Development of a Mobile Application for Measuring Heart Rate from the ECG Signal

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Trabajo de Grado para optar al título de Ingeniero Electrónico

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Content

List of figures

List of tables

Resumen

Título: Desarrollo de una Aplicación Móvil para la Medición del Ritmo Cardíaco a partir de la Señal ECG *

Autor: Yarlin Daniela Gelvez Rodriguez, Elmer Eduardo Rocha Jaime **

Palabras Clave: Aplicación Móvil, ECG, Ritmo Cardíaco, USB OTG, Variabilidad del Ritmo Cardíaco.

Descripción: En los últimos años, el uso de aplicaciones móviles y teléfonos inteligentes se ha convertido en parte de la vida cotidiana de las personas. Del mismo modo, ha habido un aumento sustancial en la tasa de mortalidad por enfermedad cardíaca. Teniendo en cuenta lo anteriormente mencionado, en este artículo se desarrolla una aplicación móvil que mide el ritmo cardíaco (RC) y calcula la variabilidad del ritmo cardíaco (VRC) a partir de una señal electrocardiográfica (ECG) de una manera no invasiva a través de tres electros superficiales. La señal ECG analógica es amplificada y filtrada con un circuito que se diseñó, es llevada a un smartphone basado en Android a través de un microcontrolador (MCU) con el protocolo USB On-The-Go (OTG) y mostrada en una gráfica en tiempo real en la aplicación desarrollada, todo el sistema es alimentado por una pequeña batería LiPo de 3.7V. La aplicación desarrollada entrega un documento al finalizar cada medición con el ritmo cardíaco medido de manera clínica, el ritmo cardíaco promedio y los parámetros de variabilidad en el dominio de tiempo, como la desviación estándar de todos los intervalos RR (SDNN), la raíz cuadrada de la media de la suma de las diferencias entre intervalos elevados al cuadrado RR (RMSSD) y el porcentaje de diferencias mayores de 50 milisegundos entre intervalos RR normales (pNN50).

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* Trabajo de grado

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Abstract

Title: Development of a Mobile Application for Measuring Heart Rate from the ECG Signal *

Author: Yarlin Daniela Gelvez Rodriguez, Elmer Eduardo Rocha Jaime **

Keywords: ECG, Heart Rate, Heart Rate Variability, Mobile Application, USB OTG.

Description: In recent years, the use of mobile applications and smartphones has become part of people's everyday life. Likewise, there has been a substantial raise in the mortality rate for heart disease. Considering the previously stated, this paper aims at developing a mobile application that measures the heart rate (HR) and calculates the heart rate variability (HRV) from an electrocardiographic (ECG) signal in a non-invasive way through three surface electrodes. The analog ECG signal is amplified and filtered with a circuit that was designed and carries it to an Android-based smartphone via a microcontroller (MCU) using the USB On-The-Go (OTG) protocol and displays it on a real-time graph in the developed application. The entire system powered by a small 3.7V lithium polymer (LiPo) battery. The application delivers a document at the end of each measurement with the clinically measured heart rate, the average heart rate and variability parameters in the time domain such as the square root of the mean of the sum of the squares of differences between adjacent RR intervals (RMSSD), the standard deviation of normal to normal RR intervals (SDNN), and the percentage of consecutive by the total number RR intervals that differ over 50 milliseconds from each other (pNN50).

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* Bachelor Thesis

** Facultad de Ingenierías Físico-Mecánicas. Escuela de Ingenierías Eléctrica, Electrónica y Telecomunicaciones. Director: Jorge Eduardo Quintero Muñoz, Esp en Telecomunicaciones. Co-Director: Jaime Guillermo Barrero Pérez, Msc en Ingeniería Electrónica.

Introduction

Heart rate measurement allows to keep track of an individual's heart health. A period of the normal electrocardiographic signal shown in Figure 1, shows the P wave (atrial depolarization), the QRS complex (ventricular depolarization), the T wave (ventricular repolarization), and rarely the U wave, as well as the PR intervals, QT (ventricular depolarization and repolarization) and the ST segment (ventricular repolarization) Tan et al. (2000). In electrocardiogram studies the variation in time between RR intervals is the variability of heart rate, it uses the study of this measure as an indicator of normal and pathological cardiovascular function, etc. There are different methods for measuring HRV, such as time-domain methods (statistical and geometric), frequency-domain methods (Spectral components), and nonlinear methods Malik (1996).

Processing the ECG signal determines heart rate (HR) and heart rate variability (HRV), this allows to diagnose arrhythmias or heart abnormalities.

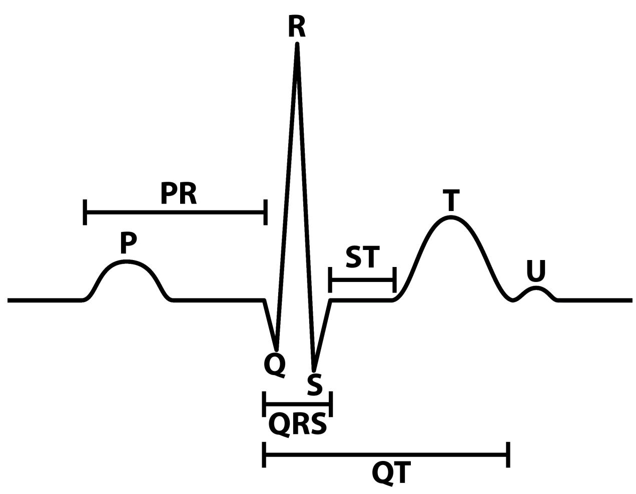

Figura 1. Typical signal of an ECG

A study Jeanne et al. (2009) suggests that the acquisition of the RR series for the measurement of HRV using an R wave detection algorithm. As a result, they get the ECG signal recording and automatic calculation of the RR series in real-time. Cardiac arrhythmias constitute a significant group within cardiovascular diseases (CDVs), it is estimated that 17.9 million people died in 2016 from CDVs, which represents 31 % of all deaths recorded globally Organization (2017). According to this, it is possible to conclude that people are not entirely aware of CDVs, therefore the need to take advantage of new technologies to identify a potential risk of heart problems. Nowadays the use of smart devices has become increasingly popular. There are several types of portable equipment that can measure heart rate by using sensors. For example, smartwatches (Galaxy Watch Samsung (2018)) or sports wristbands (Mi Smart Band Xiaomi (2019)). However, the heart rate monitoring of these devices is not always accurate. Furthermore, an image of the ECG signal would be much more valuable. There are also devices for clinical environments that capture and expand the electrical activity of the heart through electrodes (SE-12 Express EDAN (2011)) yet these devices are large, expensive and difficult to handle. In this paper, a mobile application was developed to measure the heart rate from the ECG signal using a USB OTG interface similar to that made in Tsai and Ko (2018), where they communicated an electromyographic signal detection (EMG) system with a smartphone through a microcontroller (MCU). In our work, to obtain the ECG signal we used a three surface electrodes in the right arm, left arm and lower abdomen, this topology is known as lead I, in Toinga et al. (2017) the union of the three bipolar leads (I, II, and III) form an imaginary triangle known as the Einthoven triangle, that allows the normalization of the electrocardiographic records.

Figure 2 presents the proposed system for the acquisition of the ECG signal. It comprises four key parts: an analog circuit, a MCU, a power supply and an Android-based smartphone. The signal is obtained from electrodes, then amplified and filtered, then transmitted to an MCU that digitizes it to a sample and quantifies it. In the smartphone, we developed an application program (app) for the ECG signal processing with a graphical and interactive interface. The system is powered by a power circuit from a small LiPo battery.

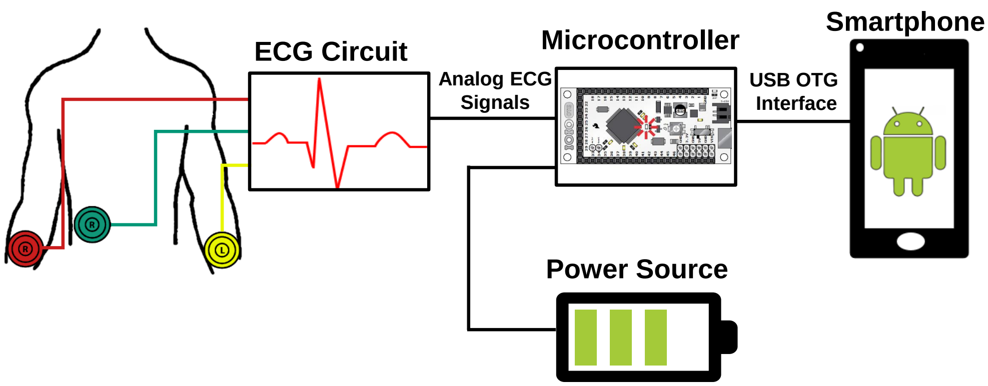

Figura 2. Acquisition system of the ECG signal

This paper is structured as follows: section II discusses analog filter design and signal amplification. Section III implements the USB OTG communication interface between an Android-based smartphone and a MCU. Section IV develops the algorithm for interpreting and calculating HR and HRV. Section V shows the design of the mobile application. Section VI explains the power source. Section VII shows the results. Section VIII discusses the results and provides some ideas for further research. Finally, Section IX concludes this work.

1. Analog filtering and signal amplification circuit

Throughout a pair of surface electrodes connected to the body we get an electrical signal that varies over time with the heartbeat, this signal has a maximum amplitude in the order of 1 mV Dharneeshkar and Vanitha (2019). However, there is an important source of electrical power interference that flows through the impedances of the electrodes and the skin on the way to the ground. If the right leg is connected to the common ground of the amplifier, the noise of 50-60 Hz appears as an input in the order of mV, but the ECG signal is a differential voltage, then this noise voltage is common to each electrode, therefore the rejection depends on the amplifier's ability to reject common-mode voltages.

To choose the elements that were used in the analog circuit, the comparison was made between other amplifiers shown in the following Table 1.

Table 1
Comparison of amplifier characteristics.

Amplifiers	CMRR [dB]	Bandwidth [MHz]	Noise at 1 kHz $[nV/\sqrt{Hz}]$
INA326	100	0.001	33
INA128	120	1.3	8
TL084	86	3	18
OPA2335	130	2	60
OPA2132	100	8	8

We chose these devices mainly because the INA326 and OPA2335 amplifiers operate from a single 5V supply. Even though some amplifiers have better parameters, they were discarded because of their supply voltage and the fact that these do not have medical instrumentation applications.

The frequency spectrum containing electrocardiography signals under normal and pathological conditions ranges from 0.05 to 100 Hz, most electrocardiography equipment for clinical diagnostic studies uses this bandwidth, even some equipment reaches up to 150 Hz. However, when heart rate studies are conducted in isolation, the bandwidth is usually between 0.5 and 40 Hz. In our case, we use a bandwidth between 0.05 to 100 Hz because it made possible to calculate the HRV Kiing-Ing Wong (2009).

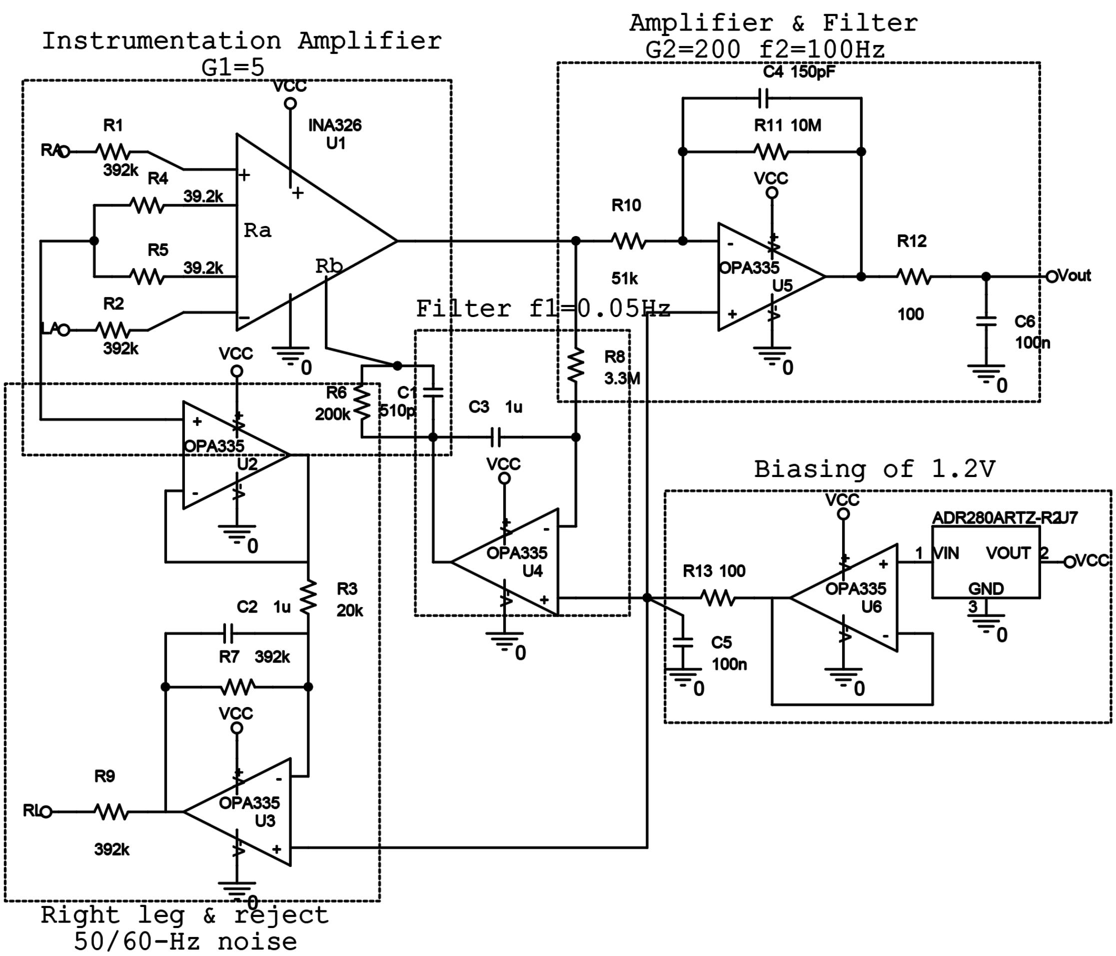

Figura 3. ECG analog circuit based in Thomas (2005)

In Thomas (2005) designed a portable high-precision analog circuit that requires only one power supply; our circuit follows this design as shown in Figure 3. The first stage performs a pre-amplification using an instrumentation amplifier that has a high input impedance and a high common-mode rejection ratio (CMRR), the gain for this stage is set to 5 using Eq. (1). In the next stage, the signal is filtered allowing only frequencies from 0.05 Hz Eq. (2). The third stage amplifies and filters the signal with a gain of 200 Eq. (3),(4), to get a total gain of 1000 Eq. (5) in the system and a bandwidth to 100 Hz. At the protection stage, a right leg safety circuit, a circuit to reject common mode voltage and 50-60 Hz noise is implemented. The last stage adapts the signal by adding a voltage DC (offset) to the analog ECG signal which is done by the constraints of the ADC.

$$G_1 = 2\left(\frac{Rb}{Ra}\right) \tag{1}$$

$$f_1 = \frac{1}{2\pi \cdot R8 \cdot C3} \tag{2}$$

$$G_2 = \frac{R11}{R10} \tag{3}$$

$$f_2 = \frac{1}{2\pi \cdot R11 \cdot C4} \tag{4}$$

$$G_s = G_1 \cdot G_2 \tag{5}$$

Figure 4 shows the final result of the circuit designed as a printed PCB board.

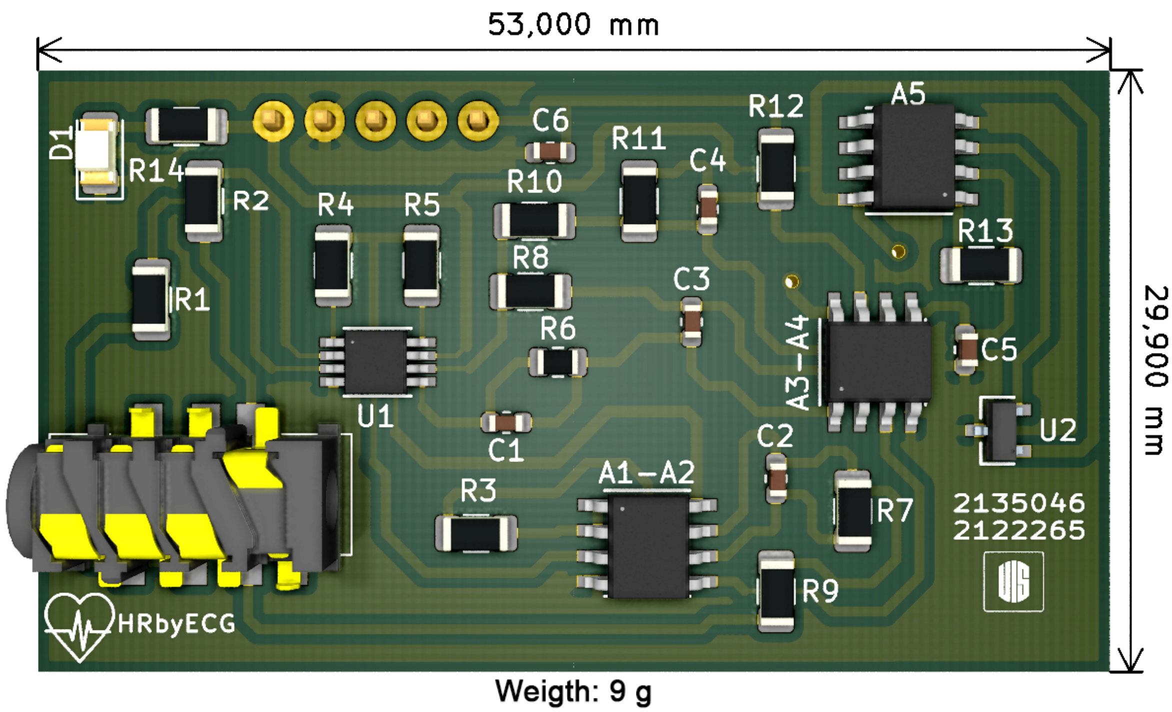

Figura 4. ECG circuit PCB board

2. USB OTG communication

Raspberry Pi, Arduino, and ESP32 are a series of commonly used of boards because of their simplicity, each of these boards have their limitations and purposes. The ESP32 is usually low cost and power-efficient; it has wireless connectivity (WiFi/Bluetooth) and is designed for portable applications and the Internet of Things (IoT) among other features. However, we used the IOIO OTG v2.2 board (Figure 5) which is specially designed to work with Android devices, this board is open hardware, it has an MCU PIC Microchip (2010) that can connect to an Android device using the USB OTG protocol, which needs an Android version higher than 4.4, the board has general-purpose input/output pins (GPIO) of 3.3V and some pins support up to 5V, also has ADC pins of 0-3.3V 10-bits, even supports Bluetooth adding an adapter.

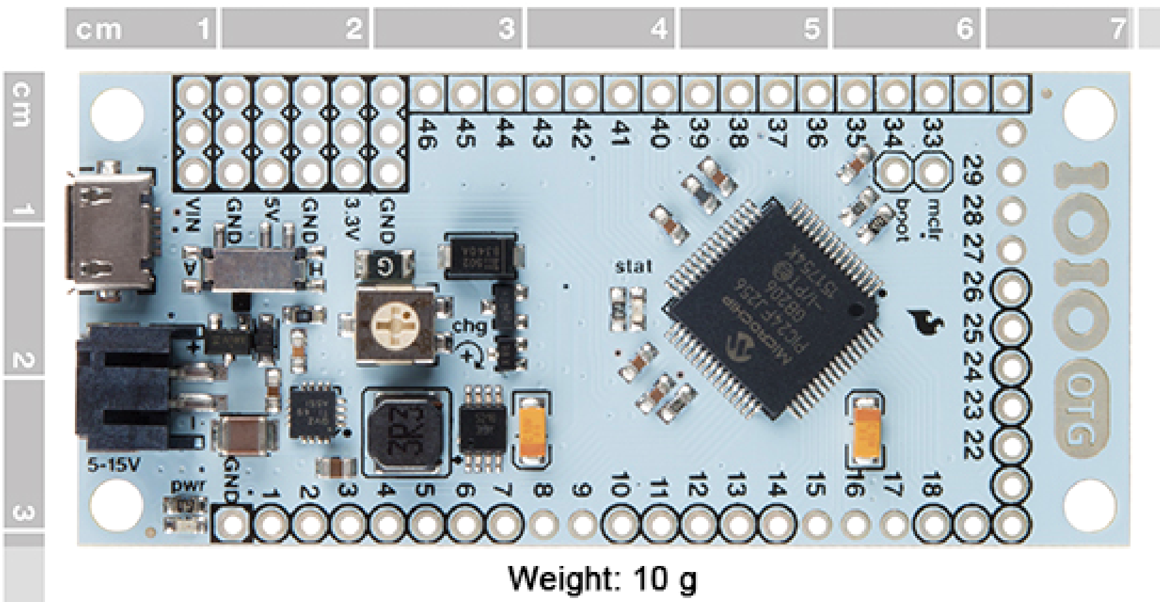

Figura 5. IOIO-OTG-v2.2. Taken from SparkFun (2011)

Regarding the communication between the board and the smartphone, the board's manufacturer SparkFun (2011) explains that IOIO-OTG acts as a USB host and supply charging current to the device. Thus, IOIO-OTG needs its own power supply, the USB host can close the bus to save power. When the board is inactive, this communication is bidirectional, the smartphone sends orders to the board which sends it back data from its GPIO pins. In our case, the IOIO board takes the analog reading of the ECG signal and sends it to the digitized smartphone.

3. Algorithm for graph ECG, calculate HR and HRV

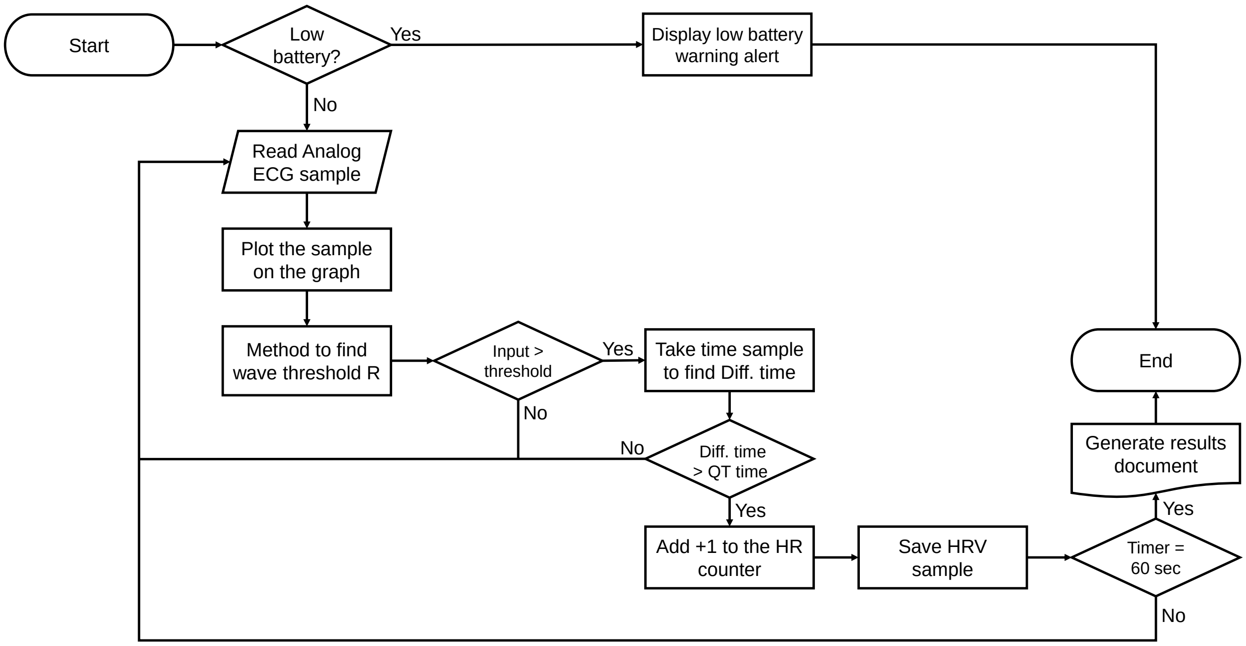

Figura 6. Flow diagram of calculate HR and HRV algorithm

Figure 6 shows the flow diagram that runs in the application. The first thing to do is check the battery voltage level, if it has a low level the app shows an alert on the screen and end the execution if the voltage level is adequate, the analog value of the ECG signal is read via an ADC of the board, this read value is graphed in the application interface, with 200 samples the wave average is calculated, to this average a percentage is added to find a threshold, as shown in Figure 7, for this reason, it is necessary to graph 2 QRS complexes before starting the measurement to have both a threshold and time reference.

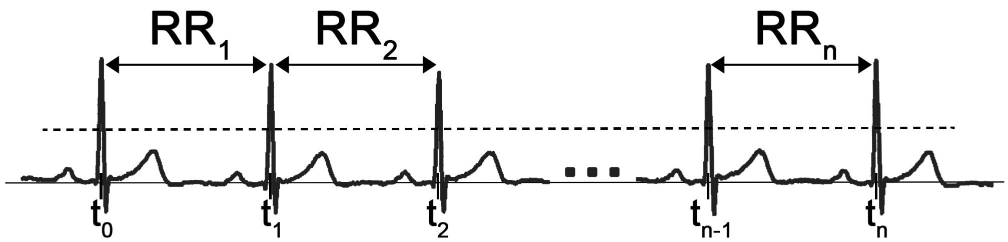

Figura 7. Time RR intervals

The current sample is compared with the threshold, to find a peak value of the R wave, when this value is obtained, a time sample is taken (t_{n-1}), to determine the time until the next sample greater than the threshold (t_n), if the determined time ($RR_n = t_n - t_{n-1}$) is greater than the average duration time of the QT segment ($RR_n > t_{QT}$), +1 is added to the pulse counter to calculate HR, and this time is saved as a sample of the RR segment for HRV measurement. For this reason, the algorithm works best with ECG signals that have the R wave magnitude greater than the T wave magnitude. All the above processes are performed every 5 ms (by Nyquist theorem) until the 60-second timer expires.

For the measurement of frequency variability we use the method in the time domain with the following parameters: the mean of the RR interval ($\overline{RR}$), the standard deviation of normal to normal RR intervals (SDNN), the root mean square of successive RR interval differences (RMSSD), and the percentage of consecutive RR intervals that differ over 50 milliseconds from each other (pNN50) Capdevila Ortís et al. (2008).

To measure these parameters, we use the following standard equations defined in Tarvainen et al. (2009), Eq. (6) represents the mean value of the RR intervals where N is the number of measured RR intervals and RR_n is the value of the n-interval.

$$\overline{\text{RR}} = \frac{1}{N}\sum_{n=1}^{N} \text{RR}_n \tag{6}$$

Eq. (7) calculates the heart rate from the average of RR intervals. In Eq. (8) SDNN reflects the standard deviation of the overall variation (both short and long term). The RMSSD parameter of Eq. (9) associates the measurement of short-term variability (beat per beat). The pNN50 of Eq. (10) value provides information about spontaneous variations in heart rate.

$$\overline{\text{HR}} = \frac{60}{\overline{\text{RR}}} \tag{7}$$

$$\text{SDNN} = \sqrt{\frac{1}{N-1}\sum_{n=1}^{N} (\text{RR}_n - \overline{\text{RR}})^2} \tag{8}$$

$$\text{RMSSD} = \sqrt{\frac{1}{N-1}\sum_{n=1}^{N-1} (\text{RR}_{n+1} - \text{RR}_n)^2} \tag{9}$$

$$\text{pNN50} = \frac{\text{NN50}}{N-1} \times 100\,\% \tag{10}$$

After completing the one-minute measurement, a document is delivered with the graph signal on intervals, the clinical heart rate (HR), and the heart rate from HRV ($\overline{\text{RR}}$), the maximum and minimum values of the RR segments, the RMSSD, SDNN, and pNN50 parameters.

4. Mobile application

We developed the application in the official Android IDE: Android Studio, where the application its two extensions are mainly worked, the extension .xml and .java. The IOIO board includes a GitHub repository needed to develop Android applications.

The app's interface was designed to be comfortable and user friendly. Also, information is provided to facilitate the interpretation of the ECG signal and give a diagnosis.

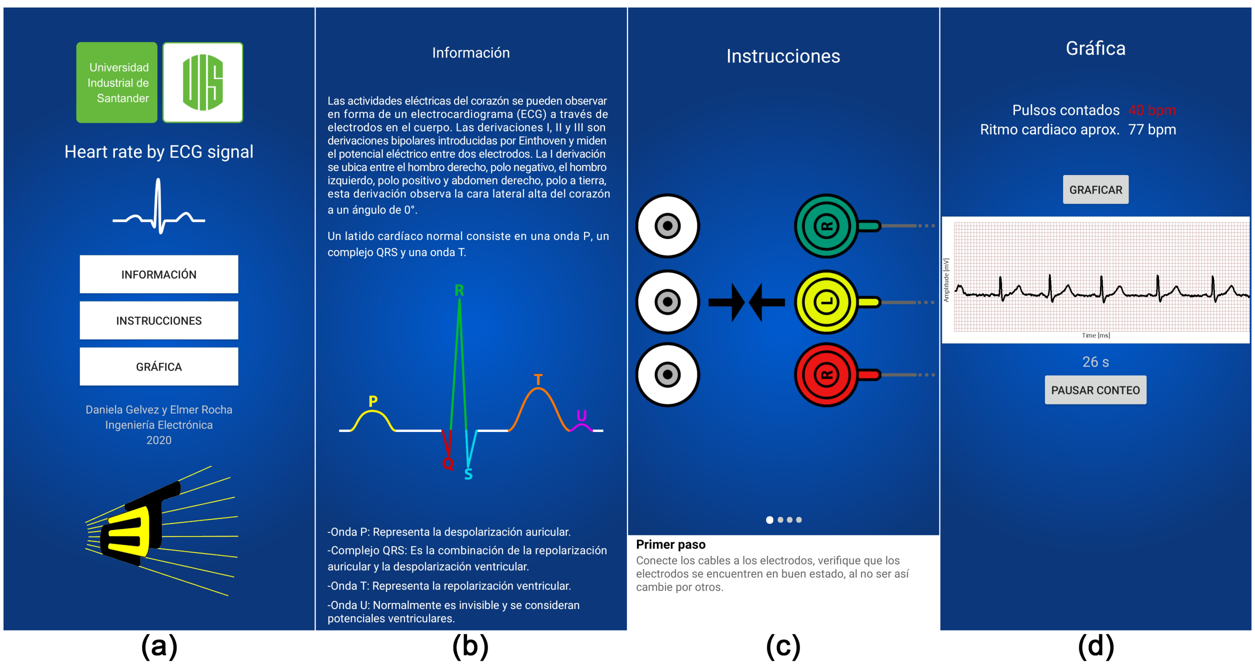

Figura 8. Visual design of the application: (a) Home screen; (b) Information; (c) Instructions; (d) Graph

Figure 8 shows the application interface with its respective activities; (a) displays the app's home screen with a 3-button-menu to go to basic information, instructions for connecting the device, and a graph to get the ECG signal. In (b), it gives information regarding ECG and HR.

Whereas (c) instructs how to connect the device. Finally, (d) is the activity where the ECG signal is graphed, HR and HRV are measured. The user gets a diagnosis based on its basic data, generating a PDF file that can be shared through email, social networks, or cloud drives. Also, the measured data is generated in text format .csv. An example of the results document is shown in Figure 9.

Figura 9. Results document sample

5. Power Source

As the IOIO board needs to operate in USB host mode, it requires an external power source. Therefore, the choice of the 3.7V LiPo battery was based on the area, weight, capacity, and type of battery. Although there are other 3.7V batteries with higher capacity, these are heavier and have a larger area. Even 7.4V batteries with 10000mAh are too large (70mm x 35mm x 18mm), heavy (85g) and they require a special charger. Small 6V (4LR44) batteries alkaline were also discarded because they were not rechargeable and had a greater environmental impact.

Table 2 shows the characteristics of the battery choice.

Table 2
Battery characteristics.

3.7V LiPo battery	Characteristics
Capacity	600mAh
Nominal Voltage	3.70V
Charge Limited Voltage	4.20V
Discharge Cut-off Voltage	2.80V
Dimensions	9mm x 20mm x 35mm
Weight	15g

However, the IOIO board requires to be powered with a minimum of 5V, therefore a step-up DC/DC Instruments (2013) circuit was implemented to raise the voltage from 3.7 to 5V, it operates at a frequency of 600 kHz, as Fig 10.

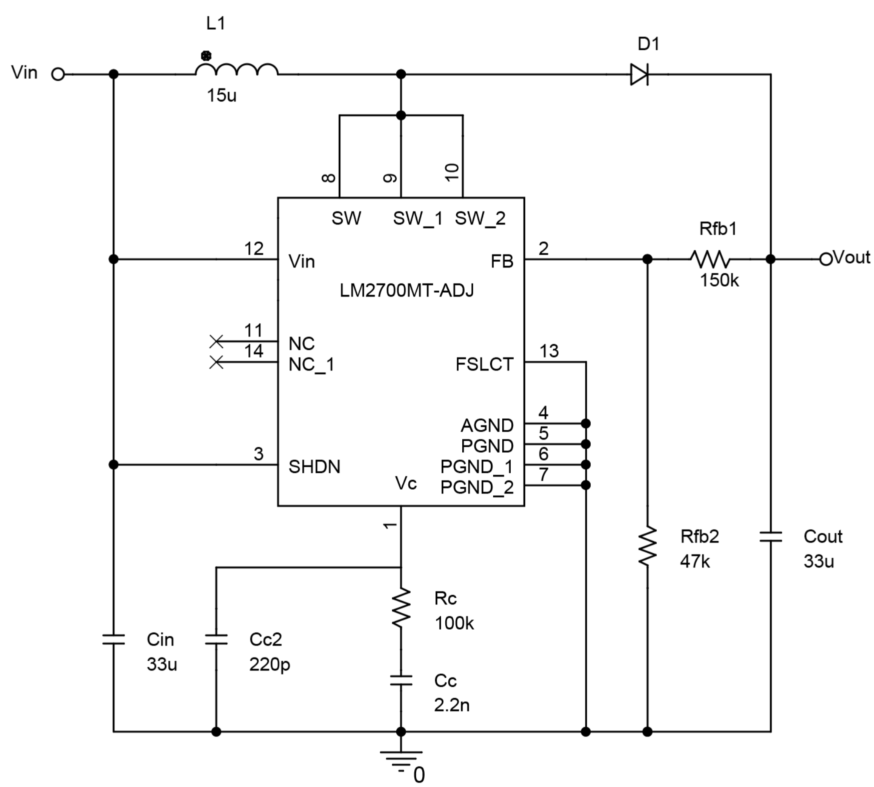

Figura 10. Boost analog circuit. Taken from Instruments (2013)

Where the following Eq. (11) is used to set the output voltage:

$$Rfb1 = Rfb2 \cdot \frac{Vout - 1.26}{1.26} \Omega \tag{11}$$

Also, a small micro USB battery charger SparkFun (2015) was used to charge the 3.7V LiPo cell with a current of 500 mA. Additionally, a resistive divider was implemented to lower the battery voltage Eq. (12) and measure it through an ADC of the IOIO board to determine whether if it needs charging. If the voltage is at the limit, a warning sign is sent in the app to protect battery life.

$$V_{sen} = \frac{300k\Omega}{300k\Omega + 1M\Omega} \cdot V_{Bat} \tag{12}$$

Figure 11 shows the final result of the boost circuit as a printed PCB board.

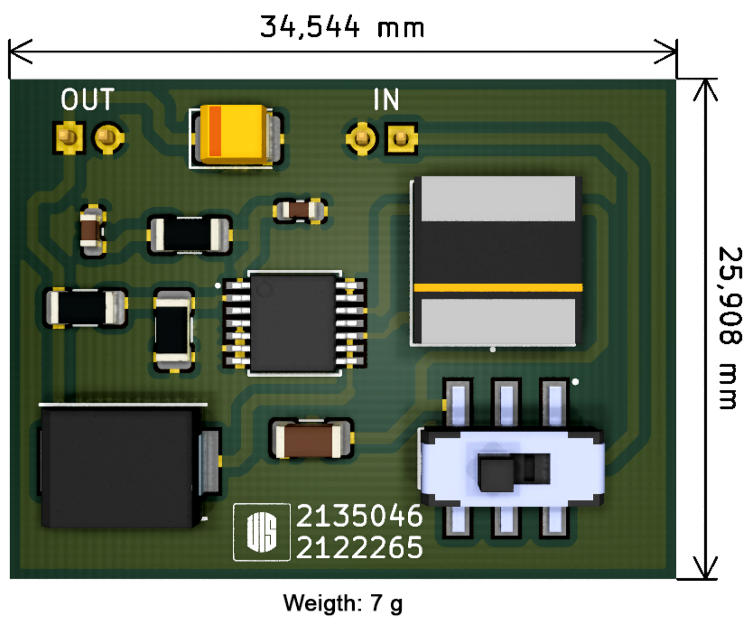

Figura 11. Boost converter PCB board

The ECG signal acquisition system consumes approximately 1.87 W with an average current of 380mA measured several times through a current sensor. When USB OTG communication is not active, the system consumes approximately 1.1 W. We should note that using a step-up DC/DC increases the power consumption of the entire system.

Theoretically and ideally, the battery lasts on 1,578 hours (1 hour 34 minutes), this is estimated using the Eq. (13).

$$Battery\ life\,[h] = \frac{Battery\ capacity\,[mAh]}{Power\ consumption\,[mA]} \tag{13}$$

However, in a practical context the faster a battery discharges, the more energy gets lost by internal resistance. In our case, the battery lasts approximately 60 minutes.

6. Results

This section describes the different tests that were performed on the device using signals from an open-access database. We carried the simulations of the data out by loading the records into the smartphone in comma-separated values (csv) format, fixing the data for proper interpretation. To do so, we simulated three records with different pathologies each. Finally, we performed a test using the hardware designed for ECG signal acquisition. In each ECG graph presented in this section, the horizontal separation on the time axis is 40 ms, and the vertical separation on the amplitude axis is 1 mV.

6.1. Bradycardia and Tachycardia

Bradycardia is the decrease in normal heart rate, meaning the heart beats less than 60 times per minute. Tachycardia is the increase in heart rate when it is at rest, meaning that the heart beats over 100 times per minute Clinic (2019). The collected data is in the database (MIT-BIH Normal Sinus Rhythm Database Goldberger et al. (2000b)) where it includes 18 ECG records. For this simulation data from 16265 record Goldberger et al. (2000a) was used, we loaded this data into the application by modifying the sample rate to generate bradycardia and tachycardia.

Figure 12 shows the slow-paced ECG plot (bradycardia) in the lead II.

Figura 12. Ecg signal of the B16265 record

Figure 13 shows the fast-paced ECG plot (tachycardia) in the lead II.

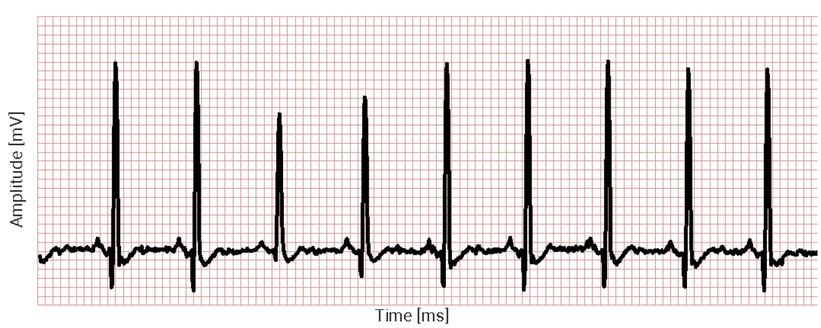

Figura 13. Ecg signal of the T16265 record

The results are organized in the Table 3.

Table 3
Simulation B16265 and T16265 results performed.

Register	Heart rate	Heart rate variability
B16265	Clinical: 52 bpm	Max: 56 bpm (1079 ms)
	Diagnosis: Bradycardia	Min: 48 bpm (1261 ms)
		HR: 52 bpm (1148 ms)
		RMSSD: 37 ms
		SDNN: 41 ms
		pNN50: 16%
T16265	Clinical: 124 bpm	Max: 135 bpm (445 ms)
	Diagnosis: Tachycardia	Min: 94 bpm (635 ms)
		HR: 124 bpm (482 ms)
		RMSSD: 27 ms
		SDNN: 22 ms
		pNN50: 2%

6.2. Ventricular tachyarrhythmia

Ventricular tachyarrhythmias are heart rhythm disorders. These disorders are caused by abnormal electrical signals in the heart. To carry out our simulation, we considered the ECG database (Creighton University Ventricular Tachyarrhythmia Database Goldberger et al. (2000b)) that contains 35 records taken from people who experienced episodes of arrhythmias, it labels all heartbeats normal (although some are ectopic). Particularly, we used the data from cu17 record Nolle et al. (1986).

Figure 14 shows an accelerated changes in the measurement of RR intervals in the lead I.

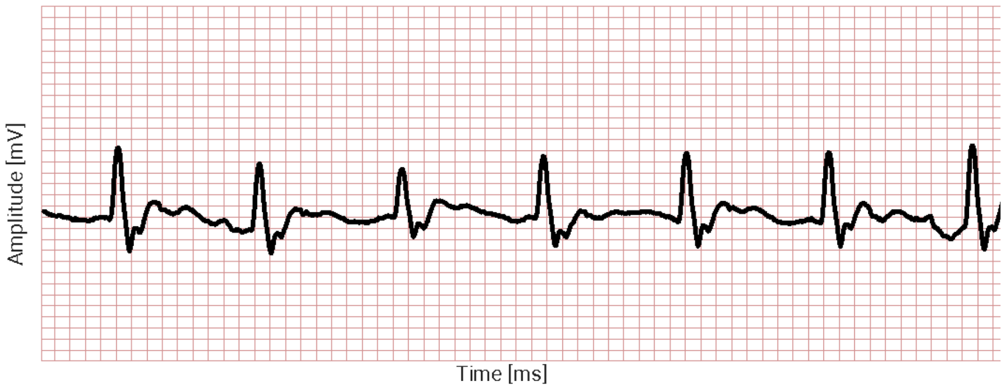

Figura 14. Ecg signal of the cu17 record

The results are organized in the Table 4.

Table 4
Simulation cu17 results performed.

Register	Heart rate	Heart rate variability
cu17	Clinical: 79 bpm	Max: 96 bpm (622 ms)
	Diagnosis: Normal	Min: 73 bpm (825 ms)
		HR: 79 bpm (763 ms)
		RMSSD: 57 ms
		SDNN: 39 ms
		pNN50: 17 %

6.3. Supraventricular arrhythmia

Supraventricular arrhythmias are one of the major causes of several cardiovascular diseases. Considering this, we collected the data (MIT-BIH Supraventricular Arrhythmia Database Goldberger et al. (2000b)) containing 78 records of some kinds of arrhythmia. Here, we used the data from 803 record Greenwald et al. (1990).

A change in pace is clear in Figure 15 and frequency variability in Table 5.

Figura 15. Ecg signal of the 803 record

The results are organized in the Table 5.

Table 5
Simulation 803 results performed.

Register	Heart rate	Heart rate variability
803	Clinical: 60 bpm	Max: 186 bpm (323 ms)
	Diagnosis: Normal	Min: 41 bpm (1466 ms)
		HR: 60 bpm (1005 ms)
		RMSSD: 297 ms
		SDNN: 191 ms
		pNN50: 36 %

6.4. Normal sinus rhythm

Using the circuit proposed in this paper, we tested its operation, performing the following steps. First, we placed the electrodes in the right arm (RA), left arm (LA) and lower abdomen (RL). Second, we inserted the plug to the ECG board and then the USB OTG cable was connected to the board and smartphone. Third, we verified the device connection. Finally, we pressed the app button to start the measurement. Figure 16 shows the ECG signal of the lead we got in this test.

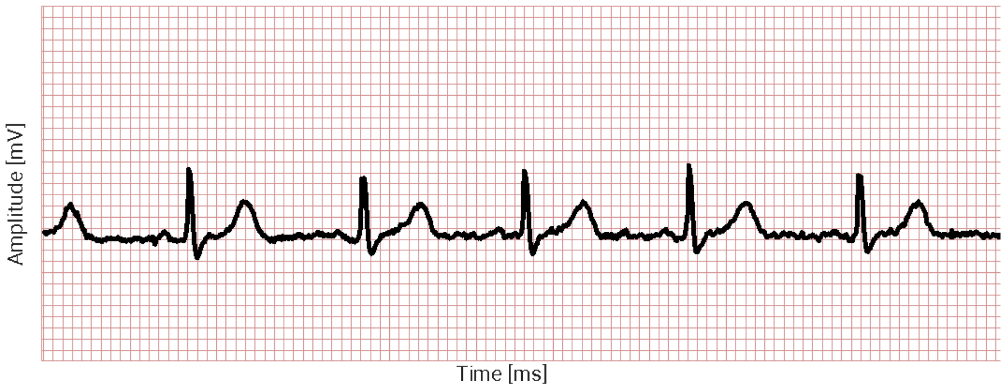

Figura 16. Ecg signal of the dg01 record

The results are organized in the Table 6.

Table 6
Simulation dg01 results performed.

Register	Heart rate	Heart rate variability
dg01	Clinical: 82 bpm	Max: 91 bpm (662 ms)
	Diagnosis: Normal	Min: 74 bpm (814 ms)
		HR: 82 bpm (736 ms)
		RMSSD: 33 ms
		SDNN: 30 ms
		pNN50: 11 %

Figure 17 shows the final prototype of the system for the acquisition of the ECG signal. Inside the box are the PCB boards and out of it the elements needed for measurement.

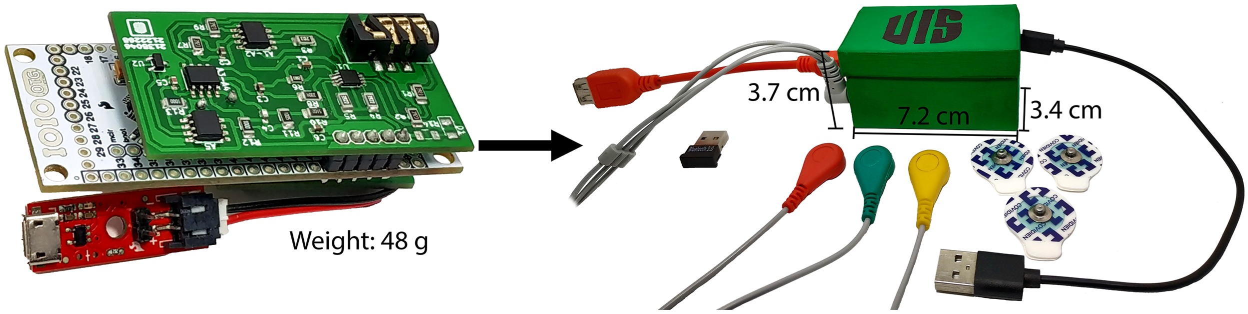

Figura 17. Prototype of the device

To test the results of the work carried out in this paper, we measured with the prototype as shown in the Figure 18

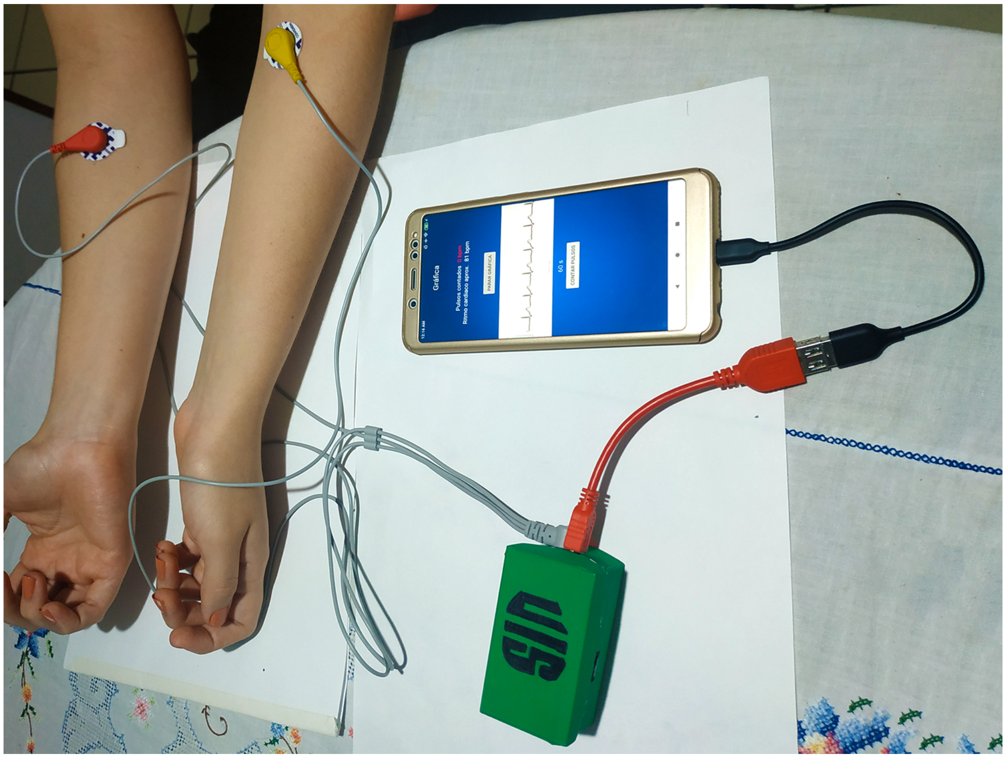

Figura 18. Prototype test

Exporting the measured signal data in .csv text format allows the user to use this file for further analysis of the ECG wave.

The following links lead to videos where we test the operation of the device developed in this document: prototype test (https://youtu.be/p28VvllEFl8), and the way to simulate ECG signals of the data base (https://youtu.be/rYNkWkIRIdA).

Table 7 shows the list of the component used for this system. Total cost of this system is USD 96.14.

Table 7
Cost of the equipments.

Component name	Quantity	Unit price USD	Total cost
IOIO OTG v2.2	1	40.95	40.95
INA326	1	0.50	0.5
OPA2335	3	0.70	2.1
Resistor	16	0.004	0.064
Capacitor	9	0.004	0.036
33u Tant capacitor	1	0.3	0.3
ADR280ARTZ	1	2.74	2.74
Audio Jack	1	0.38	0.38
Electrodes	3	0.15	0.45
Battery	1	3.97	3.97
LM2700MT-ADJ	1	2.05	2.05
15uH Coil	1	0.19	0.19
Diode Schottky	1	0.39	0.39
Switch	1	0.57	0.57
Bluetooth dongle	1	3.0	3.0
PCB Printing	2	12.7	25.4
LiPo charger	1	8.95	8.95
Electrodes cable	1	4.1	4.1
Total cost			96.14

7. Discussion

To get the ECG signal it is important to know the noise affected by the measurement, in Friesen et al. (1990) they describe some cases in which the signal is affected by different noises. During our work, we found some of these. Electrode contact noise is a transient interference for loss of contact between the electrode and the skin, the location of the electrodes in any part of the extremities caused by distinct traces in the ECG signal and therefore it is necessary that their position is correct. Motion artifacts are changes in the baseline and are caused by changes in the electrode's impedance skin with the movement of the electrode. When we perform the measurement and the algorithm does not detect the RR series, it is for the lead that is being used or that the magnitude of the QRS complex is less than the magnitude of the T wave, therefore, it is necessary for the R wave to exceed the threshold so that the measurement can be performed correctly.

We compared the simulations made using the graphs obtained with the record Goldberger et al. (2000a) Nolle et al. (1986) Greenwald et al. (1990) (Figure 19), this data was useful to determine the HR and its HRV. However, as the patients' record is unknown (information such as age, sex, disease, weight) a better diagnosis can't be given. The only important information available of each record are the duration of the ECG recording, the number of samples and the sampling frequency used for testing.

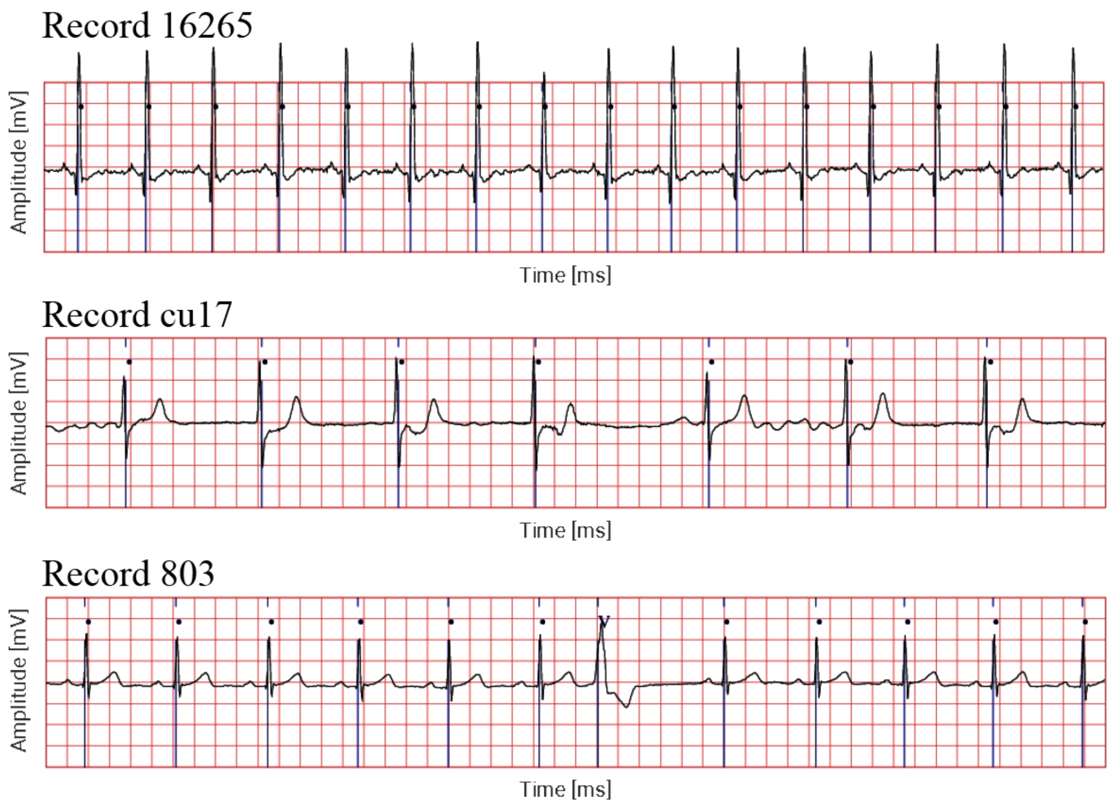

Figura 19. Waveforms plot from databases. Taken from Goldberger et al. (2000b)

To simulate database logs, the application was adapted by changing the reading of an analog port by reading a file in .csv format loaded directly into the smartphone's memory.

To not use a wired connection between the smartphone and the ECG acquisition system, the IOIO board can be used wirelessly by adding a USB Bluetooth dongle to its port. This generates electrical isolation between the system and the smartphone. Repository Ben-Tsvi (2013) warns that latency via Bluetooth is significantly higher than wired connection (10 ms order). The Bluetooth bandwidth is significantly lower than wired connections (order of 10 kB/sec), this means that analog readings can lead to a delay. Also, some dongles exhibit lower bandwidths than others.

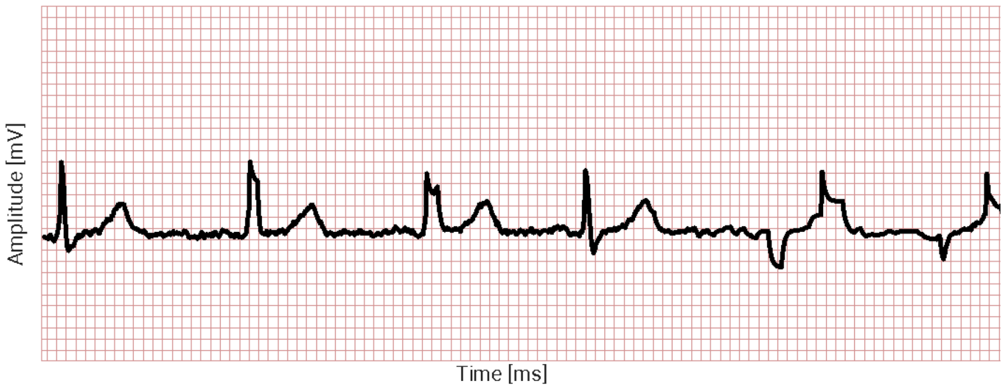

Figura 20. Ecg signal of the dg01 record using Bluetooth

Figure 20 shows an ECG graph transmitted via Bluetooth. Using a dongle to test the acquisition system, we noticed that the signal got more noise, making the calculation of HR and HRV more complex. The latency of communication via Bluetooth deforms the ECG signal graph. Bluetooth communication makes the device portable, however, USB OTG communication ensures less latency and less data loss.

To make a comparison of the obtained ECG signal, we can test the system with an ECG simulator. This instrument used to calibrate electrocardiographs in the health area, can deliver three signals (RA, LA, and RL), simulate arrhythmias and varying heart rate. However, it was not possible to carry out this comparison because at the time this paper was written, there was a mandatory national order of confinement because of the SARS-CoV-2 virus health emergency. Future research will focus on measuring more ECG leads, as they can lead to a more advanced study on heart abnormalities to prevent more deaths worldwide from heart disease.

On the market, several devices can display the ECG signal; Apple Watch Series 5 Apple (2019) can calculate the HR and alert about cardiac irregularities, but this device is of great cost approx. US $400, also exists on the market the ECG Dongle device Healthline (2016), which uses OTG and with four electrodes measures the six limb leads with a cost of approx. US $105, additionally if a person wishes to have an electrocardiogram test in Colombia, the cost is approx. US $14.1 Consultorsalud (2020).

8. Conclusion

In Subin et al. (2017), they proposed an ECG monitoring system based on an Android application with software filters. In Ahamed et al. (2015) using a smartphone with Android and Arduino operating system, they designed a low-cost portable ECG monitoring system. In our work, we developed and implemented a portable ECG signal detection system with the IOIO OTG v2.2 (board focused on developing Android-based smartphone applications). Through the usage of a USB OTG interface data transmission was achieved. For this protocol it was necessary to use an additional source to power the system with a charging device.

The application we developed is used to measure heart rate in two modes: clinical protocol and through the frequency variability, where some results are yielded to be able to monitor them. We expect this application to open a new door between healthcare and the user to prevent future diseases.

We chose a IOIO board to achieve the proposed objective that was the communication using the USB OTG protocol and for its compatibility with the Android system. However, the content of this paper can be applied to other modules with microcontrollers that have connectivity via Bluetooth or Wi-Fi integrated and not necessarily with USB OTG.

We can make the PCB design smaller by using two layers and through-hole technology, but the cost of printing these boards is too high for just one board, the entire system design can be smaller using a smaller microcontroller on the same PCB board as the analog circuit. Regarding energy issues, using a microcontroller that requires less supply voltage would significantly lower the power consumption of the system using the same battery election in this paper without a boost circuit.

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