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Experimental data acquisition and processing system for ECG signals

Natal - RN
October, 2021

Abstract

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Keywords: oi. turo. bem.

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1 Introduction

An electrocardiograph (ECG) is an instrument responsible to record the electrical activity of the heart, considering that electrical signals are generated before the heart mechanical functions and generated by nerve impulse stimulus. In this context the ECG provides information about those signals that allows to detect a variety of cardiac disorders [1, 7].

The ECG machine has became an essential instrument throughout the years once it is a noninvasive, simple to record and minimal cost device [2]. Regarding its importance, the objective of this work is to design an experimental setup for an ECG machine, highlighting all project steps so anyone can learn experiment with a simple and accessible device for the analysis of cardiograph signals.

The first step of the ECG design is to get to know how the cardiac signal is generated, its specifications and the noises associated with the data acquisition.

Once define those guidelines, the next step is to develop a reliable source of cardiac signals, both in the software and hardware domain, also known as a simulator. A simulator from the software point of view can be used to learn how to process the signal with multiples digital signal processing algorithms in a higher level of abstraction. From the hardware point of view there is a need to generate a cardiac signal in the analog domain so it can be acquired by the proposed data acquisition system (DAQ). For the latter a consolidated ECG device could be used, but once the project relies in the simplicity of it resources, a simple circuit serves its purposes.

The third step concerns the ECG design itself. The sub steps are:

- Design the low noise preamplifier(instrumentation amplifier), since the acquired signal by the electrodes has low voltage;
- Design a low-pass analog filter serving as an antialising filter to avoid out of band noise contamination after the signal is digitized;
- Design a high-pass filter to eliminate the DC component and ...

- Design a Notch filter (reject band filter) centered in 60 Hz to remove any interference from the power grid;
- Data acquisition by an analog-to-digital converter (ADC);
- Program the micro controller interface between the analog front end and the digital part responsible for analyzing, processing, storing and displaying the final graphical output.

Where the first five sub steps integrate the analog front end that is responsible for conditioning and digitizing the signal.

The final step is to validate the system with experiments and discuss the results.

2 Specifications

2.1 Cardiac signal

The electrical potentials of the cardiac signal are acquired by various electrodes connected in the surface of the skin of the patient. Those combined generate the cardiac signal, as can be seen in the normal wave pattern of figure 1, with voltage differences in the order of $1\,mV$ between given points in the body [8].

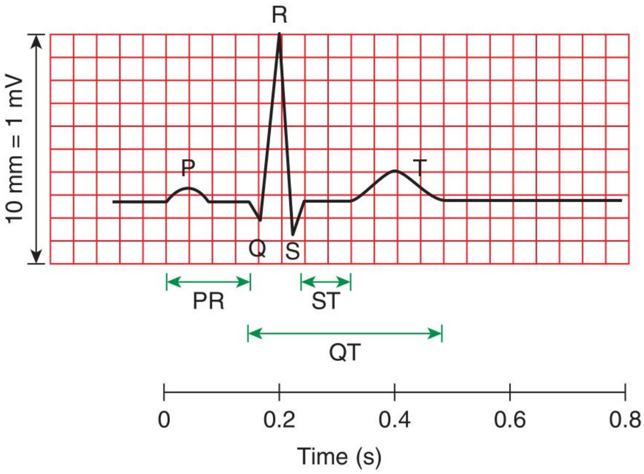

Figure 1: Normal waveform pattern of cardiac signal obtained in ECG. Source: Khandpur [8].

Regarding the figure 1, each letter has a proper meaning for each step in the cardiac cycle. Those are:

- The P wave represents the depolarization of the atrial muscles;
- The QRS section is a combination of the atria repolarization between QR and the ventricles depolarization between RS;
- The T wave represents the repolarization of both ventricles;

The interval PR represents the actrial systole, of which the diastole lasts from R until the next P wave. The ventricular systole occurs between R until the end of the T wave and its diastole lasts until Q [12].

The interval QRS representing the time taken by the heart impulse to travel from the inter ventricular system then through the walls of the ventricles is the most critical one when thinking in the design of an ECG machine since it has the higher frequency of all the cardiac signal and lasts about 0.05 to 0.1 seconds [8].

Those characteristics leads to a deeper specification of the ECG regarding the signal. Khandpur [8] defines the frequency range of the signal varying from 0.05 to 150 Hz. By the Nyquist sampling theorem, it is recommended that the sampling frequency used in digitization to be at least two times the higher frequency in the signal, i.e 300 samples/s. Despite this, Khandpur [8] exerts that a sampling rate of 200 samples/s is satisfactory, yielding 12 to 20 samples for the QRS interval. Therefore the system designed in this work will attend to the Nyquist criteria and beyond, using a sampling frequency of 500 samples/s. The disadvantage in this approach is that compared to the sampling frequency proposed by Khandpur [8], the present system will need 2.5$\times$ more space due its higher sampling frequency.

Regarding the bit resolution to store the cardiac signal, Khandpur [1] suggests two approaches: to use low-noise and high-gain amplifiers, enabling the use of low-resolution 16-bit ADC; or using a low-gain amplifier with a high-resolution 24-bit ADC. The first approach will be used in this project for the sake of a more elaborated amplifier design and signal conditioning, instead just using a better ADC.

Since the arrangement of electrodes is not the main focus of this work, the system will stick with a bipolar leads arrangement. This arrangement uses two electrodes placed in the right and left arm to capture the signal and send to the input of an instrumentation amplifier and other electrode as reference placed in the left leg [8]. This means that there is no need for a multiplexing system for multiples channels of electrodes, like in a typical 12-lead ECG [6].

2.2 Associated noises

In the chain of processing, the cardiac signal must be filtered in a way to remove certain noises and interference inherent to the data acquisition. The most common source are: interference from the power-lines (power grid) as a tone in $50/60\,Hz$ (depending on the region of the globe); noise generated mechanically due to the contact between electrodes and skin, motion artefacts firing random derived from the patient movement and muscle contraction (voluntary or involuntary); additive white Gaussian noise (AWGN) derived from thermal sources; or electromagnetic interference from other electronics devices that can extends to the RF spectrum or higher [1].

It is essential to a ECG instrument to maintain clear of those noises and interference in a level of approximately $10\,\mu V$ peak to peak to ensures ECG applications in diagnostics [1].

Each source of noise and interference can be treated in a specific manner.

Power-line interference can be solved designing a Notch filter (reject band) with cutoff frequency set to $50/60\,Hz$ (this project will validate the filter to $60\,Hz$).

Baseline wanders and muscle contraction are phenomena of low frequencies and they can be eliminated with the help of a high-pass filter. In the previous section the lower frequency defined to the cardiac signal was $0.05\,Hz$, so a high-pass filter with this value as cutoff frequency should be able to remove those interference without risking rising too much the cutoff and attenuating the P or T waves [8, 1, 7]. Sahin, Fidel, and Perez-Castillejos [10] suggests the use of a $0.5\,Hz$ cutoff frequency in a fourth order Butterworth filter, therefore allowing the use of a simpler filter with relaxed requirements and alerting that this cutoff may be tuned to meet the requirements. The lower cutoff frequency will be used in this project.

AWGN and electromagnetic interference are phenomena of mainly high frequencies, therefore can be eliminated with a low-pass filter that might be designed with the anti aliasing filter. The anti aliasing filter is a analog and low order filter, but can also be used to eliminate electromagnetic interference since it has a frequency band way higher than the signal. The AWGN is uniform thought out all frequencies, so the previous high-pass filter will contribute to eliminate some of the noise. The cutoff frequency selected for this filter will be 200 Hz which comprises the signal band.

An alternative idea is to merge both low-pass and high-pass filters into a band-pass filter. The problem associated with this strategy is that it is not possible to select different filter orders to the low-pass and high-pass segment. Even though this project will stick with a band-pass filter combining the high-pass and low-pass filter cutoff frequencies, thus simplifying the circuit.

Below there is table 1 showing a summary of the analog filters to be designed in further sections.

Filter	Cutoff frequency	Order
Notch (analog)	60 Hz	X
Band-pass (analog)	0.05 Hz to 200 Hz	X

Table 1: Summary of filters.

3 Methodology

3.1 System overview

The idealized system follows the sketch of the figure 2

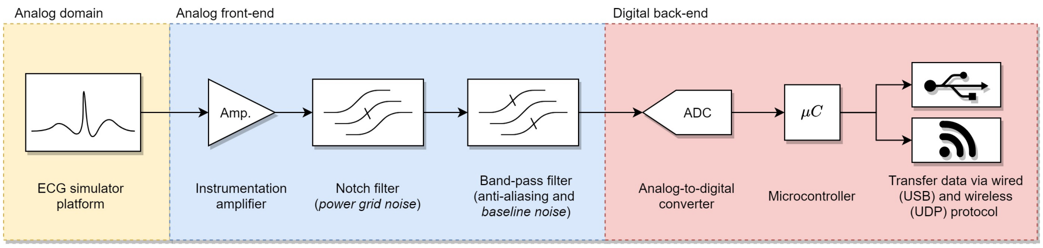

Figure 2: Idealized system for acquiring, processing and transferring ECG data.

3.2 Preamplifier

The amplifier architecture used is an instrumentation amplifier of which resembles the topology of a differential amplifiers but with the addition of buffers on each input. The complete circuit can be seen in the figure 3 below.

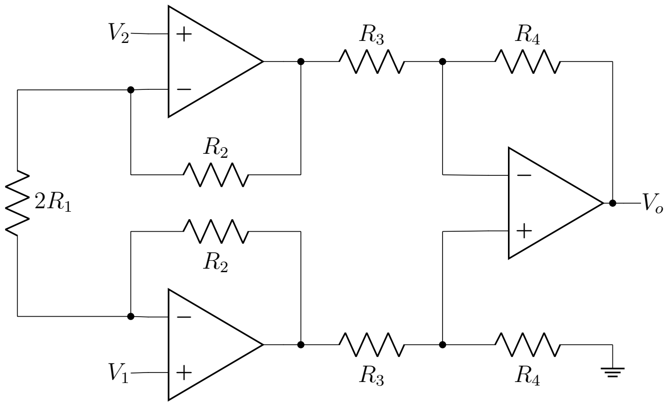

Figure 3: Instrumentation amplifier.

The output of the instrumentation amplifier can be written as equation 1:

$$V_o = \mathrm{DG}(V_1 - V_2) + \mathrm{CMG}\left(\frac{V_1 + V_2}{2}\right) \tag{1}$$

Where DG is the differential gain and CMG is the common mode gain given by equations 2 and 3:

$$\mathrm{DG} = \frac{R_4}{R_3}\left(1 + \frac{R_2}{R_1}\right) \tag{2}$$

$$\mathrm{CMG} = \left(\frac{R_4}{R_3 + R_4}\frac{R_1 + R_2}{R_1} - \frac{R_2}{R_1}\right) \tag{3}$$

In ideal condition the common mode gain is null with the traditional choice of $R_1 = R_3$ and $R_2 = R_4$, and the final output turns to be $V_o = \mathrm{DG}(V_1 - V_2)$.

The problem is that CMG will never be null due to imprecision in the resistance value of the resistors. Despite this, there is a metric called Common Mode Rejection Ratio (CMRR) given by the equation 4 in dB that indicates how much times higher the differential gain should be relative to the common mode gain. In the case of ECG, this ratio must be higher than $+100\, dB$ [8, 3]. Also a input impedance of at least $10\, M\Omega$ in the input buffers is recommended [1].

$$\mathrm{CMRR} = 20\mathrm{log}\left(\frac{\mathrm{DG}}{\mathrm{CMG}}\right) \tag{4}$$

Usually the differential gain for the preamplifier in a ECG is 500 [8]. A better approach would be to distribute this gain along a multistage amplifier to avoid distortion due to non linearities [11]. Despite this statement, the project will be based in a single amplification block for the sake of simplicity,

.

The differential gain for the project should be close to 506 with the resistors $R_2 = R_4$ with $22\,k\Omega$ resistors and the resistors $R_1 = R_3$ with $1\,k\Omega$ resistors.

Considering the gain, the choice of resistors can be illustrated by the table 2. Also the operational amplifier LMH6629 [5] was chosen once it is SMD, has ultra-low noise and has a slew rate of $1600\,V/\mu s$ that can follow the variation of the QRS wave.

Component	Identifier	Number of components
Resistor	$22\,k\Omega$	4
Resistor	$1\,k\Omega$	4
Opamp IC	LMH6629	3

Table 2: Summary of electronic components for the preamplifier.

3.3 Analog filters

3.3.1 Notch filter

The Notch filter topology used in this project follows the circuit of figure 4. Besides implementing the low-pass and high-pass segment, this filter has a buffer so it isolates from the output and at the end, the output is fed back fractioned to adjust the filter's quality factor (Q).

The cutoff frequency of the previous Notch filter can be found by the equation 5. Keeping in mind that the cutoff frequency is $60\,Hz$, the capacitor and resistor could be $100\,nF$ and $13.2\,k\Omega$. Once the found resistance does not match with a commercial value, the resistor R_N can be used with a series association of a $13\,k\Omega$ and $200\,\Omega$ resistor.

$$f_c = \frac{1}{4\pi R_N C_N} \tag{5}$$

The quality factor (Q) is a function of the cutoff frequency and the bandwidth of the filter, given by the equation 6 where B_W is the bandwidth

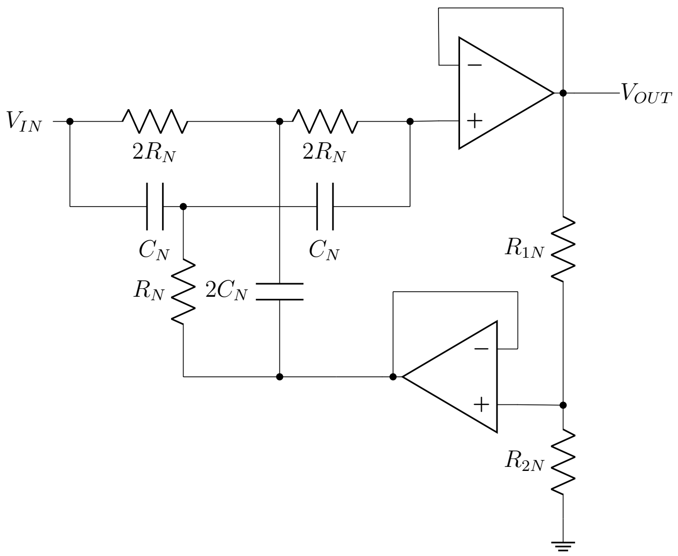

Figure 4: Notch filter.

in Hertz. Setting a bandwidth of 1 Hz with reject band between 59.5 and 60.5 Hz, which is a band that the power grid voltage can assume in the worst case during anomalies according to ANEEL [4]. Henceforth the quality factor is found to be $Q = 60$ and the remaining resistor can be found by the equation 7 as $R_{1N} = 180\,\Omega$ and $R_{2N} = 47\,k\Omega$.

$$Q = \frac{f_c}{B_W} \tag{6}$$

$$1 - \frac{1}{4Q} = \frac{R_{2N}}{R_{2N} + R_{1N}} \tag{7}$$

The table 3 summarizes the components choices for the Notch filter.

Component	Identifier	Number of components
Resistor	$47\,k\Omega$	1
Resistor	$180\,\Omega$	1

Component	Identifier	Number of components
Resistor	$13\,k\Omega$	5
Resistor	$200\,\Omega$	5
Capacitor	$100\,nF$	4
Opamp IC	LMH6629	2

Table 3: Summary of electronic components for the Notch filter.

3.3.2 Band-pass filter

Once stated before, the high-pass and low-pass filter were merged into a single band-pass filter. One possible topology for the band-pass filter would be merged a cascade association of a high and low-pass filter, but this choice wastes improvement possibilities. Instead the topology of choice is the one in the figure 5

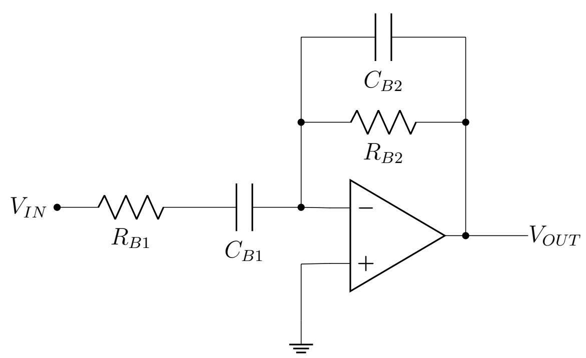

Figure 5: Band-pass filter.

The filter stated above is an inverting active band-pass filter. Its equations are: equation 8 to select the gain between input and output, which in this case will be set to one, resulting in $R_{B1} = R_{B2}$; equation 9 to select the capacitance of C_{B1} based in the lower cutoff frequency ($f_L = 0.05\,Hz$); and equation 10 regarding C_{B2} based in the higher cutoff frequency ($f_H = 200\,Hz$).

$$\text{Gain} = A_v = -\frac{R_{B2}}{R_{1B}} \tag{8}$$

$$f_L = \frac{1}{2\pi R_{B1} C_{B1}} \tag{9}$$

$$f_H = \frac{1}{2\pi R_{B2} C_{B2}} \tag{10}$$

An inverting amplifier with unitary gain can be placed in the output of the band-pass filter to correct the polarity of the resulting signal, or instead just inverting when taking the output between V_{OUT} and ground node.

The values found for the components are summarized in the table 4, considering $R_{B1} = R_{B2} = 500\,k\Omega$, $C_{B1} = 6.36\,\mu F$ and $C_{B2} = 1.59\,nF$. The capacitor C_{B1} was broken into a parallel association of four capacitors $2.2\mu F$, $2.2\mu F$, $1.8\mu F$ and $0.1\mu F$. The same for C_{B2} but with a parallel association of $680pF$, $680pF$, $220pF$ and $10pF$.

Component	Identifier	Number of components
Resistor	$500\,k\Omega$	2
Capacitor	$2.2\,\mu F$	2
Capacitor	$1.8\,\mu F$	1
Capacitor	$0.1\,\mu F$	1
Capacitor	$680\,pF$	2
Capacitor	$220\,pF$	1
Capacitor	$10\,pF$	1
Opamp IC	LMH6629	1

Table 4: Summary of electronic components for the band-pass filter.

3.4 ECG simulator

Prior to the tests of the digital processing algorithms, there is a need to simulate typical ECG signal with a good model in the software domain. The high level abstraction allow it to debug faster than deploying and testing direct in the hardware.

The model used was the one presented by Quiroz-Juárez et al. [9], whose consists in the use of three different nonlinear oscillators related to the cardiac natural pacemakers where three coupled oscillators represent the action potentials of the SA node, AV node and His-Purkinje complex, as illustrated in the figure 6 extracted from Quiroz-Juárez et al. [9].

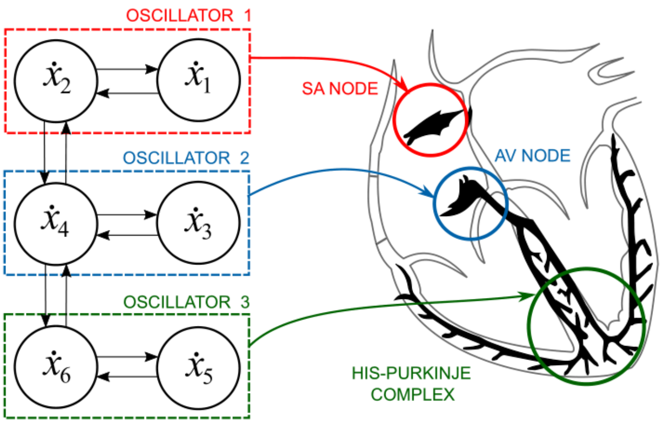

Figure 6: Diagram relating the cardiac natural pacemakers to non linear variables. Source: Quiroz-Juárez et al. [9].

Quiroz-Juárez et al. [9] also derive an ODE system of nonlinear equations to represent the given oscillators, as illustrated in the equation 11.

$$\begin{cases} \dot{x_1} = \Gamma_t \cdot (x_1 - x_2 - Cx_1x_2 - x_1x_2^2) \\ \dot{x_2} = \Gamma_t \cdot (Hx_1 - 3x_2 + Cx_1x_2 + x_1x_2^2 + 2\beta(x_4 - x_2)) \\ \dot{x_3} = \Gamma_t \cdot (x_3 - x_4 - Cx_3x_4 - x_3x_4^2) \\ \dot{x_4} = \Gamma_t \cdot (Hx_3 - 3x_4 + Cx_3x_4 + x_3x_4^2 + 2\beta(x_2 - x_4)) \\ \mathrm{ECG}(t) = \alpha_1x_1 + \alpha_2x_2 + \alpha_3x_3 + \alpha_4x_4 \end{cases} \quad (11)$$

The whole description is left for the original paper, but is known that a correct choice of the coefficients results in changes in the ECG waveform. Quiroz-Juárez et al. [9] comes with a set of coefficients that calibrates the ECG in different modes, varying from the normal cardiac rhythm and also represent pathologies such as: sinus tachycardia, atrial flutter, ventricular tachycardia and ventricular flutter. The table 5 illustrates the coefficients used by Quiroz-Juárez et al. [9] to generate each mode, starting from the initial excitation of $\mathbf{x} = \{0, 0, 0.1, 0\}$.

Mode	H	Γ_t	α_1	α_2	α_3	α_4
Normal rhythm	3	7	-0.024	0.0216	-0.0012	0.12
Sinus tachycardia	2.848	21	0	-0.1	0	0
Atrial flutter	1.52	13	-0.068	0.028	-0.024	0.12
Ventricular tachycardia	2.178	21	0	0	0	-0.1
Ventricular flutter	2.178	13	0.1	-0.02	-0.01	0

Table 5: Coefficients for the ECG simulator to access each mode. The coefficients $C = 1.35$ and $\beta = 4$ are constants along all the simulations. Source: adapted from Quiroz-Juárez et al. [9].

With this representation, a code in Python was created to simulate the ECG in software and further a firmware for the Espressif micro controller ESP32 was also created, allowing the internal 8-bit DAC (digital-to-analog converter) to continuously output the ECG simulation given by the model above. It was also predict a switch button in the micro controller to switch between each pathology. In both cases a fixed sampling period of 10^{-4} was used. The simulation results will be shown in further sections.

3.5 ECG embedded acquisition system

3.6 ECG viewer

4 Results

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5 Discussions

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6 Conclusions

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A Schematics