PC BASED ECG MONITORING SYSTEM

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Keywords: PC ECG, Digital Electrocardiography.

Abstract

A project on creating and research the framework needed for building a PC based ECG monitoring system.

1 Introduction

1.1 About the Project

The aim of the project is to create the framework needed for building a low cost and simple PC based ECG monitoring system. We first started by researching on the basics of electrocardiogram (ECG) and its theory. We then explore how to obtain and process the signal from the human body. We also looked at how to digitise the signal so that it can be fed to the computer. Various ways of integrating the system with a computer and how the host computer can play a part in the system are also explored.

1.2 Basic ECG

Most of the modern 12 Lead ECG monitoring systems are based on the Einthoven's triangle, Wilson central terminal and Goldberger Augmented Leads.

The main idea is to monitor the electrical signals of the heart so that the physician can gauge whether the heart is functioning normally.

The electrical behaviour monitored by the ECG system is caused by the polarisation and the depolarisation of the heart muscles. However, when the electrical signals reach the skin, it would only be about 0.1mV to 0.5mV.

Of the 12 leads, the 1st 3 leads are derived from the Einthoven's triangle. Where,

Lead I = V left arm – V right arm (1)

Lead II = V left foot – V right arm (2)

Lead III = V left foot – V left arm (3)

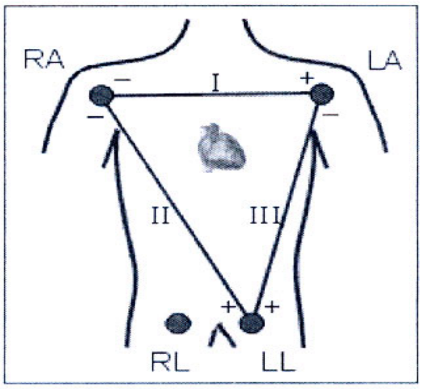

Figure 1 Einthoven's Triangle

The next 3 are the Goldberger's Augmented Leads. It is the measurement of the 3 corners of the Einthoven's triangle with reference to Wilson's central terminal (CT). They can be derived from the 1st 3 Leads.

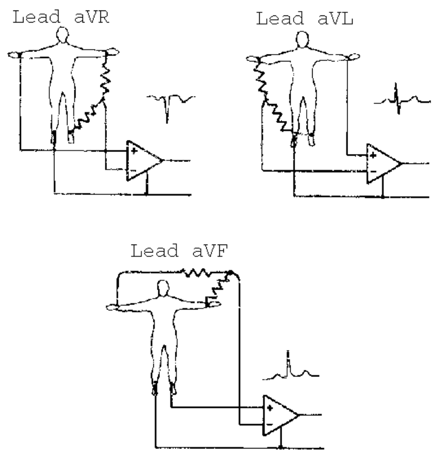

Figure2 aVR, aVL and aVF measurement

The next 6 will be the chest leads. These are points along the chest measure with reference to the CT.

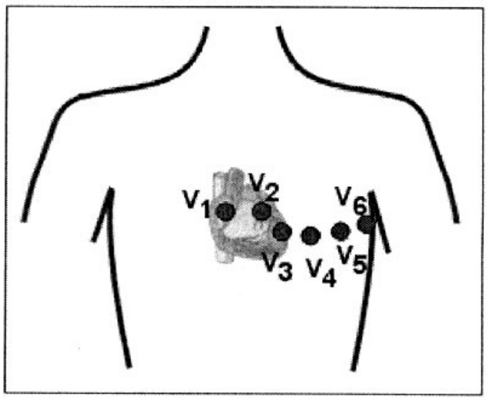

Figure3 Position of the chest leads

2 Overall Scope

In the past, a string galvanometer is used to obtain the ECG readings. As we have seen in 1.2, ECG readings can be obtained by taking the potential difference between two points. Therefore, we can use an Instrumentation Amplifier for our purposes. Since all 12 leads are obtained by simply taking the potential difference between different points of the body, our tests shall concentrate on obtaining the Lead I signal for our research of the ECG monitoring system framework.

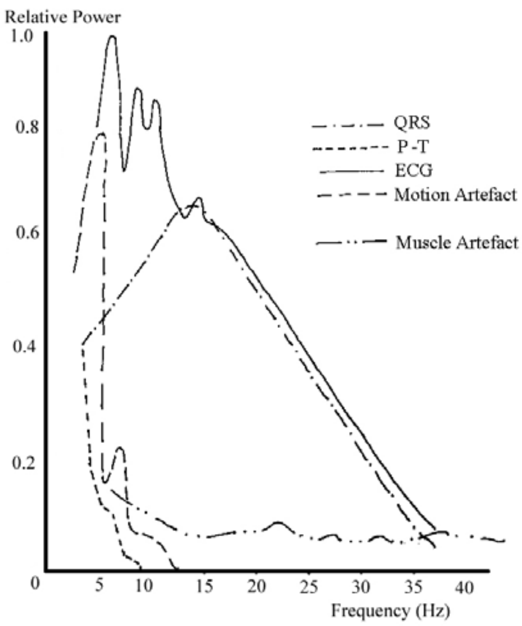

Figure4 Power spectra density of ECG

From figure 4, we can see that the signal we need to obtain is from around 2.5Hz to 40Hz. The British Standard 60601 also states that an ECG device should have frequency response from 0.67Hz to 40Hz. There is no point measuring signals beyond 40Hz as the ECG signals get overwhelmed by the muscle artefacts. As the signal is in terms of millivolts, high CMRR is needed to diminish the size of artefact introduced by electrode or skin resistance. Therefore in our selection of components, we must select those which have a flat frequency response in the range of 0.67Hz to 40Hz and high CMRR.

We also looked at the design of the probes and electrodes itself. As we are detecting very low voltages, special considerations have to be taken.

As we need not consider the signals above 40Hz, this project also looks at the various filters needed to remove noise and clean up the signal.

Lastly, for the analogue portion of the project, we touched on how else the signal has to be processed for the ADC.

For the digital portion, we looked at implementing the microcontroller, the ADC and how to interface the data with the PC.

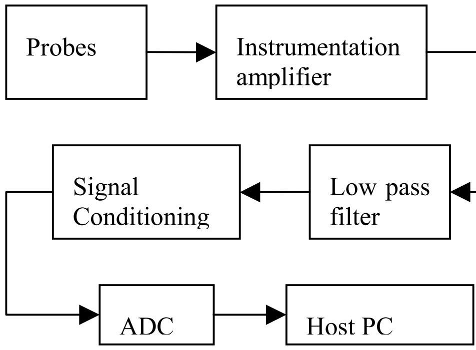

Figure5 Basic Framework of the PC based ECG monitoring system

3 Probes and Electrodes

As the signal we have to pick up is in terms of millivolts, it is important that our electrodes have good contact with the skin, is sensitive and offers low resistance. For our case, we look towards readily available commercial products to speed up our development.

As the signal the electrodes picked up is in millivolts, the probes have to be shielded from the 50 Hz electrical noise and other noise present in the air. We purchased shielded cables for this purpose.

Figure6 Disposable Electrodes from MSB with clip

For our 1st prototype, we use crocodile clips to connect to the electrodes. We discovered that the crocodile clips are not really secure and there is no full contact between the clips and the electrodes connection tabs due to the jagged clips. We then proceed to file down the teeth of the crocodile clips to improve contact. We discovered that by doing so, the signal strength received by the IA increased. This is most probably due to that the signal we are working with is so small, any marginal decrease in resistance will result in a stronger signal. Therefore, we proceed to purchase a commercially available clip that is designed for the sole purpose of connecting to ECG electrodes.

The new clips offered better contacts with the electrodes tabs and gave even better stronger signals. In the early part of the project, we face a lot of problem with noise and artefacts due to loose connections between the probe and our circuit. To avoid such problems, we also switched to a 15 pin D type connector to ensure that the connection is secure.

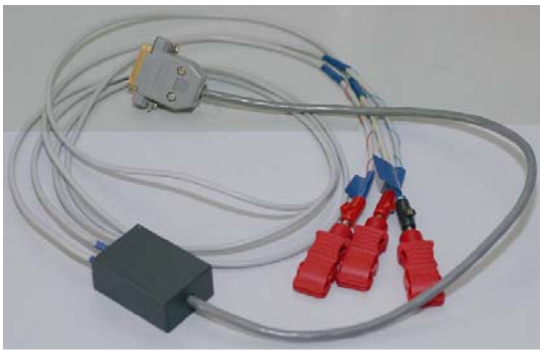

Figure7 Picture of probes and the new clips

4 Instrumentation Amplifier

Since the basic idea of ECG is to monitor the potential difference between 2 points on the body, we can use an instrumentation amplifier (IA) for our purposes.

Initially we tried a common IA to see if it can suit our purpose. However, the frequency response from 0 Hz to 20Hz is not flat. We observed that from 20 Hz to 0 Hz, the signal starts to attenuate. We noticed that a lot of operational amplifier datasheets do not have material on the frequency response of the amplifiers at such low frequencies or they have bad low frequency response. We had to look through a whole range of amplifiers to find one that suit our needs.

In the end, we settled on an IA that is specially designed for the purpose of amplifying ECG signals. It has high CMRR and the RF resistor of the IA is laser trimmed and built-in, resulting in that we only need to tamper with the RG and accurate gain control.

A connection from RG to the right leg is established to increase the CMRR and create a common ground between the body and the circuit. This setup is recommended by the manufacturer of the IA.

Initially, we used a gain of 100 for the IA. However, with a gain of 100, we discovered that the 50 Hz noise is so strong that it is almost impossible to filter it off. The 50 Hz noise is about 40% of the peak amplitude of the ECG signal. Furthermore, amplifying the signal now would also mean amplifying the 50 Hz electrical power noise. We decide to try a higher gain so that the ECG signal would be a lot stronger than the 50 Hz signal.

We proceeded to experiment with a gain of 1000, and found that the ECG signal did become stronger than the accompanying 50 Hz noise which makes our processing of the signal easier in later stages.

We could further increase the gain for the IA to minimise the electrical noise in comparison with our signal but doing so would be decreasing the value of RG. The value of RG for a gain of 1000 is already quite small. Increasing the gain of the IA would mean that the value of RG would be even less, which may make the gain of the IA unpredictable or uncontrollable as RG would be affected by wire resistance.

Thus we set our gain at 1000 for our IA.

5 Low Pass Filter

The signal required is from around 0.67 Hz to 40 Hz, we need to filter out the unnecessary noise. As we wanted to retain the integrity of the shape of a signal, a filter design that has a maximally flat amplitude response is needed. Because of the fore mentioned requirements, a Butterworth filter designed using the

Sallen-Key circuit topology is selected as our low pass filter.

We placed our cut off frequency at 60 Hz as we fear that placing it too close to the frequencies we want to sample may cause those frequencies to be attenuated. We started out using a 2nd order low pass filter. Unfortunately, the low pass filter did not attenuate the noise to the extent we wanted and also attenuated part of the signal we want to preserve.

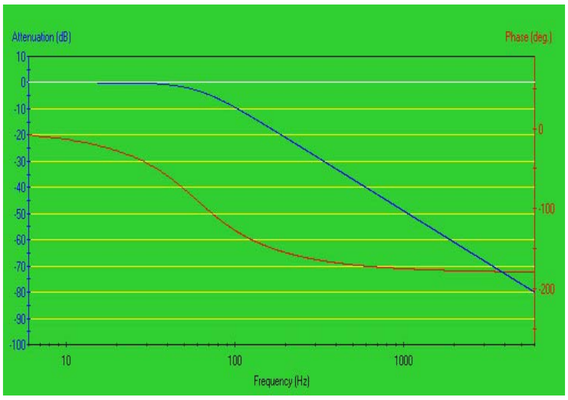

Figure8 Frequency response of the 2nd order low pass filter

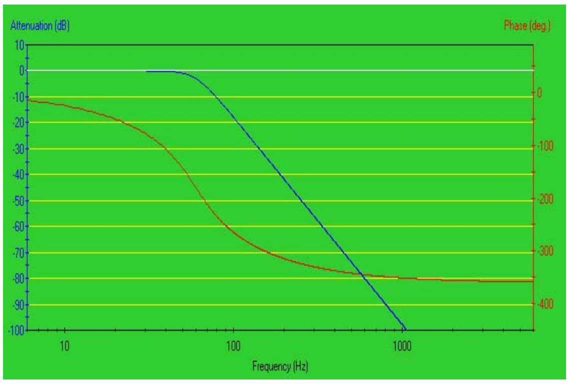

Figure9 Frequency response of the 4th order low pass filter

We proceed to test out the 4th order filter. The 4th order filter is able to attenuate the noise and yet left the signal desired untouched.

6 Notch Filter

Despite the low pass filter mentioned earlier, we still suffer from a very strong 50 Hz electrical noise. Our only course of action here is to apply a notch filter to attenuate the 50 Hz noise.

For the project several types of notch filters were tested. They are the boot strapped Twin-T notch filter, “normal” the Twin-T notch filter and the ZXF36L01 variable Q filter.

The behaviour of the notch filters are determined by the Q factor of the filter. Q factor determines the width of the notch. The higher the Q the more narrow the notch; however as the Q increase the depth of the notch decrease also. Therefore our aim here is to obtain a notch that is as narrow and as deep as possible.

In our test with the boot strapped Twin-T notch filter, we discover that the notch filter cannot go into very high Q and is quite unstable.

The ZXF36L01 managed a Q of 50, but unfortunately, it has the same problem as most operational amplifiers we have seen, a very bad low frequency response. In our test, the ZXF36L01 starts to attenuate from 20Hz to 0 Hz, practically blocking out what we wanted to retain.

This leaves us the “normal” Twin-T notch filter. Using operational amplifiers, we managed to create a Twin-T notch filter with an adjustable Q of up to 50. We did not fix our Q variable as we wanted to be able to adjust the amount of filtering needed for any given situation.

7 Signal Conditioning for the ADC

The signal we obtained from the IA needs to be digitised so that it can be sent to the PC for processing. However, the signal is not ready for the ADC after coming off the notch filters.

7.1 DC offset removal

One problem that prevents the signal to be fed straight into the ADC is the DC offset. We discovered that the DC offset created varies each time.

The 1st approach we took is to develop a system that detects the amount of DC offset and fed this signal into the negative input of an IA, while the signal itself goes to the positive input of the IA. The resultant signal that comes out of the IA is the signal without the DC offset.

To achieve this, we fed the signal into a simple RC circuit, which is essentially a low pass filter, to extract the DC component of the signal before going into the negative input of the IA.

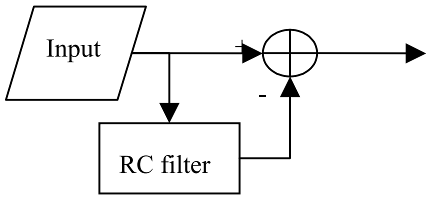

Figure 10 Block diagram of prototype DC offset remover

However, if the R is large, there will be a drop in potential of the DC component. But to keep R small, C has to be large. A large C would result in a very slow response. Our experiments proved that having a large C creates a very slow response as sometimes the circuit needs to react fast as the DC level changes when the patient moves around.

We then decide to take a 2nd approach by treating the DC component as a waveform and filter it out. As the DC component is a flat level voltage level, we can say that it is a waveform of 0 Hz. Therefore, if we apply a high pass filter, we can remove the DC component.

As the signal ranges from 0.67Hz onwards, we feel that a simple RC high pass filter circuit does not attenuate fast enough for the 0 Hz frequency to be negated. Our experiments of an RC high pass filter at 0.25 Hz proved this. We implemented a 2nd order Butterworth high pass filter using Sallen-Key circuit topology and tested it. The DC component is effectively removed and the response is much faster than the previous design using the IA.

7.2 2nd Stage Amplifier and DC offset adjustment

Even after an amplification of 1000 at the IA, the signal is only around 400mV in amplitude. Also, the signal has a negative component; therefore we must give the signal a DC component so that the whole signal is within the range of the ADC and amplify the signal to make full use of the ADC's voltage range.

As we do not have data on what is the maximum possible amplitude an ECG signal can be, we have to leave some leeway on the top and the bottom of the signal. We decided that a gain of 4 would be able to do just that.

Amplifying the signal turns out not to be a simple task of using an operational amplifier as all the operational amplifiers we tried have bad low frequency response, the 0 Hz – 20 Hz gets attenuated, we therefore used the IA.

Tying the positive input of the IA to the signal and negative input to the ground, we have created an amplifier for the signal. The signal, however, still has no DC component to push it into the ADC's range. To achieve this, we made full use of the IA's negative input by feeding in a – 0.5V DC signal. This will create an output signal = (input signal + 0.5V) x 4. In this way, the signal will have a DC offset of 2V.

8 Analogue to Digital Conversion

The ADC built into the microcontroller is used for this project. After looking at other references and products, we agreed that an 8-bit resolution is enough for our purpose. What we had to find out is the sampling rate needed.

In Figure 18 below shows the graph of a typical Lead I signal. The letters PQRST were given to each distinct point of the wave for identification.

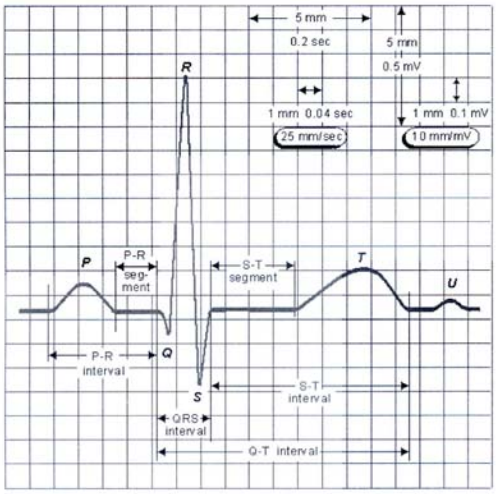

Figure 11 PQRST graph

As we can see, the ECG is not a regular waveform; therefore the sampling rate must be high enough to be able to allow us to recreate the waveform. In order to do that, we shall concentrate on the QRS portion of the waveform. The QRS is not only the narrowest portion but the most important. The heart beats per minute is based on how many QRS spikes are detected per minute. In order to have an algorithm to extract the heart beats per minute from the ECG graph, we need to have an accurate QRS read out.

Assuming what we have in Figure 11 is a typical ECG waveform; the QRS is only 1/18 of the whole waveform. If the maximum heartbeat rate to be detected is 190 beats per minute, then the QRS would be

$$60 \div 190 \times 1/18 = 17.54ms \qquad (4)$$

Taking 3 times the Nyquist rate, we need

$$\frac{1}{17.54ms \div 6} = 342Hz \qquad (5)$$

Therefore our sampling rate needs to be at least 350Hz.

The microcontroller which we intend to use ultimately can sample at 6 KHz, which far exceeds what we need.

9 Microprocessor

For our project, we must choose a microprocessor that has most functionality and devices onboard so that it does not complicated the system design or increases its size. The microprocessor must also be powerful enough to take on some data processing to lessen the host PC's load. If we are able to transfer the bulk of the routine processing to the microprocessor, we may be able to lessen the load on the host PC such that it may even be able to be connected and monitor several such ECG monitoring systems concurrently. Also important is that the microprocessor must be easy to develop and a wide range of tools are available for us to carry out our development.

The microcontroller we selected for our project is an 8 bit microcontroller with a 10 bit ADC. As it is a Flash reprogrammable device, development would be a lot easier.

10 Possible Future Developments

Although this project is to research and create a framework for a PC based ECG monitoring system, we have already researched on and planned what is needed for the future.

10.1 Full 12 Lead Monitoring

Our basic ECG system can only monitors the 3 basic leads, but there are a total 12 leads to monitor.

For the 1st 6 Leads, which is Lead I to III, aVR, aVL and aVF, can all be derived from Lead I and III. We only need to monitor Lead I and Lead III to generate the other 4 Leads. The other 6 chest Leads cannot be derived mathematically from each other, we have to create a connection to monitor them.

In total, 4 physical connections are needed, but in order to get ECG readings, we need to measure potential difference between certain pairs of connections. There are a couple of approaches to do this, and they all require multiplexing to a certain degree.

10.2 QRS detection

One important feature needed in the ECG device is QRS detection, or in layman terms, heartbeat rate counting. We can implement the QRS detection using software.

Modern personal computers are getting more powerful and cheaper. We can use the microprocessor's power to implement the QRS detection function. This way, we can save costs incurred in using hardware implemented on the ECG device to do the same work.

10.3 Using digital filters

The filters used here are not ideal and in some cases allow some of the noise in or worse, attenuated the signal we wanted to retain. The 50 Hz noise posed the most problems. Some ADCs in the market has digital filters incorporated into them. Digital filters can also remove noise injected during the A to D conversion process.

10.4 Waveform analysis by PC

Currently our software can only extract and record the data from the ECG device. Given the power of modern PCs, there are much potential for what it can do.

Tasks like waveform matching and waveform behaviour detection can allow the software to learn and identify anomalies in the ECG and other illness.

12 Conclusion

From this project, we have gained a lot of experience regarding instrumentation and operation amplifiers, ADCs and their applications. Although we have only built the basic framework of an ECG device and have it connected to the host PC using a serial bus. We have learned enough to develop it into a full blown system. There is much yet to improve and work on in this field.

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