WIRELESS DEVICE FOR NONINVASIVE RECORDINGS OF CARDIO-RESPIRATORY SIGNALS

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Abstract: This work describes a portable device that acquires two ECG leads, pulse photopletismography waveform and respiratory flow waveform that are sent, using wireless Bluetooth protocol, to a notebook where they are shown on the screen in real time and are also stored into the hard disk. Example of recording during cardiac autonomic activity blockade is presented. The results show that the developed system is a suitable tool to study autonomic modulation of the heart rate variability in different scenarios.

1 INTRODUCTION

The autonomic activity is usually assessed by means of the heart rate variability (HRV) that has been applied to several clinical studies related to cardiological and non-cardiological conditions. Indexes obtained from the HRV in the time and frequency domains are used to point out changes of the autonomic nervous system (ANS) activity. By means of spectral analysis, power measurements of low-frequency (LF=0.04 to 0.15Hz) and high-frequency (HF= 0.15 to 0.4Hz) components of the HRV are employed to quantify the activities of the sympathetic (SNS) and parasympathetic nervous systems (PNS), respectively, during cardiovascular regulation (TASK FORCE, 1996).

Despite their widespread use, there is no consensus on the suitability of these indexes as autonomic outflow markers (Parati et al., 2006). Breathing strongly modulates the HF components of the HRV and, in a less significant way, the LF components as well (Brown et al., 1993). These modulations occur due to the direct neural coupling among breathing, HR control centers of the brain and venous return changes induced by the intrathoracic pressure variation (Saul and Cohen, 1994). Therefore, HF power is simultaneously affected by the PNS activity and by the breathing. The arterial blood pressure (ABP), by means of the baroreflex mechanism, also acts on the HR control centers of the brain, contributing to HR oscillations in the LF and HF bands (Cavalcanti, 2000).

From the above, it is possible to state that LF power, HF power and the LF/HF ratio do not provide reliable figures on the ANS activity and sympathovagal balance (Parati et al., 1995). These arguments suggest that the HRV has limited value to characterize the autonomic system role on the cardiovascular regulation (Chen and Mukkamala, 2008). To obtain indexes that may better characterize the autonomic cardiovascular regulation, additional physiological data have been used (Chen and Mukkamala, 2008). Pulse *photopletismography* waveform (PPG) and respiratory flow waveform (RFW) may provide valuable clinical information to overcome HRV limitations.

This work describes a portable device that acquires ECG, PPG and RFW that are transmitted to a computer by radio-frequency. The device portability makes it suitable for application in different clinical scenarios. Examples of clinical recordings using the developed system are presented.

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2 SYSTEM DESCRIPTION

The developed system consist of a portable electronic device and software for Windows® OS. The device samples ECG, PPG and RFW that are sent to a notebook. Software presents the acquired signals on the screen and records them into the hard disk in real time.

2.1 Hardware

The portable device has transducers and circuits to acquire, process and transmit the physiological signals to the notebook. It can be divided in three parts: acquisition and conditioning unit, control unit and communication interface unit.

The acquisition and conditioning circuits process ECG, PPG and RFW signals. In order to prevent data loss due to movement artifacts or poor skin-electrode contact, ECG from 2 leads are acquired and recorded. Connected to the ECG lead wires, there are passive first order low pass (LP) RC filters with cut-off frequency (f_c) of 8 kHz to attenuate electromagnetic interference (EMI). The next stage consists of an instrumentation amplifier (IA) AD620 (Analog Devices) with a gain (G) of 100. The amplified ECG is applied to a Butterworth band-pass filter ranging from 0.5 to 100 Hz that was built by cascading a first order high-pass (HP) and a third order LP filters. It removes half-cell potential and attenuates interfering signals such as EMG. The ECG signal is further amplified (gain from 10 to 15 adjusted by a potentiometer) and added to DC offset (1.25V). The conditioned ECG ranges from 0 to 2.5V, being within the dynamic range of the analog to digital converter (ADC). The circuits described above were implemented with the operational amplifier TLC2254 (Texas Instruments) that has low power consumption (35µA typical). Figure 1 shows the block diagram of the ECG amplifier.

For the PPG acquisition, the circuit shown in Figure 2 was assembled. An infrared light emitting diode (LED) with the wavelength of 850 nm (SFH4252 - OSRAM) and a photodiode (OPT101 – Burr-Brown) are placed on opposite faces of the subject's finger to measure the light attenuation produced by the blood perfusion and other tissues.

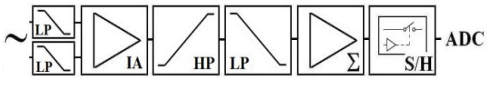

Figure 1: Block diagram of the ECG circuit, containing low-pass (LP) filters, instrumentation amplifier (IA), high pass (HP) filter, baseline displacement (Σ), sample-and-hold (S/H) and analog to digital converter (ADC).

The electric current supplied to the LED is controlled by a potentiometer in order to adjust the signal intensity for each individual. A LP filter (f_c=10Hz) avoids noise interference. The PPG generated by the light attenuation has low (related to skin absorption, sensor displacement and long-term changes of the mean arterial blood pressure) and high frequency components (associated to finger blood volume changes during the heart beat). To get the blood volume curve (PPG-AC) with a better resolution, a HP filter (f_c=0.8 Hz) attenuates the DC component of the PPG. This allows the components of higher frequency to be amplified by an adjustable gain (G) in the next stage without saturation. A LP filter (f_c=10Hz) removes noise of higher frequencies and an offset of 1.25 V is added to the signal to fit its amplitude to the ADC range.

RFW is obtained with a circuit based on thermal anemometry (Figure 3). A Wheatstone bridge contains a heated (70° C) glass sealed bead NTC thermistor (QTMB-16C3 - *Quality Thermistor*). The thermistor is placed ahead of the volunteer nostril. When there is no air flow, the bridge is balanced and its output is 0V. The breathing cools the NTC, changing its resistance and the bridge balance. The power supplied to maintain the NTC temperature constant is an indirect measurement of the RFW.

As for the ECG amplifier, passive first order LP RC filters (f_c = 8 kHz) removes the EMI. The bridge output is amplified by an IA (AD620), with an adjustable gain G, and applied to the transistor base that controls the emitter voltage supplied to the bridge. The emitter voltage, proportional to the square root of the flow velocity, is also amplified and filtered by a LP anti-aliasing filter (f_c =6 Hz). 1.25V offset is added to the signal to make its range compatible to the ADC input.

After their amplification and filtering, the five signals are simultaneously acquired by a sample-and-hold IC (SMP-04) at a frequency of 1 kHz. A microcontroller (ADuC841) that has an ADC built-in converts the signals, one by one, to a 12 bits word. It carries out the conversion in 8 µs with a voltage resolution of 0.61mV (1LSB =2.5 V/4096).

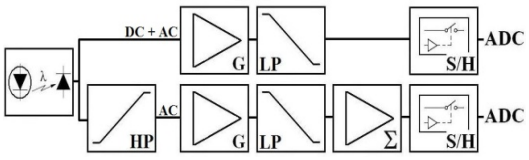

Figure 2: The PPG waveform is sampled by two different channels. One acquires the whole PPG waveform; the other attenuates the DC component to allow further amplification that improves the resolution of the higher frequency components.

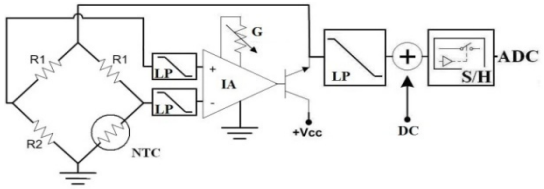

Figure 3: Respiratory flow waveform circuit.

The wireless data transmission is carried out by a Bluetooth module (v1.2 Protocol - KC11 — KCWireFree) that receives the sampled data from the microcontroller serial port. The data transmission occurs in the bypass mode at rate of 115 kbps, reaching the Bluetooth receiver module connected to the notebook USB up to a distance of 100m.

2.2 Software

Software for the notebook was developed in C++ programming language for Windows® OS.

To establish the wireless communication, it is firstly necessary to run a Windows application that makes the Bluetooth modules (portable device and computer) to recognize each other and creates a virtual serial port COMx to be accessed by the acquisition software. The software main window allows the user to select the virtual serial port to receive the data. The received waveforms are plotted and stored in real time. Since Borland C++ Builder graphic libraries are relatively slow for real time applications, the PlotLab library (Mitov Software) was used.

The instantaneous heart rate (IHR), calculated from one of the ECG channels, are also shown on the screen (beats per minute) during the signal acquisition to better supervise the subjects' clinical state. Six windows simultaneously show the ECG1, ECG2, IHR, PPG, PPG-AC and RFW (Figure 4). Each channel is recorded into the hard disk in a separated file using binary format (16 bits Intel PCM). Subject's data (name, age, height and weight), as well as the date and the time of the clinical trial, can be typed and stored along with the signals.

3 EXPERIMENTAL TRIALS

This section describes the use of the developed system during clinical trials to assess the effect of SNS and PNS blockade on the HRV. These experiments are being carried out in the Medical School Hospital of the Federal University of Santa Catarina/Brazil after being approved by its Research Ethics Committee (Project 529/10). Written informed consent was obtained from the subjects who took part in the experiments.

Two ECG leads, PPG, PPG-AC and RFW are recorded for each experimental setting. The volunteers are divided in two groups. Initially, the signals are recorded in supine and standing postures for both groups. An interval of five minutes between these measurements allows the hemodynamic recovery of the subject. These data are used as reference for the other measurements described below. For the first group, intravenous bolus of esmolol (3 mg/kg) is administered by an infusion pump at consecutive intervals of 3 minutes, being the HR observed. The procedure is repeated until the HR stops dropping (β-sympathetic blockade). Following that, the volunteer´s signals are again recorded for the supine and standing postures. At last, the subjects receive 0.03 mg/kg of atropine (Jose and Taylor, 1969) to produce a total autonomic blockade (double blockade) and their signals are again recorded for the same two postures.

For the second group, the same protocol is repeated, but with the drugs applied in reverse order. Atropine is injected first, blocking vagal stimulus. Afterwards, esmolol is administered to bring the total autonomic blockade about. Throughout the recordings, the respiratory activity is guided by a metronome, following three breathing patterns. For each experimental condition (control, β-sympathetic or vagal blockade and total autonomic blockade), the subjects are asked to breathe at the fixed rates of 12 and 15 breaths/min (respiratory frequency of 0.2 and 0.25 Hz, respectively) during two minutes (each rate) and at a random rate, during six minutes. The random breathing phase followed a Poisson distribution with periods from 1 to 15 seconds, mean of 5 s (Berger et al., 1989). The random breathing protocol allows a better analysis of the autonomic activity since the respiratory spectra broadens, preventing the coupling between the respiratory frequency and the HF or LF HRV components. The metronome is an independent application software

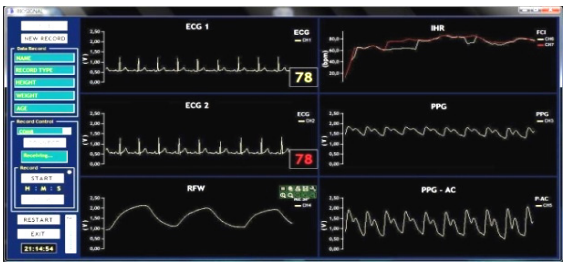

Figure 4: Software main window during acquisition of signals from a volunteer.

Table 1: Spectral power measurements of the HRV bandwidths for 2 subjects breathing at random rates.

(Subject 1)	Breath	LF	HF	LF/HF	(Subject 2)	Breath	LF	HF	LF/HF
CONTROL		(s^2/Hz)	(s^2/Hz)	(unitless)	CONTROL		(s^2/Hz)	(s^2/Hz)	(unitless)
Supine	Random	5,3E-03	3,00E-03	1,77E+00	Supine	Random	1,30E-03	7,58E-04	1,72E+00
	Paced 12	2,3E-03	5,70E-03	4,04E-01		Paced 12	5,90E-04	5,51E-04	1,07E+00
Standing	Random	5,9E-03	3,70E-03	1,59E+00	Standing	Random	1,89E-03	5,30E-04	3,57E+00
	Paced 12	5,8E-03	5,60E-03	1,04E+00		Paced 12	1,37E-03	5,86E-04	2,34E+00
VAGAL-BLOCKADE					BETA-BLOCKADE				
Supine	Random	1,6E-06	1,69E-06	9,36E-01	Supine	Random	3,31E-04	8,49E-05	3,89E+00
	Paced 12	3,8E-07	9,20E-07	4,13E-01		Paced 12	1,15E-04	1,04E-04	1,10E+00
Standing	Random	4,7E-06	1,17E-06	4,06E+00	Standing	Random	1,36E-03	4,64E-04	2,93E+00
	Paced 12	9,5E-06	2,95E-06	3,24E+00		Paced 12	4,99E-04	5,49E-04	9,09E-01
DOUBLE-BLOCKADE					DOUBLE-BLOCKADE				
Supine	Random	3,8E-07	3,57E-07	1,06E+00	Supine	Random	1,30E-05	5,33E-05	2,45E-01
	Paced 12	5,2E-07	4,62E-07	1,12E+00		Paced 12	7,02E-06	7,17E-05	9,79E-02
Standing	Random	3,8E-06	1,52E-06	2,50E+00	Standing	Random	1,66E-05	5,21E-05	3,19E-01
	Paced 12	3,1E-06	1,88E-06	1,63E+00		Paced 12	1,22E-06	3,49E-06	3,51E-01

that shows the respiratory patterns mentioned above as sinusoidal waves on the screen that the volunteer has to follow.

During the experiments, the subject's arterial pressure was acquired by a sphygmomanometer in intervals of 3 minutes and annotated in another file. Such measurements provide data to investigate the role of the pressure on the HRV. These data will be also used to investigate the relation of the pulse wave transit time (PWTT), measured from the ECG and PPG-AC, as an indirect measurement of the instantaneous arterial pressure, as proposed in other works (Lass et al., 2004).

3.1 Data Analysis

A typical recording segment of 30 seconds (Figure 5) shows one of the acquired ECG channels (lead II), RR-interval, RFW, PPG-AC and PPG. The RR-interval waveform was obtained by identifying the R waves of the ECG (red dots) using algorithm based on Ghaffari et al. (2008). Their fluctuations reveal the breath modulation of the HR, that is, the respiratory sinus arrhythmia (RSA). The RSA can be better assessed by spectral analysis. The spectra of the HRV and RFW waveforms were calculated for a subject breathing at two different rates (0.2 Hz and 0.25 Hz) as shown in Figure 6. In this case, the HF spectral peak shifts from one to the other breathing rate. Therefore, if the subject reduces his breathing rate, for instance, from 0.16 to 0.12Hz, the LF power increases and the HF power decreases without modification of the ANS activity.

To illustrate the effects of the ANS blockers, spectral analysis of the HRV was carried out for a subject. The LF, HF and LF/HF ratio data obtained is shown on Table 1. The left column shows the measurements for the first group subject; the right column presents the measurements for the second

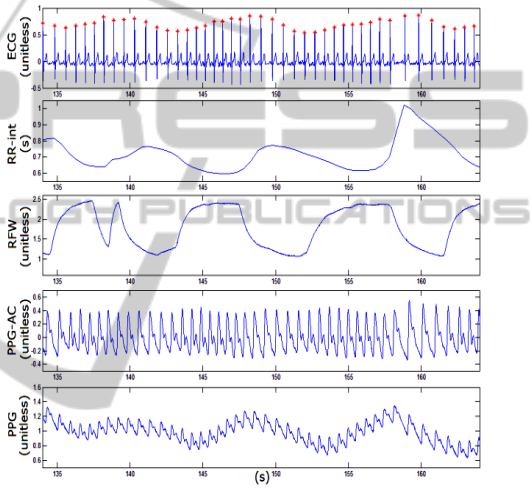

Figure 5: ECG, RR-interval, RFW, PPG-AC and PPG time series. The red dots correspond to the R wave detection used to calculate the instantaneous heart rate.

group subject. During vagal blockade, the HF power decreased markedly, as expected. However, there is difference between its values obtained during the random or paced breathing since, for the second case, the RSA is within the HF bandwidth. During the double-blockade, the results are similar. On the other hand, during the β-blockade, the LF power and LF/HF ratio did not decrease, as one could expect, not revealing sympathetic activity. These findings confirm the drawbacks of such indexes.

4 DISCUSSION

The developed device has unique characteristics since it simultaneously samples different physiological signals that are usually acquired using different equipments. Thus, cardio-respiratory

signals can be registered without any cable connection, reducing sources of artifacts and providing mobility during the recordings. The conditioning circuits demonstrated suitable gains for the desired measurements and good immunity against spurious interferences during recordings in the Hospital. Two ECG leads are acquired by the device to avoid data loss. In case of electrode displacement or deterioration of the electrode-skin interface, the RR interval can be still calculated using the other lead. Concerning the transmission distance, the Bluetooth connection proved to be reliable (no data loss) over a range of 20 m even with obstacles. The elimination of the cables besides reducing artifacts, provides more comfort to the subjects during the recordings.

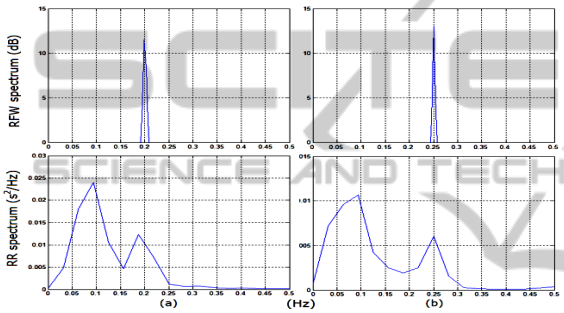

Figure 6: Example of respiratory modulation of the HRV. The breathing rate was increased from 0.2 Hz (a) to 0.25 Hz (b) as can be seen in the RFW spectrum. As result, the HF peak of the HRV is also shifted in the RR spectrum.

The PPG signal provides information on the long-term changes of the mean ABP that can be used together with the PPG-AC to better estimate the systolic and diastolic pressure in future researches.

The presented results are similar to the ones obtained by Chen and Mukkamala (2008) who used propranolol, a non-selective β blocker. Signals of other volunteers are being acquired as described above to set a data bank that will be used to investigate indexes that may better characterize the ANS activity.

5 CONCLUSIONS

A wireless non-invasive system for the recording of cardio-respiratory signals was described. Due to its portability, the system can be used in different scenarios, being a valuable tool for research.

The synchronized recordings of ECG, RFW and PPG allow a better assessment of the interactions among cardio-respiratory variables and their effect on the HRV.

Digital signal processing techniques and mathematical modelling will be applied to the acquired data to further investigate the cardiovascular regulation by the ANS.

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