ACQUISITION AND MONITORING OF MULTICHANNEL ECG

A Thesis Project

Submitted to the Department of Electrical and Electronic Engineering

Of

BRAC University

by

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In Partial Fulfillment of the

Requirements for the Degree

of

Bachelor of Science in Electrical and Electronic Engineering

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DECLARATION

We hereby declare that this thesis is based on the results found by ourselves. Materials of work found by other researcher are mentioned by reference. This thesis, neither in whole nor in part, has been previously submitted for any degree.

**Signature of
Supervisor**

11/8/10

Signature

MD. SIBBIR Awan

MD. MASHRUR SHAFIQUE KHAN

MD. ASHRAF-UL- KARIM

1.INTRODUCTION

The heart's strong pumping action is driven by powerful waves of electrical activity in which the muscle fibres contract and relax in an orchestrated sequence. These waves cause weak currents to flow in the body, changing the relative electric potential between different points on the skin. An electrocardiogram is a biophysical instrumentation device that is used to view/record the electrical activity of the heart for various diagnostic purposes. The electrocardiogram (or ECG) has been used extensively in medicine sine its invention in the early 1900's, and has since proven to be invaluable in various diagnostic applications, such as the detection of irregular heartbeat patterns (i.e. fibrillation or arrhythmia), heart murmurs (or other abnormal heart sounds), tissue/structural damage (such as valve malfunction) and coronary artery blockage. Other applications of the ECG are very effective in areas of sports medicine, or sports therapy, in tracking the heartbeat through various levels of physical activity to assist the patient in attaining a desired, optimum heart rate.
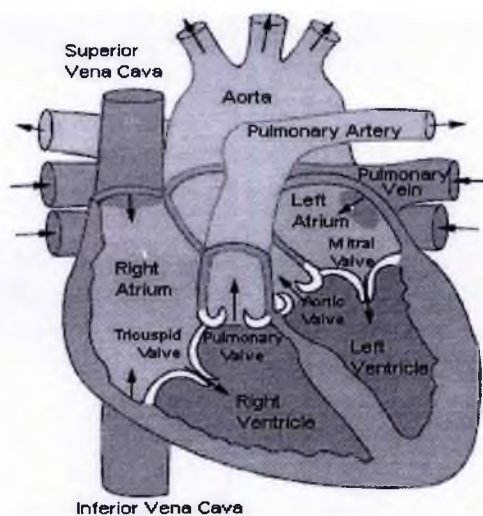

Figure 1.0a.(Heart

Therefore, while the concept of an ECG is not a novel one, the attraction of this project lay in the challenge to build a simple, compact, operational medical device at a low cost. The basic design theory is as follows: -

➤ The electrical activity of the heart is detected using electrodes placed on the surface of the chest cavity. These electrodes act as bio-transducers to convert the signal from its existing form in the body (ionic) into electrical current in the wires.

➤ The generated signal is put through an amplifier to allow for observations, measurements, and recordings to be made. This stage is extremely important, as the cardiac signal is very small, i.e. on the order of mili-volts, thus a large amplification is necessary for any use to be made of the signal.

➤ The amplified signal is then sent to the atmega8 for Analogue-to-Digital conversion, signal manipulation, calculation of beats per minute (displayed using 3, 7-segment LED displays), data logging (RS232communications) and analogue signal output (DAC) for a visual display of the ECG. Note that an oscilloscope can be used to provide a visual output.

This report will detail the development and implementation of a low-cost microcontroller base ECG monitor.

2.OBJECTIVE

The ECG is the most widely used and invaluable clinical tool for the detection and diagnosis of a broad range of cardiac conditions, as well a technique that has contributed to the understanding and treatment of virtually every type of heart disease. More than 7 million ECG's are performed in the Bangladesh each year, making ECG the most commonly performed cardiovascular diagnostic test. ECG is a very important and effective heart monitoring system. Heart patients can get better

medical care if they can keep one handy. Unfortunately, commercial ECG machine are too costly to use at home for a poor country like Bangladesh. Our area of interest is to implement such device which can perform like conventional ECG machine and by which we will be able to locate the exact abnormality of human heart, but with low price so that cardiac patients can keep one and check their cardiac report whenever they want for maintaining proper diet suggested by cardiologist.

3. MESUREMENT AND SETUP

3.1 12-Lead Systems:

As being pointed out on earlier, the electrocardiogram (ECG) is the recording on the body surface of the electrical activity generated by cardiac muscle, a potential difference, that's why it can be measured by placing two electrodes on the surface of the body. Because of the spatial dependence of the electric field of the heart, different pairs of electrodes at different locations generally yield different voltages. Thus it is important to have certain standard for clinical evaluation of the ECG (Webster, 1998).

The term *'Lead'* is defined as a spatial arrangement of a pair of electrodes, or an equivalent pair that is combination of several electrodes through a resistive network. One lead is labelled '+' and the other '-'. The electrode placement defines the recording direction of the lead, which is called the lead axis or angle. The axis is determined by the direction when going from the negative to positive electrode. The ECG recorder computes the difference (magnitude) between the positive and negative electrodes. A good mathematical tool for representing the measurement of a lead is the vector. A vector is defined as an arrow whose head points in the positive direction. The length of the arrow is proportional to the

magnitude of the lead (Richard and William, 2002).

The most commonly used clinical ECG-system is the 12-lead ECG (10 electrodes) system, which consists of the three standard limb leads (I, II, III), the three augmented referenced limb leads (aVR, aVL, aV_F) and the six Wilson terminals referenced chest leads (V_1 V_2, V_3) V_4, V_5, V_6) (Joseph, 2000). (Figure 3.1)
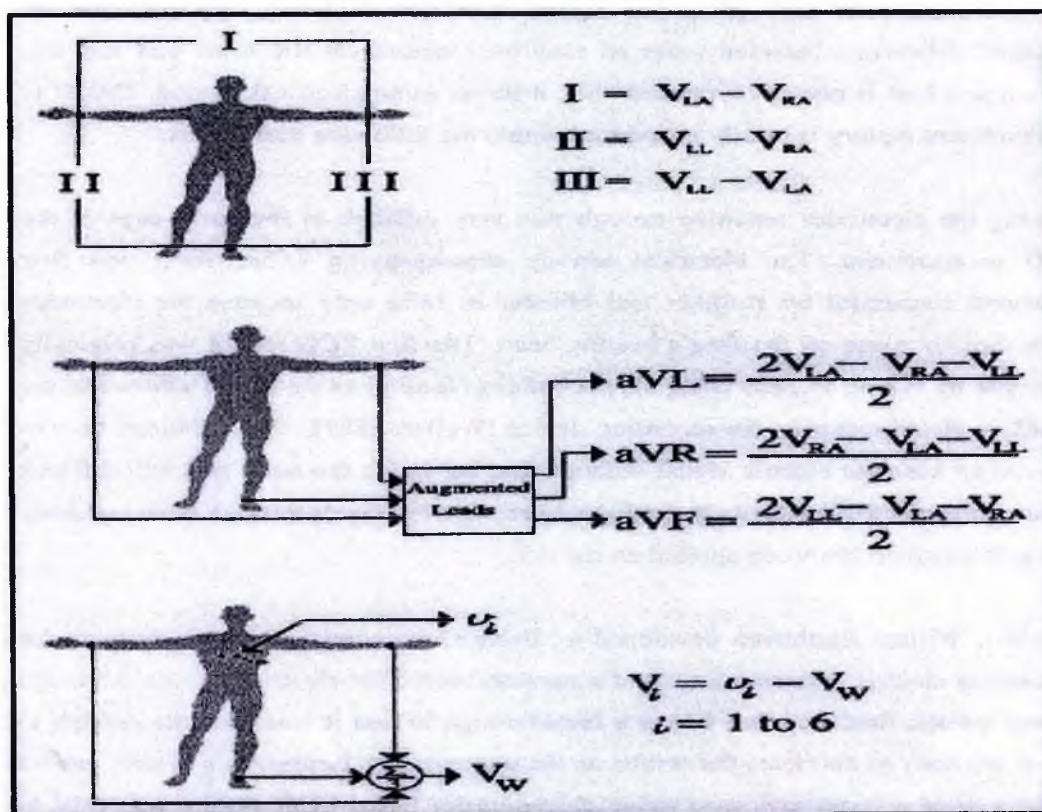

Figure 3.1 The standard 12-lead ECG (Joseph, 2000)

Where, I = the voltage of Lead I

II = the voltage of Lead II

III = the voltage of Lead III

V_{LA} = the potential at the left arm

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V_{RA} = the potential at the right arm

V_{LL} = the potential at the left foot

aVR= the augmented voltage of Lead aVR

aVL= the augmented voltage of Lead aVL

aVF= the augmented voltage of Lead aVF

Vi= the voltage of the six chest leads

Vi= the potential at the six chest leads

V_w= the potential at the Wilson central terminal

3.2 Bipolar Leads (Einthoven Triangle):

The three standard limb leads are bipolar leads, which mean they measure the potential difference between pairs of electrodes placed on the arms and left leg. Although a lead is placed on the right leg, it serves as an electrical ground. The ECG measurement history is briefly introduced within the following paragraphs. Making the electrodes sensitive enough was very difficult in the early days of the ECG measurement. The electrical activity accompanying a heart-beat was first measured successful by Kolliker and Mueller in 1856 only because the electrodes were directly place on the frog's beating heart. The first ECG record was originally observed by Waller in 1889 using his pet bulldog (Jimmy) as the signal source and the capillary electrometer as the recording device (Waller, 1889). Nevertheless, how to acquire an accurate electric signal without entering inside the body was still difficult because the electrical activity is getting weaker when it travels through bone and body tissue to reach an electrode applied on the skin.

In 1901, Willem Einthoven developed a 'String Galvanometer' - an instrument for measuring electric current -that could accurately record the electric activity. Although it was not the first recorder; it was a breakthrough in that it was accurate

enough to allow anybody to duplicate the results on the same patient. Einthoven's system proved to be a great success and soon string galvanometer based ECG system was used in clinical practice worldwide. Since that time the ECG has become a very powerful tool in diagnosing disorders of the heart.

Einthoven stated that the heart is in the centre of an equilateral triangle whose aATMEGAes are the right arm, left arm and left leg. The lead vectors therefore also form an equilateral triangle called *Einthoven's Triangle*. The Einthoven's Triangle is defined as a configuration of three standard limb leads (Figure 3.2). Lead I goes from right to left arm; lead II from right arm to left leg; lead III from left arm to left leg.
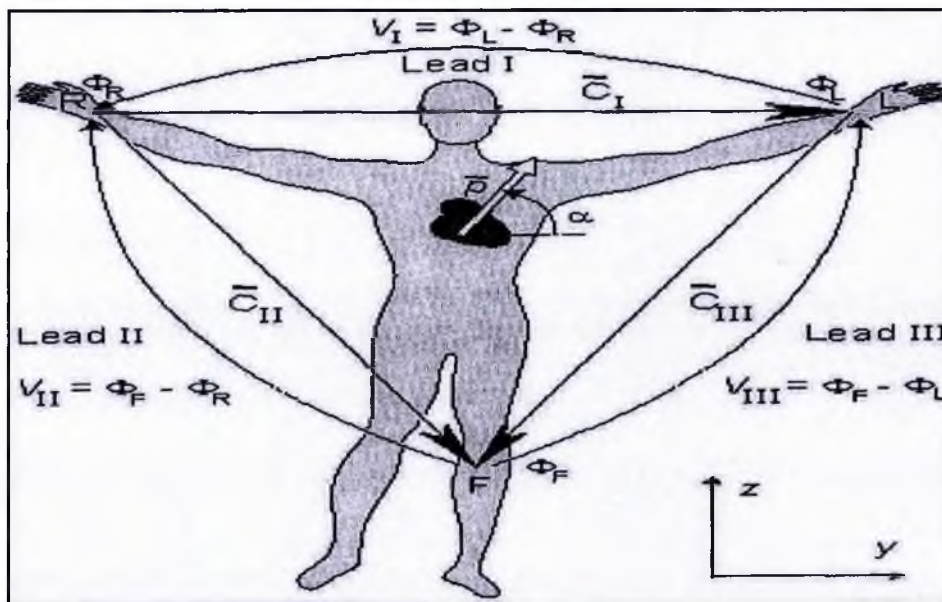

Figure 3.2 Einthoven limb leads and Einthoven triangle (Jaakko and Robert, 1995)

$$I: V_1 = \Phi_L - \Phi_R \tag{1}$$

$$II: V_{II} = \Phi_F - \Phi_R \tag{2}$$

$$III: VIII - \Phi_F . \Phi_L \tag{3}$$

Where V_I = the voltage of Lead I

K_{II} = the voltage of Lead II

F_{III} = the voltage of Lead III

ϕ_L = potential at the left arm

ϕ_R = potential at the right arm

ϕ_F = potential at the left foot

According to Kirchhoff's law these lead voltages have the following relationship (Einthoven's Law):

$$V_I + V_{III} = V_{II} \tag{4}$$

Therefore, if any two leads are known at a given time, the third can be determined

mathematically (Jaakko and Robert, 1995).

3.3 Unipolar Leads:

Unlike the three standard limb leads, the three augmented referenced limb leads and the six Wilson terminals referenced chest leads are unipolar leads, which means they record the voltage between a single 'exploratory electrode' placed on the body and an electrode that is built into the electrocardiograph and maintained at zero potential (ground). The major difference from the bipolar leads is therefore that a negative electrode of the unipolar leads is imaginary but not a real physical electrode. The three augmented referenced limb leads are derived from the same three electrodes of the three standard limb leads (Left arm, right arm, and left leg). Hence, there are together 10 electrodes required in the 12-lead ECG system; four limb electrodes and six chest electrodes which produce six limb leads (I, II, III, aVR, aVF, and aVL) and six chest leads (VI, V2, V3, V4, V5, and V6).

The concept of a 'unipolar' recording is first suggested by F.N. Wilson. He suggested that unipolar potentials should be measured with respect to the Wilson

Central Terminal (WCT) as a reference point. The WCT, which represents the average of the limb potentials, is formed by interconnecting two arm and left leg electrode through a 5k resistance (Figure 3.3). A unipolar chest lead is therefore created by using the WCT as the negative electrode, and an exploring electrode on the chest as the positive electrode.
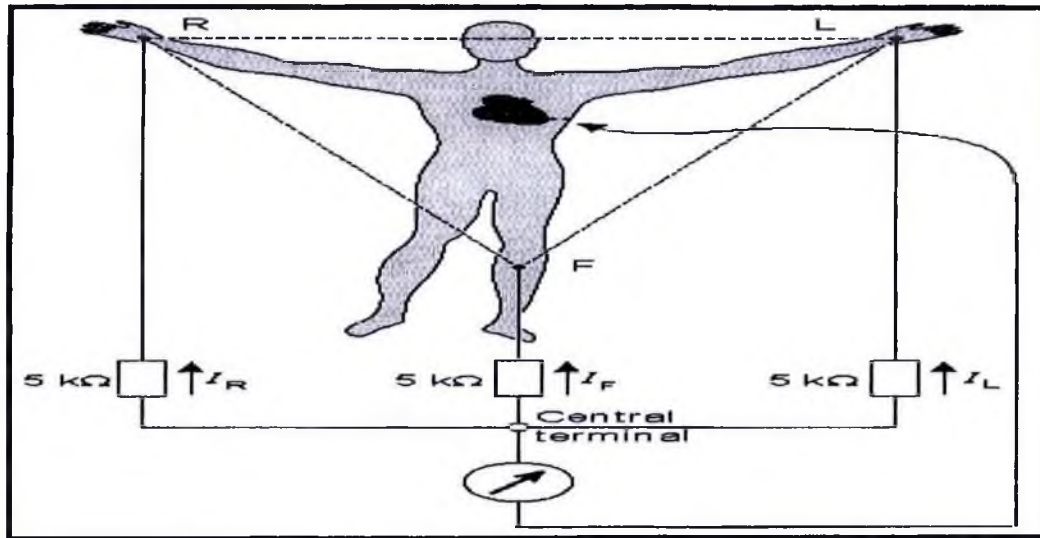

Figure 3.3 Wilson Central Terminal (WCT) (Jaakko and Robert, 1995)

Because there is no lead current in an ideal voltmeter, the total current into the central terminal from the electrodes must add to zero to satisfy the conservation of current. Accordingly, we acquire that:

$$I_R + I_L + I_F = \frac{\Phi_{WCT} - \Phi_R}{5000} + \frac{\Phi_{WCT} - \Phi_L}{5000} + \frac{\Phi_{WCT} - \Phi_F}{5000} \tag{5}$$

From which it follows that:

$$\Phi_{WCT} = \frac{\Phi_R + \Phi_L + \Phi_F}{3} \tag{6}$$

8

The location of the chest electrodes is given by Figure 3.4, which are labelled one through six, starting from the midline position.
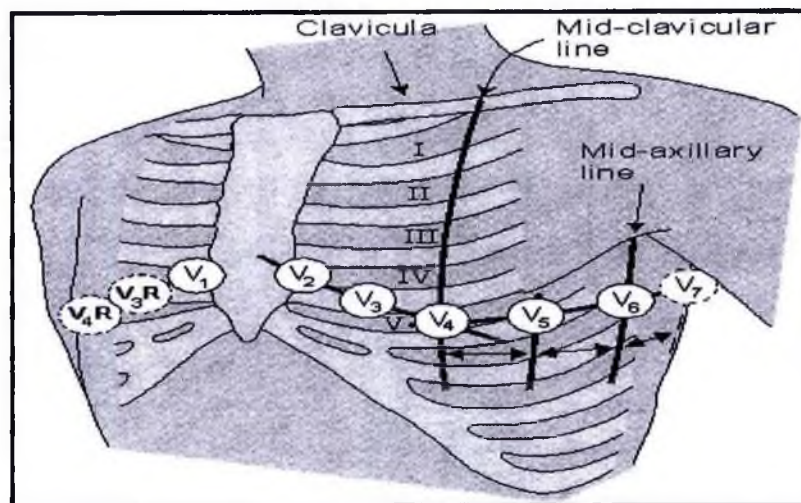

Figure 3. 4 Location of the chest electrodes (Jaakko and Robert, 1995)

Where VI: Fourth intercostal space to the right of the sternum

V2: Fourth intercostal space to the left of the sternum

V3: Directly between leads V2 and V4

V4: Fifth intercostal space at mid-clavicular line

V5: Level with V4 at left anterior axillary line

V6: Level with V5 at left mid-axillary line

In the three augmented referenced limb leads, however, the reference point is not using the WCT, because the recorded amplitude is extremely low. For instance, the measurement from the left leg (foot) gives:

$$V_F = \Phi_F - \Phi_{WCT} \frac{2\Phi_F + \Phi_R + \Phi_L}{3} \tag{7}$$

In 1942, Goldberger E. suggested that the signals can be amplified or augmented by omitting the 5k resistance from the WCT. Thus, lead aVR, aVL, and aVF are called augmented limb leads. The augmented aVF lead is therefore

created by using the left foot as its positive electrode, and the average of the right arm and left arm as its negative electrode. The augmented aVL and aVR are created in the same way (Figure 3.5).
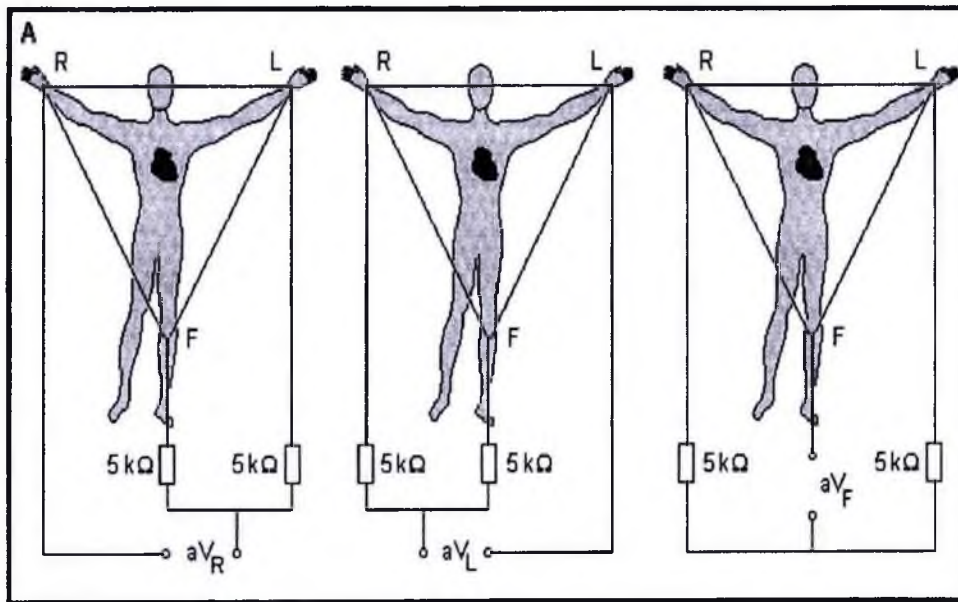

Figure 3.5: Goldberger augmented leads (Jaakko and Robert, 1995)

As an example, the equation for the augmented lead aVF is:

$$V_F = \Phi_F - \Phi_{aVF} = \Phi_F - \frac{\Phi_R + \Phi_L}{2} = \frac{2\Phi_F - \Phi_R - \Phi_L}{2} \qquad (8)$$

A simple comparison of Equation (3.7) with Equation (3.8) shows the augmented signal to be 50% larger than the signal with the Wilson central terminal chosen as reference (Jaakko and Robert, 1995).

3.4 Electrodes:

Electrode is not the same concept as lead An electrode is a physical patch which connects to the patient; meanwhile, a lead is a specific vector in which voltage is measured. ECG electrodes are used for sensing bioelectric potential (electrical

activity) as caused by cardiac muscle. The electrical activity can be seen as a constant DC electric field or a constant flux of charge-carrying particles or current. The electrodes work as transducers converting ionic current flow from the body into the electron flow of the metallic wire, and consequentially ECG

signal can be diagnosed after amplified and processed. A high ionic concentration gel is therefore normally used in the skin-electrode interface to increase conductivity. The choice of material is important as well because the small electrical charge at the skin-electrode interface vary with different electrode materials. The best currently available are gold, platinum, stainless steel, while the most common used is the silver-silver chloride electrodes.

4. ECG INSTRUMENTATION

4.1 ECG requirement:

Because the ECG machine must deals with extremely weak electric signals, and the noise from everywhere will intensively affect the acquisition of ECG signal, there are some particular requirements for ECG instrumentation. The Committee on Electrocardiography of the American Heart Association and lots of other organization have made recommendations for the standardization of the ECG requirement. Some of these recommendations for the ECG instrument design used in this thesis are as follows.

1.   The instrument should have capability to sense low amplitude signals should be in the range of 0.05 - lOmV, as the normal ECG amplitude is ±2 mV.

2.   The input impedance between an electrode terminal and ground should not less than 5 MQ at 10 Hz, as the ECG signal has high source impedance.

3. The instrument's freq response should have a band width from O.lHz to 150Hz.

4. The instrument should not allow currents (leakage currents) greater than 10 uA to flow through the patient.

5. The isolation methods should be used to keep the patient from being part of the AC circuit in the case of a patient-to-power-line fault.

6. The instrument is recommended to have a high common-mode rejection ratio (CMRR) on the preamplifier stage.
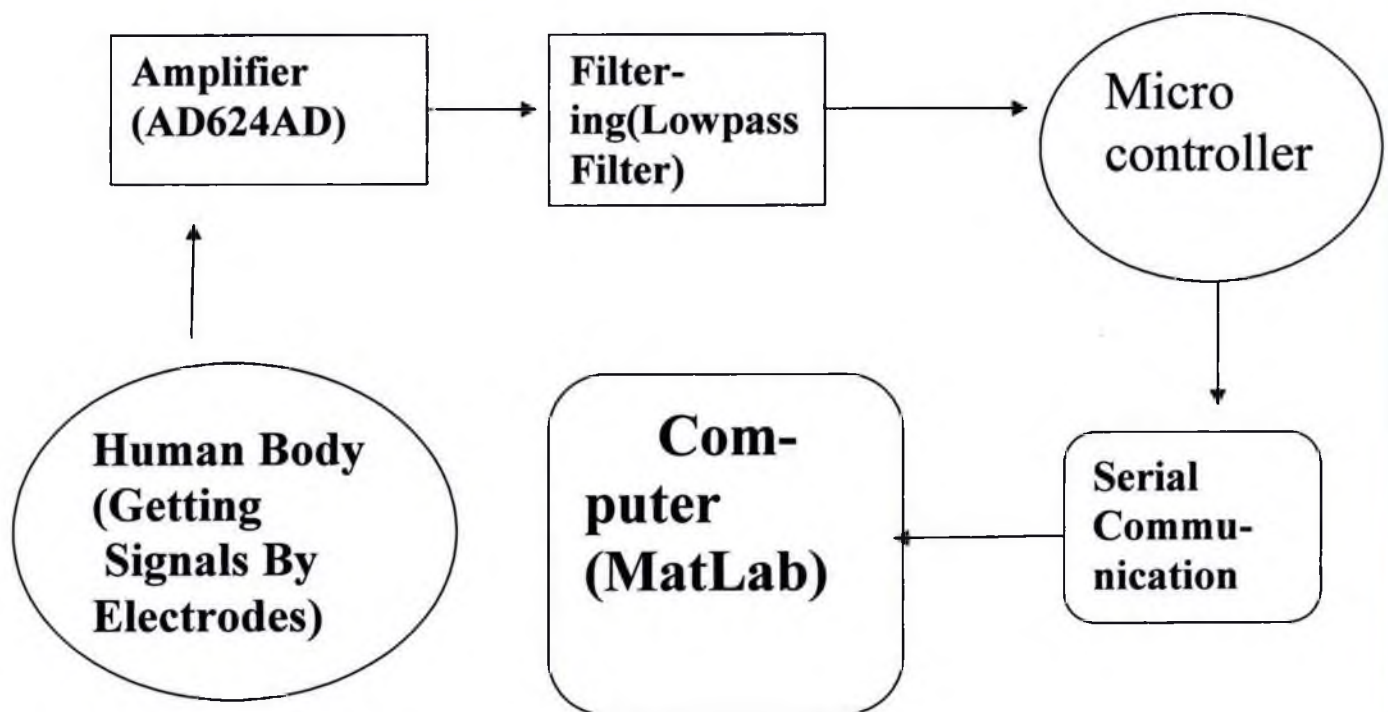

Figure: Block Diagram Of The Project

4.2 ECG Instrument Design:

The ECG signals are characterized by high source impedances, very small signal voltage, significant interference and noise, and a modest frequency range. All

these characters determine the ECG measurement machines are different from the industrial measurement machines. ECG instrument should do such jobs as follow: converting the biopotential into electric signal; amplifying the small electric signal to a level at which it can appropriately being diagnosed; suppressing the interferences, especially the AC main interference; processing the ECG signal for further used; displaying, recording, or storing this physiological quantity; the last but not the least is it must be electrically safe for operate and patient.

4.3 Bio-potential Electrode:

The first stage is the bio-potential electrode, which converts the bio-potential into electric signal. The common used silver-silver chloride electrodes are attached on the human body with electrolyte jelly. On the skin side of the electrode interface, a drift of ions is formed as the ECG signal spreads throughout the body. On the metal side of the electrode, a little electric current is formed as the metal ions ionizing or unionizing to maintain an electronic potential equilibrium. This reaction can be represented by the following equation.

$$Ag \Leftrightarrow Ag^+ + e^- \tag{9}$$

The result is a voltage drop across the electrode-electrolyte interface that varies depending on the electrical activity on the skin. This voltage drop causes electrons and anions to move across the interface into the skin. The differential potential between a pair of electrode (a lead) is therefore formed.

4.4 Instrumentation Amplifier:

The second stage is the instrumentation amplifier, which amplifies the small difference between the signals and suppresses the common-mode interferences at the two electrodes with a common-mode rejection function.

The heart's strong pumping action is driven by powerful waves of electrical activity in which the muscle fibres contract and relax in an orchestrated sequence. These waves cause weak currents to flow in the body, changing the relative electric potential between different points on the skin by about 1mV. The signals can change sharply in as little as one fiftieth of a second. So boosting this signal to an easily measured one-volt level requires an amplifier with a gain of about 1,000 and a frequency response of at least 50 hertz. At first it appears that an operational amplifier could be used. But two vexing subtleties make most op-amps unsuitable. First, when two electrodes are placed at widely separated locations on the skin, the epidermis acts like a crude battery, generating a continuously shifting potential difference that can exceed 2V. The cardiac signal is small in comparison. Second, the body and the wires in the device make good radio antennas, which readily ATMEGA up the 50Hz hum that emanates from every power cable connected to the mains supply. This adds a sinusoidal voltage that further swamps the tiny pulse from the heart and because these oscillations lie so close to the frequency range needed to rack the heart's action, this unwanted signal is difficult to filter out. Both problems generate equal swells of voltage at the amplifier's two inputs. Unfortunately, op-amps usually can't reject these signals. To ensure that this "common-mode" garbage (whose amplitude, can be over 1,000 time greater than the cardiac signal) adds no more than a 1 percent error, a CMRR (Common-Mode Rejection Ratio) of at least 100,000 to one (100decibels) is required. This precision eludes most op-amps. When an application calls for both high gain and a CMRR of 80 dB or greater special devices known as "instrumentation amplifiers" are required. The AD624AD from Analog Devices (see [W12]) when set to a gain of 1,000 has a CMRR exceeding 110 dB. It is available In PatuaTuli for TK-1500. Clearly at bit expensive, hence another option is the AD620A it is also available in PatuaTuli for TK-1000
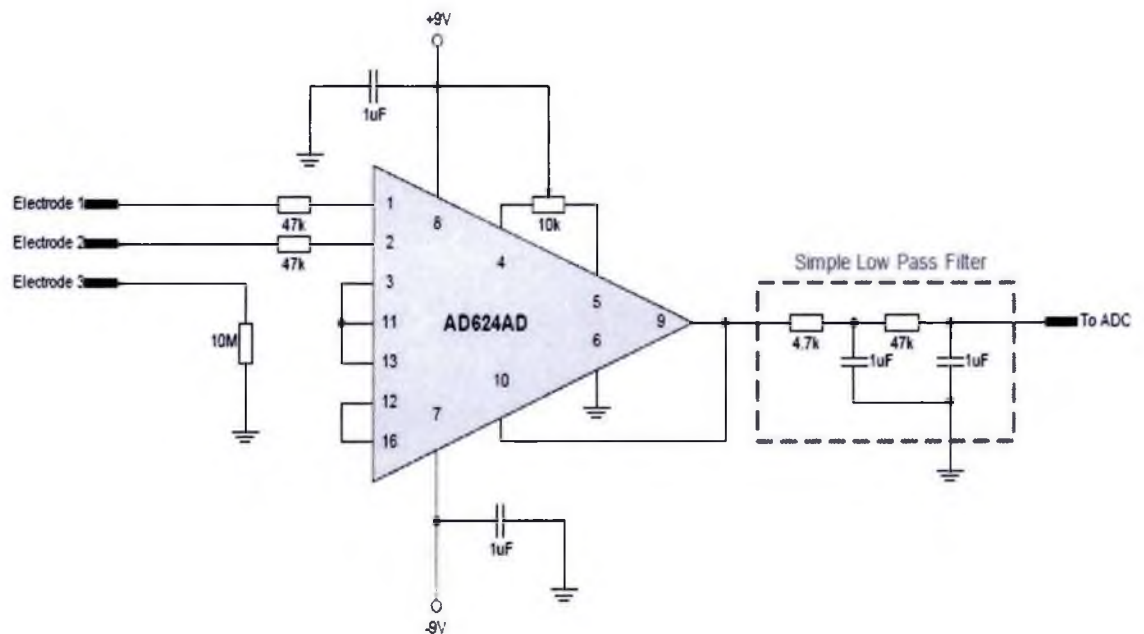
Amplifier :

Figure 4.4 Simple ECG Amplifier

Figure 4.4 shows a simple ECG amplifier using the AD624AD instrumentation amplifier. A gain of 1,000 is selected by shorting certain pins together as shown. The two-stage RC filter weeds out frequencies higher than about 50 hertz. A 3 lead cable connects the circuit to the electrodes and two wires are required to connect the output to an ADC for sampling.

Basically this is the amplifier diagram of the two channels, we made twelve channels with five more amplifier circuit. We couldnot been able to do the full test for the conductivity reason, but we have checked each of the amplifier circuit by taking pulses from hands.
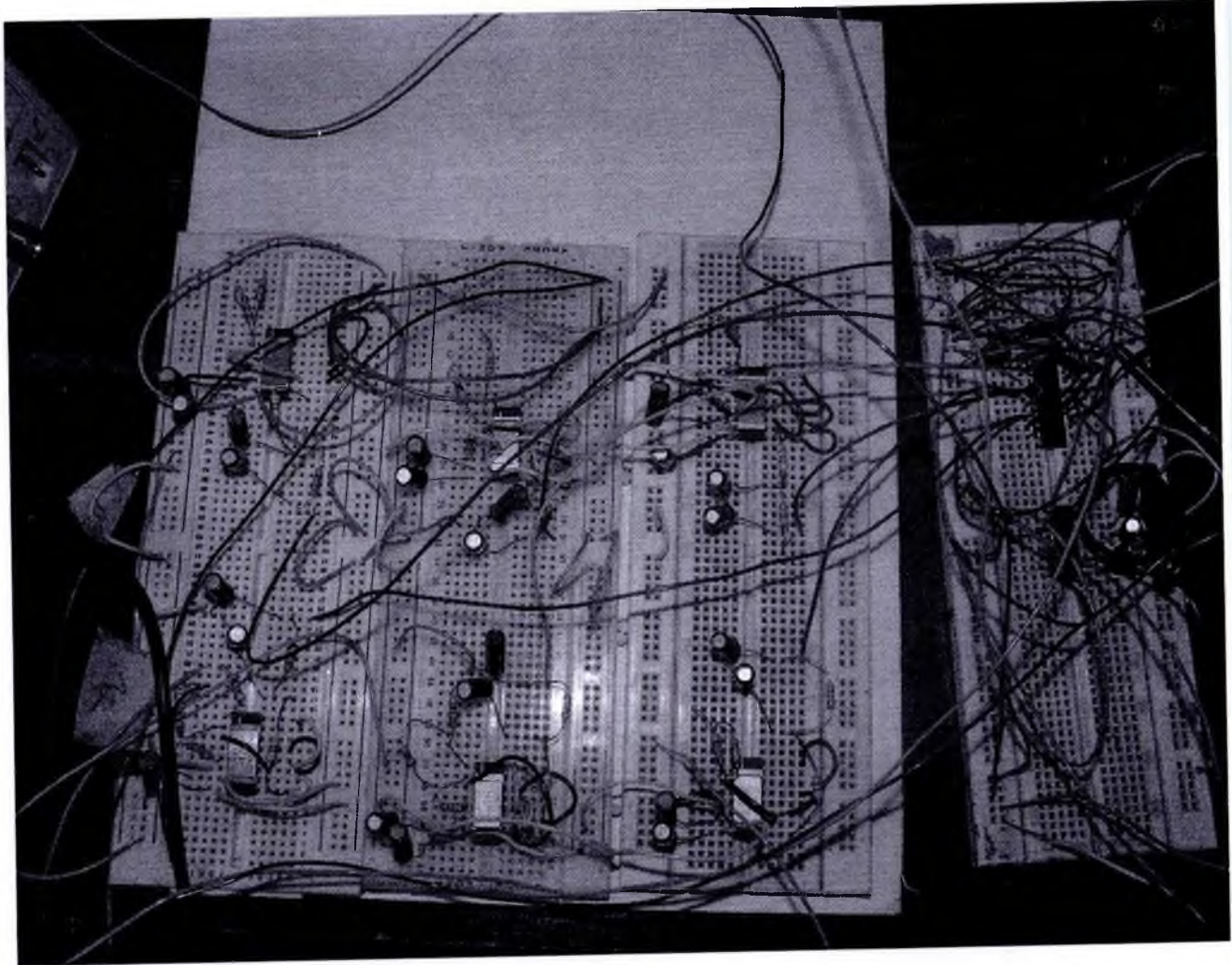
Figure 4.5: Implemented circuit

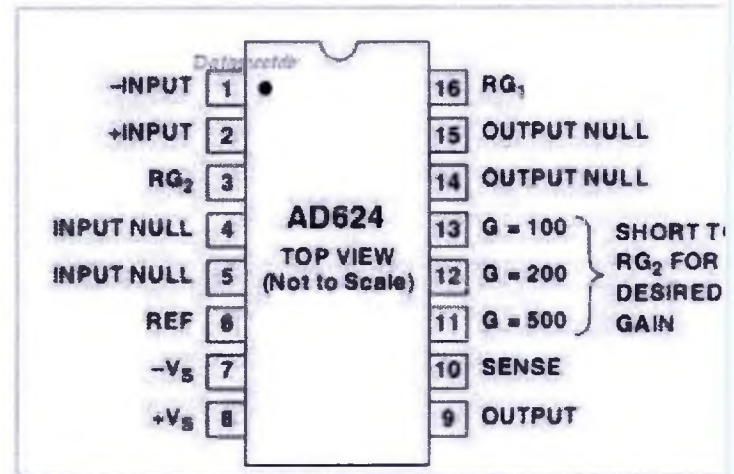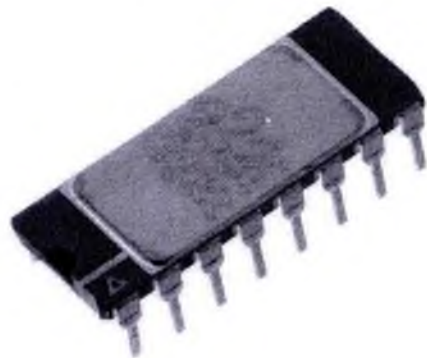## 4.6 Precision Instrumentation Amplifier:

Figure4.6:Precision Instrumentation Amplifier

FEATURES:

Low Noise: 0.2 mV p-p 0.1 Hz to 10 Hz

Low Gain TC: 5 ppm max (G = 1)

Low Nonlinearity: 0.001% max (G = 1 to 200)

High CMRR: 130 dB min (G = 500 to 1000)

Low Input Offset Voltage: 25 mV, max

Low Input Offset Voltage Drift: 0.25 mV/8C max

Gain Bandwidth Product: 25 MHz

Pin Programmable Gains of 1, 100, 200, 500, 1000

No External Components Required

Internally Compensated

The AD624 is a high precision, low noise, instrumentation amplifier designed primarily for use with low level transducers,including load cells, strain gauges and pressure transducers. An outstanding combination of low noise, high gain

accuracy, low gain temperature coefficient and high linearity make the AD624 ideal for use in high resolution data acquisition systems. The AD624C has an input offset voltage drift of less than 0.25 mV/°C, output offset voltage drift of less than 10 mV/°C, CMRR above 80 dB at unity gain (130 dB at G = 500) and a maximum nonlinearity of 0.001% at G = 1. In addition to these outstanding dc specifications, the AD624 exhibits superior ac performance as well. A 25 MHz gain bandwidth product, 5 V/ms slew rate and 15 ms settling time permit the use of the AD624 in high speed data acquisition applications.

5.Analog To Digital Conversion

5.1Digital Sampling

Digital sampling requires an ADC (analogue-to-digital converter) to converter analogue voltages to binary representation. The sampling rate specifies the number of samples taken per second. Figure 5.1 demonstrates clearly how an analogue waveform is digitally sampled and displayed onto the screen (LCD,Computer Monitor, or a CRT using a DAC etc...).
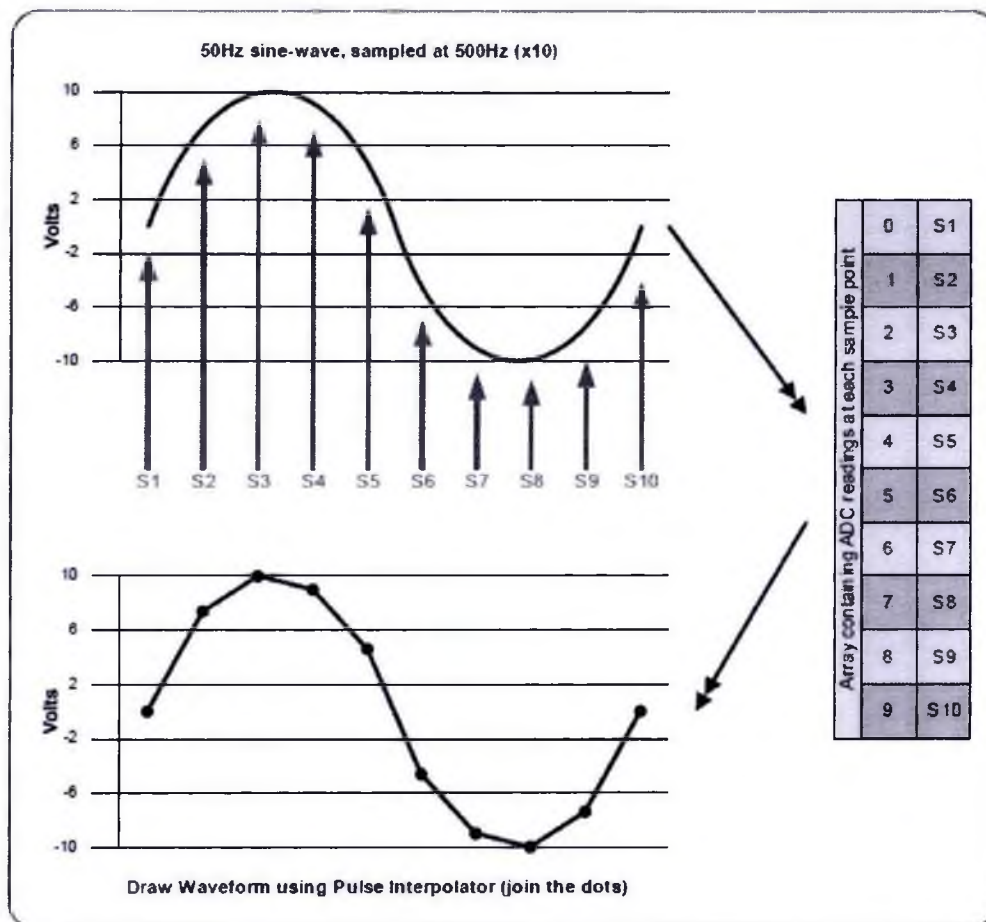
Figure 5.1a. Example showing how a sine-wave is digitally sampled

5.2 Aliasing

Aliasing is an undesirable effect that can occur when digital sampling analog voltages. This is the display of an apparent signal which does not actually exist, usually caused by under-sampling. Many samples should be taken per cycle (Nyquist theorem states that *"to define a sine wave, a sampling system must take more than two samples per cycle".*) to ensure an accurate representation of an analog signal in a digital memory. If only one sample is taken per cycle, or one sample per several cycles, then aliasing occurs. For example say a waveform is

being sampled every three cycles, these samples may form together, particularly when using pulse interpolation (join the dots), to look like a valid waveform.
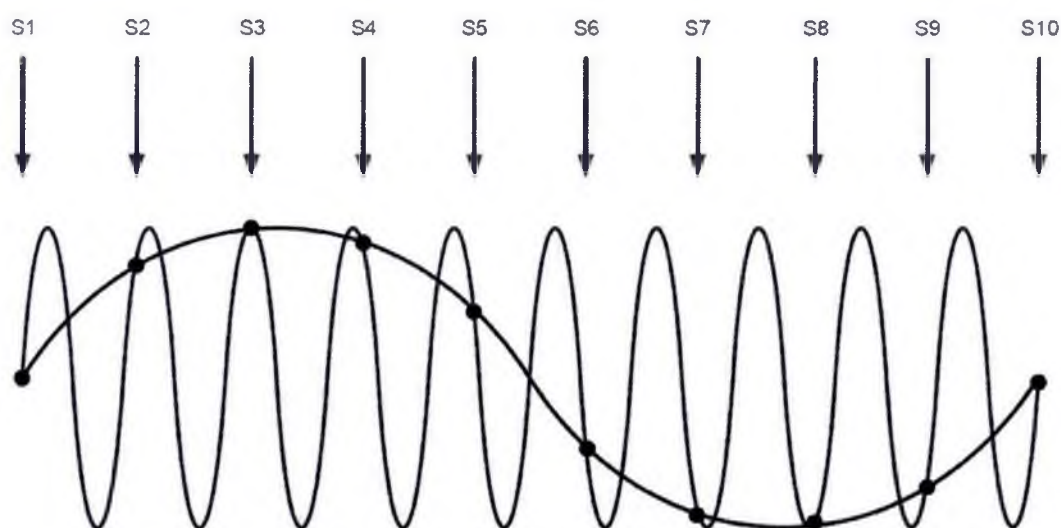

Figure 5.2a. Demonstrating aliasing, red is the real waveform, while blue is an alias.

Figure 5.2 a clearly demonstrates how false signals (aliasing) are created. The red waveform is the real waveform, notice that the waveform is under sampled (see green arrows for sample points). The black dots shows were the real waveform (red) has been sampled, by joining the dots, it is clear that a perfect sinewave is created (blue), which is an alias of the original signal. Note that it is impossible to tell that the blue signal is an alias. There is nothing that can be done after sampling to correct aliasing; hence the solution is to filter out high frequencies by sending the input signal through a low-pass filter. Ideally all frequencies above half the sample rate should be filtered out.

5.3 Microcontroller :
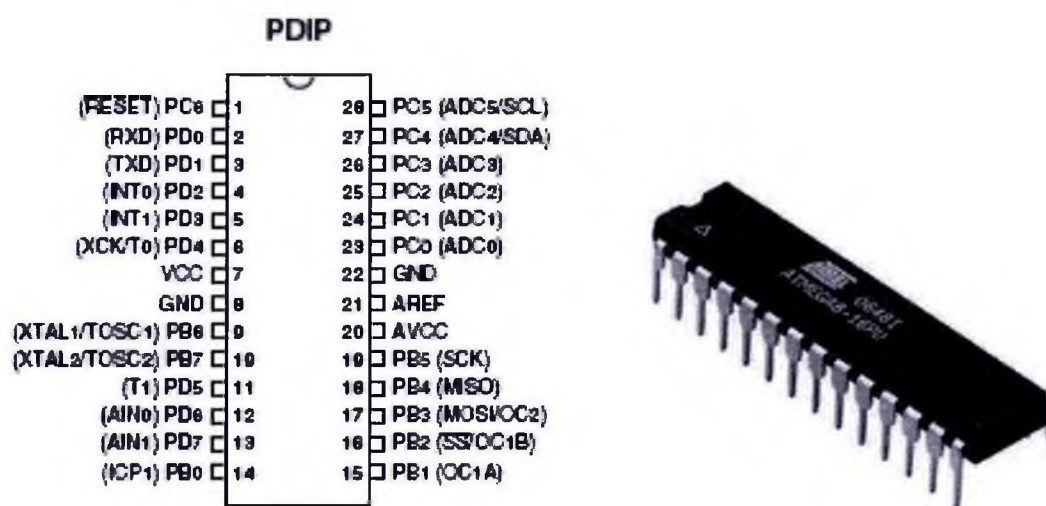
FIGURE 5.2: Pin Configuration Of ATMEGA8

Features

• High-performance, Low-power AVR® 8-bit Microcontroller

• Advanced RISC Architecture

   ✓ 130 Powerful Instructions – Most Single-clock Cycle Execution

   ✓ 32 x 8 General Purpose Working Registers

   ✓ Fully Static Operation

   ✓ Up to 16 MIPS Throughput at 16 MHz

   ✓ On-chip 2-cycle Multiplier

• High Endurance Non-volatile Memory segments

   ✓ 8K Bytes of In-System Self-programmable Flash program memory

   ✓ 512 Bytes EEPROM

- ✓ 1K Byte Internal SRAM
- ✓ Write/Erase Cycles: 10,000 Flash/100,000 EEPROM
- ✓ Data retention: 20 years at 85°C/100 years at 25°C(1)
- ✓ Optional Boot Code Section with Independent Lock Bits

In-System Programming by On-chip Boot Program

- ✓ True Read-While-Write Operation
- ✓ Programming Lock for Software Security

• Peripheral Features

- ✓ Two 8-bit Timer/Counters with Separate Prescaler, one Compare Mode
- ✓ One 16-bit Timer/Counter with Separate Prescaler, Compare Mode, and

Capture

Mode

- ✓ Real Time Counter with Separate Oscillator
- ✓ Three PWM Channels
- ✓ 8-channel ADC in TQFP and QFN/MLF package

Eight Channels 10-bit Accuracy

- ✓ 6-channel ADC in PDIP package

Six Channels 10-bit Accuracy

- ✓ Byte-oriented Two-wire Serial Interface
- ✓ Programmable Serial USART
- ✓ Master/Slave SPI Serial Interface
- ✓ Programmable Watchdog Timer with Separate On-chip Oscillator
- ✓ On-chip Analog Comparator

• Special Microcontroller Features

- ✓ Power-on Reset and Programmable Brown-out Detection
- ✓ Internal Calibrated RC Oscillator
- ✓ External and Internal Interrupt Sources

    ✓ Five Sleep Modes: Idle, ADC Noise Reduction, Power-save, Power-down, and

Standby

• I/O and Packages

    ✓ 23 Programmable I/O Lines

    ✓ 28-lead PDIP, 32-lead TQFP, and 32-pad QFN/MLF

• Operating Voltages

    ✓ 2.7 - 5.5V (ATmega8L)

    ✓ 4.5 - 5.5V (ATmega8)

• Speed Grades

    ✓ 0 - 8 MHz (ATmega8L)

    ✓ 0 - 16 MHz (ATmega8)

• Power Consumption at 4 Mhz, 3V, 25°C

    ✓ Active: 3.6 mA

    ✓ Idle Mode: 1.0 mA

    ✓ Power-down Mode: 0.5 µA

    ✓ 8-bit with 8K Bytes.In-System Programmable Flash ATmega8

6. Serial Interface and Outputs

6.1RS232 Serial Interface

RS232 is simple, universal, well understood and supported, but it has some serious shortcomings as a data interface. Its origins predate modern computers and it contains many features that are not relevant to the modern user. It can control very old primitive modems and has many control signals to do this in hardware, but often it is used without these old control and status lines. Its major feature is that it does not require the transmission of a clock, the reception of a 'start bit' is enough to cause the receiver to time all its actions from this one edge. This is called asynchronous transmission.

RS232 allows a 5% difference in transmitted timings and receiver chip timings. This is important if using a ATMEGA-8 as the datasheet specifies the % error of

the baud rate generator at certain baud rates (the higher thebaud rate, the higher the % error), as long as this error is less than 5% the RS232 standard is capable ofcoping. Electronic data communications between elements will generally fall into two broad categories: single-endedand differential. RS232 (single-ended) was introduced in 1962, and despite rumours for its early demise, hasremained widely used.
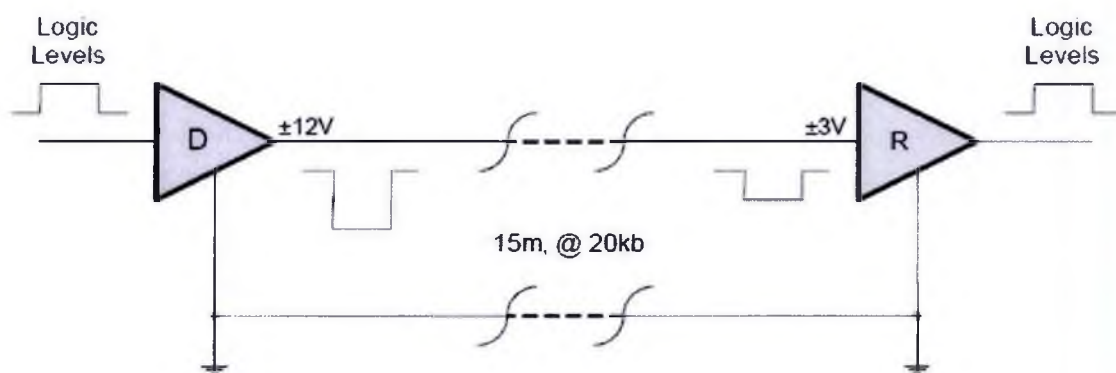

Figure 6.1a. Illustration of RS232, 1 driver and 1 receiver

RS232 data is bi-polar, e.g. a +3 to +12 volt indicates an SPACE (ON) while a -3 to -12 volt indicates an MARK (OFF). Modern computer equipment ignores the negative level and accepts a zero voltage level as the MARK (OFF) state. This means circuits powered by 5 VDC are capable of driving RS232 circuits directly; however, the overall range that the RS232 signal may be transmitted/received is dramatically reduced. The output signal level usually swings between +12V and -12V. The 'dead area' between +3v and -3v is designed to absorb line noise. This dead area can vary for various RS232 like definitions, for example the definition for V.10 has a noise margin from +0.3V to -0.3V. Many receivers designed for RS232 are sensitive to differentials of 1v or less.
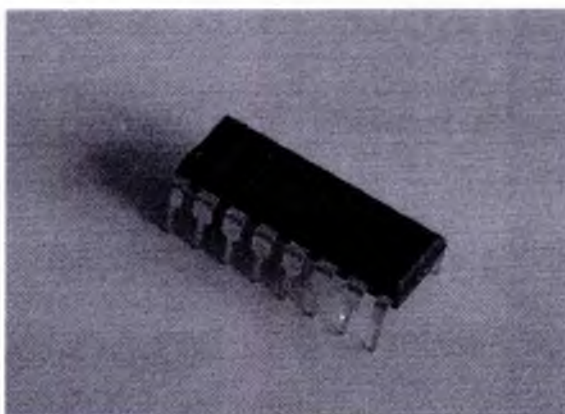
Pin	Signal	Pin	Signal
1	Data Carrier Detect	6	Data Set Ready
2	Receive Data	7	Request to Send
3	Transmit Data	8	Clear to Send
4	Data Terminal Ready	9	Ring Indicator
5	Signal Ground		

Figure6.2: Serial Port

6.2 Max232

Typical line drivers / receivers chips for RS232 are the Maxim MAX232 or–MAX233 chips (seehttp://www.maxim-ic.com) the original specification states that RS232 should drive 50 feet, but modern line driver/receivers can manage much better than this. In this project we have used MAX232 as driver of the serial port communication.

PIN CONFIGURATION

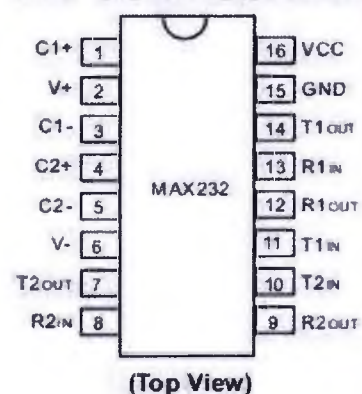|	MAX232	
C1+ 1		16 VCC
V+ 2		15 GND
C1- 3		14 T1out
C2+ 4		13 R1in
C2- 5		12 R1out
V- 6		11 T1in
T2out 7		10 T2in
R2in 8		9 R2out

(Top View)

Figure6.3 : Pin Configuration Of Max 232

This is the table which is showing the driving capacity of RS232

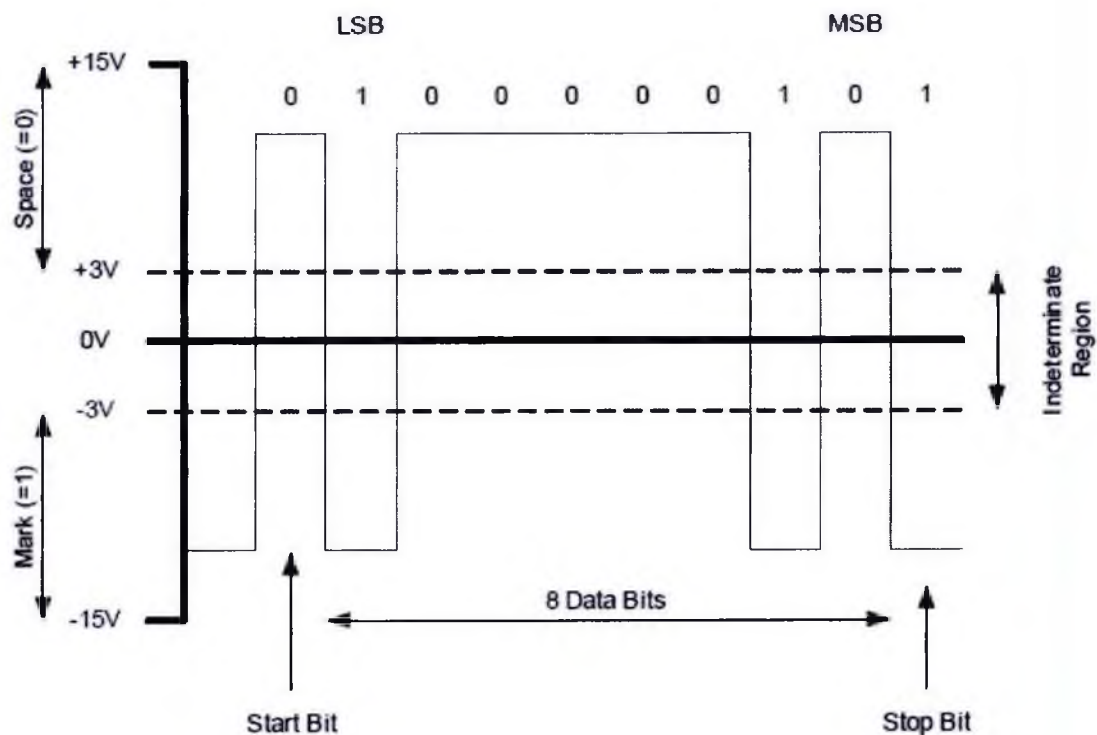| Baud Rate	Max Distance Shielded Cable	Max Distance Un-shielded Cable
110bps	5000feet	3000feet
300bps	5000feet	3000feet
1200bps	3000feet	3000feet
2400bps	1000feet	500 feet
4800bps	1000feet	250 feet
9600bps	250feet	250 feet

TABLE: 1

Figure6.4: Illustration of how data is transmitted over RS232

6.3Baud-Rate Generator, BRG

This is basically a programmable 8-bit counter followed by a switchable frequency ÷4 flip flop chain which can be set up to give the appropriate sampling and shifting rates for the desired baud rate, based on the

ATMEGA 8's crystal frequency XTAL (e.g. for 20MHz, XTAL = 20) giving: -

Baud rate(Low Speed Mode)=xtal/64*?(x+1)

Baud rate(Low Speed Mode)=xtal/16*?(x+1)

X=xtal*10^6/64*Baud

It may be advantageous to use the high baud rate (BRGH = 1) even for slower baud clocks as this may reduce baud rate error in some cases.

6.4 HARDWARE DEVELOPMENT

The purpose of this project is to design, built and test a **low-cost** Acquisition and Monitoring of Multichannel ECG. The main reasoning behind hardware development was to keep the hardware cost to an absolute minimum. It was decided to use the ATMEGA 8 (flash version), because it has enough RAM for time-compressed memory without the need for an external RAM chip. *"The ATMEGA8 is a high-performance FLASH microcontroller that provides engineers with the highest design flexibility possible. In addition to 8192x14*

words of FLASH program memory, 256 data memory bytes, and 368 bytes of user RAM, ATMEGA8 also features an integrated 8-channel 10-bit Analogue-to-Digital converter.

The ATMEGA's ADC is used for data acquisition, a MAX232 buffer is used to convert the TTL serial logic of the ATMEGAs UART to the correct RS232 format (connect to a PC for data logging).

6.5 Design Considerations

Accuracy, dependability, and precision are an absolute must if the device were to be used for diagnostic, or other medical purposes. Any small fluctuation in the waveform generated could carry critical diagnostic value, thus it is extremely important that the clinician can confidently and fully rely on the equipment. This means that the ECG must faithfully display the cardiac signal exactly as it exists in reality, such that any irregularity detected did in fact arise from an unhealthy cardiac cycle, *not* from the equipment that was used. Therefore, there were many special considerations that had to be taken into account when designing the ECG Monitor.

Noise:

First and foremost in these considerations is the effect of *noise*. Noise interference in the signal detection process would be detrimental to the experiment, as the ECG signal is at such small amplitudes it could easily be masked by noise related fluctuations. Therefore in order to detect the signal accurately, there must be strict limitations on the acceptable level of noise allowed, and every possible attempt must be made to minimize this level and reduce the effects of noise on the data acquisition process.

Signal Amplitude:

Another consideration that strongly influenced the design of the ECG is the fact that the cardiac signal generated has a very small peak amplitude. (As stated above, this is the very signal attribute that makes noise control so vital). Considerable amplification is necessary if there is any use to be made from the cardiac signal in terms of analysis and output. Also, the small size of the signal plays a very influential role in the approach to creating a system of visual output. Caution has to be taken to effectively differentiate between actual changes in the signal amplitude as opposed to a random variation in noise amplitude.

Low frequency:

Because the signal that is generated from the cardiac muscle has such a low bandwidth, it is very important that the ECG have a good low frequency response. This is because any shifts in the frequency of the detected signal, especially the S-T portion of the waveform, carry critical diagnostic value.

6.6 **Outputs And Results:** After implementing the amplifier circuit we took the analog output. Here is the analog output of the amplifier circuit.
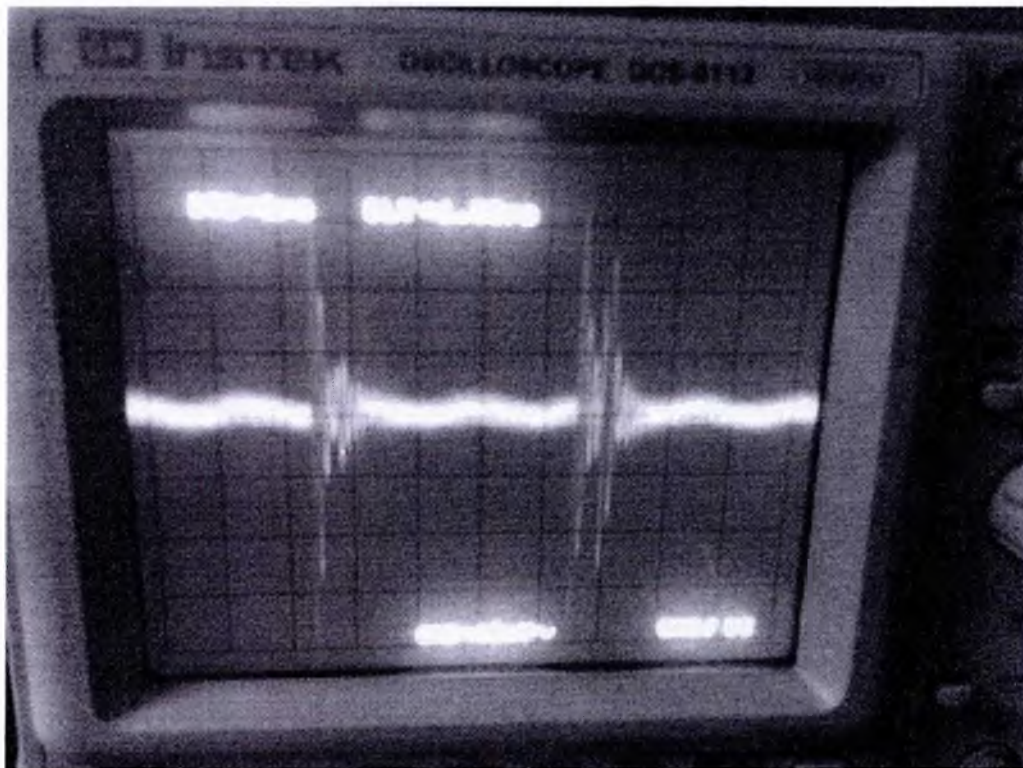
Figure 6.5: Analog Output

After using the microcontroller for digital sampling we got the digital wave from the digital oscilloscope. In figure 6.5 ... we can see the output of the digital oscilloscope.
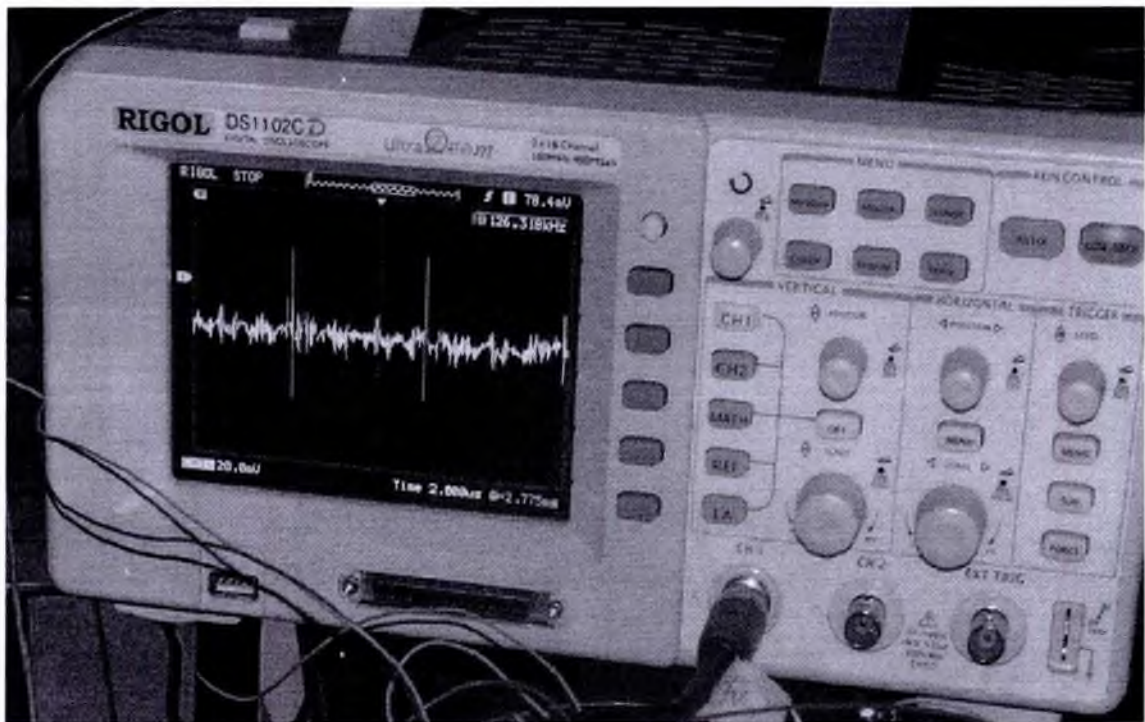
Figure6.6: Digital Output

Finally after serial communication we got the final output from Matlab. In figure we can see the final output.
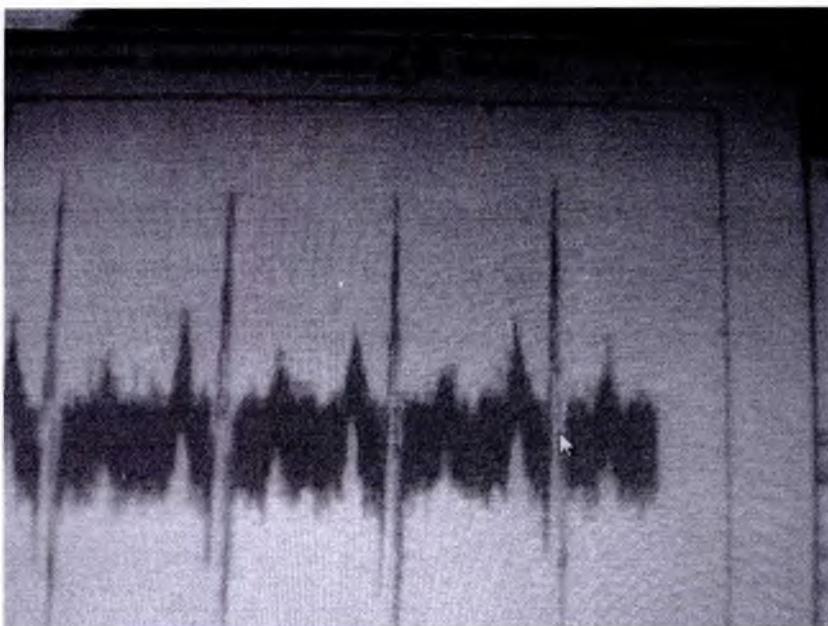

Figure6.7: Serial Output

7.1 Further Developments:

For furthur developments We would like to design the whole circuit in Printed Circuit Board. For Low conductivity we had to go through So many problems as the desired wave shape was changing with the movements of the hands, so we for that we will try to use ECG clamps and wires with the marginal budget. For filtering we willtry the matlab Filters for digital output.

7.2 Conclusion:

To conclude this project, the main objective has been achieved as a satisfactory level. All the two-electrode amplifier circuits and heart rate detector circuit were developed and constructed successfully and working properly. This project has enhanced the author plenty of knowledge about the ECG, from the principles of the physiology to the technique of circuit developing. This project will also enable the readers further understand the ECG principles, the technique applied for data acquisition and monitoring, especially using the two-electrode amplifier technique for making 12 channels.

Bibiliography:

http://www.digchip.com/datasheets/parts/datasheet/041/AD624AD-pdf.php

http://innovexpo.itee.uq.edu.au/2002/projects/s358223/thesis.pdf

http://www.idt.mdh.se/utbildning/exjobb/files/TR0702.pdf

http://library.epfl.ch/en/theses/?nr=3565

http://www.mathworks.com

http://www.atmel.com/dyn/resources/prod_documents/doc2486.pdf

http://www.captain.at/electronic-atmega16-serial-port.php

http://www.codeproject.com/

Codes For Analog To Digital Conversion(ADC):

```c
#include <avr/io.h>
#include <avr/interrupt.h>
#include <util/delay.h>
#ifndef F_CPU
//define cpu clock speed if not defined
#define F_CPU 8000000
#endif
//set desired baud rate
#define BAUDRATE 9600
//calculate UBRR value
#define UBRRVAL ((F_CPU/(BAUDRATE*16UL))-1)
//define receive parameters
#define SYNC 0XAA// synchro signal
#define RADDR 0x44


#define LEDON 0x11//LED on command
#define LEDOFF 0x22//LED off command
void Display(unsigned int in);
void Display1(unsigned int in1);
#define SEGMENT_DATA_DDR DDRB
#define SEGMENT_DATA_PORT PORTB
#define SEGMENT_CONTROL_DDR DDRD
#define SEGMENT_CONTROL_PORT PORTD
#define sbi(address,bit) (address |= (1<<bit));
#define cbi(address,bit) (address &=~ (1<<bit));
#define SEG1 cbi(PORTD,PD7) sbi(PORTD,PD6)
#define SEG2 cbi(PORTD,PD6) sbi(PORTD,PD7)
void delay_ms(unsigned int ms){
```

```c
    while(ms){
        _delay_ms(1.000);
        ms--;
        }
}
void USART_Init(void)


{
    //Set baud rate
    UBRRL=(uint8_t)UBRRVAL;            //low byte
    UBRRH=(UBRRVAL>>8);        //high byte
    //Set data frame format: asynchronous mode,no parity, 1 stop bit, 8 bit size
    UCSRC=(1<<URSEL)|(0<<UMSEL)|(0<<UPM1)|(0<<UPM0)|
        (0<<USBS)|(0<<UCSZ2)|(1<<UCSZ1)|(1<<UCSZ0);
    //Enable Transmitter and Receiver and Interrupt on receive complete
    UCSRB=(1<<TXEN);

}
void USART_vSendByte(uint8_t u8Data)
{
    // Wait if a byte is being transmitted
    while((UCSRA&(1<<UDRE)) == 0);
    // Transmit data
    UDR = u8Data;
}
void Send_Packet(uint8_t cmd)
{
    USART_vSendByte(cmd);//send increment command
}
void InitADC()
{
```

```c
ADMUX =(1<<REFS0) ;//| (1<<REFS1) ;                    // For Aref=AVcc;

ADCSRA=(1<<ADEN)|(1<<ADPS2)|(1<<ADPS1) | (1<<ADPS0); //Rrescalar div factor =128

}

uint16_t ReadADC(uint8_t ch)

{

  //Select ADC Channel ch must be 0-7

  ch=ch&0b00000111;

  ADMUX|=ch;

  //Start Single conversion

  ADCSRA|=(1<<ADSC);

  //Wait for conversion to complete

  while(!(ADCSRA & (1<<ADIF)));

  //Clear ADIF by writing one to it

  ADCSRA|=(1<<ADIF);

  return(ADC);

}

int main()

{

  InitADC();

DDRB=0x00;

  int adc_val,adc_val1,adc_val2,adc_val3,adc_val4,adc_val5;

  while(1)

  {

        InitADC();

  adc_val=ReadADC(0);

                InitADC();

  adc_val1=ReadADC(1);

                InitADC();

  adc_val2=ReadADC(2);

                InitADC();
```

```c
        adc_val3=ReadADC(3);
                InitADC();
        adc_val4=ReadADC(4);
            InitADC();
        adc_val5=ReadADC(5);
////////////////////

        if(PINB & (1<<PB0))
        Send_Packet(adc_val);
                    if(PINB & (1<<PB1))
        Send_Packet(adc_val1);
                    if(PINB & (1<<PB2))
        Send_Packet(adc_val2);
                    if(PINB & (1<<PB3))
        Send_Packet(adc_val3);
                    if(PINB & (1<<PB4))
        Send_Packet(adc_val4);
                    if(PINB & (1<<PB5))
        Send_Packet(adc_val5);
                    ////////////
    }
        return 0;
}
```

Serial Port Interface Using MATLAB:

Reading and Plotting ECG Signal From Hardware:

CODE :

```
clc

clear all

% Creating serial obj ..

s=serial('com6')

%Configuring properties (baudrate-databits-stopbits-InputBuffersize and Outputbuffersize) ..

set(s,'baudrate',128000,'databits',8,'stopbits',1) ;

set(s,'InputBufferSize',1000)

%connecting

fopen(s)

% Reset RTS pin to begin Transmission

s.requestTosend='off';

s.requestTosend='on';

% Reading Binary data From ECG Hardware ..

data=fread(s)

% To see Data in Binary ..

databin=dec2bin(data)

%Plotting the received Array of bits ..

plot(data, 'DisplayName', 'data', 'YDataSource', 'data');

% Disconnecting and cleaning ..

fclose(s)

delete(s)
```