IMPROVEMENTS IN ELECTROCARDIOGRAPHY SMOOTHENING AND AMPLIFICATION

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Abstract - The electrocardiogram (ECG or EKG) is a graphic recording of the time-variant voltages produced by the myocardium during the cardiac cycle. The P, QRS, and T waves reflect the rhythmic electrical depolarization and re-polarization of the myocardium associated with the contractions of the atria and ventricles. The electrocardiogram is generally used clinically in diagnosing diseases of the heart. Hence, it must be very accurate. The ECG waveform is a periodic signal with bandwidth of 0.05 Hz to 100 Hz. Amplitude is typically 1 milli-volt peak to peak in the presence of much larger (1000 times larger) external high frequency noise plus 50/60 Hz interference common mode voltages (common to all electrode signals). We present a method to eliminate much of the noise using a pre-amplifier design with high common mode rejection ratio and high input impedance. We verify our results using computer simulation of the signal via the software MULTISIM 9.0.

1. Introduction

The electrocardiogram (ECG or EKG) is a graphic recording or display of the time-variant voltages produced by the myocardium during the cardiac cycle. [3] It is the physiological measurement of the cardiovascular systems. Cardiovascular system is the transport system of the body, by which food, oxygen, water and all other essentials are carried to the tissues and cells and their waste products are carried away. It comprises of blood, blood vessels (arteries, capillaries and veins), and the heart. ECG was originally observed by Waller in 1889 using his pet bulldog as the signal source and the capillary electrometer as the recording device. In 1903, Einthoven enhanced the technology by employing the string galvanometer as the recording device and using human subjects with a variety of cardiac abnormalities. [9]

2. Basic EKG Waveform

The record of the bio-potentials generated by the muscle of the heart is the electrocardiogram and the basic waveform recorded for a normal person is shown below:

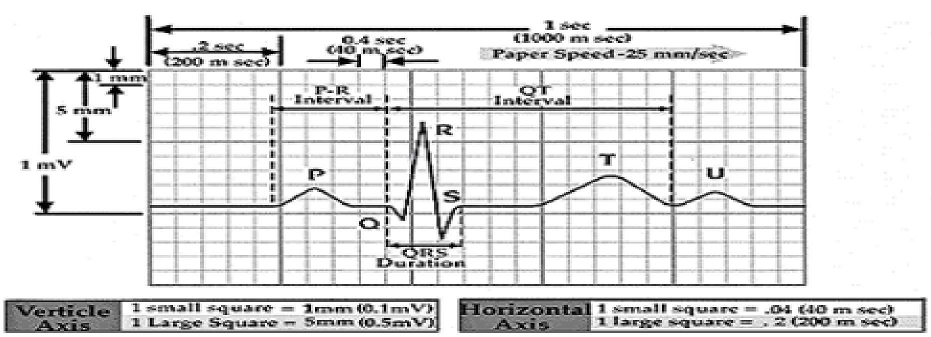

Figure 1: A typical Electrocardiogram waveform [3]

Figure 1 shows a typical ECG as it appears when recorded from the surface of the body. Alphabetic designations have been given to each of the prominent features. These can be identified with events related to the action potential propagation pattern. The horizontal line preceding the P wave is called as the isopotential or the baseline. The P wave represents the depolarization of the atria .The QRS complex is the combined result of the repolarization of the atria and the depolarization of the ventricles, occurring simultaneously. The T wave is the wave of ventricular repolarization whereas the U wave is generally the result of after potentials in the ventricular muscle. The P-Q interval represent the delayed time in the fibers neat the AV node.

Some normal values for the amplitudes and durations of the parameters of the wave are as follows:

Amplitude:	P wave	0.25 mV
	R wave	1.60 mV
	Q wave	25% of R wave
	T wave	0.1 to 0.5 mV
Duration:	P-R interval	0.12 to 0.2 sec
	Q-T interval	0.35 to 0.44 sec
	S-T segment	0.05 to 0.15 sec
	P wave interval	0.11 sec
	QRS interval	0.09 sec

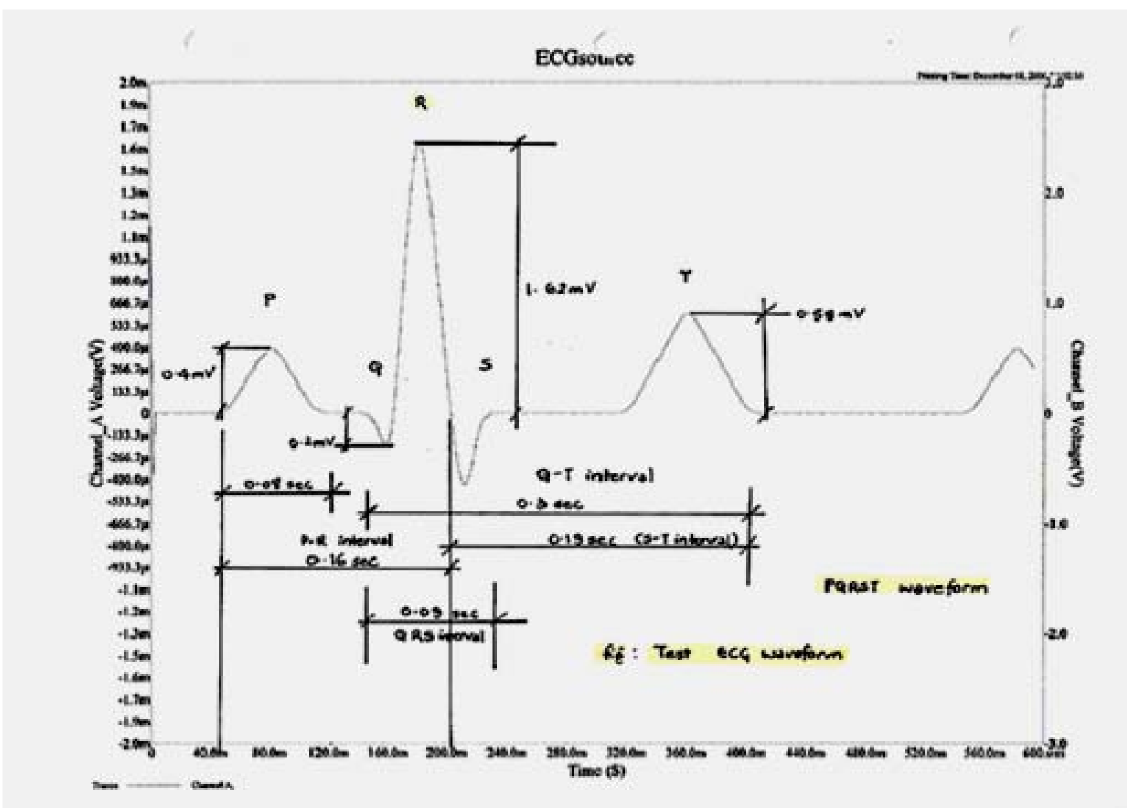

Figure 2: ECG signal used to test the designed circuits

3. Noises in ECG

The main noises found in an ECG waveform are the common mode signals. The common mode voltage (CMV) in ECG is composed of two components:
a. DC electrode offset potential
b. 50 or 60 Hz ac induced interference

This 50 or 60 Hz interference also known as Hum interference is caused by magnetic and electric fields from power lines and transformers cutting across ECG electrodes and patients. Hum currents flow in signal, common, and ground wires via capacitive coupling between the field and the system.

In spite of numerous improvements over the years, noise disturbances have proved hard to remove. *This paper proposes a pre-amplifier design, which is highly successful at minimizing the hum in the ECG recordings and provide gain to them.*

4. Designing

4.1 The ECG Preamplifier

An ECG preamplifier is a differential bioelectric amplifier. Amplifiers used to process bio potentials such as electrocardiogram, electroencephalogram, electromyogram are known as bioelectric amplifiers. The input circuitry consists of the high input impedance input of the bioelectric amplifier, a lead selector switch, a 1-mV calibration source, and a means for protecting the amplifier against high voltage discharges from defibrillators used on the patient. The amplifier may be a bioelectric instrumentation amplifier, though in all modern machines, one of the isolation amplifier designs is used for patient safety.

4.2 ECG Pre-Amplifier Designing

We have used Multisim 9 Student SUITE from Electronics Workbench to perform the simulations of the designed circuits. A design for a 1 Lead ECG pre-amplifier has been proposed in this part. The main high points of the design are the high CMRR (≈ 80 dB) and good frequency range of operation (0.05 Hz to 45 Hz). The circuit also overcomes the dc electrode offset that it might come across.

We have divided the design onto four parts:
a) Input Differential Amplifier
b) Intermediate Differential Amplifier
c) Amplification stage (Common Emitter Amplifier) plus the low output impedance CC stage
d) Filter

During the designing, the stages were analyzed individually and then the complete design analysis was done. It is always a good practice to analyze the individual parts, before going for the complete analysis. The individual circuit property might be different when the additional circuits are cascaded as the next stages. The input impedance of the succeeding stage plays a big role. Hence, it is better to design all the stages with very high input impedance. Here, only the final design has been discussed.
Power rails: ± 25 V

The Bipolar Junction Transistors used for the designing are National 2N3906 (PNP) and Zetex Q2N2222A. All the circuits are tested with a normal ECG waveform shown in Figure 2.

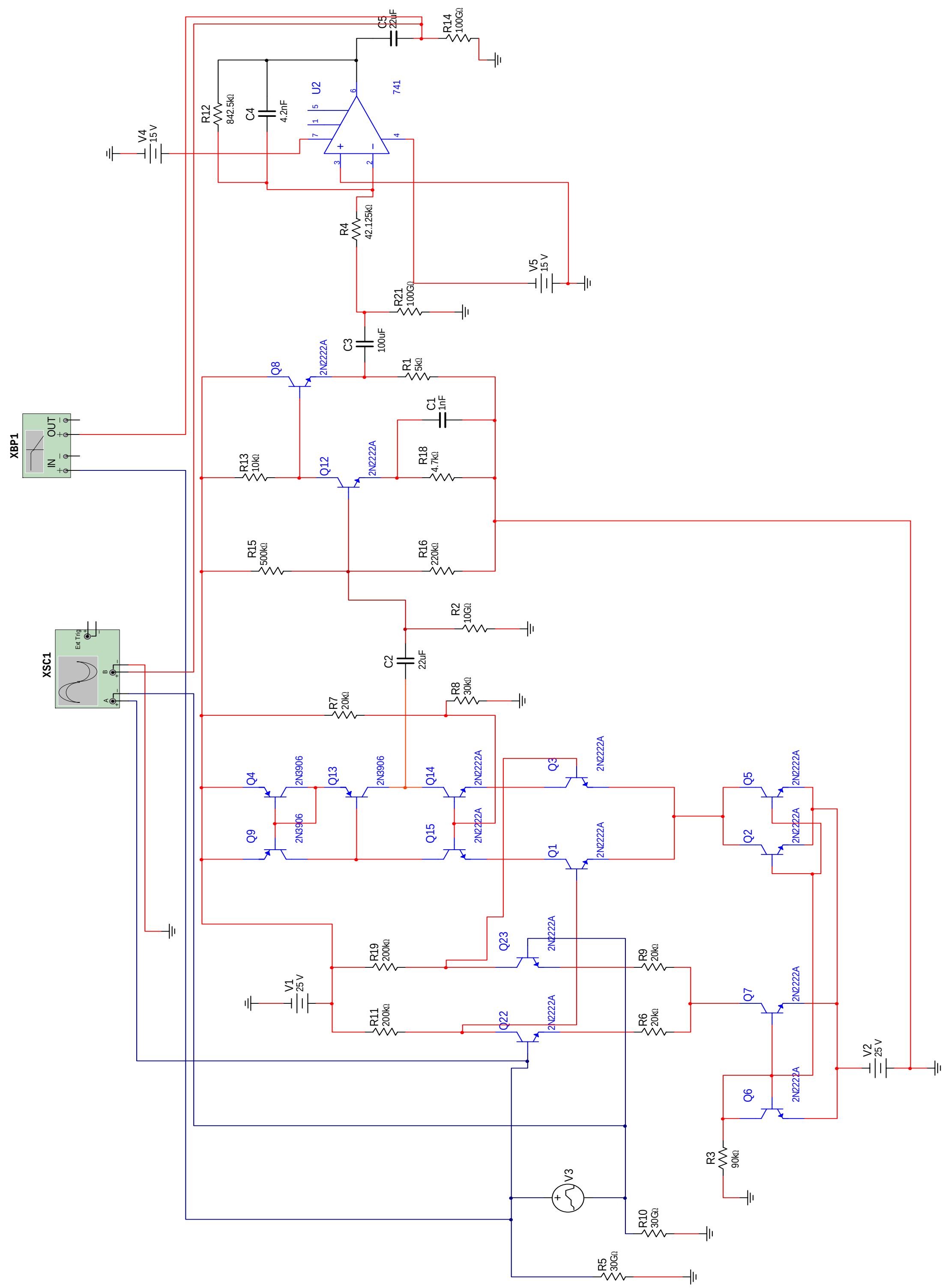

Figure 3: Designed 1 Lead ECG Front End (Monitoring Mode)

Figure 3 shows the designed 1 Lead ECG Front End pre-amplifier used in monitoring mode. The initial two stages (differential amplifiers) are designed with low gain but with high common mode rejection capability keeping in mind the possible presence of high value dc offsets. If the high gain is provided right at the first stage, it might take the other stages ahead into saturation. The main purpose of the first two stages is to convert a noisy differential ECG signal to a single mode noise free signal. The third stage and the filter provide the amplification. The type of filter used depended on the use of the pre-amplifier (monitoring or diagnosing). The design proposed is for a monitoring mode, which required the operation range up to 45 Hz, a low pass filter with a cut off of 45 Hz was used.

Since an ECG signal is a differential signal taken from two points on the body or taken with respect to some reference point, a differential amplifier is used as the first stage. The noises present in an ECG signal are the strong common mode noises such as the 50- or 60-Hz electromagnetic interference and the dc offsets due to the electrode skin contact. Hence the differential amplifier also serves as a common mode noise rejecter as much as anything else. The electromagnetic interference could be very strong (nearly 1V) compared to the weak ECG signal (close to 1mV mark). Hence the initial stages of all the amplifiers must have a very high common mode rejection ratio, which is the ratio of the differential gain to the common mode gain.

The first differential amplifier used in our design is a double-ended input (differential input) double-ended output differential amplifier. The stage was biased with a current source of 0.267 mA to give a differential gain of less than 10. The observed gain from simulation was 3.08. [Simulation results are shown in the next section]. The second stage in the design which changed the double-ended signal to a single ended signal (measured w. r. t. ground) is a cascode differential amplifier. This stage provides the very high differential gain compared to the common mode gain, resulting in a very high Common Mode Rejection Ratio of 80 dB.

The next stage is introduced to amplify the noise free signal. A common emitter amplifier with a gain of -455.68 was designed to this job. But the gain seen in simulation is much less (-2.125) than the analyzed gain due to the presence of an emitter bypass capacitor. This stage was also tested with a higher frequency (1 kHz) signal, for which it gave the gain close to the estimated value. Hence the major gain had to retrieved from the next stage i.e. filter. The low pass filter with a cut off frequency of 45 Hz follows the Common Emitter-Common Collector amplification stage. This restricts the use of the amplifier to the monitoring purpose only as for diagnostic purposes the amplifiers with response higher than 100 Hz are used. The filter used could be replaced by another Low pass filter with higher cutoff frequency along with a 60-Hz notch filter if it is to be used for other than monitoring purpose. The pass band gain for the used filter is -20. Hence the overall gain for the designed amplifier is -56.25. The feedback ratio of the filter could be changed for the change in gain.

The lower and upper 3dB frequency points for the proposed design are 38.5662 mHz and 47.4091 Hz respectively. The main high point of the design is the CMRR of 80 dB, which is very high, compared to the simplicity of the design.

The specifications of the proposed design are listed below (Simulations shown in the next segment):

Amplifier Specifications:

Voltage rails: ± 25 V ±15 V (filter)	Gain: +2.8125 (without filter) -56.25 (with filter)
Input resistance: 8.8 MΩ	Frequency range: 38.5662 mHz - 47.4091 Hz (monitoring mode)
Output resistance (Amplifier): 25.44 Ω	
CMRR at 60 Hz: 77.4 dB	

Table 1: Specifications of the designed ECG preamplifier

4.3 Testing the circuit with an EKG signal generated using MATLAB – Simulation and Results:

The design was checked for various cases with a pure ECG signal as well as with an ECG signal with different noises. It worked very well as it rejected most of the noise and demonstrated high CMRR

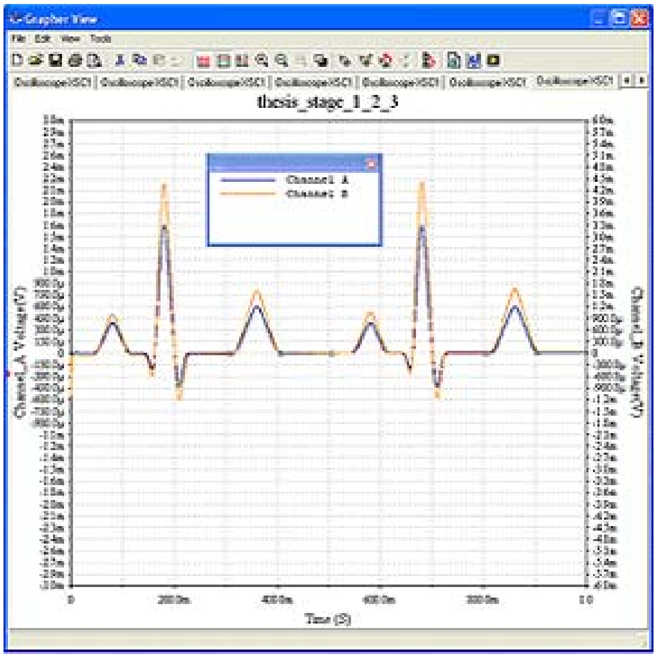

Figure 4: Testing with a normal ECG wave
(I/P and O/P plot – before the filtering stage) Vs Time
Observation: Comparing the R wave peaks
Peak (Channel A –Input) = 1.6 mV
Peak (Channel B –Output) = 4.5 mV
Gain = 2.8125

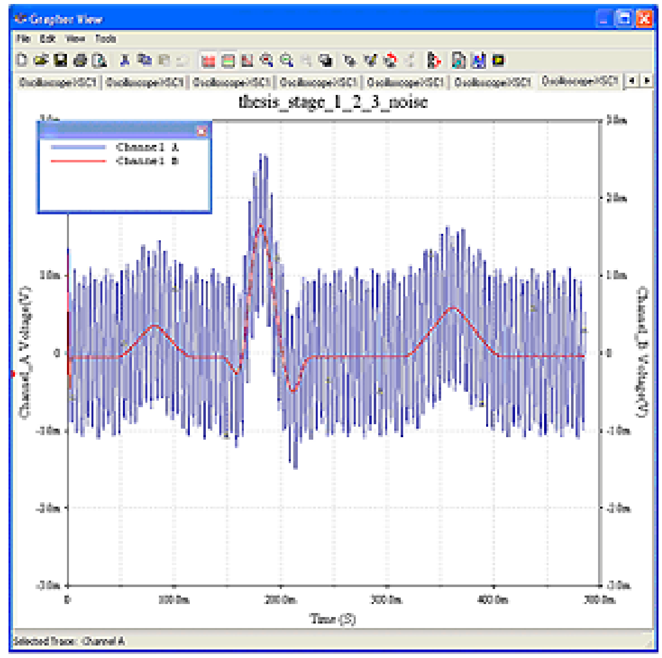

Figure 5: Testing with a noisy ECG wave
(I/P and O/P plot – before the filtering stage) Vs Time
Observation: Channel 1: ECG + [0.1mV peak 60 Hz sine wave + 1mV peak 200 Hz sine wave]-common mode noise
Channel 2: Smooth amplified ECG waveform

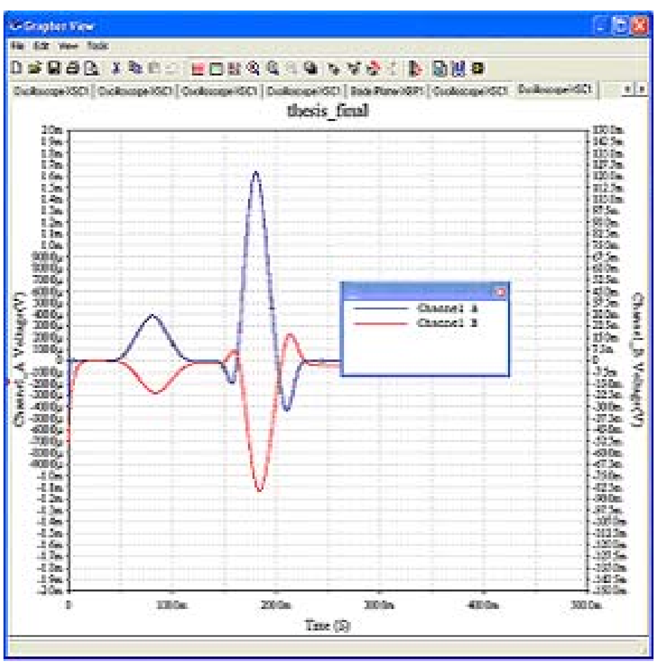

Figure 6: Testing with a normal ECG wave
(I/P and O/P plot – Final) Vs Time
Observation: Comparing the P wave peaks
Peak (Channel A –Input) = 0.4 mV
Peak (Channel B –Output) = -22.5 mV
Gain = -56.25

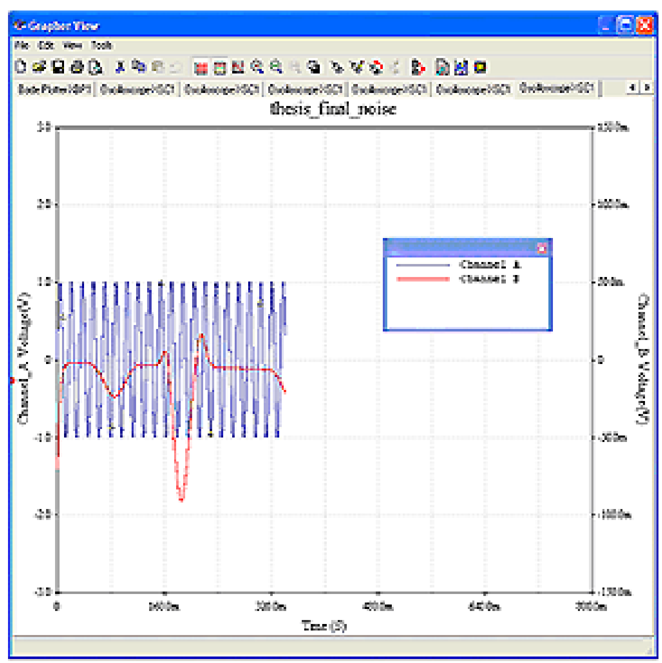

Figure 7: Testing with a noisy ECG wave
(I/P and O/P plot – Final) Vs Time
Observation: Channel 1: ECG + [0.3mV peak 200 Hz sine wave + 1V peak 60 Hz sine wave]-common mode noise
Channel 2: Smooth amplified ECG waveform
Peak (Channel B –Output) = 92 mV (for R wave)
Original R wave peak = 1.6 mV
Gain = -57.5

Figure 4-7 show the simulation results when the circuit shown in Figure 3 was tested with various noisy ECG signals. Figure 4 shows the output of the amplifying stage (before the filtering), when it was tested with a normal ECG wave. Comparing the R wave peaks, the gain seen is 2.8125. Figure 5 shows the clean output at the same point when the input is a noisy one (ECG signal + common mode noises: 0.1mV peak 60 Hz sine wave + 1mV peak 200 Hz sine wave).

Figure 6 shows the final output, when the circuit was tested with a normal ECG wave. The gain achieved by the design is -56.25. Figure 7 shows the pure, noise-free ECG signal as an output, when a noisy test signal (ECG signal + common mode noises: 0.3mV peak 200 Hz sine wave + 1V peak 60 Hz sine wave) was given as an input. The designed circuit purely removed all the common mode noises and provided a gain of -57.5.

4.4 Frequency Response

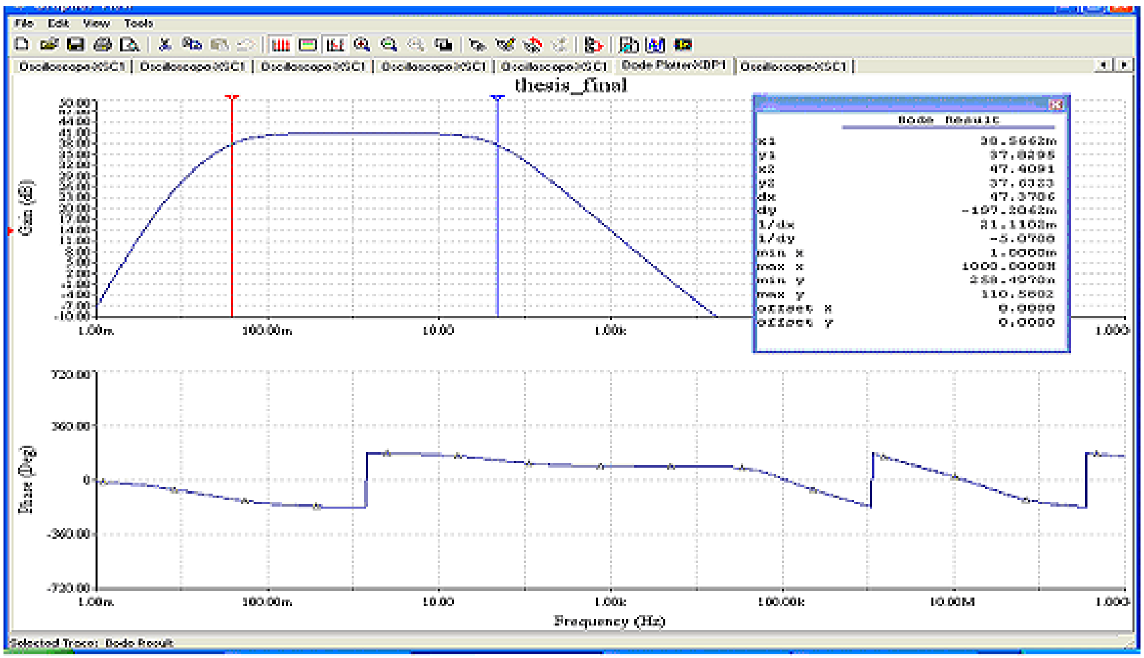

Figure 8: Frequency Response
Observation: Maximum Gain = 38.8726 dB
Lower 3dB cutoff frequency = 38.5662 milli Hz
Upper 3dB cutoff frequency = 47.4091 Hz

Figure 8 shows the frequency response of the proposed design. The lower and upper 3dB points are 38.5662 mHz and 47.4091 Hz respectively. The type of filter used depends on the use of the pre-amplifier (monitoring or diagnosing). The design proposed is for a monitoring mode, which required the operation range up to 45 Hz, a low pass filter with a cut off of 45 Hz was used.

5. Future Improvements

The achieved CMRR of nearly 80 dB is very high compared to the complexity of the design. The first two stages could be designed for better differential gain and hence a better Common Mode Rejection Ratio, had there not been a problem with the possible offset voltage. A dc restorator amplifier can be introduced in feedback to null out the dc offset, which will apply a negative correction voltage to the input of the first differential amplifier as soon as the output tried to swing very high driving the amplifying stage into saturation. Zener diodes with high breakdown voltages can also be used to save the circuitry from the high voltages from the defibrillator.

6. Conclusion

In this paper we have discussed Electrocardiogram (EKG), its significance, noises present during its recording and their elimination. The main noises present in an ECG recording are the common mode voltages (CMV) composed of two components (1) dc electrode offset potential and (2) 50 or 60 Hz ac-induced interference. We have proposed a design of an ECG pre-amplifier that can be used for monitoring mode. The amplifier has a high CMRR of 77.4 dB at 60 Hz and the operating frequency range of 38.5662 milli Hz to 47.4091 Hz.

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Biographies

Manan Joshi has received his MS degree in Electrical Engineering from University of Bridgeport in Dec 2006. Currently he is pursuing his PhD in Computer Science & Engineering at the University of Bridgeport. His research interests are in the field of Analog Electronics, Medical Electronics, Computer Networking and Wireless Communications.

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Lawrence V. Hmurcik is Professor and Chairman of Electrical Engineering at the University of Bridgeport, Bridgeport, CT. He earned his Ph.D. in semiconductor devices at Clarkson University in 1980. He worked in Diamond Shamrock's research division for 3 years before joining the University of Bridgeport in 1983. Dr. Hmurcik has 50 publications and 5 grants. He is also a professional consultant with 240 case entries, including 14 appearances in Court and Legal Depositions. Dr. Hmurcik's interests have changed over the years: starting in Solar Cell technology in 1977, Dr. Hmurcik is currently pursuing work in Medical Electronics and Electric Safety.