Low Cost Data Acquisition System for ECG

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ABSTRACT
The objective of this research work is to develop a low cost circuitry for Electrocardiogram (ECG) data acquisition system that outputs optimum ECG data with very low noise and minute spectral information loss. The experimental results demonstrate that by the careful choice electrodes and placement of electrodes, it is indeed possible to collect pure ECG signal. By using this system, acquires the filtered and amplified ECG data through Line-In port of sound card of computer through MATLAB software. In the design of ECG data acquisition used the instrumentation amplifier packed IC module instead of separate opamp module which results in high common mode rejection ratio up to 80db and minimizes the offset voltage, to very low value up to 25µV and in the design of the notch filter circuits uses UAF42IC, and in the design circuitry of low pass, high pass filter uses the opamp TLO84ICs. The ECG data recorded by developed system useful for purposes of diagnosis of heart related diseases and also researchers study purposes.

Keywords
Electrocardiogram, LABVIEW, Notch Filter, CMRR

1. INTRODUCTION
Electrocardiogram (ECG) signals plays a vital role in clinical diagnosis especially for diagnosing heart related diseases and disorders such as, cardiovascular disease (CVD), pulmonary disease, sudden cardiac arrest (SCA), etc [1]. ECG signal is generated by a nerve impulse stimulus to a heart. The current is diffused around the surface of the body and build on the voltage drop, which is a normally 0.0001 to 0.003volt and the signals are within the frequency range of 0.05 to 100 Hz [1] [2]. ECG signals are usually recorded at the surface of the body and processed to give important information about the electrical activity of heart.

A typical ECG tracing of a normal heartbeat consists of a P wave, a QRS complex and a T wave. Usually, the signal which is acquired from the human body is of very low potential and difficult to analyze the signal variance. Hence, necessary amplification is required before processing the ECG signal to derive any give useful information about the cardiac abnormalities. The elements of ECG complex Zeli Gao et.al, developed an 2 lead ECG device with lead I configuration, Right Leg driven circuit and used total gain of 1000. The active filter is used to obtain ECG signal with a frequency range of 0.05 Hz to 150 Hz and used NI USB 6008DAQ card to integrate with LAB VIEW [3].

In [4], the researchers have developed a lead II ECG data acquisition device with a cut off frequency of low pass and notch filter with a value of 150 Hz and 60 Hz, respectively. They used ADuC831 DAQ system to be integrated to J Free Chart for ECG data acquisition. Steve et. al used lead II configuration with a gain of 987[5]. These researchers acquired ECG signal with a frequency range of 0.1-50Hz and implemented inverting amplifier before hardware integration to my DAQ to be read by LABVIEW software [5]. Above mentioned literature works, motivated us to develop low cost ECG data acquisition .

Based on the literature, this paper research work consists of several stages such as, pre-amplification, isolation, filtering and second stage of amplification to acquire the ECG signals using three ECG leads [6-7]. In general, multi-stage amplifiers are required to amplify the ECG signals with a larger gain. Meanwhile, the amplifiers should have a high common-mode rejection ratio (CMRR) to amplify the ECG signal. This amplifier usually amplifies the most useful information of heart activities along with inherent noises developed in a system during data acquisition. These noises are filtered using both high and low pass filters, to extract the ECG signal between 0.05 Hz and 113 Hz.

To develop data acquisition system we employed an instrumentation amplifier to reject common mode signals and optimize its CMRR using resistance tuning and matching. A single order high pass filter of 10 Hz is used in the feedback of Instrumentation Amplifier output stage to prevent the saturation of data due to Base line noise and motion Artifact. Then to remove other noise such as ECG and other biomedical signal interference, 6th order unity gain chebyshev Active high pass filter(Sallen-Key Architecture) and to remove thermal noise a 4th order Butterworth active low pass filter is employed. To overcome the problem with non-ideal operation of op-amps at high frequency one single order passive low pass RC filter is also employed. To remove line frequency components a new approach is used than the conventional notch filter as twin-t filter and Inductor Simulated Notch Filter. After design and development of acquisition system, we acquire the filtered and amplified ECG data through Line-in port of sound card of computer using MATLAB software.

Our study is organized as follows: Section 1 is the introduction, Section 2 describes the block diagram of single channel ECG signal acquisition system, Section 3 presents experimental results and discussion and Finally, the conclusion is made in Section 4.

2. DEVELOPMENT OF ECG SIGNAL ACQUISITION SYSTEM
A detailed study of physiological origin of ECG signal, thorough study of characteristics of ECG signal, amplitude, time and frequency domain properties of ECG signal and sources of noise during its acquisition provided us insight into the complex nature of ECG signal. The use of surface electrodes for measuring the action potentials accompanying muscle contraction requires high-gain amplifiers having flat frequency response and high input impedance. The amplitude of ECG signal varies from few micro volts to few mill volts. The usable energy of signal is limited to 2 Hz to 100 Hz frequency range, with dominant energy being 10 to 60 Hz range [4]. The amplitude, time and frequency domain

properties of ECG signal depends on the following factors according to The various sources of noise that the raw ECG are inherent noise in the electronics components in the detection and recording equipment, motion artifacts, and inherent instability of the signal.

Another important factor is cross talk that should be taken care in the design of ECG signal processing system. Cross talk can be dealt with choosing the appropriate size of the electrodes conductive area. A circuitry was proposed for a single channel surface-ECG data acquisition system that outputs optimum ECG data with very low noise and minute spectral information loss is shown in Figure 1 below.

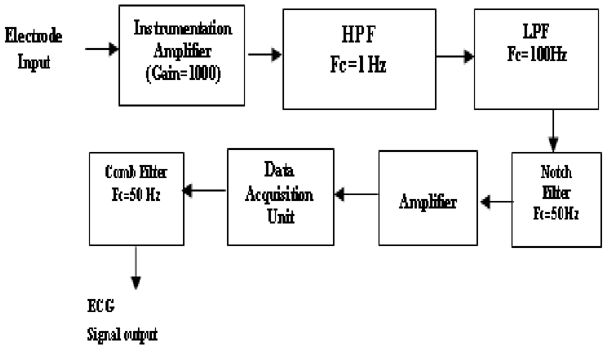

Fig1: Block diagram of single channel ECG data acquisition system

2.1 Instrumentation amplifier improvement

In our conventional design circuitry made with the instrumentation amplifier with separate operational amplifier module (TL082 JFET Op-Amp). But this circuitry was highly sensitive and error prone due to components mismatch. Since discrete resistance components possess slight mismatch from the specified values, this leads to low CMRR (Common mode rejection ratio). This type of circuitry also involves lot of discrete passive components. The circuit we made gives a CMRR of 30 dB.

In the improved circuit, we have chosen an Instrumentation Amplifier packed IC module INA101 from Texas Instruments (14-pin plastic and ceramic DIP)[5]. This IC module comprises of high accuracy resistances inside which provides very fine matching of sensitive resistances and results in high CMRR up to 80 dB practically (100 dB specified). This also reduces the discrete peripheral resistance count. Additionally, it reduces the circuit complexity, minimizes the offset to a very low value (specified 25µV max), and reduces the noise.

2.2 INA Saturation Prevention

During testing we found that sometimes our signal gets saturated in spite of being passed through a single order high pass filter of cutoff frequency 1 Hz and subsequently through a 6th order low pass filter of cutoff frequency (20 Hz). We analyzed this situation and found that the saturation cause is Movement artifact. This effect occurs during the movement of the body part from which the ECG data is being acquired. The Figure 2 shows the Block diagram of Instrumentation amplifier circuit.

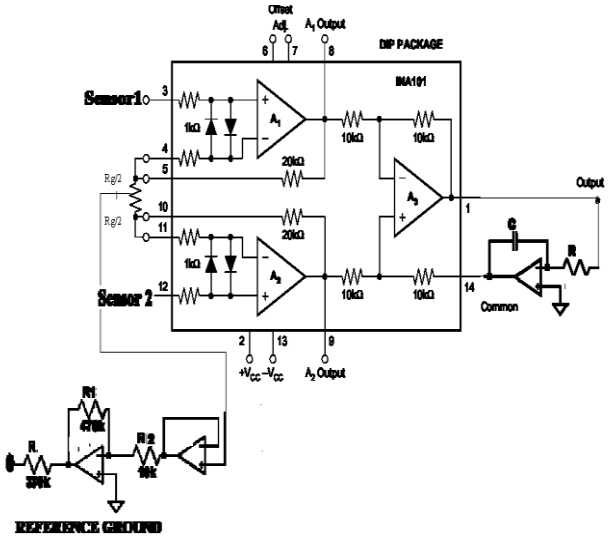

Fig. 2:Instrumentation amplifier circuit with reduced components

skin under contact of sensors moves causing an error signal of significant amplitude and has frequency components of around 10Hz. ideally this 10 Hz signal should pass through INA output terminal. But because of the high amplitude of this artifact signal, the output of difference amplifier (the last unit of Instrumentation Amplifier) gets saturated. To prevent the saturation, we have to somehow prevent the motion artifact noise at the output of Instrumentation amplifier. and also by preventing this signal to pass through difference amplifier and this strategy leads to the prevention of saturation during testing, which may be caused due to any movement of skin.

2.3 High-pass filter frequency alteration

On taking output directly after the Instrumentation amplifier stage in the computer through sound card and observing the frequency spectrum of the acquired signal, we found that the major part of ECG signal starts from above 2-3Hz and it possesses substantially high and significant power spectral density (around 25Hz). According to the papers and texts available [6] on ECG signal, we had constructed a High pass filter with a cutoff frequency of 1 Hz. But as per the definition of Butterworth filter, this is the 3 dB frequency and this leads us to the significant attenuation of signal be above this range., which contains a majority of the signals information. To prevent the attenuation and distortion of the signal, so we decreased our high pass cutoff frequency to 1Hz. .

2.4 Addition of an overshoot in High pass filter

Another important source of noise is power Line frequency of 50 Hz which is comes from power supply, or power lines of the walls of rooms.. But to remove the power line noise of 50Hz, we utilize a notch filter that leads to the drastic attenuation in the nearby frequencies also. In the lower part of frequency spectrum this decrement starts slowly from 20 Hz and then drastic decreases in the nearby frequencies of 50 Hz. This again leads us to the spectral distortion and loss of information content. To overcome this problem we changed our high pass filter design such that it gives slight overshoot in the frequency range of 20-45 Hz .This technique amplifies the signal of this frequency range slightly as in fig, 3, and gives less spectral distortion. Since this region has very high power spectral density components compared to the frequencies that lie in the upper part of line frequency, it comprises of significant amount of information about signal shape and characteristics. Hence, there is no need to apply such techniques in the upper part of the signal.

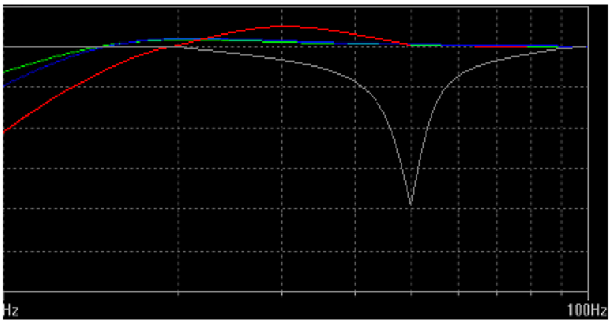

Fig. 3. Output of notch filter and with Overshoot at 30Hz

2.5 Change in Notch Filter Design

Traditional design of notch filter that was Twin T notch filter design gave the 20dB notch at 50 Hz and 3dB bandwidth of 30Hz (35 Hz-65 Hz). We changed our design of the notch filter where we applied an Inductor simulation technique by op-amps, capacitances and resistances. The filter response was 30 dB notch at line frequency with a 3 dB bandwidth of 20 Hz (40 Hz-60 Hz).Finally we employed a readymade Active filter module UAF042 from Texas instruments [7]. This IC contains low pass filter along with High pass filter circuit. Only few external resistances are required to make a filter. Using this IC we subtracted a low pass filter output to High pass filter by a difference amplifier. This high pass filter and low pass filter response exactly match at 50 Hz and the difference amplifier subtracts this particular frequency.

Since this involves high precision resistive components, the output dependency on the external parts reduces. Finally we are able to achieve 40dB attenuation at line frequency and approximately 5 Hz, 3 dB bandwidth (47.5 Hz to 52.5Hz) as shown in fig, 4. Below are the different designs and their responses. The UAF42 is a monolithic, time-continuous, 2nd-order active filter building block for complex and simple filter designs. It uses the classical state-variable analog architecture with a summing amplifier plus two integrators.

This topology offers low sensitivity of filter design parameters f_o (natural frequency) and Q to external component variations along with simultaneous high-pass, low-pass and band-pass outputs. A notch filter is easily realized with the UAF42 and six external resistors. The pass outputs, At $f = f_{notch}$, both of these outputs times their respective auxiliary operational amplifier is used to sum both the high-pass and low-gain at the summing circuit are equal in magnitude but 180° out of phase. Hence, the output goes to zero. The notch frequency for the notch filter is set by the Following calculations:

$$f_{notch}=\sqrt{A_{LP}/A_{HP} * R_{Z2}/R_{Z1}}$$

Where, $\mathbf{A_{LP}}$ = gain from input to low-pass out at **f = 0Hz**.

$\mathbf{A_{HP}}$ = gain from input to high-pass out of **f >> $\mathbf{f_0}$.**

Typically, $A_{LP}/A_{HP} \cdot R_{Z2}/R_{Z1}$ is equal to one. This simplifies

f_{NOTCH}, to be, i. e $f_{notch} = \mathbf{f_0}$.

$$f_o=\frac{1}{2\pi CR_f}$$

Where, $R_f = R_{f1} = R_{f2}$ and $C = C_1 = C_2$

For C=1nF, and $f_o = 50$ Hz, $R_F = 1.6M\Omega$.

The –3dB bandwidth, as shown in Figure 4, can be set by the following calculations.

BW–3dB = $\mathbf{f_{notch}/Q}$ where, BW –3dB = $f_H – f_L$.

The filter Q can be determined by setting R_Q to a value given by,

$$R_q=\frac{25k\Omega}{Q-1}$$

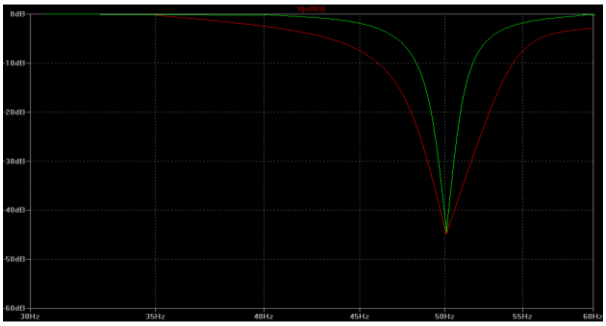

Fig 4: The Red signal shows the response of Simulated Inductor Notch filter and Green represents the response of UAF42 Notch filter response.

2.6 Change in Notch filter design

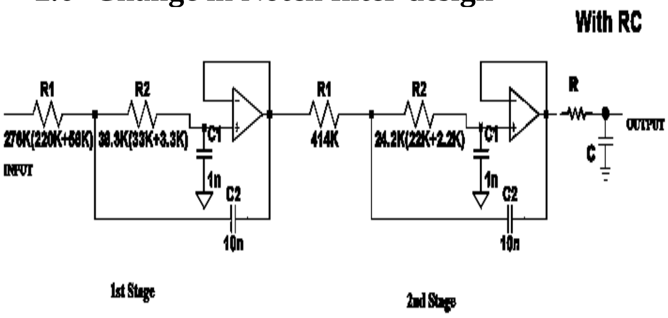

Fig 5: Low pass filter with an extra passive filter at the output.

We have used Sallen Key Architecture in making of Low pas filter. This filter topology has certain disadvantages: As at high frequency, where we expect the response to keep attenuating at –40 dB/dec, the filters actually turn around and start passing signals at increasing amplitudes.

3. EXPERIMENTAL RESULTS AND DISCUSSION

Muscles which give the substantial amount of signal are selected for ECG data recording. We placed one differential electrode on the Left leg, other differential electrode on the Right arm and placed reference electrode on the Right leg for Lead II ECG Recording. After the electrodes are placed, the procedure is explained to the subject asked to practice several times while data are recorded. ECG data for desired Lead of ECG is recorded. Raw ECG signal has been acquired using line-in port of the sound card with built-in programmable ADC and sample and hold mechanism. The line-in port of the sound card of Personnel computer was configured to sample at sampling frequency of f_S = 2000 Hz and at 16 bit resolution. MATLAB code used for recoding ECG data. Fig. 6, shows waveform of ECG signal obtained from a developed hardware module and Figure-7 shows the Normalized FFT of ECG data.

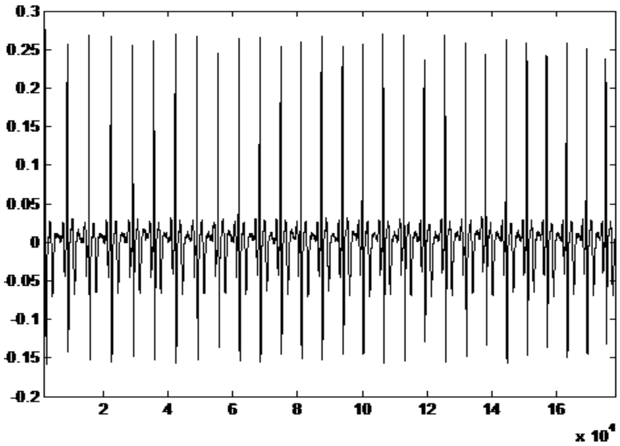

Fig 6:A typical recorded ECG Signal

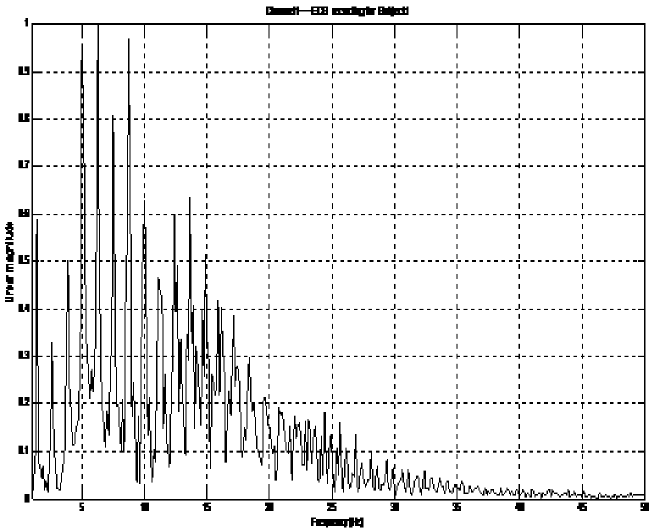

Fig 7: Normalized FFT of recorded ECG data

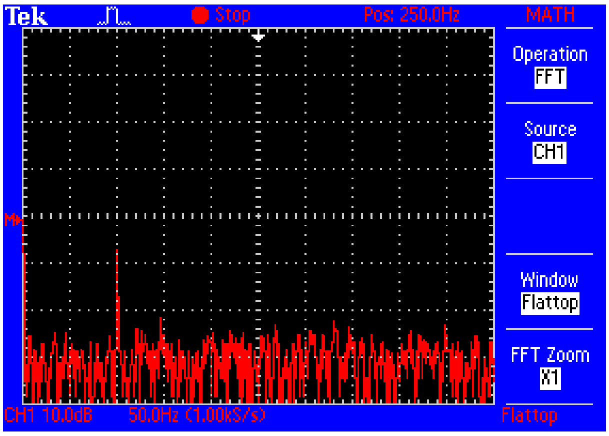

Fig 8: Instrumentation amplifier CMRR performance for output signal

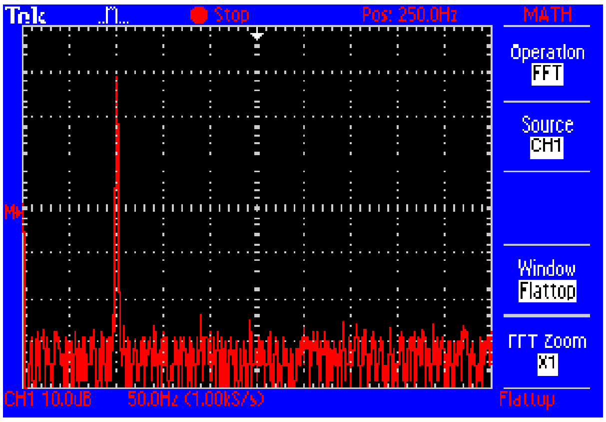

Fig 9: Instrumentation amplifier CMRR performance for input signal

4. CONCLUSION

This paper research work successfully demonstrated the low cost data acquisition for ECG signal. developed Low cost circuit shows the less noisy and less distorted signal with very little spectral information loss and implemented on the general purpose printed circuit board (PCB).

Design and implementation of each stage is compared with the conventional circuits and the optimum design approach is chosen. To process the data in computer we used the line-in port of the sound card of computer. and implemented a MATLAB code to collect the real time data through line port of computer and acquire it for further processing with adjustable sampling frequency and ECG data is analyzed by FFT algorithm.

5. REFERENCES

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