Detection of the R wave using analogue multiplier

Detección de la onda R empleando multiplicador analógico

CÓRDOVA-MANZO, Jesús Fausto†*, CÓRDOVA-ESCOBEDO, Jesús Fausto, MENDOZA-GONZÁLEZ, Felipe and GOMEZ-RODRIGUEZ, Cristian

Universidad Veracruzana, Faculty of Engineering, Coatzacoalcos-Minatitlán Region, Mexico.

ID 1st Author: *Jesús Fausto, Córdova-Manzo* / **ORC ID:** 0000-0001-6284-6990, **Researcher ID Thomson:** AAV-9602-2021, **CVU CONACYT ID:** 1149068

ID 1st Co-Author: *Jesús Fausto, Córdova-Escobedo* / **ORC ID:** 0000-0002-7456-6897, **Researcher ID Thomson:** S-6737-2018, **CVU CONACYT ID:** 511561

ID 2nd Co-author: *Felipe, Mendoza-González* / **ORC ID:** 0000-0003-1172-6782, **Researcher ID Thomson:** S-6747-2018, **CVU CONACYT ID:** 947336

ID 3rd Co-author: *Cristian, Gomez-Rodriguez* / **ORC ID:**0000-0001-9124-6037, **Researcher ID Thomson:** S-6787-2018, **CVU CONACYT ID:** 210208

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Abstract

In the present work, a novel methodology is proposed for the detection of the R wave of the electrocardiogram (ECG) making use of the analog multiplier AD633 and function in the methodology of the detection algorithm of the Pan-Tompkins R wave. With the proposed methodology, different ECG leads were simulated, achieving favorable results in respect of the R wave detection. The circuits present in this work were simulated in Proteus. The contribution of this work is to design a novel way to perform R wave detection with analog elements.

Analog multiplier, R wave, electrocardiogram

Resumen

En el presente trabajo se propone una innovadora metodología para la detección de la onda R del electrocardiograma (ECG) haciendo uso del multiplicador analógico AD633 basándose en la metodología del algoritmo de detección de la onda R Pan-Tompkins. Con la metodología propuesta, se simularon diferentes derivaciones del ECG logrando resultados favorables en la detección de la onda R. Los circuitos presentes en este trabajo fueron simulados en Proteus. La contribución de este trabajo es diseñar una novedosa forma de realizar la detección de la onda R con elementos analógicos.

Multiplicador Analógico, Onda R, electrocardiograma

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* Correspondence to the Author (Email: fcordova@uv.mx)
† Researcher contributing as first author.

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Introduction

An electrocardiogram (ECG) is a test in charge of recording the electrical signals of the heart; It is used to detect heart problems and evaluate the health of the heart.

The electrical activity is recorded from the patient's body surface and is drawn on a paper by means of a graphic representation, where different waves are observed that represent the electrical stimuli of the atria and ventricles. The medical equipment with which the electrocardiogram is obtained is called an electrocardiograph.

Figure 1 Lead II electrocardiogram showing premature ventricular contraction (PVC).
Source: https://es.dreamstime.com/ image213247971

For the acquisition of electrical activity by the electrocardiograph, electrodes placed on the patient's skin are needed. These electrodes will be attached to the electrocardiograph by cables. With 10 electrodes, 12 leads are obtained.

This results in 12 traces of the heart's electrical impulses from different points in the body. There is the possibility of adding extra leads by adding more electrodes to the body surface, however, the basic electrocardiogram consists of a minimum of 12 leads.

The electrocardiogram of a healthy person has a characteristic tracing. Due to this very particular morphology of the signal, there are changes in said tracing and the specialist can determine if there is a problem with the heart.

The ECG is useful for measuring the heart rhythm, the size and position of the atria and ventricles, any damage to the heart, and the effects that certain drugs or devices implanted on the heart may have.

The alterations shown in the trace are essential for the detection and analysis of cardiac arrhythmias. It is very useful in acute episodes of coronary disease, such as myocardial infarction.

It is a simple, available, fast test that does not cause any discomfort and there is no risk for the patient since no type of current is sent, it only detects the electrical activity that is generated in the heart itself.

ECG waves
(My EKG, 2021)

The electrocardiogram records the electrical activity of the heart in line tracings on paper. The peaks and valleys traced are called waves.

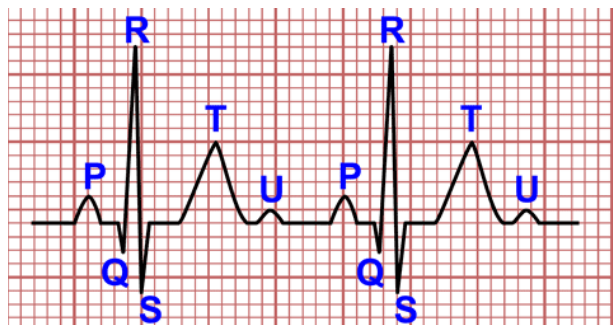

Figure 2 ECG waves. P, Q, S, R and T, different types of waves that the electrocardiogram produces
Source: MyEKG.com

The P wave is the first wave of the cardiac cycle, it records electrical activity through the atria (upper chambers of the heart). It is composed of the superposition of the electrical activity of both atria. The initial part corresponds to the depolarization of the right atrium and its final part to that of the left atrium

The T wave represents the moment when the lower chambers of the heart are electrically restored and ready for the next muscle contraction.

The U wave is positive of low amplitude, it appears immediately behind the T wave. It signifies the repolarization of the papillary muscles in precordial leads.

QRS complex

Group of waves that result from the depolarization of the ventricles. Its duration ranges from 0.06 s to 0.10 s. Depending on the derivation, it takes different morphologies.

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Q wave: Represents the first wave of the QRS complex and is negative.

R wave: it is the first positive wave of the QRS complex, sometimes preceded by a negative wave (Q wave). If there is another positive wave in the QRS complex, it is called R '.

S wave: it is the negative wave that appears after the R wave.

QS wave: when a complex is completely negative, without the presence of a positive wave, it is called a QS complex. It is usually a sign of necrosis.

R 'and S' waves: when there is more than one R wave or more than one S wave, they are called R 'and S'.

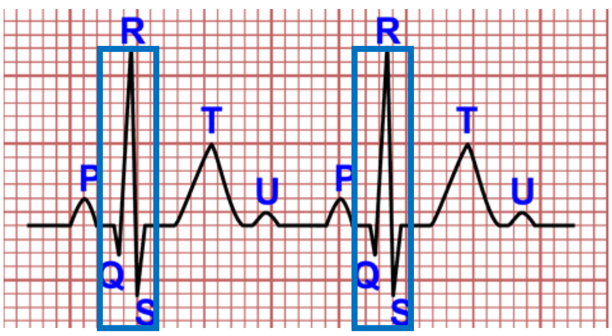

Figure 3 ECG signal where the QRS complex stands out
Source: MyEKG.com

Analog multipliers
(Cancino de Greiff, 2021)

An analog multiplier is a circuit with 2 inputs, which generates a voltage output (V) represented by the following formula.

$$v_o = kv_xv_y \quad (1)$$

Where:

v_o is the exit $[V]$

v_x, v_y they are entrance $[V]$

k is a constant with $[V^{-1}]$

The polarity of the multiplier inputs are classified as:

- Four-quadrant multiplier: Both inputs can be bipolar.
- Two-quadrant multiplier: One input is unipolar and the other bipolar.
- One-quadrant multiplier: The two inputs are unipolar.

The characteristics of a multiplier are described in terms of its precision and its linearity.

- The precision of a multiplier represents the maximum deviation of the ideal multiplier output from the ideal transfer function.
- The linearity of a multiplier is measured as the maximum output deviation, relative to the line that best approximates the multiplier output curve with respect to one of the inputs, when the other is kept constant at its maximum value.

To process these analog signals, the circuit is often required to take two analog inputs to produce an output proportional to that of your product. When multiplying these two analog signals, the following elements that make up these signals must be taken into account:

1. The magnitude, 2. The frequency and 3. The phase.

An integrated circuit capable of performing the aforementioned operation is the AD633.

AD633
(CALDAS, 2021)

The AD633 is an analog signal multiplier integrated circuit, capable of performing operations in all 4 quadrants, with a 1 MHz bandwidth. It does not require external components and extensive calibration, with an error of 2% at full scale. Generate 0V with a buried Zener (Texas Instruments, 2015).

The following expression represents the output:

$$\boxed{W = \frac{(X1-X2)(Y1-Y2)}{10\ V} + Z} \quad (2)$$

Where:

W is the output of AD633

$X1, X2, Y1, Y2$ are input signals

Z is the OFFSET voltage

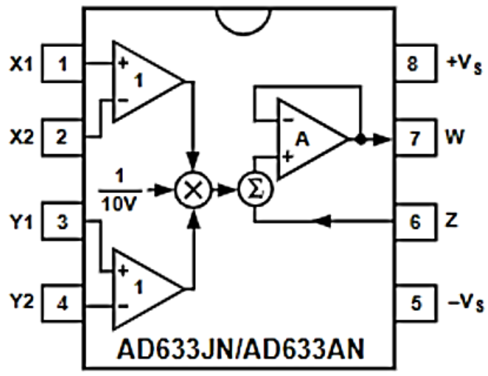

Figure 4 AD633 Block Diagram
Source: Datasheet AD633

Highlights

1. The AD633 is inexpensive. It is offered in 8-lead plastic packages.

2. It is stable and reliable due to its monolithic construction and laser calibration.

3. It allows high resistance values in MΩ, making the signal source load negligible.

4. The power supply voltages range from ± 8 V to ± 18 V. The internal scale voltage is generated by a stable Zener diode; multiplier precision is essentially supply insensitive.

Applications involving R wave detection

The pacemaker is an electronic device whose purpose is to make the heart beat, using electric shocks that replace the cardiac conduction system itself and guarantee a synchronous and efficient beat. Pacemaker placement depends on the presence or absence of significant symptoms or signs attributable to bradycardia. (Carrasco & Villeda, 2000)

Pacemakers use algorithms that allow detecting the time between each R wave. If the time that exists between each R wave corresponds to tachycardia, then electrical shocks are sent to reestablish normal heart rhythm.

Similarly, in the case of tachycardias, the defibrillator is available. Defibrillation is based on applying an electrical current abruptly and briefly to reverse rapid cardiac arrhythmias; situations in which the number of heartbeats increases excessively or there is disorganized electrical activity. As can be concluded, like pacemakers, defibrillators must be able to determine in a correct way that the patient is in tachycardia and for this they rely on algorithms for detecting the R wave to determine the heart rhythm and verify if it is in effect. have an arrhythmia or not.

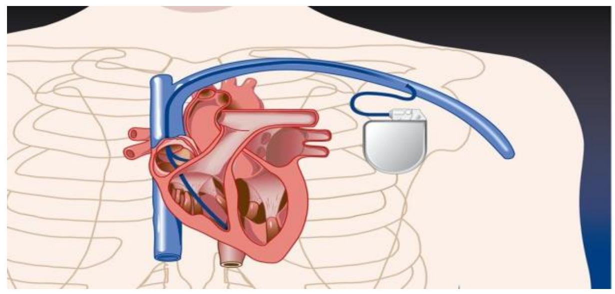

Figure 5 Schematic of an implantable pacemaker
Source: http://www.insuficiencia-cardiaca.com

Methodology

The study was divided into 5 phases: the first is the Database, where the vector of values from a real ECG study was collected. Additionally, a signal treatment was carried out to adapt it to admissible values for pulse width modulation (PWM).

The second stage is the Deployment of the data whereby means of the analog writing that the microcontroller performs using PWM and a second order RC low-pass filter we obtain the ECG signal.

The last three phases were the description and implementation of the Variant of the Pan-Tompkins algorithm where the band-pass filter was designed to clean the signal; ECG signal squared and voltage level assigned to achieve R wave detection.

Phase 1.- Database

The vector of ECG lead I values was obtained from the Physionet online database. A 10 second study was downloaded with a 2 millisecond sample time, this study contains too much data (5000 items).

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Therefore, it was decided to reduce the number of samples to 100. This was achieved considering only the first 2 seconds of the study, which correspond to 1000 data. Of these 1000 pieces of data, one element out of ten was assigned to a new vector that finally has 100 pieces of data.

It is important to mention that the values of the vector that generates the pulse width modulation (PWM) in the Arduino microcontroller, are integers that go from 0 to 255.

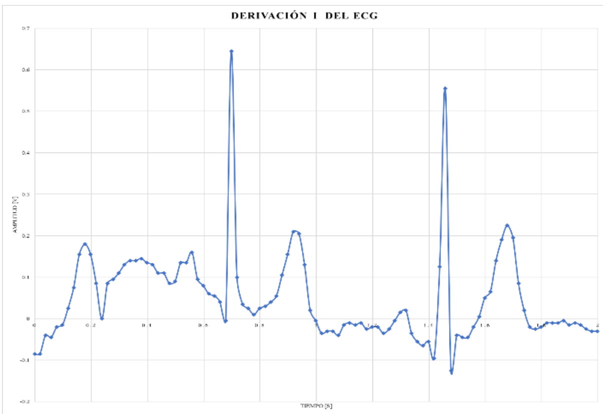

Figure 6 ECG lead I vector
Source: Own elaboration

This vector consists of decimal values that are both positive and negative, so a conditioning of the vector had to be performed to display the data on the scale from 0 to 255.

An OFFSET was applied to the entire signal, thus avoiding possible negative components that could not be displayed by the Arduino's PWM. For this, the most negative value was found and it was subtracted from the entire vector, resulting in the graph shown in figure 7.

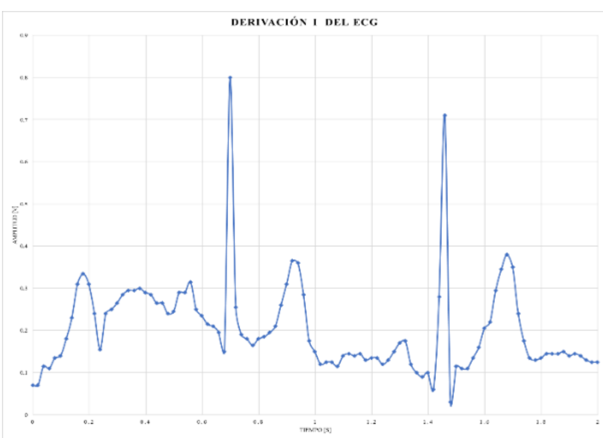

Figure 7 ECG Lead I without negative values
Source: Own elaboration

The maximum amplitude of the signal was made unitary, as can be seen in figure 8.

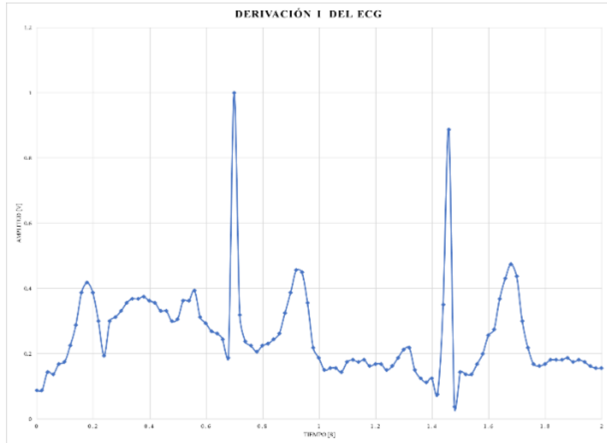

Figure 8 ECG vector with maximum amplitude of 1
Source: Own elaboration

Finally, this entire vector was multiplied by 255, with rounding to obtain whole numbers. It is this last vector, the one that can be declared in the Arduino board program, since it complies with the admissible values for the PWM, as shown in figure 9.

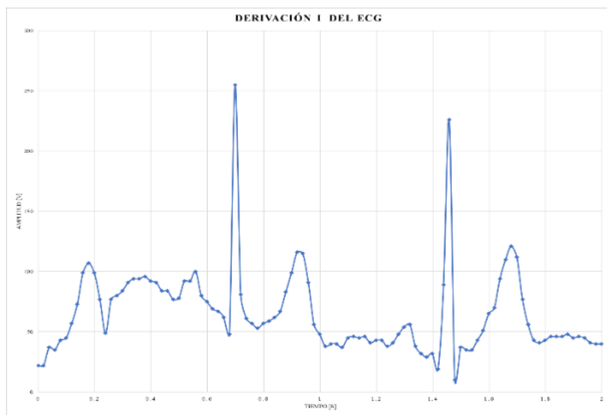

Figure 9 Lead I of the ECG with acceptable values for PWM (Integers ranging from 0 to 255)
Source: Own elaboration

Phase 2.- Deployment of the data

To display the data, the microcontroller's analog write with a timeout was used. The calculation of this wait was determined as follows:

$$delay = \frac{2000ms}{100\ datos} = 20\ ms/datos \tag{3}$$

For the signal period to be 2 seconds, there must be a wait between each of the 100 20-millisecond data.

To visualize lead I, we have to fit a second order RC low-pass filter since the PWM alone is nothing more than square waves.

The cutoff frequency will be, the number of cycles of the signal between the total period of the signal in seconds).

$$f_c = \frac{2\,ciclos}{2\,s} = 1\,Hz \tag{4}$$

For practical purposes a higher cutoff frequency will be used without affecting the integrity of the signal.

This frequency is determined by the values of capacitors and resistors with which the filter was built.

With 6.8 kΩ resistors and 1 μF capacitors, the corresponding cutoff frequency is:

$$f_c = \frac{1}{2\pi RC} = \frac{1}{2\pi(6.8\,k\Omega)(1\mu F)} = 23.4051\,Hz \tag{5}$$

In this way, the ECG lead I signal can be recovered in the simulation..

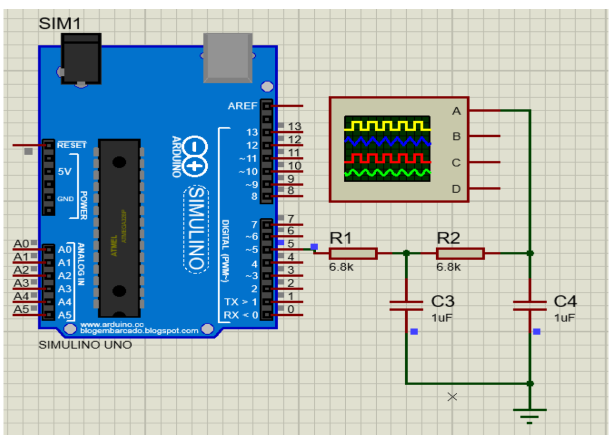

Figure 10 ECG lead I signal generator. It consists of the Arduino that generates the signal and a second-order RC low-pass filter
Source: Own elaboration

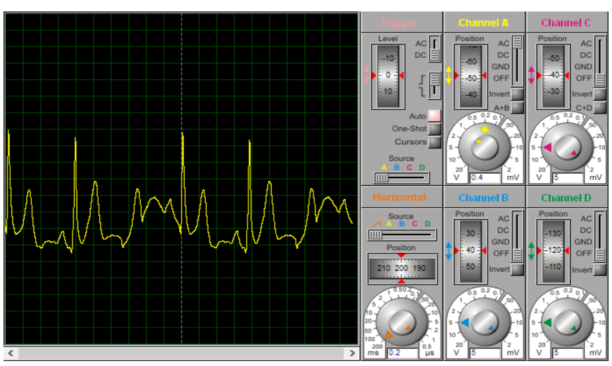

Figure 11 ECG lead I signal displayed on the oscilloscope
Source: Own elaboration

Phase 3.- Variant of the Pan-Tompkins algorithm

The Pan-Tompkins algorithm is a sequential process that allows us to detect the R wave of the ECG. This sequence of steps is: Step 1.- The band-pass filter eliminates any unwanted noise. Step 2.- A derivative is made to emphasize those parts of the signal where there are sudden changes in slopes. Step 3.- A squared elevation is performed to make all the components of the signal positive and further accentuate the results of the derivative (Tompkins, 1992).

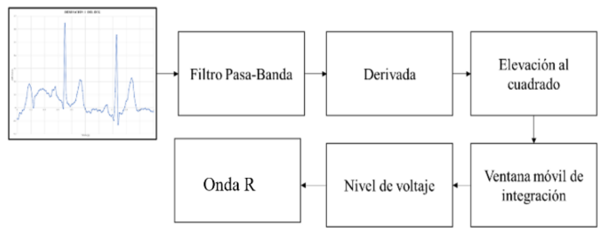

Figure 12 Pan-Tompkins algorithm
Source: Own elaboration

It is based on the accentuation of sudden changes in slope, emphasizing them thanks to the derivative and the squared elevation, giving the R wave a high amplitude with which it is easier to detect said wave. Although the derivative and the mobile integration window could be implemented, in the present methodology it was decided to omit these stages, in such a way that the variant of the Pan-Tompkins algorithm for the detection of the R wave is shown in figure 13.

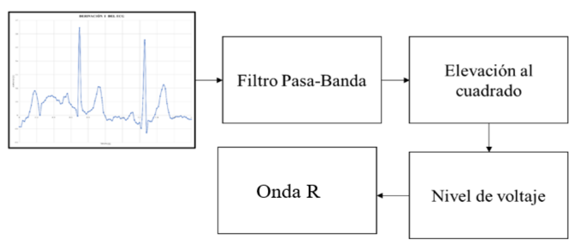

Figure 13 Variant of the Pan-Tompkins algorithm
Source: Own elaboration

Phase 4.- Implementation of the variant of the Pan-Tompkins algorithm: Design of a band-pass filter from 0.5 to 150 Hz.

To eliminate possible undesirable noise and clean the signal, a band-pass filter composed of the response of a high-pass filter in series with an active second-order low-pass filter MFB with cutoff frequencies ω_1 = 0.5 Hz and ω_2 was used. = 150 Hz respectively (Webster, 2010).

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These frequencies are at which an electrocardiogram operates (Webster, 2010).

Design of the active second-order high-pass filter MFB

Considering that we want to implement an approximation of the Butterworth type, whose quality factor Q = 0.7071, whose constant k = 1, a unity gain of the filter (| A | = 1.1), with a cut-off frequency f_c = 0.5 Hz and setting the value of the capacitors at C = 1 μF, we proceed to calculate the real values of the remaining elements:

Calculating the constant m

$$m = \frac{1+\sqrt{1+8Q^2(A-1)}}{4Q} \quad (6)$$

$$m = \frac{1+\sqrt{1+8(0.7071)^2((1.1)-1)}}{4(0.7071)} = 0.7719$$

For resistance R_1

$$R_1 = \frac{m}{2\pi k f_c C} \quad (7)$$

$$R_1 = \frac{0.7719}{2\pi(1)(0.5\,Hz)(1\times 10^{-6}F)}$$

$$\boldsymbol{R_1} = 2.457\times 10^5\Omega = 245.7\times 10^3\Omega = \boldsymbol{245.7k\Omega}$$

For resistance R_2

$$R_2 = \frac{R_1}{m^2} \quad (8)$$

$$\boldsymbol{R_2} = \frac{245.7k\Omega}{(0.7719)^2} = \boldsymbol{412.37k\Omega}$$

For resistance R_a

$$R_a = \frac{AR_2}{A-1} \quad (9)$$

$$\boldsymbol{R_a} = \frac{(1.1)(412.37k\Omega)}{(1.1)-1} = \boldsymbol{4536.07k\Omega}$$

For resistance R_b

$$R_b = AR_2 \quad (10)$$

$$\boldsymbol{R_b} = 1.1(412.37k\Omega) = \boldsymbol{453.61k\Omega}$$

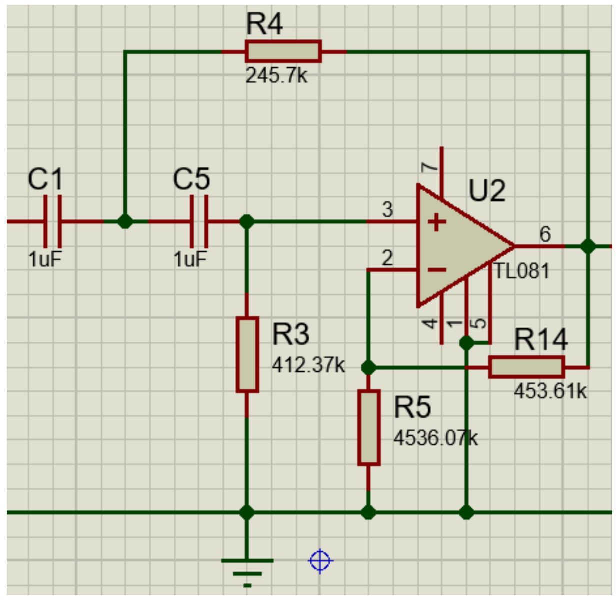

Figure 14 Sallen Key 2nd order active high-pass filter with cutoff frequency at 0.5 Hz and 1.1 gain
Source: Own elaboration

MFB second-order active low-pass filter design

Considering that it is desired to implement an approximation of the Butterworth type, whose quality factor Q = 0.7071, whose constant k = 1, a unity gain of the filter (| A | = 1), with a cut-off frequency f_c = 150 Hz and setting the capacitor value at C_1 = 1 μF, we proceed to calculate the real values of the remaining elements:

For resistance R_3

$$R_3 = \frac{Q(A+1)}{\pi k f_c C_1} \quad (11)$$

$$R_3 = \frac{0.7071(1+1)}{\pi(1)(150\,Hz)(1\times 10^{-6}F)}$$

$$\boldsymbol{R_3} = 3.001\times 10^3\Omega = 3.001k\Omega = \boldsymbol{3k\Omega}$$

For the capacitor C_2

$$C_2 = \frac{1}{4\pi k f_c Q R_3} \quad (12)$$

$$C_2 = \frac{1}{4\pi(1)(150\,Hz)(0.7071)(3\times 10^3\Omega)}$$

$$\boldsymbol{C_2} = 2.5009\times 10^{-7}F = 250.09\times 10^{-9}F = \boldsymbol{250\,nF}$$

For resistance R_1

$$R_1 = \frac{R_3}{A} \quad (13)$$

$$\boldsymbol{R_1} = \frac{3k\Omega}{(1)} = \boldsymbol{3k\Omega}$$

For resistance R_2

$$R_2 = \frac{R_3}{A+1} \qquad (14)$$

$$\boldsymbol{R_2} = \frac{3k\Omega}{(1)+1} = \frac{3k\Omega}{2} = \boldsymbol{1.5k\Omega}$$

For resistance R_4

$$R_4 = 2R_2 \qquad (15)$$

$$\boldsymbol{R_4} = 2(1.5k\Omega) = \boldsymbol{3k\Omega}$$

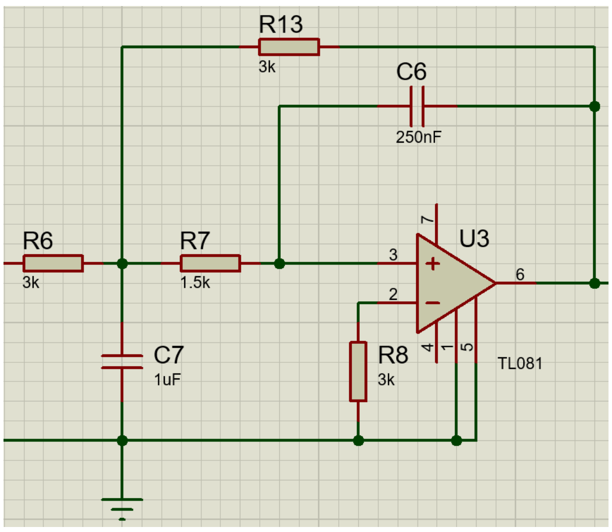

Figure 15 Active low-pass filter of 2nd order type MFB with cut-off frequency at 150 Hz and unity gain
Source: Own elaboration

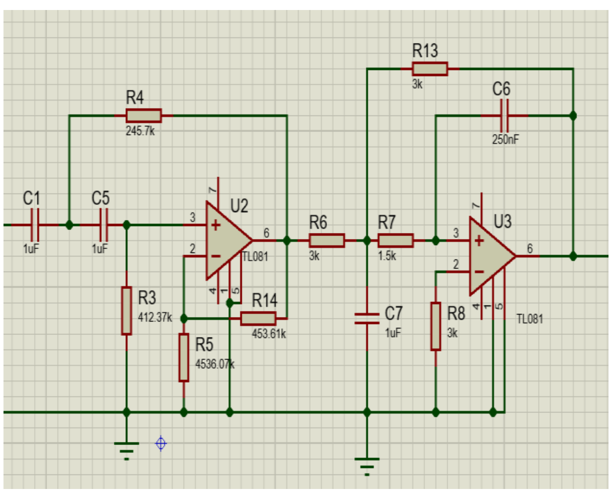

Figure 16 Band-pass filter composed of a cut-off frequency 0.5 Hz high-pass filter and a 150 Hz low-pass filter.
Source: Own elaboration

Phase 5.- Implementation of the variant of the Pan-Tompkins algorithm: Elevation squared and Voltage level

Once the filters that will have the desired band-pass filter behavior have been designed, it remains to square the signal and assign a voltage level to achieve the detection of the R wave.

Due to the transfer function of the low-pass filter, the ECG signal is inverted. However, since the next step in the variant of the Pan-Tompkins algorithm involves squaring the signal, it is well possible to proceed with the output of the filters without inverting it. In this case the signal is inverted as shown in the following figure.

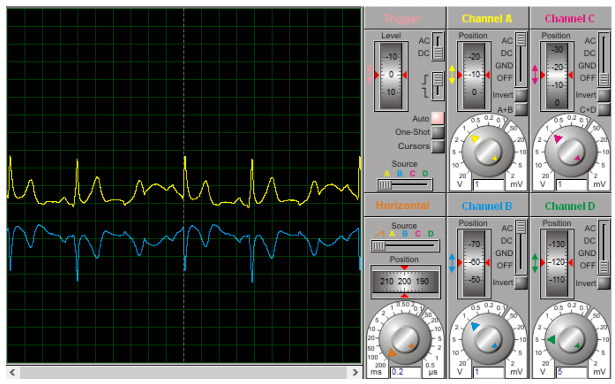

Figure 17 ECG lead I signal displayed on the oscilloscope (yellow) and signal resulting from applying the band-pass filter (blue).
Source: Own elaboration

To square the signal, use was made of the AD633 analog multiplier, whose output voltage equation is given by:

$$W = \frac{(X1-X2)(Y1-Y2)}{10\,V} + Z \qquad (16)$$

Where:

W is the output of the AD633

X1, X2, Y1, Y2 are input signals

Z is the OFFSET voltage

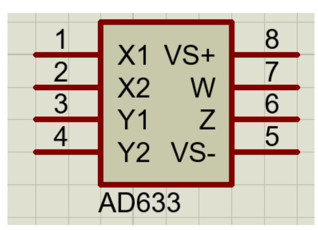

Figure 18 AD633 in Proteus
Source: Own elaboration

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Since we want to square the signal, inputs X1 and Y1 will be the same, while X2, Y2 and Z will go to ground.

$$W = \frac{(X1)(Y1)}{10\ V} = \frac{(X1)(X1)}{10\ V} = \frac{X1^2}{10\ V} \tag{17}$$

According to the AD633 data sheet to polarize the AD633, it is necessary to supply with +12 V for V_{s+} and -12 V for V_{s-}. Both sources must go in series with a 10 µF capacitor.

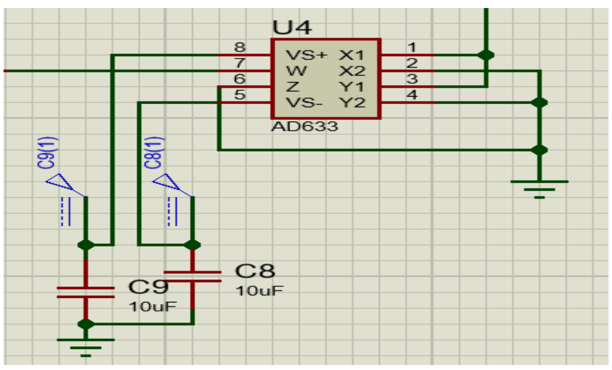

Figure 19 AD633 setup to square the signal.
Source: Own elaboration

One option to simulate the AD633 is to select SIMPLE instead of AD633 (by default). This option guarantees accuracy and precision in simulation tests.

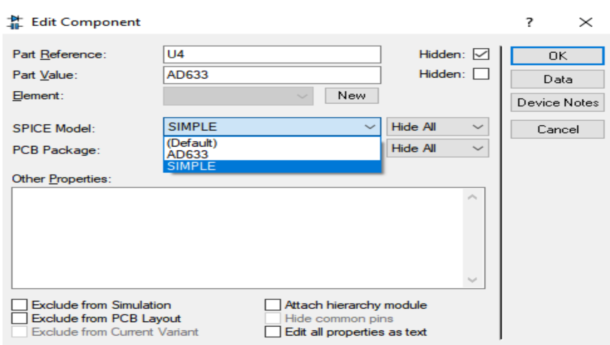

Figure 20 AD633 SPICE Model configuration
Source: Own elaboration

With these considerations, it was possible to obtain the squared ECG signal at the output. However, as seen earlier in the AD633's output voltage equation, the product is being divided by 10, so it is necessary to amplify the signal 10 times to compensate for this factor.

An op amp in non-inverting configuration or a pair of op amps can be used as inverters, in this case a pair of op amps was used as inverters.

Design of operational amplifiers in inverter configuration

The voltage gain is given by

$$G = -\frac{R_f}{R_i} \tag{18}$$

The gain of the first OpAmp in inverter configuration is -5, if R_i = 1kΩ, then:

$$\boldsymbol{R_f} = 5R_i = 5(1k\Omega) = \boldsymbol{5k\Omega}$$

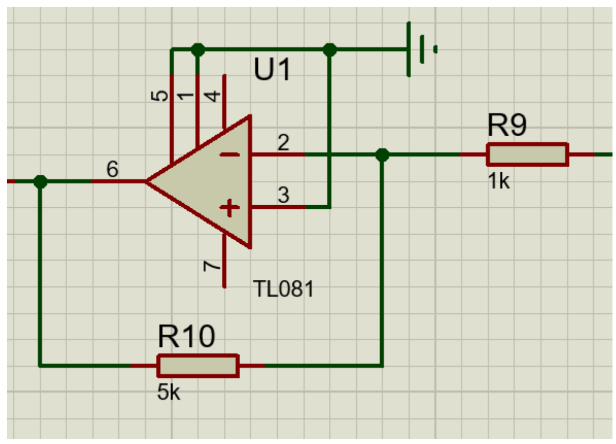

Figure 21 OpAmp in inverter configuration and -5 gain
Source: Own elaboration

The gain of the second OpAmp in inverter configuration is -2, if

R_i = 1kΩ, then we proceed to calculate R_f:

$$\boldsymbol{R_f} = 2R_i = 2(1k\Omega) = \boldsymbol{2k\Omega}$$

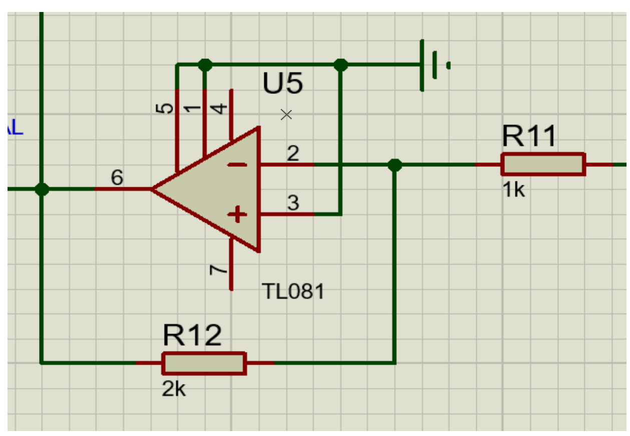

Figure 22 OpAmp in inverter configuration and gain of -2
Source: Own elaboration

The joint action of both operational amplifiers, in inverting configuration, results in an amplified signal 10 times, which compensates the factor of the output equation of the AD633.

The results of squaring the ECG signal are shown below in Figures 23 and 24:

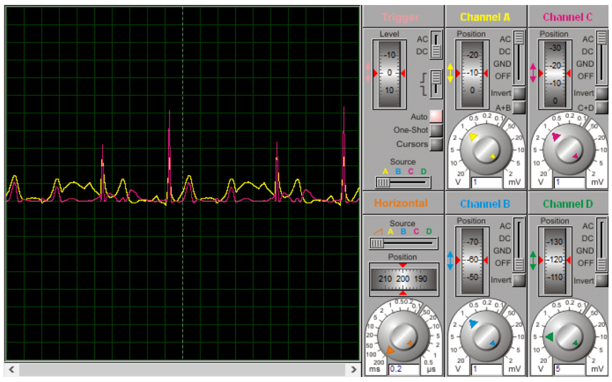

Figure 23 ECG lead I signal displayed on the oscilloscope (yellow) and ECG signal after filtering squared (magenta)
Source: Own elaboration

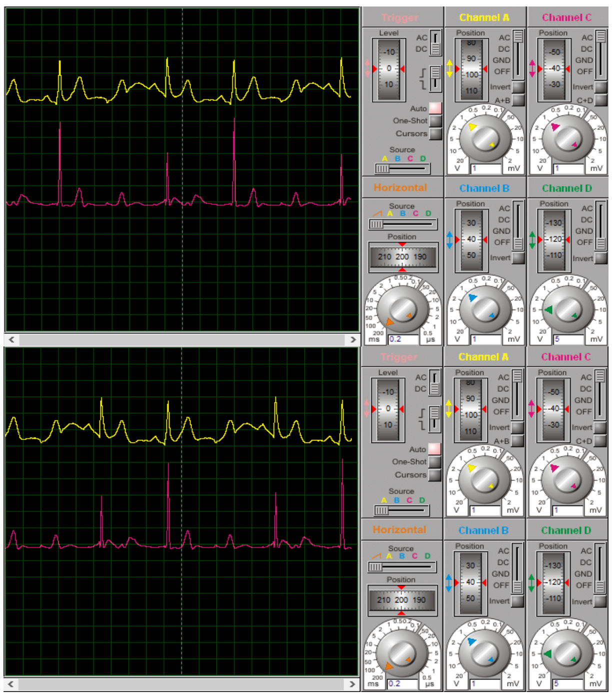

Figure 24 ECG lead I signal displayed on oscilloscope (yellow) and ECG signal after filtering squared (magenta)
Source: Own elaboration

As the last step of the variant of the implemented algorithm, a voltage threshold is assigned such that it is able to discriminate those peaks that are not of our interest and to identify only the one corresponding to the R wave. To achieve this purpose, an operational amplifier was used in comparator configuration. With this setup, you are constantly comparing the value of the squared ECG signal against a reference voltage. As a consequence, the OpAmp will output positive saturation in those parts of the signal where the threshold is exceeded (as expected from the R wave) and will return negative saturation to the output in those parts of the signal where the voltage of reference is higher.

The voltage being compared against can be varied with a potentiometer to achieve different results. The results of having considered our reference voltage of 9 Volts are the following: (see figure 25)

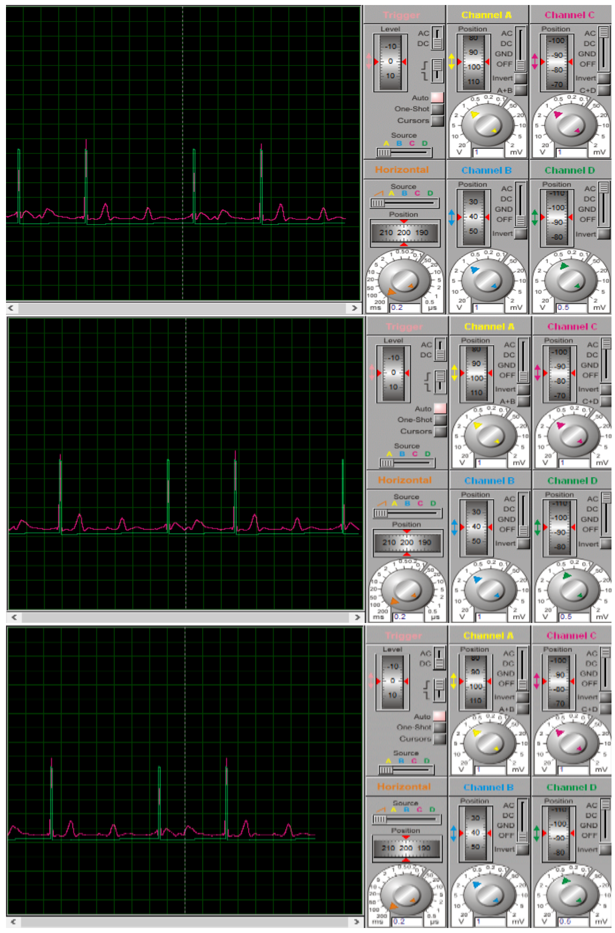

Figure 25 ECG signal after filtering squared (magenta) and comparator output (green). Oscilloscope at 0.2 seconds per division.
Source: Own elaboration

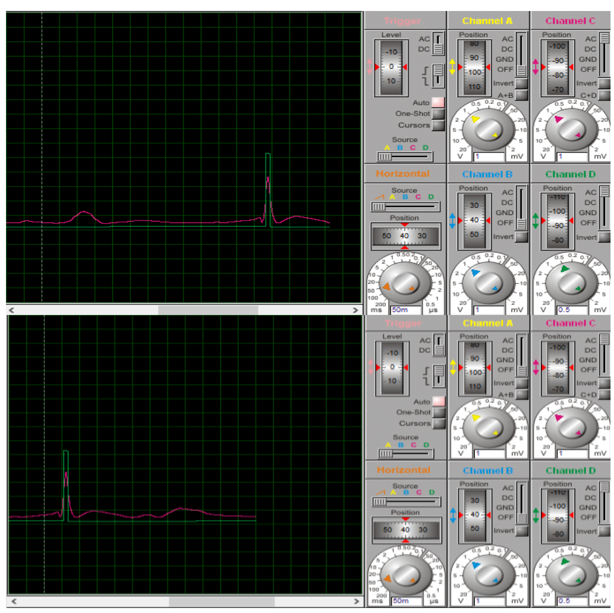

Figure 26 ECG signal after filtering squared (magenta) and comparator output (green). Oscilloscope at 50 milliseconds per division
Source: Own elaboration

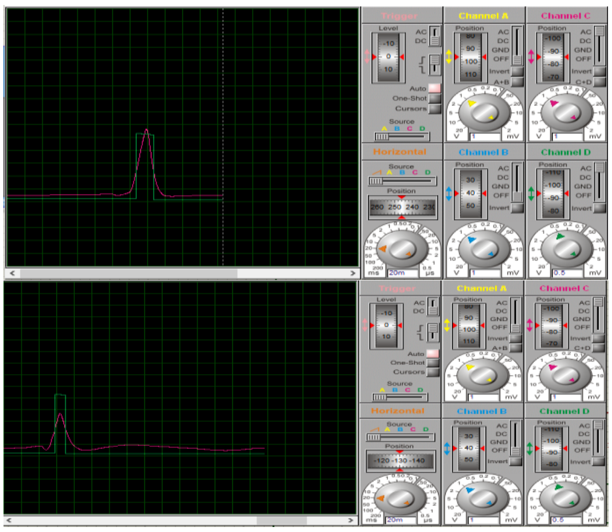

Figure 27 ECG signal after filtering squared (magenta) and comparator output (green). Oscilloscope at 20 milliseconds per division.
Source: Own elaboration

R wave detection

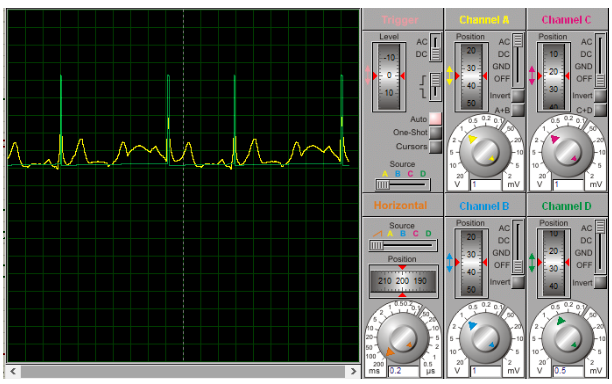

Figure 28 Results displayed on the oscilloscope, where the detection is made by the comparator amplifier (green), which “encloses” the R wave of lead I of the ECG (yellow)
Source: Own elaboration

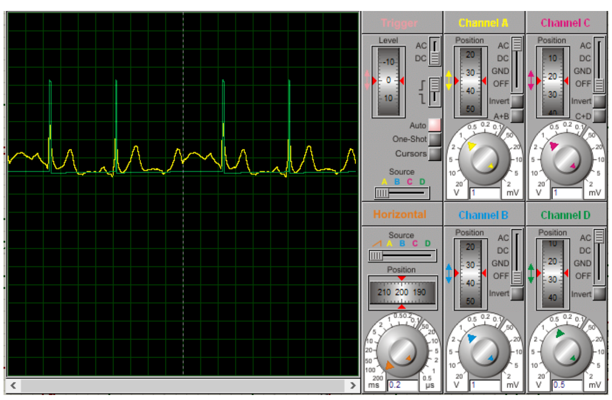

Figure 29 Results displayed on the oscilloscope, where the detection is made by the comparator amplifier (green), which “encloses” the R wave of lead I of the ECG (yellow)
Source: Own elaboration

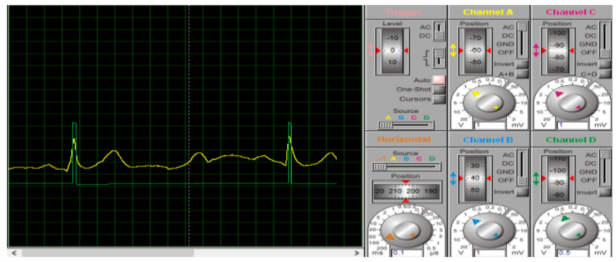

Figure 30 Results displayed on the oscilloscope, where the detection is made by the comparator amplifier (green), which “encloses” the R wave of lead I of the ECG (yellow)
Source: Own elaboration

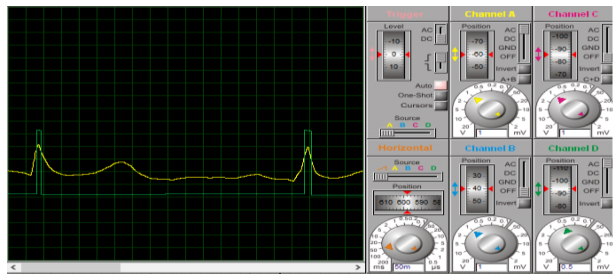

Figure 31 Results displayed on the oscilloscope, where the detection is made by the comparator amplifier (green), which “encloses” the R wave of lead I of the ECG (yellow)
Source: Own elaboration

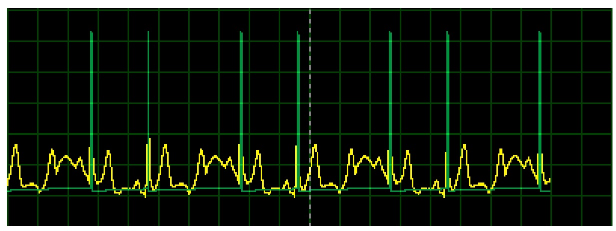

Figure 32 Results displayed on the oscilloscope, where the detection is made by the comparator amplifier (green), which “encloses” the R wave of lead I of the ECG (yellow)
Source: Own elaboration

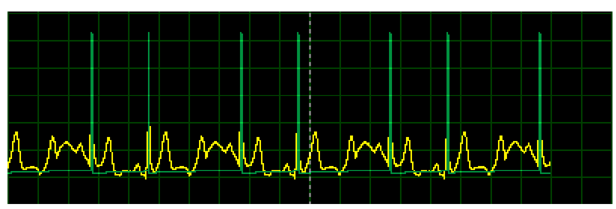

Figure 33 Results displayed on the oscilloscope, where detection is carried out by the comparator amplifier (green), which “encloses” the R wave of lead I of the ECG (yellow)
Source: Own elaboration

Results

The results of the simulations are correct and coincide with what was expected. The oscilloscope of the simulator shows how it is possible to detect the R wave of the lead I signal of the ECG.

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With regard to the detection of the R wave in a real ECG study for lead I, this methodology can be physically implemented with the previously described analog elements, obtaining results in accordance with expectations and therefore, a correct functioning of the detector.

Conclusions

Taking as a basis the Pan-Tompkins algorithm for the detection of the R wave was a very useful tool since it is a methodology that has been developed previously, achieving favorable results in the multiple studies that have implemented it.

A poor implementation of the R wave detector, at the simulation level, does not represent any risk. However, physically the detector may lack precision and miss that the patient is suffering from an arrhythmia, which leads to more serious complications such as fainting, cardiac arrest and eventually death.

The results obtained with the simulations show a correct behavior and an ideal operation of the R wave detector for lead I.

Developing this work for multiple derivations and verifying that the obtained corresponds to what is expected, allows the general validation of the Pan-Tompkins algorithm. that presents good results regardless of the derivation with which it is being treated, but it is worth corroborating this variant of the algorithm implemented with analog elements in different derivations to verify its effectiveness, precision and accuracy.

It is feasible in the future, the inclusion of analog elements to achieve the detection of the R wave and thus estimate the heart rate of the heart. The determination of the heart rhythm has applications for example in the pacemaker or in equipment such as the defibrillator.

The inclusion of detectors built with analog elements would achieve a possible reduction in the price of these medical devices, since the use of microcontrollers for heart rhythm determination tasks would be neglected.

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