Cardiovascular Disease and Sleep Apnoea: a Wearable Device for PPG Acquisition and Research Aims

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Abstract

Many scientific research try to relate PPG signals to other physiological parameters, such as breathing rate, blood pressure, blood vessel elasticity, blood viscosity and other cardiovascular compliances and diseases. The aim of this work is to develop a platform composed of a wearable device and a digital framework for continuous acquisition and processing of photoplethysmography (PPG) signals for scientific researches on cardiovascular disease (CVD) and sleep apnoea (SA).

We developed a MATLAB-based framework for off-line algorithm research and a battery powered microcontroller-based, bluetooth enabled, wearable device for PPG signal acquisition. The first problem we had to face was the type of PPG sensor to use. The best solution we found was the new Nellcor SpO2 forehead reflectance sensor, called Max-Fast. Moreover, we improved the Max-Fast sensor with a MEMS three-axis accelerometer for motion artefacts detection.

Thanks to the wireless communication link to a PC, the wearable device can be used in different scenarios such as clinical environments, dangerous situations, sport activity, during sleep, and also at home in telemedicine systems. At the same time, the digital signal processing framework allows new algorithms research on automatic analysis of PPG signals in the field of CVD and SA.

1. Introduction

Photoelectric plethysmography, also known as PPG, is nowadays widely used in medicine to monitor some physiological parameters, usually related to the cardiovascular system. Discovered by Hertzman in 1938 [1], this very simple technique uses a photosensitive detector applied to the skin and two small light sources, usually one in the red and one in the infrared spectrum. The light scattered is emitted in the tissue and partly absorbed. Part of the light comes out again through the skin and is detected by the photoelectric cell. The PPG technique for medical monitoring is usually applied in a device called pulse oximeter, which measures only the oxygen saturation of the blood (i.e. SpO2) and the heart rate. These measures allow a rapid detection of hypoxemic events and help the monitoring of the progression of the patient's medical condition. For example, pulse oximeters are used to monitor patients during surgical operations, rehabilitation sessions, sedation, pain management, or sleep states. However, there are many scientific research trying to relate PPG signals to other physiological parameters, such as breathing rate, blood pressure [2], blood vessel elasticity, blood viscosity and other cardiovascular compliances and diseases [3]. This work deals with cardiovascular disease (CVD) and sleep apnoea (SA), two apparently different illnesses, but in fact strictly related.

CVD represents one of the major causes of death all over the world. Moreover, hypertension and obesity resulting from unhealthy lifestyle, increase the risk of cardiovascular diseases and CVD mortality. In Europe, for example, 40% of deaths are due to disorders related to diseases of the cardiac muscle and of the vascular system supplying heart, brain, and other vital organs.

Sleep apnoea is a common sleep disorder characterized by brief interruptions of breathing during sleep [4], and they can vary from 10 seconds up to one minute [5]. SA is associated with a wide range of health implications and increases cardiovascular diseases and mortality. It has been linked to depression, irritability, sexual dysfunction, learning and memory difficulties, high blood pressure (hypertension), heart attack and stroke [6]. Moreover, the effects of sleep apnoea on cardiovascular system (blood pressure for example) are not confined to sleep. Patients with SA have usually high blood pressure levels [7]. Hence, a continuous monitoring of sleep apnoea disorder is recommended in order to have a better control of cardiovascular outcomes.

The gold standard in diagnosing and monitoring of SA is overnight polysomnography (PSG). It is a non-invasive method that uses a lot of sensors for the recording of different physiological parameters [8]. However, PSG is an expensive and time consuming procedure. The most promising method for home diagnosis of SA and CVD is the heart rate variability (HRV) analysis [9]. In this paper we present a microcontroller-based wearable device for real-time acquisition and a MATLAB-based digital

framework for processing of PPG signals for HRV and new algorithms research purposes. Thanks to the wireless communication link to a PC, the usability and the long time measurement feature, the wearable device can be used in different scenarios such as clinical environments, dangerous situations, sport activity, during sleep, and also at home in telemedicine systems.

2. System overview

2.1. The wearable head ribbon

Figure 1 shows the prototype of the developed wearable device.

Figure 1. Some snapshots of the developed wearable device. The figure shows the flexible kapton PCB embedding the Nellcor Max-Fast forehead reflectance sensor, the insertion of the PCB in the Nellcor Max-Fast headband, and the easy way to wear the system.

The development of such type of device needs specific requirements such as usability, reliability of the signals acquired, reduction of motion artefacts and the necessity to long-term acquisition performance. In order to satisfy these strong requirements, the first problem we had to face was the type of PPG sensor to use. We tested commercial pulse oximetry sensors, using as a reference signal the PPG outing from a classical fingertip sensor (Fig. 2). We also developed some reflectance sensor to be applied to various part of the body in order to find a reliable and comfortable part to sense the PPG signal and to design a device easy to wear.

The best solution we found was the use of the new Nellcor SpO2 sensor, named Max-Fast (Fig. 3).

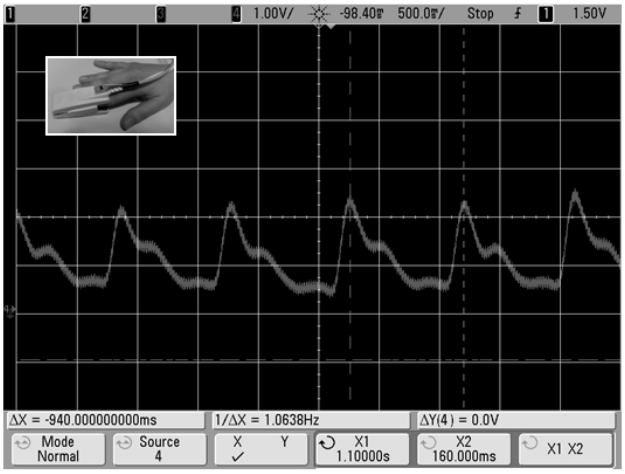

Figure 2. Screenshot of the oscilloscope display for the analog amplified PPG signal coming from a standard fingertip pulse oximeter sensor.

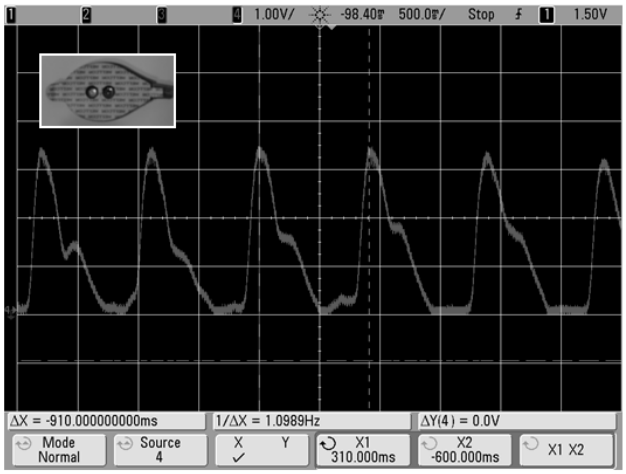

Figure 3. Screenshot of the oscilloscope display for the analog amplified PPG signal coming from a Nellcor Max-Fast pulse oximeter reflectance sensor.

Thanks to a comfortable cotton headband, the forehead sensor is easy to wear. Moreover, the Max-Fast [10] reflectance sensor presents a reduced motion artefacts noise compared to traditional sensors such as the fingertip or the finger wrapped sensors. It also allows long time measurement with continuous high signal strength due to the insensibility of the sensor to pressure necrosis, a typical phenomenon that in other sensors, like ear sensors, reduce the signal amplitude during long term acquisition. Moreover, we improve the Max-Fast sensor with a MEMS three-axis accelerometer (i.e. Parallax MMA7455L) placed behind the sensor for motion artefacts detection and future implementation of algorithms for motion artefacts denoising.

2.2. Block diagram of the wearable device

The core of the device is represented by a very simple, high performance, low power, Digital Signal Controller (DSC), i.e. the Microchip dsPIC33FJ64GP802. This is a 16-bit 40MIPS microcontroller, with 64KB program memory, 16KB data memory, and enhanced with DSP hardware. It also embeds several peripherals such as a Direct Memory Access (DMA), a Real Time Clock, a 12-bit 500sps ADC, a I2C communication bus, and two serial USART, both of them employed in this device.

The block diagram of the developed device is shown in Figure 4.

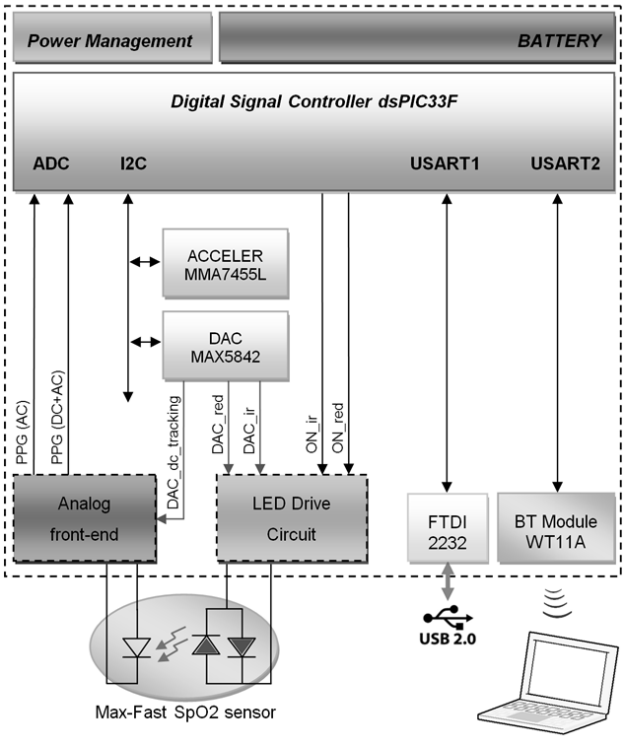

Figure 4. Block diagram of the developed wearable device. The figure shows the core of the device (i.e. the dsPIC33F) connected through its embedded peripherals to the others circuits mounted in the PCB, and the way the device is connected to the Nellcor Max-Fast forehead sensor and linked to the PC.

The DSC manages all the functionalities of the system, starting from the two USART embedded channels in order to perform a wired and a wireless link. The first USART is connected to an on-board USART/USB converter (i.e. FTDI 2232) only for debugging purpose, while the second USART is used to manage an OEM Bluetooth module (i.e. Bluegiga WT11A) for the real-time PPG sending to a remote PC. Through the exploitation of the I2C bus, a three axis accelerometer placed behind the Max-Fast sensor is used to detect the corrupted motion artefact segments and to allow the research of new motion artefacts reducing algorithms based on accelerometer data (e.g. [11]). Moreover, through the I2C bus the DSC manage three out of four 16-bit DAC outputs (i.e. Maxim MAX5842) in order to drive the two source LED, red and infrared, of the SpO2 sensor, and to delete the DC component of the PPG signal coming from the analog front-end amplifier of the SpO2 sensor's photodetector. The source LED drive circuit and the analog front-end we used is published by Texas Instruments in a famous application note about pulse oximetry [12]. The circuitry is composed of two switching PNP transistors (driven by two different I/O lines of the DSC, i.e. ON_red and ON_ir) in order to switch on only a light source at a time, and of two NPN transistors, driven by two voltages generated by the DAC (i.e. DAC_red and DAC_ir), in order to modulate the amount of light emitted by the LEDs and avoid the saturation of the first stage amplifier of the analog front-end.

The analog front-end of the device is a very simple two stage amplifier. The first stage is a common trans-impedance amplifier, able to amplify the photo-current coming from the SpO2 sensor's photodetector (i.e. PPG(DC+AC) signal). The DC component of the signal PPG(DC+AC), caused by the lesser oxygen bearing parts of the body tissue and scattered light, is removed in the second stage amplifier by subtracting a digitally generated offset by the DAC_dc_tracking output calculated through an IIR filter in the DSC. In this way, the AC component of the PPG (i.e. PPG(AC) signal) is digitalized using the built-in 12-bit ADC at a programmable sample frequency, and wireless sent to the PC through the Bluetooth OEM module.

2.3. The MATLAB framework

Previous scientific works have demonstrated that 90% of automated classification of SA events can be correctly identified with HRV analysis in the time and frequency domain [9]. HRV analysis is the standard to describe variations of instantaneous heart rate and the normal-to-normal (NN) intervals (all intervals between adjacent QRS complexes).

Starting from the tachogram of the short-term 5-minute recordings of PPG signals, the MATLAB framework perform an HRV analysis in both time and frequency domain. In the time domain method, each NN intervals are determined and simple time domain statistical variables are calculated. In the frequency domain, the power spectral density (PSD) analysis of the tachogram provides basic information on power distribution in the different frequency bands. In short-term recordings, three main spectral components are distinguished: VLF, LF,

and HF components [9]. Figure 5 shows the processing flow implemented in the MATLAB framework and Tab. 1 and Tab. 2 show an example of the output of the framework.

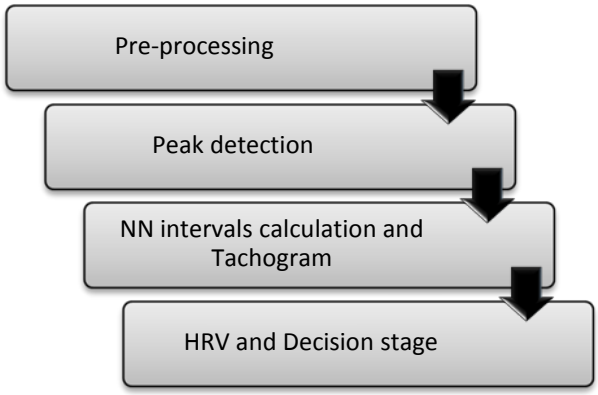

Figure 5. HRV processing flow implemented.

- Pre-processing stage: the framework implements a Butterworth 4th order IIR filter in order to reduce noise and baseline drift of the PPG signals;
- Peak detection stage: the framework implements two different types of algorithms in order to feed the HRV script with the best input data: the first one is based on a classical 3-points algorithm for QRS detection and the second one is a simple first derivative and threshold algorithm;
- NN intervals calculation and Tachogram: starting from NN intervals, tachogram is calculated and normalized;
- HRV analysis and Decision stage: starting from NN intervals and tachogram, statistical analysis in time domain and frequency domain variables are calculated and analyzed; a predefined threshold is applied to the LF/HF ratio in order to determine if sleep apnoea events has occurred.

Table 1. Time-domain output of the framework.

Variable	Value
HR	68 bpm
SDNN	41,2
NN50	109

Table 2. Frequency-domain output of the framework.

Variable	Value
VLF	50,7
LF	176,2
HF	139,7
LF/HF	1,26

Acknowledgements

This work is supported by Region of Sardinia, Young Researchers Grant, PO Sardegna FSE 2007-2013, L.R.7/2007 “Promotion of the scientific research and technological innovation in Sardinia”.

References

[1] Hertzman AB. The blood supply of various skin areas as estimated by the photoelectric plethysmography. American Journal of Physiology 1938; 40:124–328.
[2] Fortino G, Giampà V. PPG-based methods for non invasive and continuous blood pressure measurement: an overview and development issues in body sensor networks. MeMeA 2010; 10–13.
[3] Angius G, Raffo L. A sleep apnoea keeper in a wearable device for Continuous detection and screening during daily life. Computers in Cardiology 2008; 433–436.
[4] Young T, Palta M, Dempsey J, Skatrud J, Weber S, Badr S. The occurrence of sleep-disorder breathing among middle-aged adults. New England Journal of Medicine 1993; 328:1230-1235.
[5] Shahar E, Whitney CW, Redline S, Lee ET, Newman AB, Nieto FJ, O’Connor GT, Boland LL, Schwartz JE, Samet JM. Sleep-disordered breathing and cardiovascular disease. American Journal on Respiratory and Critical Care Medicine 2001; 163:19-25.
[6] Sin DD, Fitzgerald F, Parker JD. Risk factors for central and obstructive sleep apnea in 450 men and women with congestive heart failure. American Journal on Respiratory and Critical Care Medicine 1999; 160:1101-1106.
[7] Somers VK, Dyken ME, Clary MP, et al. Sympathetic neural mechanisms in obstructive sleep apnea. Journal of Clinical Investigation 1995; 96:1897-1904.
[8] Haponik EF, Smith PL, Meyers DA, Bleecker ER. Evaluation of sleep disordered breathing. Is polysomnography necessary? American Journal of Medicine 1984; 77:671-677.
[9] Drinnan MJ, Allen J, Langley P, Murray A. Detection of sleep apnoea from frequency analysis of heart rate variability. Computers in Cardiology 2000; 27:259-262.
[10] Fernandez M, Burns K, Calhoun B, George S, Martin B, Weaver C. Evaluation of a New Pulse Oximeter Sensor. American Journal of Critical Care 2007; 16:146–152.
[11] Chan V, Underwood S. A Single-Chip Pulse oximeter Design Using the MSP430. http://www.ti.com
[12] Poh MZ, Swenson NC, Picard RW. Motion-tolerant magnetic earring sensor and wireless earpiece for wearable photoplethysmography. IEEE Transaction on Biomedical Engineering 2010; 14:786–794.

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