COMPARISON OF FILTER APPROXIMATION METHODS FOR ECG FILTERING – AS PER AHA RECOMMENDATIONS

A Thesis submitted in partial fulfillment of the

requirements for the award of degree of

Master of Engineering

in

Electronic Instrumentation and Control

THAPAR UNIVERSITY

Submitted by

Sachin Gupta
Regd. No.: 801051017

Under the Guidance of

Mr. Vishal P. Arora

Lecturer

Department of Electrical and Instrumentation Engineering

Thapar University

(Established under the section 3 of UGC act, 1956)

Patiala, 147004, Punjab, India

July 2012

DECLARATION

I hereby certify that the work which is being presented in the thesis entitled "**Comparison Of Filter Approximation Methods For ECG Filtering – As Per Aha Recommendations**" in partial fulfillment of award of degree of **Master of Engineering** in **Electronics Instrumentation and Control** submitted in Electrical and Instrumentation Engineering department, Thapar University, Patiala is an authentic record of my own work carried under the supervision of **Mr. Vishal P Arora**, Lecturer, Department of Electrical and Instrumentation Engineering, Thapar University, Patiala, Punjab.

Date: 12-July-2012

Sachin Gupta

Sachin Gupta
Regd. No.: 801051017

I hereby certify that the above statement made by the student is correct to the best of my knowledge and belief.

Date: 13-July-2012

Vishal

Mr. Vishal P. Arora
Lecturer
Department of Electrical and
Instrumentation Engineering
Thapar University, Patiala
Punjab

Countersigned By

S. Ghosh

Dr. Smarajit Ghosh
Head of Department
Department of Electrical & Instrumentation Engineering
Thapar University, Patiala
Punjab

Dr. S. K. Mohapatra
Dean of Academic Affairs
Thapar University, Patiala
Punjab

ACKNOWLEDGEMENT

I sincerely acknowledge the efforts of all those, whose contribution paved the path for the successful completion of this report.

This dissertation could not have been written without Mr. Vishal Preet Arora, Lecturer, EICED, Thapar University, who not only served as my supervisor but also encouraged me and acknowledged my work. I sincerely thank his selfless efforts to help me complete this thesis. His trust that he showed on me permitted me to undergo for the project in the field of Biomedical Engineering and also for his full co-operation and guidance throughout the project work.

I would like to acknowledge the cooperation and constant support of my family and friends.

I would also like to thank our head of the department, **Dr. Smarajit Ghosh** for his anticipated support during the course of study and thesis.

Sachin Gupta
Regd. No. 801051017

Abstract

Healthcare is one of the fastest growing sectors. Physicians use various parameters from a patient to diagnose the ailment for its treatment and cure. ECG is one of the most vital parameter for monitoring and diagnosing different types of diseases and disorders related to the heart. Till now various monitoring systems have been developed for this very reason. From the background study it has been seen that each system has its own specifications. Sometimes they use only a standard single order filter approximation method. Each system has its own frequency related specifications like frequency cut off and range selection. When analysed, their output is most of the time not satisfactory. So in this thesis, various filter approximation methods have been designed using the specifications which are provided by American Heart Association (AHA); a well known and prominent organisation of scientists in USA who underlines the various specifications of designing the ECG systems. After Analysis, the 1^{st} order High Pass filter, inverse chebyshev Low Pass Filter has been found to give better response. The Fast Fourier transform (FFT) analysis of output signals from filters have been done for feature extraction.

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LIST OF ABBREVIATIONS

AHA	American Heart Institute
AV	Atrio Ventricular
BAN	Body Area Network
ECG	Electro Cardiogram
EEG	Electro Enceplogram
FFT	Fast Fourier Transform
HPF	High Pass Filter
IA	Instrumentation Amplifier
LPF	Low Pass Filter
PPG	Photo Plythesmogram
SA	Sino Atrial

Chapter 1

INTRODUCTION

1.1 Brief History of Monitoring System

Monitoring of physiological signals like ECG, PPG, SPo2, EEG, Blood Pressure, and Heart Rate etc became the essential part of biomedical science from several decades. Since the time doctors and medical technologists understood the relation of physiological signals with the diagnosis of patient's health, the monitoring systems have evolved to help us in diagnose and analyse the physiological changes in patient's body using the depiction of various physiological signals. It is a system built around hardware and software which uses complex algorithms to extract and display the signals in their analogous form [1].

Till now there is an Evolutionary growth in monitoring devices. In 1950's the monitoring devices had a CRT with appropriate analog circuits to measure physiological parameters and then show them on CRT screen. Here a Cardioscope module is shown in Figure 1.1 which was used to monitor ECG, and Heart rate [2].

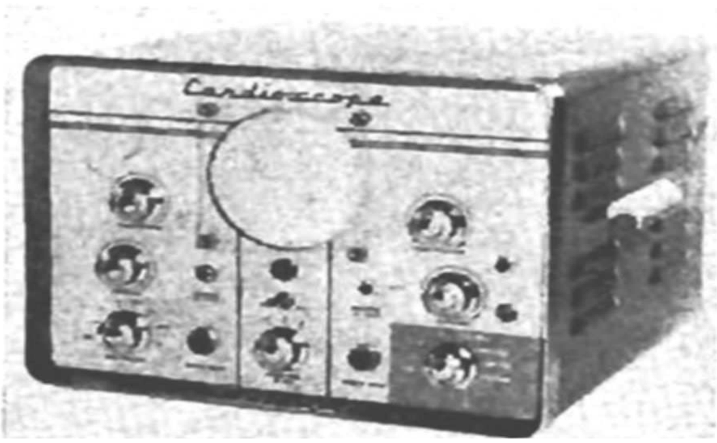


Figure 1.1: A Cardioscope Monitoring module

With the advancement, in 1970's there was a significant improvements in the presentation of the vital parameters and information on the monitoring screen. This was a result of the incorporation of digital electronics and, eventually, microprocessors. Real time (centrally controlled and processed) arrhythmia analysis machines appeared at the beginning of the decade and evolved over the duration of the 1970's. After that various Standards, and recommendations was established for improved performance criteria for physiologic monitor systems began appearing from several bodies such as the "Specification for Biomedical Monitoring Systems" X-1414 of 1970 from the Veterans Administration, American Heart Association. Now due to advancement in Digital Electronics world, there was possible to install memory elements to store real time monitoring data. Even due to this Holter monitoring devices comes in to existence. The advantage of this scheme was that the stored data could be written to the screen quickly and repeatedly at the same location well before it would fade out resulting in a “non-fade” persistent display. Waveform data was presented either as a moving wave that scrolls from right to left or as a stationary waveform that is overwritten by the trailing left edge of a narrow "eraser bar" which travels from left to right. Modern monitors use one display method or they permit selection of either of these modes of display. Some memory monitors also allowed data to be cascaded. This meant that an ECG trace would continue on the line below thereby providing more visible information over a greater time period. The display of most memory monitors could also be stopped or “frozen” for review of the appearance of a particular waveform. Also, printing or viewing of several seconds preceding an alarm condition was possible with some systems. Here in Figure 1.2, fully equipped modern digital monitoring system is shown.

Figure 1.2: A Modern Digital Monitoring System

1.2 Background

In the field of Biomedical Engineering, measuring bio signals like Bioelectric signal, Biomechanical signal, Biomagnetic signal, Biochemical signal etc. from human body is a customary and efficacious way to diagnose the diseases and condition of a patient. As ECG signal is the most vital & essential parameter for a doctor and heart specialist to get familiar with condition of a patient. Several types of instruments and devices have been developed till now to measure ECG signal. Measuring ECG signal is a non-invasive way, a basic ECG Signal acquisition system is shown in Figure 1.3 [1].

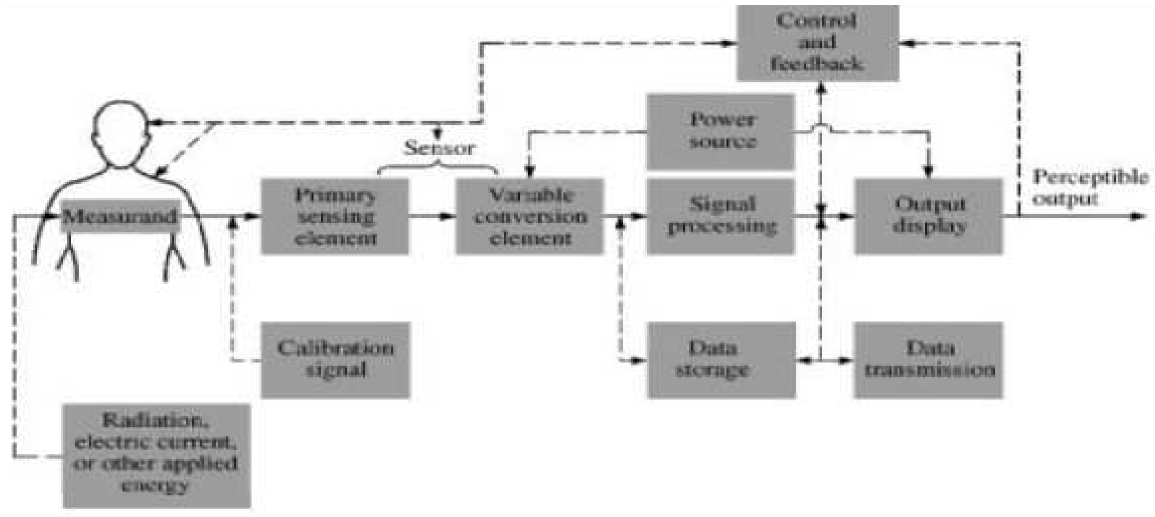


Figure1.3: A Basic Approach to Measure ECG

There is rapid growth in system development to measure ECG signal using wired and wireless approach. So there are several existing systems are explained below.

The Figure 1.4 presents a complete basic design methodology for ECG signal measurement and monitoring system. In general, the ECG system comprises four stages which are as following [3]:

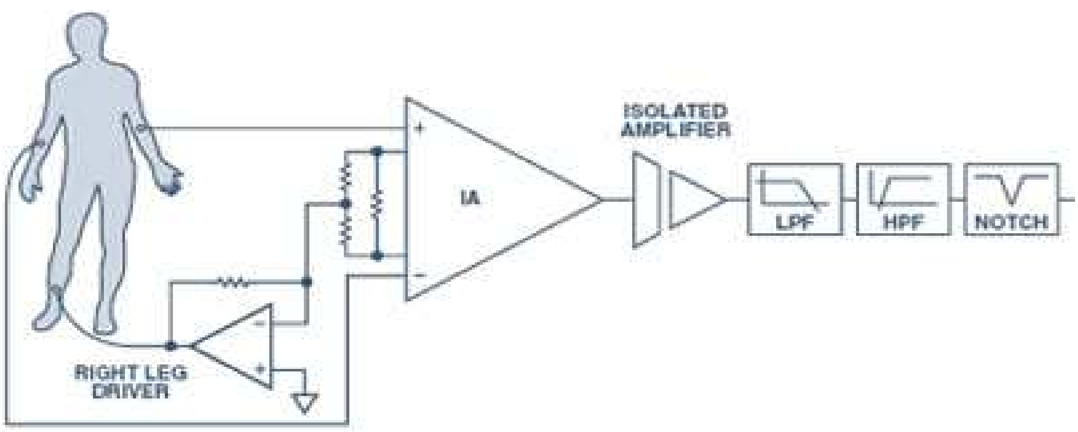


Figure1.4: Basic Stages of ECG Signal Acquisition System

(1) The first stage consists of a transducer (Electrode) like $AgCl^-$ electrode, which convert ECG into electrical voltage when it comes in contact with Skin. The voltage is in the range of 1 mV ~ 5 mV.

(2) The second stage is an instrumentation amplifier, which has a very high CMRR of minimum 90dB to remove noise and artifacts while ECG signal acquisition and high gain of upto 1000 to amplifies the weak signal.

(3) The third stage is Filter section. These filters are active filters which consist of filters in a series. First filter is LPF of 150Hz to cover QRS segment. Second filter is HPF of 0.5Hz to 1Hz to remove motion artifacts. Third filter is Notch filter to remove 50Hz Noise and RF noise.

(4) The fourth stage is monitoring and Recording section. Generally a Digital Controlled oscillator with strip chart is used for monitoring. For recording purpose, a mechanism is used in which a pen is derived with a driver to trace ECG signal on a graph paper.

Ovidiu Apostu, Bogdan Hagui et al. [4] this paper presents the development of a system for wireless ECG monitoring and alarm using Zigbee. The system is intended for home use by patients that are not in a critical condition but need to be constant or periodically monitored by clinicians or family. Patient monitoring is the cornerstone of proper medical care. It provides clinicians the much needed information about a person's current health status, so that they can act accordingly if anything goes wrong. Nowadays, complex patient monitoring systems offer the possibility of continuously monitoring of biological signals, analyze them, interpret them and take the required action for treatment. This system measures, records in real-time the electrical activity of the heart while preserving comfort of the patient. As this project was just developed for only monitoring purpose, this is a single lead ECG measuring system. The filters are designed using 0.5 to 35 Hz. The Zigbee wireless technology is used to transmit the signal because it consumes less power.

E Monton, J.F Hernandez et al. [5] in this paper they described that, Patient data monitoring is a key issue for health and disease management. The use of wireless sensors within a body area network (BAN) makes this task seamless and easy. A BAN system for ECG measurement is presented, which allows the connectivity of a wide range of heterogeneous body sensors to a portable hub device that is connectable to external networks. This BAN is based on the use of Zigbee/IEEE 802.15.4 standard technology and off-the-shelf modules. Characterised by its low power consumption, low cost, and ability to connect a wide range of heterogeneous sensors, this system can substantially improve the performance of different services, especially those that are health related. In this system the bandwidth for ECG filter is .0001 to 90 Hz. The BAN presented here has been designed and developed to fulfil two major requirements: usability and comfort, and the capacity to integrate a great variety of sensors. The system is to be used by people carrying out their daily tasks in their normal environment.

Alvaro Alesanco and Jose Garcra [6] This paper presents a complete study of wide-area wireless ECG transmission for real-time cardiac tele-monitoring taking into account both technical and clinical aspects, in order to provide recommendations for real-time monitoring considering both channel parameters and the tolerance of cardiologists to the effects of interruptions introduced during transmission. By using extensive wireless simulated scenarios, the compressed ECG signal is monitored on reception. A new protocol [real-time ECG transmission protocol, reliable ECG transmission protocol is used to perform the retransmissions of erroneous packets, introducing a monitoring buffer that mitigates possible negative effects. Assessments by cardiologists have shown that the maximum percentage of time for which the monitoring process could be stopped without their feeling uncomfortable is around 15% with a maximum monitoring delay of 3 or 4 s, depending on the scenario in question. Taking into account these values and the results obtained in the simulations, it is a straightforward step to obtain working areas for the wireless channel parameters where transmission is not recommended from a clinical point of view.

Dina Simmunic, Slaven Tomac et al. [7] The objective of the work was to make a simple wireless one channel limited (3-lead) home electrocardiogram (ECG) transmission system for home and ambulance use. The wireless ECG monitoring system significantly improves the quality of life of the cardiac patients, reflected primarily in the permanent monitoring. In case of an accident, an immediate alarm is being transmitted to the physician. A long-term monitoring facilitates the capturing of sporadic events and therefore is an important contribution for the improvement of the therapy and, consequently, for the health of the patients. The task has been accomplished by Bluetooth technology, ECG detector and personal computer as monitor. For Signal acquisition standard single order filters are used with the frequency range of 0.05 Hz to 100 Hz.

Thaddeus R. F. Fulford-Jones, Gu-Yeon Wei et al. [8] in this paper they told that , Sensor devices ("motes") which integrate an embedded microprocessor, low-power radio and a limited amount of storage have the potential to significantly enhance the provision of emergency medical care. Wearable vital sign sensors can wirelessly monitor patient condition, alerting healthcare providers to changes in status while simultaneously delivering data to a backend archival system for longer-term storage. As part of the Code Blue initiative at Harvard University, they previously developed a mote-based pulse-oximetry module which gathers data from a non-invasive finger sensor and transmits it wirelessly to a base station. To expand the capabilities of the mote for healthcare applications, they introduced ECG on Mica2, the first custom-designed electrocardiograph sensor board to interface with this platform. They additionally present Vital ECG, a collection of software components which allow the capture and wireless transmission of heart activity traces. They present preliminary test results which validate their approach and suggest the feasibility of future enhancements. They use standard filter for ECG monitoring with multiple feedback configuration. INA 321 is used as difference amplifier.

M Oehler, V Ling, K Melhorn et al**.** [9] in this paper they described that, Capacitive sensors can be employed for measuring the electrocardiogram of a human heart without electric contact with the skin. This configuration avoids contact problems experienced by conventional electrocardiography. In this study, they integrated these capacitive electrocardiogram electrodes in a 15-sensor array and combined this array with a tablet personal computer with standard analog filters. By placing the system on the patient's body, they measure a 15-channel electrocardiogram even through clothes and without any preparation. The goal of this development was to provide a new diagnostic tool that offers the user a reproducible, easy access to a fast and spatially resolved diagnostic 'heart view'.

1.3 Need of the Project

After the study of existing monitoring systems, it has been observed that they used only standard single order filters. As monitoring devices are made by various organisations, each organisation has set different frequency cut-off for filter section. Their frequency response is not suitable according to the specification and standards given by various medical bodies. So there must be a system which should be developed to improve these shortcomings.

1.4 Objectives

The objective of our thesis is to design filter circuits according to recommendations given by various standard organization to improve the filtering quality in ECG acquisition section of monitoring System.

1.5 Technical Focus

To achieve this goal we developed various active filter circuits and then tested.

1. To analyse their response for given ECG signal.
2. Comparing then to choose the best filter configuration.
3. Integrate them to build a complete system.
4. To get the FFT analysis for feature Extraction.

1.6 Problem Statement

1.6.1 Cost Effective: The cost of system should be low. So it can be easily affordable and implemented.

1.6.2 Low Power System: To have low power system, consideration will be given to have component with lower current required.

1.7 Tools Required:

To accomplish the project we used hardware and software tools.

1.7.1 Hardware: In hardware (Components), an Instrumentation Amplifier AD8429 is used. Active filters are designed using OPA355 operational amplifier. Capacitors and resistors are used with minimum tolerance value.

1.7.2 Software: In software we use MULTISIM 11.0.1 workbench to design and test the amplifier and active filter circuits. Labview 8.6 is used for signal Processing (FFT analysis)

Chapter 2

LITERATURE REVIEW

2.1 The Heart

Heart is a muscular organ that pumps blood to the body. Heart is at the center of circulatory system. This system consists of a network of blood vessels, such as arteries, veins, and capillaries. These blood vessels carry blood to and from all areas of human body. Regarding the action of heart, an electrical system controls heart's cambers and valves and uses electrical signals to contract the heart's walls. Blood is pumped into the circulatory system when the walls are contracted. Then Inlet and outlet valves in heart chambers ensure that blood is flowing in the right direction. The electricity which helps the heart to work is originated due to the biopotential in the cells of human body [10].

2.1.1 Biopotential

Human body is made up from tissues and tissues are made up from cells. Heart muscle, muscles of arms and legs are the examples of tissues. Cells are connected to each other through an axon, and dendrites as shown in Figure 2.1.

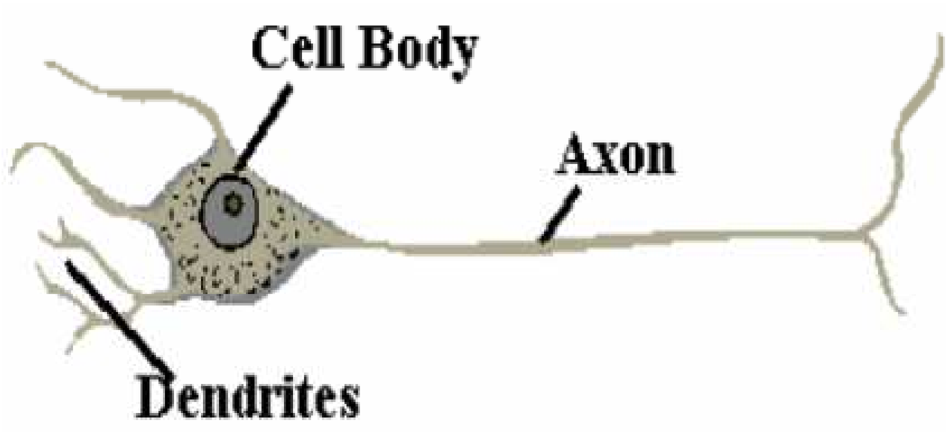


Figure 2.1: Interconnection of Cells through Axon and Dendrites

The cells have ions and electrical potentials are created by the flow of ions (Na+, Ka+, Ca+ and Cl^-) in and out of cells due to the change in ionic concentration inside and outside the cell membrane. The flow of these charged ions creates potential differences between the inside and outside of cells (-65mV to +40mV). These potential differences are called bioelectric potentials or action potential. The cell body contains the nucleus or command centre of the cell, the axon, which is responsible for transmitting the action potential along the cell, and the dendrites, which are responsible for receiving inputs to the cell in the form of neuro-transmitters. Nerve and muscle cells in the body communicate with each other via action potentials. Action potentials are voltage impulses that propagate along a nerve or muscle and may cause neurotransmitter release when the action potential reaches a specific area of the nerve cell [11]. A typical action bio potential recorded from a muscle is shown in figure 2.2

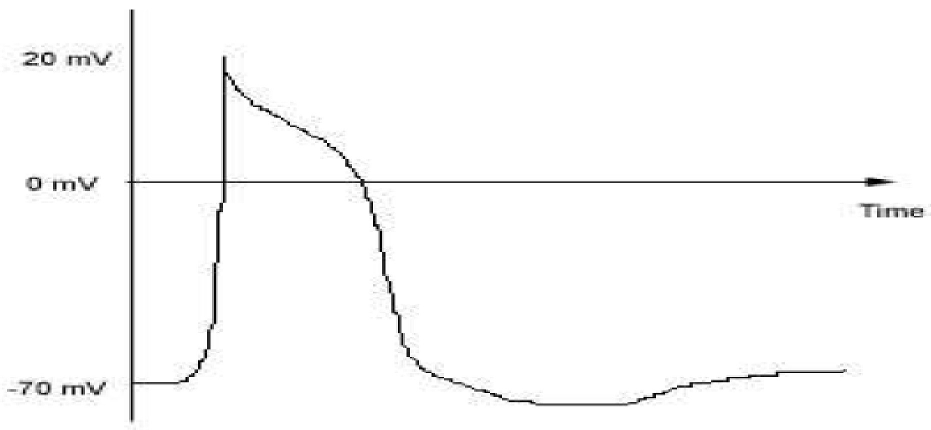


Figure 2.2: A Recorded Bioelectric Potential Voltage

2.1.2 Depolarization and Repolarization

To understand the cardiac signals, a brief description of anatomy and working of heart is essential. The heart is divided into four chambers (Figure 2.3). The two upper chambers, the left and right atria, are synchronized to act together. Similarly, the two lower chambers, the ventricles, operate together.

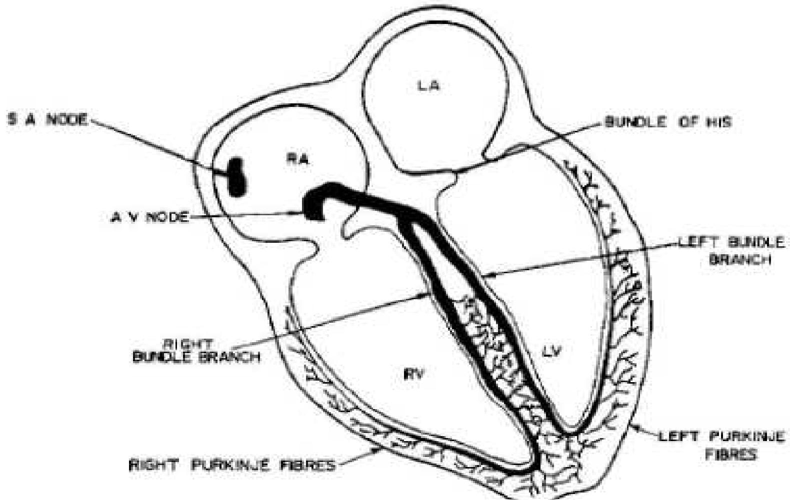


Figure 2.3: Heart Chambers

The right atrium receives blood from the veins of the body and pumps it into the right ventricle. The right ventricle pumps the blood through the lungs where it is oxygenated. The oxygen-enriched blood then enters the left atrium, from which, it is pumped into the left ventricle. The left ventricle pumps the blood into the arteries to circulate throughout the body. The contraction of any muscle requires electrical stimulation. This is true for the different cardiac muscles of the heart too. The electrical signal for these contractions originates near the top of the right atrium at point called sinoatrial (SA) node. It is also called the pacemaker and is a group of specialized cells that spontaneously generate action potentials at regular rate. To initiate the heart beat, the action potentials generated by the SA node propagate in all directions along the surface of atria. The wave terminates at a point near the center of the heart called the atrio-ventricular (AV) node. At this point, some special fibers act as a “delay line” to provide proper timing between the action of the atria and ventricles. Once the electrical excitation has passed through the delay line, it rapidly spreads to all parts of both the ventricles by the bundle of His. The fibers in this bundle, called Purkinje fibers, divide into two branches to initiate action potentials simultaneously in the powerful musculature of the two ventricles. The ventricles relax thereafter, to complete one cycle. This cycle is repeated over and over again. The simplest of all the signals that can be used to detect any malfunctioning of heart is its sound [12] [13]. It can be heard using a plain stethoscope or can be recorded in the form of a trace on the paper by phonocardiogram. The traces are shown in Figure 2.4

Figure 2.4: Normal and Murmur Sound from Heart

2.2 ECG

ECG (Electrocardiogram) is the most vital parameter related to Heart to diagnose and analyse the condition of a patient. The method to measure ECG was discovered by Dr. Willem Einthoven. ECG is basically measuring and recording of electrical impulses, generated from heart on a graph paper [14]. A regular pattern of signals originate on the graph paper. From this graph paper various parameters are measured like

- The rate and regularity of heartbeats as well as the size and position of the chambers.
- The presence of any damage to the heart.
- The effects of drugs or devices used to regulate the heart.

2.2.1 Einthoven Triangle and 3 lead ECG

An initial breakthrough came when Willem Einthoven, working in Leiden, Netherlands, used the string galvanometer that he invented in 1903. This device was much more sensitive than both the capillary electrometer that was earlier used. Rather than using today's self-adhesive electrodes Einthoven's subjects would immerse each of their limbs into containers of salt solutions from which the ECG was recorded. The earlier machine is shown in Figure 2.5

Figure 2.5: Dr. Einthoven's ECG Machine

According to Einthoven's law heart's electrical activity represents the vector quantity. The electrical signal generated by the heart appears throughout the body and on skin surface. Here is a dipole movement of electrical signal. There is a potential difference between two electrodes. This potential signal is measured by placing electrodes on the heart surface. Different electrodes are placed on the body surface to measure potential difference. Different electrode pair gives different voltages because of the spatial dependence of the electric field of the heart. Basically we need many electrode pairs to cover the hearts vector views like frontal or lateral view, posterior view, inferior view or transverse view. Three basic leads make the frontal plane. The three lead vector triangle is shown in Figure 2.6

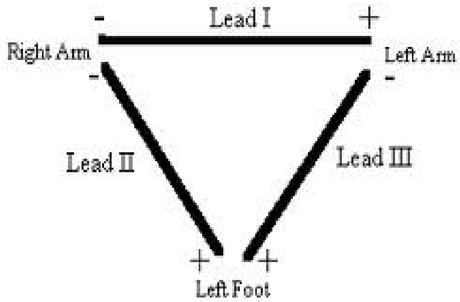


Figure 2.6: 3 Lead Vectors

In ECG one lead is equal to combination of two electrodes. This is called bipolar leads. So, three electrodes are placed one on left leg, one on right arm, one on left arm. Very often one electrode is connected to right leg and grounded or connected to special circuit [1][15]. Resultant lead pairs are shown in Table 2.1

LEAD	POSITON	VIEW	HEART CHAMBER
Lead 1	From RA to LA	Lateral	Left ventricle, left atrium
Lead 2	From RA to LL	Inferior	Left and right ventricle
Lead 3	From LA to LL	Posterior	Right and left ventricle

Table 2.1: Chamber View of Heart

This lead vector is called Einthoven's triangle. Now the scalar signal on each lead acts as a voltage source. We can write Kirchhoff's voltage law for three leads. Shown in equation

$$I-II+III=0 \quad (1)$$

$$I=II-III \quad (2)$$

$$II=I+III \quad (3)$$

$$III=I-II \quad (4)$$

Now we can calculate component of a particular cardiac vector according to this triangle and determine its projection along each side.

2.2.2 Signal Intervals and Bandwidth

As the sinoatrial node generates the stimulation to heart's different chambers, then ECG signal is occurred with different segments like amplitude levels and frequency contents as shown in Figure 2.7.

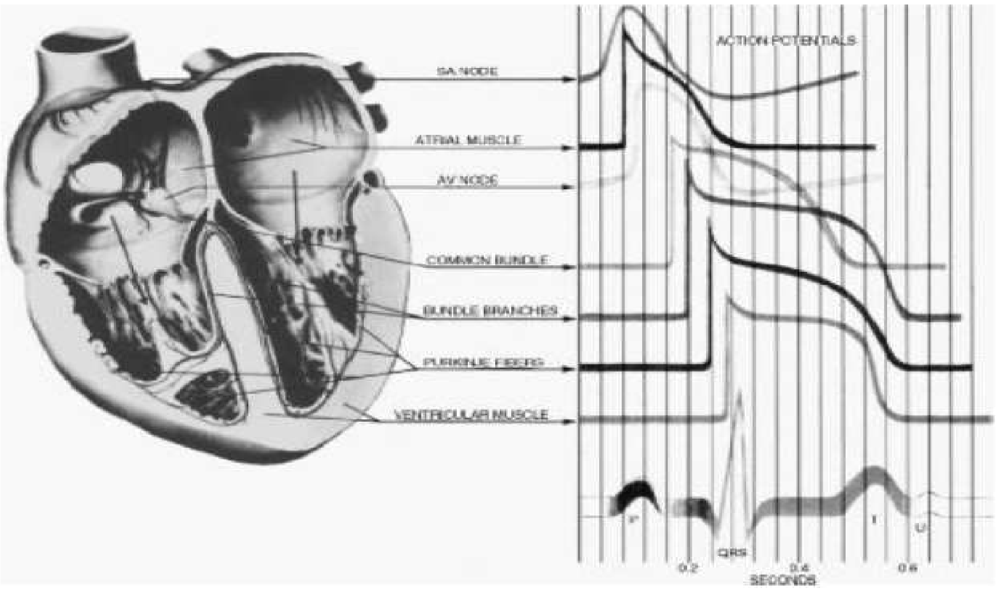
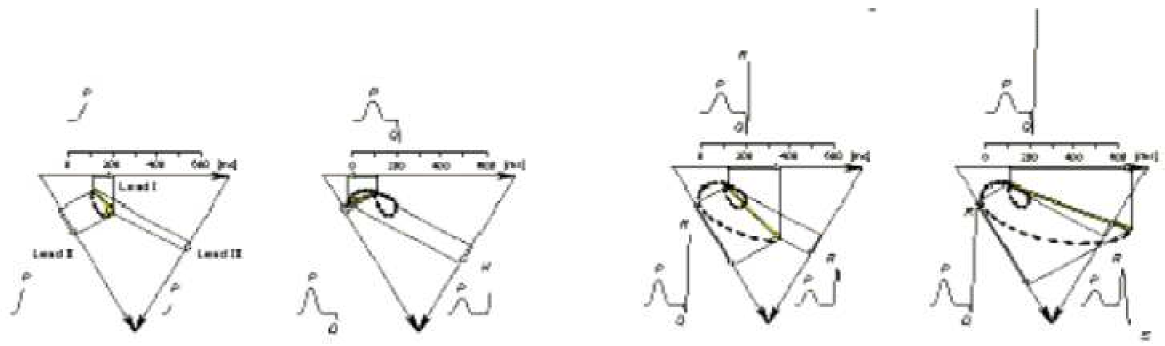


Figure 2.7: Different ECG Segments from Heart Chambers

As the electrical currents that are generated for depolarization and repolarization of heart spread throughout the body and can be measured by an array of electrodes placed on the body surface. The sequence of depolarization and repolarization of the atria and ventricles is represented by the different waves of the ECG. The wave of depolarization that spreads from the SA node throughout the atria is represented by the P wave and is usually 0.08 to 0.1 seconds (80-100 ms) in duration. Within the AV node the conduction velocity of the impulse is greatly retarded. The time taken by impulse here is represented by a brief isoelectric (zero voltage) period after the P wave. The P-R interval, which normally ranges from 0.12 to 0.20 seconds in duration, is the period of time from the onset of the P wave to the beginning of the QRS complex. This interval represents the time between the onset of atrial depolarization and the onset of ventricular depolarization. If the P-R interval is >0.2 sec, a conduction defect (usually within the AV node) is present (first-degree heart block). The ventricular depolarization is represented by QRS complex. The duration of the QRS complex is relatively very short (0.06 to 0.1 seconds) thereby indicating that ventricular depolarization normally occurs very rapidly. Prolonged QRS complex (> 0.1 sec), indicates impaired conduction within the ventricles. It can be occurred under two circumstances. One, with bundle branch blocks and second whenever a ventricular foci (abnormal pacemaker site) becomes the pacemaker driving the ventricle. Such an ectopic foci nearly always results in impulses being conducted over slower pathways within the heart, thereby increasing the time for depolarization and the duration of the QRS complex. ST segment is the isoelectric period following the QRS. It is the time at which the entire ventricle is depolarized and roughly corresponds to the plateau phase of the ventricular action potential. A depressed or elevated ST segment is important in the diagnosis of ventricular ischemia or hypoxia. The T wave represents ventricular repolarization and is longer in duration than depolarization, as the conduction of the repolarization wave is slower than the wave of depolarization. The Q-T interval represents the total time for ventricular depolarization and repolarization to occur and therefore roughly estimates the duration of an average ventricular action potential. Depending upon heart rate, this interval can range from 0.2 to 0.4 seconds. At high heart rates, ventricular action potentials shorten in duration, which decreases the Q-T interval. A relatively prolonged Q-T interval can be diagnostic for susceptibility to certain types of tachyarrhythmia. To know this, the Q-T interval is divided by the square root of the R-R interval (interval between two ventricular depolarizations) and is expressed as a "corrected QT (Q-Tc)". This makes assessment of the Q-T interval independent of heart rate. Normal corrected Q-Tc intervals are less than 0.44 seconds. Several ECG tracings are recorded simultaneously from different electrodes placed on the body to produce different characteristic waveforms. Different views of ECG signal with Heart pumping action is shown in Figure 2.8



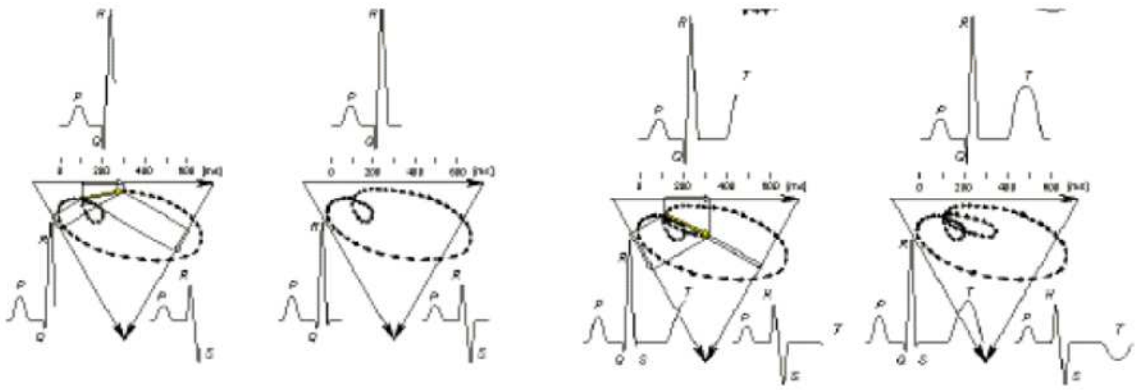


Figure 2.8: ECG signal with Heart Working Action

From the different time intervals, the bandwidth of ECG signal is from 0.5Hz (due to minimum Heart "beat/minute" measured previously in studies) to 250 or 500 Hz (Due to QRS intervals)[13]

2.2.3 Interference and Noise

The ECG signals are typically in the millivolt range, so they are more susceptible to large amounts of interference, from a variety of sources. The interference sources can be divided into 3 distinct groups [1] [11] [12].

- Noise originating from sources external to the patient
- Interference originating from the patient
- Unwanted Potentials as well as interference originating from patient-electrode contact.

2.2.3.1 External Noises

This noise is produced by the external sources. There are mainly two types of external types of noises which produce noise in the ECG signal**.** These are electrostatic and electromagnetic noise.

2.2.3.1.1 Electrostatic Sources

Regarding the electrostatic source, if a charged body is brought up close to an uncharged body, then equal & opposite charge develops on the uncharged body. For example if an unearthed body is close to any electronic device which is connected to the mains supply voltage, the body will create a surface charge of equal & opposite potential even though no current is flowing between the two bodies. This phenomenon is also known as Electrostatic Discharge. As the mains potential has a frequency of 50 Hz, the induced potential will also have this frequency. Other sources of electrostatic charge are operating table, other persons, electronic equipment during ECG measurement.

2.2.3.1.2 Electromagnetic Induction

Interference is occurring in the wires which are carrying AC currents, is due to the generation of a magnetic field by the flow of a current, all conductors carrying mains currents are surrounded by electromagnetic fields. As the main supply has 50 Hz frequency .The 50 Hz mains interference is a difference in potential, relative to ground, that is impressed upon any patient/subject in proximity to the wire carrying alternating (50Hz) main supply current. The patient takes on a potential that is neither that of ground, nor that of mains, but rather, somewhere in between. As the mains current is fluctuating (AC), the induced voltage in the subject is also fluctuating. This effect is minimized by the fact that the electromagnetic field generated by the live wire is to a large degree cancelled out by the neutral cable flowing adjacent to the live cable but it should be in opposite direction.

2.2.3.2 Internal Noises

Internal noises are generated due to naturally generated artefact. There are mainly two types of artefacts.

2.2.3.2.1 Originating from the Patient

The electrodes are placed on the patient’s body surface (when body is in rest). The signal is measured with less noise, because at that time there is no unwanted activation of axon and hence no generation of unwanted ions. If the body is in moving condition, then ions produce unwanted electric signal at body surface. Hence at the time of ECG measurement body should be in stationary condition.

2.2.3.2.1 Originating from Patient-Electrode Contact

The ECG electrodes are not like a passive non-invasive conductor. The placement of any metal adjacent to an electrolytic solution (gel on ECG pads combined with surface of skin) produces an electrochemical half-cell, similar to that of a battery, resulting in potentials on the surface of the skin. Hence, if a differential amplifier is connected to a pair of such electrodes it will amplify any difference in potentials. if the cells are identical then the output will be zero. If the potential difference is not same, the output will be non-zero. Then any difference between the two electrodes will be amplified. Additionally, the small current produced by the offset potential may result in polarisation. Polarisation of the electrode will further distort any signal. Mainly this noise is called half-cell noise. [16]

2.2.4 ECG Sensors

Mainly to measure ECG, electrodes are placed on human body's skin according to Einthoven's triangle method. Electrodes are like transducers which converts ionic current into electronic current in the electrode. Mainly electrodes are classified into two categories first is polarized and second in unpolarized. Polarized electrodes are very poor conductors of ionic current due to their composition. But unpolarized electrodes are highly conductive to ionic current. Hence they are widely used. Some times unpolarized electrodes provide over potential without any energy exchange. Hence silver-silver chloride electrode (Ag-Agcl) electrode is used because of minimum over potential in the electrode. There are various types of electrodes available to measure ECG waveform like metal plate electrode, Metal disc electrodes, Disposable form pad electrode, and Metallic suction electrode. Disposable electrodes and Metallic suction electrodes are widely used because their cost is less. Presently there are also Capacitive electrodes (active electrodes) are used for measurement purpose. Various types of sensor are shown in Figure 2.9, 2.10, 211.

Figure 2.9: Disposable Electrodes

Figure 2.10: Suction type Electrode

Figure 2.11: QUASER's Capacitive Electrode (Active)

Hence electrodes are placed on the chest or limbs according to the Dr. Einthoven's triangle law to measure potential difference between two points [1] [12] [17].

2.3 ECG Standards

There are various companies and organisations which are designing and developing biomedical devices for monitoring and diagnoses. Some times their specification and calibration parameters are different for their own devices. So there should be a common standard so that they choose common parameters. For ECG monitoring devices, the standards and recommendations are given by American heart association. The American Heart Association is a national, voluntary, non-profit health organization dedicated to reducing death and disability from heart disease and stroke. This organisation is a group of prominent scientists in USA which are dealing with cardiovascular science, research, standards and recommendations for medical devices related to ECG. This pioneering group of physicians and social workers formed the first Association for the Prevention and Relief of Heart Disease in New York City in 1915. Till now AHA recommended various standards for various biomedical devices [18].

2.3.1 AHA Recommendation

Based on the analysis of Diagnostic and recording ECG measuring devices. AHA introduced the standards and recommendations for ECG recording devices. Mostly ECG measurement devices choose frequency range from 0.5Hz (due to minimum Heart "beat/minute" measured previously in studies) to 250 or 500 Hz (Due to QRS intervals) for filtering purpose. But AHA recommended frequency from .05 Hz to 100 Hz. Because according to AHA T wave components have frequency range of .67 to 1 Hz. The 0.05 Hz cut off increases the baseline wandering, but it will capture the significant amplitude segments which help to diagnose various Cardiac Disorders. Baseline wandering can be recovered by digital filters. The QRS complex fundamental frequency is approximately near to 30 Hz and may vary up to 100 Hz in case of adults. Increasing the cut off to higher frequency will lead to more accurate representation of amplitude but also chances of distortion and phase nonlinearity may increase. Now AHA recommended various standards for ECG's preamplifier section. First the gain should be flat in pass band (1 to 30 Hz) with variation of ± 0.5db (it is acceptable) with minimum phase shift. It will reduce the displacement in the ST segment and duration of QRS will be maintained. Secondly the 3db point should be less than or equal to at .67 Hz and grater or equal to at 150 Hz. So that roll off will be achieved [18] [19] [20].

Chapter 3

ACTIVE FILTERS APPROXIMATION

3.1 Introduction to Filters

Filters are analog circuits those allow a desired band of frequencies to pass through while attenuating the rest of them, when a signal is applied at their input terminals. Basically filters are of four types [22] [23].

- **Low Pass Filter**: - Allows a band of frequencies ranging from zero to f_{pass} to pass through without any attenuation, where f_{stop} is the cut off frequency. The response of low pass filter is shown in Figure 3.1

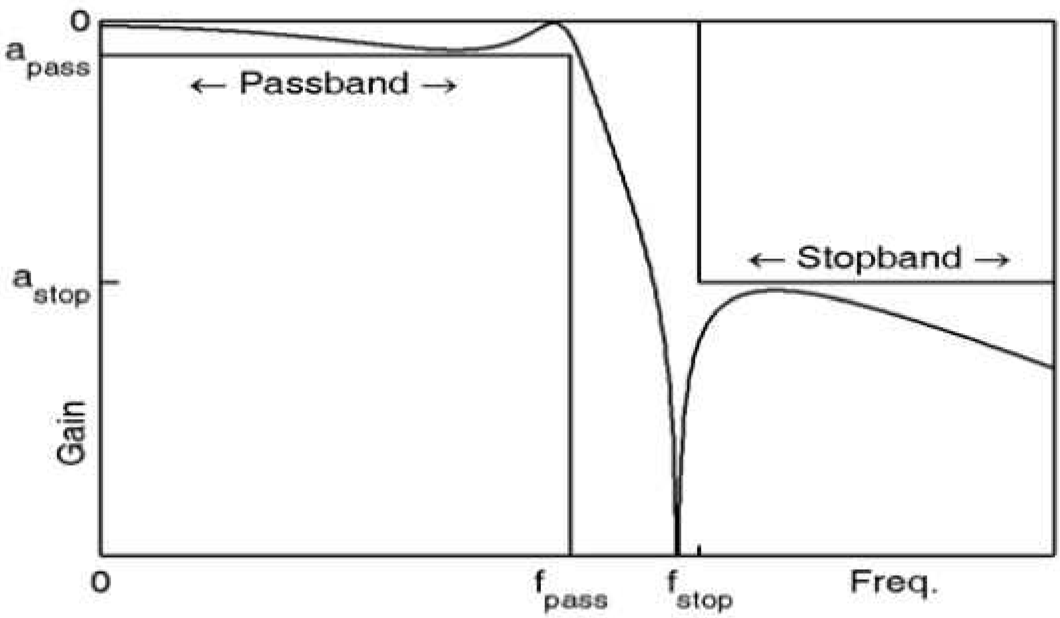


Figure 3.1: Low Pass Filter Response

- **High Pass Filter**: - Allows a band of frequencies ranging from f_{pass} to infinity to pass through without attenuation, f_{stop} being the cut off frequency. The response of High pass filter is shown in Figure 3.2

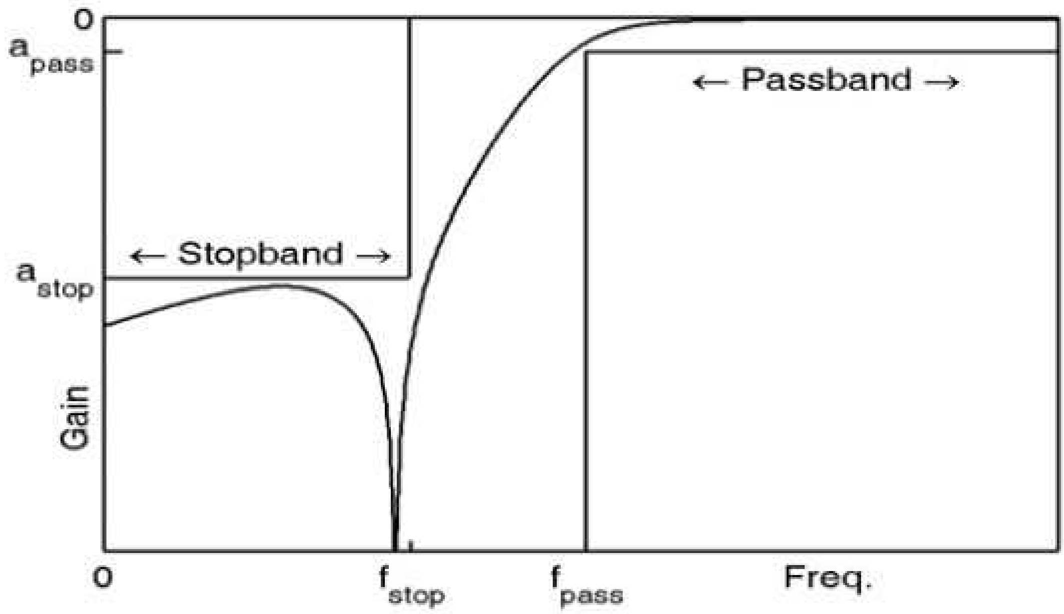


Figure 3.2: High Pass Filter Response

- **Band Pass Filter**: - Allows a band of frequencies ranging from $f_{pass\,1}$ to $f_{pass\,2}$ to pass through without attenuation where $f_{pass\,1}$the lower cut off is while $f_{pass\,2}$ is the higher cut off frequency. The response of Band Pass filter is shown in Figure 3.3

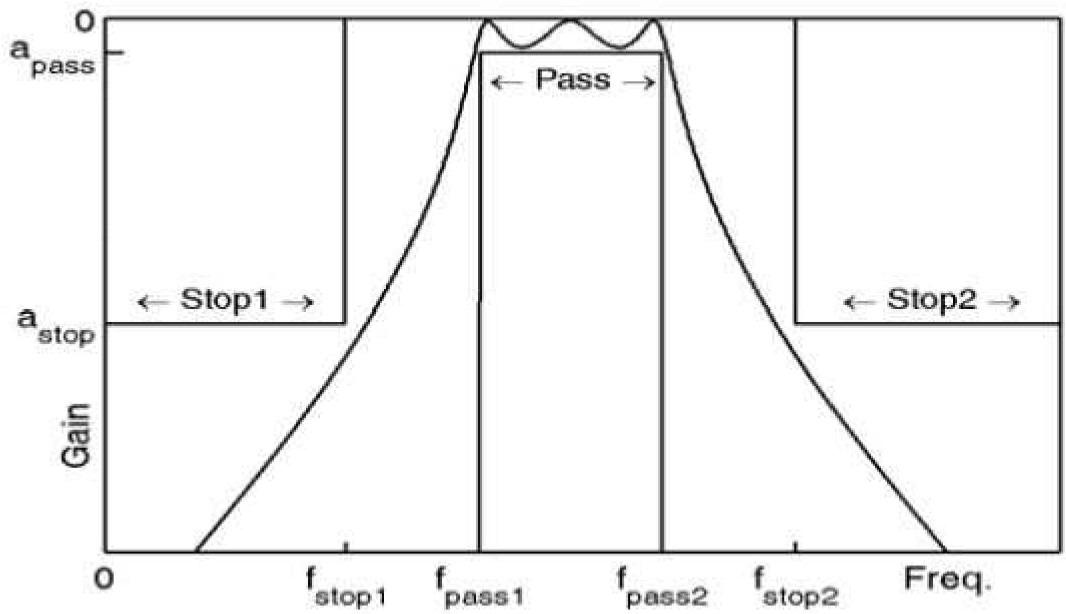


Figure 3.3: Band Pass Filter Response

- **Band Stop Filter**: - Attenuates a band of frequencies ranging from $f_{stop\,1}$ to $f_{stop\,2}$ while allowing the rest of the frequencies to pass through it. The response of Band stop filter is shown in Figure 3.4

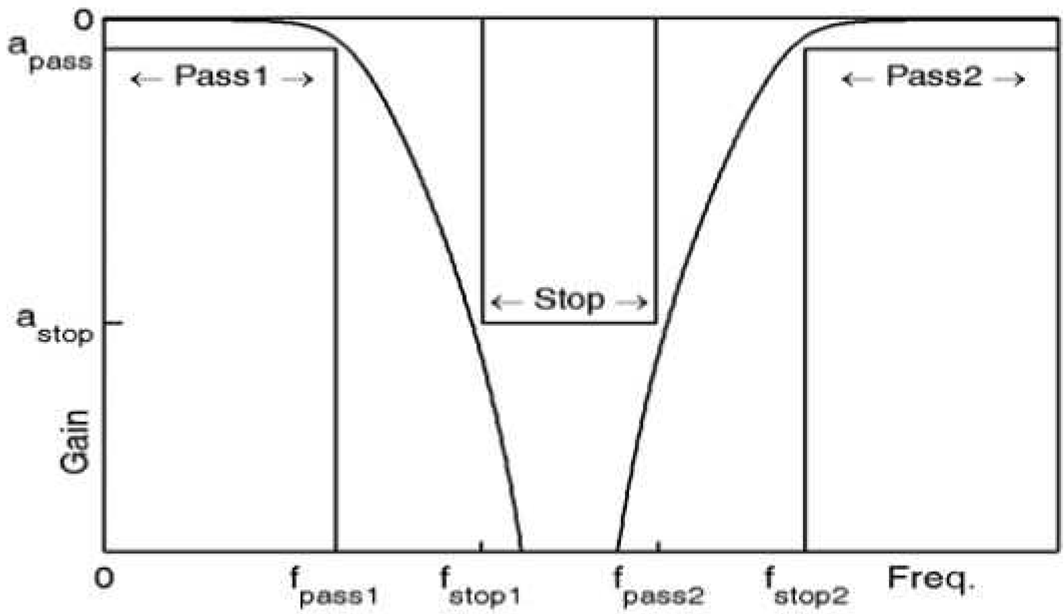


Figure 3.4: Band Stop Filter Response

Filters are of passive and active types. Passive filters consist of electrical elements like resistors, inductors, capacitors etc. combined in a circuit. This utilizes the frequency selective properties of inductors and capacitors for the purpose of filtering. The impedance of an inductor increases with increase in frequency of the signal while that of a capacitor decreases. Thus by using these elements in a circuit arrangement, frequency selective attenuation can be obtained. The cut-off values of the filter circuit depend upon the value of the circuit elements being used. Also the capacitive filters are preferred over the inductive filters because of the inductors being costlier and much more bulky. For low frequency signals active filters are preferred.

3.1.1 Active Filters

Active filter are analog by nature. These filters are built using the combination of resistances and capacitors with an amplifier (Operational filters). So these filters are called active filters. As a simple filter is made from passive components, the gain will be low at the output of filter due to attenuation. So an amplifier with these filters again raises the amplitude level of the signal. Also this amplifier prevents the load impedance of the following stages from affecting the characteristics of the filters.

When ECG signal is measured using electrodes, it contains various types of Noises and artefacts like Baseline wandering, Motion artefacts, Radio frequency Noise. Then this signal is passed from the filters to remove these noises and artefacts. As the response of ideal filter is unattainable so the best way is to approximate it. There are mainly four classifications for Active filters approximation which can be designed on two types of configurations (Topologies) one is Sallen key and second is multiple feedback type. The main disadvantage of multiple feedback configurations is that it gives 180° phase shift to signal, But Sallen key configuration doesn't introduce any phase shift to signal. So Sallen key configuration is an ideal configuration to design and approximate various active filters. The analog active filter approximation configurations are classified as [22] [23] [24].

- First order filters Approximation (A Standard Configuration)
- Butterworth Filter Approximation
- Chebyshev Filter Approximation
- Inverse Chebyshev Filter approximation

3.2 AHA Filter Specifications

American Heart Association recommended various specifications and standards for ECG Signal recording and measurement. According to AHA various parameters like HPF cut off frequency and LPF cut off frequency regarding the designing of analog active filters are given below[19][20][21].

- For High Pass Filter cut off frequency should be .05 Hz (To capture T wave components)
- For Low Pass Filter cut off frequency should be 100 Hz (For QRS Complex intervals)
- Gain should be flat from 1 Hz to 30 Hz (Variation ± 0.5 db)
- Phase shift should be minimum

3.3 Instrumentation Amplifier

The ECG signal was taken from the PhysioNet databank with sampling rate of 200Hz and digitisation of 12 bits. It is a single-lead ECG. Due to biopotential, the ECG signals are in the weak (mV) ranges and hence an instrumentation amplifier is used. An instrumentation amplifier is designed using two operational amplifiers with signals provided to V_+ of the op-amps. Instrumentation amplifier is also called difference amplifier because it's the difference between the two input signals which is actually amplified. A basic schematic of Instrumentation amplifier is shown in figure 3.5. Regarding the characteristics of instrumentation amplifier, its input impedance is very high so it is act like a input buffer. Its output impedance is very low to so that it can deliver full current to further stages. It has very low DC offset, low drift, low noise, and very high open-loop gain. It has high common-mode rejection ratio so that a common signal like noise is removed. The instrumentation amplifiers also provide necessary gain to a signal that it can further be amplified by voltage amplifiers. As a revolutionary growth is designing compact and small IC's, the instrument amplifiers are embedded in small IC package. With the required characteristics, there are various instrumentation amplifiers made by various companies in the market. But AD8429 IA made by Analog Devices is best when compared by other instrumentation amplifiers. The basic Instrumentation amplifier configuration is shown in Figure 3.5

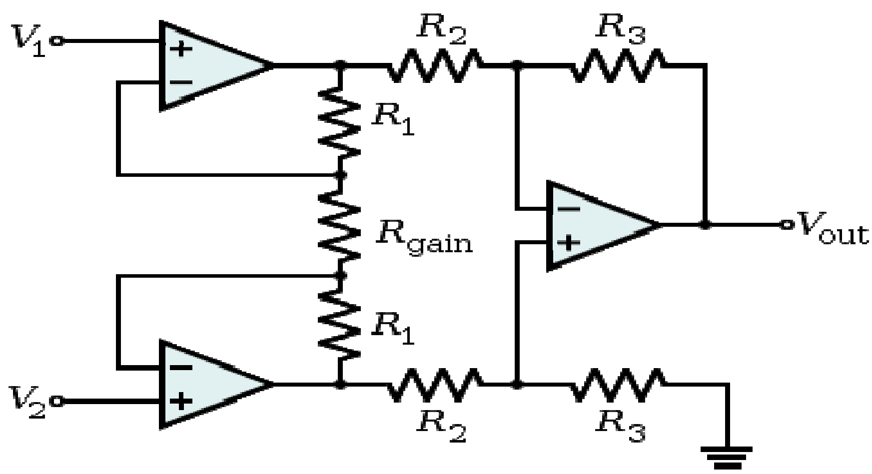


Figure 3.5: Instrumentation Amplifier

3.3.1 AD8429

The AD8429 is an ultralow noise, instrumentation amplifier designed for measuring extremely small signals like ECG signal. It delivers ultralow input noise performance of 1 nV/Hz. The high CMRR of the AD8429 prevents unwanted signals from distortion. In AD8429, CMRR increases as the gain increases, and offering high rejection to noise and common signals. The high performance pin configuration of the AD8429 allows it to reliably maintain high CMRR at frequencies well beyond those of typical instrumentation amplifiers. It can amplify fast changing signals. Its current feedback architecture provides high bandwidth at high gain, for example, 1.2 MHz at G = 100. The AD8429 was designed for excellent distortion performance. In AD8429, the gain can be set from 1 to 10,000 with a single resistorR_g. A reference pin allows the user to offset the output voltage. This feature can be useful to shift the output level when interfacing to a single supply signal chain. The AD8429 performance is specified over the extended industrial temperature range of −40°C to +125°C. It is available in an 8-lead plastic SOIC package [25]. The 8 pin package is shown in Figure 3.6

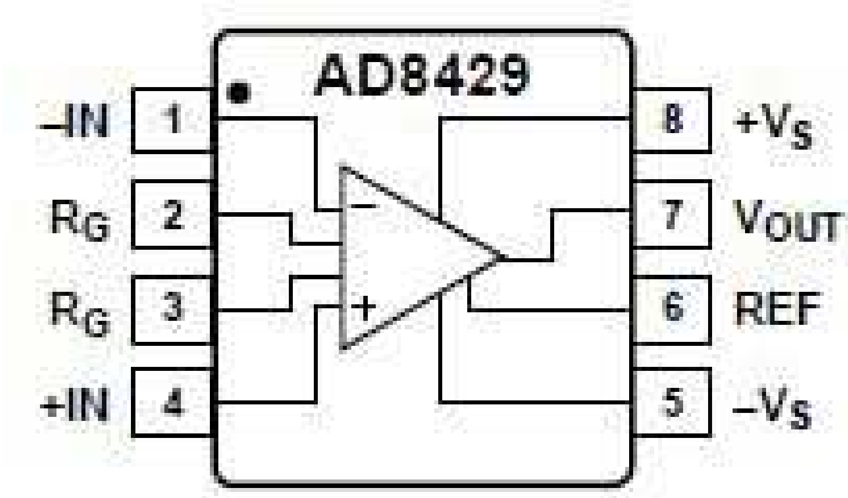


Figure 3.6: Pin Configuration of AD8429 Instrumentation Amplifier

The pins are described in Table 3.1

Pin No.	Name	Description
1	-IN	Negative input terminal
2,3	R_G	Gain setting resistor. Place a resistor across the terminals to set the gain. Here G=1+($6k_{\Omega}/R_G$)
4	+IN	Positive input terminal
5	-V_S	Negative input Power terminal
6	REF	Reference voltage. To shift the level of output
7	V_{OUT}	Output terminal
8	+V_S	Positive input Power terminal

Table 3.1: Pin Description of AD8429

3.3.1.1 Comparison with other Instrumentation Amplifiers

Here various parameters and specifications of various Instrumentation amplifiers are compared with AD8429 shown in Table 3.2

Types	AD8429	AD620	AMP01	INA114
CMRR	140db	130db	125db	115db
Noise	1nV/Hz	9nV/Hz	10nV/Hz	11nV/Hz
Input bias Current	150nA	0.5nA	4nA	2nA
Accuracy	0.02%	0.15%	0.18%	0.18%
Power Requirement	7mA	1.3mA	4.8mA	2.2mA

Table 3.2: Comparison of Various Instrumentation Amplifiers

From the comparison it is concluded that the response of AD8429 is good regarding the CMRR and Noise.

3.4 OPA 355

For designing of various Active filter configurations OPA 355 operational amplifier is used. An OPA355 series is a high-speed, voltage-feedback CMOS operational amplifier. These are designed for video and other applications requiring wide bandwidth. An OPA355 is a unity gain stable amplifier and it can drive large output currents. OPA355 has a digital shutdown (Enable) function. This feature provides power savings during idle periods and places the output in a high-impedance state to support output multiplexing. Its Differential gain is 0.02% and differential phase is 0.05°. The OPA355 is optimized for operation on single or dual supplies as low as 2.5V (1.25V) and up to 5.5V (2.75V). Its Common-mode input range can extends 100mV below ground and up to 1.5V from V+. It can support wide dynamic range variation [26]. The pin diagram of OPA 355 is shown in Figure 3.7

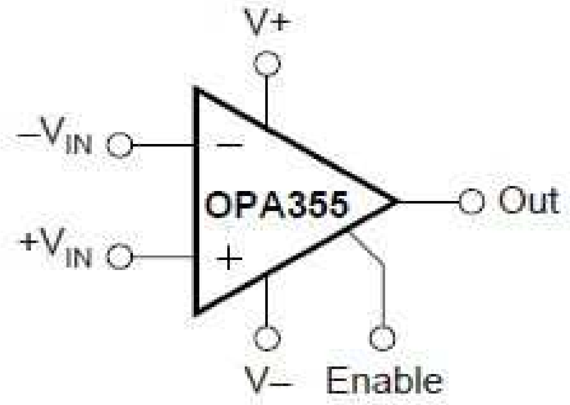


Figure 3.7: Pin diagram of OPA355

The pins are described in Table 3.3.

Name	Description
-V_{IN}	Negative Input Signal Terminal
+V_{IN}	Positive Input Signal Terminal
+V	Positive Voltage Supply Terminal
-V	Negative Voltage Supply Terminal
Enable	A Reference Voltage to make Operational Amplifier Active (For Power Saving Purpose)
V_{OUT}	Output Signal Terminal

Table 3.3: Pin Description of OPA355

3.5 Filter Approximation Method

Attaining the ideal response of a filter is not possible. Hence the best way is to approximate it to near values. To determine the values of components, first determine the order of filter then normalize its transfer function then unnormalize its approximation transfer function to convert it into another configuration. For example to convert low pass filter into high pass filter just unnormalize the transfer function.

3.5.1 First order Filters

a) Low Pass Filter

The transfer function of 1st order LPF is given as

$$H_{i,L}(S) = \frac{A_0}{a_i S+1} \tag{1}$$

Where A_0 is Gain of filter, a_1 coefficient is the complex pole location of filter, S is complex frequency variable. The values of Components are determined using AHA recommendations [22] [23] [24]. The values after calculations are shown in Table 3.4

S. No	Filter	R	C	Gain
1	LPF	1.59kΩ	1μf	Unity

Table 3.4: Component Values

The designed circuit of LPF is shown in Figure 3.8

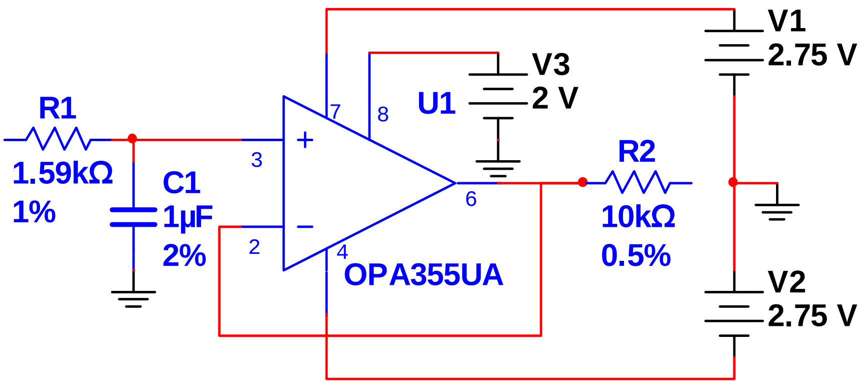


Figure 3.8: Circuit diagram of LPF

b) High Pass Filter

The transfer function of 1st order HPF is given as [22] [23] [24].

$$H_{1,H}(S) = \frac{A_\infty}{a_i S+1} \tag{2}$$

The values are calculated using AHA recommendations. The values after calculations are shown in Table 3.5

Sr. no.	Filter	R	C	Gain
1	HPF	3.2MΩ	1µf	Unity

Table 3.5: Component Values

The designed circuit of HPF is shown in Figure 3.9.

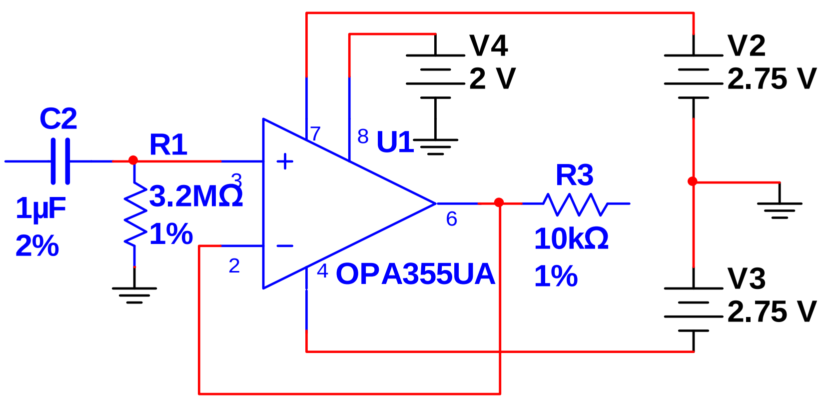


Figure 3.9: Circuit diagram of HPF

3.5.2 Butterworth Filter

The Butterworth filter has flat gain response in pass band and flat transition after cut-off frequency. No other approximation has a smoother transition through the pass band to the stop band. The phase response also is very smooth. The roll-off is not sharp as compare to chebyshev and inverse chebyshev filter. This approximation Method is selected, when low phase distortion and moderate selectivity are required [22] [23] [24].

The magnitude response of filter is given in equation (3). Where n is the order of the approximation function, ω_o is the pass band gain edge frequency of filter. ϵ is the pass band gain adjustment factor.

$$\left|H_{B,n}[j({}^{\omega}/_{\omega_0})]\right| = \frac{1}{\sqrt{1+\epsilon_0{}^2.({}^{\omega}/_{\omega_0})^{2n}}} \tag{3}$$

Where

$$\epsilon_0 = \sqrt{10^{-0.1\,\alpha_{pass}} - 1} \text{ (Pass band gain adjustment factor)}$$

The filter order n_B of LPF and HPF is calculated the equation (4). In this formulation, note that it is the ratio of the stop band and pass band frequencies .which is important. Hence with a given set of gains will require the same order whatever the edge frequencies. The value of *n* calculated using this equation must always be rounded to the next highest integer in order to guarantee that the specifications will be met by the integer order of the filter design

$$n_B = \frac{\log\left[(10^{-0.1.\alpha_{stop}}-1)/(10^{-0.1.\alpha_{pass}}-1)\right]}{2.\log(\Omega_r)} \tag{4}$$

Where $\Omega_r(3\ db\ cut\ off\ frequncy) = \frac{\omega_{stop}}{\omega_{pass}}$ for LPF and $\frac{\omega_{pass}}{\omega_{stop}}$ for HPF

The generalized transfer function of Butterworth approximation filter is given in equation (5). Where R is radius of circle in pole zero Location, $B_{1m}$and B_{2m} are quadrants.

$$H_{B,n}(S) = \frac{R.\prod_m B_{2m}}{(S+R).\prod_m(S^2+B_{1m}.S+B_{2m})} \tag{5}$$

m = 0, 1.... [(n-1)/2] -1 (n odd)

With the reference of these equations and AHA recommendations values are calculated for both the HPF and LPF shown in Table 3.6

Sr. No.	Component	LPF	HPF
1	Order	3	3
2	R1,R2,R3	1.6K,1.6k,1.6K	3.2M,63.7K,15.9K
3	C1,C2,C3	1µf,1µf,1µf	1µf,100µf,100µ
4	Gain	2	2

Table 3.6: Component Values.

The designed circuit diagram of both the Butterworth LPF and HPF are shown in Figure 3.10 and 3.11

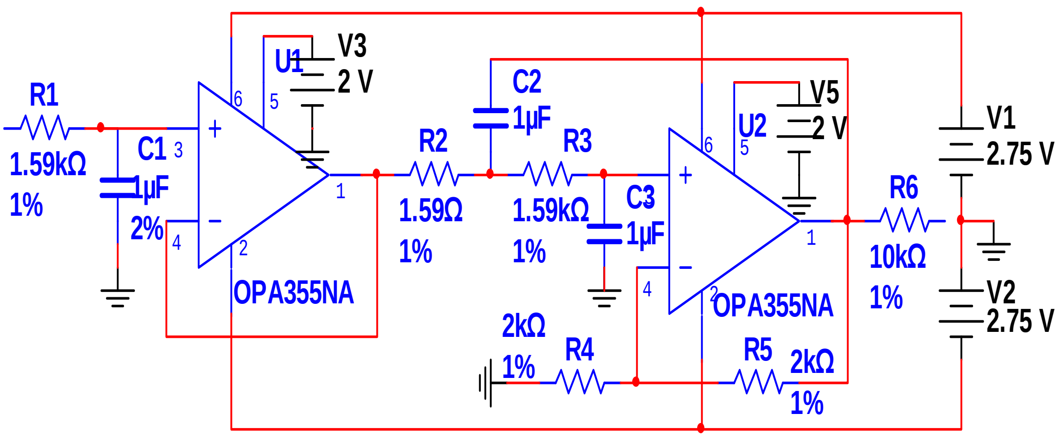


Figure 3.10: Circuit diagram of 3rd Order Butterworth LPF

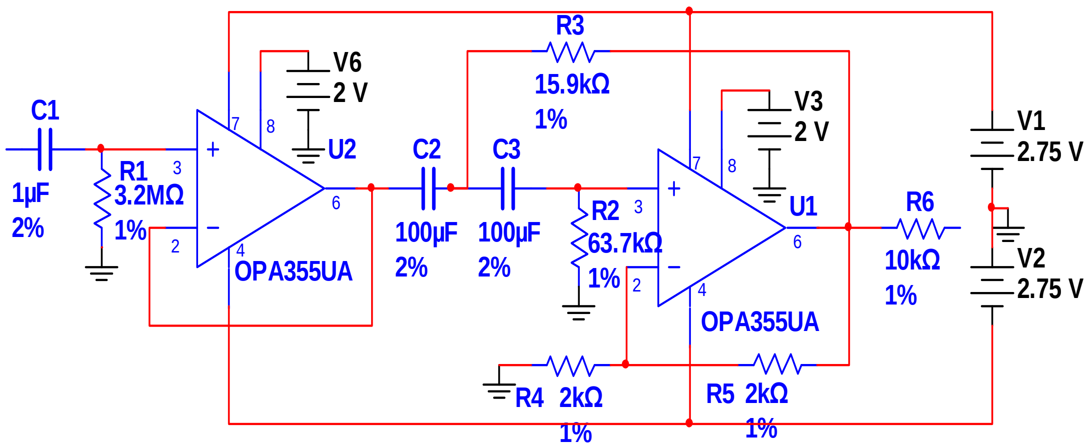


Fig 3.11: Circuit diagram of 3rd Order Butterworth HPF

3.5.3. Chebyshev Filter: The Chebyshev approximation function also has an all-pole transfer function like the Butterworth approximation. However, unlike the Butterworth case, the Chebyshev filter allows variation or ripple in the pass-band of the filter. This reduction in the restrictions placed on the characteristics of the pass band enables the transition characteristics of the Chebyshev to be steeper than the Butterworth transition. Because of this more rapid transition, the Chebyshev filter is able to satisfy user specifications with lower-order filters than the Butterworth case. However, the phase response is not as linear as the Butterworth case, the magnitude response of Chebyshev approximation filter is given in the equation (6). Where n is the order of the approximation function, ω_o is the pass band gain edge frequency of filter. ϵ is the pass band gain adjustment factor [22] [23] [24].

$$\left|H_{C,n}[j(\omega/\omega_0]\right| = \frac{1}{\sqrt{1+\epsilon^2 {C_n}^2.(\omega/\omega_0)}} \tag{6}$$

Where $\epsilon = \sqrt{10^{-0.1\,\alpha_{pass}} - 1}$

And $C_n(\omega)$ is the normalised chebyshev polynomial ($\omega_0 = 1$) given as

$$C_n(\omega) = \cos[n.cos^{-1}(\boldsymbol{\omega})], \boldsymbol{\omega} \leq \boldsymbol{0} \tag{7}$$

$$C_n(\omega) = \cos\mathrm{h}[n.cosh^{-1}(\boldsymbol{\omega})], \boldsymbol{\omega} \geq \boldsymbol{0} \tag{8}$$

The order n_C of LPF and HPF is calculated the equation (9). In this formulation, note that it is the ratio of the stop band and pass band frequencies .which is important. Hence with a given set of gains will require the same order whatever the edge frequencies. The value of n calculated using this equation must always be rounded to the next highest integer in order to guarantee that the specifications will be met by the integer order of the filter design.

$$n_C = \frac{Cosh^{-1}[\sqrt{(10^{-0.1.\alpha_{stop}}-1)/(10^{-0.1.\alpha_{pass}}-1)}]}{\mathrm{Cosh}^{-1}\,(\omega_{stop}/\omega_{pass})} \tag{9}$$

Where $\Omega_r = \frac{\omega_{stop}}{\omega_{pass}}$ for LPF and $\frac{\omega_{pass}}{\omega_{stop}}$ for HPF

The generalized transfer function of Butterworth approximation filter is given in equation (10). Where $B_{1m}$and B_{2m} are quadrants,

$$H_{C,n}(S) = \frac{\sinh(D).\prod_m B_{2m}}{(S+\sinh(D)).\prod_m (S^2+B_{1m}S+B_{2m})} \quad (10)$$

m = 0,1,.... [(n-1)/2] -1 if n = odd.

With the reference of these equations and AHA recommendations, component values are calculated for both the HPF and LPF which are shown in Table 3.7

Sr. no.	Components	LPF	HPF
1	Order	3	3
2	R1,R2,R3	2.9K,1.74K,1.74K	17.1K,995Kk,854k
3	C1,C2,C3	1µf,1µf,1µf	100µf,100µf,100µf
4	Gain	2.41	2.41

Table 3.7: Component Values

The designed circuit of LPF and HPF are shown in Figure 3.12 and 3.13 respectively.

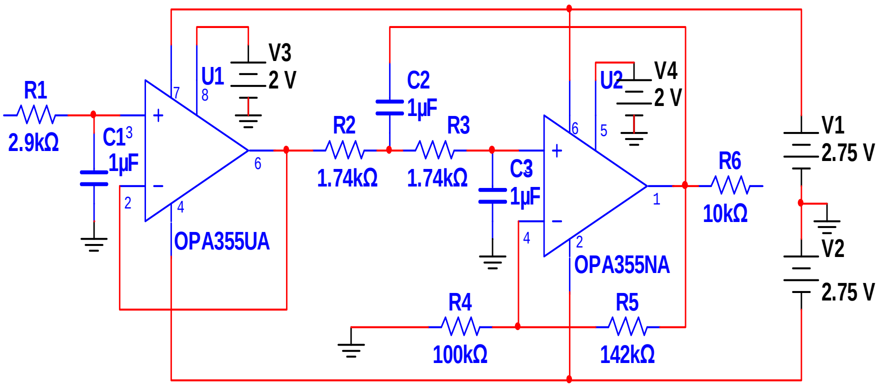


Figure 3.12: Circuit diagram of 3rd Order Chebyshev LPF

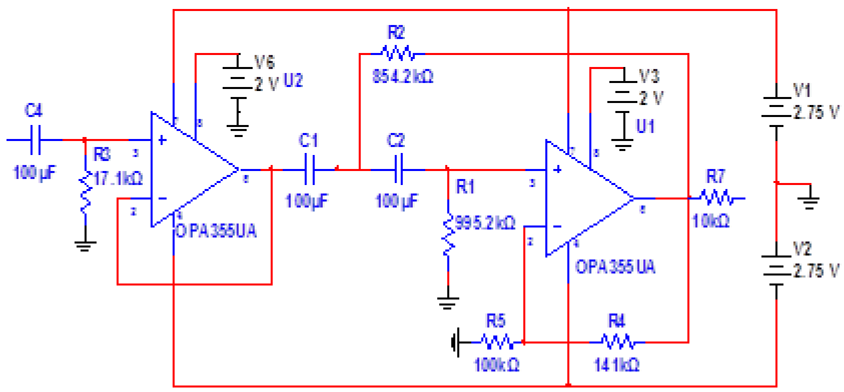


Figure 3.13: Circuit diagram of 3rd Order Chebyshev HPF

3.5.4. Inverse Chebyshev Filter: The response of Inverse chebyshev approximation filter is opposite to chebyshev approximation filter's response. It is also called Chebyshev II approximation filter. It has flat pass band gain and has ripples in stop band. This filter has complex zeros on $j\omega$ axis in s-plane.so its quadratic factor is identical to band stop filter function. The inverse Chebyshev approximation provides better transition characteristics than the Butterworth filter and better phase response than the standard Chebyshev.

The magnitude response of filter is given in equation (11). Where $C_n(\omega)$ is the normalised chebyshev polynomial, n is the order of the approximation function, ω_o is the pass band gain edge frequency of filter. ϵ_I is the pass band gain adjustment factor [22][23][24].

$$\left|H_{I,n}[j(^{\omega}/_{\omega_0})]\right| = \frac{\sqrt{\epsilon_I{}^2.C_n{}^2(\omega_0/\omega)}}{\sqrt{1+\epsilon_I{}^2 C_n{}^2.(\omega_0/\omega)}} \tag{11}$$

Where $\epsilon_{I=}\frac{1}{\sqrt{10^{-0.1\alpha_{stop}}-1}}$

The order of filter is determined through the equation.

$$n_I = \frac{Cosh^{-1}[\sqrt{10^{-0.1.\alpha_{stop}}-1/10^{-0.1.\alpha_{pass}}-1}]}{\text{Cosh}^{-1}\,(\omega_{stop}/\omega_{pass})} \tag{12}$$

Where $\Omega_r = \frac{\omega_{stop}}{\omega_{pass}}$ for LPF and $\frac{\omega_{pass}}{\omega_{stop}}$ for HPF

The generalized transfer function of inverse chebyshev filter is given in equation (13). There are many quadrants in the transfer function equation.

$$H_{I,n}(S) = \frac{[\sinh\,(D_i)^{-1}.\prod_m(S^2+B_{2m}).\prod_m(S^2+A_{1m}.S+A_{2m})}{(S+[\sinh(D_i)]^{-1}.\prod_m(A_{2m}).\prod_m(S^2+B_{1m}.S+B_{2m})} \tag{13}$$

M=0,1....[(n-1)/2]-1

With the reference of these equations and AHA recommendations values are calculated and shown in Table 3.8.

Sr. no.	Component	LPF	HPF
1	Order	3	3
2	R1, R2, R3,R4	58.5K,27.5K,27.5K,13.7K	7.4M,7.4M,3.7M,3.18M
3	C1,C2,C3,C4,C5	.01µ,.01µ,.02µ,.01µ,.02µ	10µ,10µ,20µ,1µ
4	Gain	2.19	1.99

Table 3.8: Component Values

The designed circuit of LPF and HPF are shown in Figure 3.14 and 3.15 respectively.

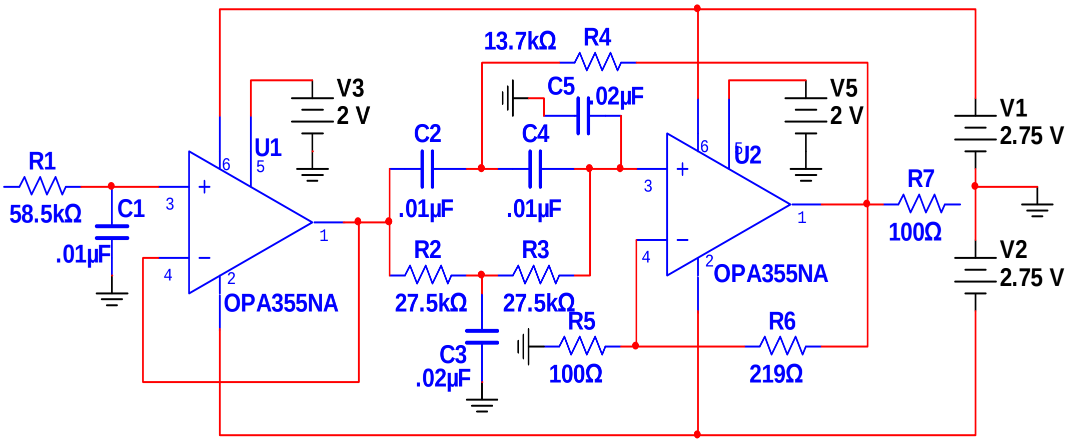


Figure 3.14: Circuit diagram of 3rd Order Inverse Chebyshev LPF

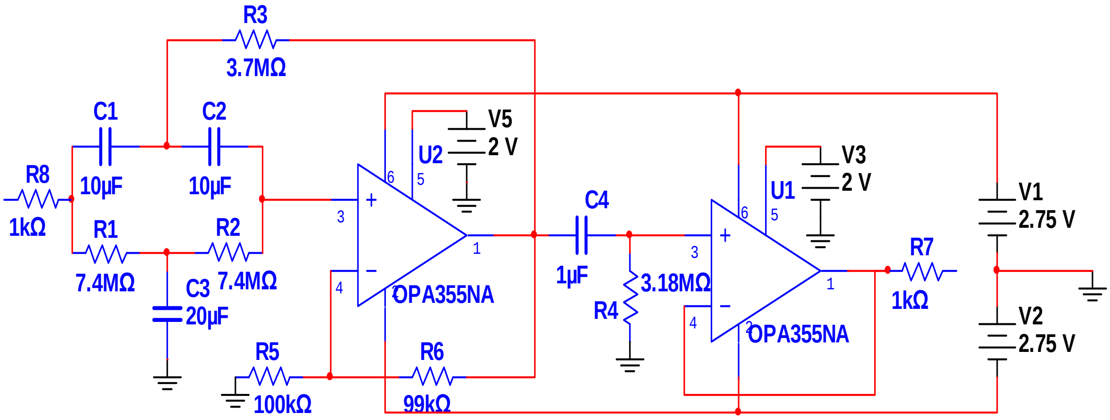


Fig 3.15: Circuit diagram of 3rd Order Inverse Chebyshev HPF

Chapter 4

TESTING

4.1 Testing

After the filters were designed, their testing and analysis was performed on National Instruments Multisim 11.0.1 workbench and LabView 8.6. With the help of this graphical software, filters were designed and simulated. For analysis, Bode Plot Response (Magnitude and phase plot) and Output Response were performed on Multisim 11.0.1 and then FFT analyses were performed on LabView 8.6.

4.2 Multisim 11.0.1

National Instrument's Multisim 11.0.1 is a circuit design workbench. It contains almost all the components and tools for designing. In Multisim, circuits are designed and then simulate to perform different responses and analysis like Transient Response, Bode Plot, Phase Response etc. It contains Virtual Instrument like Voltmeter, Ammeter, Oscilloscope, Multimeter etc to measure values. This workbench is very easy to use and it is flexible to add new components and instrument [27].

4.2.1 Bode Plot

A Bode Plot is a Graph which shows frequency response of a system. It is a combination of Bode's Magnitude Plot (which shows the magnitude of the frequency response gain) and Bode's Phase Plot (which shows the frequency response phase shift of a system) [28]. Bode Plot is present in Multisim 11.0.1's tool Palate. It is very simple and accurate method for graphing gain and phase-shift response of a circuit. Bode Plot is an important tool to analyse these filters. The axis of graph is shown in Figure 4.1

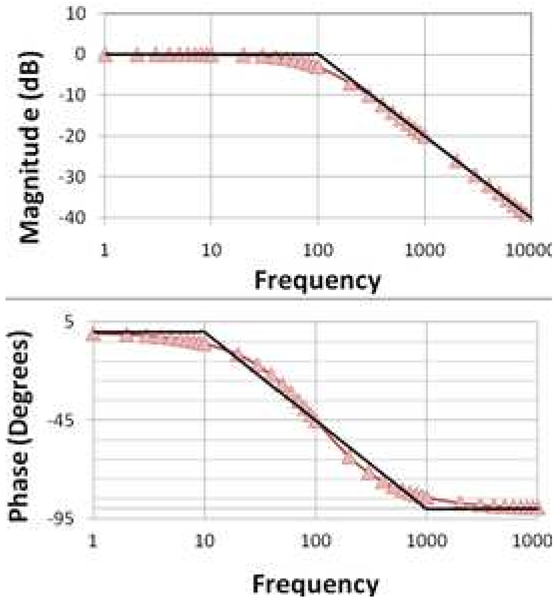


Figure 4.1: Bode's magnitude and Phase Response Graphs

With the help of Bode Plot, Both Magnitude and Phase Response of filter circuits are calculated and then compared on different point on frequency axis using AHA recommendations.

4.2.2 Output Response of Filters

After analysing the Bode Plot using AHA recommendations, an ECG signal is given to filter's input terminal to get the output shape of ECG signal [27]. The ECG signal is virtually given to filters in Multisim 11.0.1. The procedure to make input ECG source in Multisim11.0.1 is given below as

- An ECG signal file of .DAT format is downloaded from www.physionet.org. This website contains recorded physiological signals.
- Now open this file, it contains data in two columns (one column for time and one for ECG values). Then copy the given two columns of data in to Notepad(.txt format).
- Now in Multisim11.0.1, Go to Components and then select voltage source then select Piecewise linear voltage source. The symbol of Piecewise linear Voltage source is shown in Figure 4.2

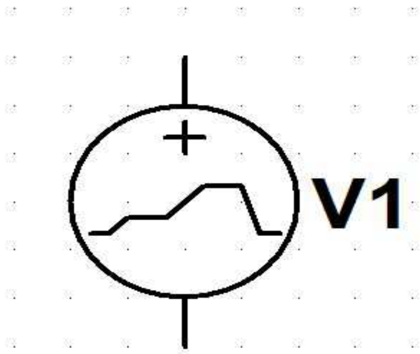


Figure 4.2: Symbol of Piecewise linear Voltage Source

- Now Double click on it. An option will open. Then goto initialize from file option. Now select the notepad file. Now the data will be loaded in the Voltage source. The view is shown in Figure 4.3

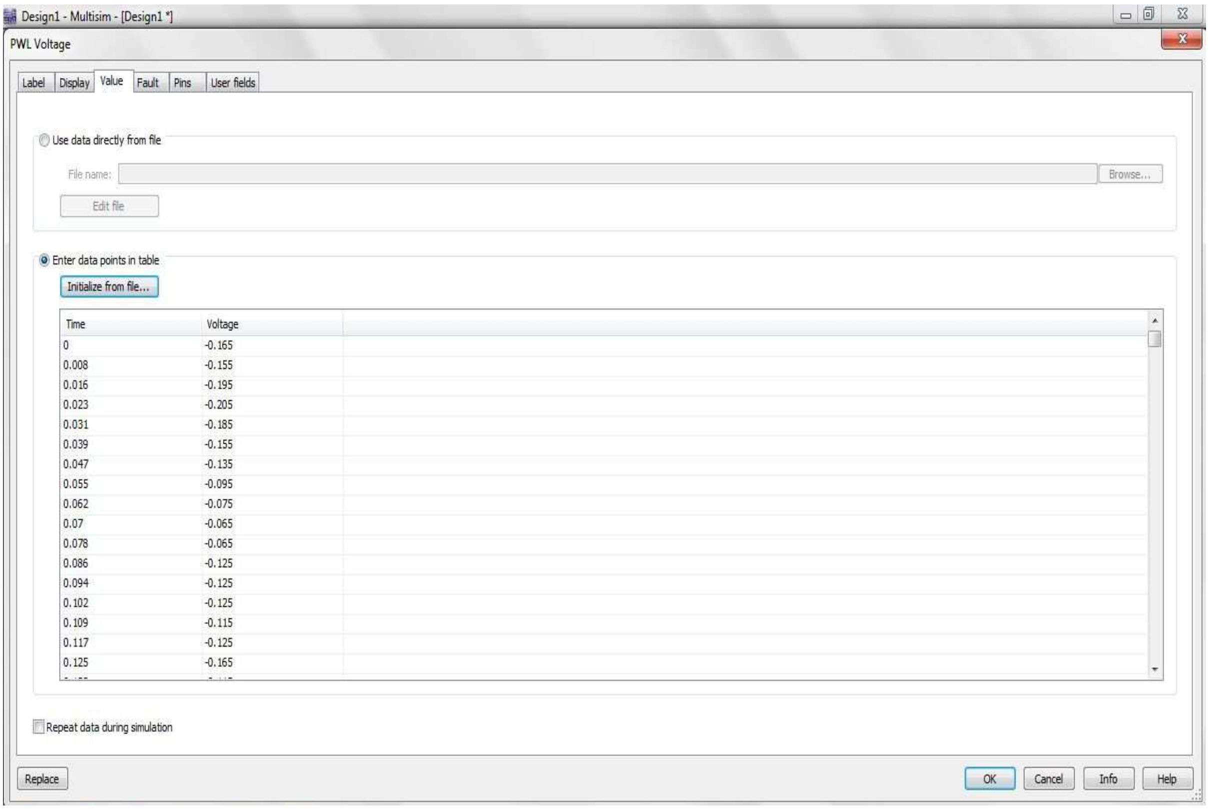


Figure 4.3: A view of options of Piecewise linear Voltage source

Now it is become an ECG signal Source. By connecting an oscilloscope an input signal can be seen. The simulated input signal is shown in Figure 4.4

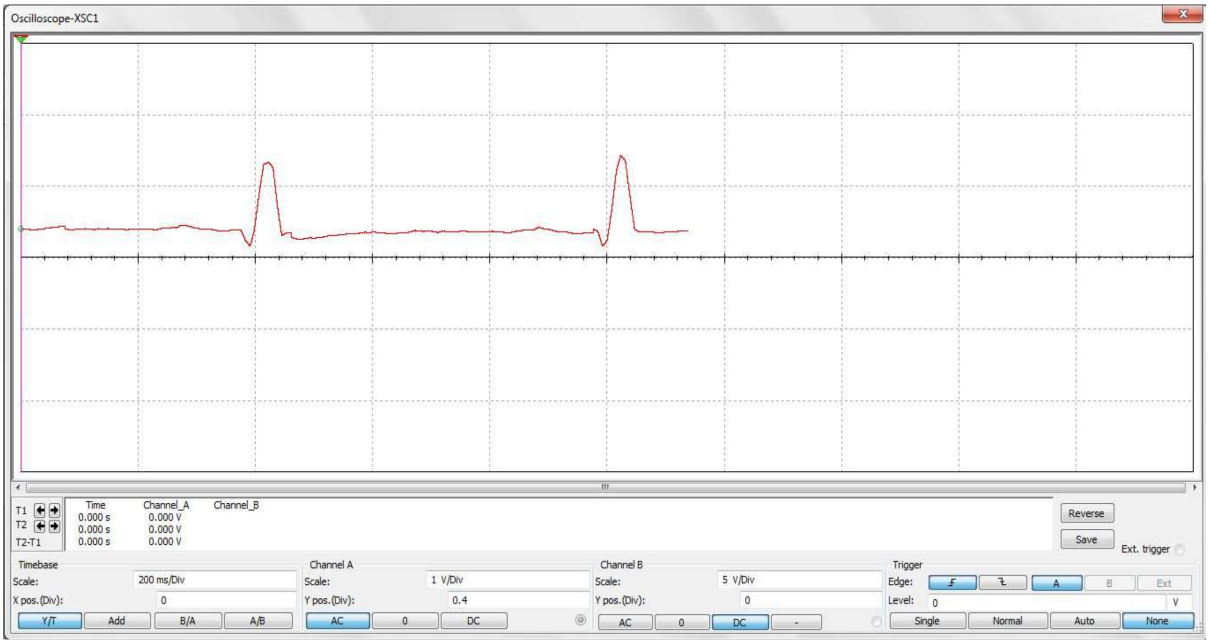


Figure 4.4: Simulated input ECG signal in Multisim 11.0.1

This input signal is given to filters and output is checked at output terminals using oscilloscope.

4.3 Labview 8.6

Labview 8.6 is a system design platform and development environment for a visual programming language from National Instruments. It is also called Graphical Programming. In Labview there are various tool boxes like for signal processing, control system etc [29]. They have built in VI files. In Labview, there is no need of conventional procedural programming. In labview8.6 output signal of Filters are further processed for feature extraction.

4.3.1 Fast Fourier Transform

As there are other methods to calculate DFT. A Fast Fourier transform (FFT) is an efficient algorithm to compute the Discrete Fourier transform of a signal. The DFT is extremely important in the area of frequency (spectrum) analysis. Because it takes a discrete signal in the time domain and transforms that signal into its discrete frequency domain representation. Without a discrete-time to discrete-frequency transform it is not possible to compute the Fourier transform. In Labview 8.6, there are signal processing tools (built in signal processing functions) to compute the FFT of a signal using Labview is an easy approach for signal processing [30].

4.3.2 Feature Extractions from FFT

Conventionally ECG signals are acquired by ECG acquisition devices and then these devices generate a printout of the lead's outputs on a graph paper. After that a cardiologist analyse the ECG data for checking the abnormality in the signal. But now automatic ECG processing is the main focus. Now the main point of concern is how to extract the features from ECG signal so that these features can be used for Automatic Diseases Diagnosis and to identify symptoms of various heart diseases like Nocturnal hypoglycaemia, Ischemia etc. there are various complex algorithms for ECG signal feature extraction. FFT is one of them which is very efficient [31] [32] [33].

4.4 Multisim and Labview Connectivity

As the filters are tested and analysed by giving the ECG signal on the oscilloscope. There is a option on oscilloscope to save the simulated output ECG signal in different formats. For Labview usage, store the file in .lvm format [27] [29]. The view is shown in Figure 3.20

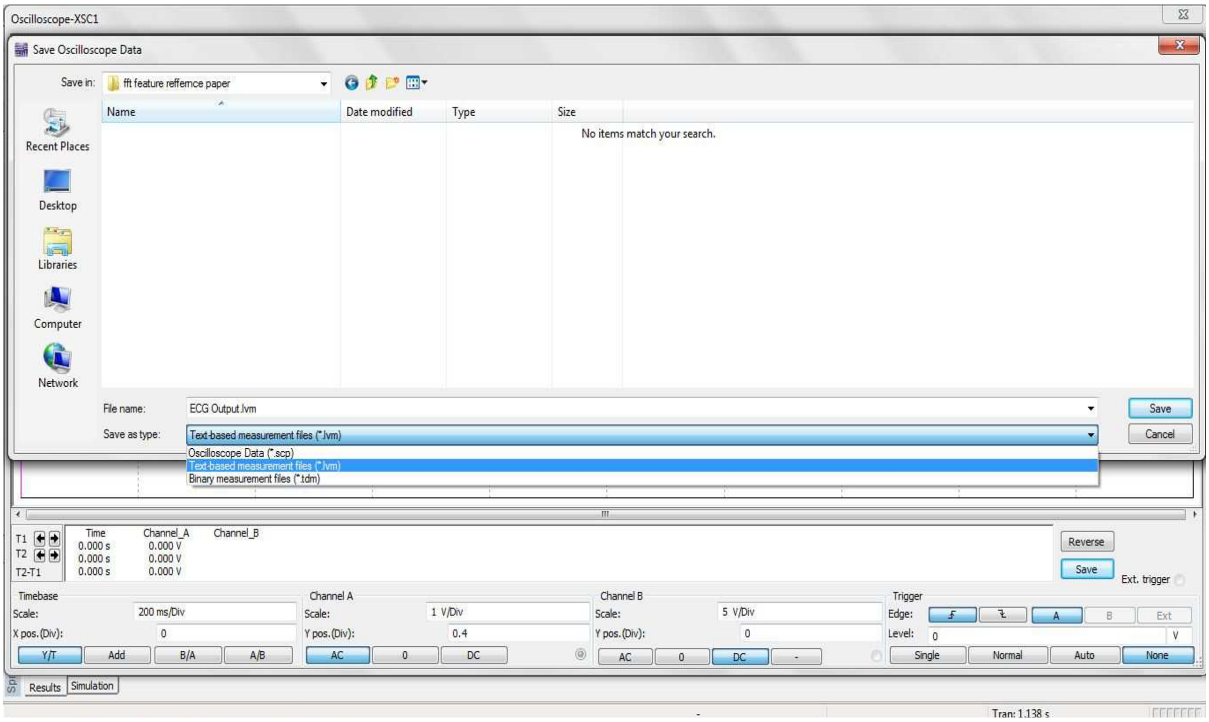


Figure 4.5: Save option in Multisim 11.0.1 for Labview usage

Now for FFT analysis in Labview, the VI is designed. The Block Diagram and Front panel is shown in Figure 4.6 and 4.7. The FFT graph is plotted between amplitude (db) and frequency.

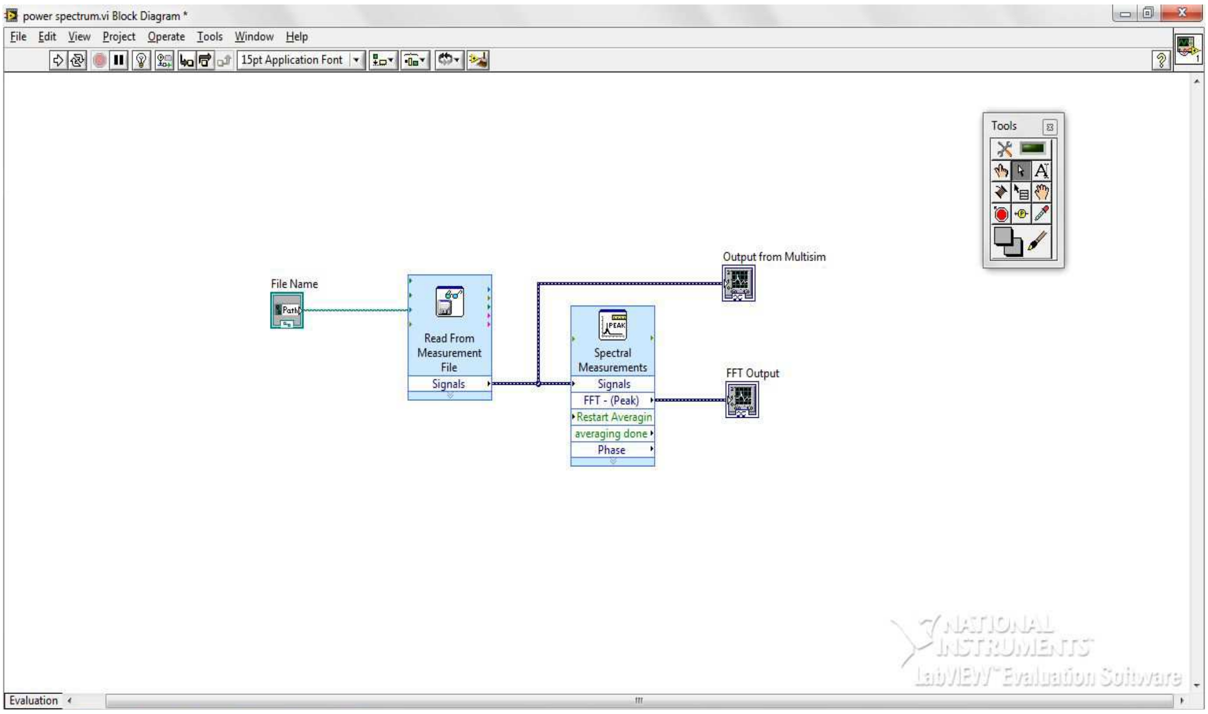


Figure 4.6 Block Diagram for FFT

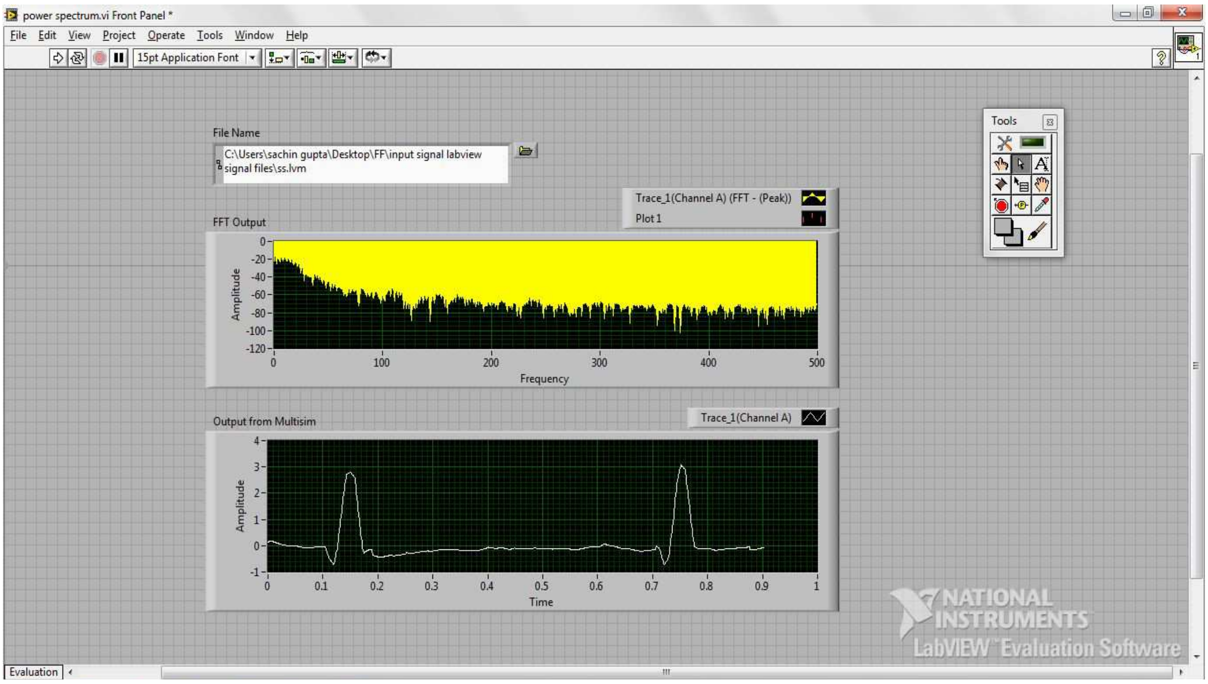


Figure 4.7 Front Panel for FFT

In front panel, the file (Simulated output ECG signal from filters) is selected through the file path option and then run the VI. The graphs show the ECG signal and FFT of signal in front panel.

Chapter 5

RESULTS AND DISSCUSSION

5.1 Introduction

The Filters were designed and then analysed using Bode Plot Response on Multisim 11.0.1 and FFT analysis were performed using Labview 8.6 according to AHA recommendations. The output of filters was tested by sending an ECG signal at their input terminals and output was measured using Oscilloscope and then compared. The Resultant analyses are given below.

5.2 Bode Plot Response of High Pass Filters

5.2.1 Bode Plot Response of 1st Order HPF

- The Magnitude versus Frequency Response of 1st order HPF (figure 3.8) is shown in Figure 5.1

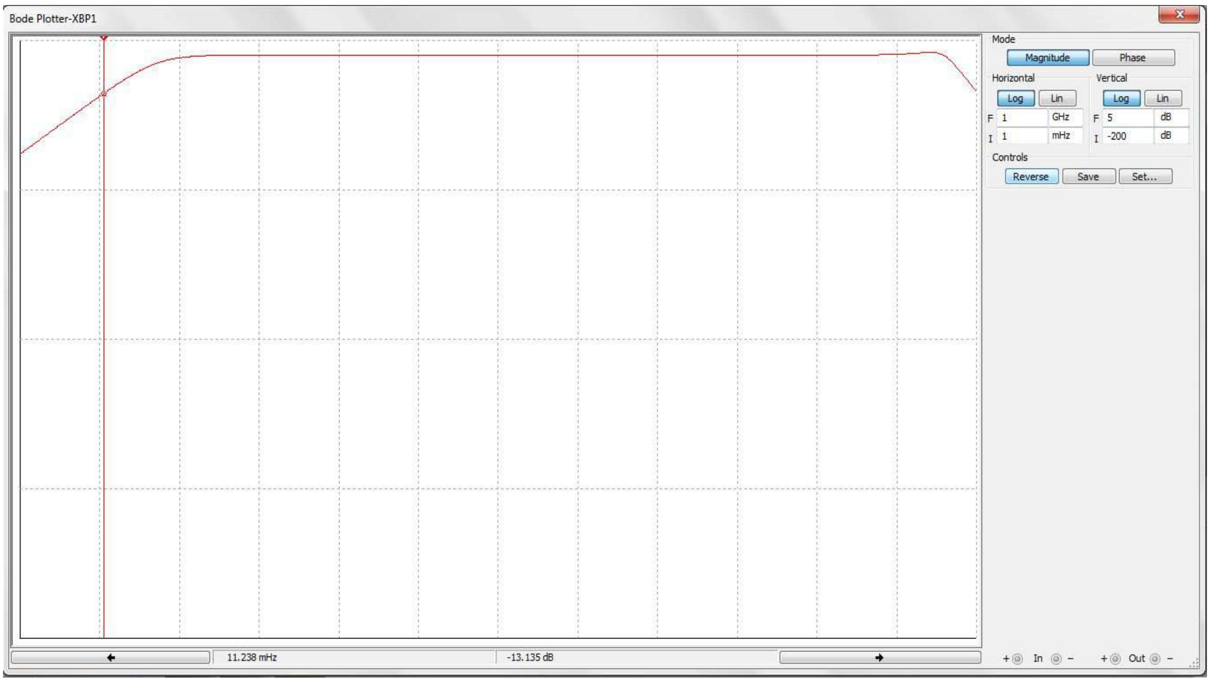


Figure 5.1: Magnitude versus Frequency Response of 1st order HPF

- The Phase Angle versus Frequency Response is shown in Figure 5.2

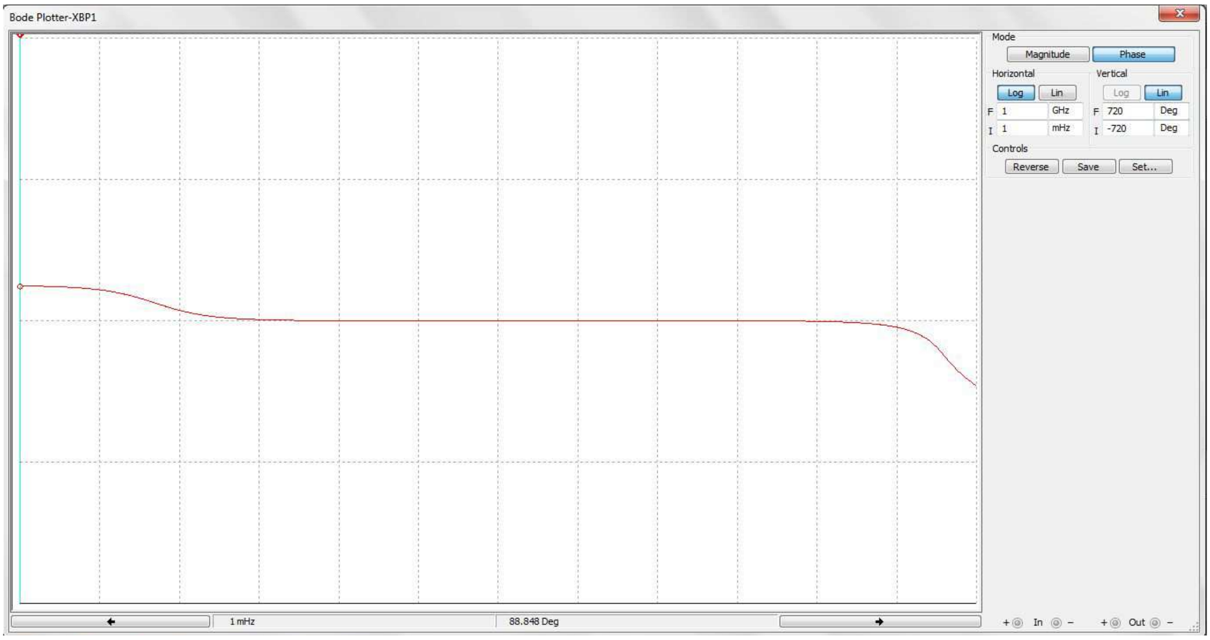


Figure 5.2: Phase Angle versus Frequency Response of 1st order HPF

The Phase Margin and gain margin of 1st order HPF is shown in Table 5.1

Cut off Frequency	Phase Margin	Gain Margin
0.05Hz	43°	Flat

Table 5.1: Phase and Gain Margin of 1st order HPF

5.2.2 Bode Plot Response of Butterworth HPF

- Magnitude Versus Frequency Response is shown in Figure 5.3

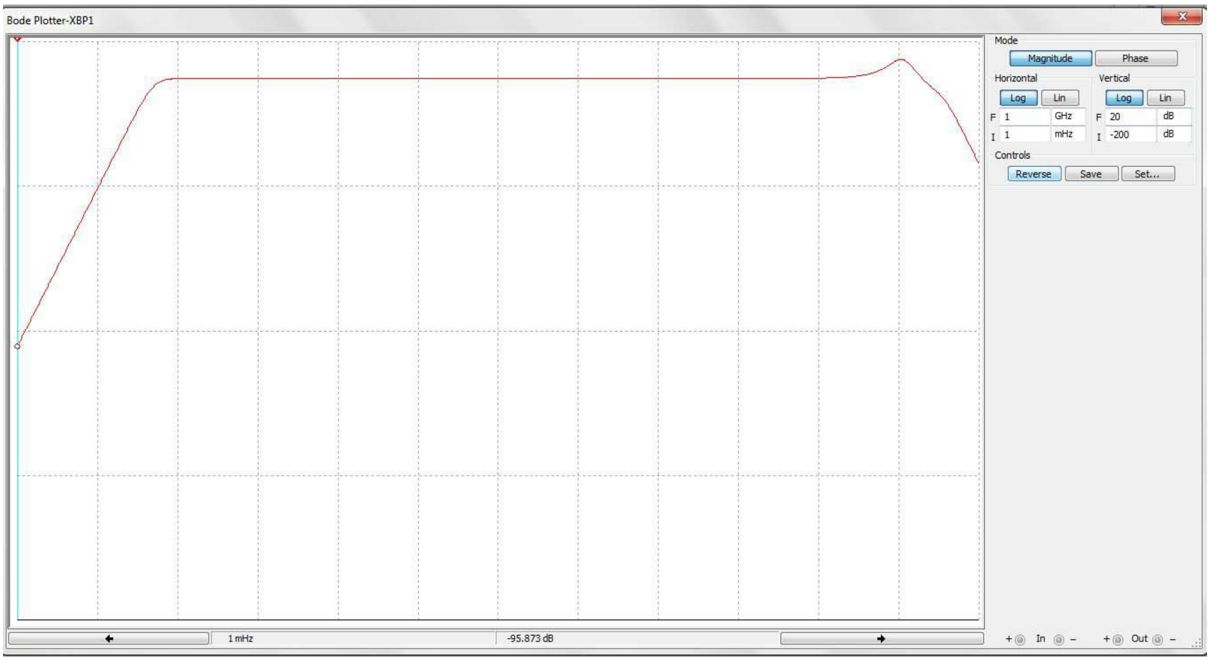


Figure 5.3: Magnitude versus Frequency Response of Butterworth HPF Filter

- The Phase Angle versus Frequency Response is shown in Figure 5.4

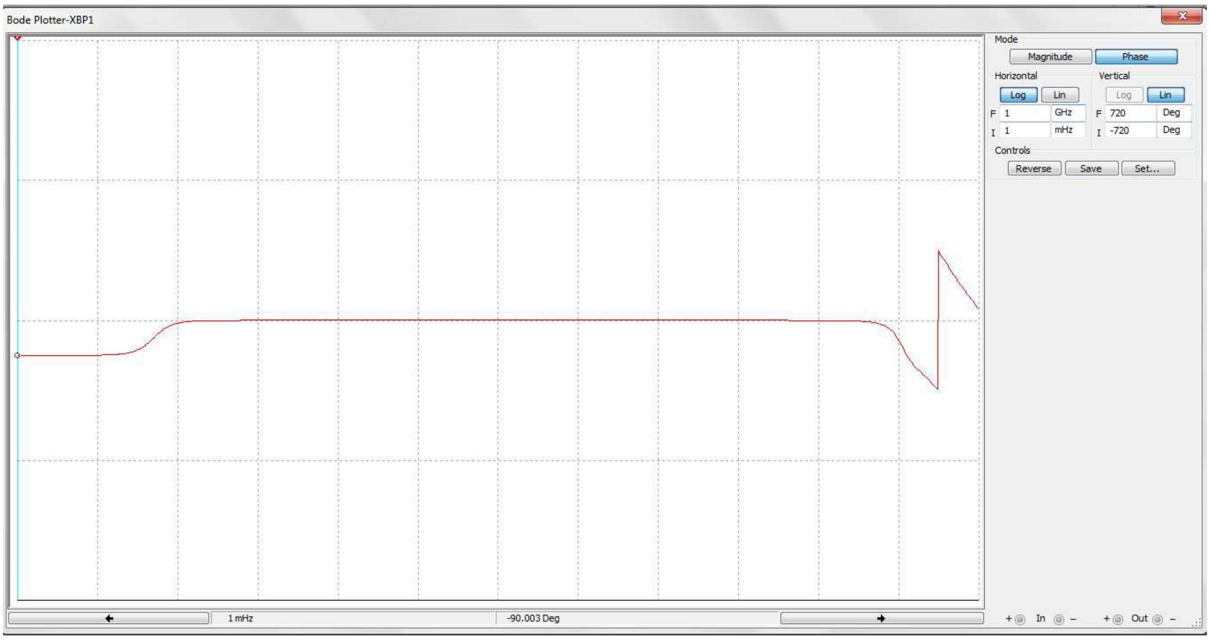


Figure 5.4: Phase Angle versus Frequency Response of Butterworth HPF

The Phase Margin and gain margin of Butterworth HPF is shown in Table 5.2

Cut off Frequency	Phase Margin	Gain Margin
0.05 Hz	-102.6°	+0.002db

Table 5.2: Phase and Gain Margin of Butterworth HPF

5.2.3 Bode Plot Response of Chebyshev HPF

- Magnitude Versus Frequency Response is shown in Figure 5.5

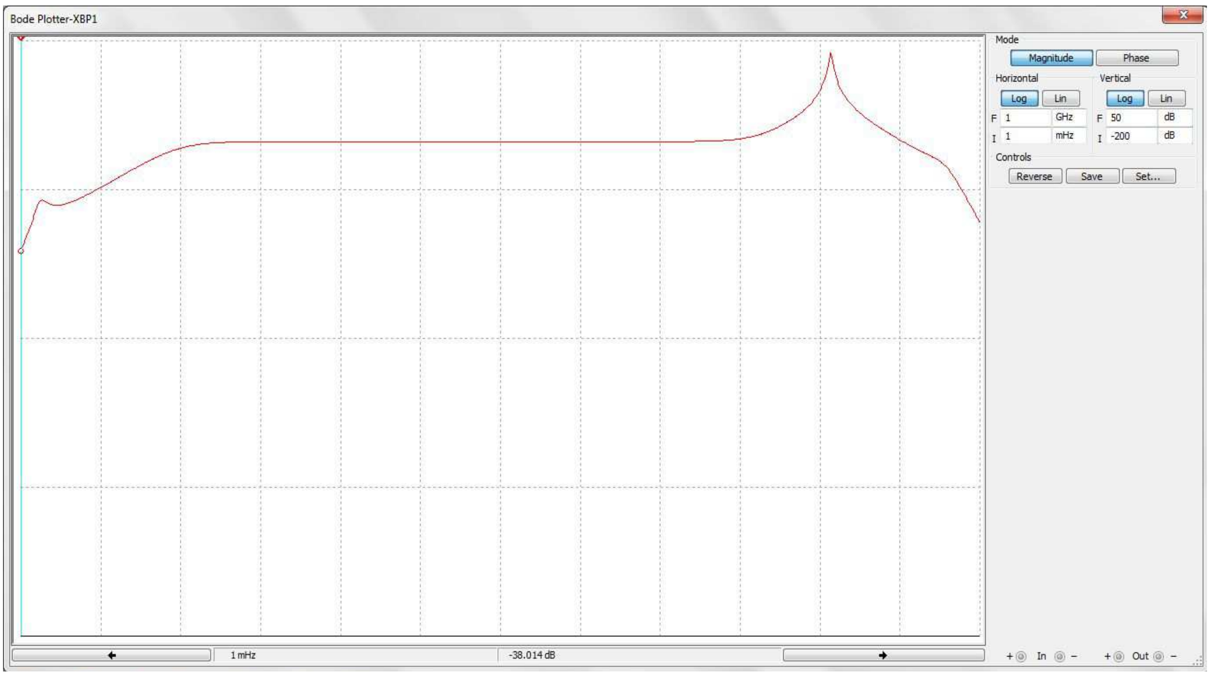


Figure 5.5: Magnitude versus Frequency Response of Chebyshev HPF

- The Phase Angle versus Frequency Response is shown in Figure 5.6

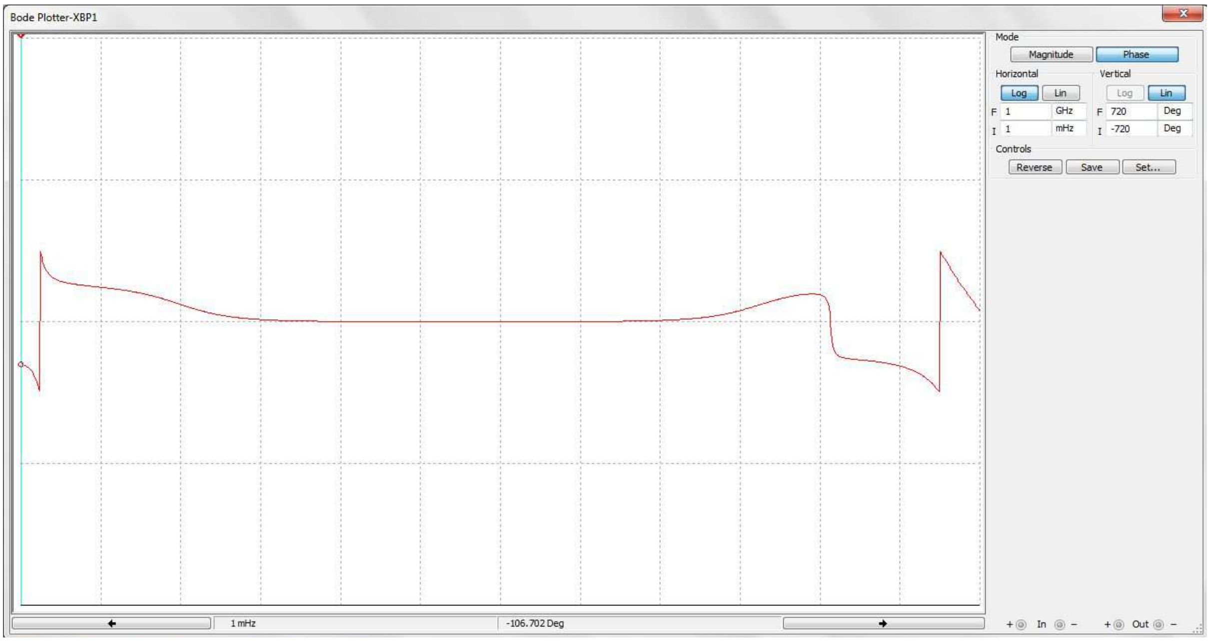


Figure 5.6: Phase Angle versus Frequency Response of Chebyshev HPF

The Phase Margin and gain margin of Chebyshev HPF is shown in Table 5.3

Cut off Frequency	Phase Margin	Gain Margin
0.05Hz	122.1°	-5.8db

Table 5.3: Phase and Gain Margin of Chebyshev HPF

5.2.4 Bode Plot Response of Inverse Chebyshev HPF

- Magnitude Versus Frequency Response is shown in Figure 5.7

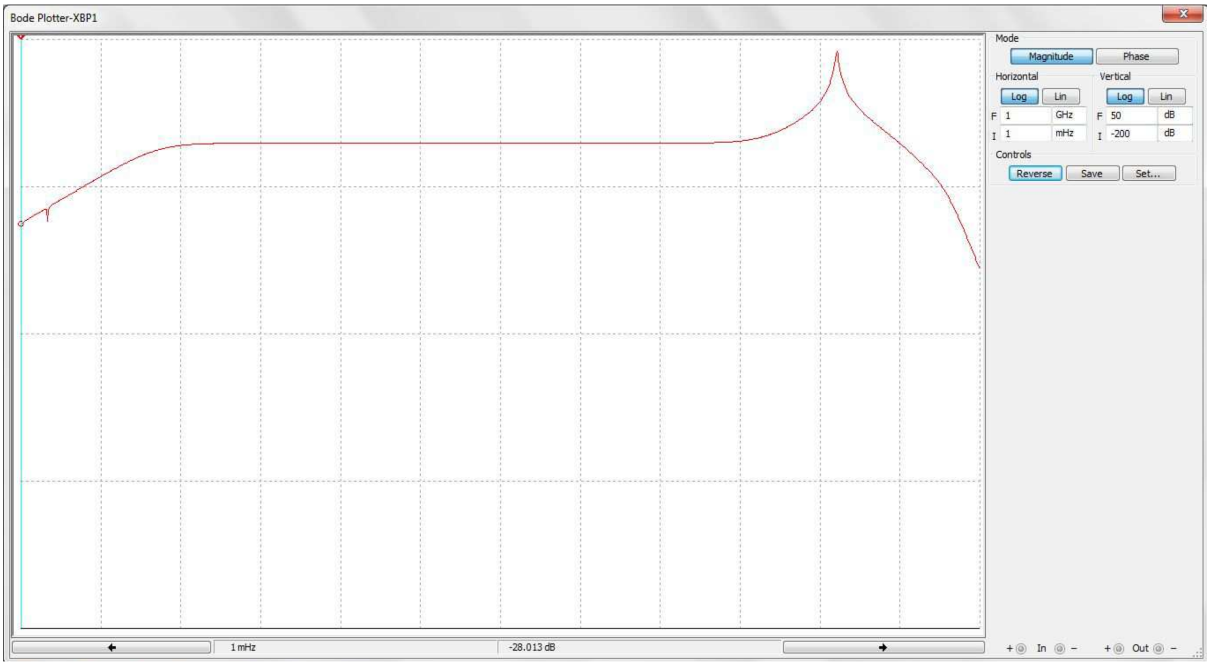


Figure 5.7: Magnitude versus Frequency Response of Inverse Chebyshev HPF

- The Phase Angle versus Frequency Response is shown in Figure 5.8

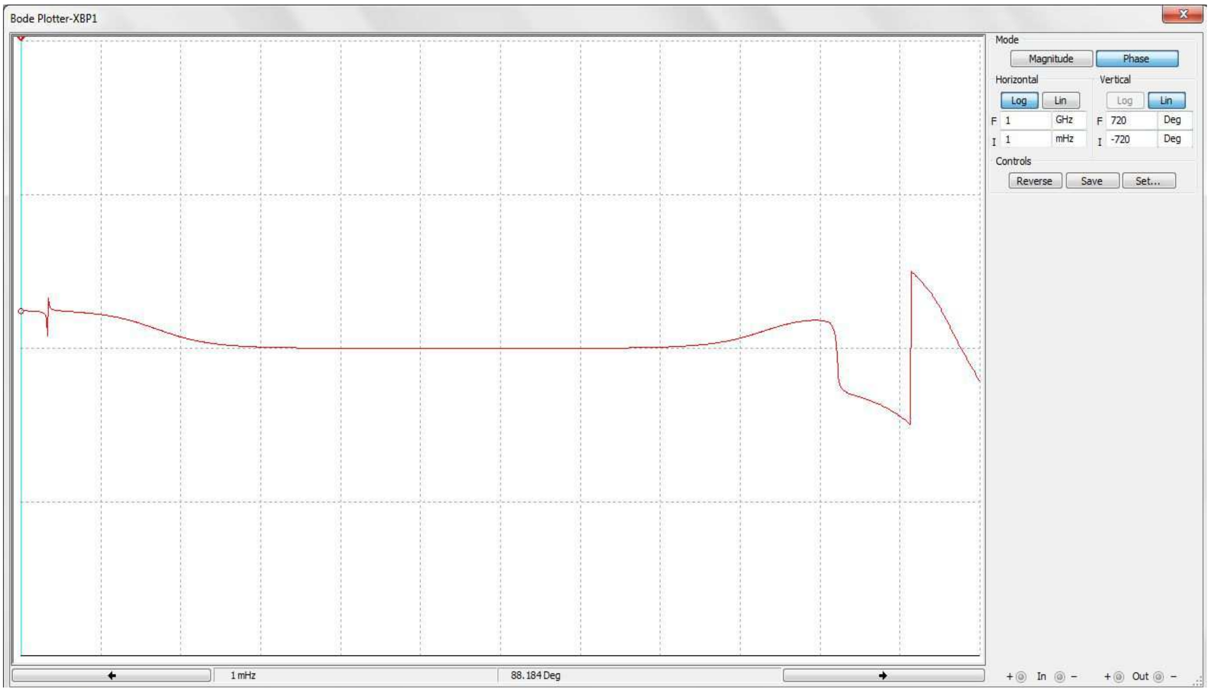


Figure 5.8: Phase Angle versus Frequency Response of Inverse Chebyshev HPF

The Phase Margin and gain margin of Inverse Chebyshev HPF is shown in Table 5.4

Cut off Frequency	Phase Margin	Gain Margin
0.05Hz	84°	6db

Table 5.4: Phase and Gain Margin of Inverse chebyshev HPF

5.3 Analysis of Bode Plot Response of HPF's using AHA Recommendation

The Analysis of Bode Plot for all HPF's are shown in Table 5.5

Sr. no.	Filter	f_{pass} at -3db point	Phase Shift at 0.05Hz	Phase Shift at 0.5Hz	Max Gain Deviation up to 0.5Hz	Ripple Presence
1	1st order	.049Hz	43°	5.6°	Flat	No
2	Butterworth	.048Hz	-102.6°	124°	+.002db	No
3	Chebyshev	.092Hz	-122.1°	10.7°	-5.8db	More
4	Inverse Chebyshev	.018Hz	84°	5.9°	6db	More

Table 5.5: Analysis of Bode Plot for HPF's using AHA Recommendation

5.4 Output ECG Wave Shapes from HPF's

The input ECG signal and output of Filter are shown in Figure 5.9, 5.10, 5.11, 5.12, 5.13.

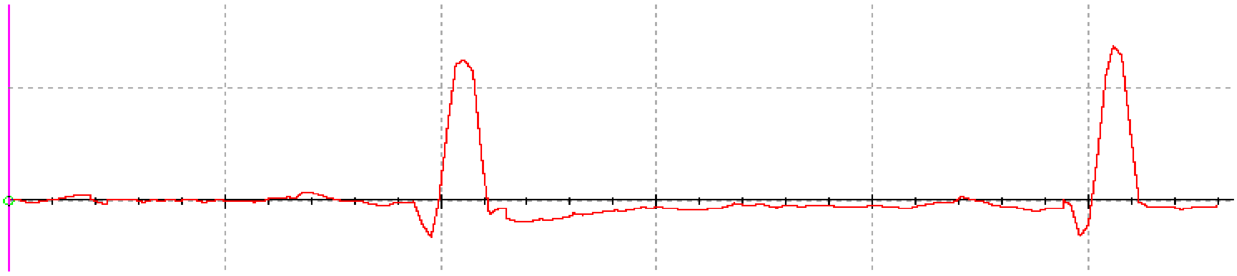


Figure 5.9: Input ECG signal given to HPF's

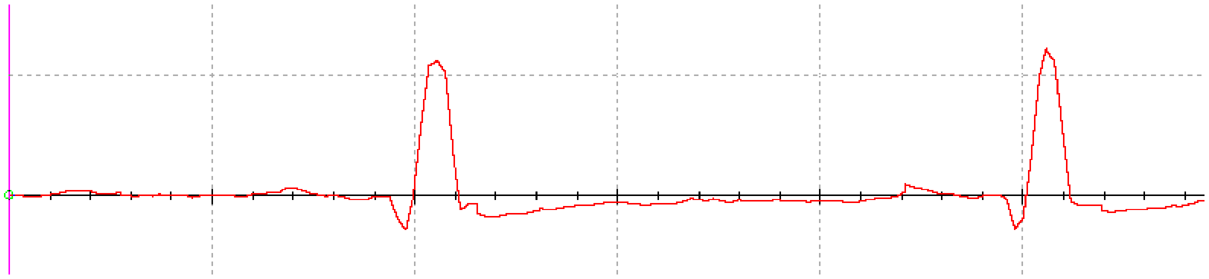


Figure 5.10: Output ECG signal from 1st order HPF

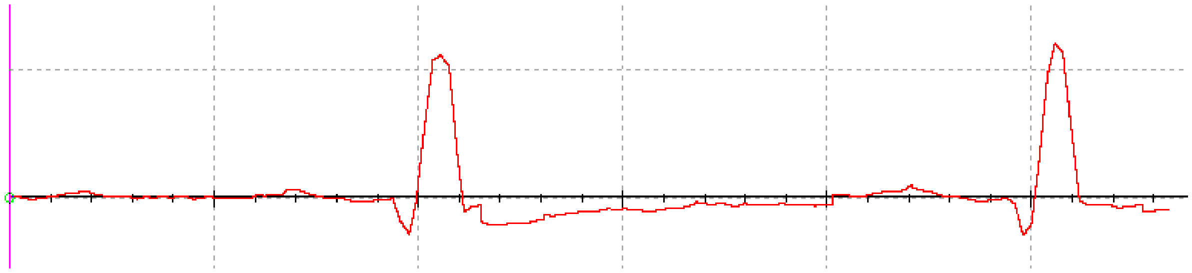


Figure 5.11: Output ECG signal from Butterworth HPF

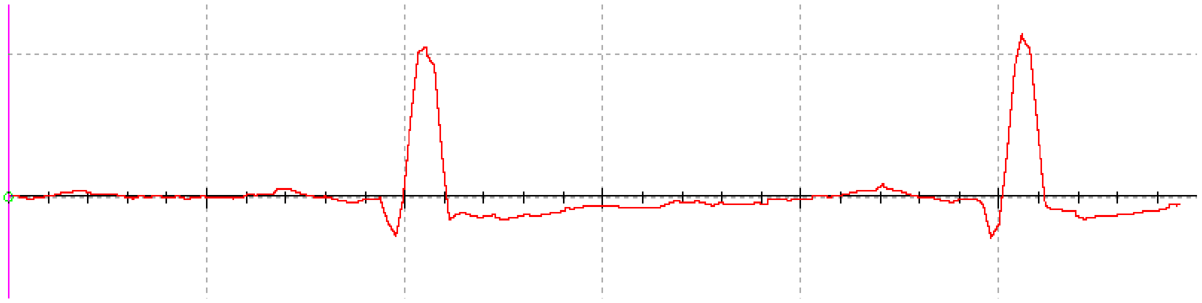


Figure 5.12: Output ECG signal from Chebyshev HPF

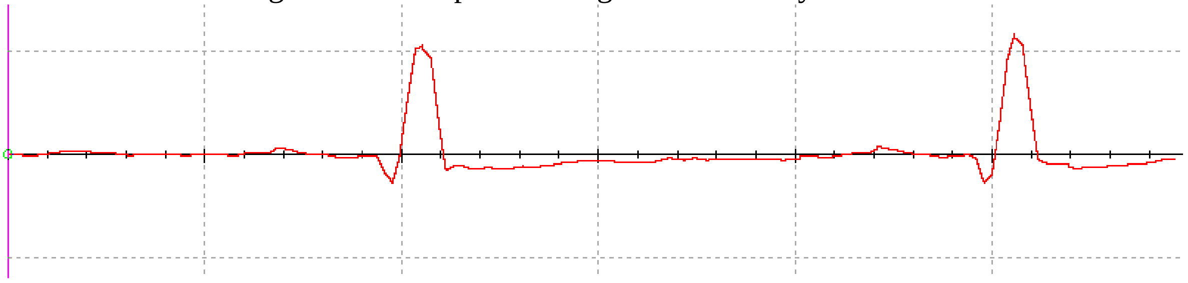


Figure 5.13: Output ECG signal from Inverse Chebyshev HPF

5.5 Bode Plot Response of Low Pass Filters

5.5.1 Bode Plot Response of 1st Order LPF

- Magnitude Versus Frequency Response are shown in Figure 5.14

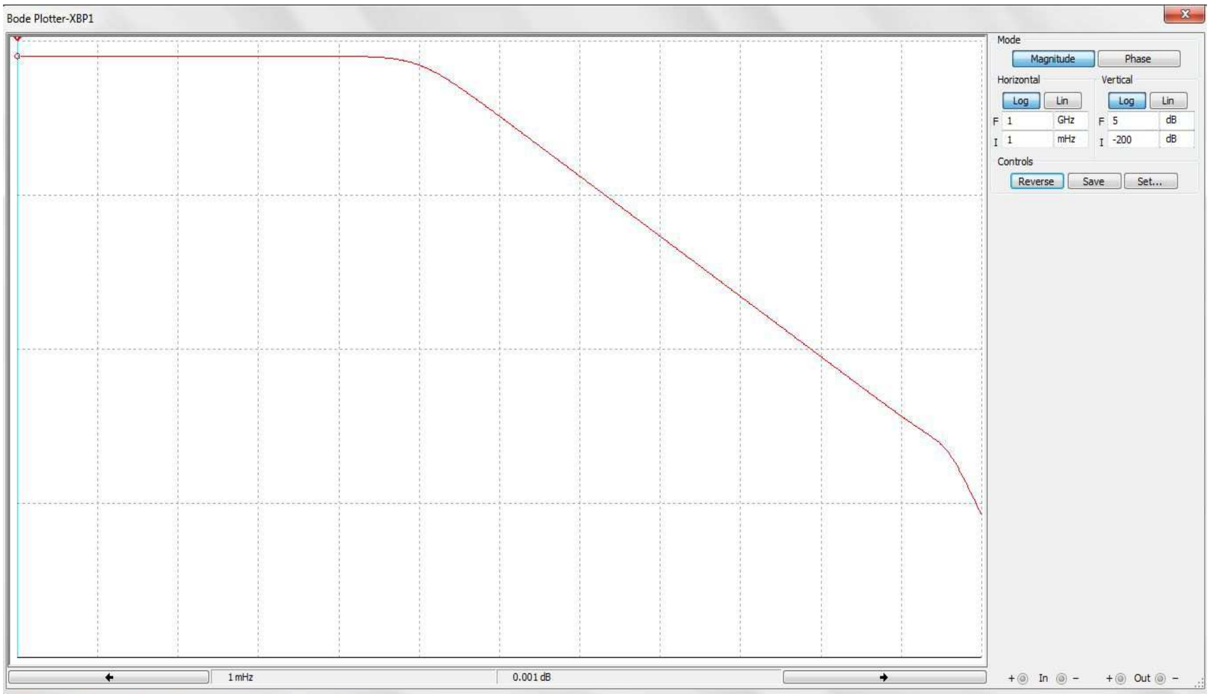


Figure 5.14: Magnitude versus Frequency Response of 1st order LPF

- The Phase Angle versus Frequency Response is shown in Figure 5.15

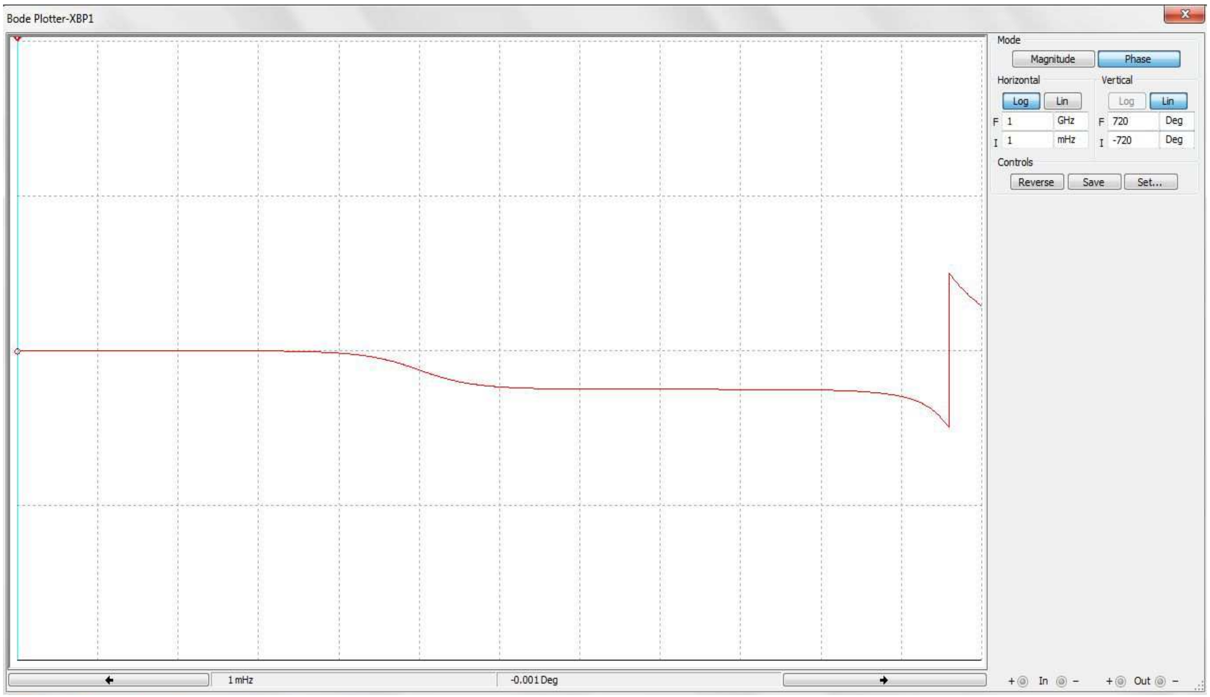


Figure 5.15: Phase Angle versus Frequency Response of 1st order LPF

The Phase Margin and gain margin of 1st order LPF is shown in Table 5.6

Cut off Frequency	**Phase Margin**	**Gain Margin**
100Hz	-16.9°	Flat

Table 5.6: Phase and Gain Margin of 1st order LPF

5.5.2 Bode Plot Response of Butterworth LPF

- Magnitude Versus Frequency Response are shown in Figure 5.16

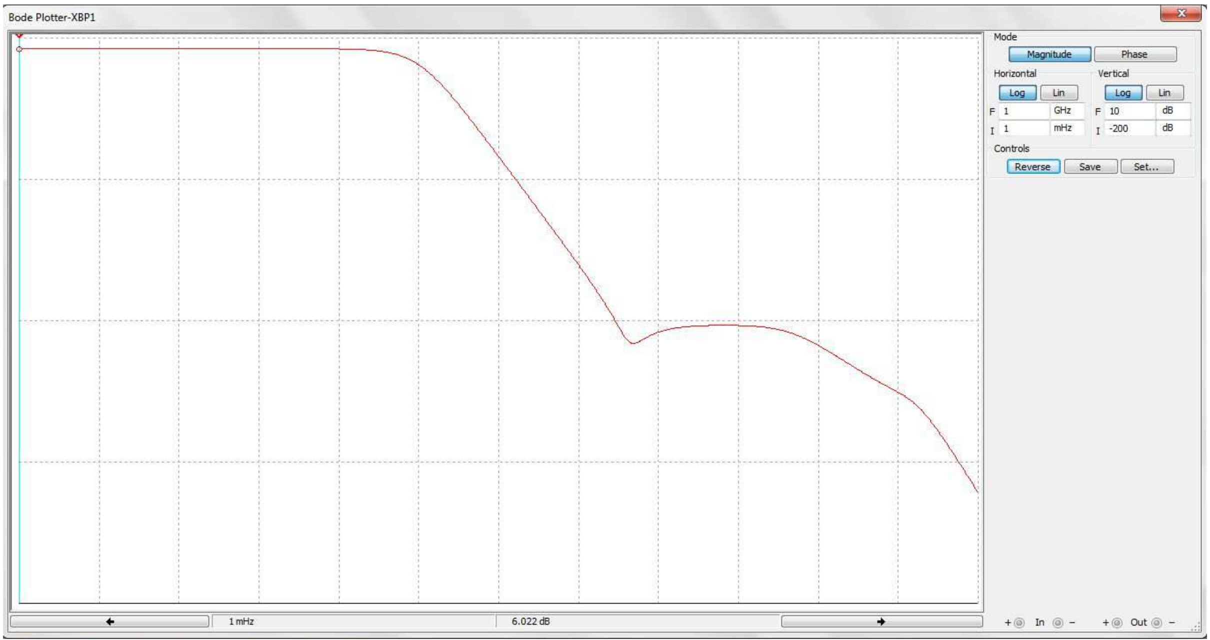


Figure 5.16: Magnitude versus Frequency Response of Butterworth LPF

- The Phase Angle versus Frequency Response is shown in Figure 5.17

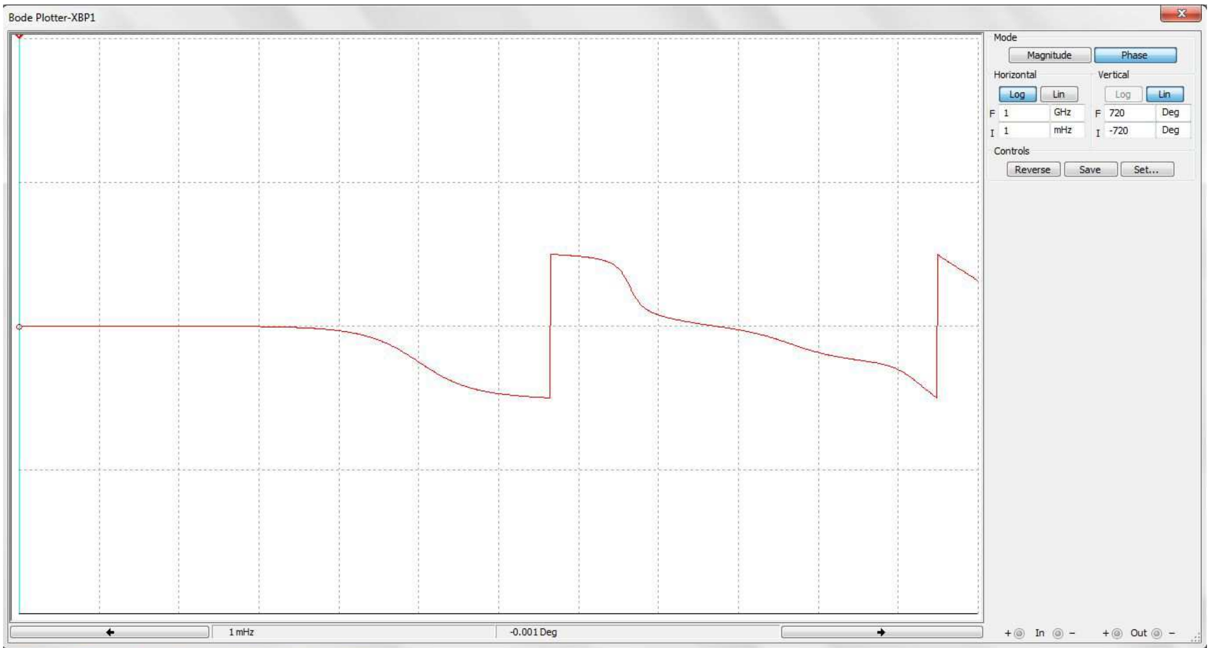


Figure 5.17: Phase Angle versus Frequency Response of Butterworth LPF

The Phase Margin and Gain Margin of Butterworth LPF is shown in Table 5.7

Cut off Frequency	Phase Margin	Gain Margin
100Hz	-34.2°	+.02db

Table 5.7: Phase and Gain Margin of Butterworth LPF

5.5.3 Bode Plot Response of Chebyshev LPF

- Magnitude Versus Frequency Response are shown in Figure 5.18

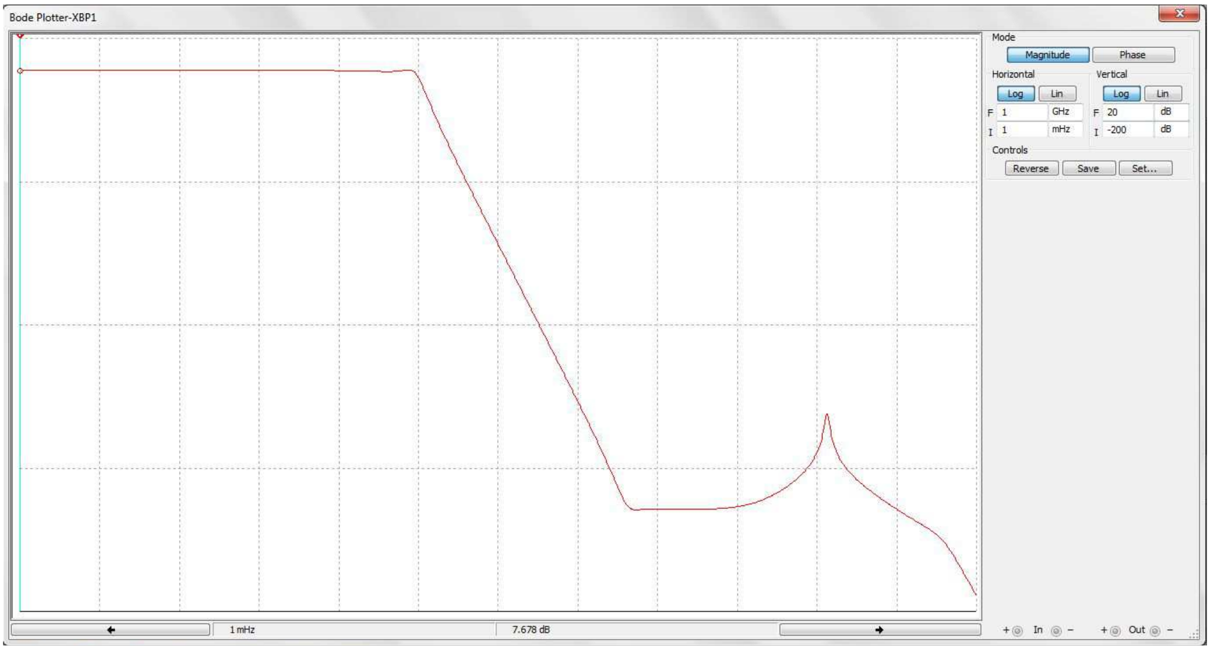


Figure 5.18: Magnitude versus Frequency Response of Chebyshev LPF

- The Phase Angle versus Frequency Response is shown in Figure 5.19

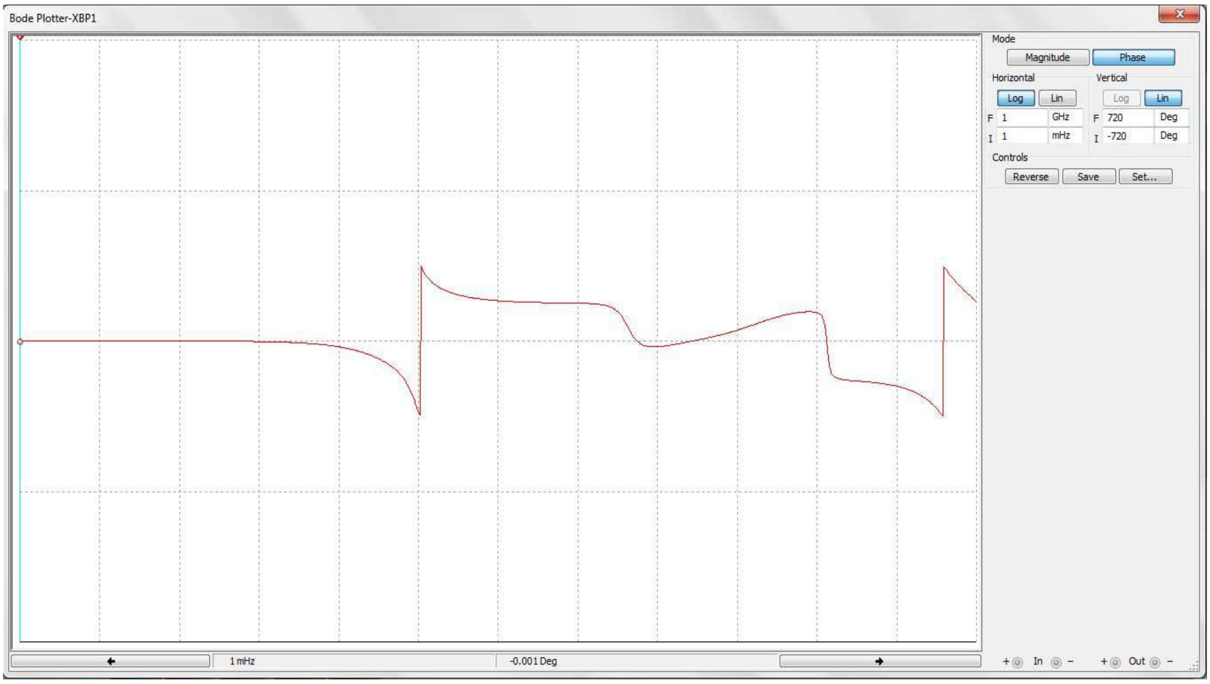


Figure 5.19: Phase Angle versus Frequency Response of Chebyshev LPF

The Phase Margin and Gain Margin of Chebyshev LPF is shown in Table 5.8

Cut off Frequency	Phase Margin	Gain Margin
100Hz	-41.7°	+.2to.6db

Table 5.8: Phase and Gain Margin of chebyshev LPF

5.5.4 Bode Plot Response of Inverse Chebyshev LPF

- Magnitude Versus Frequency Response are shown in Figure 5.20

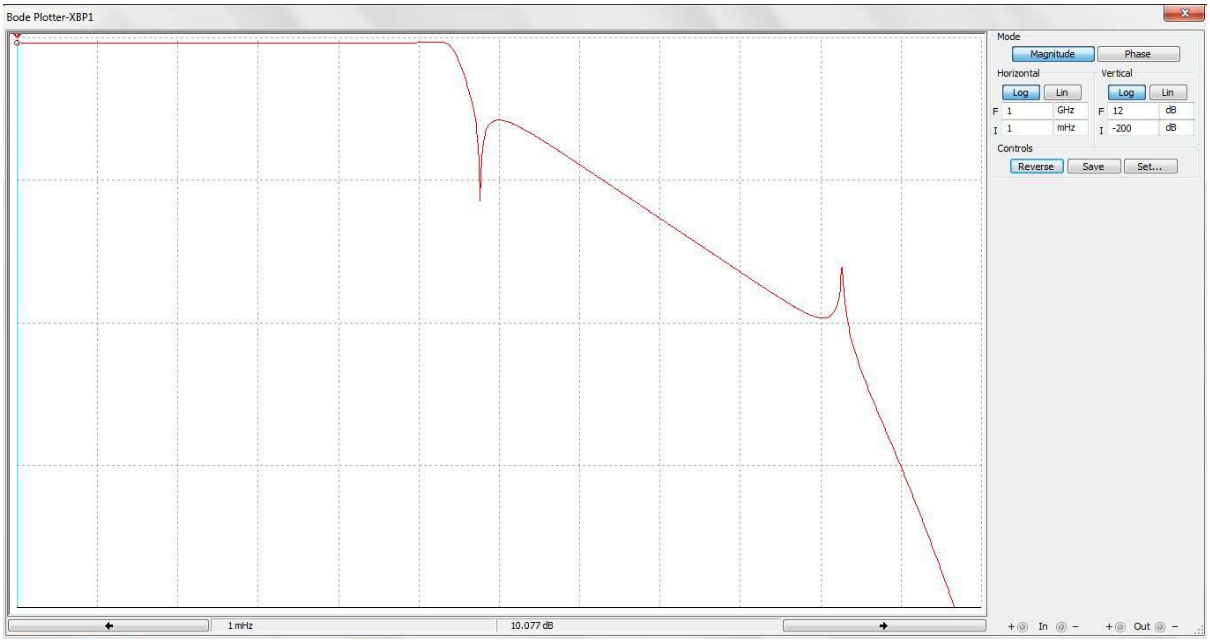


Figure 5.20: Magnitude versus Frequency Response of Inverse Chebyshev LPF

- The Phase Angle versus Frequency Response is shown in Figure 5.21

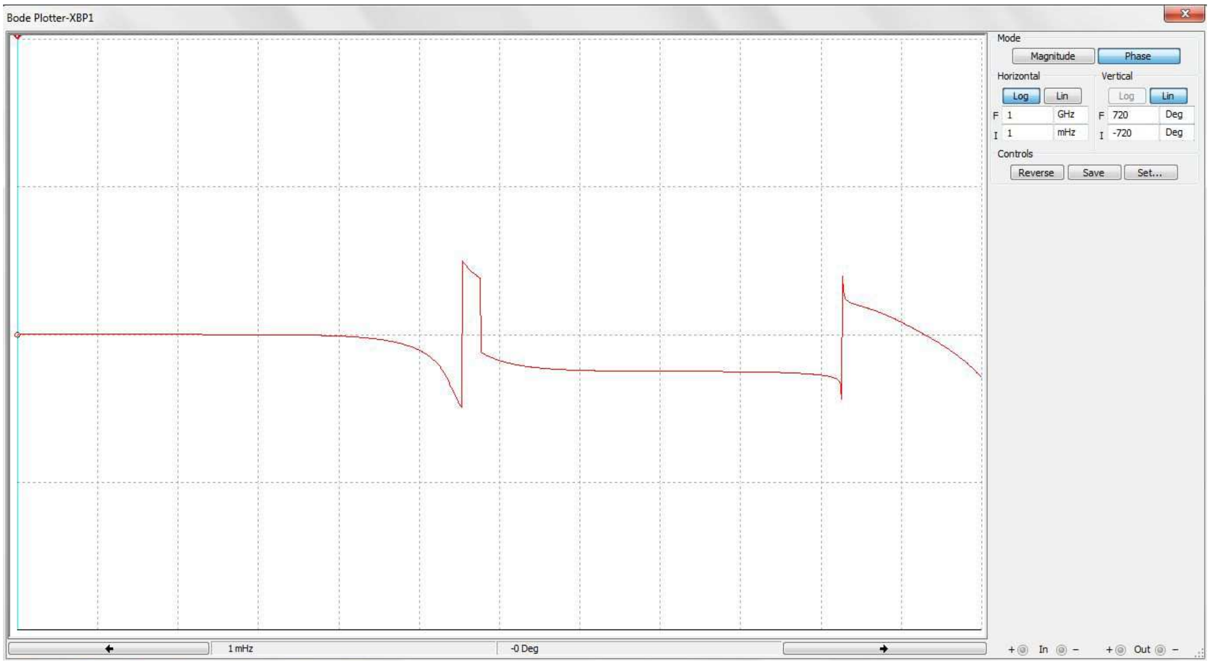


Figure 5.21: Phase Angle versus Frequency Response of Inverse Chebyshev LPF

The Phase Margin and Gain Margin of Inverse Chebyshev LPF is shown in Table 5.9

Cut off Frequency	**Phase Margin**	**Gain Margin**
100Hz	-11.2°	+.2db to .6db

Table 5.9: Phase and Gain Margin of Inverse chebyshev LPF

5.6 Analysis of Bode Plot Response of LPF's using AHA Recommendation

The Analysis of Bode Plot for all LPF's are shown in Table 5.10

Sr. no.	**Filter**	f_{pass} **at -3db point**	**Phase Shift at 30Hz**	**Phase Shift at 100Hz**	**Max Gain Deviation upto 30Hz**	**Ripples Presence**
1	1^{st} order	107.9Hz	-16.9°	-45.1°	Flat	No
2	Butterworth	137Hz	-34.2°	-91°	+0.02db	No
3	Chebyshev	131Hz	-41.7°	-169°	+.2to +.8db	More
4	Inverse Chebyshev	374Hz	-11.2°	-38.9°	+.2to +.6db	Moderate

Table 5.10: Analysis of Bode Plot for HPF's using AHA Recommendation

5.7 Output ECG Wave Shapes from LPF's

The input ECG signal and output of Filter are shown in Figure 5.22, 5.23, 5.24, 5.25, 5.26.

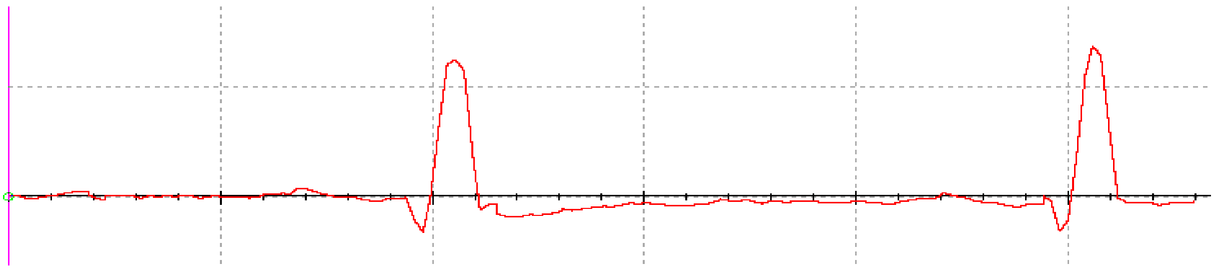


Figure 5.22: Input ECG signal given to LPF's

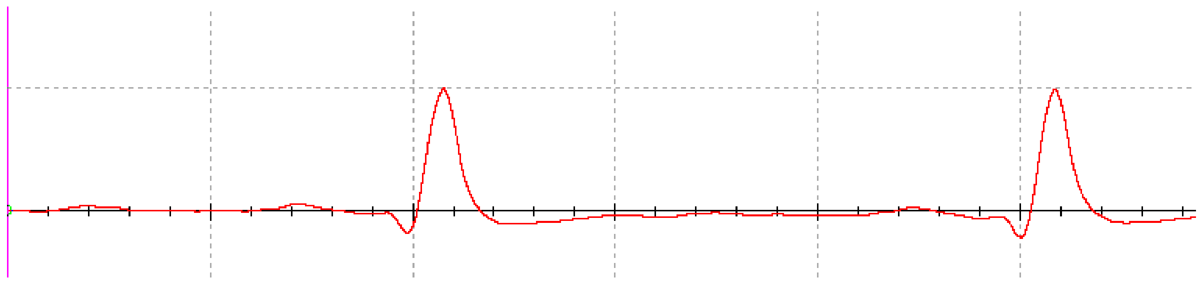


Figure 5.23: Output ECG signal from 1^{st} order LPF

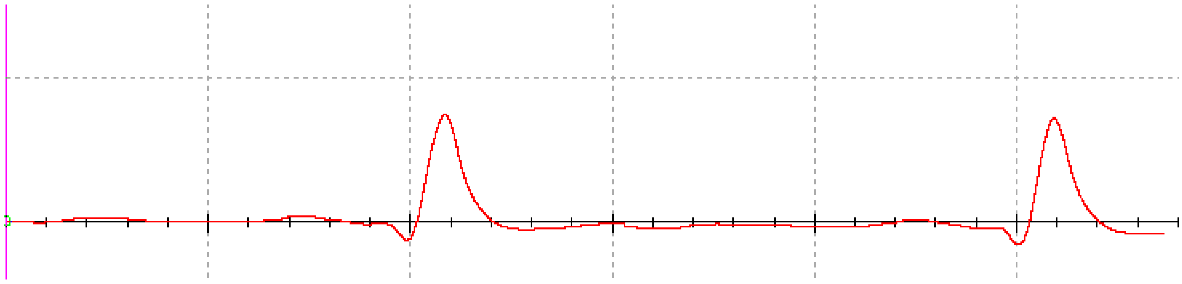


Figure 5.24: Output ECG signal from Butterworth LPF

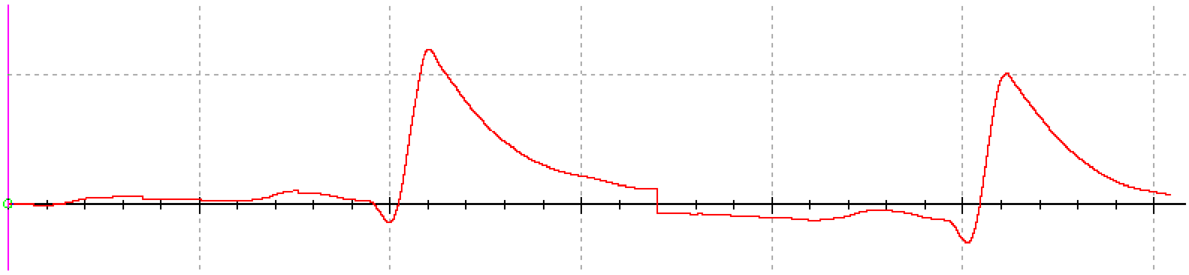


Figure 5.25: Output ECG signal from Chebyshev LPF

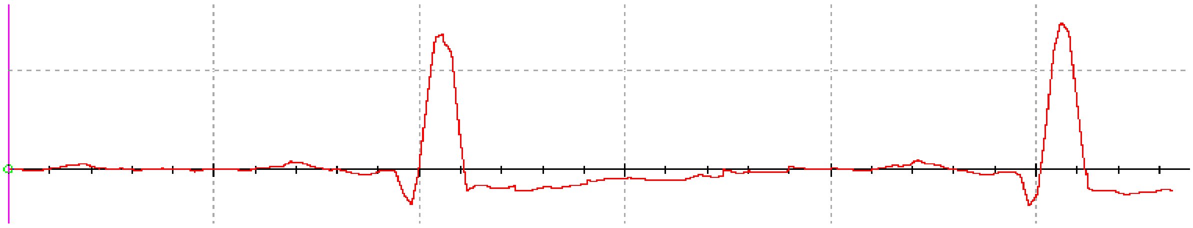


Figure 5.26: Output ECG signal from Inverse Chebyshev LPF

5.8 FFT Analysis

5.8.1 FFT analysis of HPF's

- The FFT of Input signal is shown in Figure 5.27

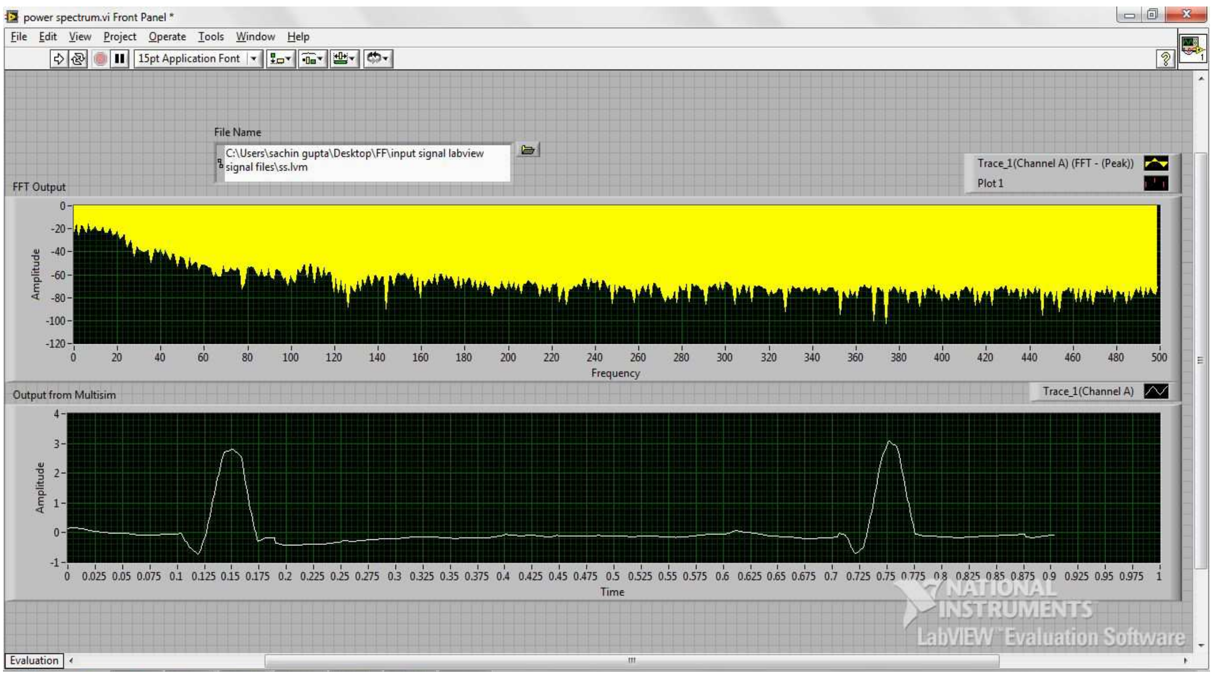


Figure 5.27: FFT of Input Signal (Given to Filter's)

- The FFT of Output ECG signal from 1st order HPF is shown in Figure 5.28

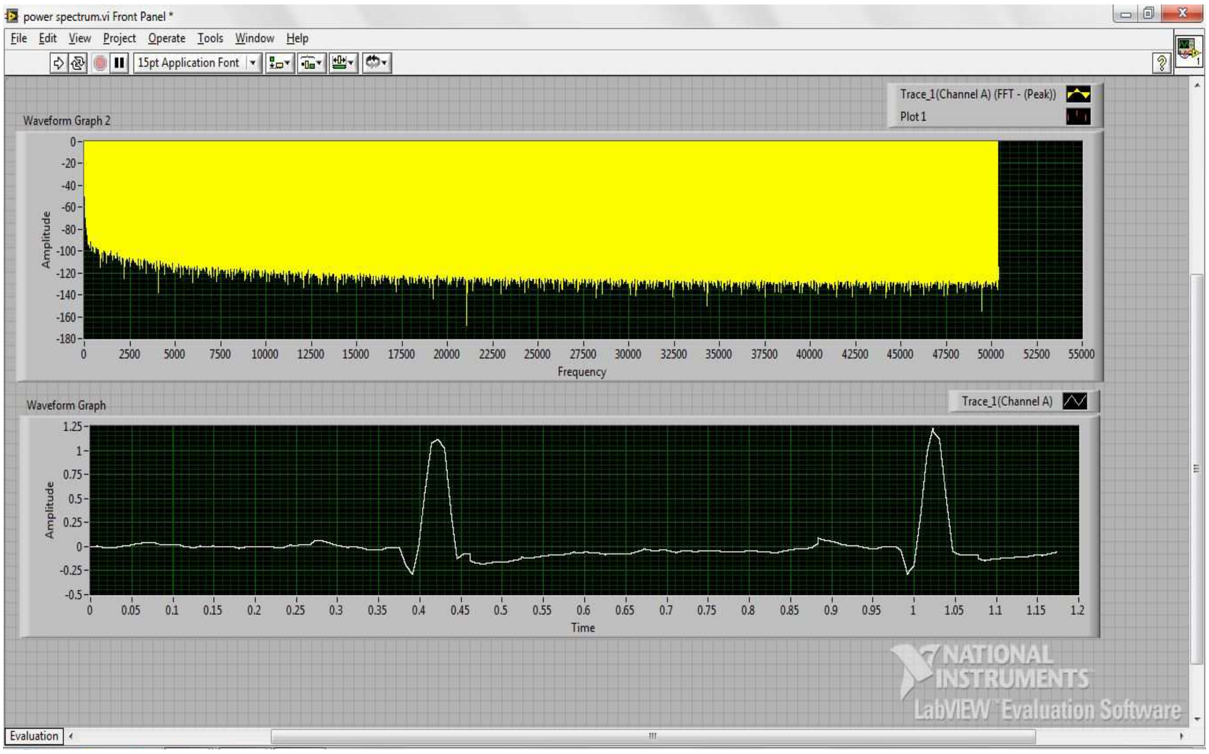


Figure 5.28: FFT of Output ECG signal from 1st order HPF

- The FFT of Output ECG signal from Butterworth HPF is shown in Figure 5.29

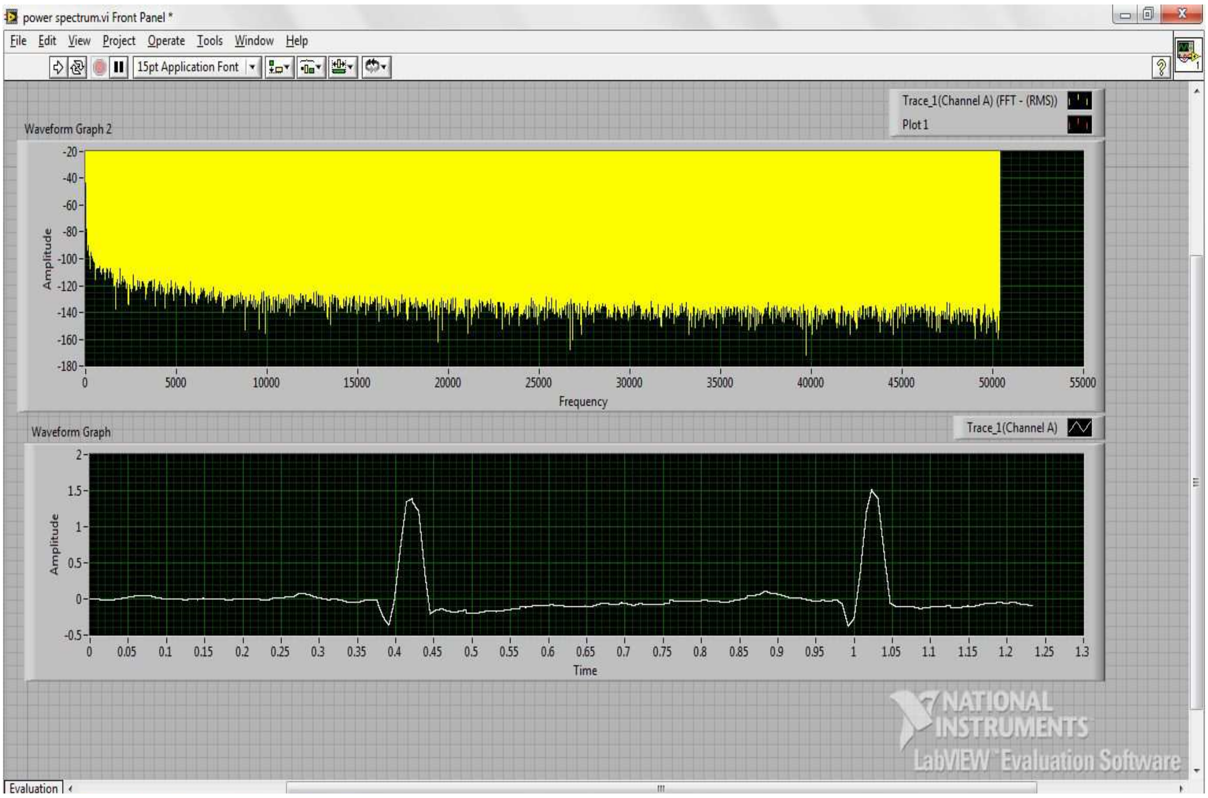


Figure 5.29: FFT of Output ECG signal from Butterworth HPF

- The FFT of Output ECG signal from Chebyshev HPF is shown in Figure 5.30

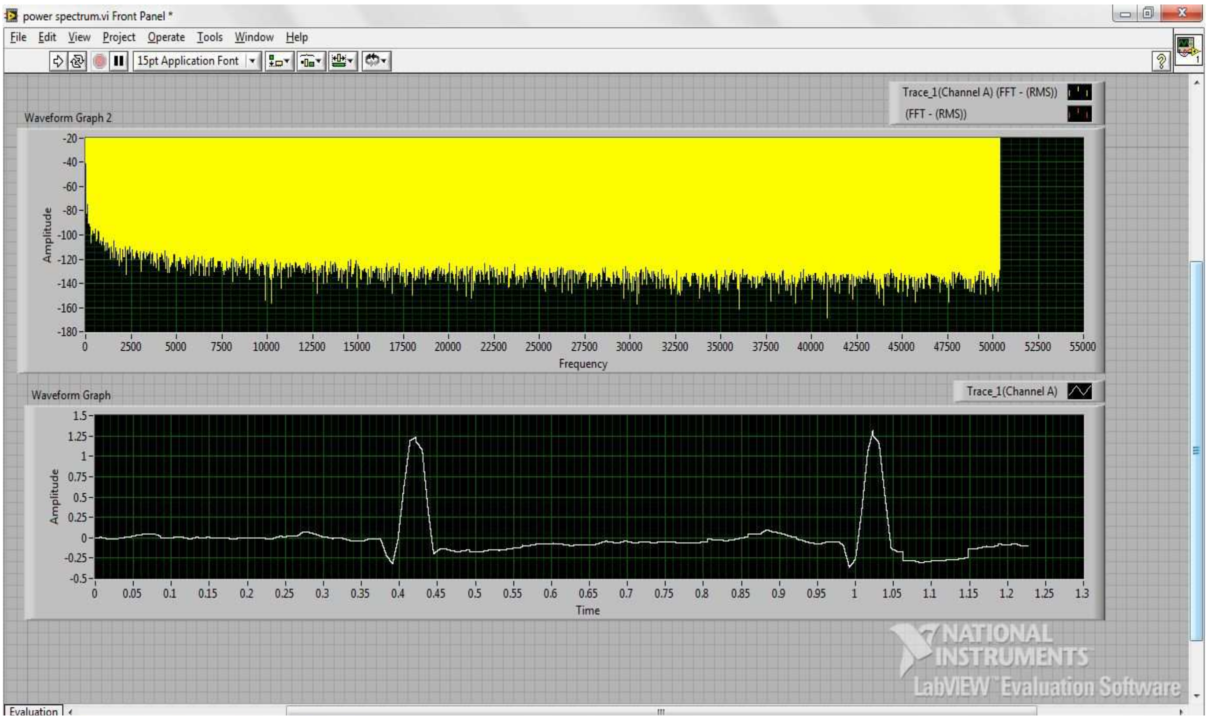


Figure 5.30: FFT of Output ECG signal from Chebyshev HPF

- The FFT of Output ECG signal from Inverse Chebyshev HPF is shown in Figure 5.31

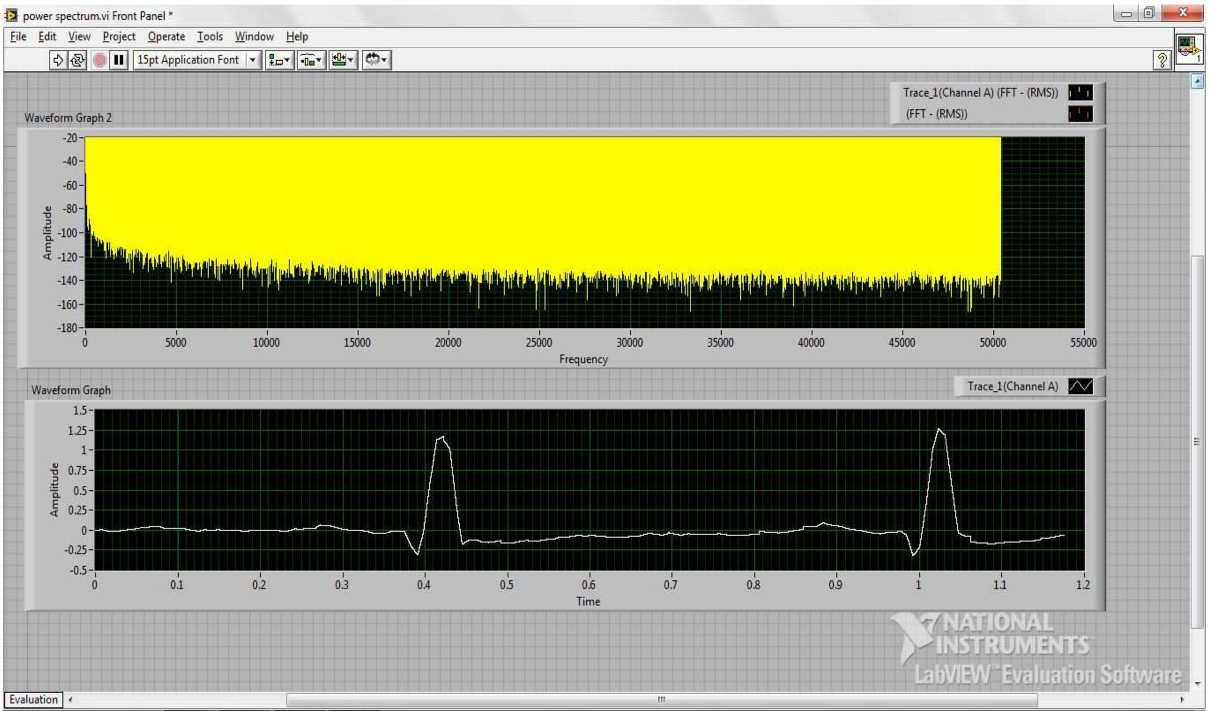


Figure 5.31: FFT of Output ECG signal from Inverse Chebyshev HPF

5.8.2 FFT Analysis of LPF's

- The FFT of Input signal is shown in Figure 5.32

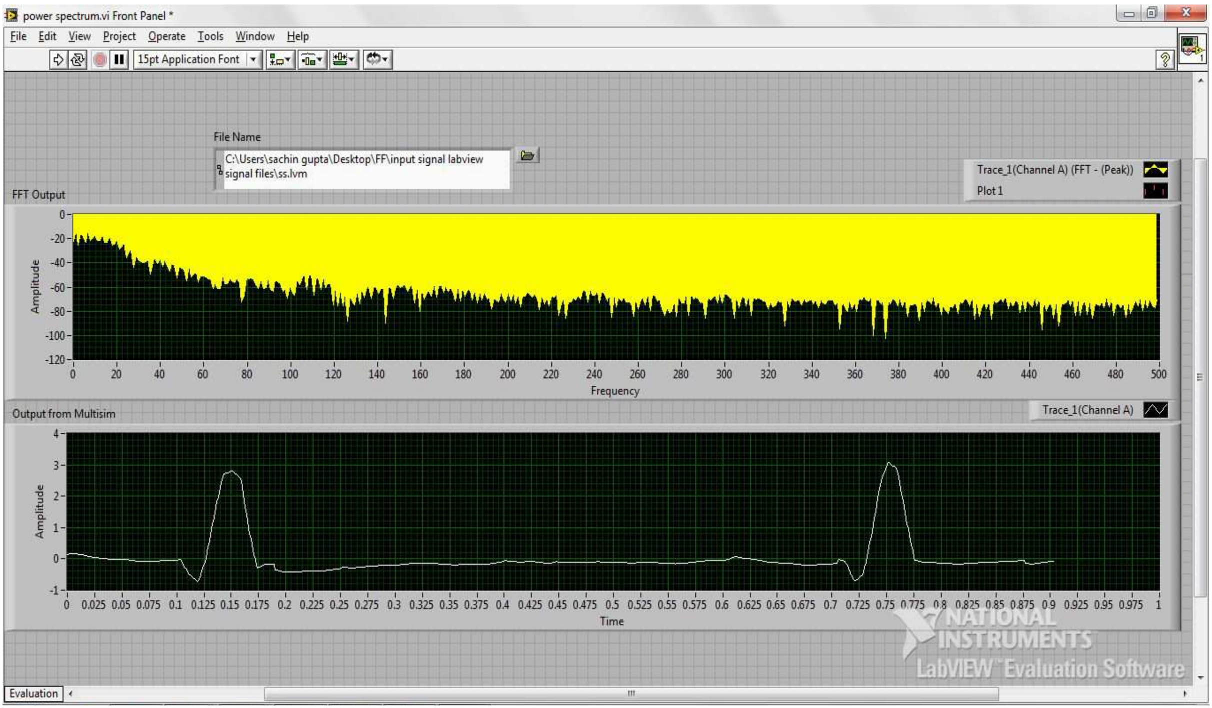


Figure 5.32: FFT of Input Signal (Given to Filter's)

- The FFT of Output ECG signal from 1st order LPF is shown in Figure 5.33

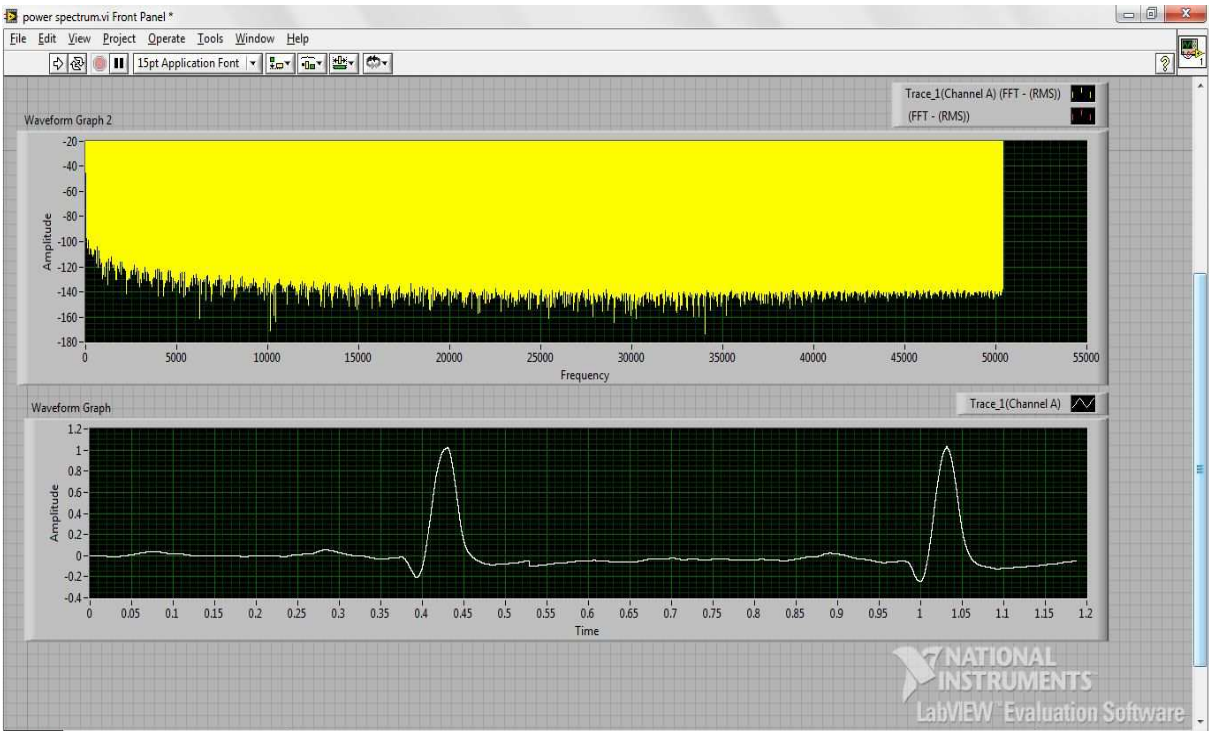


Figure 5.33: FFT of Output ECG signal from 1st order LPF

- The FFT of Output ECG signal from Butterworth LPF is shown in Figure 5.34

Figure 5.34: FFT of Output ECG signal from Butterworth LPF

- The FFT of Output ECG signal from Chebyshev LPF is shown in Figure 5.35

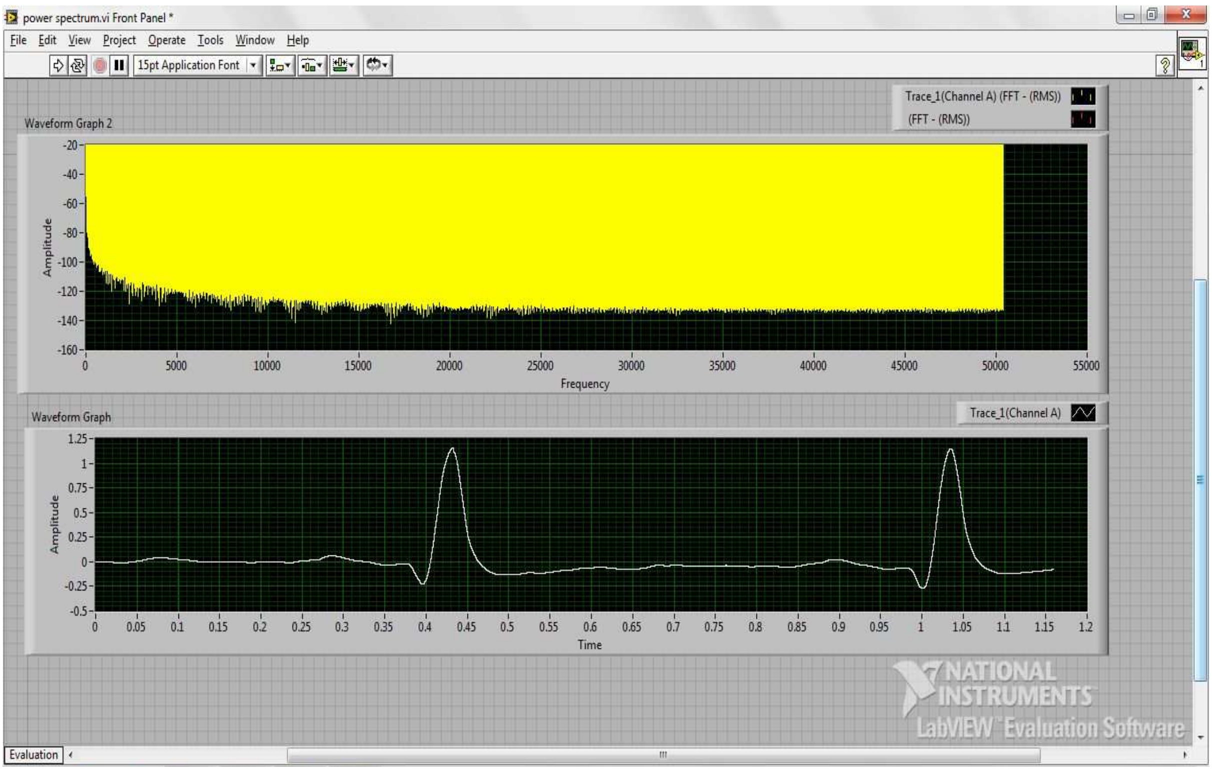


Figure 5.35: FFT of Output ECG signal from Chebyshev LPF

- The FFT of Output ECG signal from Inverse Chebyshev LPF is shown in Figure 5.36

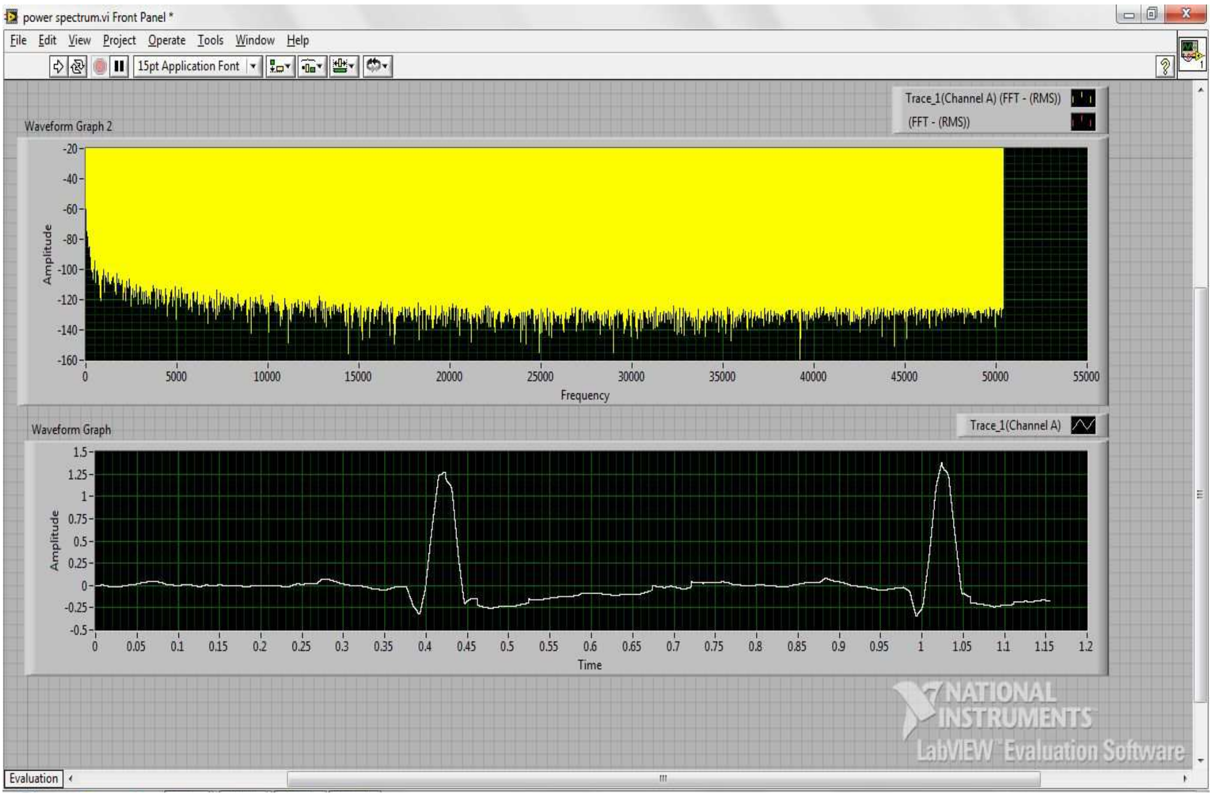


Figure 5.36: FFT of Output ECG signal from Inverse Chebyshev LPF

All the analyses were done on these filters and the best filters among the HPF's and LPF's were concluded.

Chapter 6

CONCLUSION AND FUTURE WORK

6.1 Conclusion

As ECG is the most vital parameter to diagnose the condition of Heart and identifying the disorders. From the background study of ECG Monitoring systems, conventionally standard analog filters (single order) are used. The frequency range of these systems is different for ECG signal acquisition. We designed and analysed the filters using AHA recommendations. The purpose of analytical study is to investigate the best filter response. We performed the Bode Plot analysis and the filters were tested by giving ECG signal and then FFT of filter's output signal was taken. After these analyses, we have observed that HPF's 1st order filter has good response and in LPF's Inverse Chebyshev Filter has good response in comparison to other filters for ECG filtering.

6.2 Discussion - Complete System Design

From the analysis, 1st order HPF and Inverse Chebyshev LPF have the good response. By integrating these filters with Instrumentation Amplifier AD8429, a complete system was designed. The system is shown in Figure 6.1

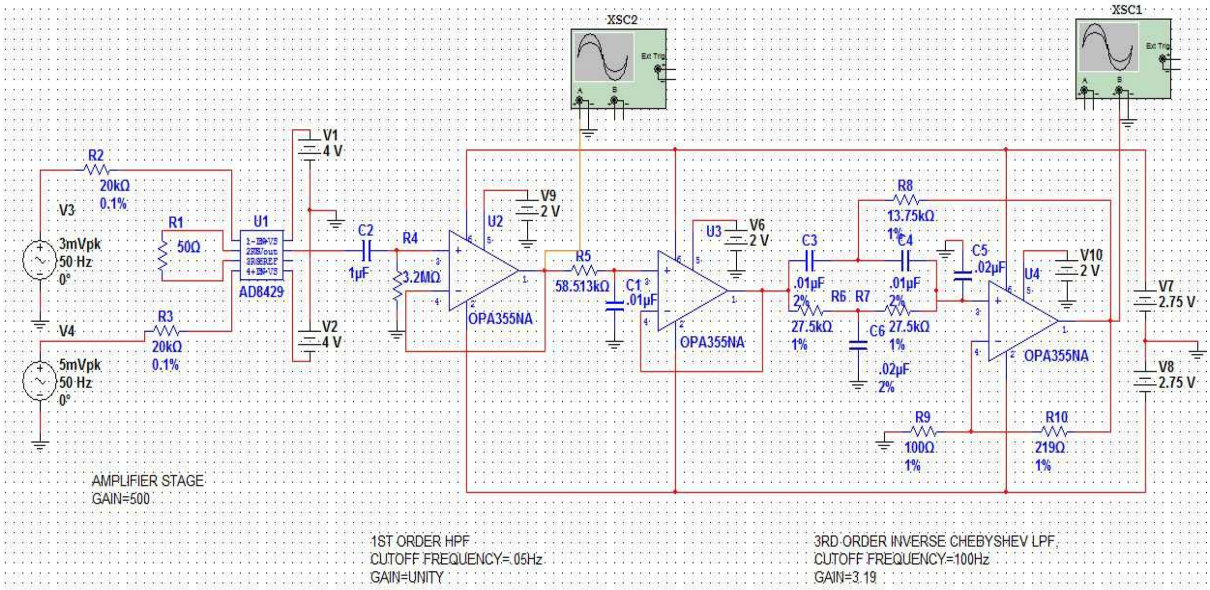


Figure 6.1: Complete System Design

6.2.1 Bode Plot Response

- Magnitude Versus Frequency Response are shown in Figure 6.2

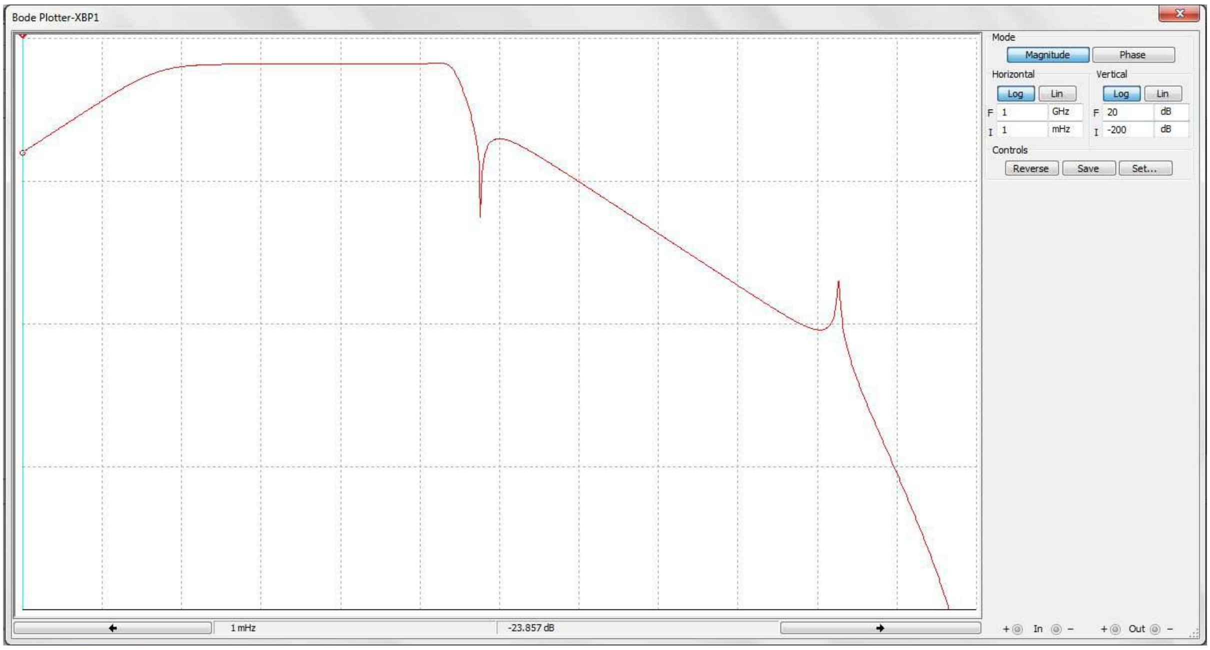


Figure 6.2: Magnitude versus Frequency Response Complete System Design

- The Phase Angle versus Frequency Response is shown in Figure 6.3

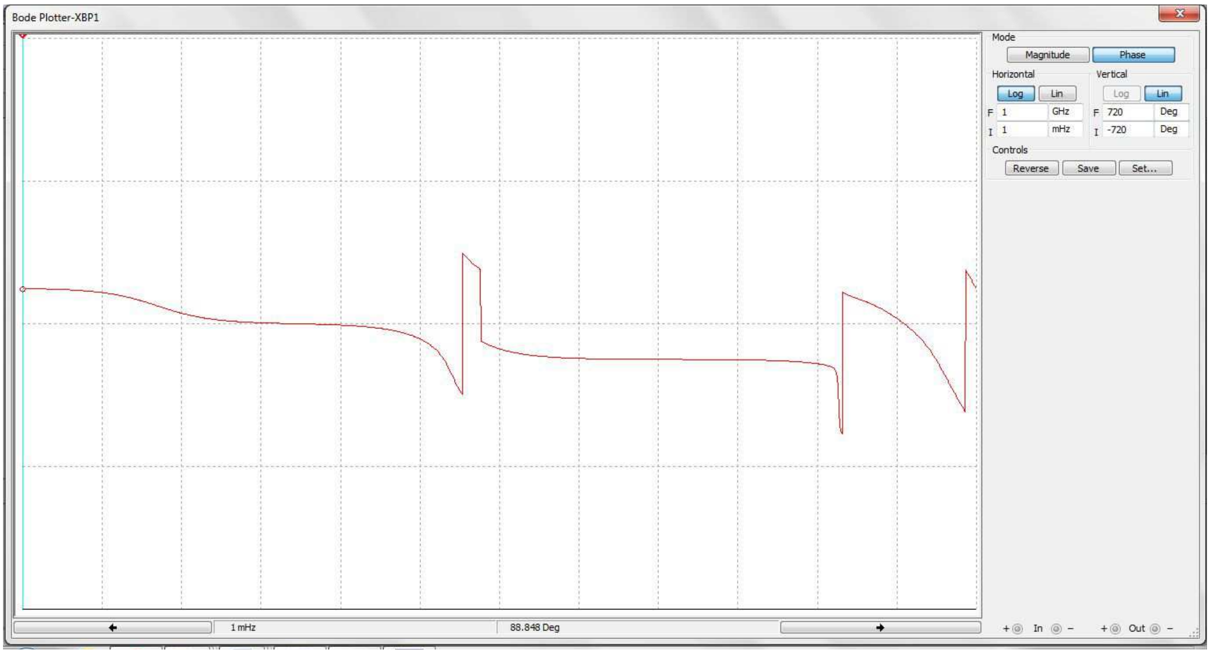


Figure 6.3: Phase Angle versus Frequency Response of Complete System Design

6.2.2 Output ECG Wave Shape from Complete System Design

Here an input ECG signal and Output of ECG signal from complete system design is shown in Figure 6.4, 6.5

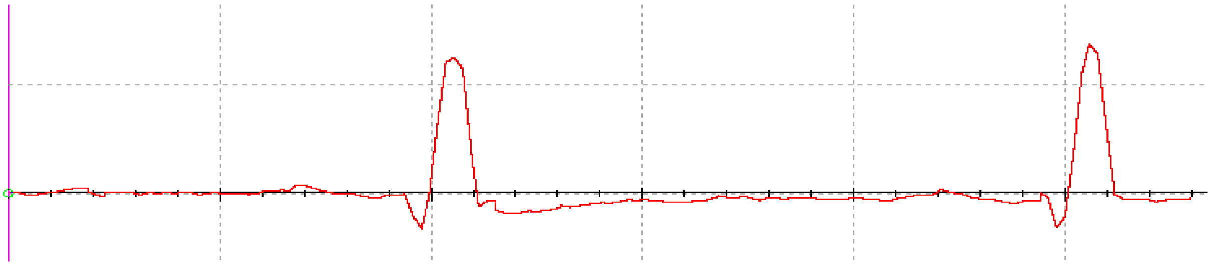


Figure 6.4: Input ECG signal to Complete System Design

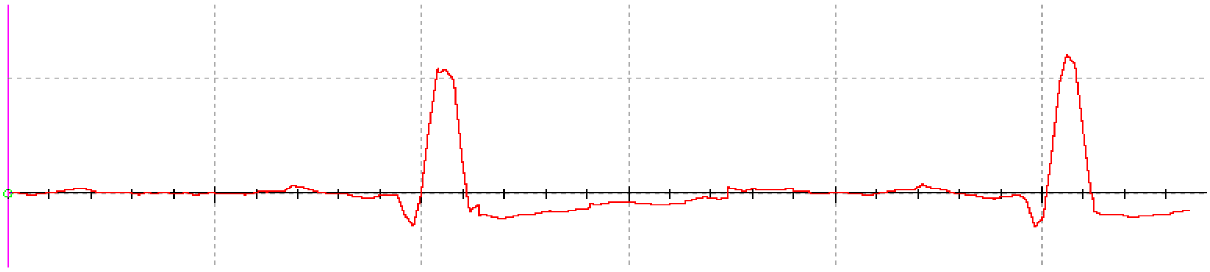


Figure 6.5: Output ECG signal from Complete System Design

6.2.3 FFT analysis of Complete System Design

The FFT analysis of Complete System design is shown in Figure 6.6

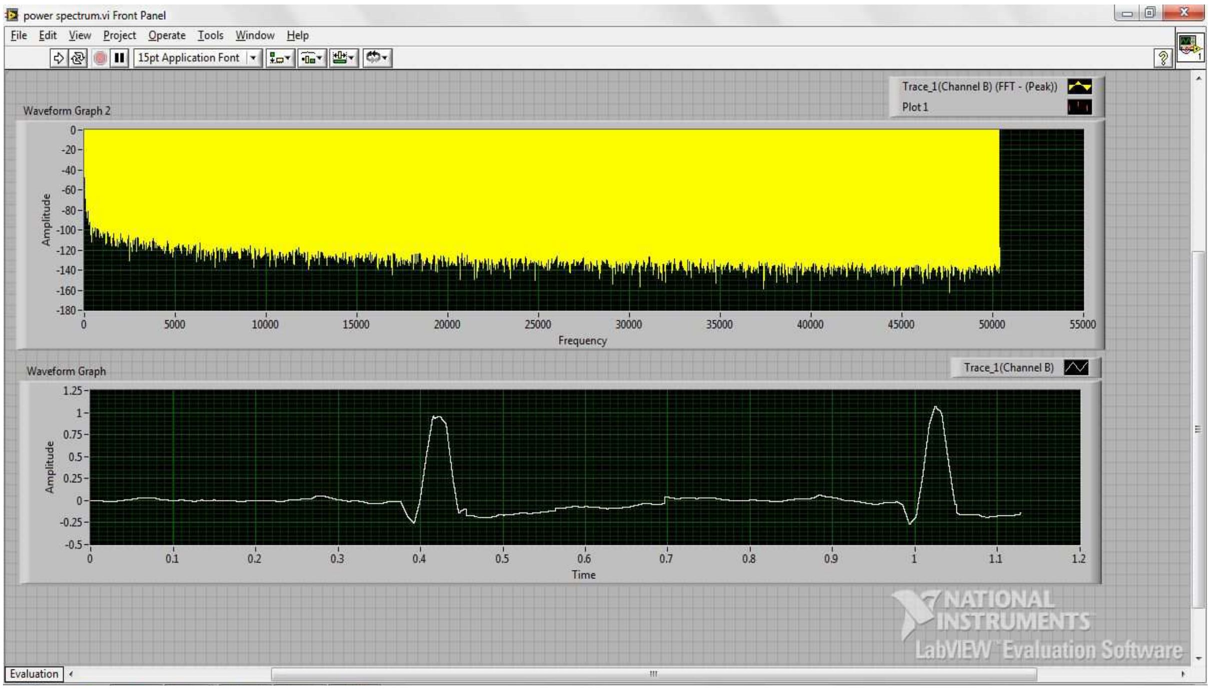


Figure 6.6: FFT of Output ECG signal from Complete System Design

6.3 Future Work

In this thesis, the single-lead ECG sampled at 200Hz and digitized in 12 bits was taken. For further more evaluation, these Filters can be tested using more number of ECG signals which contains information regarding various disorders and frequency components.

High-frequency PCB layout should be employed for these filters. Generous use of ground planes, short direct signal traces, and a suitable bypass capacitor located at the V+ pin of the filters will assure clean, stable operation. It will require more consideration to heat dissipation and low harmonic dissipation.

From the FFT analysis, more feature extraction algorithms can be employed. For example; like biometrics based human identification, FFT can also be used for the same.

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