A REAL TIME CARDIAC MONITORING SYSTEM
Arterial Pressure Waveform Capture and Analysis

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Abstract: An arterial pressure waveform recorder and analyser based on a Microchip PIC microcontroller (μC), dsPIC33FJ256GP710 is described in this article. Our purpose is to develop a dsPIC based signal monitoring and processing system for cardiovascular studies, specially dedicated to arterial pressure waveform (APW) capture. We developed a piezoelectric (PZ) probe designed to reproduce the APW from the pulsatile activity taken non-invasively at the vicinity of a superficial artery. The advantages in developing a microcontroller based system show up in decreasing the associate cost, as well as in increasing the functionality of the system. Based on a MathWorks Simulink platform, the system supports the development and transfer of program code from a personal computer to the microcontroller, and evaluation of its execution on rapid prototyping hardware. Results demonstrate that embedded system can be an alternative to be used in autonomous cardiovascular probes. Although additional studies are still required, this probe seems to be a valid, low cost and easy to use alternative to expensive and hard to manipulate devices in the market.

1 INTRODUCTION

The social and economic impact of cardiovascular diseases and the importance of efficient early diagnostic tools keep mobilizing the interest of many researchers (Laurent *et al.,* 2006). Continuous monitoring and analysis of physiological signals, as well as online interactive signal processing are essential in the management of ill patients. The term arterial stiffness denotes alterations in the mechanical properties of arteries, as the decay of elasticity in the arterial wall fibers. Much effort has focused in determining the best way to measure this parameter: pulse pressure, pulse waveform analysis and pulse wave velocity (PWV) measurements are some examples.

Historically the cuff sphygmomanometer was the first method to quantify a part of the medical information contained in the arterial pressure waveform (APW); however it provides a limited amount of information: quantitative blood pressure information at two specific points of the APW. Electrocardiography (ECG) is another widely accepted method to extract cardiovascular information but is rather limited when arterial stiffness information is concerned.

The APW morphology has gained clinical interest due the additional information obtained from the time-varying pulse waveform (Avolio *et al.*, 2010), such as the pattern of the ventricular ejection and the elastic properties of the arterial tree.

Ideally, we are looking for an instrument capable of delivering the calibrated, precise APW at the ascending aorta, even though from a remote sensing site (peripheral artery). Non-invasive assessment of APW typically uses waveforms recorded at one of two anatomical locations: the radial and the carotid artery. Carotid blood pressure is often used as a surrogate for central aortic blood pressure due its location. Van Bortel *et al.* (2001) showed that the carotid pulse pressure differ only 1.8 mmHg from central aortic pulse pressure.

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In this work we concentrate in developing a non-invasive device suitable to carotid APW measurements and to further processing, from which a great deal of clinically relevant information can be derived. The resulting instrument assumes the shape of a real time system, autonomous, with minimal human intervention, capable to respond to the time variations of the physiological signals.

Real time embedded systems using digital signal processors (DSP) in biomedical applications assumed, over the last years, an increasing importance due to the enhanced functionalities that they are capable of imparting. The development of this technology has enabled significant improvements in speed of analysis, accuracy, noise immunity, programmability, size reduction and, in addition, a decrease in cost.

Numerous cardiovascular applications have been reported in the literature: *Klig et al.* (1978) uses these systems for monitoring blood pressure and ECG signals. Bing-Nan *et al.* (2004) proposes an embedded medical advisory system for mobile cardiovascular monitoring devices that provides microcirculation information. Germano *et al.* (2009) introduces a generic architecture for developing biomedical embedded systems with special application for clinical analysis and for patient monitoring.

The Explorer 16 development board with its attached microcontroller is used with some additional hardware in order to configure a fully operational system.

The real time operating system is discussed along the paper, as well as the details of data acquisition, data pre-processing and data transmission to the host computer. In Section 2 a general embedded system design is briefly introduced while the software parts are described in Section 3. In Section 4, experimental results are shown demonstrating a very good overall performance in an almost autonomous (minimum human intervention) mode of operation.

2 EMBEDDED SYSTEM DESIGN

The microcontroller (µC) was selected from the Microchip PIC family due to its features and embedded resources. These µCs are widely available on the market at relatively affordable prices. Moreover, a wide range of programming tools are also available (Bansal et al, 2009, Smolnikar and Mohorcic 2008).

The dsPICs are a hybrid solution that combines the processing power of a DSP with the functionality of a microcontroller, which includes fast interrupt vectors, control of peripherals, general purpose I/O and can run compact code.

The dsPIC33 family, in particular, employs a powerful 16-bit architecture that integrates the control features of a microcontroller with the computational capabilities of a DSP. The dsPIC33FJ256GP710 was chosen due to its characteristics: 40 MIPS processor speed, 256 kbyte program memory and 30 kbyte of RAM.

The Explorer 16 development board (figure 1 a) is a low cost, efficient development board to evaluate the features and performance of Microchip's Microcontrollers, in particular the PIC24FJ128GA010 and the dsPIC33FJ256GP710.

Top and bottom views of a piezoelectric (PZ) probe responsible for capturing the APW at the carotid artery site, are shown in figure 1 b) and c).

The architecture of the system is diagrammatically represented in figure 2.

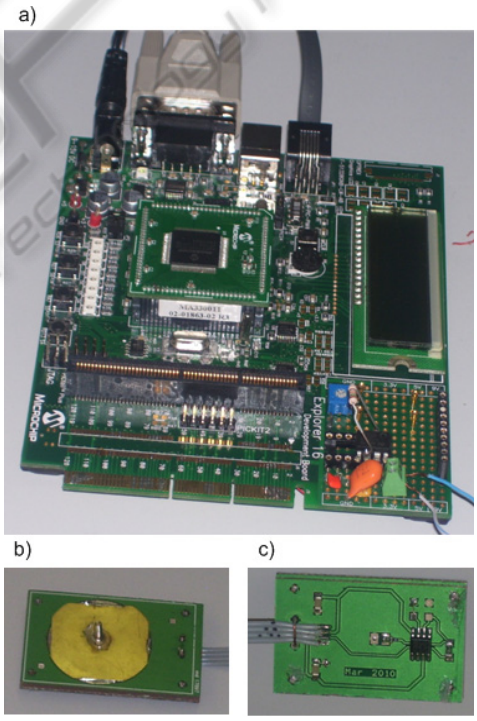

Figure 1: a) Explore 16 development board (a) and PZ probe, top and bottom views, respectively b) and c).

The signal acquisition/processing block is responsible for amplifying the sensor signal and identifying the positive peak (one of the prominent points of the APW).

The signal conditioning block has the function of

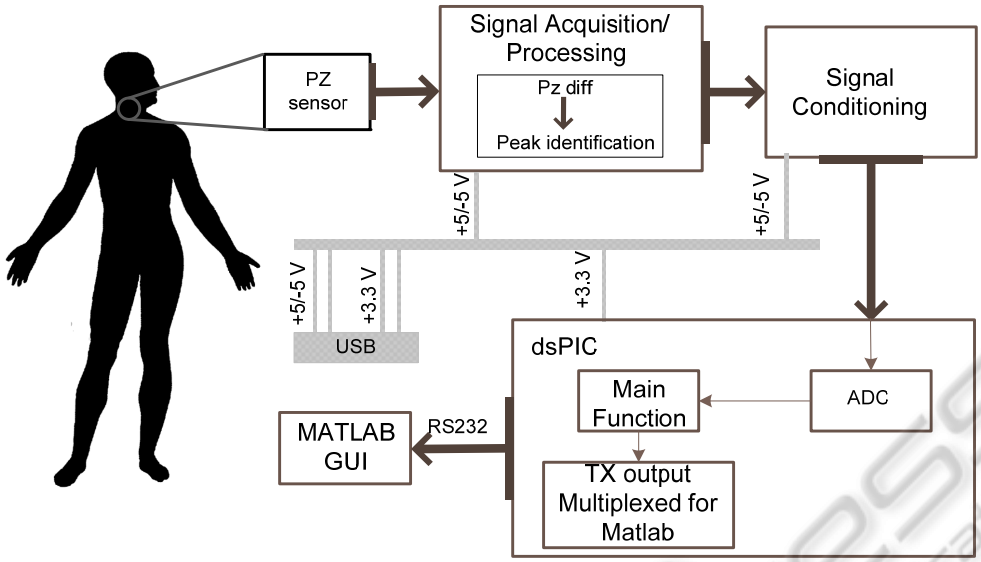

Figure 2: Workflow of the system.

supplying unipolar (positive only) values as required by the dsPIC ADC. The microcontroller is programmed to deliver the final pressure waveform and transmit data via RS232 to a Graphical User Interface (GUI) on MATLAB.

The PZ probe and rs232gui are discussed in detail in the remaining of this section.

The USB (Universal Serial Bus) is also responsible for supplying power to the system.

2.1 PZ Probe

PZ based probes have been widely used in APW measurements along the last years as a result of their characteristics: high sensitivity, high signal-to-noise ratio SNR, as well the low price associated.

A PZ element is able to convert force or pressure applied to its surface into a measurable voltage signal. From an electrical point of view the sensor can be modelled as an AC coupled voltage generator (Karki, 2000) and, consequently, it does not respond to static excitation.

Due to the above mentioned electrical characteristics of the PZ probe, the collected signal appears as a time derivative of the APW that excites the PZ sensor.

In figure 3 a) the configuration of the probe is shown. The PZ sensor (2) in use is the MURATA 7BB-12-9 sounder, this is attached on a double printed circuit board (PCB) (3).The interface between the transducer and artery (or silicon tube) is done by a PVC piece (1) (in form of a “mushroom”, with 15 mm diameter in top). The probe’s covering consist in a plastic box (OKW (ENCLOSURES)-B9002107).The final belt-mounted sensor used to carotid artery in vivo acquisitions is shown in figure 3 b).

The principle of APW measurement is based on the transmission of its mechanical energy that shows up as a displacement of the tissue surface (carotid artery) to the PZ surface.

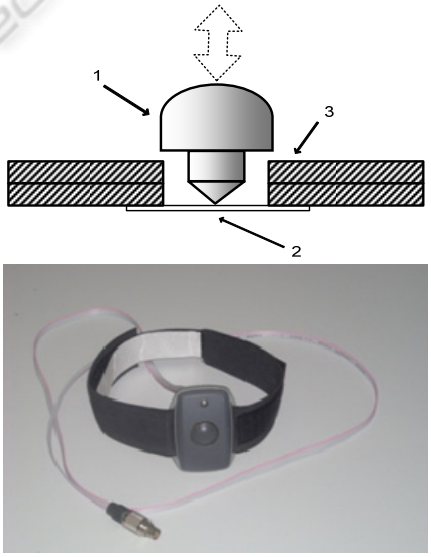

Figure 3: a) The PZ probe configuration, the arrow indicates the externally applied forces, (1) mushroom-shaped interface, (2) PZ disc sensor and (3) printed circuit board (PCB) and in b) the final probe is shown.

Figure 4 shows a typical response of the PZ (gray line) to an APW-like excitation (black line)

obtained in a dedicated test bench (Pereira *et al.*, 2009). As the microcontroller cannot sample negative voltages, the PZ signal is level shifted by a convenient DC value before being fed to the amplifier. Figure 5 depicts the level shift circuit.

Typically, if collected at the carotid site, it shows peak amplitudes of around 1V and exhibits a SNR in the order of 40 dB, allowing subsequent signal processing algorithms to run free from noise induced errors.

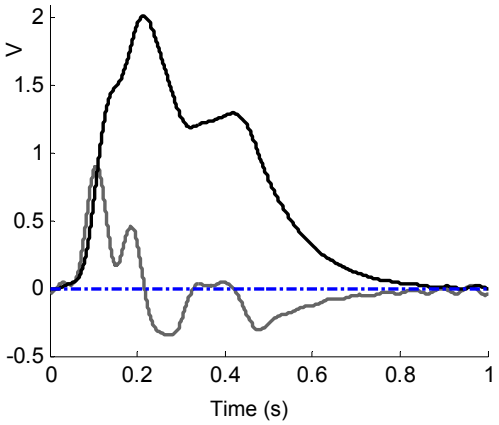

Figure 4: PZ signal (gray line) in response to APW excitation (black line). Blue line represents the DC value.

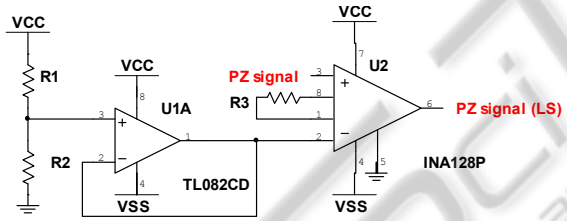

Figure 5: Level Shifter circuit used to PZ signal, before being fed to the ADC.

2.2 User Interface

As mentioned previously, data are uploaded via RS232 interface. The system is capable of sending captured data in real time with a signal acquisition rate high enough to be useful in real time hemodynamic monitoring, sampling rate of 1kHz.

This program reads the serial RS232 port and displays the data in a graph. It stores the received data in an individual text file for each measurement. This GUI is based on the one developed by Kerhuel (2010).

3 SOFTWARE MODULES

In the system design, the speed of computation and memory capacity are considered as top importance characteristics.

Microchip MPLAB v8.30 is used for building the modules using the C30 compiler for C programming, which simplifies code generation.

We also use the MathWorks Simulink platform to generate C code. It provides an interactive graphical environment in which the algorithms are developed in the form of block diagrams. With the aid of Real Time Workshop Embedded Coder it can be used to generate the target independent ANSI C code. The code generated can be included into MPLAB IDE projects. The flowchart below represents the programming stages. The program is based in the available blockset Embedded Target for PIC/dsPIC (Kerhuel, 2010).

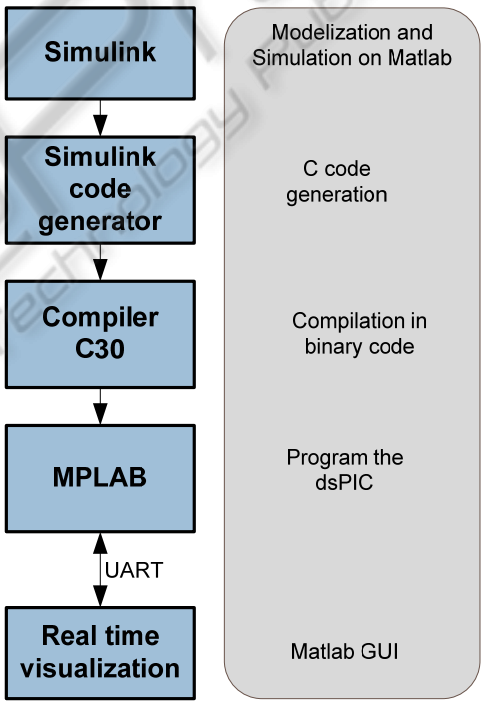

Figure 6: Flowchart of the stages of programming. The microcontroller transmits the data in real time through UART (universal asynchronous receiver/transmitter).

The C code includes the following software modules:

- Integration function
- Peak identification
- Baseline restoration
- Time Delay

3.1 Physiological Assumptions

We start by recalling that, due to the inherently capacitive nature of the sensor, its voltage signal will inevitably occurs in the shape of a time derivative of the APW and, consequently, the output stage will perform an integration of the signal in order to recover the original APW.

The prominent point's identification is an important task to identify different phases of the cardiac cycle, and for hemodynamic parameters extraction (Almeida *et al*, 2011).

As is well known, signal integration requires a periodic reset signal to avoid saturation. A time reference, for the reset signal must be identified in a pulse by pulse basis. The prominent peak in figure 4 (gray line) corresponds to the highest rise slope of the pressure waveform, few milliseconds after the beginning of the pulse (ventricular contraction). To identify this peak the PZ signal is fed to a peak detector (figure 7) formed by two peak stretchers (U3, D1, R4, C1 and U4, D2, R5, C2) and a comparator (U5).

To avoid false triggers the peak stretchers have different time constants, R4C1 and R5C2, chosen according to the typical signals, and one this is level shifted before being fed to the comparator.

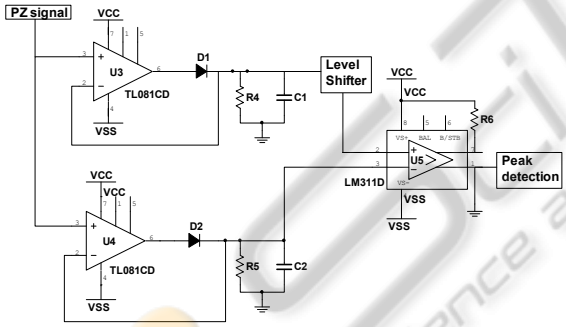

Figure 7: Peak detector circuit.

This mechanism contributes to the baseline restoration process. The elimination of baseline drift consists in forcing the foot of systolic pulses to start close to zero without affecting the shape of the signal.

Like many other biosignals, APW pulse drift essentially correlates to three sources: respiratory activity, variations in signal shape and signal jitter (defined as a random variation of the of the pulse period). If an operator holds the probe during data collection, an extra source of baseline drifts shows up due to the variations of its interaction with the patient. In *in vivo* tests a collar is used to eliminate the influence of the operator.

In a typical measuring session, the first few pulses are used just to gather the two main system adjustment parameters: time delay and baseline level, both to be used in the signal integration. Then real data collection starts for as long as possible (no discomfort for the patient that is asked not to swallow during data acquisition). Typically, one to two minutes allows the acquisition of a number of cardiac pulses high enough for the statistical processing that follows.

In figure 8 a flowchart of the system design is represented.

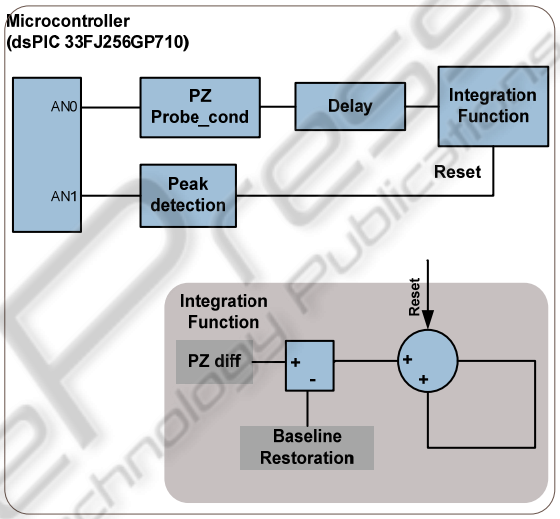

Figure 8: System overview. Acquisition/processing platform of the system.

4 RESULTS

The real time embedded system developed for cardiovascular applications was tested using cardiac simulated waveforms, synthesized using a weighted combination of exponential functions (Almeida *et al.*, 2010).

An Agilent 33220A arbitrary wave generator delivers the signal that excites the system via a test input coupled through a capacitor of the same order of magnitude of the sensor capacity itself).

In vivo tests are performed in some volunteers that granted their previous written, informed consent.

Peak detection (Figure 9) and delayed PZ waveform (Figure 10, black line) intermediate signals are shown. The relative error is computed from the data shown in Figure 11 a), comparing the integrated waveform with the excitation waveform (Agilent).

4.1 Peak Identification

The more prominent positive peak (green line) is identified, as shown in the figure 9 using the circuit described in figure 7.

As mentioned previously, this peak in the differentiated signal (PZ diff) corresponds to the highest rise slope of the arterial pressure waveform.

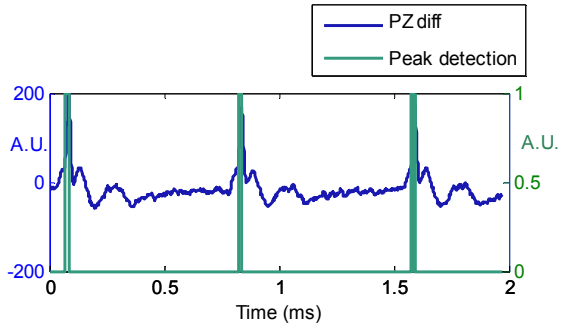

Figure 9: Results from the peak identification circuit.

4.2 Delay

The ideal delay value required, *tmin*, rising time of the most prominent peak in the PZ signal, not being possible to measure, so a time delay *td* is calculated for this purpose, To determine *td* we use information about systolic upstroke time, *tu*, of general APWs, Buteler (1961) measures the change in systolic upstroke time in patients with different diseases. The upstroke time varied between 110 ms and 230 ms. In our measurements a 110 ms time delay is used to perform the integration.

$$tu \approx 2 \times t\min \quad (1)$$

$$110 \leq tu \leq 230 \quad (2)$$

$$55 \leq td \leq 115 \quad (3)$$

Figure 10 shows the original PZ signal (gray line) and the delayed PZ signal (black line). Integration (blue line) is performed for the delayed PZ signal, as is visible.

4.3 Integration

The system is capable of recovering the APW from a PZ probe, as is shown in figure 11 where the excitation (blue) and the recovered waveforms (light blue) are shown.

Figure 11 b) plots the relative error (defined as the difference in amplitude from original and recovered waveform). Table 1 resumes the statistical parameters of the measurements. The data are

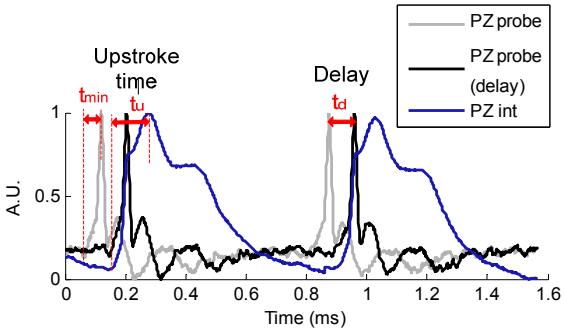

Figure 10: PZ waveform is represented in gray line and delayed PZ waveform in black line. The integrated waveform PZ int (integrated) represented in blue line was obtained from the delayed PZ waveform. *tmin* - ideal delay value, *tu*- time during of systolic upstroke, *td*- delay time used in the integration of the PZ waveform.

characterized by a mean value, a minimum and a maximum, as well as by the standard deviation (STD) for the relative error, between the excitation and the recovered waveforms. Maximum error occurs in the ascendant edge, 88 ms and minimum error occurs at 450 ms. The mean value is only 2.19 %, and the. STD is 2.57 %.

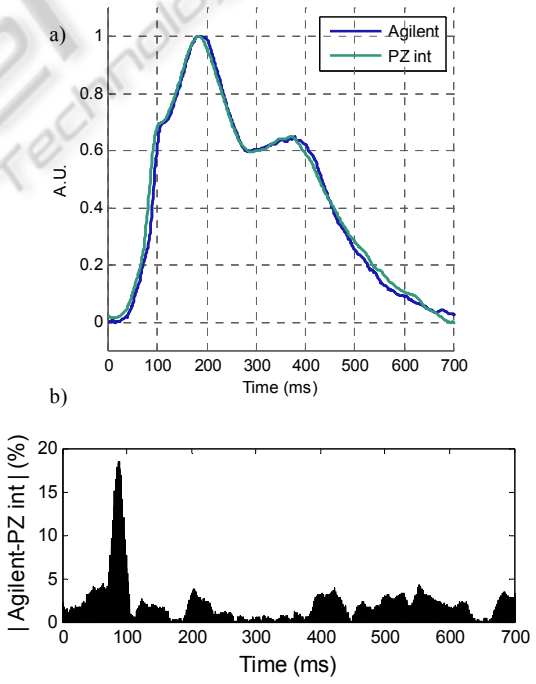

Figure 11: a) Integrated waveform (PZ int) and the excitation waveform (Agilent). b) The relative error between the original and the final waveform.

Table 1: Statistics information of measurements depicted in figure 11 b).

Maximum (%)	*Minimum (%)*	*Mean (%)*	*STD Deviation (%)*
18.520	0.001	2.1999	2.572

4.4 *In vivo* tests

To assess the capability of the system in distinguishing different points in APW morphology a small universe of volunteers was analysed.

The PZ probe is held by a collar and placed in the carotid artery site for *in vivo* data acquisition (figure 12).

The output shows a typical waveform where the most prominent points are easily identified: systolic peak (SP), reflection point (RP), dicrotic notch (DN) and dicrotic peak (DP). The effect of the baseline restoration mechanism that prevents baseline fluctuations along time can also be seen. In figure 13 a set of pulses, about 7 seconds are shown, where is possible identify a typical morphology of the APW with its prominent points identified in b).

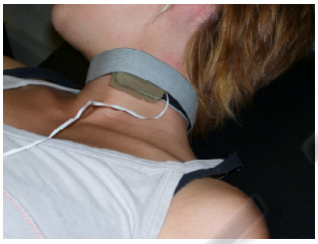

Figure 12: Our probe is held by a collar and placed over carotid artery.

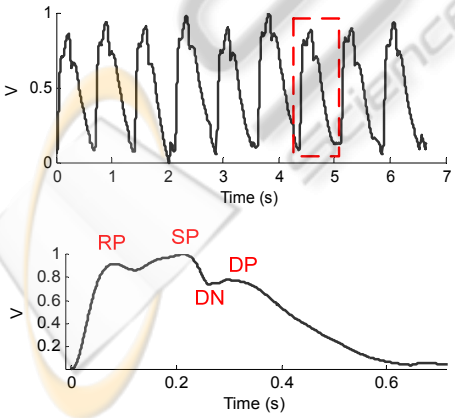

Figure 13: a) A set of pulses and in b) is shown a detail view of one pulse. RP-reflected point, SP-systolic peak, DN-dicrotic notch, DP-dicrotic peak.

5 CONCLUSIONS

In this paper, we presented the design of a real time cardiac monitoring system for APW capture. A PZ sensor was integrated in a signal acquisition circuit that communicates with a dsPIC. Application software running on the matlab was also developed to receive and plot APW signals.

Currently we are studying the clinical use of our probe, in medical environment, comparing our data with catheter collected data to prove that this system is a valid alternative with low cost associated.

Algorithms for the patient's information must be integrated in order to extract information about: heart rate, AI, PWV and reflection points (Almeida *et al* 2010, Almeida *et al* 2011, Pereira *et al*, 2010).

The proposed system was designated with user-friendly interfaces which easy the usability of this system and reduce the need of a long-time-training before usage. The minimal human intervention is a fundamental characteristic for this purpose.

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