ECE 445: Senior Design Project

Microcontroller Based Heart Rate Monitor and Arrhythmia Detector

By Enyinnaya Egejuru

Instructor: Gary Swenson TA: Bob Schoonover
12/9/2008

ABSTRACT

The design and implementation of a portable microcontroller based heart rate monitoring system is discussed in this paper. The design considerations for this project are mostly influenced by the proposed users of the system. These users are medical practitioners in developing countries, who have very limited medical infrastructure. Hence, low lost, low power, portability, and ease of use are factors that are considered at every stage of the design. This system explores a low power microcontroller, the MSP430FG4816, manufactured by Texas Instruments for signal analysis. This is a compact system capable of acquisition, amplification, and interpretation of biological signals (ECG), as well as notification whenever cardiac conditions such as tachycardia and bradycardia are experienced.

Table of Contents

1. INTRODUCTION

In the United States of America alone, heart disease is the leading cause of death in both men and women claiming up to 900,000 lives every year [1]. Worldwide, coronary heart disease, the most common type of heart disease, claims over 7 million lives every year [2]. Up to half of these deaths occur even before emergency services can step in to intervene. In countries without a healthcare system as sophisticated as that of the United States, this number is much closer to 100%. A portable system equipped to monitor heart rhythms would serve as a means for exposure of possibly fatal cardiac activity and would be a very useful product.

The goal of this project is to create an ECG monitoring and alert system that can detect cardiac abnormalities such as tachycardia, bradycardia and possibly other arrhythmias. The design of this system was greatly influenced by a specific set of users, and hence, was tailored towards them and their environments. My inspiration comes from experiences I had growing up in a developing country where the rules of the healthcare system are very different from that of the western world. Most places have little to no infrastructure, and there is a lack of basic amenities such as water, food, electricity, hospital - things that come standard in the western world. Since there is no reliable source of constant power supply, and there is a lack of medical equipment, this project focuses on the design of a portable, low power, and low cost alternative to the sophisticated cardiac monitoring systems that are found in most hospitals in the western world. These systems would be easy to operate, easy to transport and would be used to monitor admitted patients in these areas; patients who unfortunately can't afford the luxury of accommodation in the few well equipped hospitals that exist in their locale.

This device would also prove invaluable in a war zone. Injured soldiers close to the front line may need cardiac monitoring at the on-site medical facility. In a situation like this, where there is need for something portable, inexpensive, and reliable, this system will be able to provide some level of temporary cardiac monitoring.

Benefits of the device include:

- Providing immediate notification of abnormalities in cardiac activity on a monitored patient
- Its low cost and low power consumption provides a cheap and reliable method for monitoring patients in developing countries
- Microcontroller based QRS analysis

- Easily accessible
- User friendly
- Different notification sounds for different heart issues
- Portability

2. CARDIAC PHYSIOLOGY

The most functional indicator of cardiac activity is the Electrocardiogram (ECG or EKG). The electrocardiogram is a quasi-periodical, rhythmically repeating signal synchronized by the function of the heart, which acts as a generator of bioelectric events. The generated signal can be described by means of a simple dipole, which generates a field vector, changing periodically in time and space and its effects measured on the surface [3].

The ECG is used around the world today as a simple and non invasive way for diagnosing heart conditions as a deviation from the standard amplitude and phase relationships of the resulting waveforms reflect abnormality in the cardiac cycle. The term *Electrocardiogram* was coined by Nobel Prize winner, William Einthoven, who developed the fundamental function of the ECG. Adhesive electrodes applied to the chest and limbs connect to the electrograph machine that detects patterns of minute electric currents in the heart muscle and print it on a chart or display it on a computer screen.

Electrical signals from the heart characteristically precede the normal mechanical function and monitoring these signals has great clinical significance. The ECG provides valuable information about a wide range of cardiac disorders such as the presence of an inactive part (infarction) or enlargement of the heart muscle [3].

2.1 The ECG Signal

The ECG signal measures the electrical activity of specialized heart cells that generate repetitive self-induced action potentials. Each action potential generated leads to a contraction of the heart muscle and thus the heartbeat [4]. A cross-section of the heart with parts of importance is shown in Figure 1.

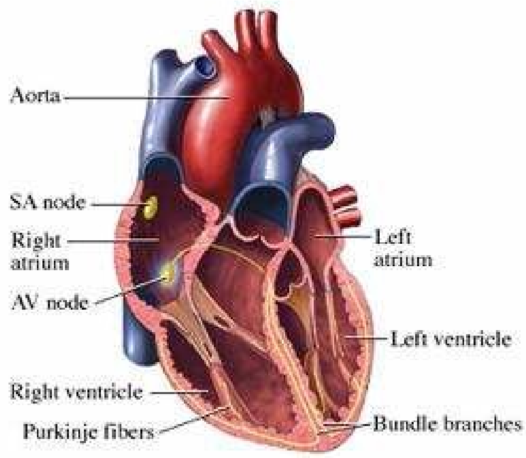


Figure 1: The Heart Conduction System

The heart is divided into four distinct chambers. The upper chambers of the heart are called the *Atria*. They are electrically stimulated first. After some delay for allowing the atria to empty their contents, the lower chambers, called the *Ventricles,* are then stimulated.

Initially, the right and left atria (also called auricles) are electrically stimulated to contract via the sinoatrial (SA) node located in the upper right atrium. The SA node is motivated through action potentials to maintain a heart rate of approximately 70 beats/minute. This action potential travels across the nerves of the atria causing depolarization and contraction, resulting in the P wave. The electrical signal continues from the SA node to the Atrioventricular (AV) node where a brief delay (~0.15 seconds) allows for atrial emptying (into the ventricles). The AV node transmits the signal to the Purkinje fibers (also called the Bundle of His) to cause ventricular depolarization and contraction, depicted as the QRS complex. Specifically, the peak of the R segment denotes the beginning of ventricular contraction. Post contraction begins a phase of repolarization, the T wave, in which blood is pumped from the ventricle chambers to the lungs (r. ventricle) or body (l. ventricle). The T wave diffuses differently than the QRS complex, beginning in the outer segments of the ventricles and propagating inward thus creating a lower amplitude signal. Occurring shortly after the T wave, the U wave is the repolarization of the Purkinje fibers (rarely detectable due to low SNR). These contractions are a result of synchronized action potentials controlling all myofibrils (muscle fibers) to produce the electrical signals collectively called the cardiac cycle.

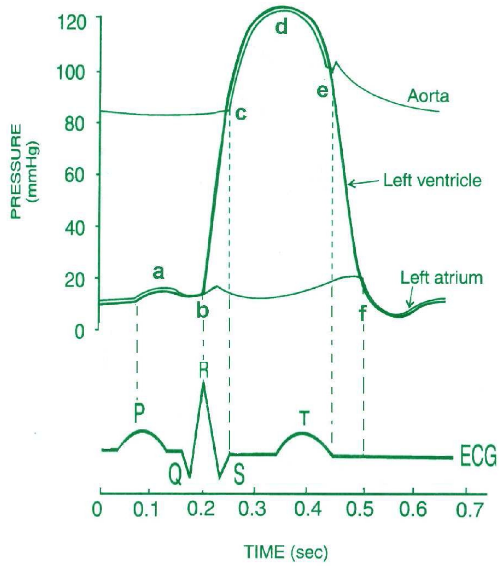


Figure 2: Wiggers Diagrams. They describe the relationship between the heartbeat and pressure changes in the heart chambers

The electrical excitation of myofibrils can also be represented as a series of pressure changes. As the atria fill with blood, the atrial pressure increases to a peak at atrial contraction (point a, figure 2). Once the blood leaves the atria, the ventricular pressure begins to increase to the point where ventricular pressure exceeds aortic pressure and the ventricles contract to release the blood (point b, c, d, figure 2). Toward the end of ventricular contraction (point e), the atria begin to fill with blood again. Beyond ventricular contraction the pressure falls until it is lower than the atrial pressure, at which point the atria contract to re-fill the ventricles (point f) [4].

2.2 Visual Representation of the ECG Signal

The polarization, depolarization, and contractions of the heart during the cardiac cycle produce signals which can be viewed on a device such as an oscilloscope. The ECG signal can be visually represented by three major waves which are synchronized with the heart activity. The focus of the project relates to the basic measure that is related to the heart rhythm, the heart rate. The heart rate simply describes the frequency of the cardiac cycle and is measured in contractions or beats per minute (bpm). Figure 3 illustrates the wave components of the ECG signal.

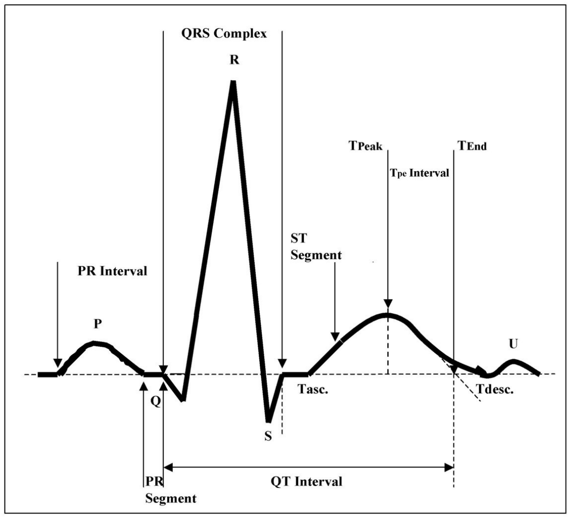


Figure 3: ECG Parameters

The *PR interval* is the duration of time between the beginning of the P wave, signifying atria depolarization, and the beginning of the QRS complex. It represents the time between the beginning of the contraction of the atrium and the beginning of the contraction of the ventricle. The *QT interval* extends from the beginning of the Q wave to the end of the T wave. It represents the time of ventricular contraction and repolarization. The *ST interval* extends from the S wave to the end of the T wave. The standard ECG parameter values are listed in the table below:

Table 1: ECG Parameters

Parameter	Range	Units
Heart Rate	60 – 100	Beats Per Minute
PR Interval	0.12 – 0.20	Seconds
QT Interval	0.39 $\pm$ 0.04	Seconds
P Wave Duration	0.12	Seconds
QRS Width	0.05 – 0.1	Seconds
T Wave Duration	0.08	Seconds

As mentioned previously, the electric field generated by the heart is best characterized by vector quantities, however, it is generally convenient to directly measure only scalar quantities, i.e. a voltage difference (in the mV order) between given points of the body. The primary signal characteristics of an ECG signal has a useful frequency range of about *0.05Hz to 150Hz*. For this reason, a good low frequency response is essential to ensure baseline stability and a good high frequency response is needed for attenuation of high frequency noise from other signals of biological origin [3].

3. SYSTEM DESIGN

This section gives a detailed description of the development of a microcontroller based heart rate and arrhythmia monitoring system. Some of the major factors considered in the design of this device were portability and low power. In order to achieve portability, there is a need to significantly minimize the amount of hardware that exists in the circuit. The use of microcontrollers is a very effective way of reducing the hardware components in a circuit as they implement hardware functions using software, often times, even more accurately and efficiently than conventional analog implementations.

For clarity, the design/implementation of this project has been categorized into three segments:

- ❖ Front End
- ❖ Core/Software
- ❖ Back End

A block diagram for the system is shown in Figure 4.

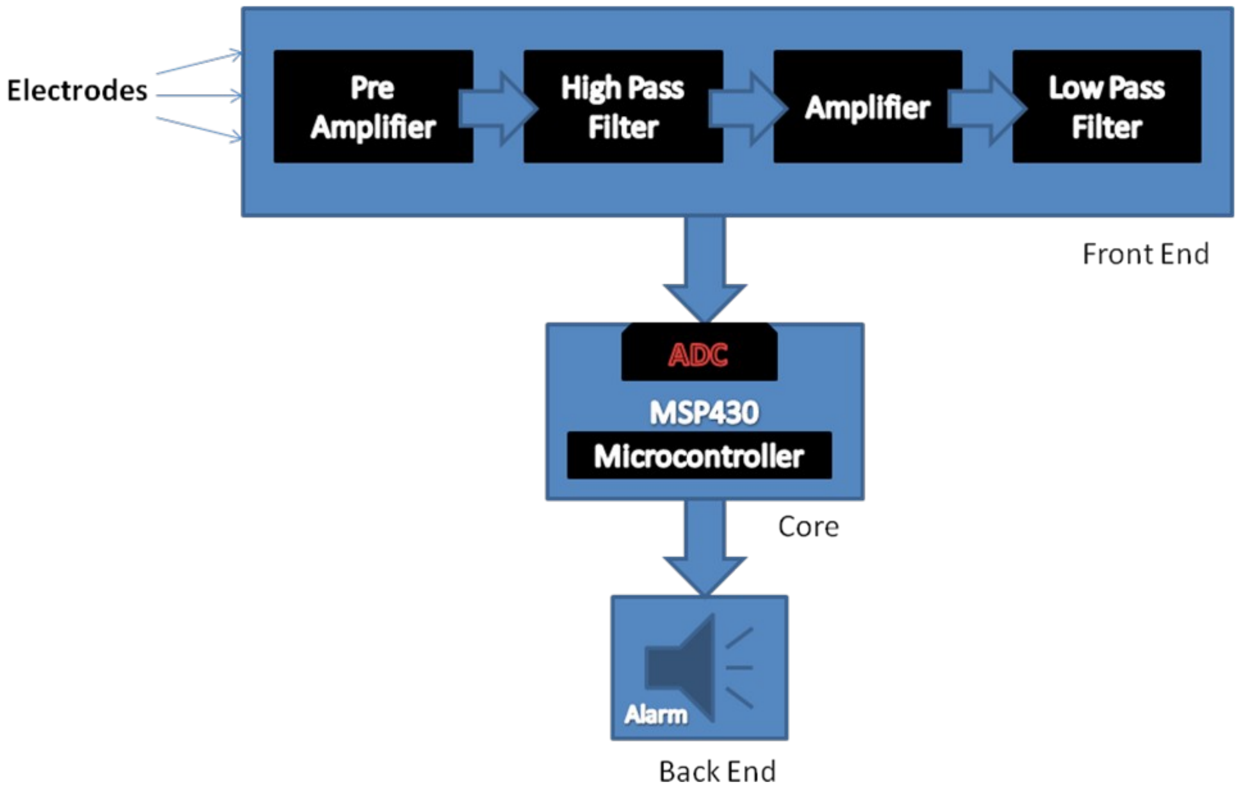


Figure 4: Block diagram for system design

Bio-signals are acquired via electrodes and passed on to the Front End segment where the signal is amplified and conditioned for interpretation. The analog signal moves from the Front End to the Core of the system which comprises a Micro programmed Control Unit (MCU). The Core is responsible for interpretation and analysis of the acquired bio-signals to get the information that is needed. After analysis, if need be, the Core communicates with the Back End segment, which is a voice and LED based notification system.

As mentioned earlier, a low power system is desired. The system to be designed should consume as little power as possible to be able to operate for a substantial amount of time on a 3V battery. If this is achieved, we would have a truly portable system that is well suited for the users in mind. This design requirement greatly influenced the choice of Integrated Circuit chips used in the system. Selecting the right hardware was one of the first challenges of the project and each choice will be discussed in detail as they arise.

3.1 Electrodes

In order to measure potentials in the body, an interface is needed between the body and the measuring device. This function is performed by biopotential. They serve as transducers to change ionic current flowing in the body to electrical current carried by electrode wires to the measuring device [4].

One of the most important characteristics of electrodes is that they should not polarize. This means that the electrode potential must not vary considerably even when current is passed through them. Electrodes made of Silver-Silver Chloride have been found to yield acceptable standards of performance. These electrodes are also nontoxic and are preferred over other electrodes like Zinc-Zinc Sulphate, which are highly toxic to exposed tissue. The Silver-Silver Chloride electrodes meet the demands of medical practice with their highly reproducible parameters and superior properties with regard to long term stability. Hence, these electrodes were chosen for signal acquisition [3].

The potential difference between electrodes can vary due to movement. This variation is known as *Motion Artifact* and causes interference in bio-potential measurements. For example, a wandering baseline (Figure 5) but otherwise normal signal is due to motion artifact. It is usually experienced right after electrodes have been applied or during patient movement and is due to a relatively slow establishment of electrochemical equilibrium at the electrode-skin interface. Poor contact between electrodes and the signal source can be another source of interference.

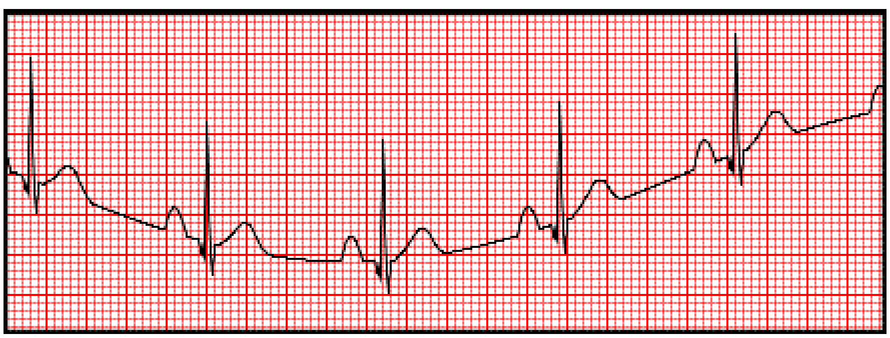


Figure 5: ECG signal with baseline drift

3.2 The ECG Leads

The resulting tracing of voltage difference at any two sites due to electrical activity of the heart is called a *Lead* [3]. In bipolar leads, on which this application is based, ECG is recorded by using two electrodes such that the final trace corresponds to the difference of electrical potentials existing between them. They have been universally adopted as the standard leads and are sometimes referred to as Einthoven leads.

In the standard lead I, the electrodes are placed on the right and left arm. The difference is potential measured between the two electrodes is in reference to a third point on the body, which is the right leg in this design.

3.3 Front End (Amplification and Signal Conditioning)

A major challenge in the design of a system like this is acquiring and measuring very small electrical signals in the presence of much larger noise components. The typical ECG signal has an amplitude between 1mV and 4mV. Although the average amplitude is only around 1 mV, there are large dc offset voltages due to electrochemical processes between the electrode/skin interface. These can be as high as 500 mV. The human body acts like the midpoint of a capacitive divider between one or more power lines and ground. Thus, common-mode voltages as high as 1.5V, can be superimposed on the body. Eliminating this source of noise is one of the major tasks of an ECG amplifier. Fortunately, the ECG signals are differential signals while the power line voltages are common-mode, so the noise can be reduced with differential amplifiers [7, 8].

There are certain requirements that come to mind when designing an amplification stage for this application. These include:

- **Frequency Response**: The ability of an amplifier to amplify input signals in the frequency range where the signals features are present is called frequency response. As mentioned earlier, the ECG signal has useful frequency components in a range of 0.05Hz to 150Hz. Hence, our amplifier should be sensitive to this.

- **Gain**: The input signal to the amplification stage will be about 1mV-4mV. In order to do any measuring or monitoring, signals in the order of about 1V are required. This means that an amplification factor of about 1000x is needed to make our signal useful.

- **Input Impedance**: A large input impedance allows for good low frequency response and low input power requirements.

- **Common Mode Rejection Ratio**: Common mode signals are interference signals present at both inputs of a differential signal. An example of this would be interference from power mains. The CMRR, as the name implies, is the ability to reject such signals. A CMRR of the order of 100-120dB with 5KΩ imbalance in the leads is desired [3].

3.3.1 Pre-Amplification

The amplifier chosen for the pre-amplification phase of the design was the **AD622 Instrumentation Amplifier** by Analog Devices. An Instrumentation amplifier is a type of differential amplifier that has been outfitted with input buffers, which eliminate the need for input impedance matching, making the amplifier particularly suitable for use in measurement.

Characteristics include very low DC offset, low drift, low noise, very high open-loop gain, very high common-mode rejection ratio, and very high input impedances. Instrumentation amplifiers are used where great accuracy and stability of the circuit both short- and long-term are required. The bio-signal is a differential, with a lot of interference, which makes the AD622 perfect for the job. The wide power supply range of 2.6V to 15V also makes it appropriate for our battery powered design. Figure 6 shows a schematic of the AD622.

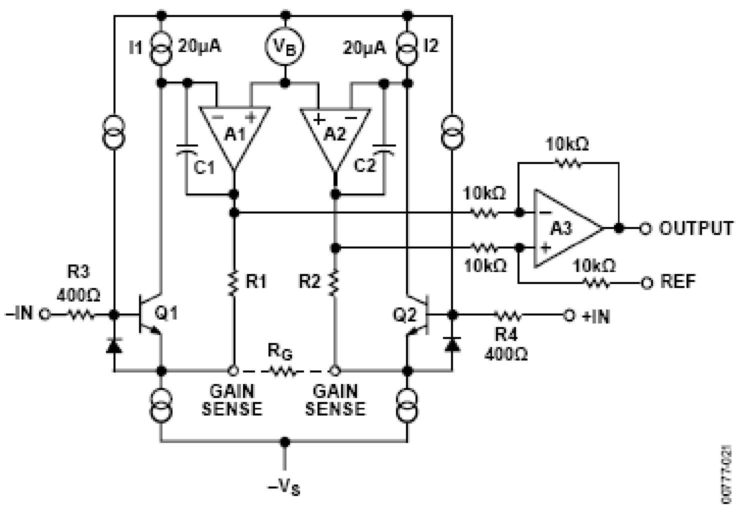


Figure 6: AD622 Schematic

The gain is controlled by a resistor R_G between the 1st and 8th pins of the IC. There is amplification by a factor of 10 in this phase. The total resistor value chosen is 5.62KΩ.

$$G = 1 + \frac{50.5K\Omega}{2.7K\Omega + 2.7K\Omega} \approx 10$$

The common mode of the body is sensed by 2 averaging resistors R_G / 2, inverted, amplified, and applied back to the right leg. This method of grounding is called the **Driven Right Leg** (DRL). The negative feedback from this setup drives the common mode voltage to a low level and the body's current flows to the OP AMP output and grounds the patient. The DRL also provides electric safety to in case high voltages travel back to patient [5]. The DRL amplifier has a gain of:

$$A = -\frac{380K\Omega}{10K\Omega} = -38$$

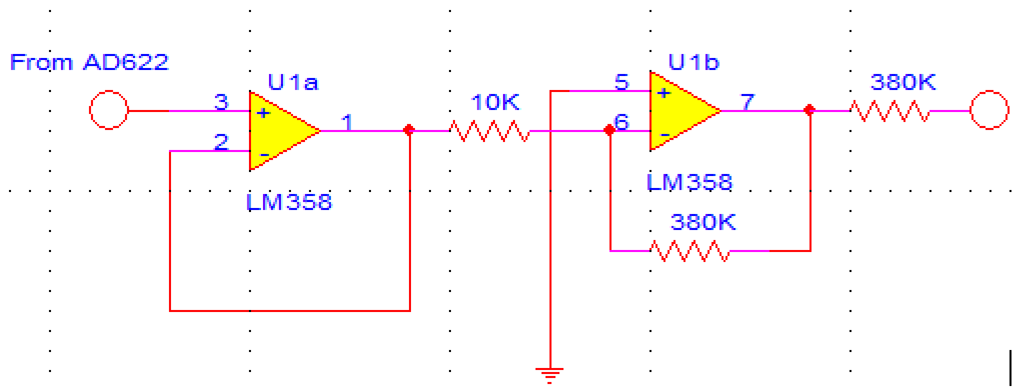


Figure 7: DRL Circuit

3.3.2 Filters

A filter is used to attenuate certain frequencies of a signal. The attenuation level depends on the type of filter used. For this circuit, we use two passive RC filters. A high pass filter with cut-off frequency of 0.05Hz is placed between the pre-amplification and gain stage. The purpose of this filter is to reduce DC offset form being amplified with the signal and amplifier saturation. A low pass filter was placed at the output of the gain stage to attenuate high frequency noise above 130Hz. The cut off frequencies are determined by the formula:

$$Fc = \frac{1}{2\pi RC}$$

Figure 8 shows the circuit formation for high pass and low pass filters.

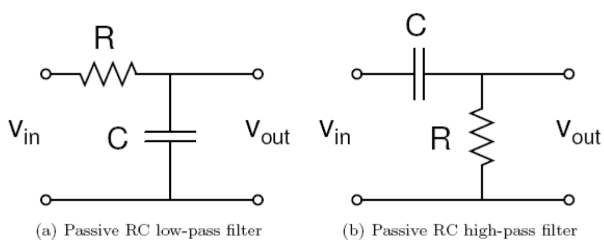


Figure 8: High Pass an Low Pass analog filters

Based on the formula above, values for R and C are calculated.

$$F_{hp} = \frac{1}{2\pi RC} = 0.05Hz$$

$$RC \approx 3$$

Therefore values for analog components R and C can be

R_{hp} = 100KΩ, C_{hp} = 30μF

Similarly,

$$F_{lp} = \frac{1}{2\pi RC} = 130Hz$$

$$RC \approx 1.2 \times 10^{-3}$$

R_{lp} = 1.2MΩ C_{lp} = 1nF

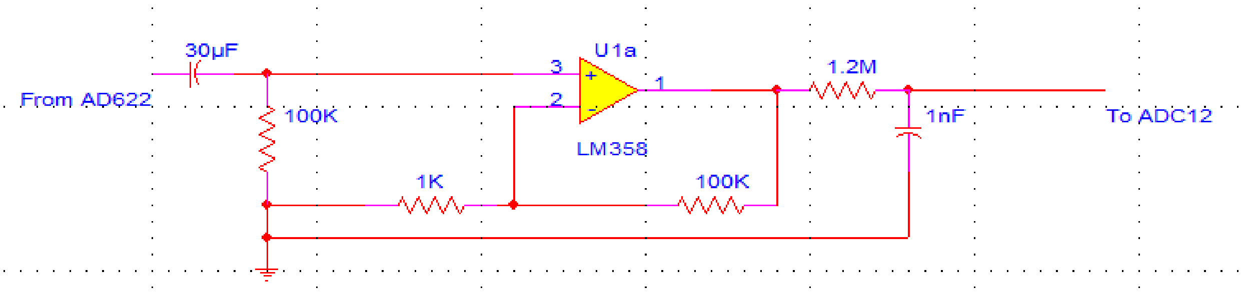


Figure 9: Filter and amplification phases

3.3.3 Amplifier/Gain

Amplification at this phase uses a simple Non-Inverting Amplifier. This phase is needed to provide the necessary amplification to render the signals being measured suitable for reading and monitoring. The signal is strengthened by a factor of 100 in addition to what we already have, making the total gain 1000. The gain of this amplifier is given by:

$$A = 1 + \frac{R_2}{R_1}$$

Resistor values R_2 = 100KΩ and R_1 = 1KΩ are chosen.

3.4 Core/Software

Signal processing and heart beat detection take place in the core. The core consists of a microcontroller that takes the analog ECG signal as input and conditions (analog-digital conversion and digital filtering) it for interpretation. The signal that triggers the backend system originates from this stage.

The choice of hardware for this function was very important. A very low power, cheap, reliable, and relatively feature full MCU was required. After much research, the MSP430FG4618 MCU by Texas Instruments was selected to perform the core functions.

The MSP430FG4618 is an ultra low power, high performance microcontroller. The chip is integrated with 16 registers and constant generators making it an excellent choice for code efficiency. It provides an avenue for designing a long lasting dependable system for analysis and interpretation of ECG signals. It comes with its own power and flash memory space for writing analysis programs as well as storing data if needed. Some of the key features of the MCU include:

- Low supply-voltage range, 1.8V-3.6V
- Ultralow power consumption
- Five power saving modes
- Wake up from standby mode in less than 6µs
- 16 bit RISC architecture
- 12 bit A/D converters
- 12 bit D/A converters
- 16 bit Timer_A
- 116KB + 256B Flash or ROM memory, 8KB RAM

Figure 10 shows a block diagram of the Core system.

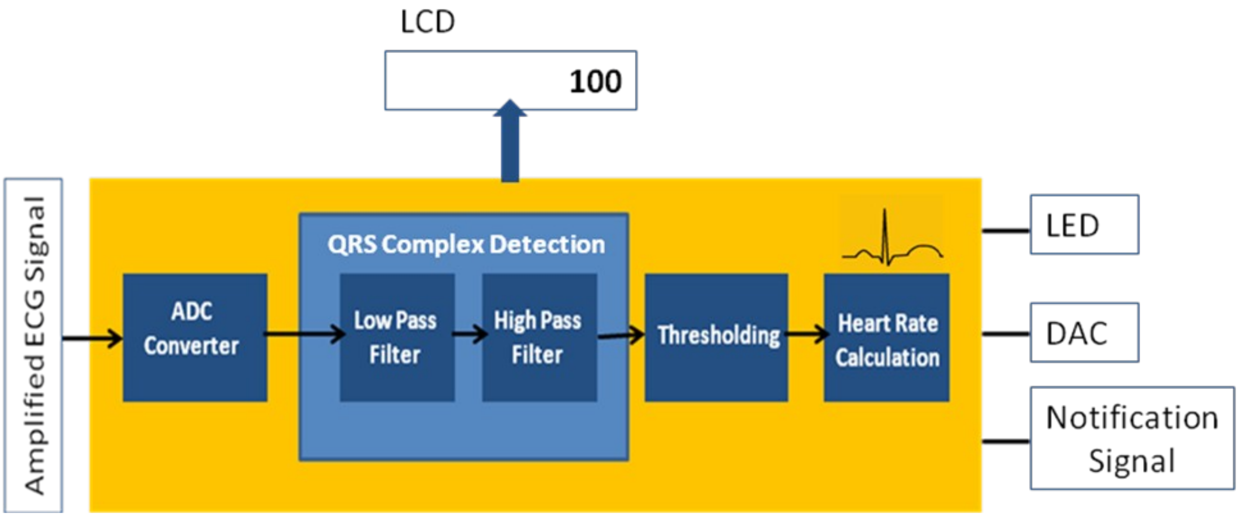


Figure 10: Block diagram for core design

3.4.1 ECG Sampling

The amplified ECG signal is fed to the on-chip analog-to-digital converter via an ADC12 analog input channel A0 on the MSP430. The ADC12 samples the signal with a sampling frequency of 512Hz or every 2ms [3]. The precise sampling period is achieved by triggering the DC12 conversions with the Timer_A pulses. Timer_A is clocked by ACLK, which is generated from the 32.768-KHz low frequency oscillator. The ADC has a resolution of 12 bits, giving us 2^{12} possible digital levels.

3.4.2 QRS Complex Detection

Calculation of heart rate from an ECG signal is based upon the reliable detection of the QRS complex. Hence, in order to accurately calculate the heart rate in real-time, the QRS complex must be detected for every beat. There are several methods for capturing the QRS complex in an ECG signal [3]:

Multiplication of Backward Difference (MOBD): This algorithm is based on the transformation of the signal by using a one point derivative and the sign- function combined to generate a signal which equals zero anytime the signal changes slope (the first derivative changes sign) over a specific time window, and equals the product of the signal values if the slope does not change over the same time window.

Pan-Tompkins Algorithm: This algorithm is based on a number of steps performed after the signal in preprocessed with a 5-15 Hz bandpass filter, implemented as a cascade of one low-pass filter and one high-pass filter. These steps involve the differentiation of the filtered signal, to obtain information of the slope of the signal, squaring to intensify the slope of the signal and minimize the false positives cause by high T waves and, at the last stage, the use of a moving window integrator which generates a signal holding information about both the slope and the width of QRS complexes.

Curve Length Transformation: This method is based on a particular type of transformation which results in a quantity that is larger in absolute value for ECG segments containing strong slope variations (QRS complexes) and tends to be smaller in presence of flat segments or slower waves (P and T waves).

Although these methods are efficient for the extraction of the QRS complex, there were difficult to implement on a microcontroller for a beginner. The method chosen for QRS detection was quick and simple…not a very efficient algorithm for arrhythmia detection, but works well for acquiring QRS complex for heart rate calculations.

The algorithm used was based on the fact that most of the energy of QRS complexes is contained in a frequency range 10-25Hz [4]. In order to isolate the QRS complex from the P and T waves, a 17 sample finite impulse response filter with a center frequency of 17Hz and a Q of 5 was implemented (A high pass filter/low pass filter configuration could also be implemented). This gives a good signal to noise ratio and isolates the QRS complex. The coefficients for the filter are obtained through Matlab and a simple convolution algorithm is implemented in the program code.

After QRS isolation has taken place, the signal is passed to the thresholding phase, where it is compared against a set threshold value. If the digital value exceeds the set threshold, then we know that a heart beat has occurred and a count is incremented. The beat interval averaged over several beats is used to calculate the heart rate for display, and alarm notification.

3.4.3 Heart Rate Calculation

The number of beats per minute is calculated using a three beat average. In the main function of the supporting software, the time scale is tracked using two variables, counter and pulse period. Pulseperiod is incremented by one during every sample period, and because we sample at a rate of 512Hz, it is relatively easy to track the timescale based on the number of counts in the pulseperiod. Whenever there is a beat detected, the counter is reset. The pulseperiod is accumulated for 3 consecutive beats, and on the third beat the heart rate is calculated using the formula below [6].

$$Heart\ Rate = \frac{1}{pulseperiod/_{3 \times 512 \times 60}} = \frac{92160}{pulseperiod}$$

3.5 Back End

The backend comprises an alarm system for notification for any occurrence of abnormality. The system has a voice notification component as well as an LED notification component. The voice component functions such that whenever a particular abnormality occurs, a custom pre-recorded voice message playback, stating the condition being experienced (Tachycardia or Bradycardia). The LED notification system displays a constant red light during normal operation and a blinking effect during message playback or arrhythmia detection.

The component selected for this function is the ISD17120 by ChipCorder. It is a high quality, fully integrated, single-chip multimessage voice record and playback device suited to a variety of electrical systems. Some of the features of this chip include:

- 120 sec messages at 8KHz sampling frequency
- PWM class D speaker driver
- LED blinking notification during playback
- Multiple message addressing
- Great sound quality.

What makes this device special is that multiple messages can be recorded in the chip, and based on the situation, the appropriate message is played using a simple selection pin. In this design, there is a voice message that says “Attention Attention, patient experiencing arrhythmia, Attention”. Figure 8 shows the circuitry for the ISD17120 chip during the record cycle.

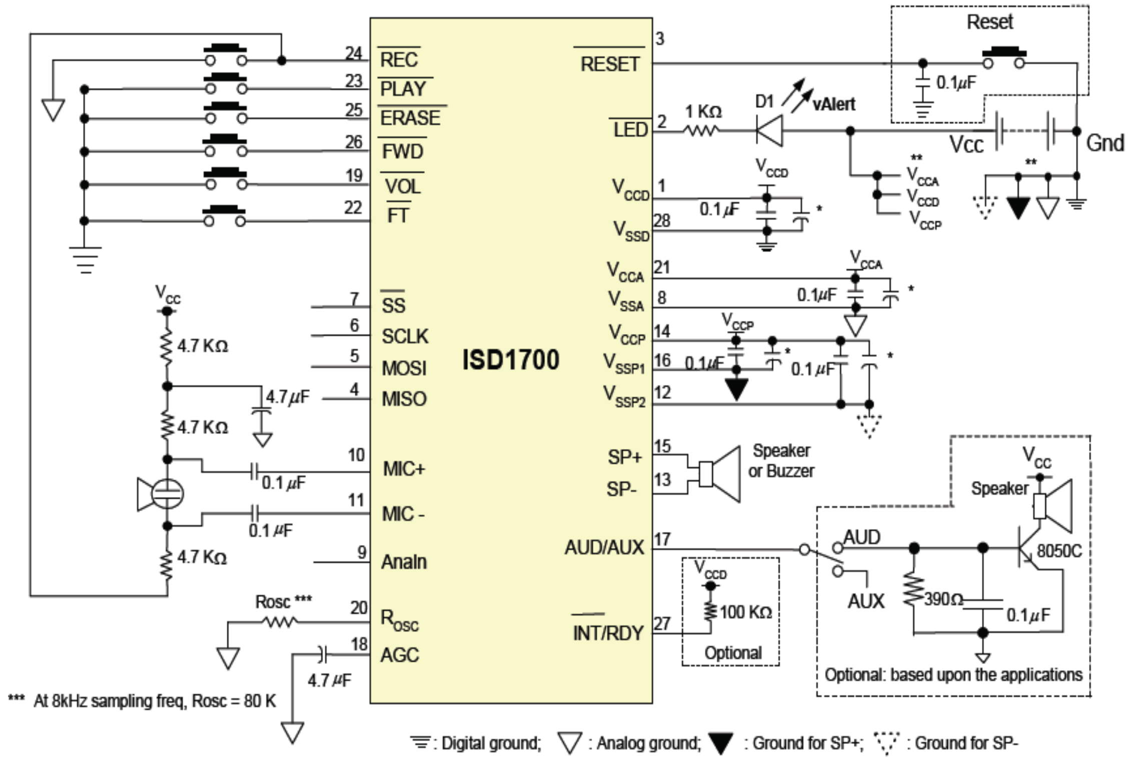


Figure 11: ISD17120 application design for record cycle

Playback occurs when the Play pin on the chip is connected to ground. The push button control style connects the pin to ground on one clock pulse, which would cause a single playback. The playback continues as long as the Play pin is connected to ground creating a looping message effect whenever an arrhythmia occurs and stays that way. Whenever the set threshold is surpassed, a signal is sent from the MCU to the chip, which, through a simple tri-state buffer, connects the Play input to ground. The figure below illustrates this concept.

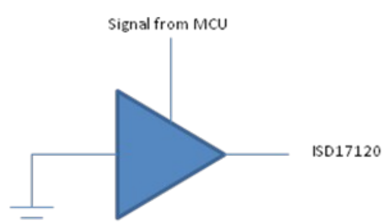


Figure 12: Tri-State Buffer Implementation

4. DESIGN VERIFICATION

4.1 Simulation

This project involves the analysis of biological signals, which in general, would involve an actual human connection for data acquisition. In order to test appropriately, there was need for some simulated bio-signals to analyze the system and determine functionality quickly and efficiently.

PhysioBank is an online physiologic signal archive where many types of digitized signals are available and free for download. A 60-second 12-lead ECG recording (16-bit resolution, 1000 Hz sample frequency) was downloaded from the site. MATLAB was used to scale the aVR and aVL signals, and then to write the data to a WAV file. The left channel of the WAV file contains aVL and the right channel contains aVR. Because aVL and aVR are augmented versions of the left arm voltage and right arm voltage, respectively, they can be used to simulate Lead I of the ECG [10].

This simulated method of ECG signal acquisition is valuable and pragmatic for testing purposes. Simulations for different kinds of arrhythmia could be acquired for testing the circuit functionality, as opposed to finding an actual person with these disorders and connecting them to a circuit. This, however, brought about a new challenge of finding a way to connect the simulated signal to the circuit as an input. This was achieved my modifying a typical headphone cable. The cable was cut at the end with the head phones and the positive, negative and ground wires were obtained. A small wire was soldered to these and they served as input to the circuit (Figure 13) and the other end as plugged into the line-out of the computer.

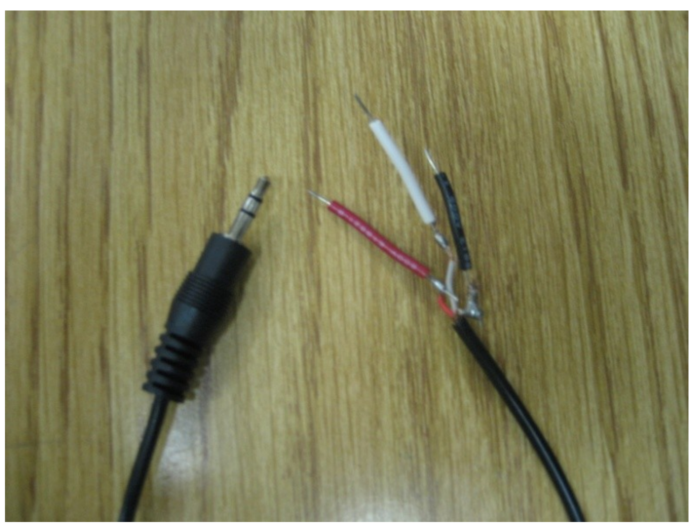


Figure 13: Modified headphone cable for ECG signal

The three wires in the headphone cable, red, white and black, correspond to the right arm, left arm and right leg respectively.

4.2 Amplification

The first phase of amplification is achieved by the AD622 instrumentation amplifier. The figures below show the functionality of this amplification phase.

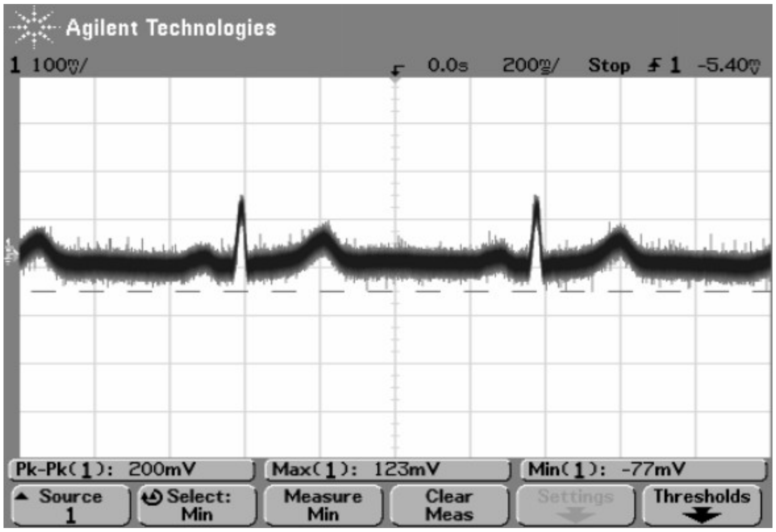


Figure 14: ECG Signal input to AD622

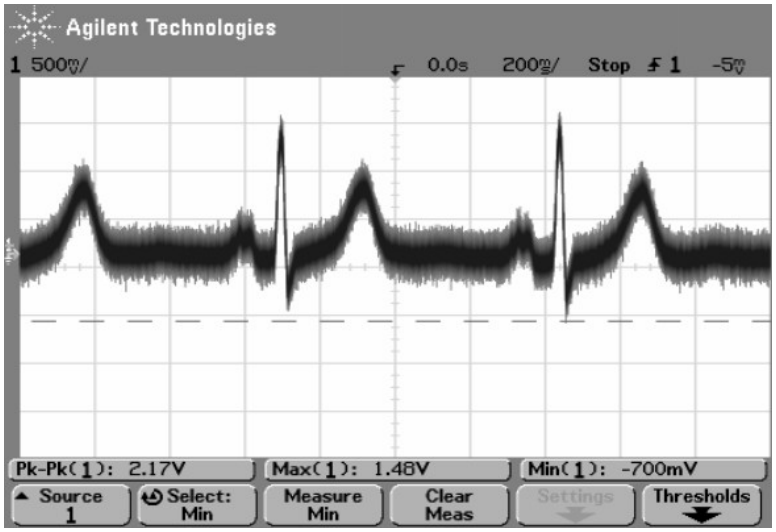


Figure 15: ECG Signal output from AD622

Although simulated signals were a very good indication of whether the system worked, there had to be test done to analyze the amplification of an actual ECG signal. The performance of the simulated signal versus the actual signal varied as expected. However, a distinct ECG signal was obtained and amplified from an actual source. This helped test the functionality of the electrodes as well as the amplifier stage as a whole.

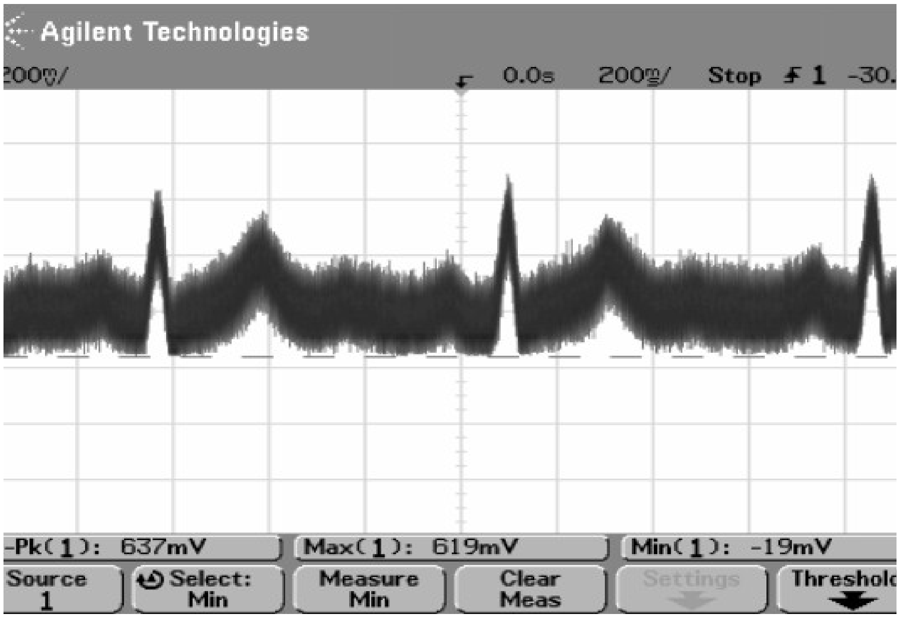


Figure 16: Actual ECG Signal from patient

Figure 17: Electrode configuration for signal acquisition

4.3 Core

The MSP430FG4618 used for this design was located on the MSP430FG618/F2013 Experimenters board by Texas Instruments. The board contains several debugging tools and ultimately led to the addition of a new feature to the original design.

An MSP Flash Emulation tool is required to code and debug the MSP430FG4618 on the board. Two separate JTAG headers are available, supporting independent debug environments. The chip uses a standard 4-wire JTAG connection allowing all port pins to be used during code debug. The debug environment used here the IAR Embedded Workbench, which was chosen for its versatility and unique debug environment. A schematic of this board is shown in the appendix.

The original design had no way of determining what was being calculated as a heart rate, and hence, it was difficult to debug the software. To solve this problem, a new feature was added to the design: an LCD screen was incorporated to display the heart rate for the user/physician to see. This addition, I discovered, was a very useful feature to include because although there is a notification system available, it is impossible for the physician to know the extremity of tachycardia or bradycardia.

When the heart rate is calculated, the individual hundredths, tens, and units have to be extracted and converted to signals for the LCD. This is done by a simple division and modulus operation the heart rate value. It is worth mentioning, however, that the presence of an LCD unit that runs constantly is very costly in terms of battery power. The lifetime of the battery is reduced by adding this feature.

4.4 Notification

The notification is triggered by a signal from the MCU. Hence, the first test for the backend system was to check if the appropriate signal was being sent at the right time. To accomplish this, the threshold level was set to a lower value than normal. Whenever the heart rate count exceeds the set threshold level, an LED was lit up on the development board indicating the signal to be sent to the ISD17120. The voice function was also tested as well as the blinking effect of the LED's. These functioned as expected, beginning playback whenever the signal was acquired from the microcontroller.

4.5 Patient Safety

In the design of any biomedical system, safety is a very important issue and consideration. The patient is kept safe by using features such as the DRL to effectively ground the patient and the use of an instrumentation amplifier to ensure that there is limited back current flowing from the circuit to the patient. The maximum current exposed to the user was about 0.6µA and this is well below the 10µA safety limit set by the American Heart Association [11].

The battery life of this system is acceptable for its application. Given the low power modes for all the devices present, the power consumption of the circuit is relatively low. In the current design, we can get up to 40hrs of analysis from a 3V power source. This number can be in the even greater if we opt to exclude the LCD feature of the design.

5. DESIGN CHALLENGES/ISSUES

There were some challenges faced in the design of this system that resulted in the system not functioning as expected.

- Cutoff distortion, where high offset voltages at the electrodes produce saturation that affect the signal significantly. Signal amplitude combined with this offset can drive the amplifier to saturation during the QRS complex, resulting in peaks being cut off.

- Open lead wires cause problems in ECG signal acquisition. They are as a result of electrodes not being connected to the patient. When this occurs, the open wires pick up relatively high voltages generated by power lines and distort the signal

- AC mains interference was a major issue in the design of this system. The 60Hz main power line frequency and its components is the most common source of interference in any electronic circuit and most especially in biomedical systems. The AC interference enters the biomedical system from the power supply and other devices connected to the mains supply. The interference can be magnetic or more commonly capacitive [8].

- There was heavy fluctuation in the heart rate value being read out by the LCD. The system in general does what it is designed for; however, our naïve QRS detection algorithm posed a problem for QRS detection. More sophisticated QRS detection and heart rate calculation algorithms exist with methods for determining threshold values for digital signals, which enable more accurate thresholding. Also, these algorithms analyze data that has been stored in the chip over several cycles to calculate the heart rate and the presence of other arrhythmias. The implemented system on the other hand, has limited resources to store the information needed for proper and accurate analysis. There is a volatile heart rate display because the system averages over only three beats for real time heart rate calculations.

- Digital filtering was not at its best, again, because of limited resources. Usually, the greater the number of taps, the more reliable and efficient the filter process is. However, due to the resources available on the MSP430FG4618, it would be impractical to have such robust filtering systems as they will slow the system down immensely.

- My inexperience in the use and implementation of analog circuits using MCU's was also a problem. There could have been some improvements that would have enabled the system function better, most especially in the digital filtering stage. More experience would allow for more sophisticated solutions for problems.

6. RECOMMENDATIONS

1. A 60Hz notch filter implemented within the microcontroller would significantly reduce the power line interference experienced by the system. A digital notch filter is much more efficient and reliable than an analog one. Analog notch filters are very sensitive and make very lousy filters. The notch filter would be placed immediately following ADC conversion. The signal would then be much cleaner and noise free for better analysis.

   An attempt was made to implement an analog notch filter; however, it was almost impossible to get a design that adequately filtered just the 60Hz noise due to the sensitivity of these analog components. The Universal Active Filter, UAF42, integrated circuit chip was explored much later in the design and could be used to implement notch filter design with minimum additional passive components.

2. Implementing a more sophisticated QRS isolation and threshold determining algorithm would render the system much more efficient. There are more efficient algorithms, such as the Pan-Tompkins algorithm, for real-time detection. With the right choice of microcontrollers and a bit of accommodation, these algorithms can be implemented on the MSP430 series.

7. COST ANALYSIS

7.1 Parts List/Cost

Part #	Description	Qty	Manufacturer	Unit Cost ($)	Total Cost ($)
AD622	Instrumentation Amplifier	1	Analog Devices	7.50	7.50
LM 358	Dual OP AMP	2	Texas Instruments	0.33	0.66
MSP430FG4618	MCU	1	Texas Instruments	9.00	9.00
Electrodes	Ag/AgCl Electrodes	3	Uni-Medical	0.20	0.60
ISD17120	Voice Chip	1	ChipCorder	10.00	10.00
Speaker	8ohm speaker	1	RadioShack	2.00	2.00
Resistors		15	Shop	0.10	1.50
Capacitors		13	Shop	0.15	1.95
Alligator to Banana cables		3	Shop	3.00	9.00
SBLCDA4	LCD	1	SoftBaugh	2.85	2.85
Battery	Battery	2	Energizer	1.00	2.00
	TOTAL				**47.06**

Based on the table above, the total cost for this device is $47.06. This is a reasonable price; however, it might be a bit expensive for the users in mind.

7.2 Labor Cost

A total of 150 hours was spent in the research, design, and implementation of the device. At a rate of 30/hr, the total labor cost is:

$$Total\ Labor\ Cost = \$30 \times 150 \times 2.5 = \$11250$$

TOTAL COST = $11250 + $47.06 = $11297.06

8. CONCLUSION

8.1 Accomplishments

All in all, this project achieved a lot of its goals. The project implemented a low cost, low power heart rate monitoring and alarm system using microcontroller technology. Lists of accomplishments include:

- Adequately acquiring biological signal
- Adequately amplifying biological signal
- ADC conversion of analog signal
- Semi functional heart rate counter
- Functional notification and alarm system
- LCD heart rate display
- Use of low power components for battery operation

8.2 Future Work

The microcontroller based heart rate monitoring system designed in this project has a lot of advantages, but can also be improved on. There is a lot of improvement that can be made to the firmware portion of the project that would result in a more reliable system as stated in the recommendations section.

Looking ahead, as microcontrollers get more and more advanced, there will be a shift from analog amplification to digital amplification. Biological signals from the electrodes can be fed directly into the microcontroller, where the front end, core, and back end work can be processed. This will significantly reduce the surface area consumed by a circuit and would lead to a smaller, more compact, and portable system.

More work can be done in the processes leading to the acquisition of these small biological signals. There are many challenges that still pose big problems in the design of systems like this. The skin/electrode interface, motion artifact, as well as AC power noise are problematic areas which would significantly increase reliability and efficiency if there are ways to minimize their effects.

REFERENCES

[1] American Heart Association. "Cardiovascular Disease Statistics" [Online Document], 2008 [Cited 9 Dec 2008], Available HTTP: <http://www.americanheart.org/presenter.jhtml?identifier=4478>.

[2] Centers for Disease Control and Prevention. "Heart Disease Facts and Statistics" [Online Document], 10 Sept 2008 [Cited 4 Dec 2008], Available HTTP: < http://www.cdc.gov/heartdisease/statistics.htm>.

[3] R.S. Khandpur. *Biomedical Instrumentation: Technology and Applications*, McGraw-Hill Companies, Inc., 2005.

[4] V. Virgilio. "Prototype of a Portable ECG Monitoring System (Holter Monitor) with Real Time Detection of Beat Abnormalities", Master thesis, Aalborg University, 2006.

[5] J. Hailong, M. Bing. "Design of Holter ECG System Based on MSP430 and USB Technology" IEEE (2007) 976-979.

[6] Murugavel Ragu. "Heart-Rate and EKG Monitor Using the MSP430FG439." Texas Instruments Application Report, Sept. 2007, Available HTTP: <http://focus.tij.co.jp/jp/lit/an/slaa280a/slaa280a.pdf>.

[7] T. Kugelstadt. "Getting the Most Out of Your Instrumentation Amplifier Design". Analog Applications Journal, 2005.

[8] W. Grossbach. "Measuring the ECG Signal with a Mixed Analog-Digital Application-Specific IC", Hewlett-Packard Journal, 1991, Available HTTP: <http://findarticles.com/p/articles/mi_m0HPJ/is_/ai_11398513>.

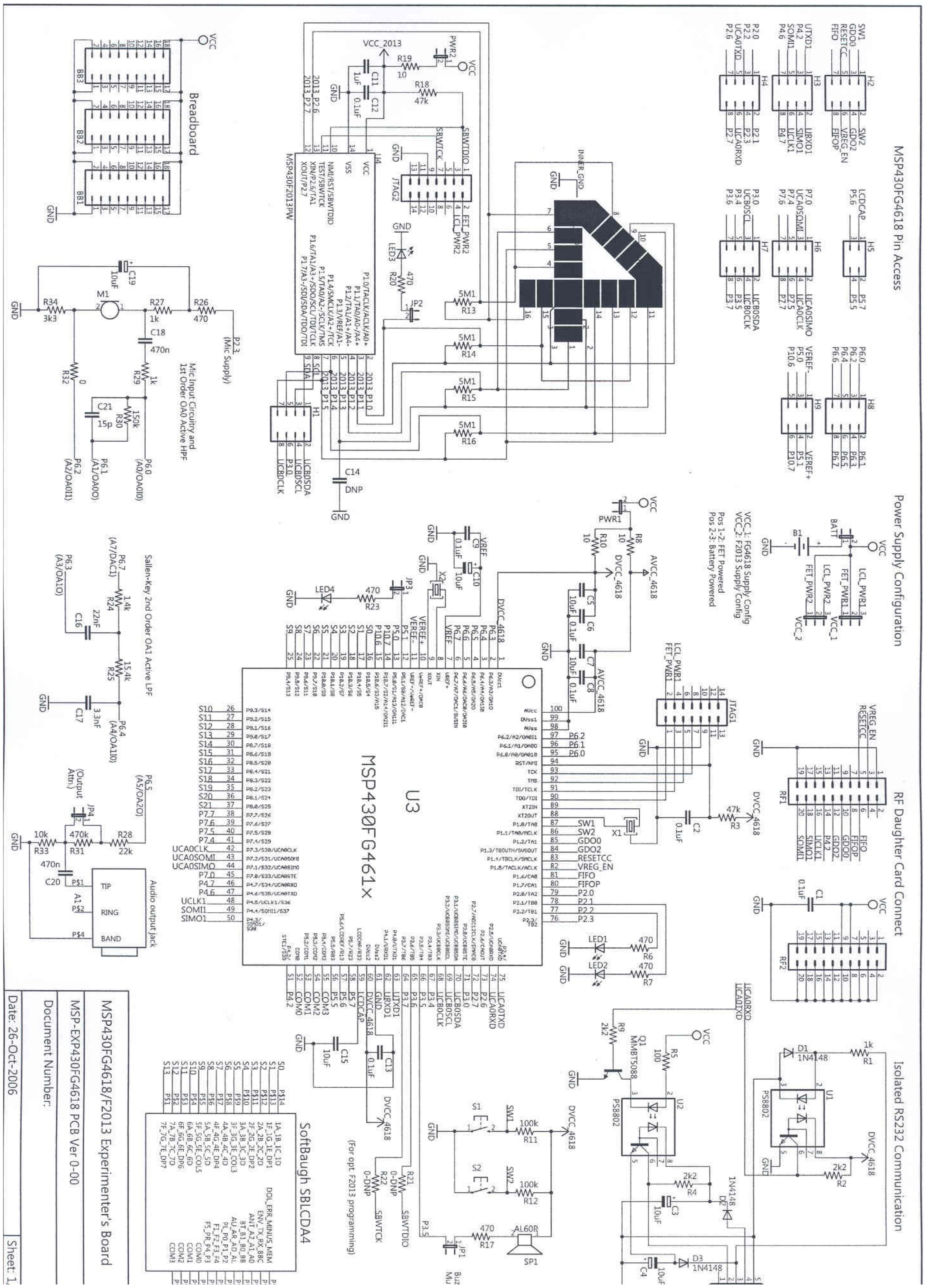
[9] Texas Instruments, "MSP430X461X- Mixed Signal Microcontroller", [Online Document], 2008 [Cited 9 Dec 2008], Available HTTP: <http://focus.ti.com/docs/prod/folders/print/msp430fg4618.html>.

[10] University of Illinois. "BIOE 415: ECG WAV how-to", BIOE 415 Course website, 2008 [Cited 9 Dec 2008], Available HTTP < http://www.bioen.illinois.edu/courses/BIOE415/labs/ecgwav.html>.

[11] M. Hynes. "Microcontroller Based ECG Monitoring System", National University of Ireland, Galway, 2007, Available HTTP: <http://geminga.it.nuigalway.ie/~0049297u/FinalReport.doc>

APPENDIX A – Experimenter's Board Schematics

MSP430FG4618/F2013 Experimenter Board Schematic



APPENDIX B – PICTURES

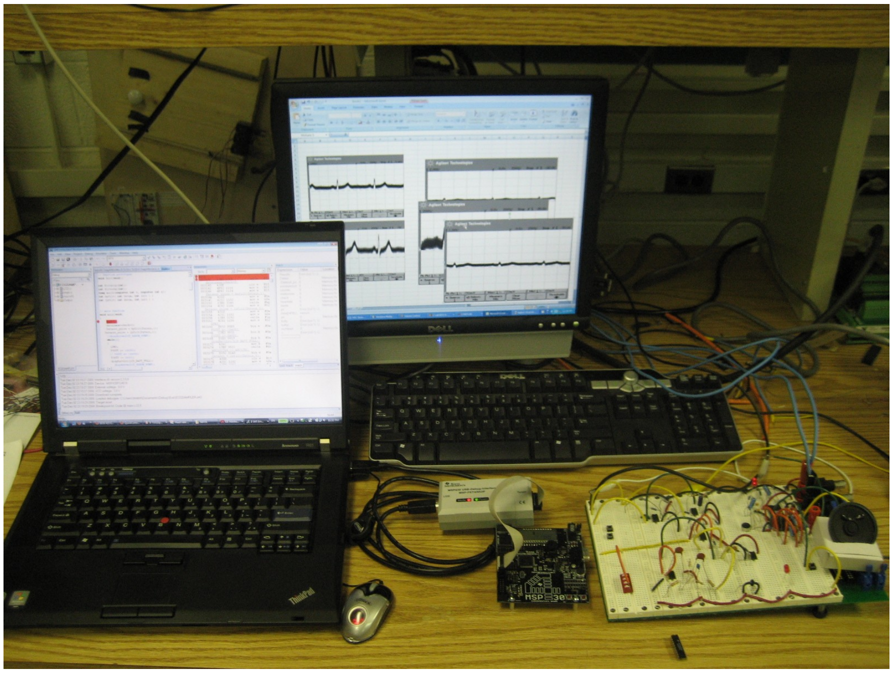


Figure 18: Set Up

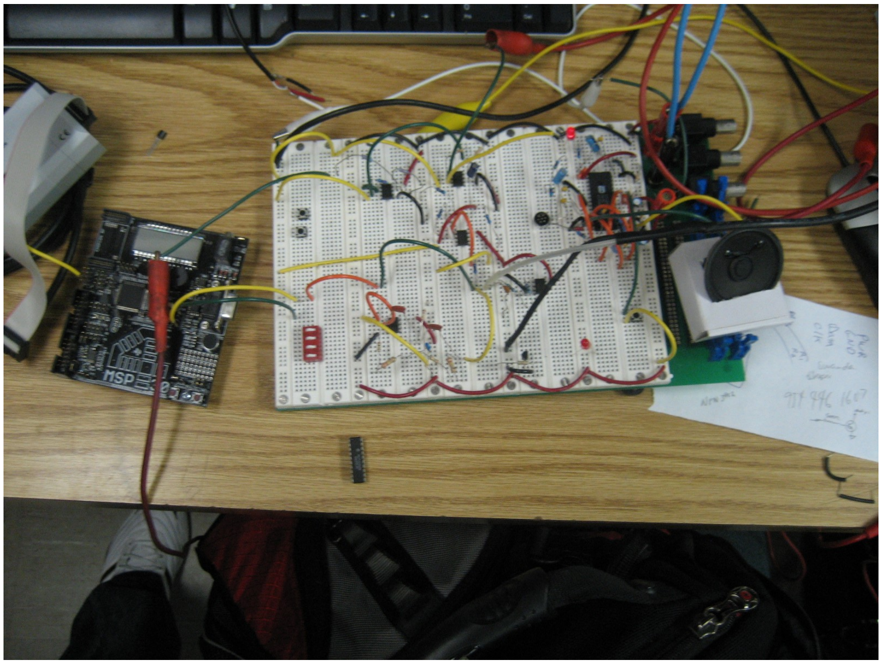


Figure 19: Close up

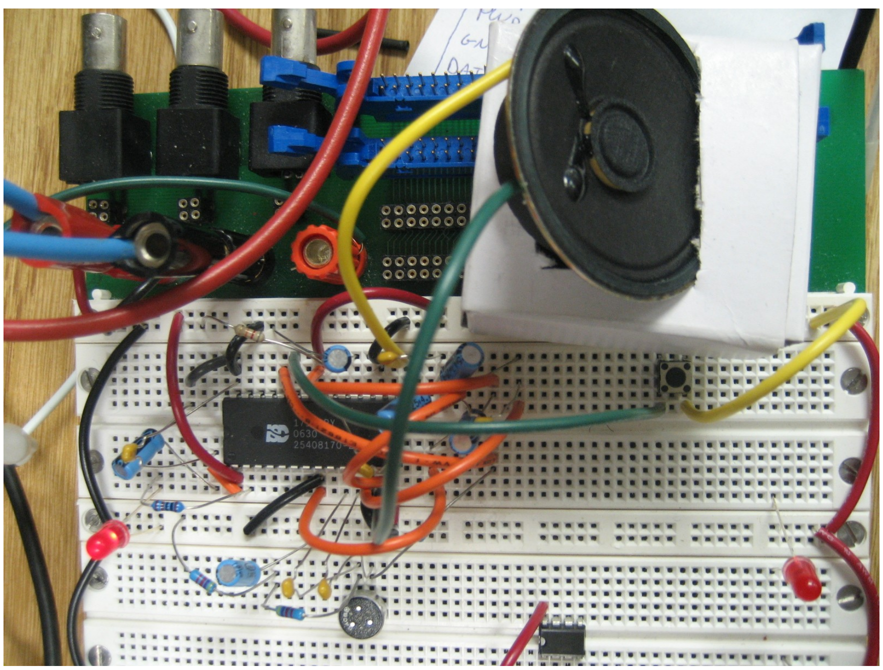


Figure 20: Notification System

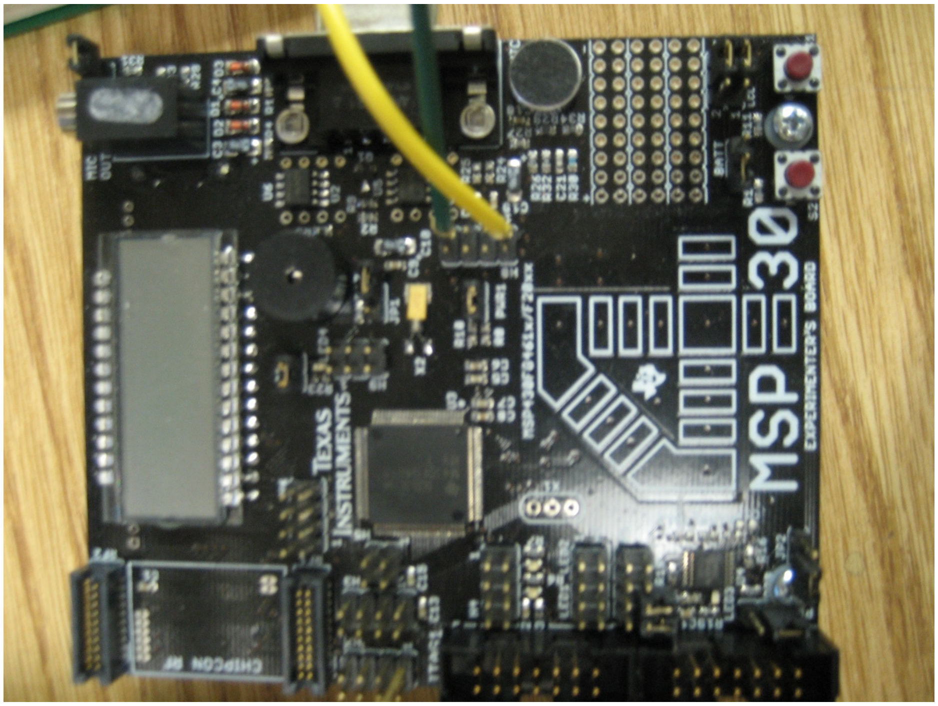


Figure 21: MSP430FG4618/F2013 Experimenters Board

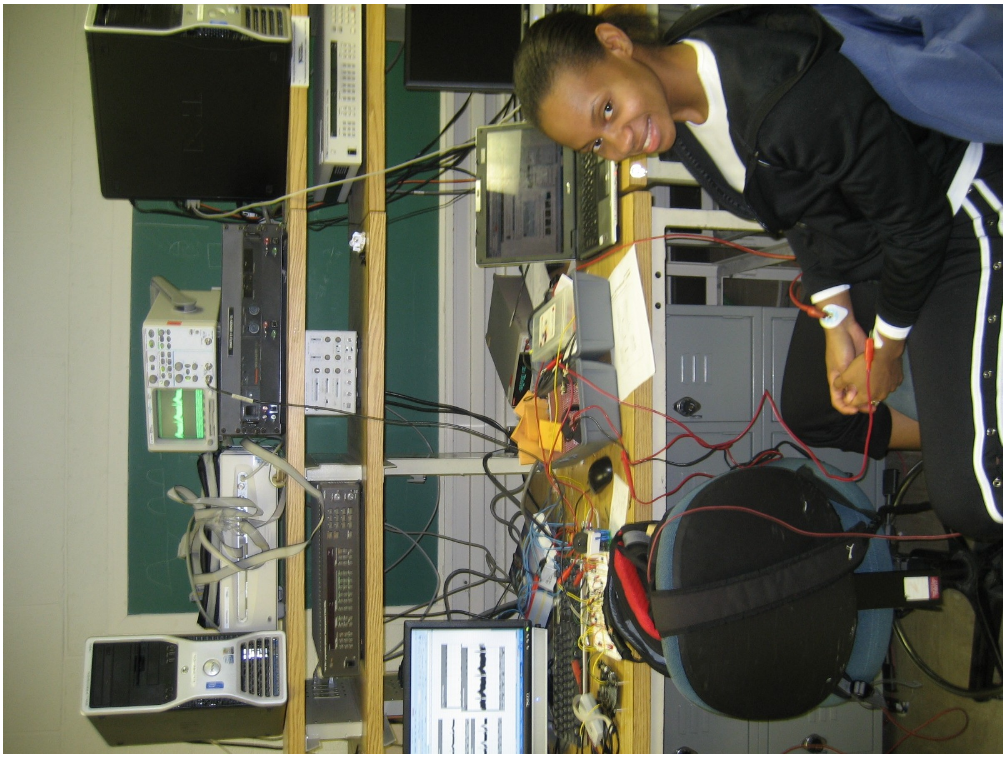


Figure 22: Signal Acquisition