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The Development of an Exercise Monitor
Prototype for the Detection of Arrhythmia
Using the Virtual Instrumentation
Capabilities of Labview

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Abstract

There is a growing and obvious need for exercise pulse monitors that not only register heart beats and record other data such as calories burned but also detect arrhythmias. Ideally, they should identify the type of arrhythmia as well. In the last fifteen years or so there has been an improvement in the quality and capacities of exercise pulse monitors; however with the improvement in the electronics and circuits and with the development of virtual instrumentation to test and develop instruments to a higher degree, the market would welcome an exercise pulse monitor with the above abilities. Therefore the purpose of this project is to construct a basic pulse monitor that can detect and identify five types of basic arrhythmias. The use of such products as LabVIEW in the area of virtual instrumentation to develop and test biomedical products has become a rapidly growing field. A search of technical electronic journals for virtual instrumentation identified over 2500 papers for a six year period. One of the reasons for this is that software products such as LabVIEW can significantly shorten the development time. The completed monitor in this project was developed using LabVIEW and the final testing was done with both simulated and real signals. A major feature of the final product is that it can be customized for a particular wearer. Although testing was not carried out with respect to this feature, it is to be assumed that if a wearer was prone to a certain type of arrhythmia that if that pattern was stored in the monitor, then there would be a greater chance of the arrhythmia being detected during an exercise period. The pulse monitor incorporates real time analysis which of course would be necessary since some arrhythmias develop quickly and can have immediate consequences. One of the reasons why such a pulse monitor could be successful is that the sampling rate that newer monitors are capable of is much higher than it was and also the features of LabVIEW allow for testing parameters that are very fast. The final monitor worked very satisfactorily. This bodes well for such a device being designed developed with significantly better accuracy and arrhythmia detecting capabilities.

Key Words: LabVIEW, Arrhythmia, ECG, Virtual Instrumentation, Supraventricular Tachycardia, Sinus Rhythm, Synthetic Instruments, CleveMed's BioCapture, real time analysis

Introduction

1. Background

Sports medicine has become an important part of the growing athletic movements over the last 40 years. It is now established as an important specialty and is used by both professional athletes and weekend tennis players; however, like mainstream medicine it is just starting to develop preventative medicine aspects. There is an important need for this. In a 27 year study that examined the sudden deaths of athletes, it was found that 57% were attributable to heart problems. [1] Atrial fibrillation (AF), a type of arrhythmia, was found in the report to be a common form of arrhythmia among athletes. Not only was it detected in pre-game situations, but it was found to occur during exercise and games were it had not been detected previously. Also, AF has been associated with a drop in cardiac output, thus leading not only to a health problem, but a lessening of an athlete's ability to perform during a game. Fortunately, AF is seen in an ECG as an irregular ventricular rhythm with an absence of P waves. Therefore, the absence of P waves makes it an easy condition to detect on a portable ECG.

In 2000, a study screened 1005 athletes engaged in 38 different sports of varying aerobic intensity. [2] Overall, abnormal ECGs, including markedly increased R- or S-wave voltages, flattened or inverted T waves, and deep Q waves, were found in 40% of athletes. If this is true of athletes, then amateurs such as high school football and soccer players and weekend joggers should wear arrhythmia detecting devices when ever possible.

Therefore the use of an accurate sports pulse monitor has become an important requirement. Unfortunately there appears to be no pulse monitor on the market that will detect and identify arrhythmia. There could be several reasons for this: legal, marketing and/or technical problems. Hospitals and laboratories have large ECG machines that can detect arrhythmia but, of course, they can not be worn. Therefore all athletes and also those who suspect they may have the potential for arrhythmia should be examined at a hospital. These people would be excellent candidates for a wearable monitor.

2. The System

Over the past 10 years many researchers have tried to develop simple minimalist ECG systems that would detect arrhythmia. The ones that have been developed have been developed are reasonably portable and computer compatible so that they can be transported and hooked up to the net or a phone service in order to be sent to a medical office for monitoring and analysis. There does not appear to be a commercial pulse monitor that is small enough and also designed to be used by athletes during their workouts. Such a system would have only two electrodes and need to incorporate small hardware, yet have embedded software that could detect at least basic arrhythmias. Some detect arrhythmia by default. For example, if the wear's heart is exhibiting triplets, then the monitor will show a sudden pulse increase from say 130 bpm to 211. Therefore, this basic indirect effect can only identify an arrhythmia that results from higher than normal pulse rates.

This project's goal is to produce a device that is small, portable, sturdy and can detect at least five types of arrhythmia. In order to develop such a system, virtual instrumentation was used to create synthetic instruments. To this end LabVIEW, MATLAB and Clevedon's BioCapture were used to develop and test such a system. Why are there no exercise pulse monitors on the market that detect arrhythmia? This is most likely due to liability issues, rather than technology difficulties.

The system is also capable of detecting and recording an individual's unique heart waveform. The next step would be to record and arrhythmic activity. Then when the monitor was worn during exercise, it would be relatively easy to compare the two sets of information.

Literature Review

There have been many projects over the years that improve on portable devices that monitor heart pulses. Even by the 1980's researchers were trying to make the devices more efficient by such means as using an AC current to improve the skin to electrode contact.[3] This type of approach was important 30 years ago but is less so now with the improvement in quality and sophistication of electronic components. In order to get a small working version the researcher had to compromise his design and components. [3] at pp364.

By the 2000's the shift was to using algorithms that had been refined on large systems and also to examining different approaches to measuring the hearts functioning. Therefore, virtual instrumentation products such as LabVIEW were used to develop systems that measure such things as heart rate variability (HRV).[4] The author states that the variations in the intervals between heart beats have been shown to be a valuable marker of the autonomic nervous system activity[5] Interestingly the author uses the Polar pulse monitor as a device to compare the results of his own HRV findings. The Polar monitors have been on sale for many years and are highly regarded; however, like all other exercise devices they do not detect arrhythmia.

A paper published in 2005 takes an approach similar to this paper. The authors used LabVIEW to design a complete PC-based Virtual Instrument as a "testing platform" for acquisition, processing, presenting, and distributing ECG data. [6] The paper reported and had many illustrations with respect to the simulated signals. These were then compared to a real ECG signal recorded by the authors. The researchers choose to focus on the QRS portion of the ECG output. This approach simplifies several problems, but would probably be suitable for an exercise pulse monitor that could detect arrhythmias at a certain level.

The use of LabVIEW as a research tool has increased dramatically over the last 15 years. A paper published in 1990 sets out the programming struggles of the originators of LabVIEW.[7] The author starts the paper with an interesting observation that puts LabVIEWs use in context.

“At a time when financial planners with little technical knowledge were whipping out complex spread sheets in seconds on their PCs, scientists and engineers were spending days getting PC controlled instruments to make the simplest measurements.”

Although the paper is short, there is a very informative description of how the creator of LabVIEW decided to deal with loops. The section shows how the solutions were as more intuitive and powerful than the current languages. When it came out, the author states, LabVIEW was acclaimed worldwide as the first viable “visual” i.e. graphical, language for instrumentation.

Much recent research has been done using LabVIEW for QRS detection and analysis. An example is [8] where LabVIEW was used to create an automated virtual instrument that was used as the main platform. As late as 2009, research was being undertaken using LabVIEW to create virtual oscilloscopes to display QRS data. The research was being carried out so as to improve noise, interconnectivity, and off line processing. [9]

Therefore, as virtual instrumentation software and electronic components improve, it is only a matter of time until a reliable and sophisticated exercise pulse monitor has the capability of detecting and identifying complex arrhythmias.

Statement of problem

1. How Electrical Circuits Work in the Heart.

What makes a heart beat? Heart beats are started by a small electric current that is in the millivolt range. This tiny electric pulse spreads rapidly throughout the heart and makes the heart beat. The electric wave in a heart beat is initiated at the top of the heart. Because of the heart muscle cell's ability to spread its electric charge to adjacent heart muscle cells, this initial wave will be enough to start a chain reaction. The electrical impulse moves in a wave. If there was just one impulse there would be no pumping action.

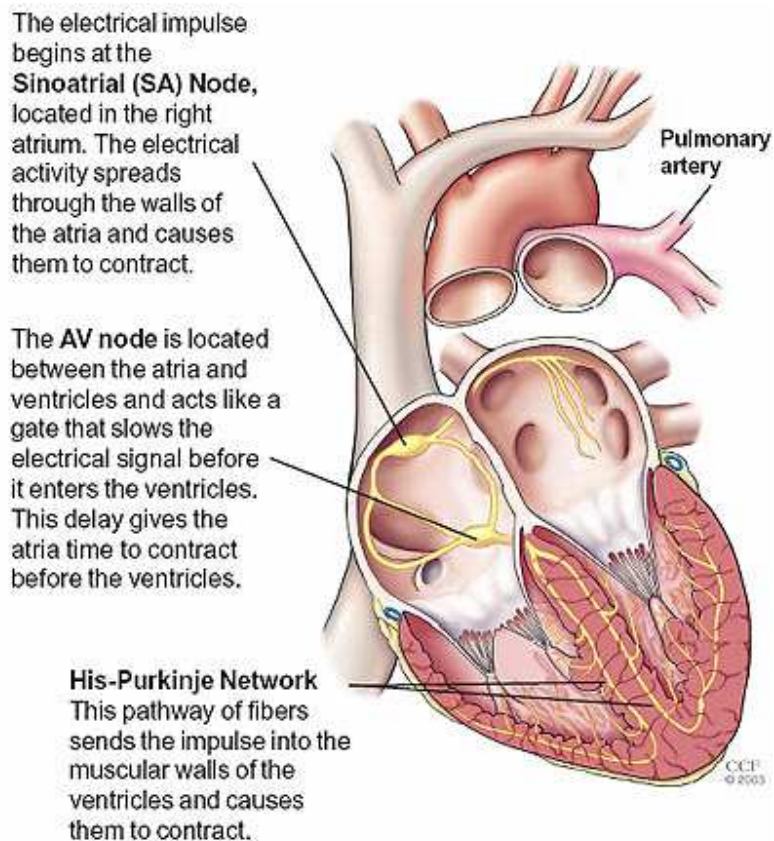


Figure 1. Heart's Electrical System¹

¹ <http://my.clevelandclinic.org/PublishingImages/heart/electricalsystem.jpg>

If the whole heart muscle contracted at the same time, there would be no pumping effect. Therefore the electric activity starts at the top of the heart and spreads down, and then up again, causing the heart muscle to contract in an optimal way for pumping blood.

2. Brief Outline of One Heart Cycle

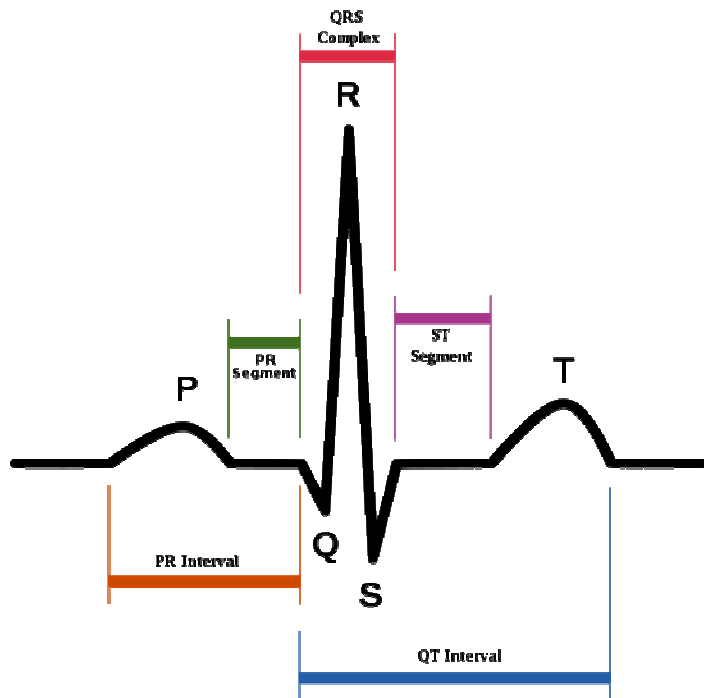


Figure 2. Normal Sinus Rhythm (normal heart beat)²

P wave

When the heart is full of blood the right and left atria receive an electric impulse and that causes both chambers to squeeze. The blood is then pumped through the open valves in the atria to the ventricles. The p wave indicates this contraction.

PQ Segment

When the electrical signal gets to the (AV) node near the ventricles it slows to allow the left and right ventricle to fill with blood.

² www.preventing-a-heart-attack.com/images

Q Wave

The signal then moves to the His in the ventricles where it divides into left and right and moves through the heart's septum. The signal then leaves the two branches through the Purkinje fibres that contact directly to the walls of the ventricles. The signal is travelling rapidly again.

R Wave

The R wave marks the outlines the contraction of the left ventricle.

S Wave

The S wave indicates the contraction of the right ventricle. This contraction is followed by the pulmonary valve opening to allow blood to go to the lungs. The left ventricle pushes the blood through the aortic valve to the rest of the body.

T Wave

The walls of the ventricle relax as can be seen by the relatively flat slope of the T wave. The heart is waiting very briefly for the next cycle.

Summary

1. **P wave** - coincides with the spread of electrical activity over the atria and the beginning of its contraction at the top of the heart pushing blood in the ventricles.
2. **QRS complex** - coincides with the spread of electrical activity over the ventricles and the beginning of its contraction- the spread of the electrical signal through the lower ventricles causing the pumping action
3. **T wave** - coincides with the recovery phase of the ventricles-while the heart muscles are relaxing and preparing for their next contraction.

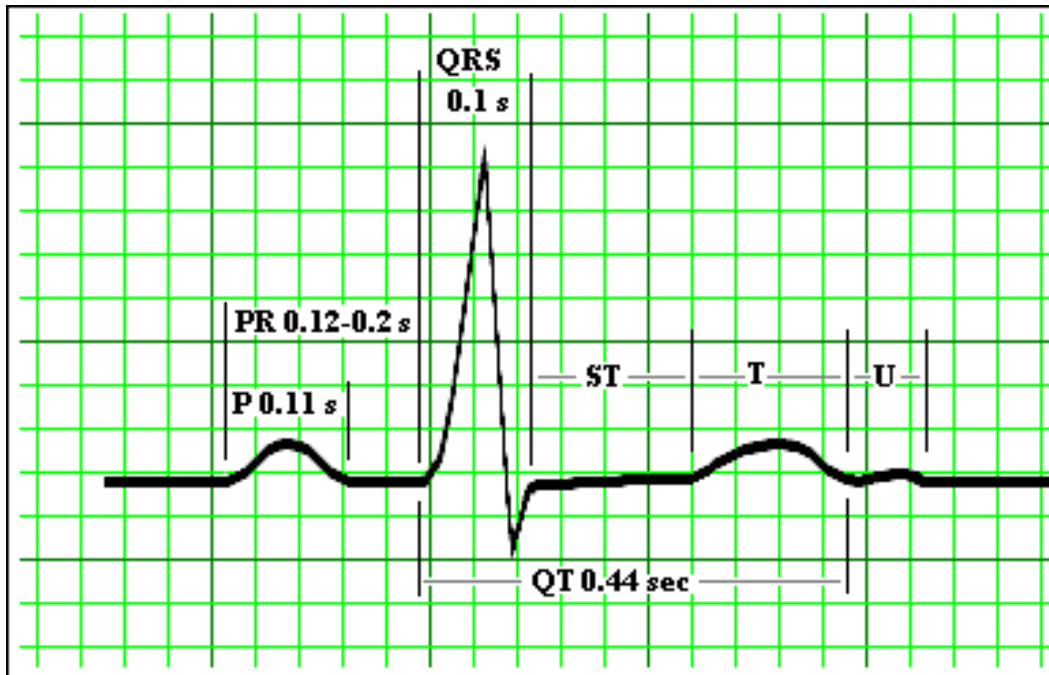


Figure 3. ECG Complexes and Intervals as Related To the Ventricles³

3. Arrhythmia

The term “arrhythmia” can describe a wide variety of conditions in which there is abnormal electrical activity in the heart. Many people who are affected do not know that they have an arrhythmic condition and are able to go into old age because it is passive. However, some arrhythmic conditions are life-threatening. With an increase in the number of middle-aged people exercising, it is presently more important to monitor for arrhythmias during exercise. Further, some arrhythmias only show up during exercise and therefore dormant conditions can be discovered by wearing a small arrhythmia detection device.

Arrhythmias can be classified as to where they begin , (the atria, AV node, or the ventricles). Those that do not originate from the ventricles are called supraventricular arrhythmias while those that come from the ventricles are called ventricle arrhythmias,

Some of the more commonly encountered arrhythmias are as follows:

³ <http://ppmceped.com>

(A) Supraventricular Arrhythmias⁴

Supraventricular tachycardia or paroxysmal SVT: Occurs when any structure above the ventricle (usually the atria or the AV node) produces a regular, rapid discharge.

Sick Sinus Syndrome Irregular firing by the SA node causes a slower-than-normal heart rate (sometimes alternating with rapid heart rates).

Atrial fibrillation: A common condition caused by electrical impulses discharged at a rapid rate from many different areas of the atria. It usually causes a fast and irregular heartbeat.

Atrial Flutter: A condition caused by a rapid discharge from a single place in the right atrium. Typically, the right atrium fires at a rate of 300 beats per minute, but only every other beat is conducted through the AV node, meaning that the ventricle is classically 150 beats per minute.

(B) Arrhythmias Arising in the Ventricle

Arrhythmias arising in the ventricle are more likely to be found in people with more serious heart disease but may also be found in healthy individuals.

Premature ventricular complex or PVC: This electrical impulse starts in the ventricle, causing the heart to beat earlier than expected. Usually, the heart returns to its normal rhythm right away.

Ventricular tachycardia: Fast and usually regular impulses come from the ventricles and cause a very rapid heart rate. This is usually a life-threatening tachycardia and needs immediate medical attention, possibly electrical shock or defibrillation

⁴ http://www.emedicinehealth.com/heart_rhythm_disorders/article_em.htm

Ventricular Fibrillation: Electrical impulses arise from the ventricles in a fast and disordered sequence. The resulting uncoordinated contractions cause the heart to quiver (appearing like a bag of worms) and lose the ability to beat and pump blood, leading to immediate cardiac arrest.

Supraventricular arrhythmias are very common in middle-aged and elderly adults. The older you get, the more likely you are to experience an arrhythmia especially atrial fibrillation.

Many supraventricular arrhythmias are temporary and not serious, especially if no underlying heart disease is present. These arrhythmias are a response to normal activities or emotions.

4. Arrhythmias Detected by the Module Constructed for this Project

Ventricular Trigeminy

Ventricular ectopics may occur in patterns, such as *bigeminy* (every other beat), *trigeminy* (every third beat), or *quadrigeminy* (every fourth beat). They may also occur in sequences, such as *couplets*, *triplets*, *salvos* (four to seven), or *episodes* (eight or more).

Figure 4 shows ventricular trigeminy; the ventricular ectopic beats (three and six) are separated by two sinus beats.

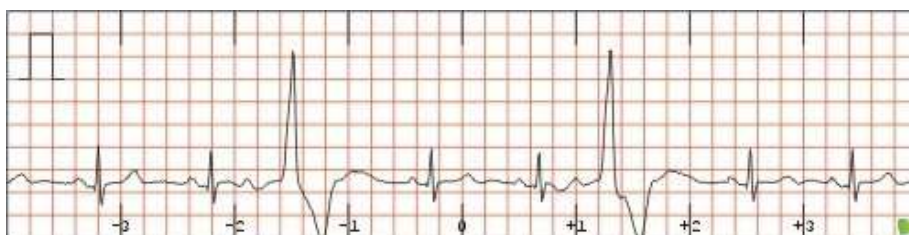


Figure 4. Ventricular Trigeminy⁵

⁵ www.cardionetics.com/cardiology/_cardioimages/rhythms/ecg_ventricula

Ventricular Flutter

In ventricular flutter there is no P wave and QRS complex and T wave merge. Often deteriorates to very dangerous Ventricular Fibrillation. This can only be corrected by defibrillation.



Sinus Bradycardia

Sinus bradycardia: The heart may slow down, yet maintain the normal pattern of rhythm (sinus), this is known as sinus bradycardia. It usually is benign and may be caused by medications such as beta blockers.

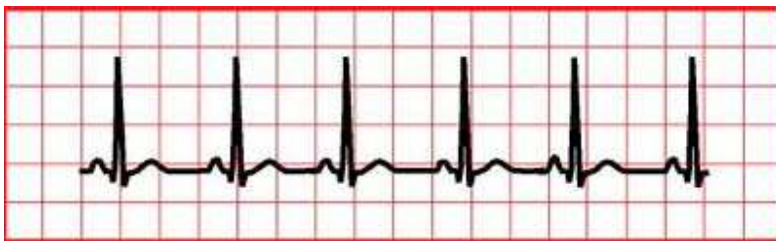


Figure 5. Sinus Bradycardia
All pink figures come from this footnote⁶

⁶ <http://pediatriccardiology.uchicago.edu/PP/abnl%20rhythm%20for%20parents%20body.htm>

Supraventricular Tachycardia

Supraventricular tachycardia (SVT). In this abnormal heart rhythm the impulse stimulating the heart is not generated by the sinus node, but instead comes from a collection of tissue around and involving the atrioventricular (AV) node. These electrical impulses from this abnormal site are generated at a rapid impulse, which may reach 280 beats per minute.

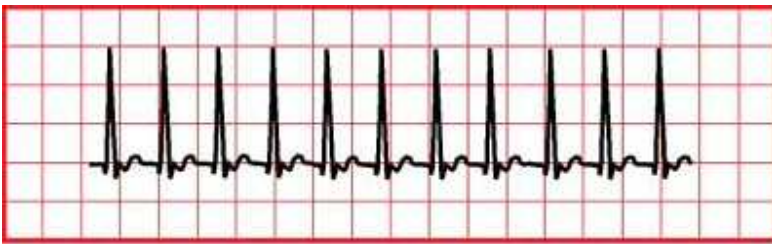


Figure 6 . Supraventricular Tachycardia

Sinus tachycardia: a fast heart rate may occur with a normal heart rhythm, this is called sinus tachycardia. This means that the impulse generating the heart beats are normal, but they are occurring at a faster pace than normal. This is termed sinus tachycardia and is seen normally with exercise or with excitement.



Figure 7. Sinus tachycardia

Basic Methodology

There are three distinct parts of the project that require a lot of development: the analog circuit, the LabVIEW virtual instrument (VI) that is responsible for data acquisition and heart rate detection, and the VI that is responsible for arrhythmia detection.

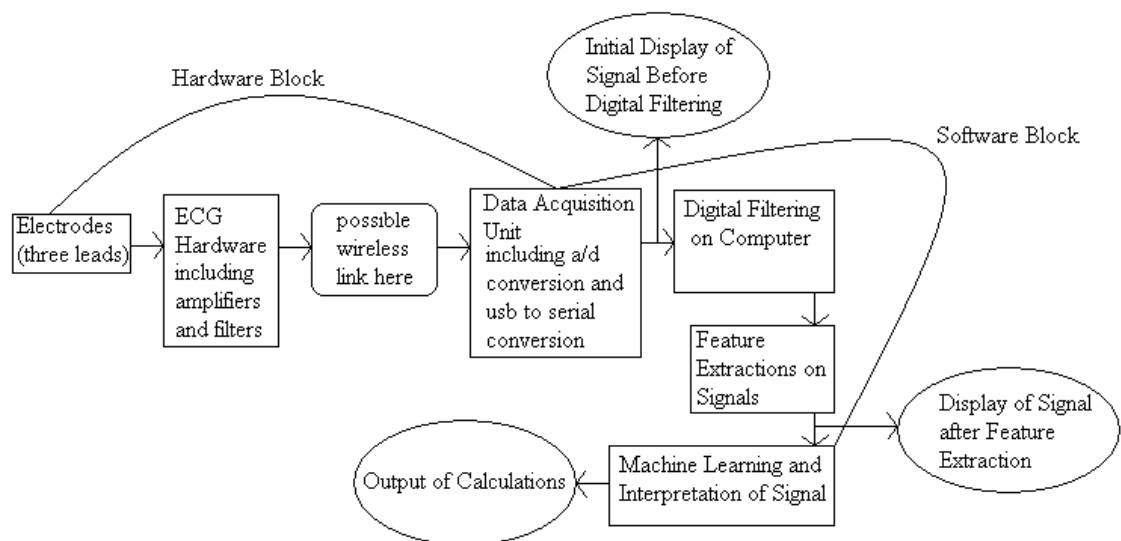


Figure 8. Flowchart of entire project.

The circuit is battery operated and is built for collection of the electrocardiogram signal from a person. One electrode is placed on each of the subject's wrists. Each electrode is an input to the AD620 differential amplifier with a gain of 10. The signal is then amplified by 100, for a total gain of 1000. Two passive high pass filters are used to eliminate DC baseline wandering. An active low pass filter is used suppress contamination from muscle artifact. The output is transmitted to the LabVIEW software on a PC by using one channel on a National Instruments Data Acquisition Unit (DAQ). The DAQ performs A/D conversion.

Using National Instrument's LabVIEW software and MATLAB, two separate VIs are created to analyze data from either a real-time electrocardiogram signal, or simulated data or previously recorded data. The VI using the DAQ reads data into the LabVIEW software, which originates from electrodes on a user. The dynamic data is digitally filtered in the VI, and outputted in a waveform chart after each filtering stage. A changing average heart rate is calculated by identifying the QRS peaks of the waveform. A few check points are put in place to prevent detecting false positives. For example, not only does the amplitude have to be over a certain threshold, but the shape of the peak has to be within a certain range.

The VI that detects arrhythmias uses data from CleveMed's BioCapture arrhythmia libraries. When the BioCapture software is run, the output voltages can be read and saved into a text file. The text files are used as input to the VI because of lack of access to an individual with a specific arrhythmia. Also, a variety of arrhythmias can be correctly determined by using these input. The data is read in real-time, in order to simulate an individual using electrodes for a real reading. The approach to decipher the waveform is to compare the shapes of certain parts of the input signal to shapes of the features of a standard heart wave and features of certain arrhythmias. The data points are analyzed in small groups to keep detailed resolution and limit false positives. The points are stored in arrays and sometimes back accessed using shift registers. Slopes between small sets of data points at certain parts in a time sequence of a data set are compared to theoretical slopes to identify possible events taking place in the shape of the waveform. If the slopes over certain time intervals are within a certain range, one check point is passed. Other check points include having the amplitude of the signal over a certain threshold, which is calculated based on previous values of the subject's unique wave; or checking whether another event has happened within a certain time frame. Many parts of the code where decisions are made are interconnected with other decision making parts of the code in order to minimize false positives. Not only are the shapes of the waveforms event criterion, but their position in time relative to other events are also important.

Scope and Project Goals

There are several goals of this project. Firstly it is desired to acquire a real-time filtered and amplified heart wave signal from a person. This is done by having the leads of electrodes as the input to an electrocardiogram circuit which has several signal conditioning stages, and having the output of this circuit detected by a computer. The next goal requires the computer to display the real-time signal on a GUI. The challenges here lie in the interface between the circuit and the computer, and also in choosing an appropriate programming language to display the data. Next, it is desired to be able to continuously update the determined heart rate. Here, the heart beats need to be identified from the signal. This needs to be done with sufficient accuracy and also use an algorithm that is adaptable for different users. Also, the method of calculating the heart rate, possibly including averages, needs to be determined. Another goal is to be able to read data from files and write recorded waveforms to files for later analysis. The main thrust of the project is to be able to detect a normal waveform, along with 5 types of arrhythmias. There are a few different routes that the code can take to accomplish this task. A method that can reliably identify important features on a waveform is needed. Lastly, the detection system should be adaptable to an individual's unique waveform.

As stated above, the main objectives are to acquire a heart signal and identify specific arrhythmias. When the code was being written, a waveform was created by plotted 100 points and looping this waveforms with different time intervals between consecutive QRS complexes. This showed that the QRS complexes could be identified by the algorithm.

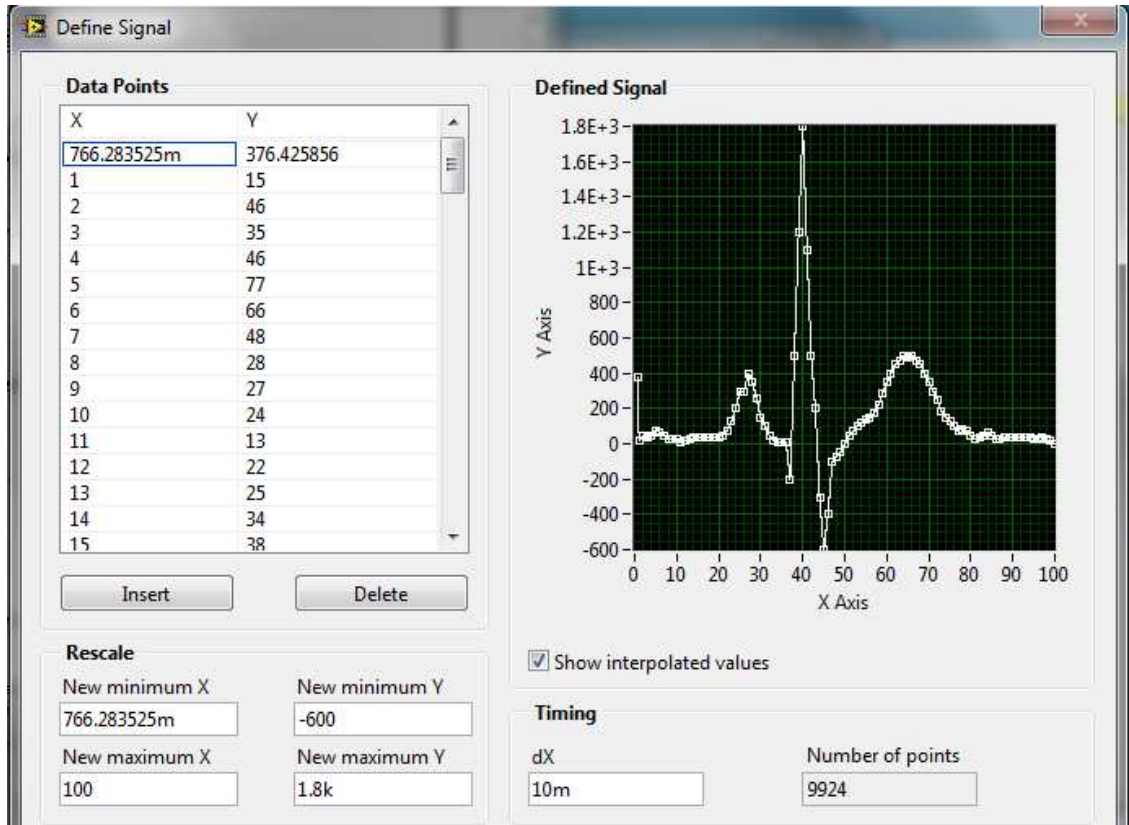


Figure 9. A simulated digital heart signal with 100 points.

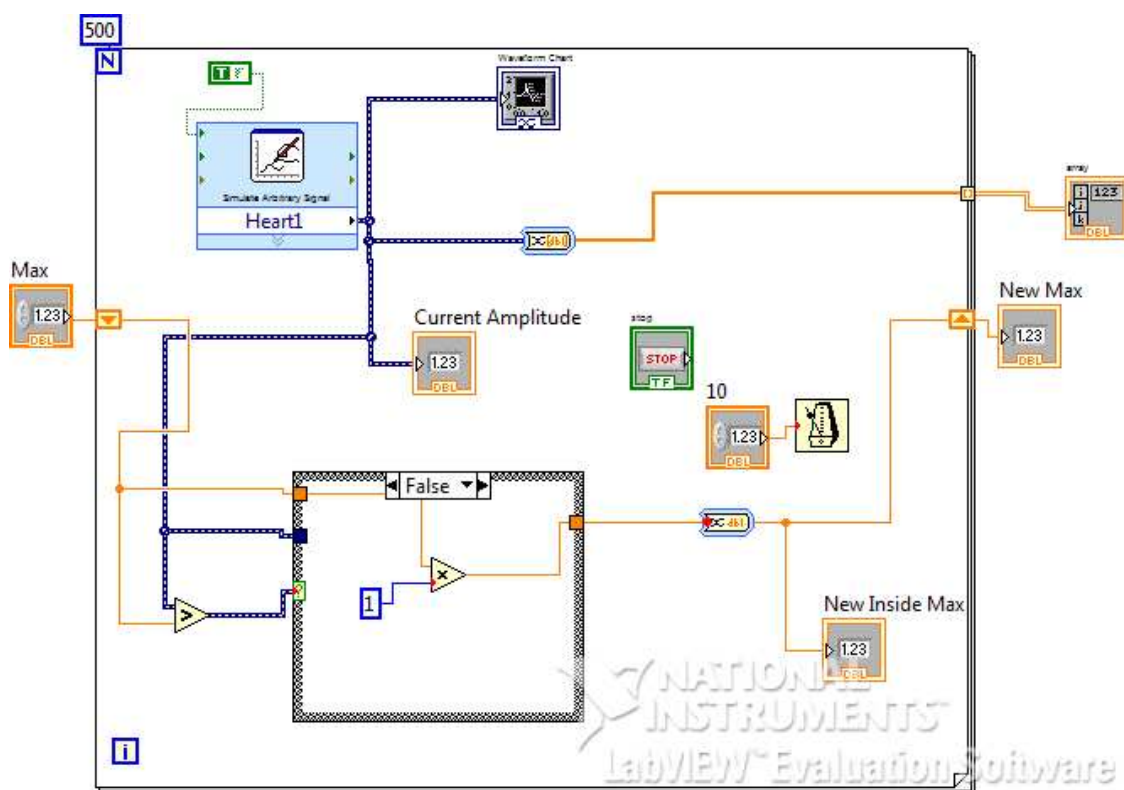


Figure 10. LabVIEW code for peak detection of a heart wave. The for loop runs for 500 counts to make 5 beats because each heart wave has 100 points.

Two types of input are currently used in the solution: a real-time signal from electrodes attached to a test-subject's wrists and saved data in a text file that originated from CleveMed's libraries of sample data. The CleveMed data is used to simulate the specific five arrhythmia waveforms chosen and one standard waveform. The test-subject's input signal shows that the project works for a real person and the software can identify the shape of a heart waveform, but the CleveMed data shows that the software can identify the five specific types of arrhythmias that are used. Therefore, if a test-subject had one of the five arrhythmias that are tested for by the software, the arrhythmia could be identified after calibrating the software to fit the individual's unique heart wave pattern. Both the text file and the signal from electrodes are read into the software

program in real-time, therefore the code that analyze these two sets of data are interchangeable.

The data from the test-subject is acquired from two electrodes, one placed on each wrist. Traditional electrocardiograms may use up to twelve leads, or sometimes five, three, or two leads. Since one goal of the project is to eventually make it portable and wireless, a two lead configuration is chosen to make the system easy user-friendly and small, while still having a high enough quality input for the purposes of arrhythmia detection.

The two leads are used as input to the electrocardiogram circuit. This circuit has an instrumentation amplifier which is a differential amplifier that can handle two inputs. The signal is then amplified by 1000, in two separate stages, and also put through a series of active and passive filters to remove the unwanted parts of the frequency spectrum. The frequencies that are allowed to pass are between 0.1 Hz and 200Hz. The DC offset is removed by the high pass filter and the unwanted higher frequencies are removed by the low pass filter. The output of the filter is then fed into the data acquisition unit (DAQ) made by National Instruments.

The data that was sent into the DAQ is read into a LabVIEW VI and stored in an array. The data is also filtered digitally to remove any lingering values outside the desired frequency range and also any 60 Hz noise. The data is put through a series of checkpoints to identify the shape of the waveform. The previous values of the waveform are compared to the current values being read in by accessing specific elements of the array. The filtered waveform, after each stage of filtering, is outputted to the front panel in the form of a waveform chart. This chart is updated in real time and has the ability to be back-accessed by the use of a time axis scroll bar. The data can also be stored in a text

file specified by the user, for later analysis. The heart rate over the last five beats is displayed to the screen, as well as the identification of one of three types of heart conditions: normal rhythm, bradycardia, or tachycardia. Also, an array is displayed on the front panel which shows the time of occurrence for every beat.

The separate LabVIEW VI that detects arrhythmias has similar code for detecting heart beats and the heart rate, but the data is read in from a text file in this case. Samples of data that have around thirty seconds of arrhythmia waveforms are saved into text files. The data originates from CleveMed software and each text file represents one type of arrhythmia. The simulated voltage levels are read into the VI and stored in an array. The waveform is displayed to the screen in real-time. No digital filtering is done in this VI. There are a number of separate branches of code that check for the different types of arrhythmias, and each branch has several checkpoints, sometimes cross-checking with other branches. Previous values of data are accessed by using shift registers. Two MATLAB script nodes are used to calculate the heart rate and to check for ventricular bigeminy and ventricular trigeminy. If a specific type of arrhythmia is detected, an LED on the front panel will light up, alerting the user that there may be a problem with their heart conduction system.

Experimental Design Procedures

1. The Circuit

The circuit is powered by two 9V batteries in series, where the ground of the circuit is taken between the two batteries, and the positive side of one battery is taken as +9V and the negative side of the other battery is taken as -9V. The batteries power the two chips used in the circuit: the AD620 and the OP491.

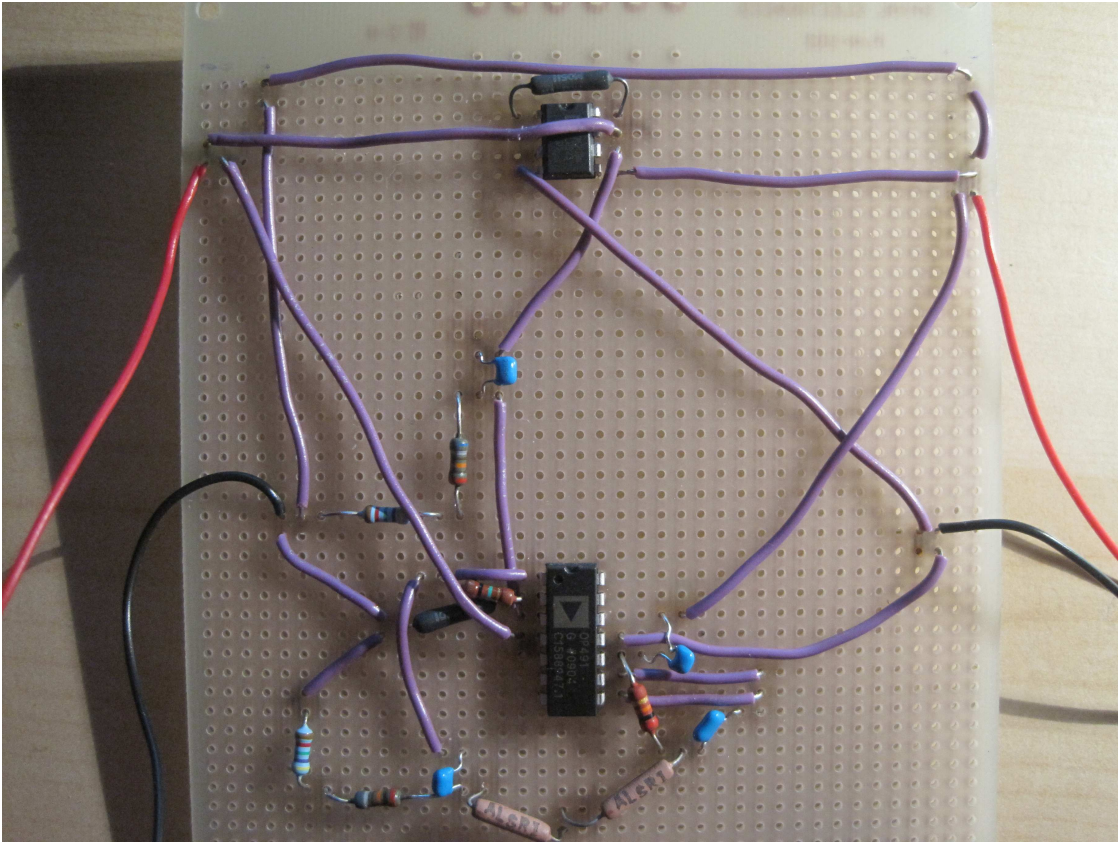


Figure 11. Soldered circuit.

The two leads that are connected to the test-subject's wrists are also linked to the positive and negative inputs of the differential amplifier in the electrocardiogram circuit. The instrumentation amplifier used is the Analog Devices' AD620. It has many properties that make it well suited to this application. The gain can be set with one external resistor. The total gain in this circuit is set to 1000, with a gain of 10 from the AD620 stage. It has low noise (0.28 mV p-p Noise for frequencies between 0.1 Hz and 10 Hz). The AD620 also has a high common mode rejection ratio of 100 dB minimum for a gain of 10. It is important for a differential amplifier to have a high common mode rejection ratio because this indicates that the device can reject signals that are common to both inputs. The desired information in an electrocardiogram is in the difference in signals between the two leads, so a clear difference between the two is important.

The output of the differential amplifier is connected to a passive high pass filter that has a cutoff frequency of 0.1 Hz. The high pass filter is intended to remove any DC offset of the signal so that the beats will be fairly aligned with one another vertically. Sometimes there is a preference to use active filters because the shape of the response can be altered. It is sometimes desired to have a sharp cutoff frequency around specific cutoff frequencies. However, the tradeoff might be a large ripple in the pass band region's amplitude. However, a passive filter has guaranteed stability. The reason that a passive filter is used in this circuit is that the resistors and capacitors used in an active high pass filter would be very large compared to the values used in the rest of the circuit.

The output of the high pass filter is put into a non-inverting amplifier of gain 100. The differential amplifier's gain of 10 is multiplied with this gain of 100 to make a total circuit gain of 1000. The non-inverting amplifier uses an op-amp from Analog Devices OP491 chip. This chip is convenient because it contains four separate op-amps inside its case. It also has no phase reversal and is well suited to battery-powered instrumentation. Its gain bandwidth product (GBWP) is 3 MHz, which is well above the 0.2 MHz of this circuit (from 200Hz x 1000).

The non-inverting amplifier is cascaded with another passive high pass filter for precaution. This is not necessary, but since the cutoff frequency is so low (0.1 Hz), it may be shifted after the last stage, so this filter removes a potential shift.

This output is fed into the last stage of the circuit: the active low pass filter. This filter also uses the OP491. Its cutoff frequency is 200 Hz. This is a high cutoff, considering that an electrocardiogram signal can be appropriately displayed using a range of 0.1-100Hz or sometimes even a smaller range. However, it is desired to keep as much of the signal as possible when analyzing the unique shape of an electrocardiogram with

arrhythmia, especially the higher frequency components that show detail in the signal. A second-order 3 dB Chebyshev filter is used because it allows for a sharp cutoff. The trade off is a few ripples near the cutoff edge. One disadvantage of using a lower order filter, such as a second order filter, is that it has a shallower cutoff slope than a filter of a higher order. However, a higher order filter may have shifting of cutoff frequencies, so the desired cutoff may not be attained. In order to make a higher order filter, several first and second order filters must be cascaded. With every cascaded filter, the cutoff frequency is shifted slightly. Also, higher order filters are more expensive, take up more room, and have a greater chance of connection problems.

The tolerances of the components are 1 and 5% resistors capacitors. These tolerances allow for high accuracy filters with cutoff frequencies that are true to the calculations. Very precise components are needed for a 60 Hz notch filter, for example. However, a notch filter is implemented in the LabVIEW software because of problems with accurate implementation in the analog circuit. Also, it took a long time to get a working circuit with the correct output. For example, the gain of the first stage read as 30 for a while, when it was supposed to be 10. The breadboard was taken a part and built from scratch and worked thereafter, probably due to better wire connections. For this reason, a circuit was built on a perforated board and soldered in for permanent, strong connections.

2. Data Acquisition Device

The output of the circuit is connected, by wires, to the data acquisition unit (DAQ). The wires extend from a coaxial cable that connects to the input channels of the DAQ. The input channel is selected as "a0", both on the board and in the program. The DAQ Assistant is used in the LabVIEW VI to interface between the DAQ unit and the code. The acquisition mode is set to one sample on demand. The sampling rate is set to 1000 samples/second or 1000 Hz. The type of signal measured is the output voltage of the circuit. The DAQ unit performs analog to digital conversion. Therefore, the sampled

data that is read into the LabVIEW VI is a continuous stream of discrete data points of double precision.

3. LabVIEW Code

LabVIEW is a graphical programming language that is commonly used for data acquisition and instrument control in industry and research labs. Two panels are used to make a complete running VI: a block diagram and a front panel. The block diagram has objects that are wired together to perform special functions. It corresponds to lines of text found in a more conventional language such as C or MATLAB. This executable code compiles into machine code. The front panel is window that the user sees and interacts with to control the program. LabVIEW is a suitable language for this project because of its real-time data acquisition graphical displays and its ease of use for users.

In order to run the VIs included, the computer has to have the 8.5 version of LabVIEW and also have MATLAB. The MATLAB R2008b version was used during the writing of the code, but other versions may also be compatible. Also, to run `CircuitInterfacingCode.vi`, a DAQ needs to be attached to an input circuit and also attached to the computer, otherwise an error occurs.

4. Circuit-Interfacing LabVIEW Code

The following is an explanation of the LabVIEW VI that is capable of interfacing with the DAQ unit. It is called “`CircuitInterfacingCode.vi`”. In this VI, the data is read in from DAQ unit, after being filtered and amplified by the analog circuit. The dynamic data is then converted into a double precision array.

The vector data is put through a series of digital filters. There are two reasons for this. Firstly, any unwanted frequencies that are not removed by the analog circuit can be further suppressed. Secondly, the parameters of digital filters can be more quickly

changed than an analog filter's parameters. This allows for more easily identifiable optimization parameters. The data is first put through a second order Butterworth filter. A Butterworth filter is chosen for maximal flatness of the response. This is a low pass filter at 200 Hz. Next, a fourth order Chebyshev high pass filter at 1 Hz removes the DC offset. Chebyshev coefficients are used for a steep cut off, as well as fourth order parameters. Also, as a precaution, a 60 Hz band stop filter is used to remove any power line interference. The signals outputted to the screen on the waveform chart are very close to one another, depending on whether the notch filter is used or not. This suggests that there is not much interference present. The waveform is displayed to the front panel after each stage of filtering. A notch filter is implemented digitally, rather than in the analog circuit, because of the precise calibration needed to remove only the 60 Hz frequency. Problems in output signal had arisen when an analog notch filter was used in a prior circuit design.

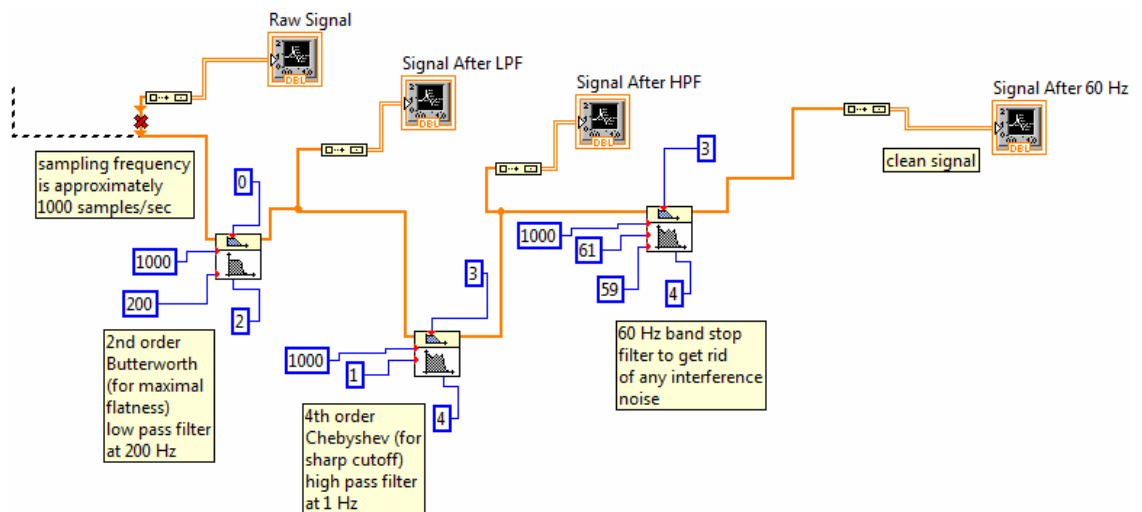


Figure 12. Digital filters in CircuitInterfacing Code.vi

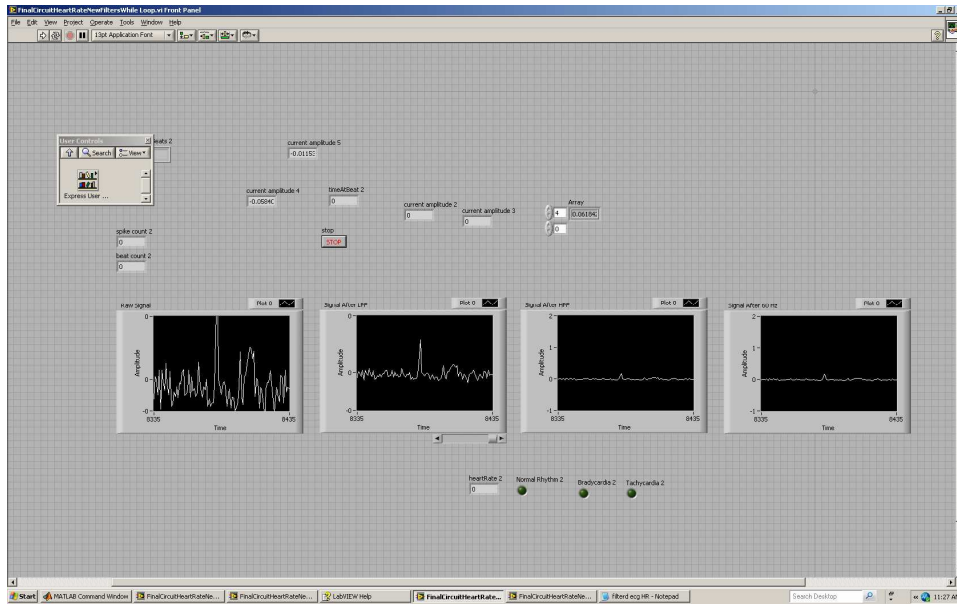


Figure 13 . ECG signal from a real person, showing filtered output after several filtered stages.

When the stop button is pressed, a dialog box appears that allows the user to save the collected data to a text file for future analysis. This is potentially useful for identifying long term trends and also characterizing an individual's unique waveform.

Three types of heart wave types are identified in this VI: normal sinus rhythm, bradycardia, and tachycardia. These three categories of heart conditions are more common to potential test subjects. In fact, it is perfectly normal to have tachycardia when one is exercising and bradycardia when one has been resting for a while. The definitions used in the code are tachycardia is a heart rate over 100 beats per minute (bpm) and bradycardia is less than 50 bpm; however, these definitions are somewhat loose in reality and may vary from one individual to another, depending on such factors as age and physical fitness levels. Normal sinus rhythm is defined as a heart rate between 50 and 100 bpm, but also having a waveform including all of the classic features of a standard waveform, without any abnormalities.

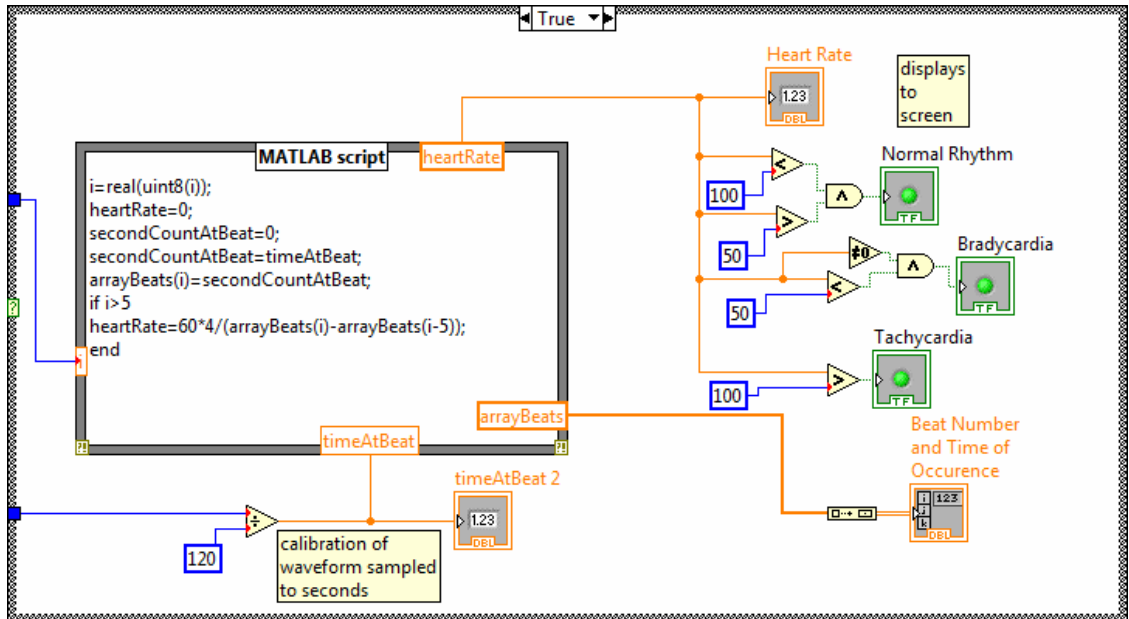


Figure 14. An example of decision code in CircuitInterfacingCode.vi

When a user attaches electrodes to their left and right wrists and the circuit is connected to the DAQ, which is connected to the computer, and the code has been initialized, the user's heart waveform is displayed to the screen in real time. The user's heart rate is detected and is usually in the normal range. If the user is very calm and still for a while, their heart rate may dip below 50, and bradycardia can be detected. Conversely, if the user moves around quickly and attains a heart rate over 100, tachycardia is identified. However, sometimes muscle electrical signals and movement of the electrodes against the skin interfere and there is noise contamination.

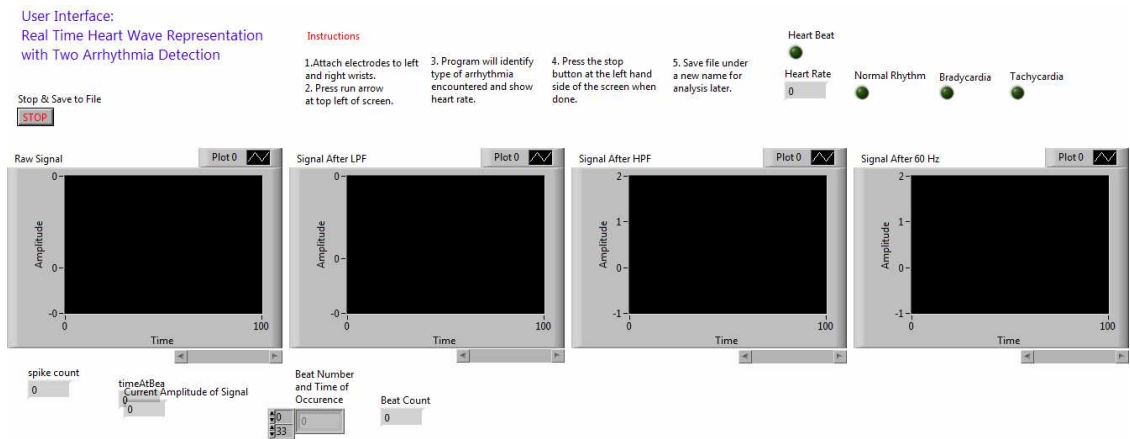


Figure 15. Graphical user interface for circuit interface code.

The other three arrhythmias are solely identified in the next VI because there are no known test subjects that have these arrhythmias. However, it is important to note that these arrhythmias can be potentially be identified by the circuit-interfacing code because the arrhythmia detection code can correctly identify them and the circuit-interfacing code can read in and do calculations on real-time signals including other types of heart waveforms.

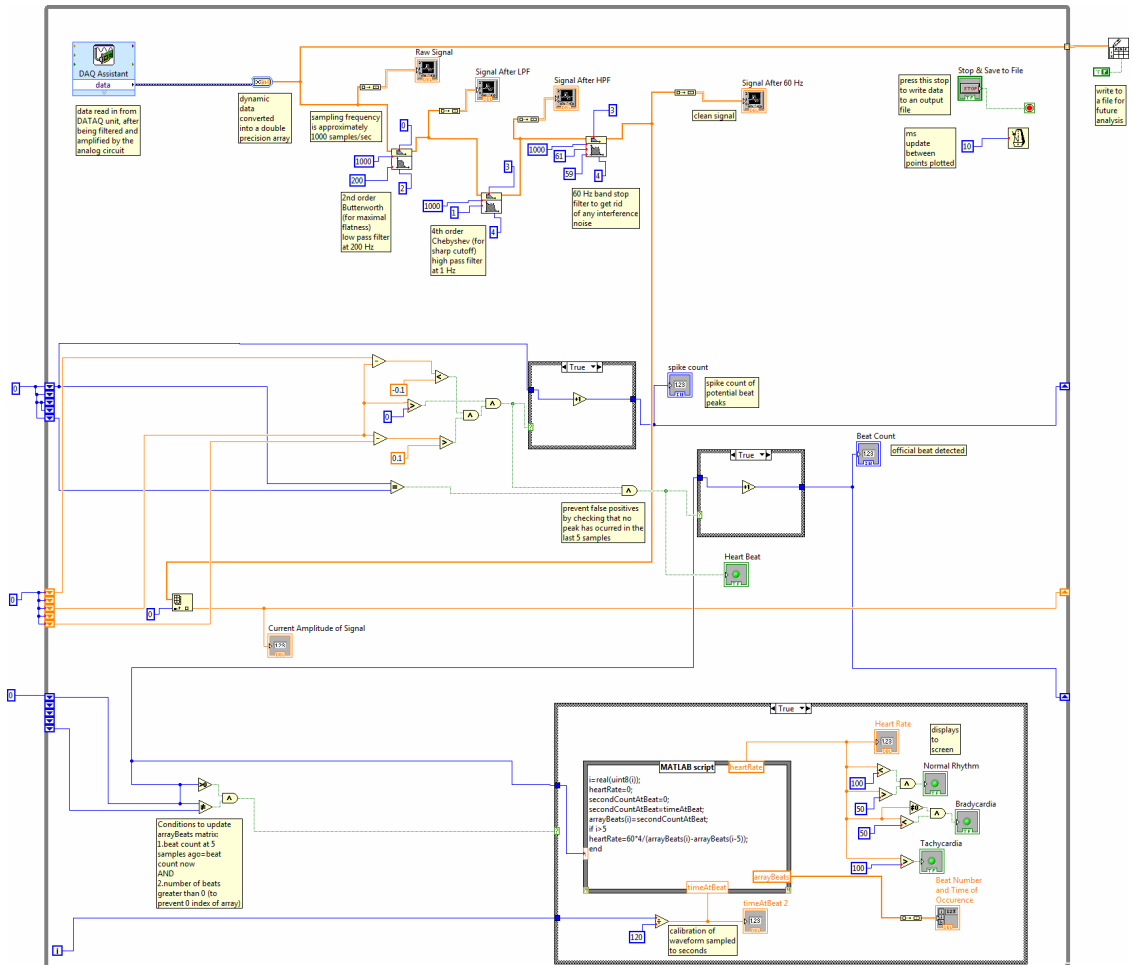


Figure 15. Graphical user interface for circuit interface code

5. Arrhythmia Detection Code

There are two VIs that detect the five types of arrhythmias: BestWorkingArrhythmiaCode.vi and CleanedUpArrhythmiaCode.vi. The former detects all five arrhythmias very accurately, but has a difficult to follow block diagram. It also has a frequency spectrum output of the arrhythmia file that is loaded. Each arrhythmia has a slightly different frequency spectrum, but not significant enough to be able to distinguish between the different types of arrhythmias. Fourier analysis is usually used in electroencephalogram analysis where there are multiple complicated signals being

analyzed. The latter has an easy to follow block diagram with comments and a more visually pleasing front panel, however, the code isn't always as accurate. The “cleaned up” version was made by writing a new VI by looking at the original version; therefore there are probably some semantic errors that are difficult to spot. Therefore, when viewing the code, the block diagram for CleanedUpArrhythmiaCode.vi better explains the thought process of the code, but the front panel of BestWorkingArrhythmiaCode.vi should be run to get more accurate results.

The following is an explanation of the arrhythmia detection code. Firstly, the user can chose a rate at which they would like the digital readings to be updated on the waveform graph. The number entered in the control field on the front panel is measured in ms. Usually 10 ms is best choice. A higher number will make the waveform go across the screen at a slower rate. Next, the user must press the run arrow at the top left of the front panel window. This opens a file-loading window where the user must choose a file to run. Each of the six possible files to choose from is a text file that contains the amplitude of a simulated heart wave from CleveMed's sample data. Once a file is chosen to be read, the Read From Spreadsheet File.vi express object is used to read in each amplitude value in the text file, one by one, until the end of the file is reached or the user stops the program. The amplitudes are read into elements of an array and the total number of elements in the text file controls the number of iterations of the main for loop of the VI. If more than 100 sampled elements have been read into the array, the main decision making part of the program can be executed. This clause prevents access of negative array indexes in element comparison calculations.

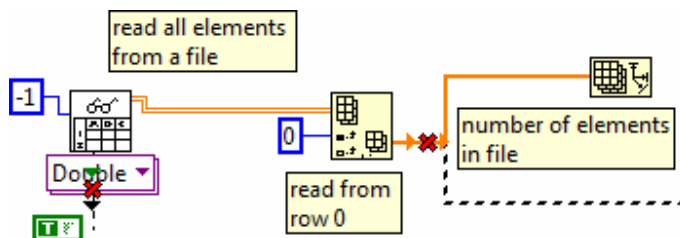


Figure 16. Data read in from a text file.

A real-time waveform is displayed to the screen by charting each element as it is read in. A running sum of elements is also computed so that the current average amplitude value of the waveform can be calculated so that a potential DC offset can be detected.

From this point on, there are many branched decision paths that are used to identify the correct arrhythmia. Some of these decisions back check with other branches so that two arrhythmias cannot be identified at the same time. Some branches do not back check, however. This is because it is technically true, for example, that a supraventricular tachycardia wave may have bradycardia properties because the time gap between successive beats is slower than a normal rhythm.

The arrhythmias are identified as certain defining features of the arrhythmias are detected. When an arrhythmia is detected, the corresponding Boolean LED lights up on the front panel. However, if a certain defining feature or event has not been detected in a specified time range, the LED turns off. Therefore, the LEDs are controlled based on feature detection.

6. Ventricular Bigeminy or Trigeminy Detection Branch

One of the branches detects ventricular bigeminy and trigeminy. It starts off by checking whether there is a fairly flat rising and falling slope present in the waveform. This is indicative of the beginning of a presumed ventricular bigeminy or trigeminy wave. This is done by back-accessing array elements that correspond to samples from previous times.

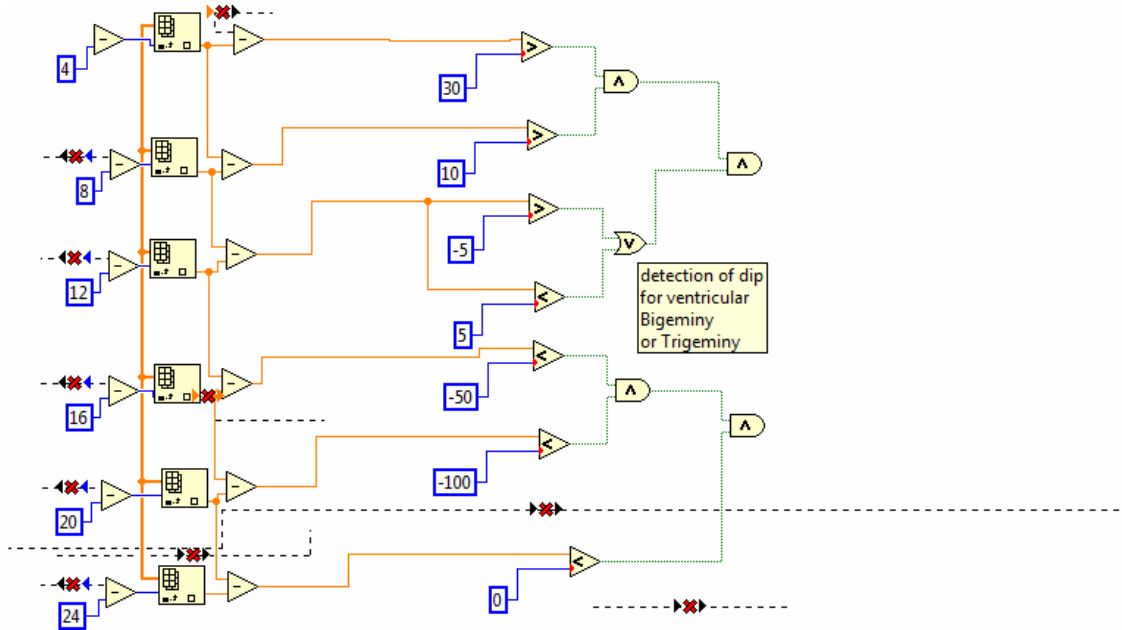


Figure 17. An example of code for identifying the shape of the waveform.

Small slopes over short sample intervals (usually four samples, which represents 4 / 1000 ms) are computed by subtracting the amplitudes at various indexes of the array and dividing this difference by the sample interval. These slopes are calculated over small, neighbouring intervals so that the waveform's shape can essentially be traced. Next, the waveform must come out of the dip with certain steepness. Also, the element at the presumed middle of the premature ventricular contraction (PVC) dip needs to be less than 90% of the current waveform average amplitude.

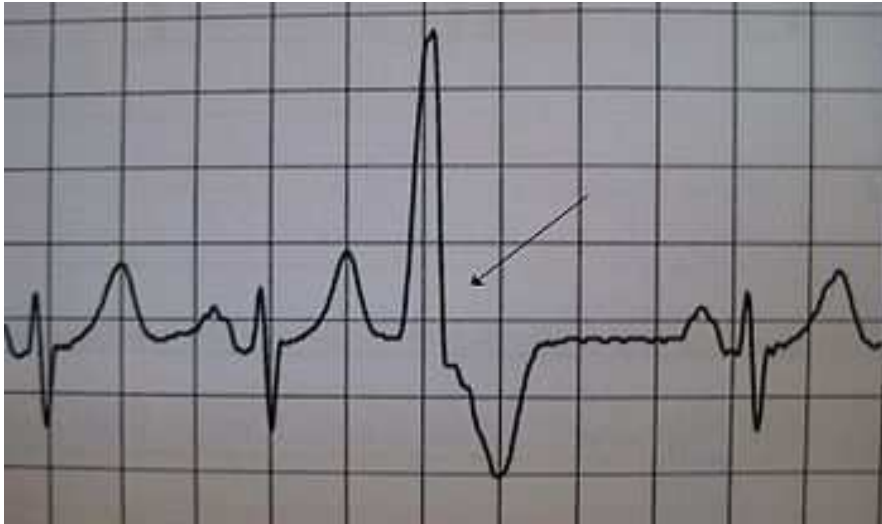


Figure 18. A premature ventricular contraction.

Next, the PVC count at the present time must be the same as the count seventy samples ago. This check point prevents detection of a feature that appears to be a PVC but is not. The feature may have multiple small ripples that are identified as a few PVCs that are close together, which would deliver a false identification of a PVC. This condition is generous enough in the time range because two PVCs would not occur that closely together. If this condition is met, the variable event2 is set to equal 2. In the heart rate detection part of the code, if a heart beat is detected, the variable event1 is set to equal 1.

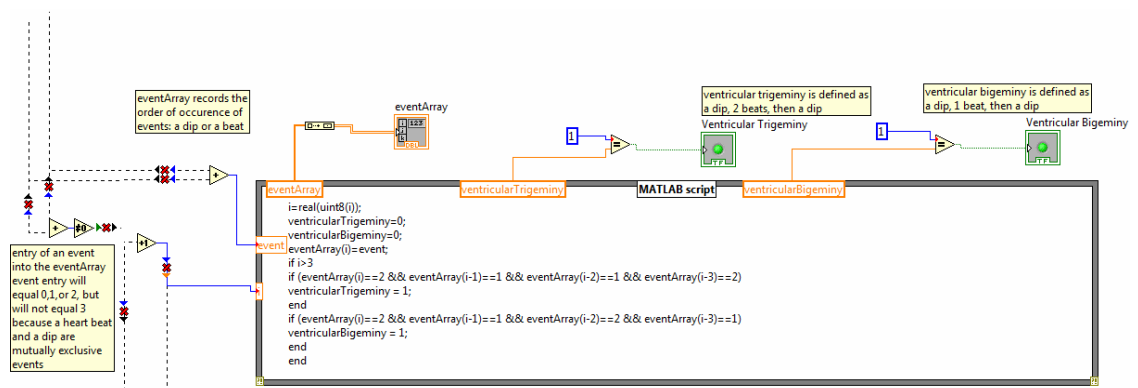


Figure 19. Code

If either of these events are detected, the order of these events is analyzed using a MATLAB script node. (A script node is used because storage of elements in an array is more straight forward in MATLAB than in LabVIEW. The value of the event (one or two) is entered into the array eventArray. This array records the order of occurrence of events: a beat or a PVC event entry will equal 1, or 2, and no event will equal 0. The addition of the values of event1 and event2 can equal 0, 1, or 2, but will not equal 3 because a heart beat and a PVC are mutually exclusive events. The MATLAB code deciphers whether the arrhythmia is bigeminy or trigeminy based off of the order of these events. If there is a PVC followed by two heart beats, followed by a PVC, the trigeminy is identified. If the pattern of a heart beat then a PVC, then a heart beat, then a PVC is identified, then bigeminy is indicated.

7. Ventricular Flutter Detection Branch

The detection of ventricular flutter involves several conditions. Firstly, the detection of shallow positive and negative slopes are needed, But either multiple shallow positive or multiple shallow negative slopes in a row must be detected, so that the waveform has consistently increased or decreased over an interval of time. If this condition was not in place, then the waveform may just vacillate around some slightly positive or negative value that appears to be a positive or negative slope. Next, the identification of ventricular flutter must happen independently of ventricular bigeminy or trigeminy. Thus, ventricular flutter cannot be identified if more than 2 ventricular bigeminy or trigeminy PVCs have been counted and also cannot be identified unless the number of ventricular bigeminy or trigeminy PVCs detected now is the same as the number 70 samples ago. This is implemented by using shift registers.

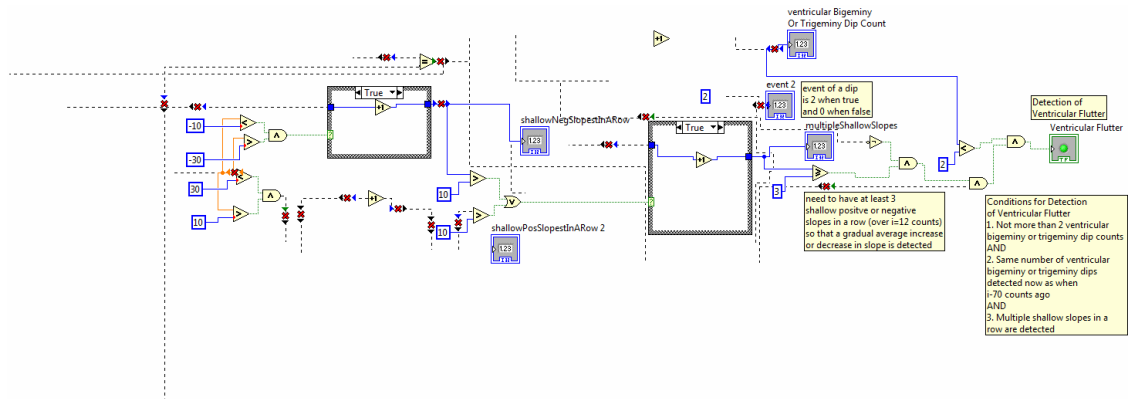


Figure 20. Ventricular flutter decisions.

8. Normal Sinus Rhythm, Bradycardia, Tachycardia, and Heart Rate Detection Branch

This branch uses similar decisions to the previous branches. The conditions for a heart beat include:

1. Slope over 4 elements between $i-32$ and $i-28 > 40$

and

2. Slope over 4 elements between $i-28$ and $i-24 < -40$

and

3. Amplitude at element 28 $> 1.1 \times$ average amplitude

and

4. Count of heart beats = count 44 i counts ago

and

5. Ventricular Bigeminy or Trigeminy Dip Count = same as when $i-70$

The first condition identifies the steep rising slope of the QRS complex. The second condition identifies the steep falling slope of the QRS complex, right after the steep rising slope. The third condition requires the amplitude of the waveform to be above 110% of the current waveform average. This prevents detection of sharp ripples in

the signal that are due to noise or may be part of other features such as PVCs. The fourth condition checks whether the count of the beat peaks is the same as 44 samples ago. This prevents sharp ripples or spikes in the QRS peak or other segments of the waveform for that matter to be counted as beats when they are not beats. Lastly, there cannot have been a PVC occurrence in the last 70 samples. Therefore, there are a few check points that need to be passed, beyond amplitude detection, in order for a heart beat to be detected. Some problems with pure amplitude detection are that a wandering baseline value is not taken into account and the spike shape of the QRS complex is not identified, allowing a shallow, high DC offset waveform to be detected as a beat.

If all of these conditions are met, then the number of beat occurrences is incremented. The MATLAB script node code stores the times at which beats occurred. The heart rate is calculated from an average of the last 5 beats. The equation for the heart rate is given by:

$$\text{heartRate} = 60 * 4 / (\text{arrayBeats}(i) - \text{arrayBeats}(i-5));$$

If the heart rate is over 100 bpm, supraventricular tachycardia is identified. If the heart rate is under 50 bpm, sinus bradycardia is identified. And if the heart rate is between 50 and 100 bpm, a normal sinus rhythm is identified.

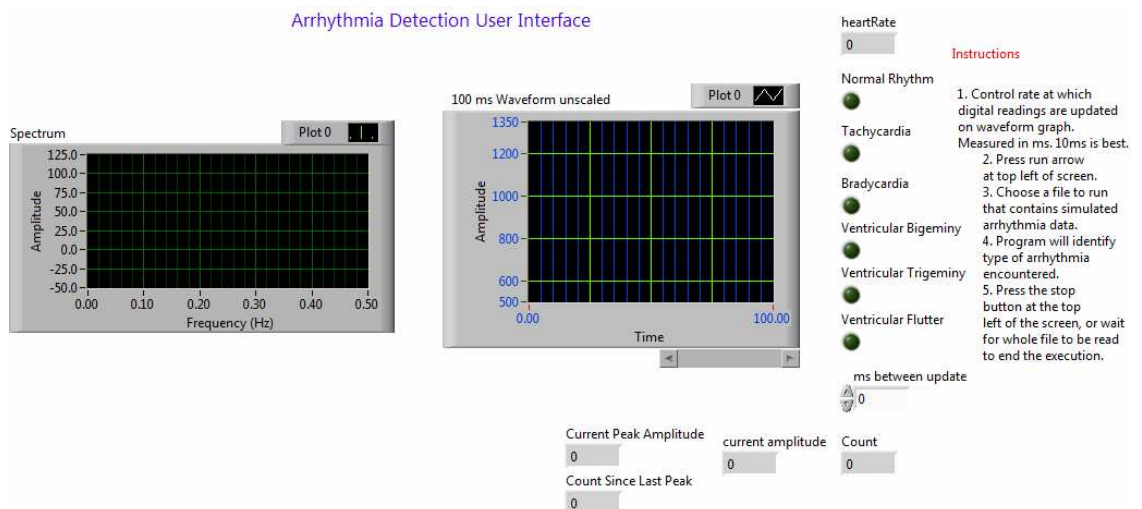


Figure 21. Arrhythmia detection GUI.

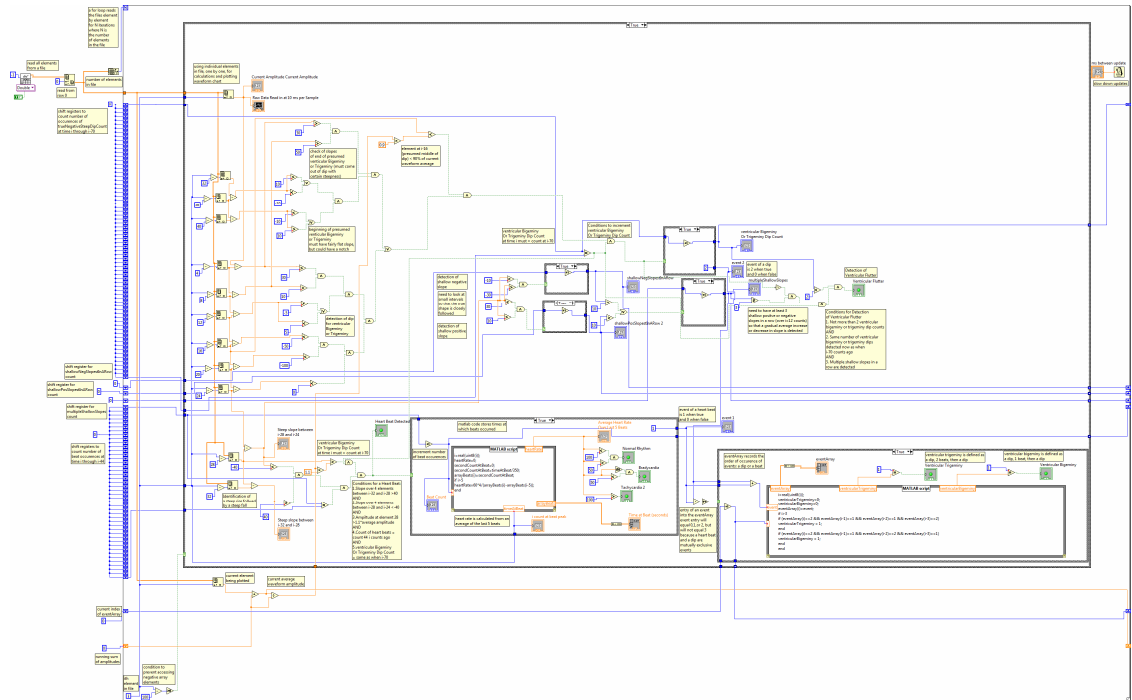


Figure 22. Block diagram of arrhythmia VI.

Conclusion and Results

All of the short term goals of the project were achieved. A real-time filtered and amplified heart wave signal from a person can be acquired. The computer displays the real-time signal on a GUI. The heart rate of the user or input file is calculated and continuously updated with sufficient accuracy and uses an algorithm that is adaptable for different users. Data can be read from different files and written to files for later analysis. The main objective of being able to detect a normal waveform, along with 5 types of arrhythmias has also been realized.

As stated previously, if a test subject had one of the identifiable arrhythmias analyzed by this system, the VIs created have the potential to identify the arrhythmia in a user by combining the arrhythmia VI with the circuit-interfaced VI. Since no test subjects were found with these conditions, saved test data was used for arrhythmia analysis.

A future development of this project would be to make it portable for use in exercise. The circuit used is small enough to be put into a portable device using only two electrodes at the wrists. The code used to identify the arrhythmias is universal enough such that it could be written in other languages, such as C, for use on a micro controller. However, LabVIEW was chosen for its user-friendly GUI, so it could still be used with a wireless interface to analyze the data at a later time on a PC or laptop.

The major advantage of using the methods used for arrhythmia detection in this project is that the parameters of the code can be easily changed and customized to an individual's unique heart waveform. An individual's waveform may change over the course of years, or the waveform may vary from person to person. By adjusting the parameters which determine whether an event has taken place in the subject's unique waveform, either a beat or particular feature of an arrhythmia, the code will detect fewer false positives.

The code detects the features of the arrhythmias with very good accuracy; however, it is still not correct all of the time. An improvement would be to increase the accuracy of the code. This is not necessarily difficult, but is time consuming. Also, certain problems, such as non-initialized heart rates LED values at the start of the running code can be changed.

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APPENDIX FOR CIRCUIT CALCULATIONS

Calculations

Head stage:

The AD620's data sheet specifies the resistor to be used with the chip by using the following formula. The gain, G, was set to 10 for this stage. R was found.

$$R=49.4K/(G-1)$$

High Pass Filters

A passive configuration was used where C was chosen and R was calculated based on a cutoff frequency of 0.1 Hz.

$$f_c=1/(2*\pi*R*C)$$

Non-Inverting Amplifier

Gain is set to be 100 so that the total gain of the circuit is now 1000. Solve for R1 and R2 using the following formula

$$G=1+R2/R1$$

Low Pass Filter

A second order active configuration is used with $f_c = 200$ Hz. Calculate R1 and R2 by choosing C1 and C2 first. (Let $C1=C2$). Chebyshev 3 dB parameters are chosen and the constants a1 and b1 are found in look up tables. The following are equations that correspond to a second order LPF.

$$R1= 1/(\pi*f_c*C*a1) \quad \text{and} \quad R2=a1/(4*\pi*f_c*C*b1)$$

Where a1 and b1 are 3 dB Chebyshev constants

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