Design of Hardware for an Electrocardiogram Analyzer

by

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Abstract

In these modern times there is a shortage of doctors in most countries around the globe with Canada having one of the worst shortages in relation to other developed Countries. This increases patient wait times, adds stress to doctors and lowers the quality of care. The ECG Analyzer is a device that integrates an ECG acquisition system with software to correctly classify and identify unhealthy ECG signals. By performing these tedious tasks for the doctor it will elevate some of the work load they have and allow them to see more patients. Design considerations that need to be taken into account when designing the hardware for ECG Analyzer are discussed including special requirements for making this a low cost system for use in third world countries. The hardware of the ECG analyzer involves three silver chloride electrodes attached to both wrists and right ankle of the patient from which the electrical activity of the heart is acquired, amplified, filtered and sent to my partner's software for processing. The main thrust of this project is on the design of the hardware used to condition the signal for use with the software.

ACKNOWLEDGMENTS

I would first like to thank my partner for being the second half of team Awesome and working through this project with me to successfully create an ECG Analyzer. I would like to next thank Dr. Doyle my advisor who provided many insights on how to approach my project and answers to any of the technical questions I had. I would also like to thank TIS technologies for providing me with many of the components I used in this project. There are also thanks in order to ECE technical staff and Dr. Patriciu who both allowed me to borrow various components for my project. Lastly I would like to thank my family and who kept me motivated and helped me deal with the stress of the project when things did not work as planned.

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Nomenclature

ECG- Electrocardiogram

SVM- Support vector machine

Op-amp-Operational Amplifier

WCT- Wilson's center terminal

Ag/Cl- Silver chloride

Chapter 1

Introduction

1.1 Background

Most processes in the human body have some sort of bioelectricity associated with them and the heart is no different. Each time the heart beats it produces electrical currents. These currents are responsible for the rate and pattern of contraction of the heart. The electrocardiogram is a transthoracic interpretation of these currents over time captured and externally recorded by skin electrodes in a non-invasive procedure (6).

The typical electrocardiogram can be seen in figure 1.1. It contains 5 main components: the P, Q, R, S, and T wave. The P wave is responsible for depolarization of the left and right atrium (6). The QRS complex is composed of the Q, R and S waves and represents left and right ventricular depolarization (6). At this same time, the QRS complex masks the P wave repolarization (6). The T wave is responsible for the repolarization of the left and right ventricles (6).

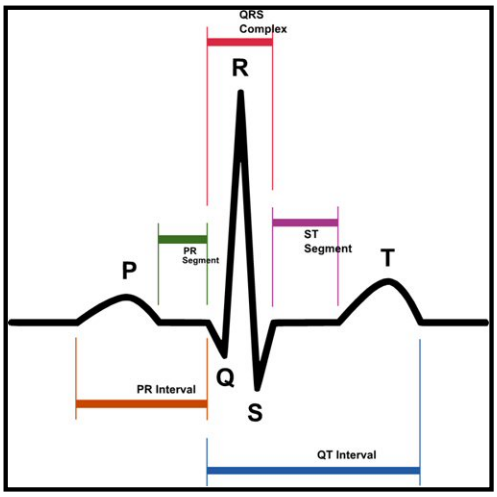


Figure 1.1: typical cardiogram with components labelled

The first electrocardiogram measurement device was developed in about 1903, when Willem Einthoven invented a new instrument called the string galvanometer (5). With it he developed an improved method for measuring the electrical changes that take place in the body upon the contraction of the heart (5). Eventually, he demonstrated that some of these waves result from contractions and electrical changes in the atria of the heart, and others from contractions and electrical changes in the ventricles (5). The string galvanometer made possible the first valid and reliable electrocardiogram, thus giving doctors one of the most valuable tools for the study of heart disease (5).

Variations in a patients ECG signal, particularly changes in the size and appearance of the QRS complex and the t/s waves will inform a trained professional what condition the heart is in (2). Also these changes give them the required information to diagnose the patient's aliment (2). One could apply the knowledge of the professional to an artificial system so that it can make the same insights and diagnosis. This concept has been explored since the early 70's with the use of large computer databases and algorithms but not until recently has it been done with great successes (7). There are systems already on the market for use by veterinarians, but how they are implemented has not been disclosed. No such system has been designed yet that is widely used in human healthcare because doctors feel that they can still diagnosis with greater success then the systems.

1.2 Objectives

The objective of this project is to create an ECG analyzer. The analyzer will take an individual's ECG signal and by using feature extraction and SVM to conclude if the individual has a problem with their heart. In particular my portion of the project is to design and test an ECG acquisition system that is affordable and cost effective. My system must provide an ECG signal that can be analyzed not only by my partner's software but also by a healthcare professional who views the output of my system.

This requires the output of the hardware to resemble the ECG signal of figure 1.1. To do this my system must successful acquire the signal using skin electrodes and then perform various signal conditioning techniques such as filtering and amplification. Once these objectives were achieved the hardware component of the ECG analyzer was completed.

1.3 General Approach to the Problem

Refer to figure 1.2 for general approach schematic. As stated above when the heart beats it produces an electric current. These currents can be acquired from the patients skin by using sensors placed in predefined areas on their body. When this signal is inputted into the system it is of very low voltage and must be amplified to be successfully condition by the ECG analyzer. Also when the signal is received it contains many frequency components, which if left untouched would have the system's output signal not resemble the signal depicted in figure 1.1. This is due to the fact that the useful signal lies in a defined frequency range and by filtering this signal I can ensure that the output signal is in this range. Also with any type of signal processing there is noise from the outside world that leaks into the system and corrupts the input signal. Again I must filter to ensure that this noise is not in our output signal and with this I achieve my goal of a signal that is similar to the one in figure 1.1. Lastly as all software is implemented in the digital domain I must convert our signal from analog to digital to achieve the final product.

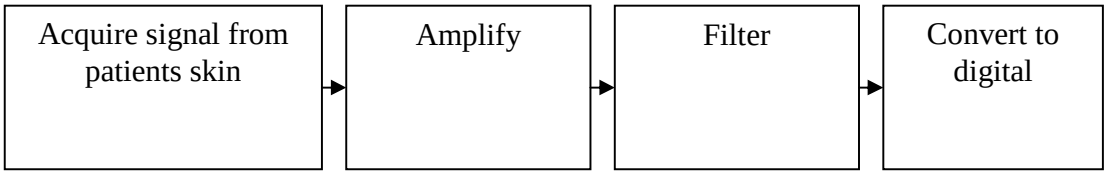


Figure 1.2 General approach to problem

1.4 Scope

The ECG Analyzer project will focus on the requirements for the acquisition of an ECG signal from a patient, amplifying and filtering the signal to follow the medical

requirements of an ECG system and finally patient safety. The acquisition of the signal component of the project will involve selecting appropriate electrodes and placing them correctly to achieve the best signal possible. It will also include appropriate selection of amplification circuitry to apply gain to the low voltage ECG signal. The filtering component is the most important as we must ensure that the signal is in a certain range for proper signal recreation. As the ECG Analyzer is connected to a patient and a computer at the same time there is a need to isolate the patients from the computer so they will not get shocked. Throughout this project I will test the theory and implement the circuit using breadboards. The Analyzer must be cost effective and inexpensive to create, as we would like to market this to third world countries where doctors are in a very short supply. By making it inexpensive does not mean the hardware is allowed to have poor signal reproduction it is the goal of the project to have a high level of signal reproduction.

Chapter 2

Literature Review

2.1 Standard 12 lead ECG

The 12-lead electrocardiogram (ECG) is a diagnostic test that helps identify pathologic conditions, especially ischemia and acute myocardial infarction (2). It provides a more complete view of the heart's electrical activity than a rhythm strip and can be used to assess left ventricular function more effectively. Patients with conditions that affect the heart's electrical system may also benefit from a 12-lead ECG (5). The standard 12-lead ECG consists of six limb leads and six chest leads (5). The electrodes to be attached on the limbs are connected to the wrists and the ankles in rest ECG recording. During the exercise ECG the electrode positions are at the ends of the collarbone and the ridges of the iliac bone. The locations for the chest electrodes according to the recommendation of The American Heart Association are as follows and can be viewed in figure 2.1(5):

- V1: Fourth intercostal space, at the right margin of the sternum.
- V2: The same space, at the left margin of the sternum.
- V3: Midway between V2 and V4.
- V4: Intersection of left mid-clavicular line and fifth intercostal space.
- V5: At the intersection of left anterior axillary line with a horizontal line through V4.
- V6: At the intersection of left mid-axillary line with a horizontal line through V4 and V5.

This type of ECG will often be used as a one-off recording of an ECG, typically printed out as a paper copy.

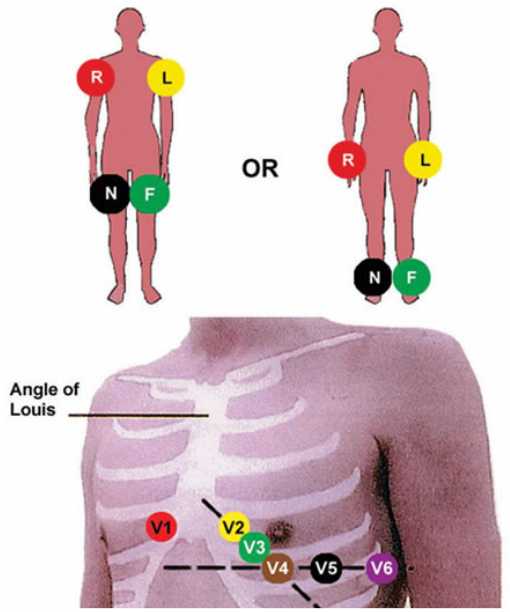


Figure 2.1: lead placement for 12 lead ECG

2.2 Unipolar ECG Lead System

Unipolar leads are based on Wilson's center terminal, which is used as a reference instead of one single reference electrode. The WCT represents a 'mean electrode' calculated from the three limb electrodes (5). The term "unipolar" originates from Wilson's aim to develop an indifferential electrode located at the center of heart (5). By removing the lead used as the measuring electrode from the Wilson's central terminal, Goldberger invented in 1942 the augmented unipolar lead system (5). These augmented leads are far from Wilson's original idea, in spite of that they have become part of the most commonly used clinical standard (5). This type of ECG is used when continues monitoring of the signal is needed and will be viewed only on the screen of an appropriate monitoring device by medical professionals. For example during an operation or while a patient is being transported in an ambulance.

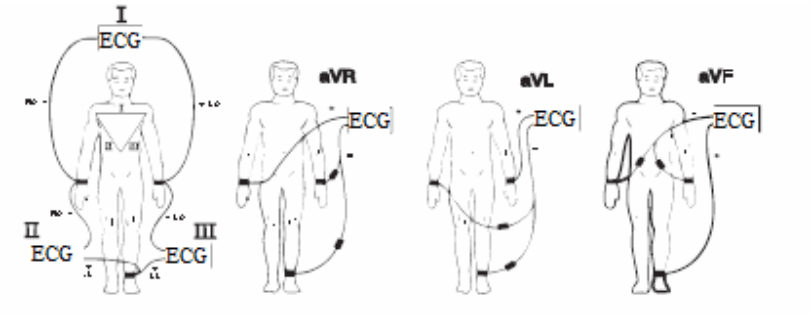


Figure 2.2 Unipolar Lead Placements

2.3 Mobile ECG System

It is sometimes necessary to monitor patient's ECG for either a long period of time or when they are performing everyday tasks. This done after they have survived cardiac arrest, ventricular tachycardia or other cardiac diseases (6). This is done as they are at a higher risk of sudden cardiac death or there is a need to recorded their ECG over time to find when their heart condition reappears. These patients may either be at home or when there is a need for a patient to be mobile in the hospital (7). By using a wireless and wearable monitoring system it is possible to alert healthcare professionals to the patient's condition so that either the necessary action for an emergency rescue can occur (7). Also problems with the heart can be identified, as some cardiac problems do not appear on a regular scan (7). Advanced monitoring solutions using telecommunication systems are used for remote ECG diagnosis. These systems can be divided into two categories real-time mode and store-and-forward mode. The systems available today are either based on standard ECG electrodes and a wired connection to a recording device or by pressing a recording device directly onto a patient's chest when a symptom arises (3). Examples of these sytems are the TTM, Quick Doc, and Holter monitor. The TTM recorder operates on a battery and is about the size of a credit card. When a patient becomes symptomatic, he or she holds the back of the card firmly to the centre of his or her chest and pushes the start button (3). Four electrodes are located on the back of the card are able to sense electrical activity and record it (3). The card can store 30 seconds of activity and can later transmit the recording across phone lines for evaluation (3). The Quick Doc ECG uses silver

electrodes attached to the patient. It acquires the signal and then sends it to the amplification and transmission stages via short-shielded wires (5). The Quick Doc ECG then transmits the information wirelessly to a receiver and finally to a computer. This allows for continuous recording of the ECG signal and the ability to detect abnormal ECG activity (5). Based on these abilities the device transmits alarm conditions to remote Clinical Alarm Station (CAS) without the help of the patient (5). A Holter monitor is an ECG recording done over a period of 24 or more hours. This done by placing three electrodes to the patient's chest and connecting them to a small portable ECG recorder by lead wires as seen in figure 2.3 (3). The patient goes about his or her usual daily activities except for activities that can damage the monitor. Also activities that cause an excessive amount of sweating should be avoided, as they would cause the electrodes to become loose or fall off during this procedure (3). There are 2 types of Holter monitoring:

- continuous recording - the ECG is recorded continuously during the entire testing period.
- event monitor, or loop recording - the ECG is recorded only when the patient starts the recording, when symptoms are felt.

Figure 2.3: Holter Monitor lead placement

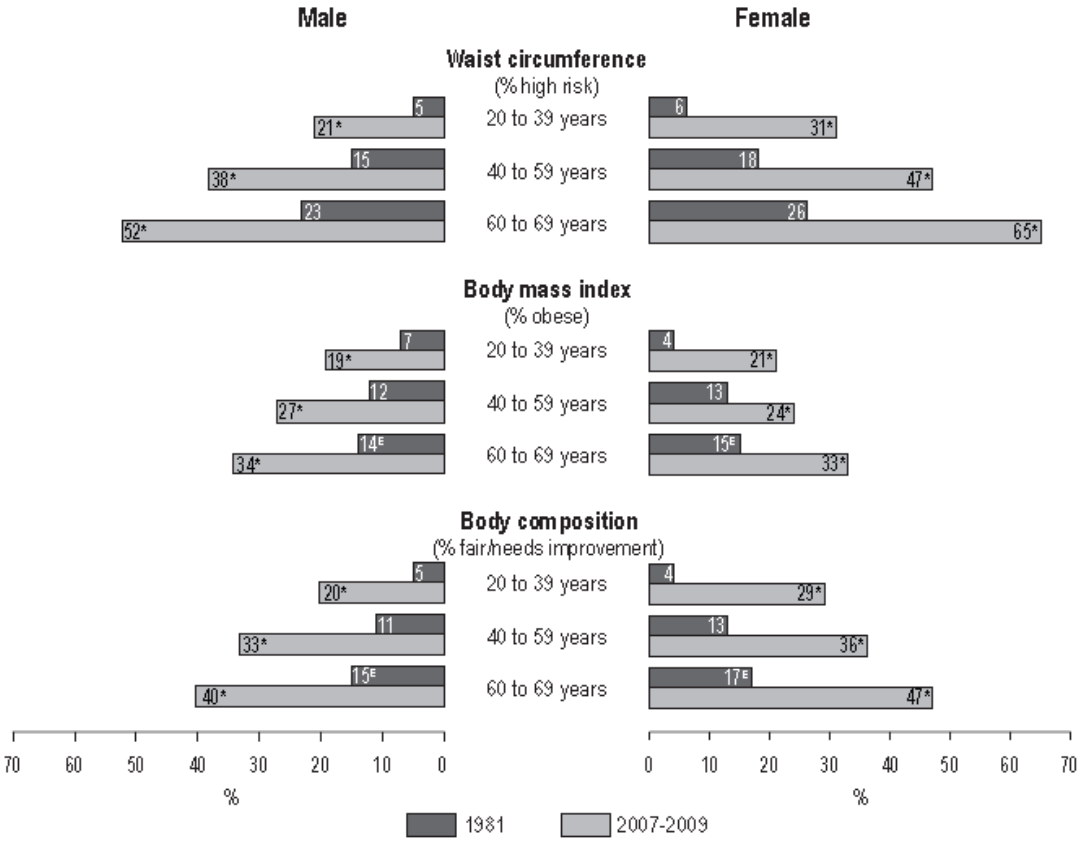
Chapter 3

Statement of Problem and Methodology of Solution

3.1 Problem

Today, five million Canadians are without a family doctor (9). A 2005 survey found that just 23 per cent of Canadians were able to see a physician the same day they needed one—placing this country last among the six studied, including the U.S., Britain and Australia (9). Canada's doctor-patient ratio is among the worst of any industrialized nation: with just 2.2 physicians per thousand people, it ranks 24th out of 28 OECD countries (well below the average of three) (9). And among the G8 countries, Canada ranks dead last when it comes to physician supply (9). It's going to get worse. The aging population—one in four Canadians will be 65 or older by 2056, compared to 13 per cent now—will put huge strains on the health care system, and little is being done to address the doctor shortage that already exists (9). If you refer to table 3.1 it can be seen that there is another problem our country is becoming less physical fit compared to pervious years.

Percentage with suboptimal health benefit ratings for selected anthropometric measures, by sex and age group, household population aged 20 to 69 years, Canada, 1981 and 2007-2009



* significantly higher than estimate for 1981 ($p<0.05$)
E use with caution (coefficient of variation 16.6% to 33.3%)
Sources: 1981 Canada Fitness Survey; 2007-2009 Canadian Health Measures Survey.

Figure 3.1:health of Canadian adults

As one can see our health system is becoming very taxed and there is not much hope for the future that this will be changing. As stated in chapter 1 I set out to create a system that could analyze the ECG signal as variations in this signal can tell a medical professional much about the health of the patient's heart. By having this system at their disposal medical professionals will shorten their time that they have to dedicate to certain patients and therefore allowing them to see more patients in a day. This will also allow them to have less stress, as analyzing an ECG is a time consuming and tedious task. There is also a need for this system in undeveloped nations, as they do not have access to the trained professionals like developed nations

do. So our system particularly the hardware must be chosen to be affordable and cost effective so that these underdeveloped nations can afford to buy this system.

3.2 Overview

An electrocardiograph is a device that can measure the electrical signals produced by the heart (2). Each event during a cardiac cycle produces a waveform that forms the ECG (2). Physician then can analyze this signal to access the state of the cardiac tissue. The ECG implores surface electrodes to acquire the minuscule voltages produced by the heart during the cardiac cycle (2). Pairs of electrodes are placed on different parts of the heart to measure the ECG from different angles. For the theory of how clinical lead placement works please refer to Appendix C.

Electrodes are required for acquisition of the ECG signal from the body. The body acts like a giant resistor and therefore the ECG signal produced in the heart has a smaller amplitude (0.5~4mV) at the surface of the body as compared to the surface of the heart (4). This means that the electrodes have to be sensitive enough to pick up the signal produced and ensure that the signal is not lost during transmission to the amplifier. There are certain requirements for the amplifiers so they provide enough amplification to the signal so that it can be analyzed by the other components of the circuit. As well it must have very high input impedance, a large CMRR and low power consumption (4). There is also a need to filter because the useful signal of an ECG lies between 0.01 Hz and 150 Hz (4). This can be dealt with by using various filter techniques but to ensure that it is in this range of frequencies the signal must be high and low passed filtered. There is also a need to remove the line noise out of the signal, as this will be used in a hospital setting where electric current is running.

The power supply used in the design of the ECG has to be able to support a current draw of 1.1mA and needs to be able to supply power to all other components of the ECG, which could require voltage in the range of +/- 5 to 15 volts. There is also a need to protect the patient, as this ECG will need to be attached to a computer so the

software can analyze the signal. This can be done successfully by isolating the patient from the computer by imploring an isolation amplifier.

3.3 Electrode theory

In order to measure and record the potentials from the heart it is necessary to provide some interface between the body and the hardware. The electrodes are this interface. Electrodes must have the capability of conducting a current across the interface between the body and the hardware (8). The electrode has to serve as a transducer to change an ionic current into an electronic current, which greatly complicates the operation of the electrode (8). The type and size of the electrode is determined by the signal being measured, the location on the body and the dimensions of the generator of the signal (8). In the case of the ECG the signal range is from 0.5~4mV and the frequency range of the signal is from 0.01~250Hz (4). Given these values modern ECG systems can implore a number of different electrodes. One common electrode implored by most ECG systems is the silver chloride electrode and I have chosen this electrode to be used for the project. The choice of our electrode and how it interacts with the patient's body determines what type of amplification hardware I use. The way the electrode interacts with the patients can be shown as an equivalent circuit and can been seen in figure 3.2.

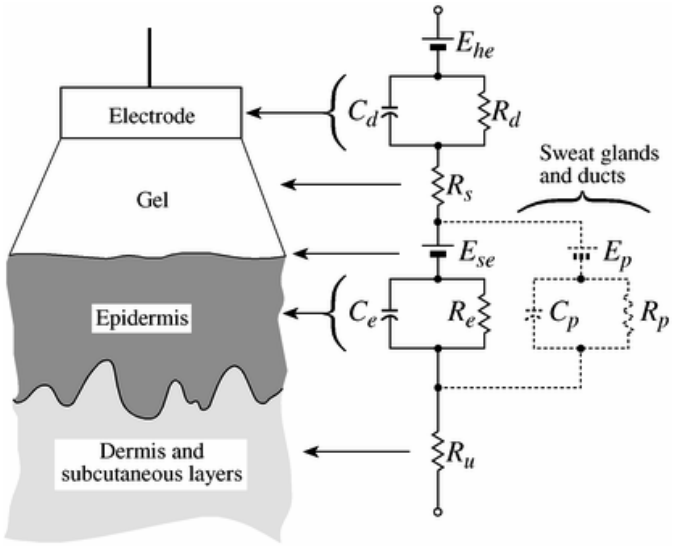


Figure 3.2: equivalent circuit of electrode skin interface

The ECG signal produced by the cardiac tissue loses signal strength because it must travel through bone and muscle and therefore the signal faces an internal resistance (8). The ECG is only acquired by measuring the difference in half-cell potentials between electrode I and electrode II to give the potential difference (in mV) (8). The dermis and subcutaneous layers are modeled as resistances since they are mainly composed of fat and have no electrical properties (8). The epidermis specifically the stratum corneum is a membrane that is semi-permeable to ions, so if there is a difference in ionic concentration across this membrane, there is a potential difference of Ese created (8). To minimize the effect of the stratum corneum it must be removed, or least part it, from under the electrode (8). This can be done by abrading the skin by vigorous rubbing of an alcohol swab in the area that the electrode is being placed. The rest of the epidermal layer is found to have an electric impendence that behaves as a parallel RC circuit (8). The sweat glands and ducts secrete fluid that contain Na+, K+, and Cl- ions, which create a potential difference between them and subcutaneous layer (Ep) (8). There is also a parallel RC combination in series with the potential, which is acquired from the walls of the sweat glands and ducts (8). If conductive gel is used it is represented as a resistance and it will improve the conductive properties of the skin. The last part of electrode theory will focus on motion artifacts and how they are created.

The silver electrode develops a double layer of charges when it is in contact with the electrolyte (8). When the electrode/electrolyte contact is disturbed during motion; the half-cell potential of one of the electrodes changes relative to the other (8). This generates a sudden potential difference between the electrodes, which results in motion artifacts in the ECG signal (8). When the motion of the electrode stops the double layer is re-established and the initial half-cell potential is obtained again. But the danger is in the fact that a sudden spike in voltage can saturate the instrumentation amplifiers. Variations in the skin/electrolyte interface can cause motion artifacts as well (8). To prevent motion artifacts ensure that the skin is prepared properly and the electrodes are placed correctly.

3.4 Amplification theory

An ECG signal has a range of 8x10^-5 to about 5x10^-3 V in amplitude which means that this signal must be amplified (8). For the system to be able to correctly produce the signal a gain of about 1000 is necessary. To successfully produce the required gain instrumentation amplifiers are the best choice. An instrumentation amplifier is a type of differential amplifier that has been outfitted with input buffers that eliminate the need for input impedance matching and thus make the amplifier particularly suitable for use in measurement and test equipment (10). Additional characteristics include very low DC offset, low drift, low noise, very high open-loop gain, very high common-mode rejection ratio, and very high input impedances (8). They also have the benefit of being able to adjust the gain of the amplifier circuit without having to change more than one resistor value. Instrumentation amplifiers are used where great accuracy and stability of the circuit is needed for both short- and long-term (10).

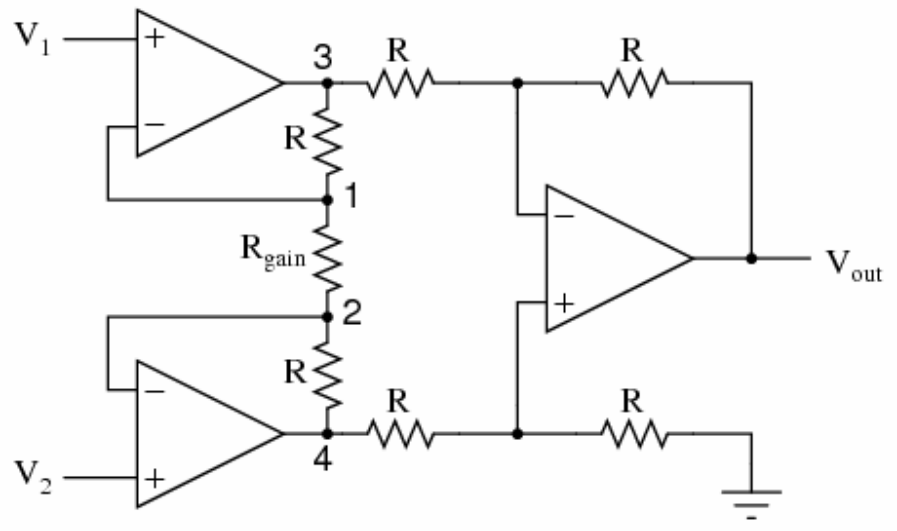


Fig 3.3: Instrumentation Amplifier (normal viewed as one opamp)

When acquired from the silver electrode the very small ECG signal will be accompanied by a large ac common-mode component (up to 1.5V) and a large variable dc common-mode component (300mV) (8). To deal with these components

it is imperative that they are met by an amplifier with a High CMRR (8). The common-mode rejection ratio specified by the Association for the Advancement of Medical Instrumentation (AAMI) is 89 dB minimum for a clinical ECG and 60 dB for an ambulatory ECG (4). As stated above the choice of an instrumentation amplifier will allow for a high CMRR and meet or surpass these requirements. The electrode/skin interface has impedances that can range from 1K to 1M ohms (4). This impedance is made up of the equivalent impedance of the electrodes, the fat volume underneath the skin which has an impedance associated with it, and the resistance of the body (4). Also there is a dependence on the skin condition. Its preparation will contribute to the impendence and if the system is worn for a long period of time the skin will change and therefore the impedance will also change (4). Finally if the signal is in the frequency range of 0.01 Hz to 1 Hz an increase in the electrode/skin impedance is expected as the capacitive component of the skin would be much higher in this range (4). If the wrong amplifier is chosen a voltage divider will appear between the electrodes and input of the amplifier, which will lead to signal loss. High input impedance on the amplifier would prevent the formation of voltage dividers and again we see the instrumentation amplifier has this quality.

3.5 Filter theory

The ECG signal contains frequency components between 0.01 Hz and upwards of 200 HZ. In a summary of performance requirements for an Electrocardiograph it is said that the minimum cutoff frequency is 150 Hz and for our application this was the cutoff frequency I chose to use (4). To ensure that the signal was between 0.01 Hz and 150 Hz high and low pass filtering had to be done. There are many different types of filtering circuits with each one responding differently to changing frequency. Some examples of these filtering circuits are the simple RC filter, Bessel filter, Butterworth filter and Chebyshev filter. Some of the low pass frequency responses of these filters can be seen in figure 3.4. Each one of these filters display different characteristics and they can be gauged by the response in the pass band, by its cutoff frequency and its rate of frequency roll off. Every filter must attenuate the input power by –3 dB at it's required cutoff frequency (10). The amount of attenuation

seen by the frequencies after the cutoff is controlled by the order of filter and type of filter chosen.

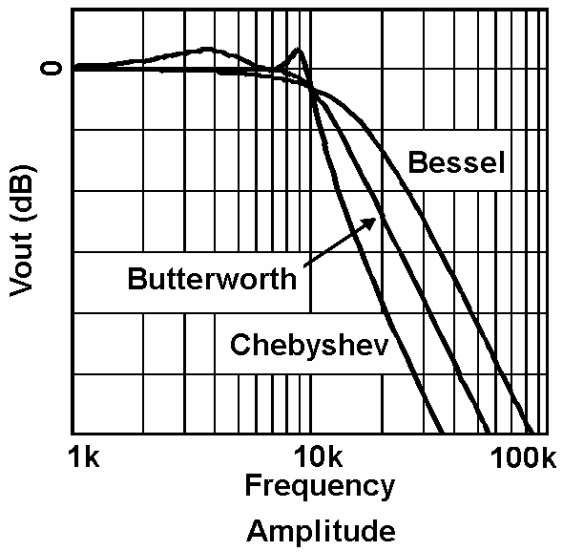


Figure 3.4: Frequency response of various low pass filters at arbitrary cutoff

To limit the signal to the 150 Hz cutoff a low pass filter will be needed. The ideal low pass filter can be seen in figure 3.5. As it can be seen we are looking to allow anything below 150 Hz to pass and essentially kill any signal above it. As the world is not ideal we have to live with a frequency roll off in the stop band and various looking frequency responses in the pass band (10). As my ECG system must interface with a computer or in the future a microprocessor a high pass Bessel filter will be used. The choice of a Bessel transfer function is motivated by the fact that it has optimal phase response. That is, it has the desirable property of near-constant group delay (which can be seen in figure 3.6) and negligible phase distortion (10). This optimality in phase response comes at the price of decreased roll-off steepness in the transition band relative to other transfer functions (10). This optimal phase response and constant group delay will lead to optimal performance of the analog to digital converter.

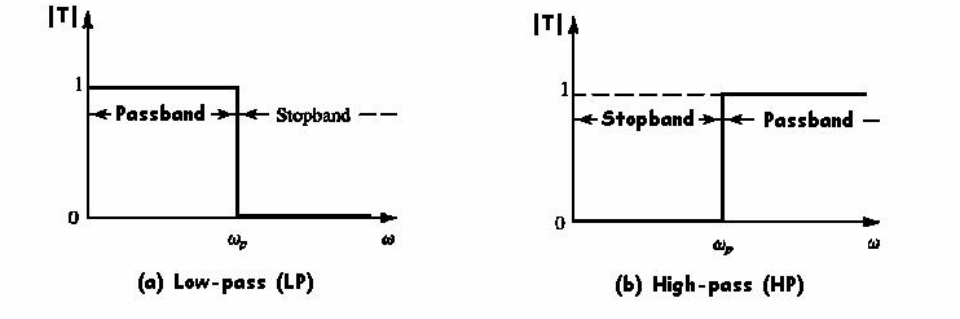


Figure 3.5:Ideal low and high pass filters

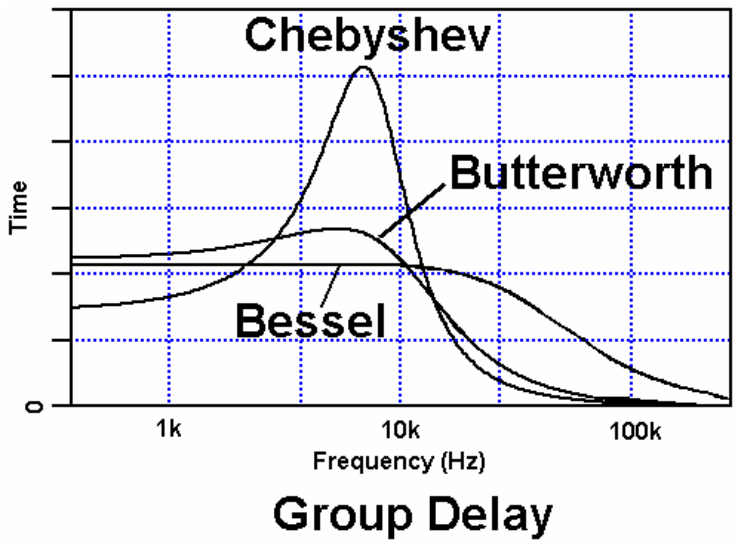


Figure 3.6: Various Low pass filter group delays

To limit the signal to the 0.01 Hz cut-off a high pass filter will be used. The ideal high pass filter response can be seen in figure 3.4. As stated for the low pass filter we are unable to achieve this ideal response and have to make sacrifices in both the pass and stop band and this is no different for the high pass filter. In this case as we are mostly trying to block the dc component of our signal a simple first order RC high pass filter can be used. This allows for the ECG analyzer to meet its goal of being cost effective with the ability to provide a quality signal to be analyzed. The reason this choice gives us the ability to be cost effective is the fact that all that is needed to create the filter is a correctly chosen resistor and capacitor. This filter can be seen in figure 3.7. The choice of The RC circuit will allow us to have a linear response in the pass band and relatively good drop off in the stop band as seen in figure 3.8.

It also should be noted that both filters will attenuate some useful signal but these signals will have larger amplitudes then the noise and therefore the ECG signal will remain intact.

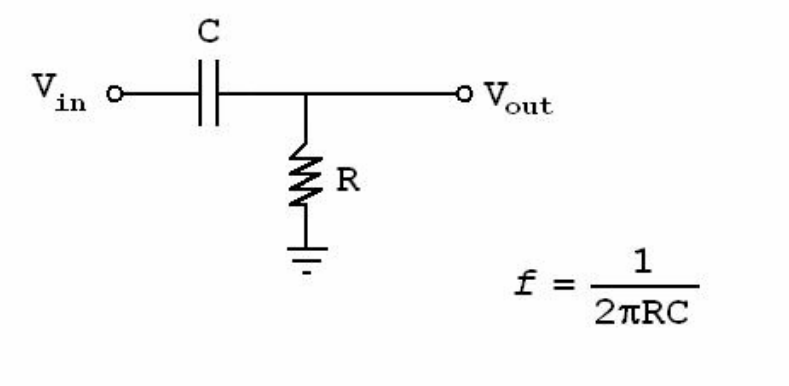


Figure 3.7: RC high pass filter

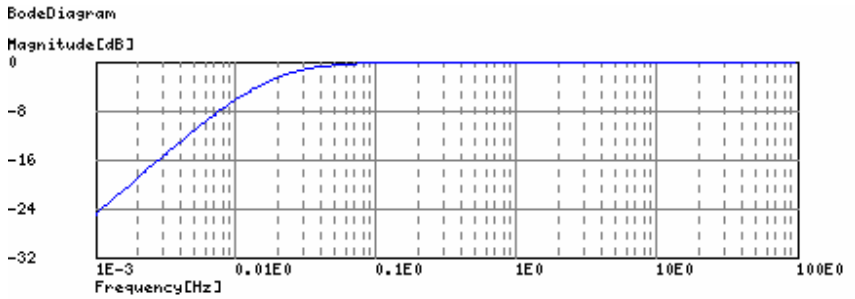


Figure 3.8: RC high pas frequency response

Our ECG Analyzer will be exploring frequencies above 60 Hz, which means we must now compete with line noise. This is a major source of interference when acquiring the ECG signal is the electrical power system (8). In a typical hospital room or physicians office power lines are connected to other pieces of equipment and are in the floors, walls and ceiling. These power lines can affect the recording of the ECG by introducing interference (noise) at the line frequency, which in Canada is 60 Hz (8). This interference can seep in by two methods the power line coupling with the leads or the patient (8). Both these process can be seen in figure 3.9. To remove this 60 Hz noise a twin notch filter can be used and its frequency response can be seen in figure 3.10. The twin-T circuit has the advantage that the quality factor (Q) can be varied via the inner gain (G) without modifying the mid frequency (fm) (10). The larger the quality factor the narrower the response will be which will allow for

minimal signal loss (10). This response will remove some useful signal of the ECG signal and attenuate some of the signal around 60 Hz but as before these frequencies will have a larger amplitude then the noise.

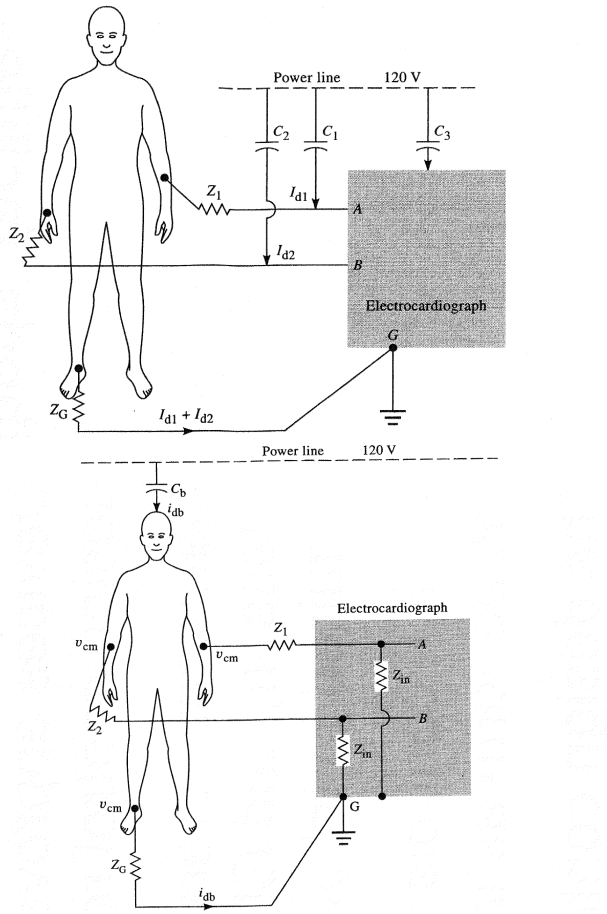


Figure 3.9: ECG lead coupling and body coupling respectively

Figure 3.10:Notch filter frequency response

3.6 Isolation theory

As this system will be interfacing with a computer there must be a barrier set up between the computer and the patient. This will keep voltage from the computer that could potential hurt the patient away from them but also ensure the safety of the internal components of the computer. This can be done by using an isolation amplifier. Isolation amplifiers are devices that break the ohmic continuity of electric signals between the input and output of amplifier (8). They usually consist of an instrumentation amplifier at the input followed by a unity gain isolation and a general model can be seen in figure 3.11 (8). The isolation can occur in three ways transformer isolation, optical isolation or capacitive isolation (8). Transformer isolation approach uses either a frequency modulated or pulse modulated carrier signal with a small signal bandwidths to carry the signal (8). The optical method uses a LED on the source side and a photodiode on the output side, which uses the brightness of the LED to deduce the voltage of the signal at that current point of time (8). Lastly the capacitive method uses digital encoding of the input voltage and frequency modulation to send the signal across a differential ceramic capacitive barrier (8).

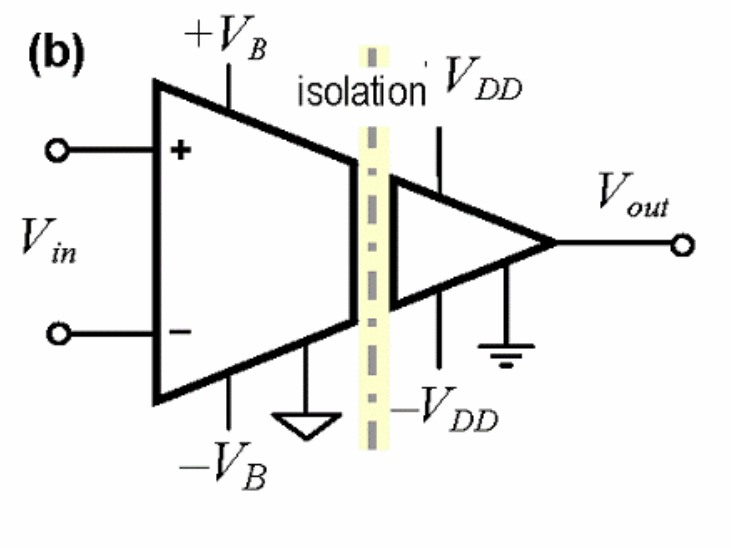


Figure 3.11: Isolation Amplifier

Chapter 4

Experimental and Design Procedure

4.1 Overview

The design of the hardware for the ECG analyzer was broken up into three individual sections. These sections are electrode and amplification, filtering and finally isolation. A block diagram of the entire hardware design can be seen in figure 4.1 and the design and testing of each of these blocks will be described in the following sections. This block diagram was the next generation of the block diagram seen in figure 1.2 and by the end of this chapter a full circuit diagram will replace this block diagram. It is to be noted when testing the circuit two power sources were used. The first was a +/-15 V source from a test board and the second was a +/- 5 V battery source.

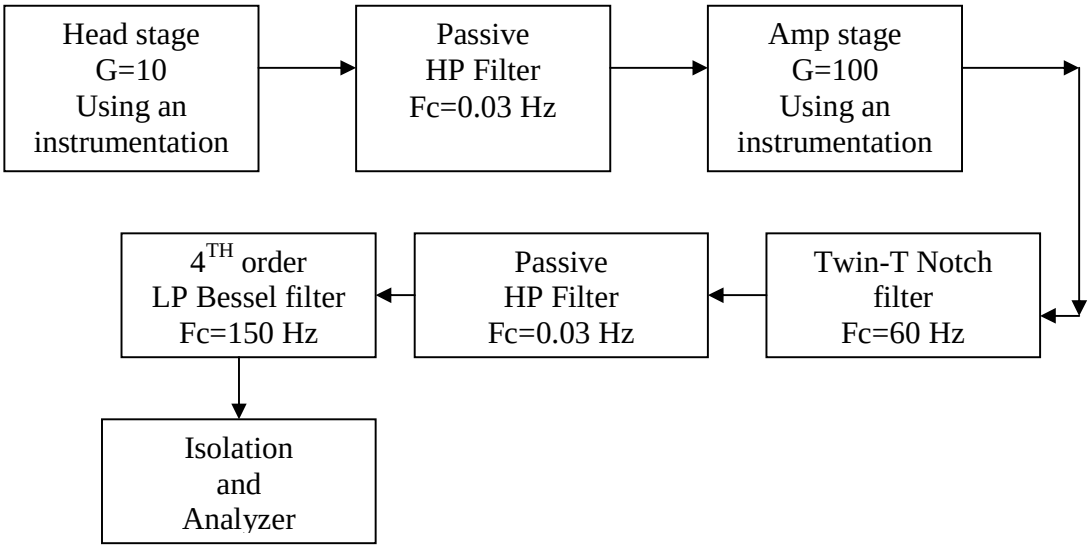


Figure 4.1: Block diagram of full hardware implementation

4.2 Electrode and Amplification

The first part of the hardware and probably the most important as they acquire the signal are the electrodes. The electrodes that were used for this project were the 3M™ Red Dot™ Wet Gel Monitoring Electrodes and can be seen in figure 4.2. Red Dot Wet Gel electrode is a round-style 3M™ Micropore™ Surgical Tape-backed electrode designed for long-term use (11). The electrodes have wet gel conductive columns and 3M™ SureSeal moisture vapour barrier caps to ensure product freshness (11). Each electrode also has a border adhesive that works well for all types of skin conditions (11). These electrodes were not chosen for any particular reason since they were donated to the project but they proved to perform very well with our 3 lead ECG system. The electrodes were placed on both the wrists and one on the right ankle. For a more in depth explanation of lead placement refer to appendix C. To prepare the skin as outlined in the theory alcohol wipes were used and the skin was rubbed until a slight redness appeared. The leads that connected the electrodes to the rest of the hardware were also donated but as the electrodes, they performed very well and directly attached to the electrodes with no modifications. The lead connected to the right wrist was attached to the positive input of the instrumentation amplifier and the lead connected to the left wrist was attached to the negative input. Lastly the lead connected to the left ankle was connected to ground. This set-up allowed us to view the heart at two different angles.

Figure 4.2:Electrodes

The ECG signal has a range of 0.5~4mV which is a very low voltage signal. With this signal being such a low voltage it needed to be amplified by a gain of 1000 to provide good signal reproduction. The amplification stage was broken up into two stages to ensure no saturation occurred. To perform amplification at both stages the pga204 instrumentation amplifier by burr brown was used which can be seen in figure 3.4. The PGA204 is a low cost, general-purpose programmable-gain instrumentation amplifier offering excellent accuracy. Gains are digitally selected: from the values of 1,10,100,1000. The digital inputs A0 and A1 select the gain according to the logic table in Figure 4.3. Logic "1" is defined as a voltage greater than 2V above digital ground potential (pin 14) which was done by attaching it to the +5 V power source. Digital ground was connected to ground. By choosing an amplifier that selects it's gains digitally this insures the gain is actually what is chosen, which other systems cant do because they suffer from the variability of resistors.

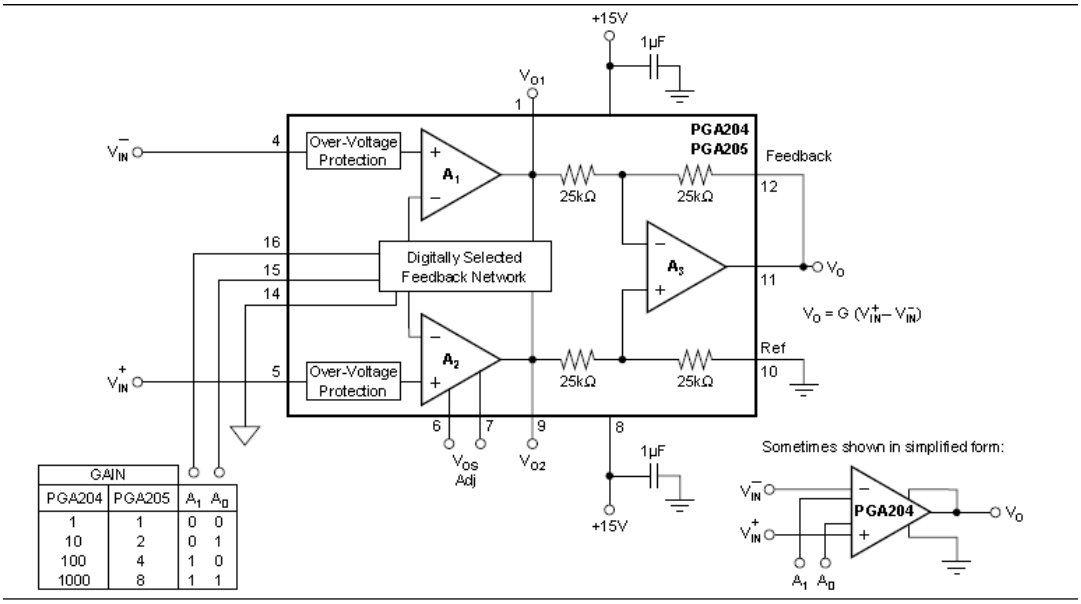


Figure 4.3:pga204

The first stage, which was the head stage in figure 4.1, was required to have very high input impedance, high CMRR and provide a gain of 10. These guidelines were achieved with the choice of this instrumentation amplifier as the pga204 has an input impedance of $10^{10}\Omega$, a selectable gain of 10 and a CMRR of 96 db, which surpasses the standard for clinical ECG. To achieve a gain of 10 A1 (pin 16) was connected to digital ground (pin 14) and A0 (pin 15) was connected to + 5 V.

The second stage, which was the amplification stage in figure 4.1, was required to provide of a gain of 100. Again the pga204 was used because of it digitally selected gain and gain accuracy. This time to achieve a gain of 100 A0 (pin 15) was connected to digital ground (pin 14) and A1 (pin 16) was connected to + 5 V.

Both stages were powered by the +/- 15 V power source to ensure that the pga204 was operating at peak performance. With the pga204 being easy to use, accurate, affordable, and able to meet the requirements of high input impedance and high CMRR it was the wise choice for our application.

4.3 Filtering

As stated in the theory section there was a need to have a signal in the range of 0.01 Hz and 150 Hz because this was the minimally acceptable frequency range to reproduce a clinically acceptable ECG signal. Also there was a need to take out the 60 Hz frequency of the signal, as this is the frequency of line noise. To achieve this various filtering techniques were used. For filters that required the use of an operational amplifier the TLC272ACP by Texas Instruments was used. The reason why this op-amp was used is because it had a bandwidth well beyond our requirements, a low supply voltage range of 3 to 6 V and finally it was cheap. These reasons allowed me to keep to my main goal of creating a cost effective ECG analyzer.

The first frequencies filtered were the ones below 0.01 Hz. To perform this task I used a first order RC high-pass filter as seen in figure 4.4. Also in figure 4.4 accompanying the filter diagram is the equation to calculate the cut-off frequency.

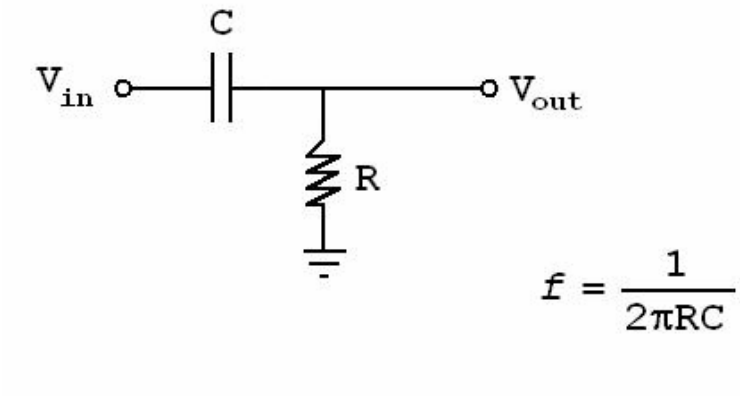


Figure 4.4: R C Low pass Filter

At my disposable I had three discap 35 V 1µF capacitors and two 1 M Ω +/- 5% resistors to create each of the high pass filters. The three capacitors were connected in parallel to achieve a total capacitance of 3µF. Connected to the end of the parallel capacitors was the output of the filter and the resistors, which, were connected in series to achieve a resistance of 2 M Ω. The end of the series resistance was connected to ground. By using the equation in figure 4.4 it can be seen that a cut-off of 0.03 Hz was achieved. This was not 0.01 Hz as stated before but many other ECG designs have a cut-off as large as 0.05 Hz and still reproduce a good ECG signal. As you we will see in the results section this was no different for a cut-off of 0.03 Hz. It can be seen in Figure 4.1 that there are two high-pass filter stages and both come after each of the amplification stages. The first high pass filter is connected to the output of the pre-amp and the input of the amplification stage. This is done as there is a DC signal accompanying our useful signal and I do not want to amplify the DC signal with a gain of a 1000 as it can have a larger voltage then the ECG signal. If the DC signal was amplified by a gain of 1000 I may have lost our signal or saturated the op-amps in the circuit. By high pass filtering I remove any DC signal as it has a frequency of 0 Hz. The second high pass filter is positioned after the notch filter and before the low pass filter. This was done to ensure before I output the signal that there is not a DC signal contained with our ECG signal. The reason that the signal could contain a DC component is because it is know that when amplifying a signal a small DC signal could be added from the amplifier.

The frequency that I worked with next was the 60 Hz line noise. This was the noise that would be inputted from other electronic devices in a hospital setting, power lines, computers etc. To remove this noise I implored the use of the twin-t notch filter, which can be seen in figure 4.5. Knowing that the cut-off frequency was 60 Hz. The design of this filter started with the selection of the resistors and capacitors as these were the variables in the cut-off frequency equation (refer to figure 4.4 for equation). By choosing a typical capacitor value of 56 nF I was able to choose a resistor value of 47.5 K Ω. With these two values I was able to specify the other resistors and capacitors attached to Vin+ of the operational amplifier as seen in figure 4.5. The next factor used was the rejection quality, which is the most important factor when designing the notch filter as it determines how narrow the notch will be. The narrower the notch the less that surrounding frequencies will be attenuated by the filter. Knowing that a Q value between 1 and 10 will provide a desirable notch I intend to use the rejection formula to find a range for the R2 value when R1 is 2 K Ω. After the range was found a potentiometer could be used to move through this range and find the most desirable response. When it came time to do this the potentiometer I had did not provide the correct range of resistance when connected to the circuit and provided an undesirable response. To solve this problem I used a 1 K Ω resistor in the potentiometers place, which corresponds to a Q of 1, and with this resistor value a desirable response was achieved. With this the notch filter was created.

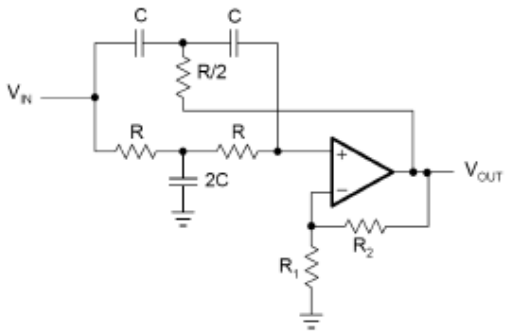
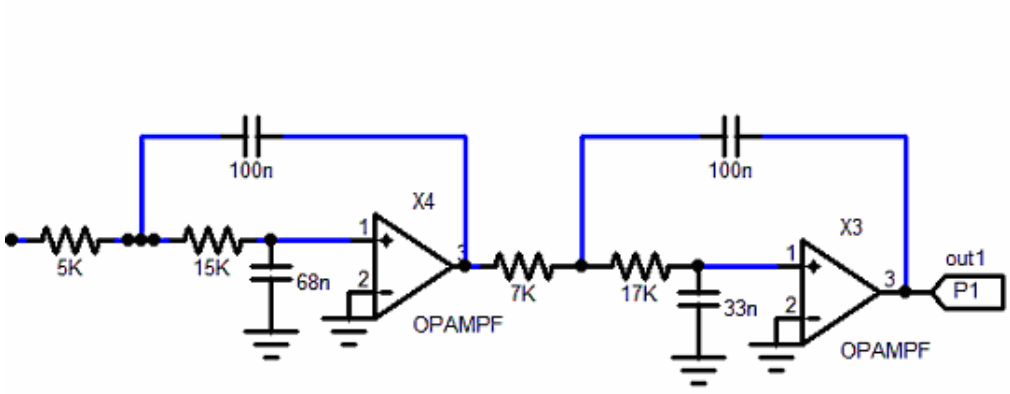
Type	Active Twin T Notch Filter
Schematic	
mid-frequency:	$f_m = \frac{1}{2\pi RC}$
inner gain:	$G = 1 + \frac{R_2}{R_1}$
pass band gain:	$A_0 = G$
rejection quality :	$Q = \frac{1}{2(2-G)}$

Figure 4.5: Notch Filter and parameters

The last filter block was the low pass filter with a cut-off of 150 Hz. This filter was required to be the most accurate and provide the best response. This was achieved by using a fourth order Bessel low pass filter and its implementation can be seen in figure 4.6. Every low pass filter such a Bessel, Buterworth etc. has the same general layout but what sets each of them apart and gives them their specific characteristics are the values of the resistor and capacitors. It is to be noted that the equations shown below are for one op-amp section which is for a 2nd order circuit response to achieve fourth order the equations are multiplied by themselves and the same process performed on the first stage is performed on the second. To find the values of the resistor and capacitors we must first look at the second equation in figure 4.6 the values for a and b are obtained from a table of Bessel coefficients. This table can be found in appendix A. The coefficients are for the fourth order response and are taken right out of the table for that order. The first equation now shows us what a and b are equal to in relation to the circuit. Using the values from the table and the relationship they share to the circuit we can choose values for the capacitors and solve for the resistances. It is to be noted that wc=2πFc. As stated above this process was repeated for the second stage of the filter.

With this the filtering stage was complete. For example calculations please refer to appendix D.



$$A(s) = 1/(1 + \omega_c C_1(R_1 + R_2)s + \omega_c^2 R_1 R_2 C_1 C_2 s^2)$$

$$A(s) = \frac{A_0}{1+as+bs^2}$$

Figure 4.6: Low Pass Bessel filter and Equations

4.4 Isolation Amplifier

As we were connecting the ECG hardware to a computer there was a need to isolate the patient from the computer to ensure patient safety. To perform this task I implored the use of an isolation amplifier, which performs the required task with ease. The isolation amplifier I chose to use was the iso124 by Texas instruments. This isolation amplifier was a high precision low cost version, which again helps me maintain my goal of creating a precise low cost ECG analyzer. It is 100% tested for high voltage breakdown and is rated for up to 1500Vrms, which is more then enough to protect the patient. This isolation amplifier uses the capacitive method of isolation that was described in the theory section. By imploring the use of this isolation amplifier we ensured the patient was safe.

4.5 Final Circuit

As I have described the entire design process it is time to see the final design and implementation of the circuit in the next two figures.

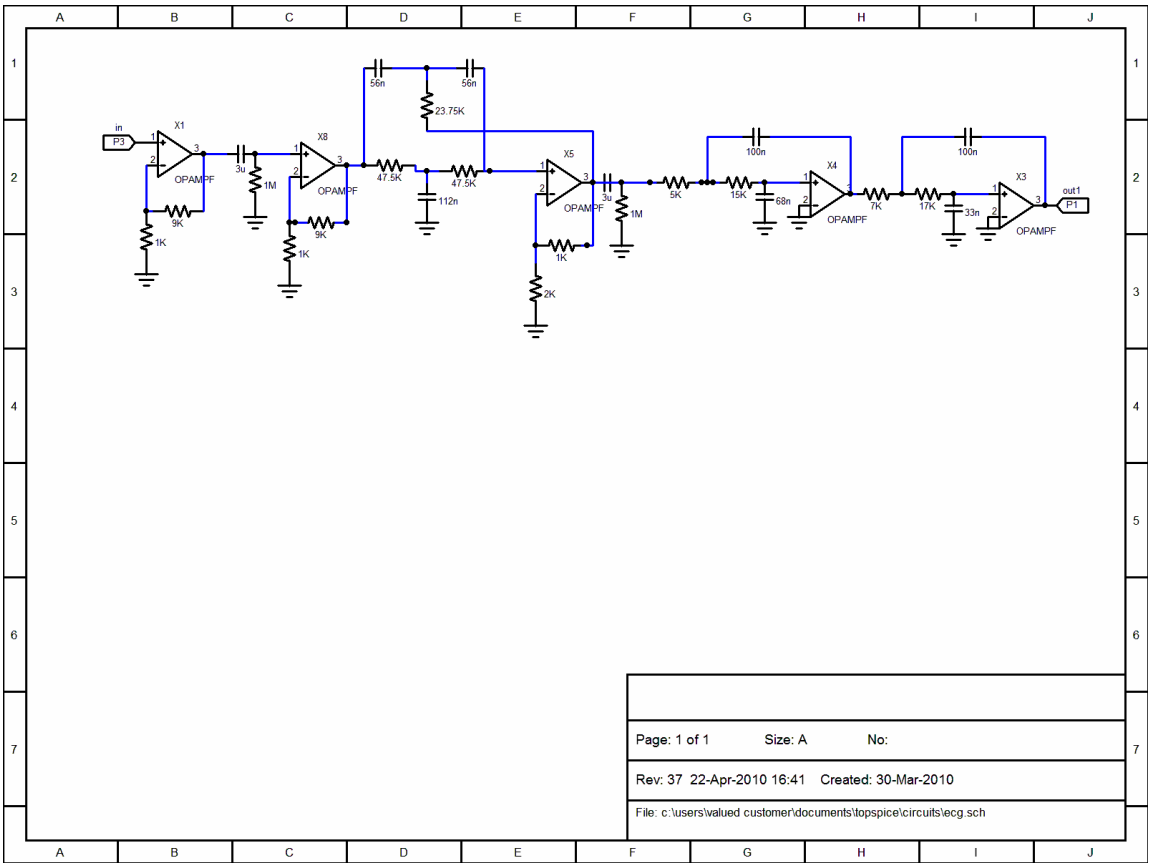


Figure 4.7: Final design of the circuit

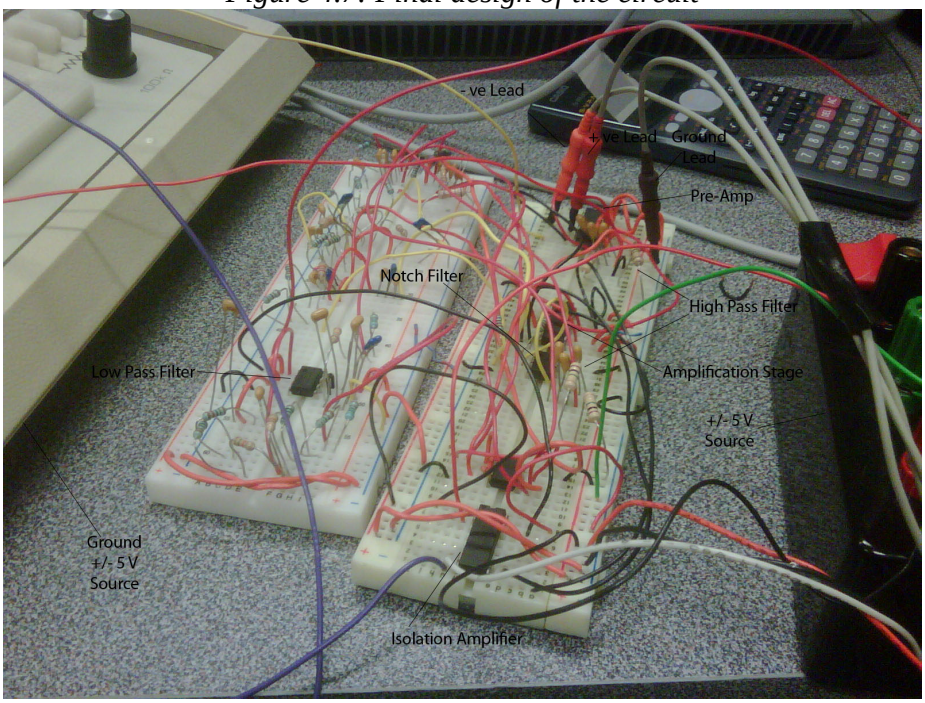


Figure 4.8: Final implementation of circuit

4.6 Testing

To test the operation of my design I performed a frequency sweep from 0 to 1000 Hz to ensure that not only my filters were working correctly but also to ensure my amplification stages were providing the correct amount of gain and final that the isolation amplifier was working. To perform the frequency sweep I connected a signal generator form the lab to the corresponding positive and negative inputs of the pre- amp stage. I then proceeded to attach the positive and negative outputs of the system to the oscilloscope in the lab. The signal that I used to test the hardware was a sin wave with amplitude of 10 mV and varying frequency. The actual amplitude outputted by the signal generator was 28 mV and knowing this allowed for a more precise dissection of the hardware. The setup for the testing can be seen in figure 4.9. It is to be noted that when performing the frequency sweep I could only have a max gain of 100 because with a gain of 1000 the lowest signal of the signal generator will still saturate the op-amps. To correctly visualize the performance of my circuit the results of my frequency sweep were graphed into a bode plot. A bode plot shows the relationship of the gain of the output in relation to the input in db and the frequency in Hertz. The bode plot can be seen in figure 4.10. It can be seen in the bode plot that the filtering amplification and isolation portions of the hardware were working correctly. The bode plot shows that very low frequencies are attenuated and this is expected because we have the high pass filter filtering out signals below 0.03 Hz. The bode plot shows the signal with a gain of about 100 before 30 Hz which is again what we should see as this is located in band of frequencies not filtered. After 30 Hz the gain begins to drop with the lowest gain at 60 Hz, which is the Notch filter performing its required task of removing the line noise. After 60 Hz the gain begins to rise again but never reaches its max gain as the low pass filter cut-off has been reached and the signal becomes attenuated again. The attenuate is sharp as expected with the choice of a 4^{th} order filter and the attenuation continues for the rest of the frequency range. The frequency sweep and bode plot visualization were great ways to test the hardware.

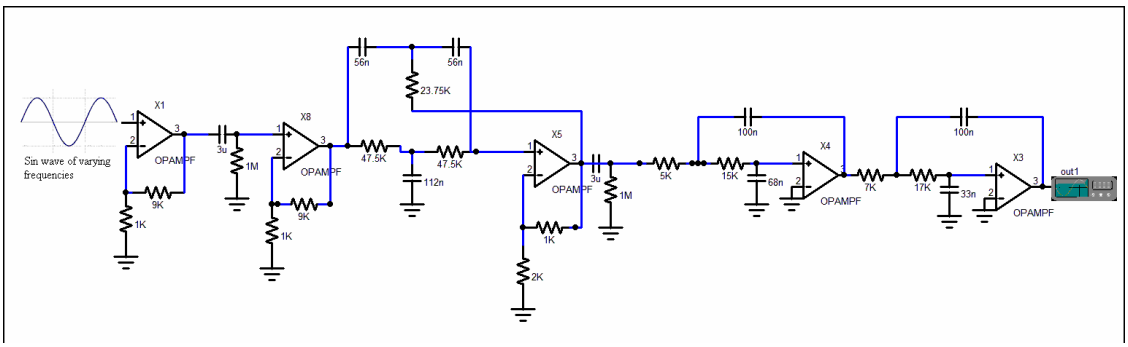


Figure 4.9: Testing setup

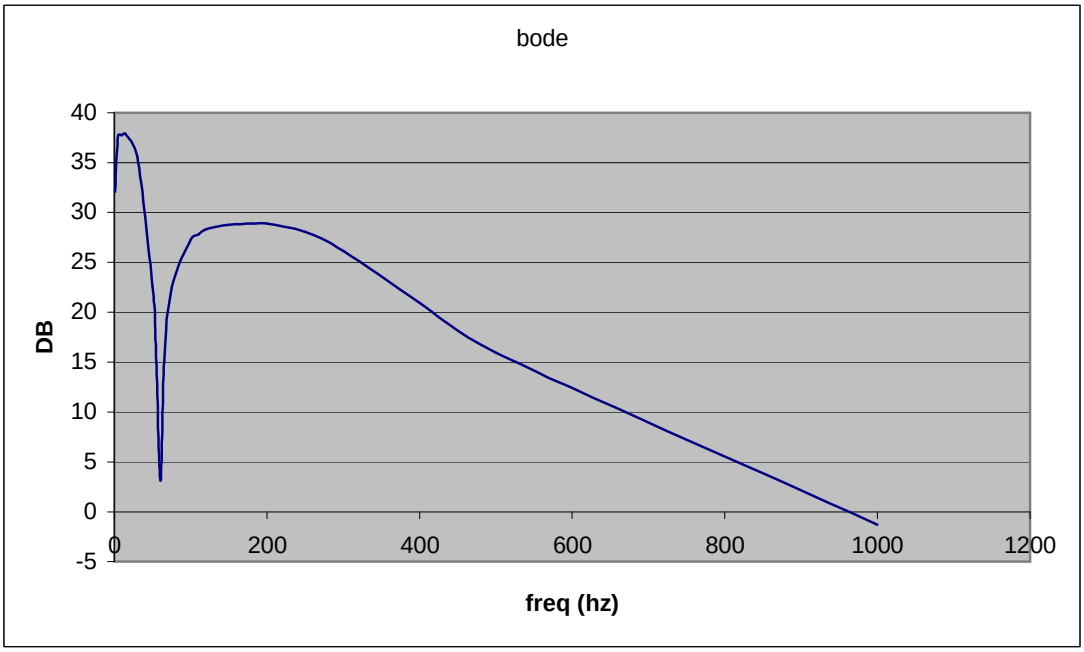


Figure 4.10: Bode plot

Chapter 5

Results and Discussions

5.1 Overview

This section of the report will critically review the results that were obtained from the ECG system when it was attached to a test subject. The test subject was connected to the ECG system through a three lead ECG setup using Ag/Cl electrodes for signal pick up. The leads were connected to the left and right wrist and the right ankle. The patients skin was lightly abraded and cleaned with alcohol wipes before the electrodes were attached to ensure the best chance of signal pick up. The left wrist electrode was attached to the negative input of the instrumentation amplifier while the right was connected to the positive end. The right ankle lead was connected to ground. Finally the electrode wires attached to the patient were short and twisted to avoid motion artifacts and to reduce electromagnetic noise.

5.2 ECG Acquisition

The test subjects for this section of results were a 21-year-old male and a 22-year-old male. Both males are healthy young men with no know heart condition and therefore will produce a healthy signal. Their signal should appear relative to close to figure 5.1 as this was used as the expected result of a healthy ECG signal. Their ECG signal will contain a P wave, QRS complex and T wave.

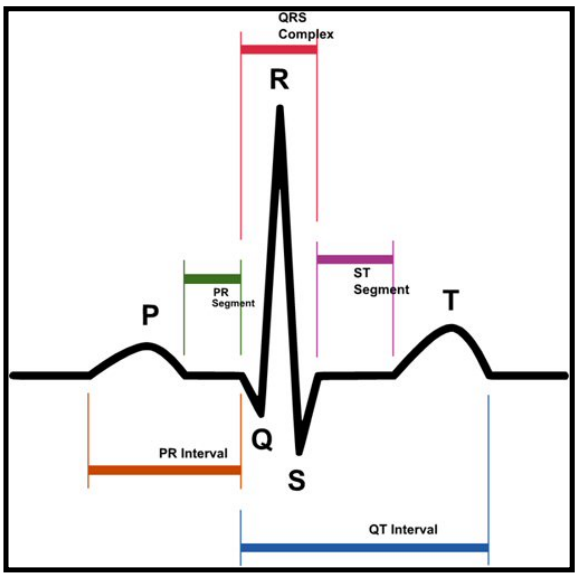


Figure 5.1: Baseline for ECG signal we hope to produce

When we first attached the ECG analyzer to a test subject, the signal that was produced on the oscilloscope and on the computer proved to be results that were not sufficient and had little relation to the baseline ECG signal of figure 5.1. The result can be seen in figure 5.2. As it can be seen the ECG signal that was produced hardly dipped below 0 v and the P, Q and S portions of the wave were missing. These portions of the wave were key features needed by my partner's software to analyze the ECG signal and not having them was unacceptable.

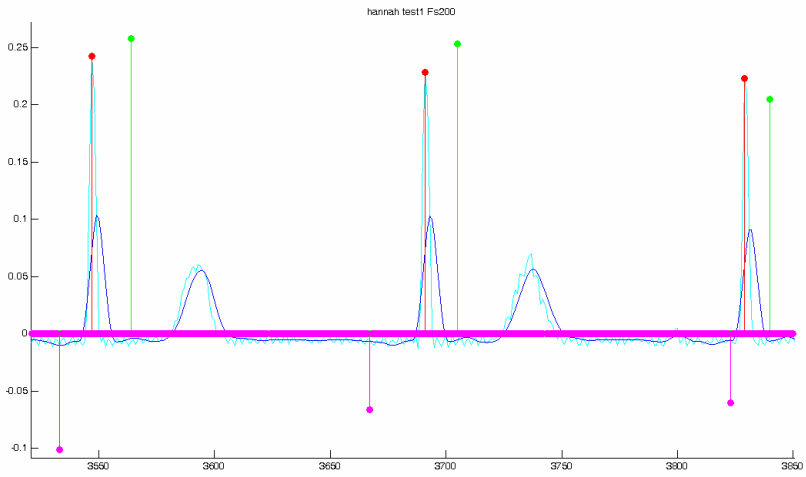


Figure 5.2: unsatisfactory ECG output of ECG analyzer

This was a very perplexing problem and one that I could not find the root of on my own. I approached Dr. Patriciu as he taught a course in instrumentation last semester and after looking at my circuit and the waveform produced he was able to notice that there was a need for the TLC272ACP to be connected between 5 V and –5 V opposed to 5 V and ground. What was occurring was that the ECG signal was being clipped and so I was unable to reproduce the desired waveform. By changing the way the op-amp was connected I was able to change the applicable voltage range of operation of the op-amp.

After the above changes were applied the desired result was achieved. I had designed and implemented a low cost ECG Analyzer that produces a very good signal for our application. One instance of this signal can be seen in figure 5.3.

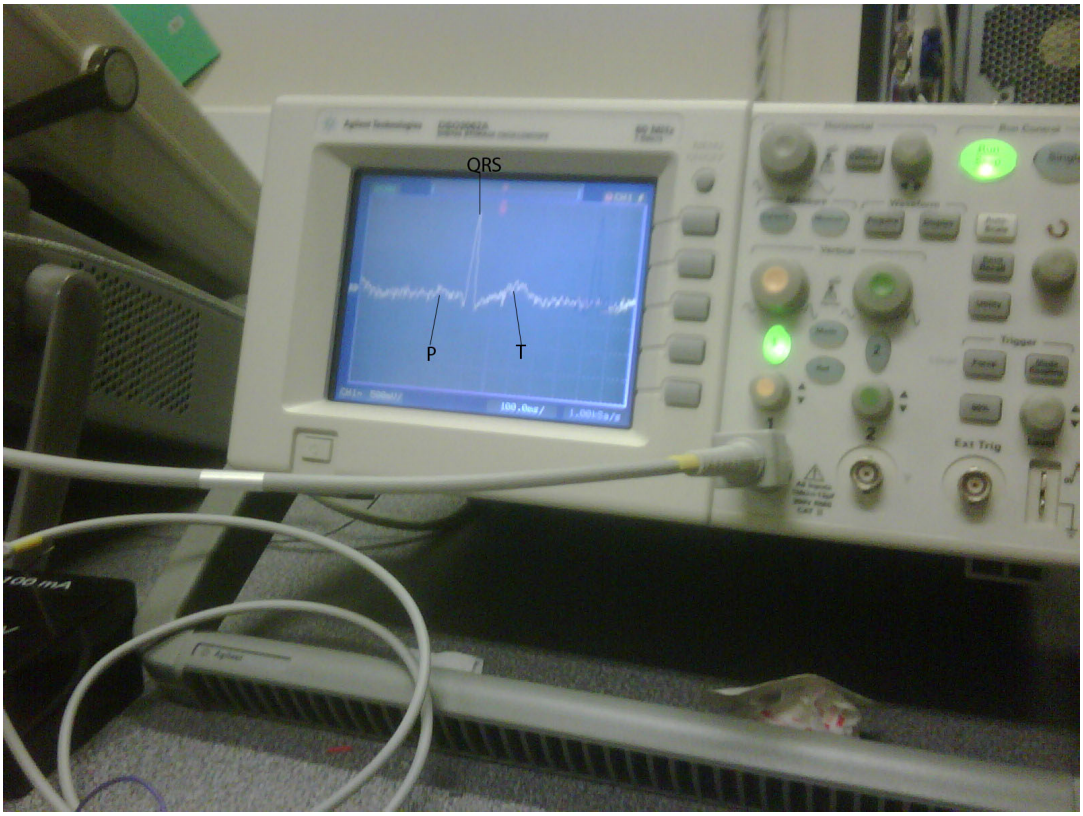


Figure 5.3: One instance of ECG signal

As one can see the signal of the ECG analyzer in figure 5.3 is producing an ECG in the form of figure 5.1. The ECG signal belongs to me and the reason it appears noisy

is because the photo was taken by me when measuring the signal and so my movement has inputted some noise into the signal. In general all the ECG signals acquired from me were noisier then others taken from the other test subject and so it can be deduced that this signal being noisy is not from the hardware setup but the patient itself. Also the P wave appears non apparent but again this is from the patient and not the hardware. It can be seen from this image that my goals were achieved.

As the purpose of this project was to create an ECG Analyzer for classification of different heart diseases we needed to acquire an ECG signal of a test subject with an unhealthy heart. This task was very hard as most people with the heart diseases we were exploring are in the hospital and don't need to be bothered by being a test subject. To prove the concept that an ECG signal could be analyzed and classified we decided on making the ECG Analyzer classify and compare two ECG signals and when given one as an input it would be able to tell which one it was. As stated above the test subjects were two young healthy males and their ECG signals can be seen in figure 5.4 and figure 5.5. These signals were the raw data taken from the ECG Analyzer hardware by using lab view and their ADC attachment. As one can see both signals are very different which is to be expected, as no two hearts are the same but they both contain the required features. As stated above and can be seen by comparing both signals my ECG signal is in general nosier. The raw data collected from these signals was then used by my partner's program to successfully classify and identify each signal.

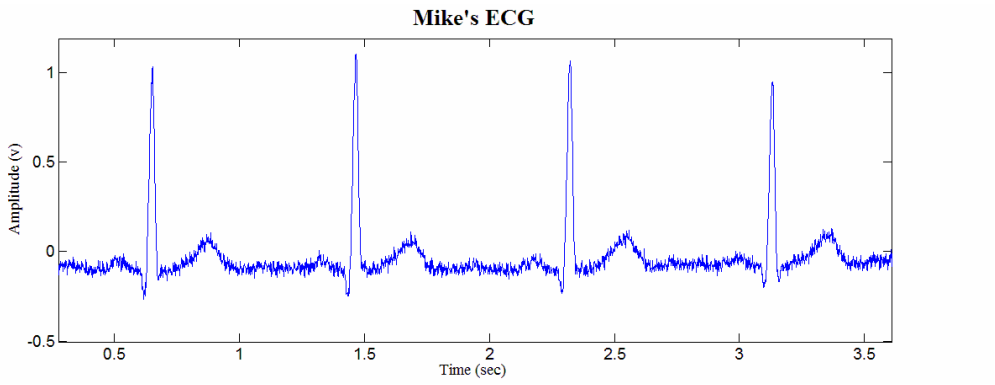


Figure 5.4: Test subject 1's ECG Signal

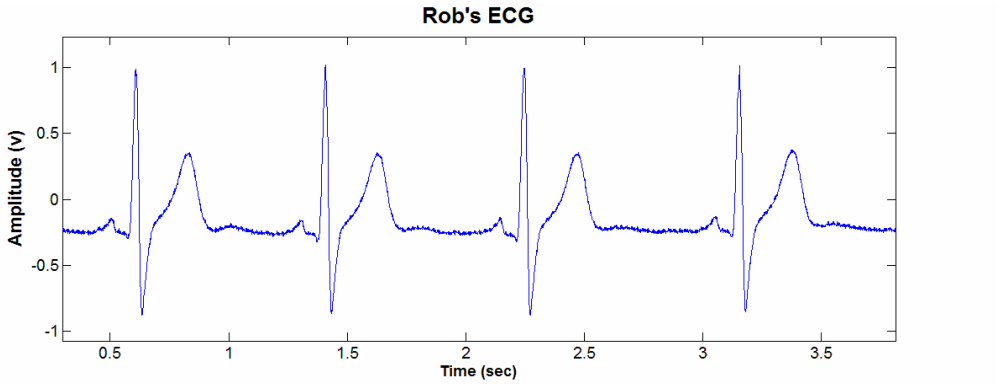


Figure 5.5: Test subject 2's ECG signal

It is also to be noted that we have a max voltage of 1 V, which proves that we are applying a large gain to the system to successfully amplify it. As these signal resemble the one in figure 5.1 it is safe to say that the filters are working and the isolation amplifier was able to send the signal across its barrier with no signal degradation.

My hardware was susceptible to some error. As stated before if the test subject was not still their movement would induce a noise in the signal and this can be better seen in figure 5.6. Also if the ground is attached in the wrong position the signal will become inverted and this is undesirable as my partners program cannot identify this and an incorrect identification will be made. This can be seen in figure 5.7. This problem has a simple solution as long as the person attaching the ECG Analyzer insured they followed the correct lead setup outlined in the report this problem will be avoided. Lastly the finally error viewed is when one of the lead wire becomes detached the signal becomes unusable. This can be seen in figure 5.8. Even though the wires are hard to detach if the patient is rough with the leads there is a possibility they can become detached. To ensure that this is less likely to occur medical tape can be attached to the top of the lead and electrode connection to hold them in place. As one can see these errors are due to patient interaction and human error and cannot be avoided.

It can be seen from these results the hardware implementation of the ECG Analyzer was successful.

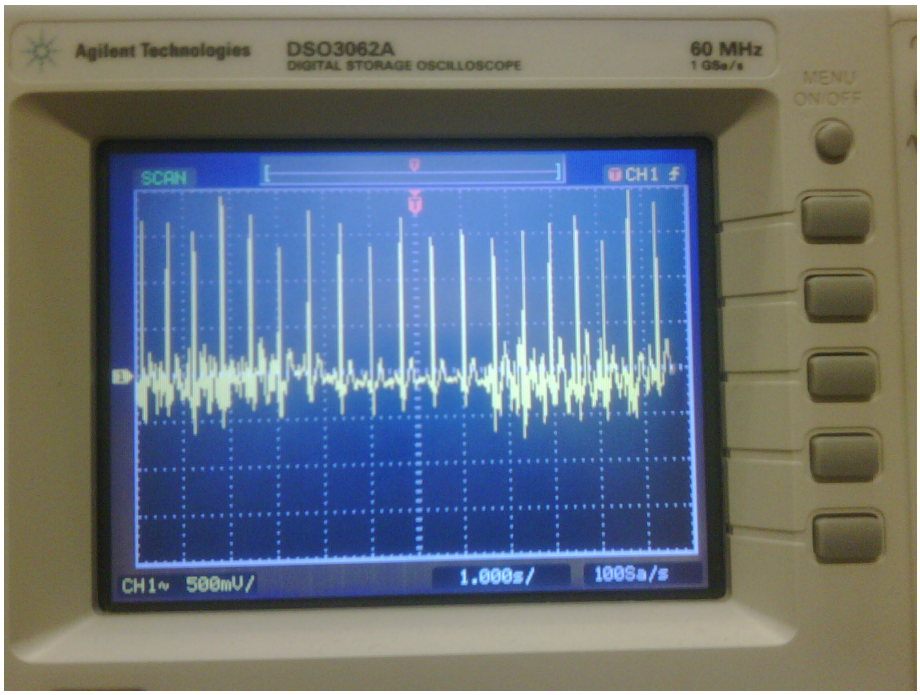


Figure 5.6: noise due to motion

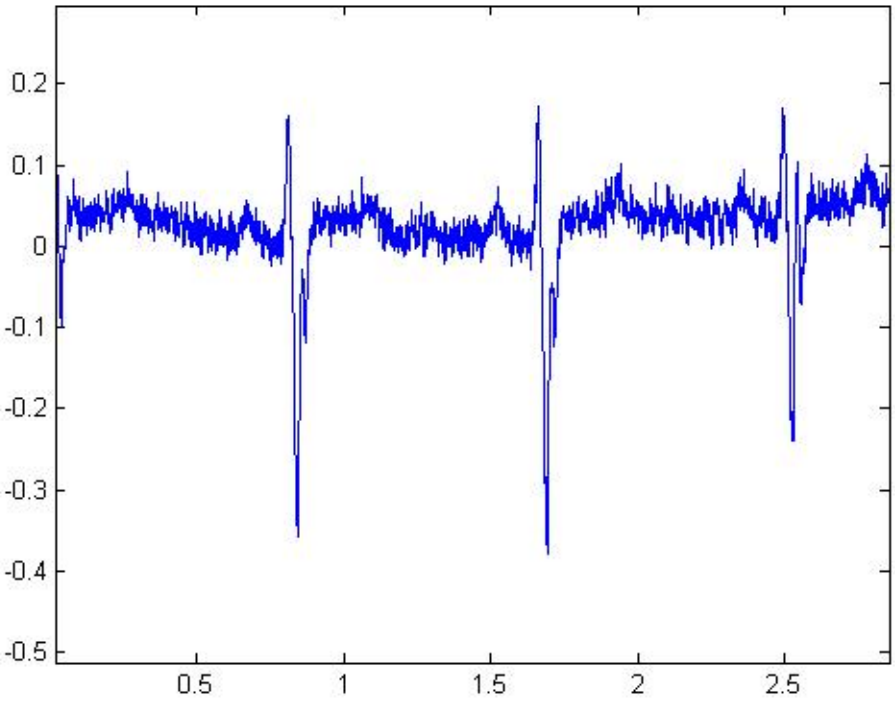


Figure 5.7: Inversion of ECG Signal due to Incorrect ECG Placement

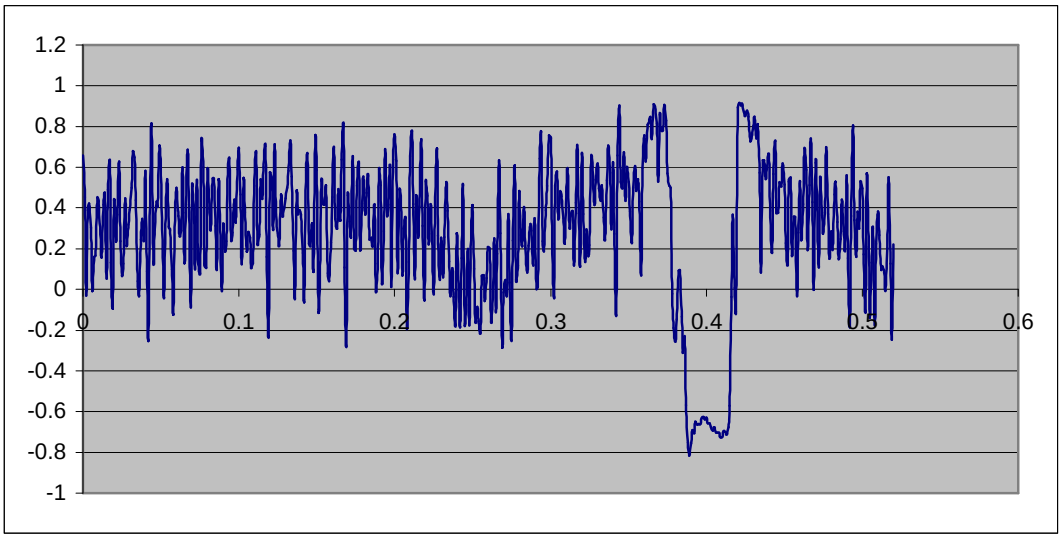


Figure 5.8: Un-useful Signal from Displacement of ECG Lead

Chapter 6

Conclusion and Recommendations

6.1 Conclusion

The ECG analyzer has been successful in achieving the major objectives set out at the beginning of the project. The Analyzer is able to acquire signals from the patients skin with specifically placed electrodes. The signal is then amplified to a useful range for my partner's software to analyze it. A high pass filter is implemented to eliminate dc and low frequencies signals that are considered to be noise in the ECG signal. As the ECG analyzer allows for signals above 60 Hz to pass a notch filter is successfully used to eliminate the line noise. Lastly meeting medical standards the ECG Analyzer eliminates frequencies above 150 Hz with the use of a high pass filter. Using minimal amount of parts and ensuring they were low cost allowed for the ECG Analyzer to be affordable for any country. This was done without lowering the standards for correct ECG signal reproduction.

6.1 Recommendations

As with most projects undertaken there is always room for improvement. The first improvement I would like to make is to eliminate the need for a computer to run the software. It was our intention as a group to implement the software on a micro controller but with problems with the micro controller we chose to use we were unable to complete this task. If the micro controller was implemented the other improvement that I would like to have made is for the ECG signal and our diagnosis to appear on a small LED screen. Lastly the patient must be still when attached to the analyzer. To allow for mobility of the patient I would have liked to have added into

the circuit a block that subtracts the EMG signal from the ECG eliminating the noise from movement.

Appendix A: Bessel Coefficients

Table 16–4. Bessel Coefficients

n	i	a_i	b_i	$k_i = f_{Ci} / f_C$	Q_i
1	1	1.0000	0.0000	1.000	—
2	1	1.3617	0.6180	1.000	0.58
3	1	0.7560	0.0000	1.323	—
	2	0.9996	0.4772	1.414	0.69
4	1	1.3397	0.4889	0.978	0.52
	2	0.7743	0.3890	1.797	0.81
5	1	0.6656	0.0000	1.502	—
	2	1.1402	0.4128	1.184	0.56
	3	0.6216	0.3245	2.138	0.92
6	1	1.2217	0.3887	1.063	0.51
	2	0.9686	0.3505	1.431	0.61
	3	0.5131	0.2756	2.447	1.02
7	1	0.5937	0.0000	1.648	—
	2	1.0944	0.3395	1.207	0.53
	3	0.8304	0.3011	1.695	0.66
	4	0.4332	0.2381	2.731	1.13
8	1	1.1112	0.3162	1.164	0.51
	2	0.9754	0.2979	1.381	0.56
	3	0.7202	0.2621	1.963	0.71
	4	0.3728	0.2087	2.992	1.23
9	1	0.5386	0.0000	1.857	—
	2	1.0244	0.2834	1.277	0.52
	3	0.8710	0.2636	1.574	0.59
	4	0.6320	0.2311	2.226	0.76
	5	0.3257	0.1854	3.237	1.32
10	1	1.0215	0.2650	1.264	0.50
	2	0.9393	0.2549	1.412	0.54
	3	0.7815	0.2351	1.780	0.62
	4	0.5604	0.2059	2.479	0.81
	5	0.2883	0.1665	3.466	1.42

Appendix B: Components List

Capacitors (nF)	**Amount**	
	10	2
	15	1
	33	1
	4.7	1
	1000	18
	56	2
	112	1
Capacitors with no polarity		and lowest tolerance

Resistors (kOhms)	
1	2
9	1
576	2
562	2
655	2
523	2
887	2
442	2
1780	2
100	1
23.75	1
47.5	1
2	1
1000	2
11	1
287	2
7.87	1
22.1	1
15	1

1 % tolerance

Amplifier	
TLC272ACP opamp	2
iso124 isolation amp	1
PGA204AP instrumentation amp	2

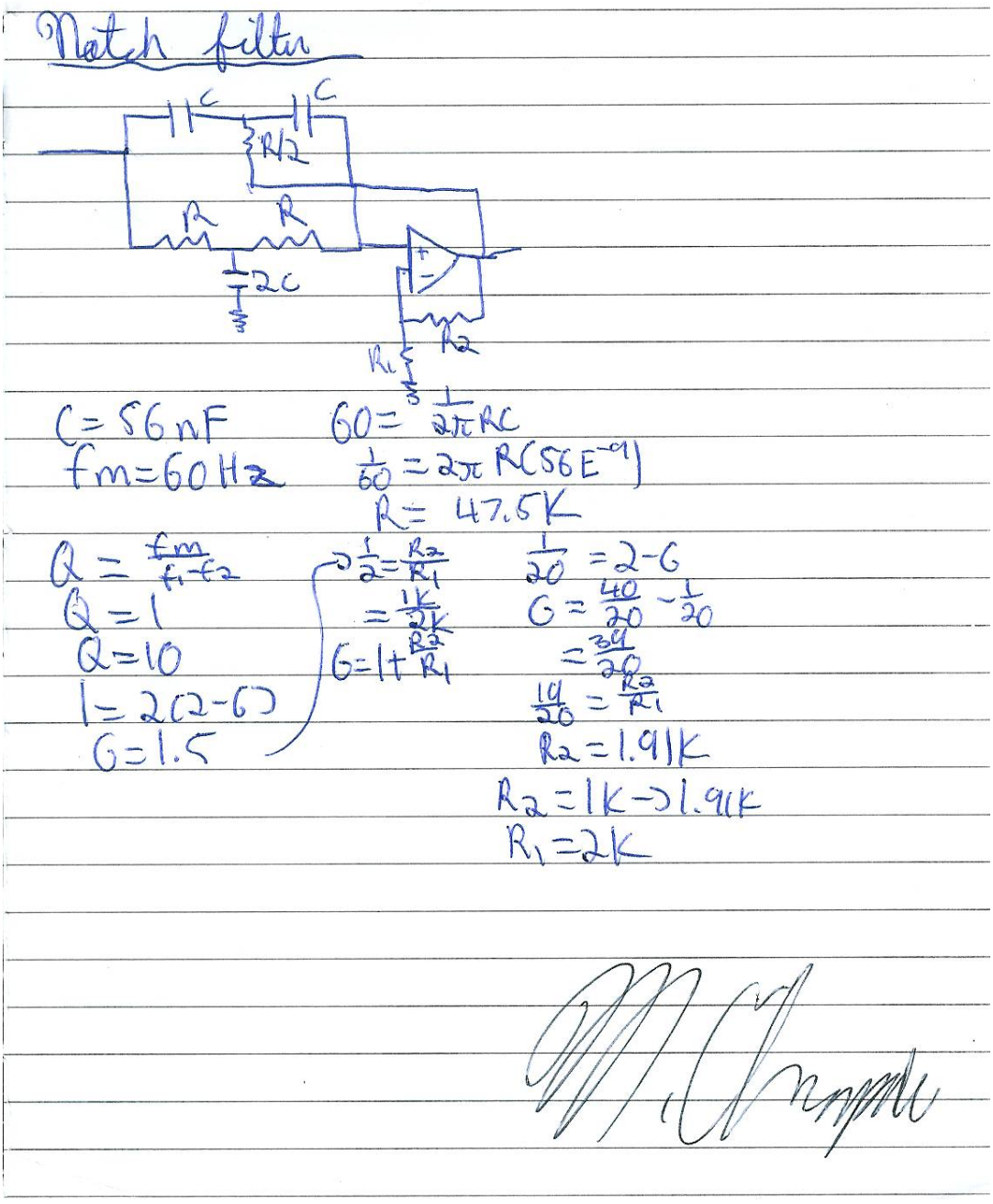
Appendix C: Lead placement explanation

There is a standard placement of electrodes when performing ECG recordings called a standard bipolar limb lead. A lead refers to the potential difference between two electrodes. The electrodes can be attached to the wrists and inner ankle, but for clinical applications are usually attached to the chest for a more accurate signal. Leads I, II, and III constitute the standard limb lead ECG. Using these three leads, we can form what is called Einthoven's Triangle. This is a representation of vectors demonstrating the formation of the ECG signal. In interpreting these measurements, each lead is assumed to be equivalent to measurements taken across all sides of an equilateral (Einthoven's) triangle. With Einthoven's Triangle, there is an equation that relates all three vectors. Einthoven's Law says that if the potentials of the first two leads are known, than the third lead can be found by adding the two vectors together. We only need to record two of the leads, as the third lead can be determined mathematically, provided that the two leads were measured simultaneously.

The ECG signal that is recorded can be derived from the Leads I-III vectors. When the ECG signal is recorded, the vector values for each of the leads changes as the atria and then ventricles contract. For example, as the QRS wave occurs, the lead I vector has a very small magnitude. This describes the slow upward growth of the lead I ECG recording. As depolarization sweeps across the atria and into the ventricles, the lead I vector begins to increase, causing the fast growth in the lead I ECG signal that is typical of the QRS complex. Then, as more of the ventricles depolarize, the lead I vector starts becoming smaller since all of the ventricular muscle has become depolarized, causing the lead I vector to have zero or slightly negative magnitude, causing the negative slope of the ECG signal in lead I. A similar analysis can be

performed on the other leads and can also explain how repolarization sweeps across the heart when the T wave occurs.

Appendix D: Calculation examples



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