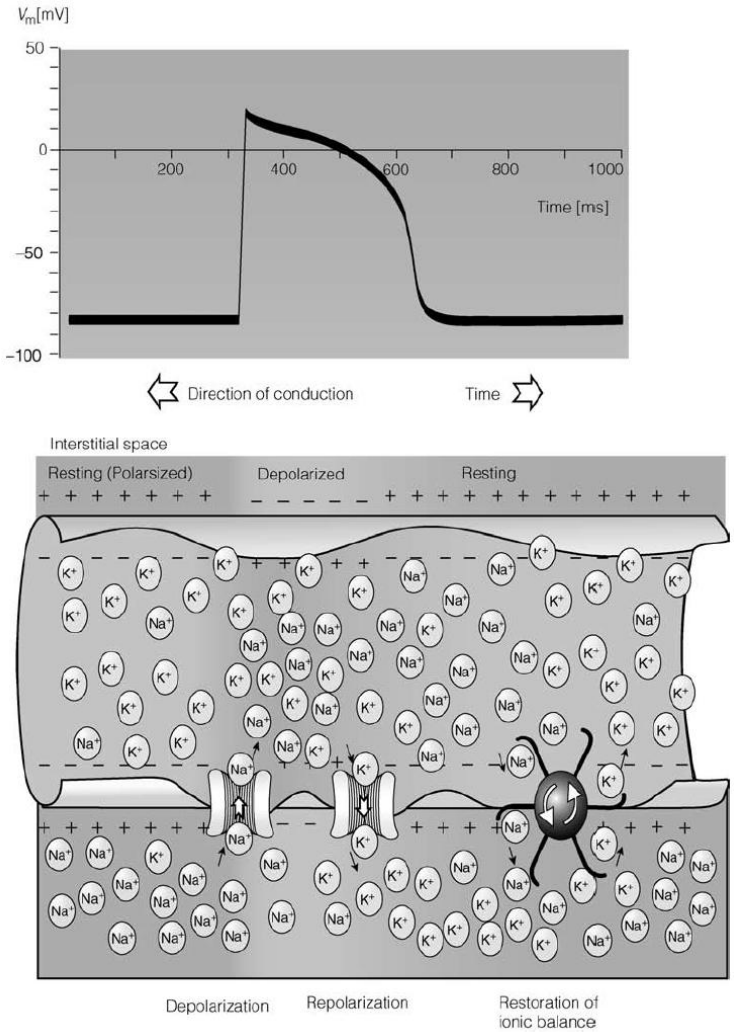

Figure2.2 The monophasic action potential, direction of conduction of the action potential, and movement of ions across the cellular membrane. [1]

The excitable cells function in large groups as a single unit and the net effect of all stimulated (active) cells produces a time-varying electric field in the tissue surrounding the bio-source. The surrounding tissue is called a volume conductor.

The electric field spreads in the volume conductor and can be detected as small voltages by means of bio-electrodes or simply electrodes placed in the tissue or on the skin. Electrodes are transducers, which convert ionic current flow in the living tissue to electrical current flow in the electro-medical instrument.

A simplified version of the volume conductor problem at the cellular level can be considered as follows. If a single excitable cell is placed in a conductive medium, it acts like a constant current source. When the bio-source becomes adequately depolarized, an action potential is generated across its membrane and it injects a current to the surrounding medium.

The conductive medium presents as a load with a long range of loading conditions depending on its geometry, temperature, and so on. The lines of current flowing out of the excitable cell into the volume conductor with a specific resistance (R), gives rise to an extracellular field potential proportional to the transmembrane current (i_m) and the medium resistance (R) according to Ohm's law. Obviously, the extracellular field potential increases with higher values of membrane current or tissue resistance.

In summary, excitable cells and tissues (bio-sources), when adequately stimulated, generate monophasic action potentials. These action potentials cause the injection of constant currents into a large medium surrounding the bio-source (considered as a point current source). As a result of the current flow in the volume conductor with specific resistance, extracellular field potentials are generated in the medium. As the resistivity of the medium increases and the radial distance from the bio-source decreases, the field potential increases. These field potentials are recorded as clinically useful signals on the body surface.

2.4 **Basics of Electrocardiography:**

The conduction system of the heart consists of the sinoatrial (SA) node, the internodal tracts, the atrioventricular (AV) node, the bundle of histidene (His), the right bundle branch (RBB), the left bundle branch (LBB), and the Purkinjie network.

The rhythmic electrical activity of the heart (cardiac impulse) originates in the SA node. This node is known as the natural pacemaker of the heart, approximately the size of the tip of a pencil, located at the junction of the superior vena cava and the right atrium. The impulse then propagates through internodal and interatrial (Buchmans's bundle) tracts.

As a consequence, the pacemaker activity reaches the AV node by cell-to-cell atrial conduction and activates the right and left atrium in an organized manner. The pacemaker action potential has a fast activation phase, a very short steady recovery phase, followed by a fairly rapid recovery phase and a characteristic slow depolarization phase leading to self-excitation as shown below in *figure **2.3***. The pacemaker cells of the SA node act as a biological oscillator.

As atria and ventricles are separated by fibrous tissue, direct conduction of cardiac impulse from the atria to the ventricles cannot occur and activation must follow a path that starts in the atrium at the AV node. The cardiac impulse is delayed in the AV node for ~ 100 ms.

It then proceeds through the bundle of His, RBB, LBB, and finally to the terminal Purkinjie fibers that arborize and invaginate the endocardial ventricular tissue.

The delay in the AV node is beneficial since electrical activation of cardiac muscle initiates its successive mechanical contraction. This delay allows enough time for completion of atrial contraction and pumping of blood into the ventricles.

Once the cardiac impulse reaches the bundle of His, conduction is very rapid, resulting in the initiation of ventricular activation over a wide range. The subsequent cell-to-cell propagation of electrical activity is highly sequenced and coordinated resulting in a highly synchronous and efficient pumping action by the ventricles. [1]

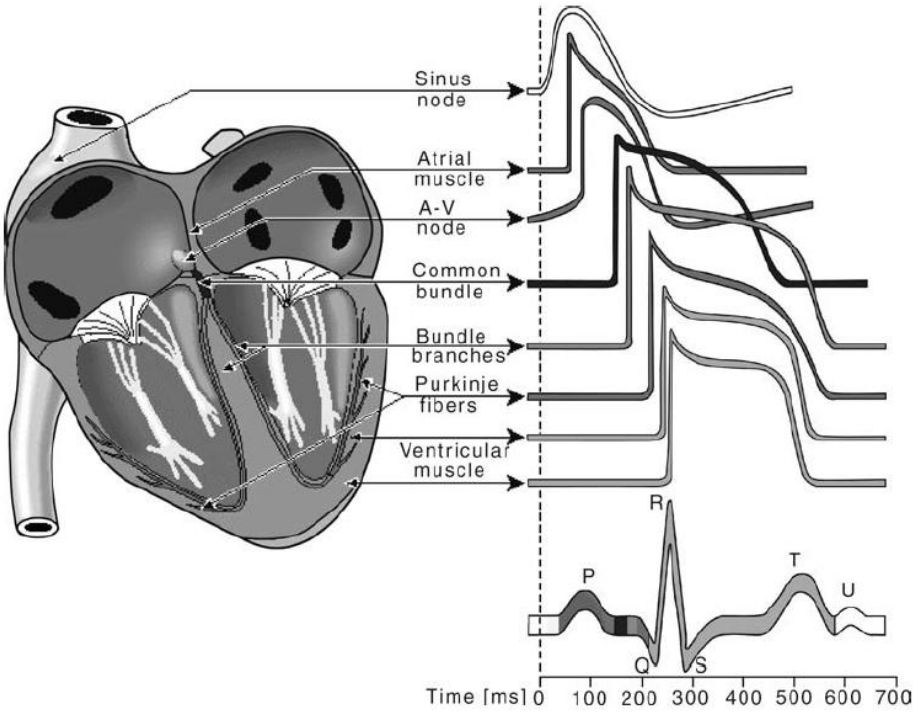

Figure2.3: Waveforms of action potentials in different specialized cells in the conductive pathway of a normal heart and their contribution with color coding to the surface ECG. [1]

Essentially, an overall understanding of the genesis of the ECG waveform can be based on a cardiac current dipole model placed in an infinite (extensive) volume conductor. In this model, an active (depolarizing) region of the tissue is considered electronegative with respect to an inactive (repolarizing) region. Therefore, a boundary or separation exists between negative and positive charges.

This is regarded as a current dipole, a current source and sink separated by a distance. According to the dipole concept, a traveling excitation region can be considered as a dipole moving with its positive pole facing the direction of propagation.

Thus a nearby recording electrode placed in the surrounding volume conductor (referenced to an indifferent electrode placed in a region of zero potential) will detect a positive-

going field potential as excitation approaches and a negative-going field potential as it passes away.

Consequently, an upward deflection in the bio-potential recording indicates the approaching of excitation (depolarization) toward the positive (recording) electrode and a downward deflection indicates a recovery (Repolarization) in the recorded signal.

As the wave of excitation (depolarization) spreads throughout the conductive pathways and tissues, specific excitation regions are synchronously excited. In the ventricles, these synchronous activation regions propagate in a temporally and spatially orderly fashion from the endocardial to the epicardial direction.

In a localized region of the heart many cells are simultaneously activated because of the high electrical and mechanical coupling between the myocardial cells. Each activation region can be viewed as an elementary dipole, and all elementary dipoles could be vectorially added to all others to form a single net dipole.

Therefore, at each instant of time, the total cardiac activity can be represented by a net equivalent dipole current source. The electric field produced by this dipole source represents the total electrical activity of the heart and is recorded at the body surface as the ECG signal.

In summary, based on the aforementioned concepts, Electrocardiographers have developed an oversimplified model to explain the electrical activity of the heart. In this model, the heart is considered as an electric dipole (points of equal positive and negative charges separated from one another by a distance), denoted by a spatiotemporally changing dipole moment vector.

This dipole moment (amount of charge time's distance between positive and negative charges) is called the cardiac vector. As the wave of depolarization spreads throughout the cardiac cycle, the magnitude and orientation of the cardiac vector changes and the resulting bioelectric potentials appear throughout the body and on its surface also changes.

CHAPTER

3

THE LEADS CONCEPT

Introduction

As discussed in previous chapter, the beating heart generates an electric signal that can be used as a diagnostic tool for examining some of the heart function. This electric activity of the heart can be approximately represented as a vector quantity. Thus it is important to know the location at which signals are detected, as well as the time dependence of the amplitude of the signals.

3.1 Cardiac Vector:

Electrocardiographers have developed a simple model to represent the electric activity of the heart .In this model, the heart consists of an electric dipole (points of equal positive and negative charges separated from one another by a distance), denoted by a spatiotemporally changing dipole moment vector, located in the partially conducting medium of the thorax.

This particular field and the dipole that produces it represent the electric activity of the heart at specific instant. At the next instant the dipole can change its magnitude and its orientation , causing a change in the electric field .Thus in this simplified model no need to draw a field plot every time to discuss the dipole field of the heart. Instead, it can be represented by its dipole moment; a vector directed from the negative charge to the positive charge and has a magnitude proportional to the amount of charge (either positive or negative) multiplied by the separation of the two charges. [2]

In electrocardiography this dipole moment, known as the cardiac vector, is represented by **M**, as shown in *figure **3.1***. As we progress through a cardiac cycle, the magnitude and direction of **M** vary because the dipole field varies.

The electric potential generated by the heart appears throughout the body and on its surface. potential difference can be measured by placing electrodes on the surface of the body and measuring the voltage between them, being careful to draw little current (ideally there should be no current at all, because current distort the electric field that produce the potential differences).

If the two electrodes are located on different equal-potential lines of the electric field of the heart, a nonzero potential difference or voltage is measured, different pairs of electrodes at different locations generally yield different voltages because of spatial dependence of the electric field of the heart. [2]

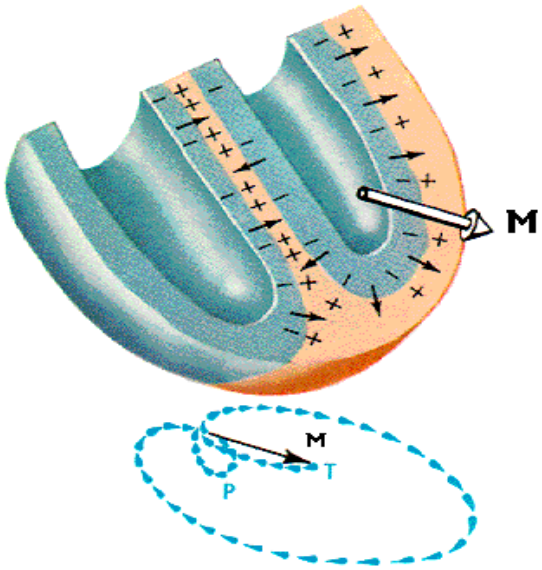

Figure 3.1: The total cardiac electrical activity represented by the net dipole (cardiac vector **M**) during different phases of the cardiac cycle.

3.2 ECG Acquisition

In order to record an ECG waveform, a differential recording between two points on the body are made, thus it is important to have certain standard positions for clinical evaluation of the ECG ,the limbs make fine guideposts for locating the ECG electrodes.

For a cardiac vector **M** as shown in *figure **3.2***, the voltage induced in a lead represented by the lead vector $\mathbf{a_1}$ ($\mathbf{a_1}$ is a unit vector) is given by the component of **M** in the direction of $\mathbf{a_1}$. In vector algebra, this can be denoted by the dot product

$$\mathbf{V_{a1} = M.a_1}$$

Where $\mathbf{V_{a1}}$ is the scalar voltage seen in the lead that has the vector $\mathbf{a_1}$, because the cardiac vector **M** is oriented in space so as to be perpendicular to the lead vector $\mathbf{a_1}$, the component of **M** along the direction of $\mathbf{a_1}$ is zero and no voltage is seen in this lead, so another unit lead vector $\mathbf{a_2}$ is considered to describe the cardiac vector uniquely, by using these two leads vector both of which lie in the same plane as the cardiac vector, the cardiac vector can be uniquely described, similarly at instant of time the cardiac vector is perpendicular to the lead vector $\mathbf{a_2}$ as shown in *figure **3.2***, at this time the cardiac vector can be described by lead vector $\mathbf{a_1}$.

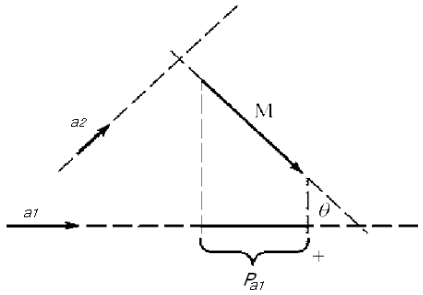

Figure 3.2: Relation between the two lead vectors a_1 and a_2 and cardiac vector **M**.

In clinical electrocardiography, more than one lead must be recorded to describe the hearts electric activity fully, in practice, several leads are taken in the frontal plane (the plane of your body that is parallel to the ground when you are lying on your back) and the transverse plane (the plane of your body that is parallel to the ground when you are standing erect).

Three basic leads make up the frontal-plane ECG, these are derived from the various permutation of pairs of electrodes when one electrode is located on the right arm (RA) ,the left arm (LA),and the left leg (LL), very often another reference electrode is also placed on the right leg (RL) and grounded or connected to special circuits.

The resulting three leads shown in *figure **3.3*** are lead **I**, LA to RA; lead **II**, LL to RA, and lead **III**, LL to LA .The lead vectors that are formed can be approximated as an equilateral triangle, known as Einthoven's triangle in the frontal plane of the body.

Because the body is assumed to be purely resistive at ECG frequencies, the four limbs can be imagined as wires attached to the torso. Hence lead **I** could be recorded from the respective shoulders without a loss of cardiac information. [3]

Another nine Leads Clinical Electrocardiography in addition to the aforementioned three bipolar limb leads (**I**, **II**, **III**) comprised the standard clinical ECG system, three augmented leads (aVR, aVL, aVF); and six unipolar chest or precordial leads (V_1, V_2, V_3, V_4, V_5, V_6).

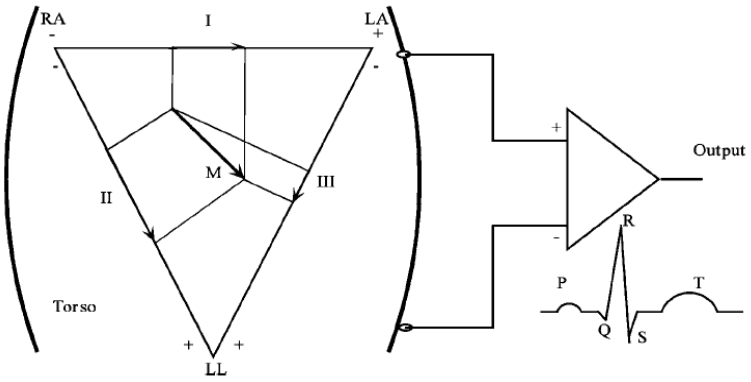

Figure 3.3: Einthoven equilateral triangle.

CHAPTER

4

ECG ELECTRODES & TRADITIONAL SYSTEM'S

Introduction

The ECG signal is one of the most important bio-signals that reveals several diagnosis, hence several attempts to measure the signal in several methods are occurred, this chapter give a study about these designs in conjunction with electrodes used to provide some interface between the body and these systems.

4.1 ECG Electrodes:

In order to measure and record potentials and current in the body, it is necessary to provide some interface between the body and the electronic measuring apparatus, bio-potential electrodes carry out this interface function. Because current in the body is carried by ions, where as in leads wire and measuring circuitry by electrons, thus the electrode must serve as a transducer which convert the ionic current into electrical current.

As in any measurement of potential, a very little current is flow in the measuring circuit for at least a fraction of the period of time over which the measurement is made, so bio-potential electrodes must have the capability of conducting a current across the interface between the body and the electronic measuring circuit. [2]

4.1.1 The Electrode-Electrolyte Interface:

The passage of an electric current between the body and electrodes can be explained by examining the electrode-electrolyte interface, from *figure.**4.1*** the electrode consist of metallic atoms C where the electrolyte which represent the body contain aqueous solution containing cation of the electrode metal C^+ and anions A^-.

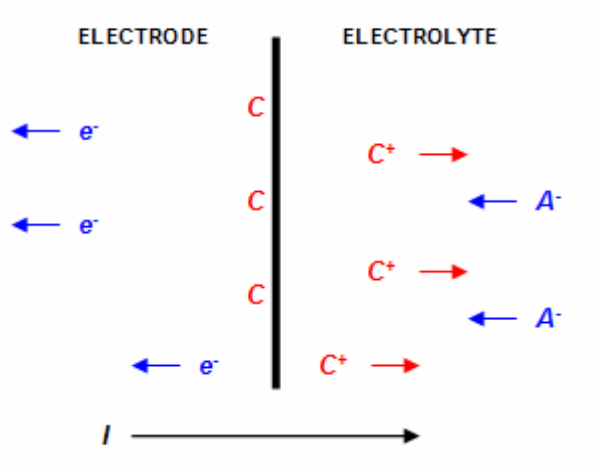

Figure 4.1: Electrode-Electrolyte Interfaces [2].

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:

$$C \rightleftharpoons C^{n+} + ne^- \qquad (4.1)$$

$$A^{m-} \rightleftharpoons A + me^- \qquad (4.2)$$

where n is the valence of cation material C, and m is the valence of anion material A, in the first *equation **4.1*** the electrode material C become oxidized to form a cation and one or more free electrons, the cation discharged into the electrolyte, electrons remains as the charge carrier in the electrode, These ions are reduced when the process occurs in the reverse direction.

In the case of the anion reaction, *equation **4.2*** 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 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.

When two ionic solutions of different activity are separated by an ion-selective semipermeable 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 ions in each of these solutions, this relationship is known as the Nernst equation [2]

$$E = -\frac{RT}{nF} \ln\left(\frac{a_1}{a_2}\right) \qquad (4.3)$$

where $\mathbf{a_1}$ and $\mathbf{a_2}$ are the activities of the ions 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]

4.1.2 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 elements are required for modeling electrode behavior, electrodes can be represented by an equivalent circuit of the form shown in *figure **4.2***. In this circuit R_d accounts 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 from 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 interfacial effects and the resistance of the electrode materials themselves, the battery E_{hc} represents the half-cell potential described above. [2]

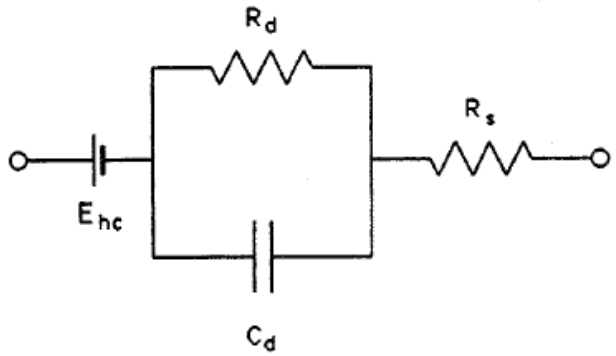

Figure 4.2: The equivalent circuit for a bio-potential electrode. [2]

It is seen that the impedance of this electrode will be frequency dependent, illustrated in *figure **4.3***. At low frequencies the impedance is dominated by the series combination of R_s and R_d, whereas at higher frequencies C_d bypasses the effect of R_d so that the impedance is now close to R_s, 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.

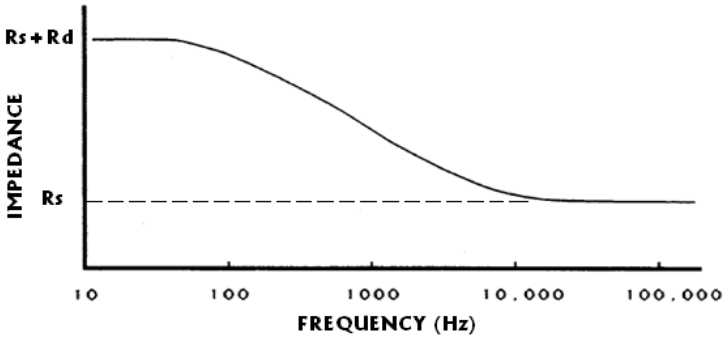

Figure 4.3: Bio-potential electrode impedance as a function of frequency. [2]

4.1.3 Body surface recording electrode:

Different types of electrodes developed over the years for recording various potentials on the body surface, this section describe the various type of these electrodes and there characteristics.

Metal-Plate Electrodes:

It is the simplest form, it consists of a metallic conductor in contact with the skin, an electrolyte soaked pad or gel is used to establish and maintain the contact, *figure **4.4*** illustrate the structure of metal-plate electrode, it consists of a flat metal plate that has been bent into a cylindrical segment, a terminal is placed on its outside surface near one end, this terminal is used to attach the lead wire to the electrocardiograph, the electrode is made of a nickel silver alloy, before it attach to the body its concave surface is covered with electrolyte gel. [2]

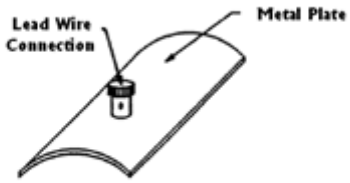

Figure 4.4: Metal-Plate Electrode. [2]

Suction Electrode:

Suction electrode is a modification of the metal plate electrode, it easier to attach this electrode to the skin to make a measurement and then move it to another point to repeat the measurement. These types of electrodes are used primarily for diagnostic recordings of bio-potentials such as the electrocardiogram. *Figure **4.5*** illustrate the construction of this electrode.

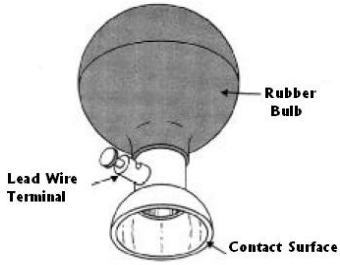

Figure 4.5: Suction Electrode [2]

They consist of a hallow metallic cylindrical electrode that makes contact with the skin at its base, a terminal for the lead wire is attached to the metal cylinder and a rubber suction bulb fits over its other base, electrolyte gel is placed over the contracting surface of the electrode, the bulb is squeezed, and the electrode is then placed on the chest wall, the bulb is released and applies suction against the skin, this electrode can be used for short periods of time, it cause irritation, although the electrode itself is quite large and the contacting area is relatively small, thus, the electrode impedance is relatively higher than the metal-plate electrode. [2]

Silver-Silver Chloride Electrode:

Silver-Silver Chloride Electrode (Ag-AgCl) electrode shown in *figure* ***4.6***. It is relatively stable in biological application. They are composed of a metal (Ag), coated with a salt of the metal (AgCl), this material (AgCl) is only very slightly soluble in water, so it remain stable, in addition some form of electrode paste or jelly is applied between the electrode and the skin, which the principal anions of this electrolyte is the Cl^-, for best result, the electrolyte solution should be saturated with AgCl so that there is little chance for the surface layer to dissolve. [2]

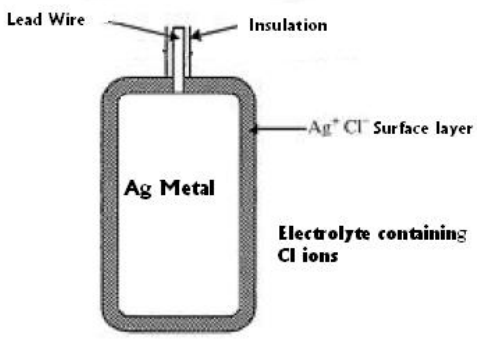

Figure 4.6: Ag-AgCl electrode shown in cross section. [2]

The behavior of the Ag-AgCl electrode is govern by two chemical reactions, the first *equation* ***4.4*** involves the oxidation of silver atoms on the electrode surface to silver ions in the solution at the interface:

$$Ag \rightleftharpoons Ag^+ + e^- \qquad (4.4)$$

$$Ag^+ + Cl^- \rightleftharpoons AgCl\downarrow \qquad (4.5)$$

The second reaction in *equation* ***4.5*** occurs immediately after the formation of Ag^+ ions, these ions combined with Cl ions already exist in solution to form the ionic compound AgCl. [2]

The half-cell potential of the Ag-AgCl can be determined by using the Nernst equation:

$$E = E^0_{Ag} + \frac{RT}{nF} \ln a_{Ag^+} \qquad (4.6)$$

Under equilibrium conditions the ionic activity of the Ag^+ and Cl^- ions must be such that their product is the solubility product. [2]

$$a_{Ag^+} \times a_{Cl^-} = K_s \qquad (4.7)$$

In biological fluids the concentration of Cl^- is relatively high which gives it an activity just a little less than unity, the solubility product for AgCl is of order 10^{-10}, so the activity of the Ag^+ is very low and of the same order of magnitude as the solubility product. [2]

$$E = E_{Ag}^{0} + \frac{RT}{nF} \ln a_{Ag^+} \qquad (4.8)$$

$$E = E_{Ag}^{0} + \frac{RT}{nF} \ln K_s - \frac{RT}{nF} \ln a_{cl^-} \qquad (4.9)$$

The first and second term on the right-hand side in *equation* ***4.9*** are constant, only the third term is determined by ionic activity of the Cl^-, which is relatively large and not related to the oxidation of Ag^+, the half cell potential of this electrode is consequently quit stable when it is placed in an electrolyte containing Cl^- as the principal anions, because this is the case of the body, Ag-AgCl electrode is relatively stable in biological applications. [2]

4.2 Traditional ECG Systems

4.2.1 Single Channel ECG System:

The traditional single channel ECG system use a multi-position switch to select the desired lead connection (I, II, III, aVR, aVL, aVF or V_1, V_2, V_3, V_4, V_5, and V_6) and apply it to a single bio-potential amplifier shown in *figure* ***4.7***, its consists of basic component (instrumentation amplifier , filter , protection circuit....), Only one ECG lead at a time could be selected and recorded with these machines via the ten electrodes mentioned in *chapter 3*. (left arm ,right arm , left leg and right leg as a fixed reference in addition to unipolar chest electrodes V_1-V_6).

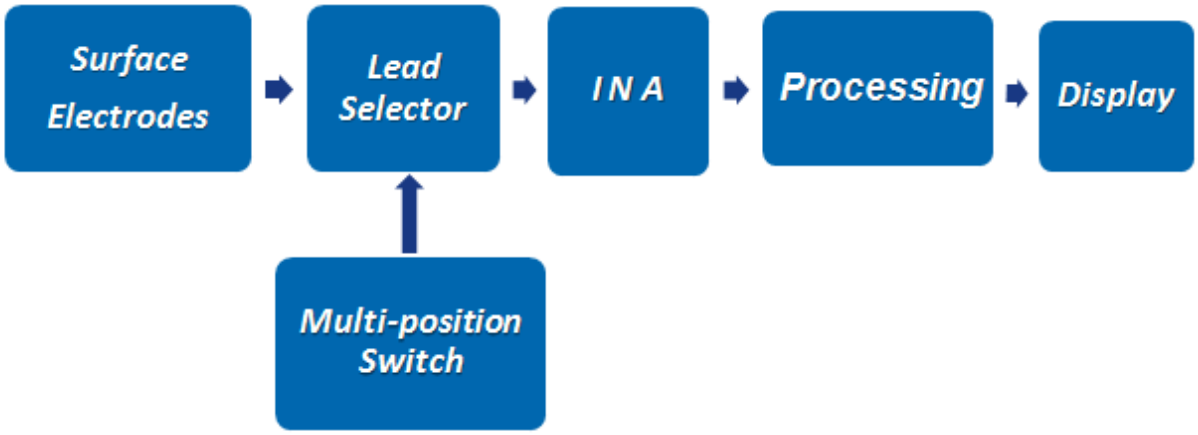

Figure 4.7: single channel ECG system block diagram.

The most obvious limitation of the system that excessive time is required to obtain the ECG signals. Hence, a multichannel ECG system is designed to provide single channel for each lead. All leads are measured, processed, and displayed simultaneously; *figure* ***4.8*** depicts an internal structure of the modern multichannel ECG systems.

4.2 Multichannel ECG System:

Multichannel ECG systems shown in *figure **4.8.*** It includes several amplifier channels and record several ECG leads simultaneously. This feature reduces the time required to complete a set of standard clinical ECG recordings. As the ECG leads are recorded simultaneously, they can be shown in their proper temporal relationship with respect to each other.

In single channel and multichannel ECG systems a fixed right leg electrode is used to attenuate the common mode voltage.

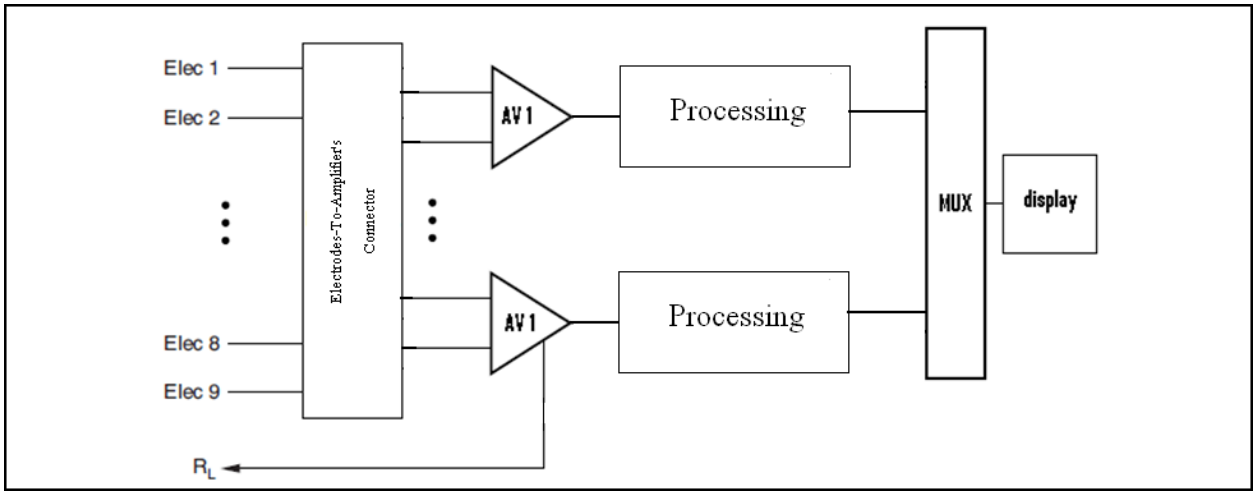

Figure 4.8: multichannel ECG system.

By describing these ECG's system, it is obvious these systems require large number of electrodes which makes the design more complicated, the goals is to measure and monitor ECG leads (I, II, III) simultaneously with minimum number of electrodes (3- electrodes RA,LA and LL). In each stage two electrodes are active and the remaining electrode acts as the reference, all these electrodes are internally switched via specific mechanism which will be described in *chapter 5*.

CHAPTER 5

SYSTEM DESIGN

5. 1 Block Diagram

5.2 Surface Electrodes

5.3 Multiplexer Circuit

5.4 Instrumentation Amplifier

5.5 Driven Right-Leg circuits

5.6 High Pass Filter

5.7 Gain Amplifier

5.8 Band-Reject Filter

5.9 Power Supply

5.10 Microcontroller

5.11 labVIEW

Introduction

In this chapter, the design process will be described through a block diagram, starting with surface electrodes, passing through all internal blocks until the three signal displayed. Traditional ECG system's have several drawbacks. This new design will provide a compromising method to reduce these drawbacks.

5.1 Block Diagram

The new ECG design defers from traditional ECG's design in that it measures and records three leads via single channel system using three electrodes simultaneously. It's consists of some common component that is used in most traditional ECG's system, such as pre-amplifier and filters, in addition this new design use the multiplexing technique, a special circuit multiplexes three electrodes (RA, LA and LL) over three terminals (positive and negative input of pre-amplifier, driven right leg output circuit).

From *figure* ***5.1***, First stage is a surface electrode which is used as a transducer which converts the ionic current into electrical current, a multiplexing circuit is used to multiplex the three electrodes over the three terminals. It's simply an electronic switch that act as a closed switch when activated, the multiplexing frequency is supplied and controlled through a microcontroller, the output signal from the multiplexing circuit is connected to a pre-amplifier (instrumentation amplifier), it meets certain basic requirement, the weak signal will be amplified by a small gain, a high pass filter is used to filter the signal from the DC offset voltage, after this filtration an amplifier magnify the filtered signal to a level that is suitable for next stages that includes a microcontroller to separate the samples and get the three signals to display it.

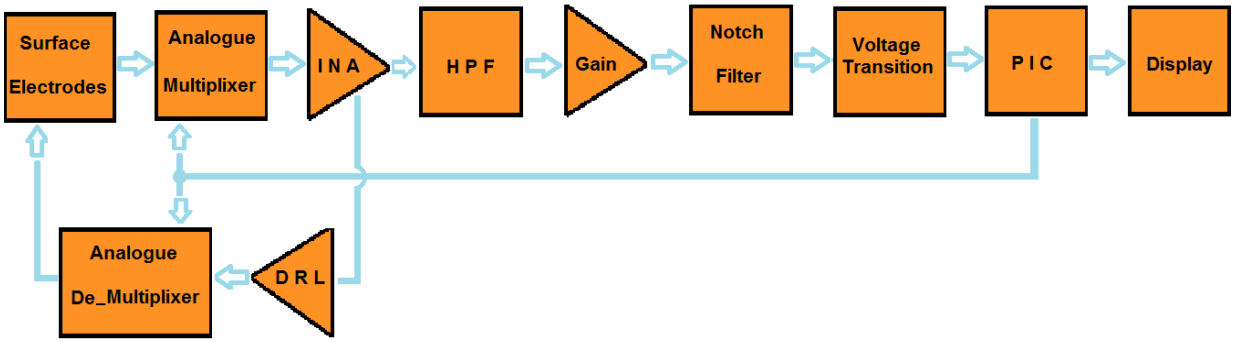

Figure 5.1: General Block Diagram.

The MUX and DE-MUX are controlled via microcontroller by providing the desired state at the desired frequency.

5.2 Surface Electrodes:

From previous study in chapter four, and according to the characteristics of each type of surface electrodes, the Ag-AgCl electrodes will be used according to their stability during biological applications.

5.3 Analog Multiplexer Circuit:

It is a device that selects one of several analog input signals, and forwards the selected input into a single line, multiplexer makes it possible for signals to share one device. Multiplexer purpose is to multiplex the three electrodes (RA, LA, and LL) over three terminals, two of them are connected to the positive and negative input of instrumentation amplifier, the third one is the right-leg drive circuit output, these technique sequentially samples each data signal and routs each signal to its proper destination. *Figure* ***5.2*** illustrate the functional block diagram of a four channel multiplexer, internally it consists of an electronic switches which activated by a synchronizing pulse from a two bit decoder (2x4), the input of the decoder determines which channel to be switched to a common output according to the following table.

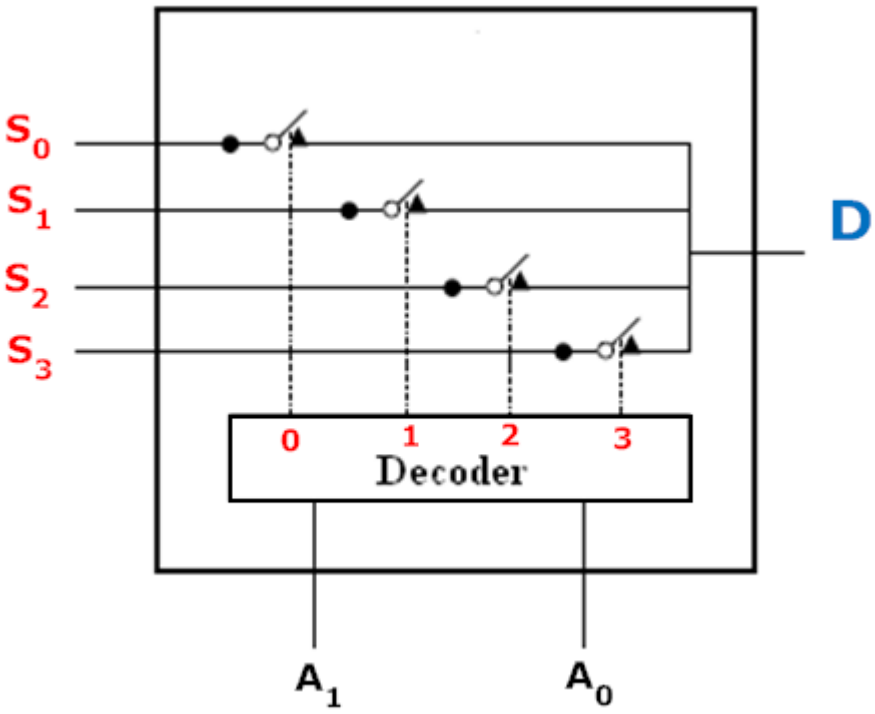

Figure 5.2: the functional Block diagram of a 4-channel multiplexer.[C]

Table 5.1: Analog Multiplexer Function table[C]

Inputs			Outputs D
$\bar{E}$	A_1	A_0	
1	X	X	None
0	0	0	S_0
0	0	1	S_1
0	1	0	S_2
0	1	1	S_3

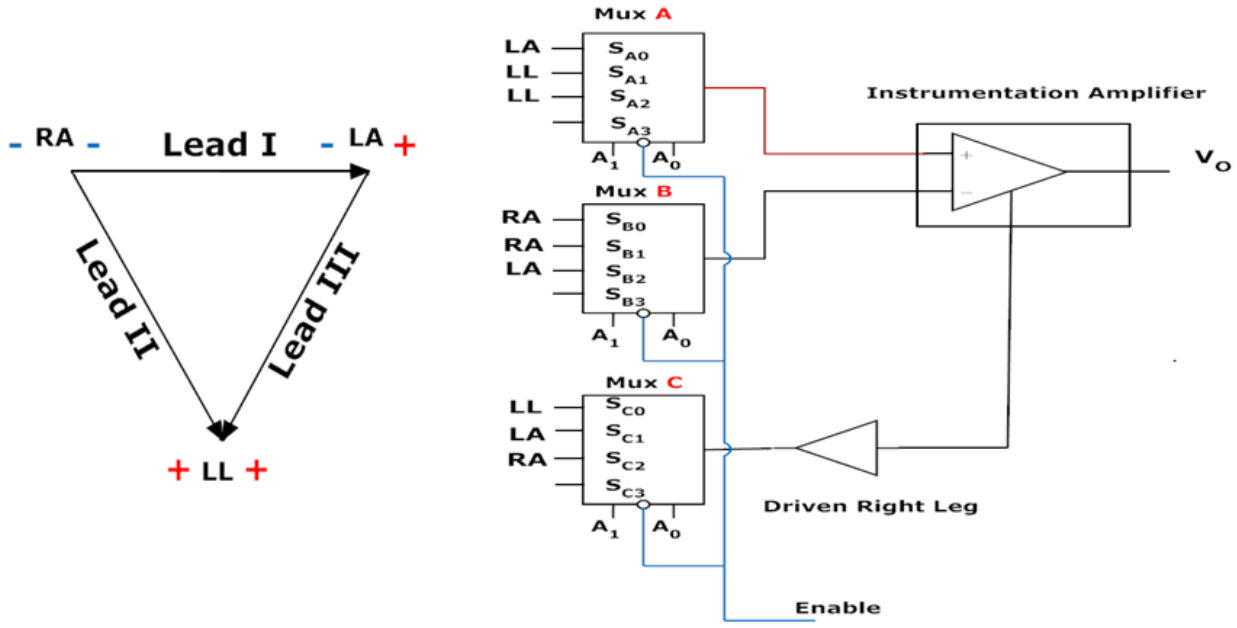

Figure 5.3: Three multiplexers connected to instrumentation amplifier.

Three multiplexers, four channel each, used in this designed illustrated in *figure **5.3*** , two of them used to routs the bio-signals from the electrodes to instrumentation amplifiers input, the third one to connect the remaining electrode to right-leg drive circuit output for attenuating common mode voltages .These process fully synchronized to get the three leads (I, II and III).

Initially when the input binary data A_0 and A_1 of the decoder is **00** respectively, channel one will be activated of each multiplexer, hence, LA is connected to positive input of instrumentation amplifier, RA is connected to negative input and LL connected to right-leg drive circuit. Truth table below describes the operation of the circuit in *figure **5.3.***

Table 5.2: Truth table

Inputs			Switch condition			
$\bar{E}$	A_1	A_0	MUX A	MUX B	MUX C	Pre-amp O/P
1	X	X	None	None	None	----
0	0	0	LA	RA	LL	$V_{LA} - V_{RA} = V_I$
0	0	1	LL	RA	Sc1	$V_{LL} - V_{RA} = V_{II}$
0	1	0	LL	LA	Sc2	$V_{LL} - V_{LA} = V_{III}$
0	1	1	-	-	-	-

From the above table, three states of four are used, the forth channel is not included, these binary states (00-10) are provided from a microcontroller counter. The counting frequency related to the sampling rate, which according to Nequist equation, each signal must be sampled at the minimum Nequist rate for its particular band width, relatively, leads signals (I, II and III) has the same band width, which is approximately (0.05 - 100) Hz, thus, the minimum sampling rate is 300 samples per seconds for each lead.

From the sampling theory, an arbitrary analog signal with a spectrum extends from dc to W Hz must be uniform sampled at a rate at least twice the highest frequency (minimum Nequist rate) in order to be recoverable by direct low-pass filtering and to prevent signal overlapping. [4]

$$f_s \geq 2\,W \qquad (5.1)$$

If $f_s = 2W$, there is no gap in the spectrum between the original spectrum and the first translated component, and a perfect block filter will required to separate the component which is does not exist. Thus, some frequency interval, called a guard band should to be provided in order that the first translated component can be rejected by a realistic filter. [4]

5.4 Instrumentation Amplifier:

ECG signal is recorded as voltages, the measurement involves voltage at very low level, with high source impedance that is superimposed high level interference signals and noise, the ECG signal must be amplified to make it compatible with subsequent stages such as filters and display, instrumentation amplifier satisfy very specific requirement, they have to provide amplification selective to the ECG signal, and reject superimposed noise and interference signal.

Instrumentation amplifier must meet certain basic requirements to acquire biosignal such as ECG, it must has high input impedance, so to provide minimal loading of the signal being measured, the input circuit of instrumentation amplifier must also provide protection to the organism being studied, any current or potential appearing across the amplifier input terminals that is produced by the amplifier is capable of affecting the ECG potential being measured, these

current from the input terminals can result in micro shock and macro shock in the patient being studied.

Instrumentation amplifier must operate in that portion of the frequency spectrum in which the ECG signals exist. The bandwidth of the amplifier must be great enough to process the signal adequately; also because the ECG signals usually have amplitudes of the order of a few millivolts or less, these signals must be amplified to levels compatible with the subsequent stages, this means that the instrumentation amplifier must have high gain of the orders of 1000 or greater. *Figure **5.4***, illustrate the block diagram of a three op-amp instrumentation amplifier, the input V_{IN+} and V_{IN-} are define through the input polarities of the difference amplifier A_3 , the instrumentation amplifier input signal consists of four components :

1. The desired bio-potential.

2. The undesired bio-potential.

3. Power line interference signal of 50Hz and its harmonics.

4. Interference signal generated by the tissue- electrodes interface.

Proper design of the instrumentation amplifier provides rejection of a large portion of the signal interference, the main task of instrumentation amplifier is to reject the line frequency interference that is electro statically or magnetically coupled into patient.

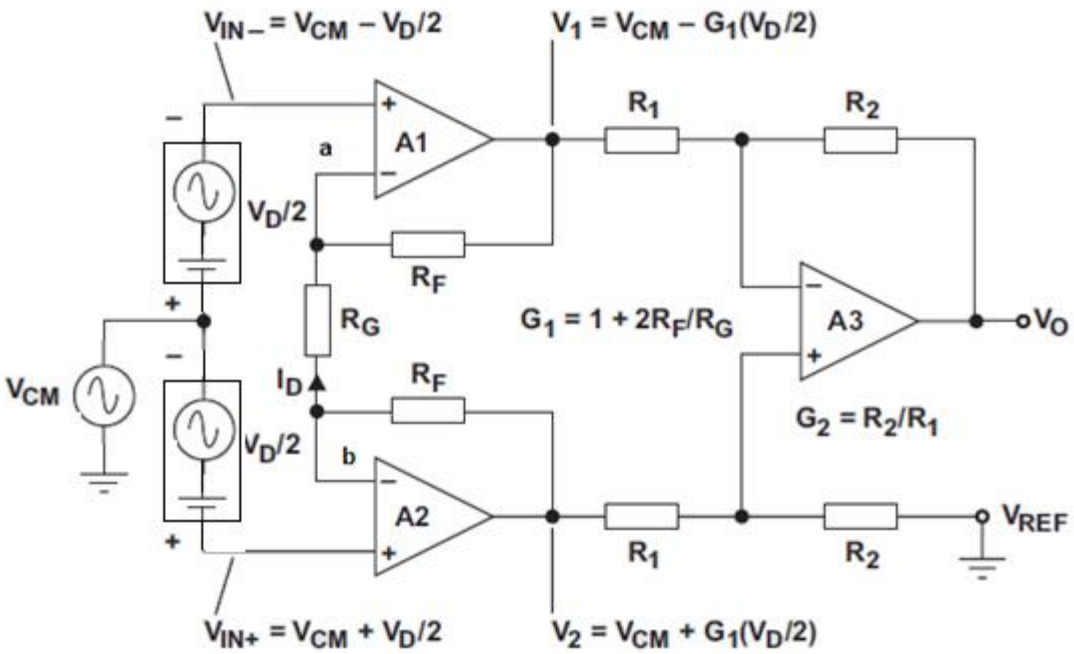

Figure 5.4: Classic Three-op-amp instrumentation amplifier and its voltage nodes

The line frequency interference signal shows only (common mode signal) very small difference in amplitude and phase between the two measuring electrodes, causing approximately the same potential at both input, and thus appears only between the inputs and ground, thus,

common mode voltage V_{CM} is common to both inputs, which is defined as the average of the sum of V_{IN}^{+} and V_{IN}^{-}.

$$V_{CM} = \frac{V_{IN+} + V_{IN-}}{2} \qquad (5.2)$$

While differential voltage which defined as the difference between the positive input and negative input

$$V_D = V_{IN}^{+} - V_{IN}^{-} \qquad (5.3)$$

Solving both equation for V_{IN}^{+} or V_{IN}^{-} and equating the received terms results in a new set of equations, which when solved for either input voltage yields

$$V_{IN}^{+} = V_{CM} + \frac{V_D}{2} \quad \text{and} \quad V_{IN}^{-} = V_{CM} - \frac{V_D}{2} \qquad (5.4)$$

In the non-saturated mode, the output the op-amp A_1 & A_2:

Apply KCL on node a:

$$V_1 = \frac{R_F}{R_G}[V_a - V_b] + V_a \qquad (5.5)$$

Apply KCL on node b:

$$V_2 = \frac{Rf}{Ra}[V_b - V_a] + V_b \qquad (5.6)$$

By using golden rules $V_a = V_{IN}^{-}$ and $V_b = V_{IN}^{+}$

$$V_1 = \frac{R_F}{R_G}(V_{IN-} - V_{IN+}) + V_{IN-} \qquad (5.7)$$

$$V_2 = \frac{R_F}{R_G}(V_{IN+} - V_{IN-}) + V_{IN+} \qquad (5.8)$$

$$V_1 = -G_1 \frac{V_D}{2} + V_{CM} \qquad (5.9)$$

$$V_2 = G_1 \frac{V_D}{2} + V_{CM} \qquad (5.10)$$

Where $G_1 = 1 + 2(\frac{R_F}{R_G})$

*Equations **5.9*** and ***5.10*** show that only the difference component $\frac{V_D}{2}$ is amplified by the input gain, while the common mode voltage V_{CM} passes the input stage with unity gain. The difference amplifier, A_3 subtracts V_1 from V_2 and amplifies the difference with the gain G_2:

$$V_o = [V_2 - V_1] G_2 \qquad (5.11)$$

Where $G_2 = \frac{R2}{R1}$

$$V_o = G_1 G_2 V_D \qquad (5.12)$$

5.5 Driven Right-Leg Circuit:

In this design a special circuit called Driven Right-Leg circuit is shown in *figure* ***5.5*** 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 ($-V_{CM}$). In this new design the circuit is the same but the location of the electrode changed, the output of Driven Right-leg circuit is connect to De-Multiplexer common inputs in order to be connected to it's proper extremities (RA, LA or LL) at proper time.

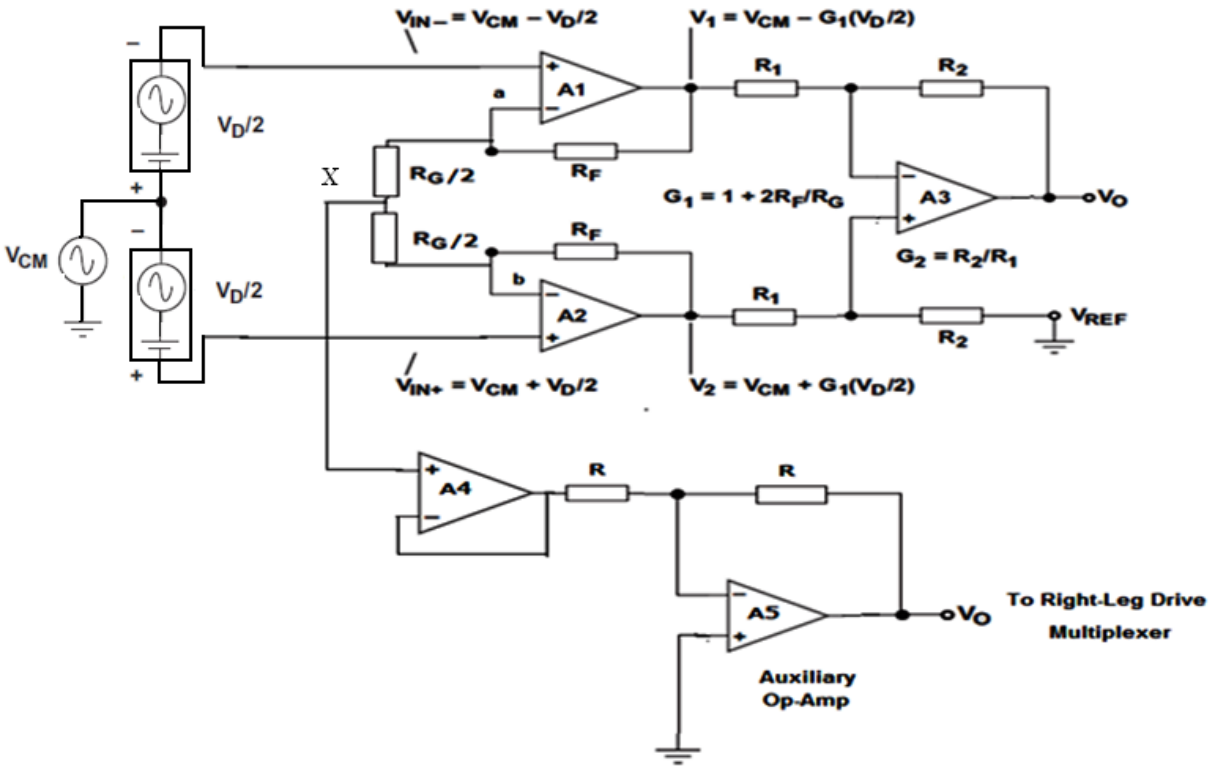

Figure 5.5: Driven Right-Leg Circuit for attenuating common-mode voltages.

The input impedance of inverting amplifier which represented by resistor **R** cause a loading effect on $\mathbf{R_G}$**,** which draw current from node **X,** these resistor will alter and change the total RG. Thus no current should be flow from node **X** through the resistor R, hence a buffer(voltage follower) is used to block the current.

$$V_{IN\ (Buffer\ )} = \frac{V_{IN+}+V_{IN-}}{2} = V_{CM} \qquad (5.13)$$

Driven Right-Leg Circuit O/P

$$V_O = -\frac{\mathrm{R}}{\mathrm{R}}\, V_{CM} = -V_{CM} \qquad (5.14)$$

The gain of the first stage of amplification (Instrumentation Amplifier) limited by the input signals of the instrumentation amplifier which as mentioned before in this section, subdivided into a common-mode voltage, V_{CM}, of up to 1.5V which comprises of 50Hz interference, and DC electrode offset potential of ± 500mV in addition to a small AC component of deferential voltage, V_D, typically is a 0.05 to 1.5 mV AC signal represent the ECG signal. *Figure* ***5.6*** illustrate instrumentation amplifier input signal component. [A]

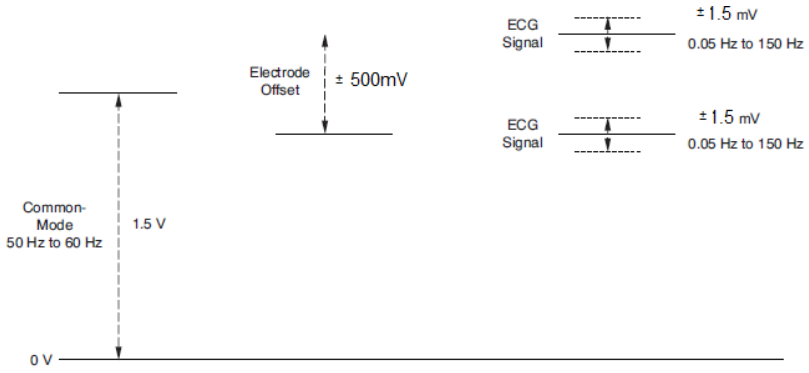

Figure5.6: Instrumentation amplifier input signal component.

The instrumentation amplifier is supplied by ±9V supply, to avoid saturation the output signal should not exceed ±7 V, at worse case the differential DC is ±500 mV and the differential AC is 1.5 mV

$$V_D = V_{D\text{-}DC} + V_{D\text{-}AC} \qquad (5.15)$$

$$V_D = 500\ mV_{DC} + 1.5\ mV_{AC} \qquad (5.16)$$

Thus the differential DC is 300 times larger than the AC signal of interest, and if untreated, instrumentation amplifier will reach saturation.

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. The solution for the above problem is performed in four steps:

1. Limit the gain of the instrumentation amplifier to avoid saturation.
2. Implement a high pass filter in the next stage to remove the DC offset.
3. Apply a high gain, boosting the AC signal of interest, $V_{D\text{-}AC}$.
4. Use the D.R.L circuit to attenuate the common mode voltage.

In the first step the gain of the instrumentation amplifier is limited to 8.5, thus, the maximum output voltage is 4.25 V DC + 12.8 mV AC, for instrumentation amplifier *ADC620* the gain is determined by *equation **5.17*** [B]

$$G = \frac{49.4\ K\Omega}{R_G} + 1 \qquad (5.17)$$

To determine R_G

$$R_G = \frac{49.4\ K\Omega}{8.5-1} = 6.6\ K\Omega \qquad (5.18)$$

5.6 High Pass Filter:

High pass filter is a device that passes high frequencies and attenuates (reduce the amplitude) frequencies lower than cut off frequency. Because the main objective of the high pass filter is to remove the DC offset and get the ECG signal at the precise level across the entire passband, a Sallen-Key, Butterworth high pass filter is used, *figure **5.7*** shows a unity Gain Sallen-Key high pass filter.

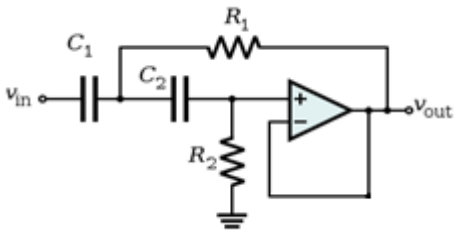

Figure 5.7: Unity-Gain Sallen-Key High Pass Filter.

The general transfer function of a high pass filter is:

$$A(s) = \frac{A}{(1+\frac{a}{s}+\frac{b}{s^2})} \qquad (5.19)$$

With **A** being the passband gain.

The cut off frequency F_L of the filter is set at 0.05 Hz and can be calculated using R_1, R_2, C_1 and C_2 as expressed in *equation **5.20***

$$f_L = \frac{1}{2\pi\sqrt{R_1 R_2 C_1 C_2}} \qquad (5.20)$$

Let $C_1 = C_2 = C$, The transfer function of the circuit shown in *figure **5.7*** is:

$$A(s) = \frac{1}{1+\frac{2}{\omega_c R_1 C}\frac{1}{s}+\frac{1}{\omega_c^2 R_1 R_2 C^2}\frac{1}{s^2}} \quad (5.21)$$

The coefficient comparison between this transfer function and *equation **5.19*** yields:

A=1

$$a = \frac{2}{\omega_c R_1 C}$$

$$b = \frac{1}{\omega_c^2 R_1 R_2 C^2}$$

For Butterworth high pass filter, a = 1.4142, b = 1. [5]

Let C = 4.7µF the resistor values for R_1 and R_2 are:

$$R_1 = \frac{1}{\pi\ f_L\ C\ a} = 957.8\ K\Omega \qquad R_2 = \frac{a}{4\pi\ f_L\ C\ b} = 478.8\ K\Omega$$

5.7Gain Amplifier's:

After removing the DC offset voltage which comes from electrodes by the high pass filter, AC component of V_D must be amplified by inverting amplifier to be suitable for the next stages before sampling.

After amplification we also need to shift the signal to the upper side so that V_D component swing between the lower reference voltage (0 V) and the upper reference voltage (5 V) as the microcontroller handles signals between (0 V) and (5 V) . Hence, the signal should be shifted to the upper by (2.5V) making the maximum voltage equal (5 V) and the minimum is (0 V), by dividing (0V-5V) area, voltages occur between (0V-2.5V) represent the negative component of V_D while voltages occur between (2.5V-5V) represent the positive component of V_D .

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

$$G_T = G_{INA} * G_{IA} \qquad (5.22)$$

To amplify the ± 1.5mV which represents the AC component of V_D to ± 2.5V , a total required gain G_T is :

$$G_T = \frac{2.5V}{1.5mV} = 1666.6$$

Where $G_T = G_{INA} * G_{IA}$.

As the gain of the instrumentation amplifier is 8.5, hence the gain of the inverting Amplifier is:

$$G_{IA} = \frac{G_T}{G_{INA}} = 196$$

An inverting amplifier is used to provide this gain which is determined by equation 5.22, *figure* 5.8 show the inverting amplifier circuit.

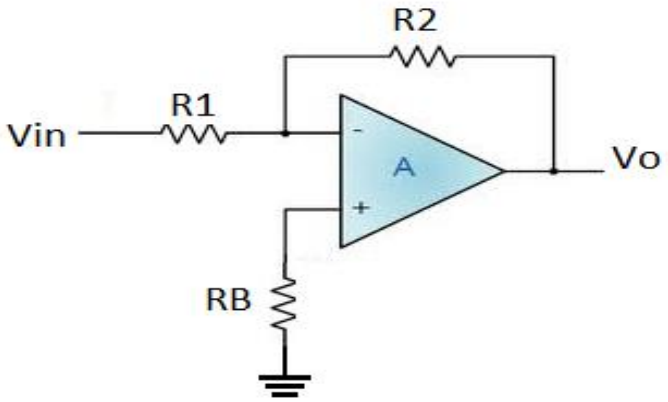

Figure 5.8: Inverting Amplifier Circuit.

The gain of the inverting amplifier is given by the following:

$$G_{IA} = \frac{-R2}{R1} \qquad (5.23)$$

By referring to standard resistor values, let R_1=1KΩ and R_2=100 KΩ, according to these values, the gain of the circuit shown in *figure* 5.8 is 100.

Another inverting amplifier with a unity gain is used to add an offset voltage to the ECG signal so that it swings between lower reference voltage (0V) and upper reference voltage (5V). *Figure* 5.9 shows the second inverting amplifier.

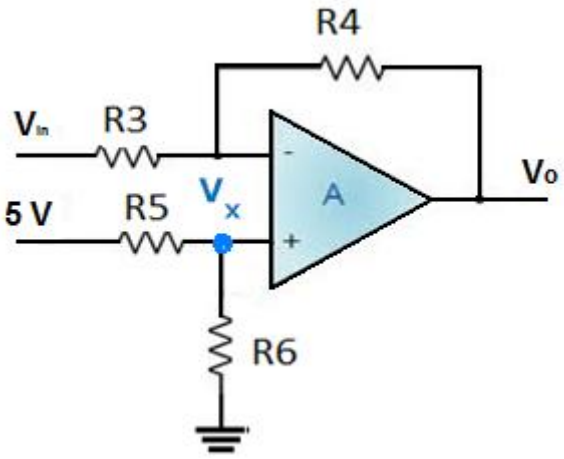

Figure 5.9: Single supply Circuit

The formula for circuit shown in *figure* 5.9 shown in equation 5.24:

$$V_o = -V_{IN} \frac{R4}{R3} + \left(1 + \frac{R4}{R3}\right) V_x \qquad (5.24)$$

The second term of equation 5.24 adds an offset voltage to the input signal. As we mentioned before we need a gain of 196 after the instrumentation amplifier, we design an inverting amplifier with a gain of 100, so the remaining gain (G_R) is:

$$196 = G_{IA} * G_R$$

$$G_R = \frac{196}{G_{IA}} = 1.96$$

This gain is provided by the next stage (band reject filter) to provide a rejection quality of 10.

From *figure* 5.9 the Remaining gain G_R is:

$$G_R = \frac{R4}{R3}$$

Let $R_3 = R_4 = 33.3k\Omega$.

This circuit also adds an offset voltage to the input signal so the output of this circuit is only positive and swing between 0V and 5V. V_x determine the offset voltage which taken from a voltage divider circuit composed of R_5 and R_6:

$$V_X = 5v * \frac{R_6}{R_5 + R_6} \qquad (5.25)$$

To add a 2.5V as an offset voltage to the input signal and by referring to equation 5.28:

$$2 * V_x = 2.5V \quad \rightarrow \quad V_x = 1.25V$$

$$V_o = -V_{IN} * 1 + 2.5 \qquad (5.26)$$

From equation 5.25:

$$1.25\ V = 5v * \frac{R6}{R6 + R5} \qquad and \qquad \frac{R5}{R6} = 3$$

From standard resistor values:

Let $R_5 = 6.8K\ \Omega$, $R_6 = 1.38\ K\Omega$

In this case R_6 composed from 1K Ω and 1k Ω potentiometer.

5.8 Band-Reject Filter:

A band reject filter is used to suppress a certain frequency rather than a range of frequencies. *figure* 5.10 shows an active twin-T second order band reject filter.

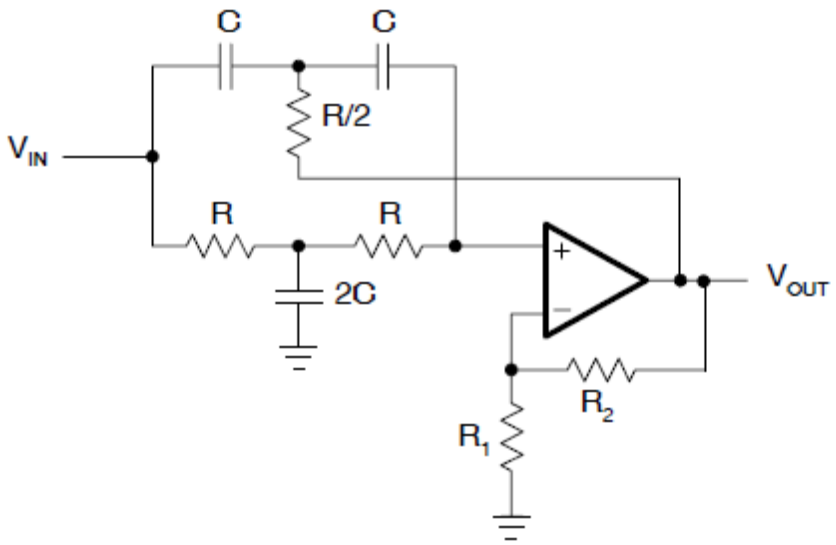

Figure 5.10: Active twin-T filter.

The general transfer function of a band reject filter is:

$$A(s) = \left(\frac{A_0(1+S^2)}{1+\frac{1}{Q}S+S^1}\right) \quad [5] \qquad (5.31)$$

Where A_0 is the passband gain, Q is the quality factor. The transfer functions of the active twin-T filter shown in *figure* 5.10 is:

$$A(s) = \left(\frac{K(1+S^2)}{1+2(2-K)\,S+S^2}\right) [5] \qquad (5.32)$$

Comparing the variables of equation 5.32 with equation 5.31 provides the quantities that determine the filter parameters:

Mid-frequency: $f_m = \frac{1}{2\pi RC}$ [5] (5.33)

Passband gain: $G = 1 + \frac{R_2}{R_1}$ (5.34)

Rejection Quality: $Q = \frac{1}{2(2-G)}$ (5.35)

The twin-T circuit has the advantage that the quality factor (Q) can be varied via the gain (G) without modifying the mid frequency (f_m).

To set the mid frequency of the band-pass, specify fm which is 50 H_Z and C, and then solve for R:

$$R = \frac{1}{2\pi f_m C}$$

According to standard values of resistors and capacitors:

Let C = 1μ F and solving for R, R = 3.3KΩ.

To design this filter with minimum rejection quality of 5, we use equation 5.37. [5]

$$R_2 = R_1(1 - \frac{1}{2Q}) \quad [5] \qquad (5.37)$$

Let R_1 = 1KΩ and Q = 5, → R_2 = 1.1 KΩ.

5.9 Power Supply:

The device needs power supply to power up the entire hardware, so need a battery that has the following characteristics:

1. Lightweight.
2. Enough supply voltage.
3. Enough supply current.

From previous characteristics and by calculating the total current for the system which is 30 mA, the 9V battery would be enough to power up the entire hardware, as shown in *figure 5.11*.

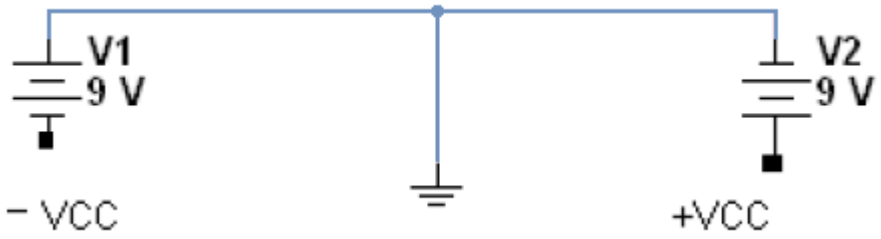

Figure 5.11: Power Supply Circuit.

5.10 Microcontroller:

A microcontroller (also microcomputer, MCU or µC) 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, simplicity is in emphasized.

In this project the main function of microcontroller is to provide the control signals to the multiplexers and synchronize all operation including the display, this synchronization represented by providing the counts to the multiplexer selection lines (A_0 and A_1) at the desired frequency via build in counter and split the three signals for displaying purpose, *figure* 5.12 describe the algorithm of the program which installed on the PIC18F4550.

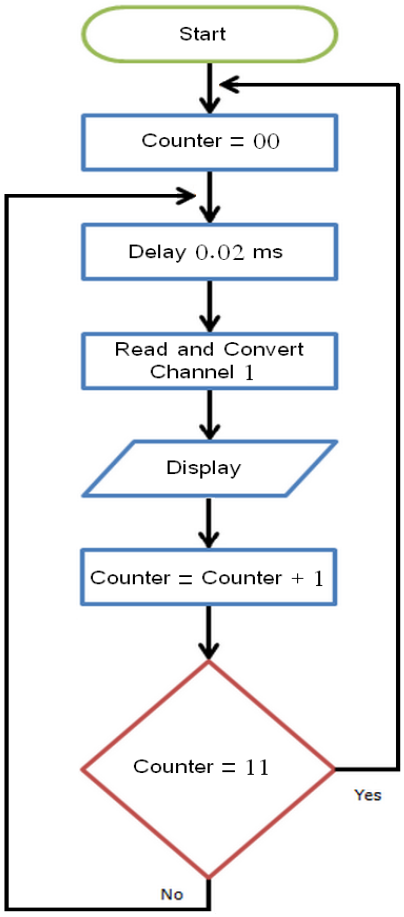

Figure 5.12: PIC18F4550 algorithm.

Microcontroller Architecture Contents:

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

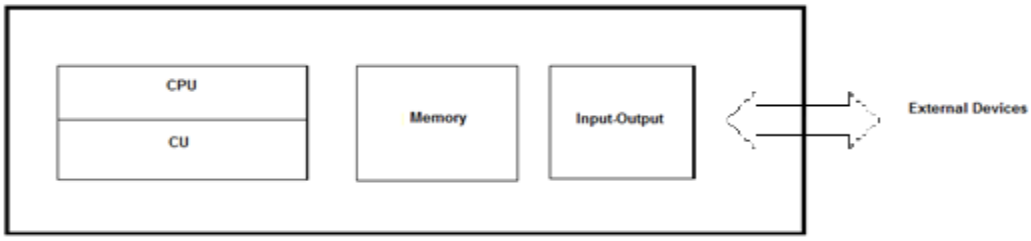

Figure 5.13: The simplest microcontroller architecture.

Hardware feature of microcontroller:

1. **Supply Voltage:**

   Microcontrollers 18F4550 operate with the standard logic voltage of 5V.

2. **Clock:**

   The clock is provided by connecting external crystal oscillator to the microcontroller. Microcontrollers 18F4550 have build-in timing circuit and they do not require any external timing components.

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).

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 13 inputs; this module allows conversion of an analog input signal to a corresponding 10-bit digital number.

5.11 LabVIEW:

LabVIEW (short for **Lab**oratory **V**irtual **I**nstrumentation **E**ngineering **W**orkbench) is a platform and development environment for a visual programming language from National Instruments. The graphical language is named "G". LabVIEW is commonly used for data acquisition, instrument control, and industrial automation on a variety of platforms including Microsoft Windows.

The function of the LabVIEW in this project is to display the three leads, the digital data from the PIC which corresponds to the leads samples provided to the DAC PORT, and according to the counter value which connected to two digital lines in the DAC, LabVIEW split the samples and display it individually. *figure* 5.14 describes the algorithm to display the three leads.

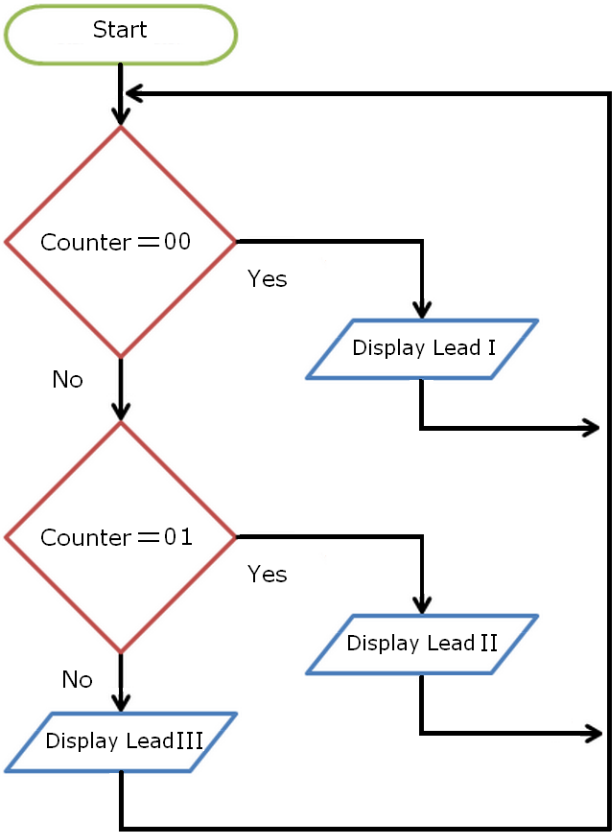

Figure 5.14: LabVIEW display algorithm.

CHAPTER

6

SYSTEM IMPLEMENTATION & RESULTS

Introduction:

Practical implementation of the project has been done in the second semester, 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.

6.1 Electrode Position:

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

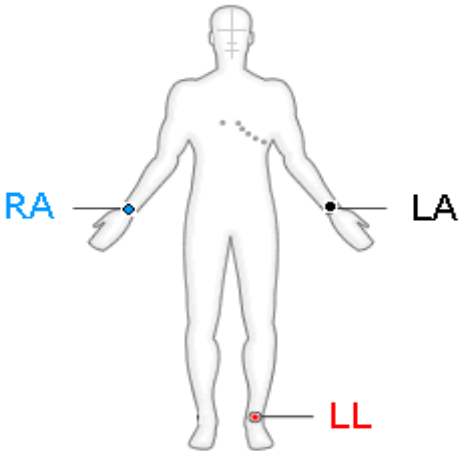

Figure 6.1: Electrodes position.

6.2 Analogue Multiplexer/De-multiplexer:

The HEF4052B is a dual 4-channel analogue multiplexer/de-multiplexer which used to change the connection of the electrodes with the input of the instrumentation amplifier (AD620) internally. *Figure 6.2* below show the 4-channel analogue multiplexer/de-multiplexer.

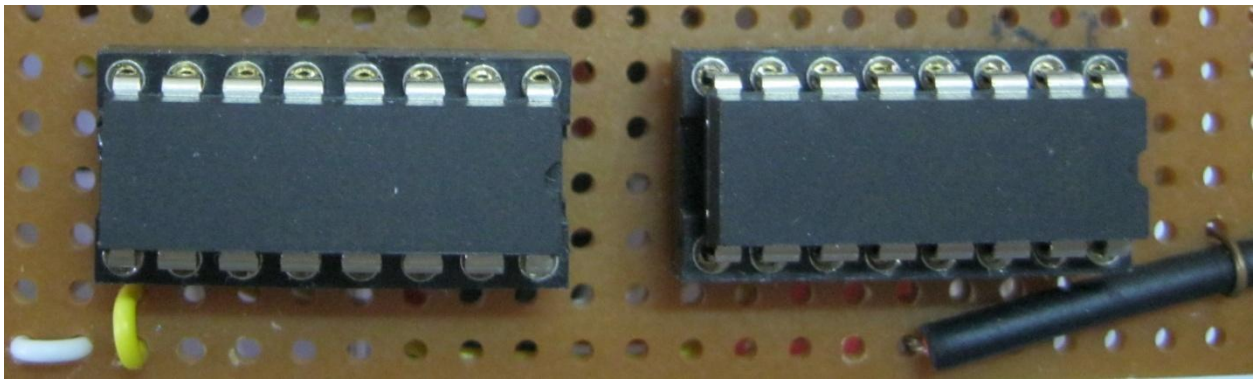

Figure 6.2: Four Channel analog multiplexer/de-multiplexer.

6.3 Instrumentation Amplifier:

Instrumentation Amplifier AD620 used to provide amplification selective to the ECG signal and reject super imposed noise and interference signal, the output of the AD620 which shown in *figure 6.3* is the difference voltage of the two electrodes.

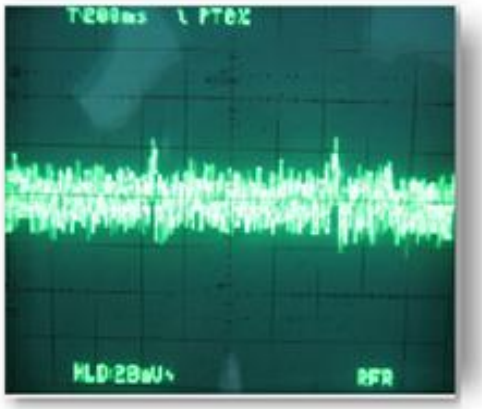

Figure 6.3: a. output of the INA for lead I

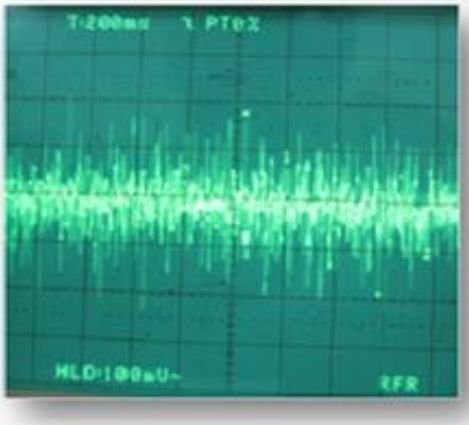

Figure 6.3: b. output of the INA for lead II

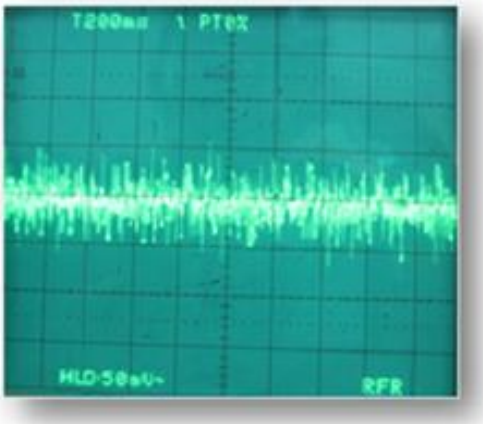

Figure 6.3: c. output of the INA for lead III

6.4 High Pass Filter:

High pass filter which shown in *figure 5.7* 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 6.5.*

Figure 6.5: a. output of the HPF for lead I.

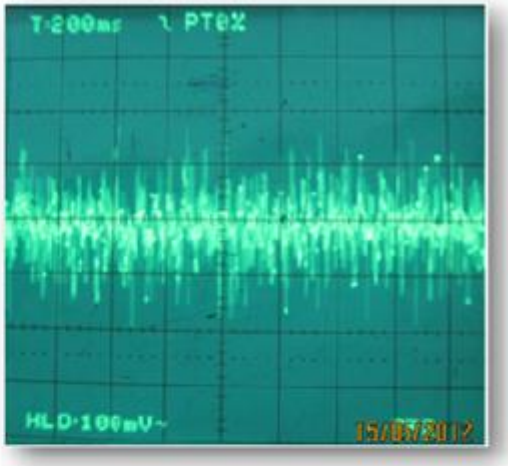

Figure 6.5: b. output of the HPF for lead II.

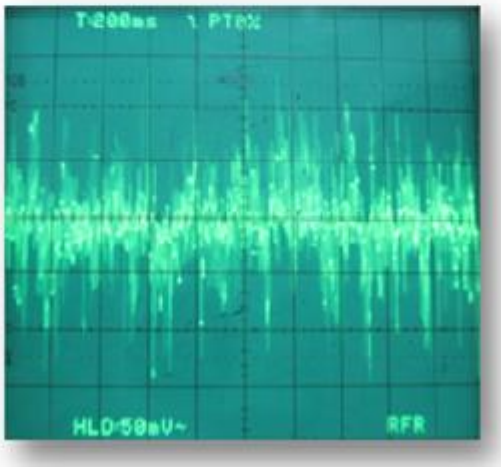

Figure 6.5: c. output of the HPF for lead III.

6.5 Inverting Amplifier:

Inverting amplifier shown in *figure 5.8* built by using TL082CN operational Amplifier, the output of the inverting amplifier is shown in *figure 6.6*

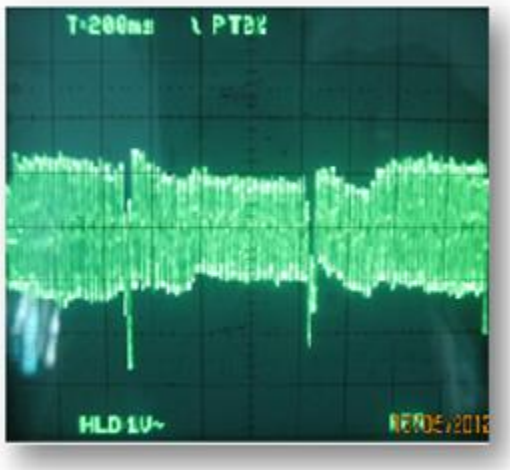

Figure 6.6: a. output of the IA for lead I.

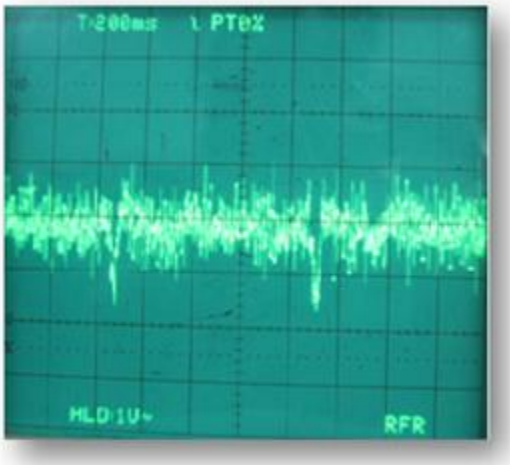

Figure 6.6: b. output of the IA for lead II.

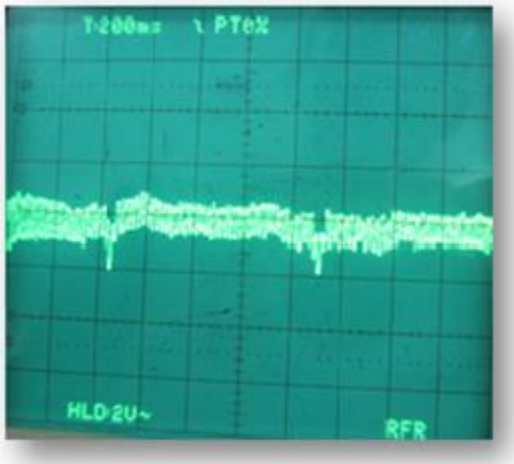

Figure 6.6: c. output of the IA for lead III.

6.6 Band-Reject Filter (BRF):

Band-Reject Filter shown in *figure 5.10* is built using TL082CN operational Amplifier. The output of the Band-Reject Filter is shown in *figure 6.7*.

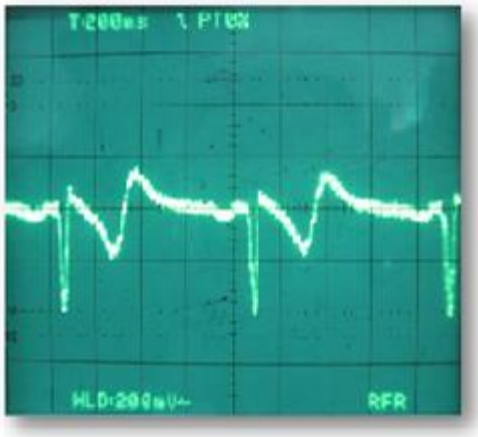

Figure 6.7: a. output of the B R F for lead I.

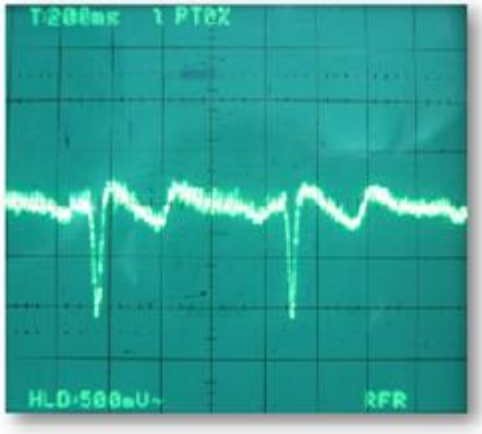

Figure 6.7: b. output of the B R F for lead II.

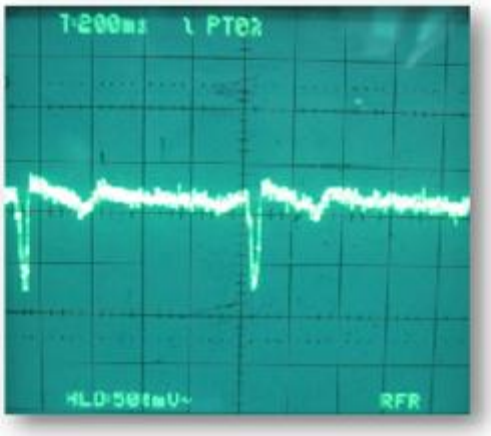

Figure 6.7: c. output of the B R F for lead III.

6.7 Voltage Translation Amplifier (VTA):

In order to shift the ECG signal by (2.5V), the circuit shown in *figure 5.9* is implemented using the TL082CN Operational Amplifier, the output of this stage described in *figure 6.8.*

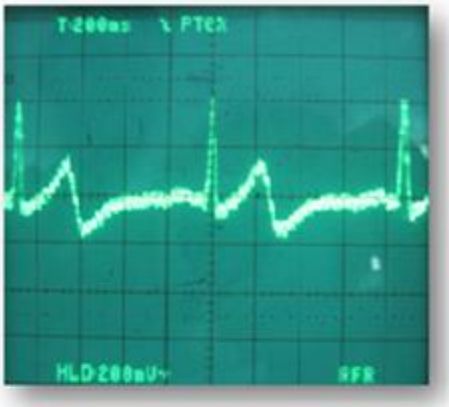

Figure 6.8: a. output of the VTA for lead I.

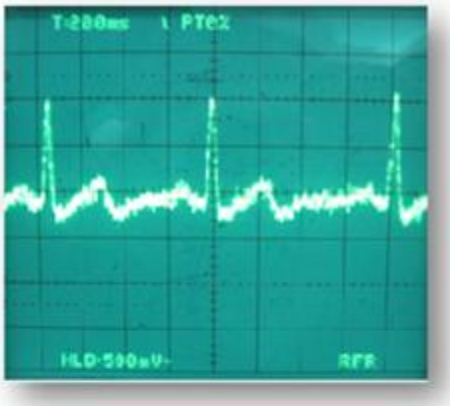

Figure 6.8: b. output of the VTA for lead II.

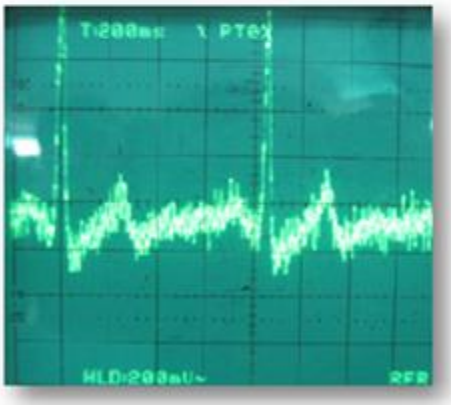

Figure 6.8: c. output of the VTA for lead III.

CHAPTER

7

Future Work and Conclusion

7.1 Recommendation

7.2 Challenges

7.3 Conclusion

7.1 Recommendations:

In this project, the system has been designed to acquire and record three leads via three electrodes simultaneously, but these leads not enough to give the specialist with a total diagnostic, so additional three leads (aV_R, aV_L and aV_F) could be included in this system .

The three leads where displayed on the PC, could be 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 fail detector circuit with conductivity indicator could be provided to this system.

7.2 challenges:

While designing the system, there are many challenge were faced, such as:

- Not all the required component for 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.
- The difficult to find the appropriate MUX with appropriate characteristics.
- 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 leads via three electrodes simultaneously, after designing, implementing and testing this system it is possible to:

- Multiplex the three electrodes over two input of the AD620 to get the three leads.
- Connect the weak Leads voltages to the instrumentation amplifier via the analogue multiplexer.
- Record and display the three leads (I, II and III) via three electrodes simultaneously.
- Attenuate the common mode voltages using driven right leg circuit connected to different locations on the body (RA, LA and LL).

By using this technique, the portability of the ECG system is strongly supplied, in addition the number of components is reduced in conjunction with the reduction of the total cost.

References

a. [1] J.G. Webster, *Encyclopedia of medical devices and instrumentation , Second Edition*, Houghton Mifflin Company, Boston, MA.

b. [2] J.G. Webster, *Medical Instrumentation, Application and Design, Second Edition*. Houghton Mifflin Company, Boston, MA. 1992.

c. [3] J.Najel , Biopotintial amplifiers , in *The Biomedical Engineering Handbook: Second Edition,* Ed. Joseph D. Bronzino, Boca Raton: CRC Press LLC, 2000.

d. [4] D.Prutchi, M.Norris , *Design and Development of Medical Electronic Instrumentation*, A john Wiley & Sons, INC, United State of America ,2004 .

e. [5] op-am for everyone, Ron Mancifini .second edition , Texas Instruments Incorporated,2009