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RESEARCH ARTICLE

Low-Power Wireless System for Continuous Measurement of Cardiovascular Parameters on a Single Limb

ALBERTO GASCÓN, ERNESTO SERRANO-FINETTI, ÁLVARO MARCO, RAMON CASANELLA, (Member, IEEE), OSCAR LOPEZ-LAPEÑA, (Member, IEEE), GEMMA HORNERO, ROBERTO CASAS, AND OSCAR CASAS, (Member, IEEE)

Aragõn Institute of Engineering Research, University of Zaragoza, 50018 Zaragoza, Spain
Instrumentation, Sensors and Interfaces Group, Electronic Engineering Department, Universitat Politècnica de Catalunya, 08034, Barcelona, Spain

Corresponding author: Oscar Casas (jaime.oscar.casas@upc.edu)

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ABSTRACT A novel low-power sensor wireless system for continuous and simultaneous measurement of the electrocardiogram (ECG) and the impedance plethysmogram (IPG) from a single limb has been designed and tested. This system is made up of an analog subsystem composed of two signal-conditioning channels, one for the ECG and one for the IPG that share the same voltage pickup electrodes, and a digital subsystem based on the BGM220PC22WGA microcontroller that is responsible for digitizing the signals and sending their data via Bluetooth Low Energy (BLE) to a processing unit. The consumption of the system has been analyzed using different settings, obtaining values that would allow a battery duration of up to 131 h, by performing four 30-s measurements every hour. The system was tested in one arm and one leg in a group of healthy volunteers to verify the functionality and to identify the best electrode positions to get high quality signals. Non-standard ECG leads are obtained in both one arm and one leg with an SNR of more than 30 dB. In the worst case, for measurements on the legs, the mean error of the RR intervals from each recording was 1.7 ms and the 95% confidence interval (±SD) was ±7.3 ms versus a reference system. This has made it possible to accurately obtain the Pulse Arrival Time (PAT) between the ECG and the simultaneously measured IPG signal, which allows more cardiovascular information to be obtained from the patient than with systems that acquire only the ECG signal.

INDEX TERMS Continuous cardiovascular measurements, electrocardiography (ECG), impedance pletismography (IPG), low-power acquisition system, wearable device, wireless.

I. INTRODUCTION

Nowadays, advances in the miniaturization of electronics, the rapid development of mobile technologies, wireless communications and embedded systems have been a great help for the development of health sensor monitoring systems, the Internet of Medical Things (IoMT). These devices facilitate continuous real-time monitoring of patients through wearable systems, which represents a substantial advance in remote care [1].

The fast pace of life, the sedentary lifestyle, the poor eating habits, the abuse of tobacco and alcohol, etc, make all the

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cardiovascular diseases (CVDs) more and more common, being the most lethal diseases in the world (32% of global deaths) according to the World Health Organization [2]. The most challenging aspect of these diseases is being able to predict them, due to their concealment and suddenness [3]. Therefore, the constant monitoring would greatly help in the fight against these diseases.

To do this, different parameters can be measured, electrocardiogram (ECG) is the most common, which provides knowledge about the heart's electrical activity as seen from a particular lead vector. At present, the clinical devices used for ECG monitoring in hospitals are very bulky, need many electrodes and do not allow free movement. There are also devices such as the Holter [4] or the SHIMMER3 [5] which, despite being accurate, also require a series of electrodes and cables to be attached to the chest. In addition, in this type of electrodes it is necessary to apply a conductive paste so that the measurement has good precision [6]. For all these reasons, these devices are neither convenient nor attractive for daily use.

In recent years, different portable devices have been proposed for ECG monitoring, using bands on the arm [7] or using devices on the ear [8], to name a few examples. Most of these portable devices work with batteries that must be recharged from time to time. Therefore, another aspect to consider is the reduction of power consumption to maximize battery life, because a frequent recharge need can be an obstacle for patients, especially the older ones. To achieve this, different strategies have been proposed [9] such as the incorporation of energy harvesting methods, the architectural optimization of the analog front-end (AFE) design -and turning on the analog-to-digital converter (ADC) with low-power routines-, the inclusion of elements that compress the digital ECG signal to reduce the number of data, or the optimization of the digital part with low-power microcontrollers (MCU). In general, these portable systems can have two different types of architecture: systems that continuously acquire physiological signals and store this information in a memory (Holter system) for subsequent communication to a central unit that processes the signal and extracts the information [10]. The fact of not having to communicate the data in real time makes it possible to significantly reduce the consumption of the system. However, the monitoring of many people requires a real-time (or as fast as possible) warning/alarm system. Therefore, it is necessary to design systems for the acquisition of more than one channel/physiological signal, with high resolution and that in real time transmits, for convenience in a wireless way, the signal to a central unit. This unit, which can be a cell phone or a computer, will be in charge of processing and alarm and/or action management. While there are many solutions for the first type of system, solutions for the second continue to be developed to reduce overall consumption, and in recent years, communication alternatives with increasingly lower consumption have appeared, such as the Bluetooth Low Energy (BLE).

With these devices, daily and continuous measurements of physiological parameters enable the monitoring of chronic patients, but also the early detection of possible diseases. For this reason, there exists an increasing number of commercial solutions that allow the detection of both the ECG signals and the photoplethysmography (PPG) signals able to track the arterial pressure wave. The PPG signal, like that obtained at the wrist, allows continuous measurements, but the ECG measurement is performed either on the chest -which causes discomfort to the patient- or on the extremities -needing two of them, as is the case of watches that measure only the ECG lead I. Therefore, a system that measures the ECG signal and a pulse wave related signal on a single extremity (either upper or lower) would pave the way for continuous measurement systems with minimal impact on the normal life of users. However, depending on the measurement position, the acquisition of a high quality PPG signal might be difficult, hence an interesting alternative is the measurement of the electrical impedance plethysmography (IPG) signal, which measures, at a single point and non-invasively, the changes in impedance due to the volume changes caused by arterial blood flow. From these signals, ECG and IPG, the heart rate, pulse rate, pulse arrival times are determined and constitute relevant parameters that yield information about the patient's cardiovascular health [11].

Single-limb ECG measurements have been reported [7] where the ECG signals were acquired from a single arm and with high enough signal-to-noise ratio (SNR) as to be able to detect the fiducial points of the ECG, necessary to detect the onset and length of the ECG waves. Although there are new commercial products [12] that measure one ECG lead in the upper arm (non-standard lead), there is no clear information on the optimal electrode position to ensure maximum signal-to-noise ratio in the measurement. Many of these studies have focused on the optimal design of the system, studying comparisons of the use of specific hardware design or dry or wet electrodes [13]. There has not been a detailed study of which positions are optimal or even the possibility of measuring in more than one position to guarantee a signal with a sufficient signal-to-noise ratio. In addition, the possibility of obtaining an ECG signal and an IPG signal in a single arm with the same shared electrodes makes it possible to obtain cardiovascular information beyond the heart rate, such as the Pulse Arrival Time (PAT), by means of a very compact system. The PAT is defined as the time taken by the blood to reach any particular body segment in the extremity from the instant of its ejection into the aorta. In the absence of noninvasive measurement of the instant of ejection blood into the aorta, the R-wave of electrocardiogram (ECG) is accepted as a valid reference point. Instant arrival of the blood volume pulse in the limb segment can be measured noninvasively by any of the plethysmographic methods, in our case the IPG signal. Other measures of great interest for cardiovascular monitoring of patients can be extracted from the PAT measurement [14], [15]. The presented system allows the acquisition of

these physiological signals and PATs, not only in the upper extremity (left arm) but also with the measurement in a single leg, not having found previous works that allow the acquisition of ECG and IPG together in only a single lower extremity.

Therefore, in this work, and in order to have continuous signal monitoring systems, we present the design of a wireless system, with very low power consumption, cost and size, for measuring the ECG and the IPG signals on a single limb, using only four electrodes, allowing the continuous monitoring of a patient and providing more cardiovascular information than current single-lead ECG systems, by being able to measure the times between the two signals (PATs). This limb might be a lower or an upper one depending on the patient's comfort, the activity level or the adaptation to the patient's ailment.

II. SYSTEM DESIGN

The main components of the developed measurement system are an AFE (Analog Front-End) for signal conditioning and a digital subsystem to manage the acquisition and wireless transmission of the measurements to another processing unit (Fig. 1).

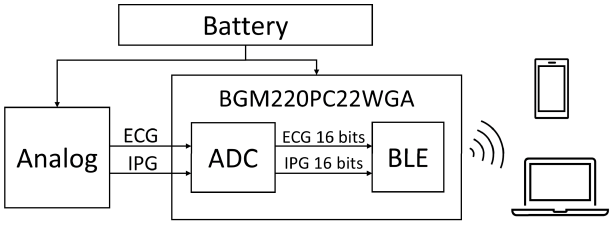


FIGURE 1. Blocks diagram of the designed measurement system, showing its main subsystems.

A. ANALOG FRONT-END DESIGN

Fig. 2 shows the block diagram of the designed AFE that implements simultaneous acquisition of the ECG and the IPG. The supply voltage of the design is 3 V. The IPG signal chain comprises a 20 kHz - 200 μA Howland current source that was applied between two electrodes -(A) and (B)- placed on a single extremity. Two voltage detecting electrodes -(C) and (D)- were placed on the same extremity and connected to voltage buffers (OPA4376, Texas Instruments, Inc.) to reduce the voltage loading effect due to the contact impedance from the electrodes and the body. The ECG signal chain shares these electrodes. An ac-coupled amplifier follows based on a fully-differential 1-kHz high-pass cut-off frequency - to eliminate the ECG contribution- and an instrumentation amplifier (AD8226, Analog Devices, Inc.) with gain G1 = 5. Then, a synchronous demodulator based on a switched-gain amplifier detects the extremity impedance and its variations, mostly attributable to cardiovascular activity. This signal was further amplified (G2 = 500) and band-pass filtered with a second-order active filter (0.5 Hz - 20 Hz) to remove offset and limit noise bandwidth.

The ECG signal chain (Fig. 2b) was band-pass filtered from 0.5 Hz to 100 Hz with a fully differential filter in order to decrease power-line interferences and increase its CMRR [16]. Besides, this bandwidth intends to preserve the signal and avoid significant waveform distortions, and is enough for ambulatory applications. The AD8226 instrumentation amplifier was used again and had a variable gain G3 in the range of 200 to 1000, to allow for different ECG measurement positions. Finally, the last stage filters and amplifies the signal with a low-pass, third order Butterworth filter with a gain G4 = 10. Its cut-off frequency is 100 Hz, almost matching the one from the band-pass filter. A Butterworth filter was chosen due to its flatness in the whole range of frequencies of interest. The op amp model used was the OPA4376 for all amplifying stages different from the instrumentation amplifier.

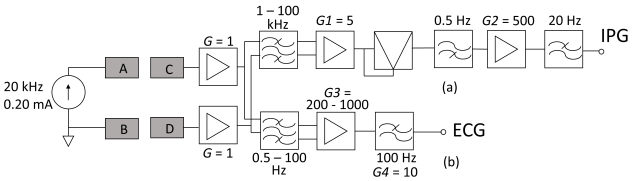


FIGURE 2. Blocks diagram of the designed AFE, showing the IGP signal chain (a) and the ECG signal chain (b).

Both signal chains have been verified in the laboratory. The electrocardiography circuit has been validated through the use of the Metron PS-420 patient simulator and the bioimpedance measurement circuit through the use of the Agilent 33220a function generator, which allows the modulation of signals with up to 0.1% of modulation index. In the ECG case, the total gain from 2000 to 10000 was validated in the specified bandwidth. The CMRR remains above 70 dB over the entire bandwidth up to 100 Hz (Fig. 3). The output total noise is lower than 17 mV (rms).

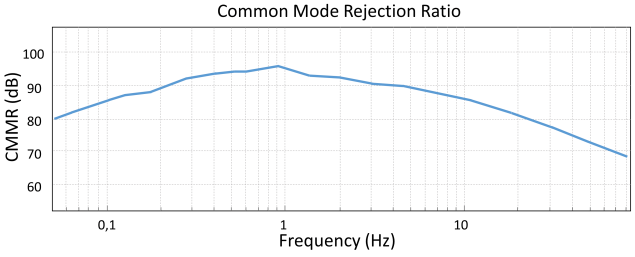


FIGURE 3. CMRR of the ECG acquisition system.

The IPG circuit was also verified. The transfer function of the sub-block consisting of G2 and the low-pass filter is shown in Fig. 4 were the gain magnitude and the phase delay are plotted. For the sub-block consisting of the high-pass filter and the instrumentation amplifier, the CMRR at 20 kHz was measured yielding a value of 51.5 dB with an electrode unbalance of 100% (10 nF in parallel with 100 kΩ). Besides, the output noise was measured with both inputs connected to the ground node yielding a value of 4 mV.

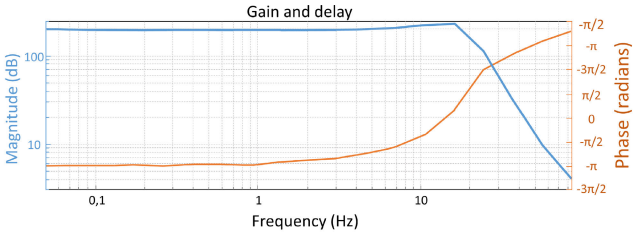


FIGURE 4. Behaviour of the IPG's circuit transfer function at low frequency.

The current demanded by the shared voltage buffers was 1.6 mA. The ECG signal acquisition circuit design adds 1.3 mA more and the IPG measurements require 5.9 mA. Thus, if both signals are being measured at the same time, the total consumption of the AFE is 8.8 mA (3.3 V).

B. DIGITAL SUBSYSTEM

This subsystem is made up of a low-cost, low-power MCU (BGM220PC22WGA) designed for low-power communication via Bluetooth Low Energy (BLE - Bluetooth 5.2). It has 5 energy modes (table 1) and a 12-channel PRS (Peripheral Reflex System) that allows direct communication between the different peripherals, reducing the computational load of the MCU and thus, the power consumption. Regarding the acquisition, it has an ADC of up to 16 channels that can be configured to provide measurements of 12 bits at 1 Msps or 16 bits at 76.9 ksps. It has 25 General Purpose I/O pins with output state retention and asynchronous interrupts. With a view to future adaptations, it is interesting that it has different input methods such as SPI or I2C.

TABLE 1. Summary of the different energy modes in the BGM220PC22WGA, power consumption with 3 V supply at 25 °C.

Energy Modes	Restrictions	Wake up sources	Power consumption
EM0	None	-	26 µA/MHz
EM1	CPU disabled	Internal IRQs	17 µA/MHz
EM2	CPU and HF Oscillator disabled	Internal IRQs	1,75 µA
EM3	CPU, HF and LF Oscillator disabled	External IRQs	1,05 µA
EM4	EFR32 MCU is completely shut down	Power-on reset, external IRQs	0.17 µA

To power the system, a very small PCB was designed so as not to limit its functionality (Fig. 5). In this way it is possible to power the MCU with a small battery which will not add much weight or bulk to the device, keeping it comfortable for the user. This device will be referred to as Smart Cardio.

However, the use of such small hardware presents difficulties for its programming and debugging, as it does not provide an easy access to the different pins. In order to carry out these tasks, two PCBs were designed and manufactured. A first one that allowed access to the Smart Cardio pins and a second one that made it possible to connect the programming pins of the Smart Cardio through a JTAG connector with the programming board.

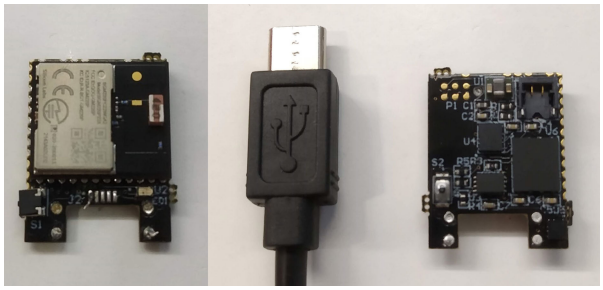


FIGURE 5. Top and bottom layer of the PCB designed. Micro USB charger added for size reference.

1) FIRMWARE DESIGNED

A program based on a state machine with three states (initialization, sleep and measuring) was developed for collecting and sending data.

In the first state (Starting), the system initialization is carried out, configuring the high and low frequency clocks, the BLE communication protocol, and other necessary MCU peripherals.

The second state (Sleep) corresponds to a sleep state. In it, a power saving mode (EM2 or EM4) is entered to reduce the consumption of the MCU and extend the time of use.

Finally, the third state (Measuring) corresponds to signal acquisition and transmission. In this mode, after connecting a central device to the peripheral and subscribing to the corresponding characteristic, the measurements values begin their continuous transmission (Fig. 6). In the transmission process, special attention was paid to achieving a lossless communication. Based on this objective, the MCU was programmed to sample the physiological signals at a frequency of 500 Hz, but data transmission was performed at a 10 times lower frequency, sending 10 samples in every message. The size of a message with 10 samples was the maximum message size that could be sent without losses, i.e. before the next measurement was initiated.

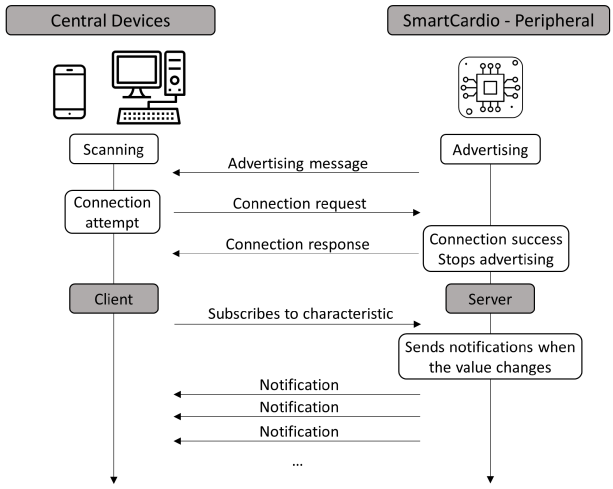


FIGURE 6. Diagram of the BLE communication process.

The normal operation of Smart Cardio begins at the Starting state after which it goes into the sleep state. It stays in this state until a timer interrupt forces it to enter the measuring state (Fig. 7).

The time that the MCU spends in the Sleep state or in a specific low-power mode depends on the application and is programmable. If an application requires constant monitoring, it does not enter Sleep, goes directly into the Measuring state, and begins transmitting data. If no constant monitoring is necessary, and the aim is to minimize power consumption, the MCU can be sent to low consumption mode EM2 and a low power timer is used to periodically wake the system up to perform measurements during a period of time. However, if the main objective is to maximize the operating time, the MCU must enter the lowest consumption mode (EM4), from which it can only get out by means of an external reset. To do this, it is necessary to include a low-power real-time clock (RTC) such as the DS12885 (Maxim Integrated) to send the necessary reset signal.

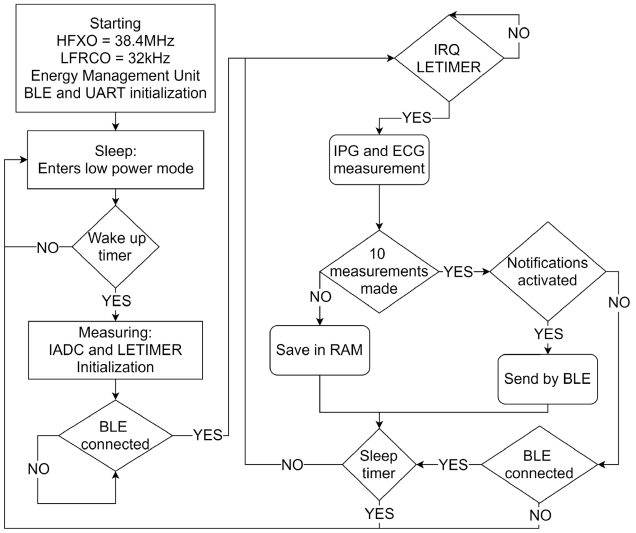


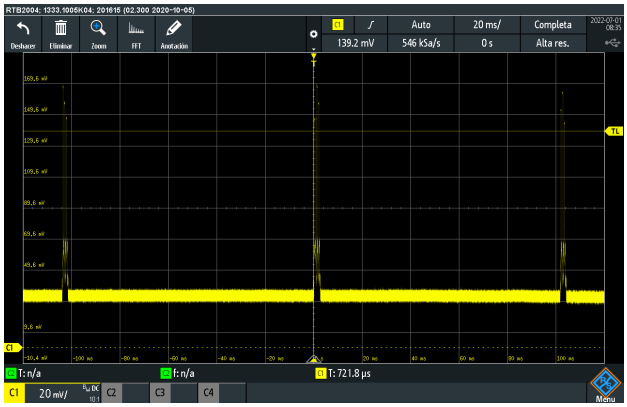
FIGURE 7. Flow chart of the firmware described in section B. Digital subsytem. Firmware design.

C. POWER CONSUMPTION ANALYSIS

To calculate the average power consumption of the system, we need to know the power consumption and duration of each activity performed. To do this, the consumption in the different states mentioned above has been analyzed using an oscilloscope and a voltage divider circuit.

In the Starting state, trains of three peaks corresponding to the advertising messages sent by the MCU can be seen every 102 ms (Fig. 8a) for each of the three channels that BLE dedicates to this task (37, 38 and 39). These trains last approximately 2 ms and in each period the consumption is around 3.2 mA (Fig. 8b.

In the Sleep state, two cases must be considered depending on the system's power mode: if on EM2, the consumption drops to 1.75 μA, and if on EM4, it drops to 0.17 μA,



(a)

(b)

FIGURE 8. Smart cardio consumption in starting mode focusing on a) the periodicity b) the shape and value of a peak train.

although the 500 nA consumed by the RTC needed to wake it up would have to be added.

Finally, in the Measuring state, the data transfer is done by trains of different number of pulses emitted periodically every 50 ms (Fig. 9a). These trains have a duration that varies between 3 ms and 17 ms with peak values of 16 mA (Fig. 9b). To determine the consumption in this state, the average number of pulses calculated after analysing the graphs obtained was close to 8, thus obtaining a final consumption of 4.9 mA. Detailed values of the consumptions of each state are shown in table 2.

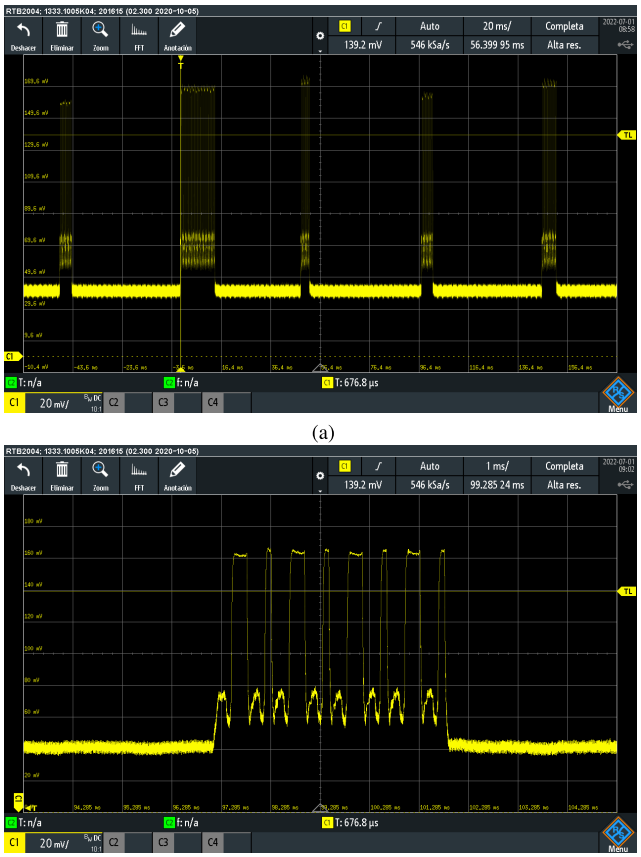
Therefore, once the consumption in each state is known, calculations have been made to determine the duration of the battery in different cases considering both the AFE and the digital subsystem consumptions. As a reference, a 60 mAh battery was taken.

In the first case, continuous monitoring was considered, i.e., to remain constantly in the Measurement state by running both the digital (4.9 mA) and analog (8.8 mA) systems continuously.

$$\begin{aligned} Duration &= \frac{\text{Battery capacity}}{\text{Average current consumption}} \\ &= \frac{60.0\ [\text{mAh}]}{(4.9 + 8.8)\ [\text{mA}]} \\ &= 4.4\ \text{h} \end{aligned} \tag{1}$$

TABLE 2. Average consumption of the Smart Cardio's digital subsystem in each of its states, considering the values of the peaks, their durations and their periodicity.

	Basic consumption $[mA]$	Peaks consumption $[mA]$	Peaks duration $[ms]$	Period $[ms]$	Average consumption $[mA]$
Starting	3.0	8.9	2.2	102	3.2
Sleep EM2	-	-	-	-	0.00175
Sleep EM4	-	-	-	-	0.00017
Measuring	4.2	10.5	5.7	50	4.9



(a)

(b)

FIGURE 9. Smart cardio consumption in measuring mode focusing on a) the periodicity b) the shape and value of a peak train.

In the second case, periodic measurements of 30 s duration were considered. Both, the period between measurements (T) and the low power state in which the MCU was put during the Sleep state (EM2, EM3 or EM4) were considered variables. In this case, the average current consumed will be obtained considering that of the Measurement state and that of the corresponding Sleep state, taking into account the time the system is in each state during a period. For the calculated example, the EM2 and EM4 modes were selected and a period of 15 min was used (900 s).

$$\begin{aligned} &Duration \\ &= \frac{\text{Battery capacity}}{\text{Average current consumption}} \\ &= \frac{60.0[\text{mAh}]}{\frac{I_{Meas}[\text{mA}]\cdot 30.0[\text{s}] + I_{Sleep}[\text{mA}]\cdot(T-30.0)[\text{s}]}{T[\text{s}]}} \end{aligned} \tag{2}$$

$$\begin{aligned} &Dur(\text{EM2}) \\ &= \frac{60.0}{\frac{(4.9+8.8)\cdot 30.0 + 1.75\cdot 10^{-3}\cdot(900.0-30.0)}{900.0}} \\ &= 130.9\ \text{h} \end{aligned} \tag{3}$$

$$\begin{aligned} &Dur(\text{EM4}) \\ &= \frac{60.0}{\frac{(4.9+8.8)\cdot 30.0 + 0.17\cdot 10^{-3}\cdot(900.0-30.0)}{900.0}} \\ &= 131.3\ \text{h} \end{aligned} \tag{4}$$

III. RESULTS AND DISCUSSION

All measurements were conducted in strict compliance with the Helsinki Declaration about ethical principles for research involving human subjects. Volunteers signed an Written informed consent was obtained prior to the study in accordance with the ethics committee approved for the CEIC Aragon Government for this project (MyGait project - 05/2022). To validate the system, measurements have been performed on 7 different volunteers: Three men and four women (34.9±9.5 years), and in two different extremities: in the left arm and in the left leg.

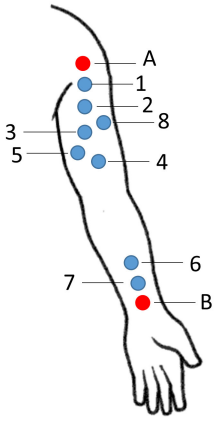


FIGURE 10. Location and numbering of the different detection electrodes (blue) on the left arm. In red the injection electrodes to IPG measures.

A. ONE ARM MEASUREMENTS

Performing different measurements with the electrodes (Skintact Solid Gel ECG Electrodes $RT-41Ag/AgCl$) in different positions of the arm, it has been checked that it is possible to measure a good quality ECG signal in the arm

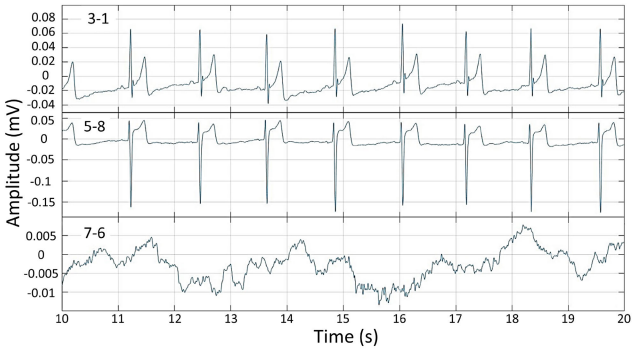


FIGURE 11. Comparison of the ECG signal with the electrodes in different positions of the arm.

with a local measure with a pair of electrodes separated by a few centimeters (Fig. 10).

If the electrodes are in the upper arm, the measured ECG shows good quality results. However, the farther the electrodes are from the heart, the smaller the amplitude of the ECG signal becomes. For this reason, at some point the amplitude is so small that it is confused with the internal noise of the measurement. Moreover, it has been checked that it is not possible to measure a quality ECG signal in the lower arm with a local measure in the wrist. The comparison between two measurements in the upper arm (3-1 and 5-8) and one in the lower arm (6-7) can be seen in Fig. 11.

TABLE 3. Signal to noise ratio for different measures of ECG in one arm.

Electrode locations	Subject 1 (SNR-dB)	Subject 2 (SNR-dB)	Subject 3 (SNR-dB)
2-1	16.7	30.8	35.1
5-3	37.8	36.4	32.6
6-1	38.8	25.1	30.1
7-4	21.4	0.0	0.0
3-1	33.0	26.7	34.3
5-8	48.3	37.4	31.3
7-6	0.0	0.0	0.0

Studying the variability between volunteers, it has been seen that the optimal position of the electrodes to measure a good quality ECG signal is different in each one. This is tested comparing the SNR of the ECG signal of three subjects (representatives of the best, worst and typical cases in the different positions) in the same position, as shown in table 3. This variability can be easily explained using the dipole model. The heart can be modeled, on a first approximation, as a three-dimensional dipole which acts as a source that generates an electric field. From this dipole, it is assumed that the unipolar tension in the body surface is a linear combination of each of the three dipole components. This assumption can be made knowing that the heart-torso electrical system is lineal and quasi-static [17]. This implies that at every instant, the electric field inside the body is in equilibrium with its sources in the heart. Hence, given a distribution of sources, an electric field is present without any regard to the source distribution at previous instants. That is, at a given time, the ECG recording is uniquely given by the state of the source at that time. As described in [18] the dipole can be represented by

$$p = p_x x + p_y y + p_z z. \tag{5}$$

On the other hand, a lead vector can be defined from the position of an electrode to the position of the other electrode, in differential leads. In the case of unipolar leads, the lead vector starts at the starting point of the dipole vector and ends at the position of the recording electrode. If this vector is denoted as $\mathbf{c}$, then

$$V = p \cdot c = p_x c_x + p_y c_y + p_z c_z. \tag{6}$$

This model allows to estimate that depending on the orientation of the dipoles, the measured tension in a particular direction can vary, which is exactly what is observed. A solution to this variability, to ensure a good measure in any subject, could be to position a patch with several electrodes in the upper arm, so that different measures can be performed at the same time and then choose the best one.

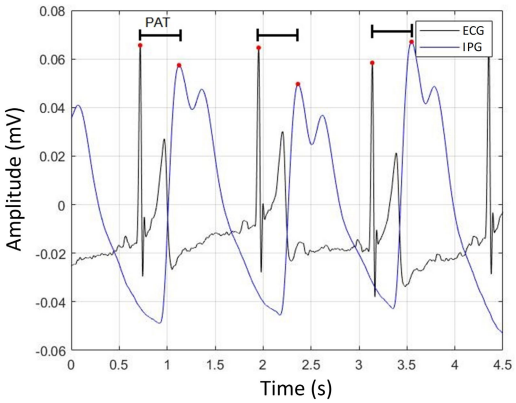


FIGURE 12. ECG and IPG simultaneous measurements on the left arm.

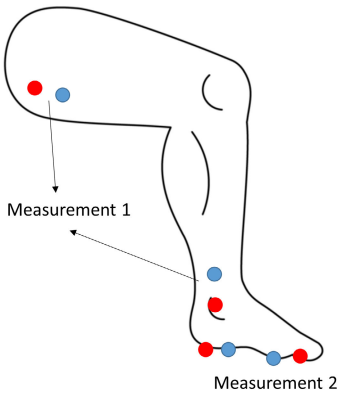


FIGURE 13. Location of the electrodes on the left leg measurements. In red the injected electrodes and in blue the detection electrodes.

IPG could be measured at all longitudinal positions, as already presented in previous papers [19]. To obtain a distal measurement, the IPG can be obtained from the voltage measured between positions 6 and 7, so that simultaneous measurement of ECG and IPG allows us to estimate temporal parameters such as Pulse Arrival Time (PAT) [20] Fig. 12.

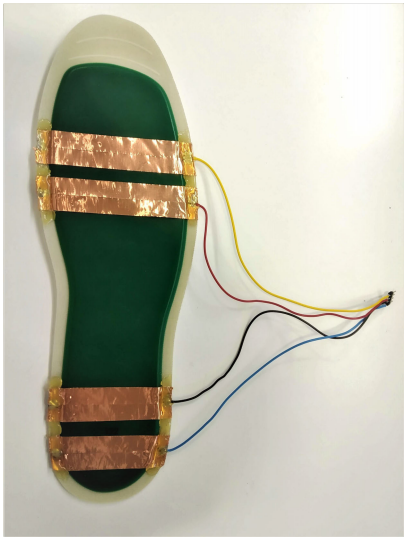


FIGURE 14. Electrodes on an insole used for the measurements.

B. ONE LEG MEASUREMENTS

Two different measurements were performed on the lower extremity (Fig. 13). On one hand, measurement 1 between the upper and lower legs allowed the simultaneous measurement of the ECG and IPG. In the case of measurement 2, due to the high level of electromiographic noise present in the ECG channel,it was only possible to directly obtain the IPG. The electrodes used for the measures are the Skintact Solid Gel ECG Electrodes RT-41 Ag/AgCl for measurement 1 and four dry copper electrodes on an insole. (Fig. 14).

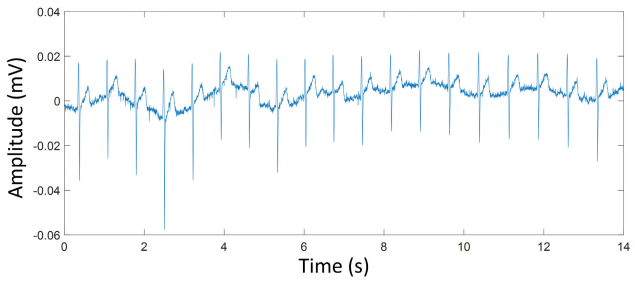


FIGURE 15. ECG obtained on the left leg.

As shown in Fig. 15, it was possible to obtain the ECG in electrode arrangement 1 with all volunteers. However, compared to the measurements on the left arm, the contribution of the electromyography signal (body support) is higher and limits the signal to noise ratio to values between 17 and 29 dB in all the volunteers. Fig. 16 shows the acquired ECG, digitally filtered (low-pass Butterworth filter of order 5 and cutoff frequency 30 Hz) and the acquired IPG. As can be seen, it is possible, as in the arm, to obtain the PAT between the two signals.

Finally, Fig. 17 shows, as an example, the IPG acquired on a volunteer with the measurement system in the shoe insole. As can be seen, although the signal is affected by the person's

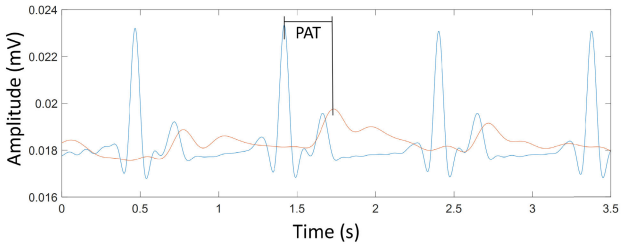


FIGURE 16. ECG and IPG acquired simultaneously on the left leg.

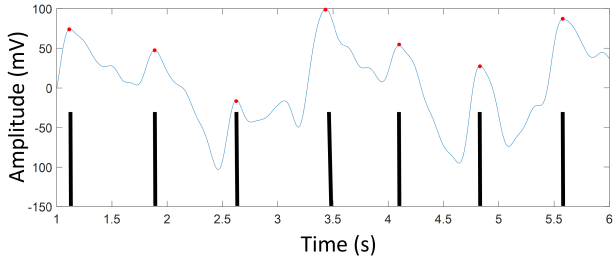


FIGURE 17. IPG acquired from the insole (the black bars give a clearer view of the heart rate).

breathing and movements, it is possible to obtain the pulse rate from the signal without any problem and accuracy [21].

In order to quantify the accuracy of the designed system in the estimation of the heart rate, the results obtained were compared with those obtained with the commercial BIOPAC MP36 system using its module SS2LB for ECG. To perform this validation, a recording of a lead 2 was acquired simultaneously with the two devices at a sampling frequency of 500 Hz and a duration of 1 min. To provide an agreement figure, we used a Bland-Altman processing for each RR time interval detected. In the worst case, that is obtained for measurements on the legs, the mean error of the RR intervals from each recording was 1.7 ms and the 95% confidence interval (± 2 SD) was± 7.3 ms.

From the results it is possible to validate that with the use of only four electrodes it is possible to obtain the ECG and IPG signal in only one extremity with a good signal to noise ratio. Although ECG results had previously been obtained in only one arm, the possibility of obtaining both signals increases the cardiovascular information that can be obtained, based on the estimation of the times between both signals. Furthermore, the possibility of obtaining this information both in one arm and in one leg makes it possible in the future to obtain the two signals in parallel, being able to characterize, from the differences between the PATs of each of them, the propagation times of the pulse wave in the thorax (abdominal aorta).

IV. CONCLUSION

In this article, a system developed for the continuous monitoring of ECG and IPG in a single limb has been presented. The acquisition system for simultaneous measurement of ECG and IPG signal is compact and energy-efficient, using

only four electrodes to perform the two measurements. The signals acquired in high resolution (16 bits) are transmitted by BLE in real time to a digital element, such as a cell phone, to be displayed and recorded. The designed system allows the realization of four series of measurements of 30 seconds every hour with an autonomy of more than five days (130.9 h).

The ECG signal measured in the extremities has a lower amplitude than that obtained in the normalized leads, being influenced by the electromyography signal. Even so, in the positions proposed in the work it is possible to obtain the ECG signal with a signal to noise ratio, which allows the use of the waveform for patient monitoring and estimation of its main parameters. The IPG signal, working in carrier and with a bandwidth lower than that of the ECG,can be obtained it with a higher signal to noise ratio, which makes possible the estimation of time intervals and related parameters, such as the PAT or the pulse wave velocity, increasingly used in the cardiovascular monitoring of patients.

The flexibility of being able to measure both a single lower and upper extremity ECG electrical signal and a pulse wave allows the proposed measurement design to be adapted to multiple applications in a convenient way. As with physiological parameter measurement systems, its most limiting factor is the motion artifacts that limit the accuracy of the measurements. This would limit its application in continuous measurements, such as those performed by a Holter [4], o similar systems [5], but it is very useful for obtaining periodic measurements (for example, n times per hour) taking advantage of the moment when there is no movement in the upper or lower extremity. This absence of movement could be detected by adding a simple inertial system, which could act as a wake-up signal.

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ALBERTO GASCÓN was born in Zaragoza, Spain, in 1995. He received the B.S. and M.S. degrees in industrial engineering and the M.S. degree in electronics engineering from Universidad de Zaragoza, Zaragoza, in 2017, 2019, and 2021, respectively. He is currently pursuing the Ph.D. degree in electronics with the Aragón Institute of Engineering Research. His research interests include the Internet of Things, artificial intelligence, and sensors networks.

ERNESTO SERRANO-FINETTI was born in Lima, Peru, in 1972. He received the Ingeniero en Electrónica degree from Pontificia Universidad Católica del Perú, Peru, in 1998, and the Ph.D. degree in biomedical engineering from Universitat Politècnica de Catalunya—Barcelona Tech (UPC), Barcelona, Spain, in 2003. He has been an Associate Professor with UPC, since 2002, where he teaches courses in electronic systems and instrumentation related topics. In 2008, he was a recipient of the José Castillejo nobility grant and joined the Biomedical Sensors Systems Group, Philips Research Europe, Eindhoven, The Netherlands, until 2009. His research interests include low-power analog design for autonomous sensors, sensor interfaces for applications ranging from smart farming to physiological measurements in humans, animals and plants, and measurements methods based on electrical impedance for non-invasive measurements of biological tissues.

ÁLVARO MARCO received the M.S. degree in electrical engineering and the Ph.D. degree in electronic engineering from the University of Zaragoza, Zaragoza, Spain, in 2000 and 2007, respectively. He is currently a Senior Researcher with the Aragón Institute of Engineering Research, Universidad de Zaragoza, where he is also a part-time Lecturer. He has participated in more than 60 research and development projects for public administrations and private companies, holding 14 patents, and he has published over 60 scientific papers, including both conferences and journal indexed articles. His research interests include machine learning, the IoT, sensor networks, ambient intelligence, and middleware applications. He has served as an Editorial Board Member for *International Journal of Distributed Sensor Networks* and *Wireless Communications and Mobile Computing* and is also a TPC member of several international conferences.

RAMON CASANELLA (Member, IEEE) received the M.Sc. degree in physics and the M.Eng. degree in electronic engineering from the University of Barcelona, Barcelona, Spain, in 2000 and 2001, respectively, and the Ph.D. degree from Universitat Politècnica de Catalunya, Castelldefels, Barcelona, in 2007. He is currently an Associate Professor in electronics with the Castelldefels School of Technology, Universitat Politècnica de Catalunya. His current research interests include electronic and biomedical instrumentation, noninvasive physiological measurements, sensor interfaces, and sensor design based on inverse-problem methods.

OSCAR LOPEZ-LAPEÑA (Member, IEEE) was born in Barcelona, Spain. He received the dual M.S. degree in physics and in electronics engineering from Universitat de Barcelona, Spain, in 1994 and 1996, respectively, and the Ph.D. degree in electronics engineering from Universitat Politècnica de Catalunya (UPC), Spain, in 2000. Since 2002, he has been an Associate Professor with UPC, engaged in teaching on analog and power electronics, where he is currently with the Instrumentation, Sensors and Interfaces Group (ISI), collaborating in the design of power supplies for autonomous sensors nodes. His current research interests include low-power converters, control theory, energy harvesting, and sensor interface circuits.

GEMMA HORNERO was born in Barcelona, in 1972. She received the degree in physics and the Ph.D. degree in science from the University of Barcelona, Spain, in 1997 and 2001, respectively. Since 2001, she has been with the Department of Electronic Engineering, Universitat Politècnica de Catalunya—Barcelona Tech (UPC), where she is currently an Associate Professor and teaches courses in several areas of electronic instrumentation. Her current developed projects and research interests include sensor interfaces, autonomous sensors, electronic instrumentation, noninvasive physiological measurements, and sensors based on electrical impedance measurements.

ROBERTO CASAS received the master's degree in electrical engineering and the Ph.D. degree in electronic engineering. He is currently a Researcher and a Teacher with the University of Zaragoza. Since 2011, he has been coordinating Human Openware Research Laboratory (HOWLab), Aragón Institute of Engineering Research. He has participated in more than 70 research and development projects for public administrations and private companies, coordinating more than 40 of them. Derived from the research done in collaboration with companies, he is a co-inventor of ten patents under industrial exploitation. He has over 30 publications in indexed journals being his fundamental research interests include electronics, communications, and embedded artificial intelligence behind the Internet of Things.

OSCAR CASAS (Member, IEEE) received the B.S. and Ph.D. degrees in telecommunications engineering from Universitat Politècnica de Catalunya (UPC), Barcelona, Spain, in 1994 and 1998, respectively. He is currently an Associate Professor in electronic engineering with UPC and teaches courses in several areas of electronic instrumentation. His current research interests include sensor interfaces, autonomous sensors, electric impedance measurements, electronic instrumentation, and noninvasive physiological measurements.