33rd Annual International Conference of the IEEE EMBS
Boston, Massachusetts USA, August 30 - September 3, 2011

A Real time, Wearable ECG and Continous Blood Pressure Monitoring System for First Responders

David M. D. Ribeiro, Márcio F. M. Colunas, Fábio A. Ferreira Marques, José M. Fernandes and João P. Silva Cunha, *Senior Member, IEEE*

The study of stress and fatigue among First Responders is a major step in mitigating this public health problem. Blood pressure, heart rate variability and fatigue related arrhythmia are three of the main "windows" to study stress and fatigue. In this paper we present a wearable medical device, capable of acquiring an electrocardiogram and estimating blood pressure in real time, through a pulse wave transit time approach. The system is based on an existent certified wearable medical device called "Vital Jacket" and is aimed to become a tool to allow cardiologists in studying stress and fatigue among first response professionals.

FIRST Responders (FR) are among the professionals that are subjected to higher levels of stress and fatigue while on duty [1]-[2]. Recent studies have shown that 45% of the "on duty" deaths in these professionals are due to cardio-vascular diseases (CVD), a rate that is three times higher when compared to other professions [3]. The hypothesis of a direct relation between these two facts is likely but not yet proved. To be able to study this hypothesis and methodologies to manage stress and fatigue events among FR, one needs to be able to continuously monitor vital signs in real "on-duty" conditions. Although blood pressure (BP) is known to be highly related to stress and fatigue levels [3]-[6], it is usually not used in ambulatory scenarios, because the available measurement methods are neither suitable nor comfortable enough for continuous monitoring in daily life activities. Blood Pressure can be measured by invasive and non-invasive methods. The invasive method is generally used at hospitals and intensive care units via intra-arterial catheterization and provides more accurate data. Non-invasive BP devices became increasingly common in the last decade. These devices are less accurate but allow a higher degree of freedom, portability and easiness of use without the need of complicated medical procedures. In this paper we present a novel wearable medical device that combines electrocardiogram (ECG) and photoplethysmography (PPG) to estimate BP in real time. This wearable device allows for comfortable monitoring of subjects and including the estimation of their arterial blood pressure in an online manner, recurring to an algorithm based on the arterial pulse-wave transit time derived from ECG and PPG waves. In recent years Literature has presented several methods and devices regarding cuff less BP estimation. Especially methods based on the use of the PPG signal have shown great potential. McCombie et al. [7] used a method with 2 PPG sensors at a know distance. Although a promising method, the analysis is done off-line. Also the absence of ECG makes this method less efficient in monitoring stress and fatigue related arrhythmias and other ECG alterations. Another author describes the use of pulse transit time (PPT) derived from ECG and PPG for estimating BP levels [8]. Although the system uses a similar approach to that is described in this paper, it lacks of portability and is therefore not yet suitable as a wearable device. Furthermore one author suggests the use of natural hydrostatic pressure changes caused by specific movements of the subject [9]. The system uses ring sensor for measuring PPG, and pressure applied to the finger. Accelerometers are used to measure the height of the BP sensor and good results are obtained. Nevertheless, signal processing is done offline.

The wearable device presented here is an extension to the Vital Jacket® system [10]-[11]. The system is able to estimate BP levels in real-time using PTT derived from ECG and PPG signals. Furthermore, with some modifications (e.g. higher sample rate), the system can be used for heart rate variability (HRV) and fatigue related arrhythmias detection [12]. A small validation study of the developed system was performed and presented in order to assess the accuracy of the vital parameters extracted from this innovative system.

I. Materials and Methods

A. Experimental setup

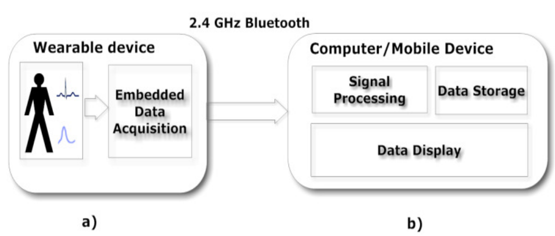

Fig. 1. System block diagram: wearable device (VJ) with sensors and embedded data acquisition box (a), handheld mobile device or laptop computer (b).

Experimental setup is based on the Vital Jacket® (VJ) that combines textile with microelectronics (figure 1a). It

Manuscript received June 24, 2011. This work was supported in part by Vital Responder Project funded by Portuguese Foundation for Science and Technology and FEDER grant :(ref CMU/PT/CPS/0046/2008).

David M. D. Ribeiro is with Institute of Electronics and Telematics Engineering of Aveiro (IEETA), Aveiro, University of Aveiro, Portugal (email: davidribeiro@ua.pt).

Márcio F. M. Colunas is with Institute of Electronics and Telematics Engineering of Aveiro (IEETA), Aveiro, University of Aveiro, Portugal (email: marcio@ua.pt).

Fábio A. Ferreira Marques was with Institute of Electronics and Telematics Engineering of Aveiro (IEETA), Aveiro, University of Aveiro, Portugal (email: fabiomarques@ua.pt).

José M. Fernandes is with the Department of Electronics, Telecommunications and Informatics and IEETA, University of Aveiro, Aveiro, Portugal (email: jfernan@ua.pt).

João P. Silva Cunha is with the Department of Electronics, Telecommunications and Informatics and IEETA, University of Aveiro, Aveiro, Portugal (email: jcunha@ua.pt).

currently has the EU CE 1011 Medical Device certification clearance, is compliant with the EU directive 42/93/CE and is produced under a ISO9001 and ISO13485 certified manufacturing process [11]. The Vital Jacket® is a long term ambulatory wearable ECG monitor and is the result of several years of development done at the Institute of Electronics and Telematics Engineering of Aveiro (IEETA). Its technology was licensed to a spin-off company called Biodevices S.A (www.biodevices.pt). Currently this device is being commercialized with focus on clinical diagnosis in cardiology and in monitoring high performance sports activity. As shown in (Figure 1), physiological signals acquired using a wearable device (VJ) are gathered in real-time through a textile embedded, custom build signal acquisition unit. It gathers ECG and PPG data but also allows to store all the data for off-line analysis or to transmit the data wirelessly to a computer or mobile device. In order to estimate the blood pressure levels in real-time, we extended the current VJ system (Figure 2) with a sensing system to acquire the PPG wave in form of a NONIN® Xpod 3012 module. The sensor is an ear probe also manufactured by NONIN®. The ECG electrodes used were manufactured by Ambu®, model Blue Sensor VL. The embedding of all these components was performed in a way that the system is wearable and comfortable, taking the form of a simple T-shirt, as is depicted in Figure 2. The combination of ECG and PPG signals allows for the estimation of a subject's BP levels. PPG and ECG signals are both acquired at a sampling frequency of 75 Hz. Blood pressure levels are calculated by software running on a computer or mobile device. An online interface was developed and BP levels are calculated and validated recurring to an online study. This study was performed on a group of 10 subjects of both genders between the ages of 22 to 62. Nine patients did not have any known cardiac issues and all had normal blood pressure levels. One subject has a known history of hypertension and diabetes (subject 7). For each subject, 60 minutes of ECG and PPG data were recorded with periodic measurements (each 10 minutes) of BP using a commercial oscillometric device (OMRON M6 Comfort). In order to ensure the quality of the PPG signal, the cuff was placed at one side (on the upper left arm), whereas the ear probe was placed on the right ear lobe. All subjects were in a sitting position and resting.

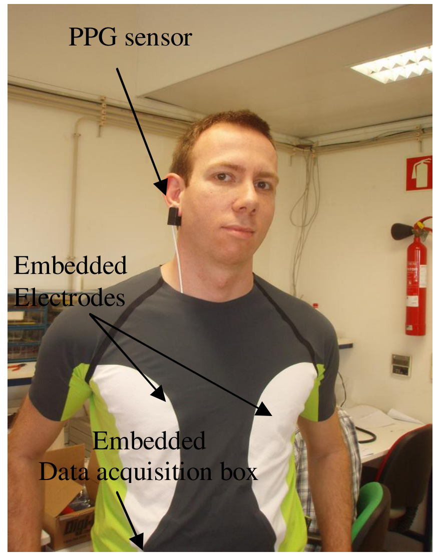

Fig. 2. Vital Jacket® and data acquisition box with embedded ear probe.

B. *Principle of operation*

The arterial pulse-wave transit time can be measured between the ECG R-wave and the PPG pulse wave, and has been shown previously to have a correlation with blood pressure [13]. Blood pressure is calculated according to the method first described in [13]. This method is based on an algorithm that uses pulse wave velocity and wave shape characteristics from continuous and synchronously acquired ECG and PPG signals. Blood Pressure is related to the inverse of the pulse arrival time (time interval from apex of ECG to onset of PPG) squared and the fractional change in pulse volume for each passing pulse [14]. Having synchronously acquired PPG and ECG data, the delay between the peak of the ECG R wave and the measured PPG pulse wave (1/Cdx) is calculated (Figure 3). Having 1/Cdx, the systolic (P_{si}) and diastolic blood pressure (P_{di}) values for the i_{th} pulse are estimated using the two following equations:

$$P_{si} = \left[k_s \times (C_{dx})_i^2\right] + k_{sys\_cal} \quad (1)$$

$$P_{di} = \left[k_d \times (C_{dx})_i^2\right] + \left[k_{IHR} \times IHR_i\right] + k_{dia\_cal} \quad (2)$$

Where $(C_{dx})_i$ is defined above and is the inverse of the delay between the R peak of the ECG and the 50% slope of the PPG wave, IHR_i is the instantaneous heart rate for the i_{th} pulse, k_s, and k_d are the fixed constants; k_{sys_cal} and k_{dia_cal} are the systolic and diastolic calibration constants, k_{IHR} is the constant related to the IHR. The introduction of the IHR factor becomes important since the BP of an individual changes with each particular heart rate, therefore diastolic BP is directly dependent on the IHR [13].

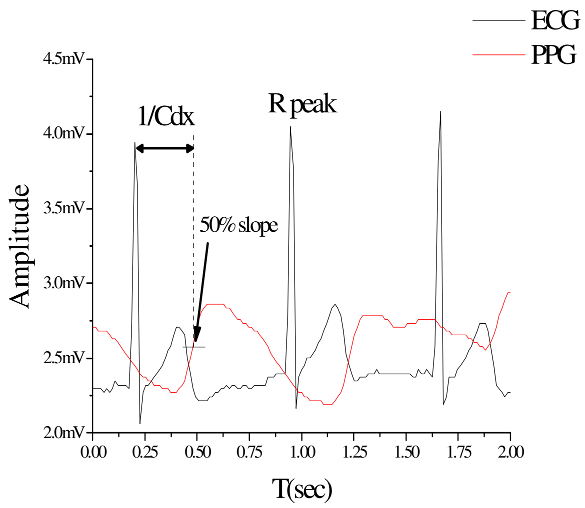

Fig. 3. Example signals obtained with 1/Cdx interval calculation

C. *Signal Processing*

In order to properly detect the QRS complex we implemented a Hamilton & Tompkins based method for an accurate ECG peak detection. This method has a near 99.8% for both sensitivity and positive predictivity on tests made using MIT/BIH and AHA arrhythmia databases. It relies on a dynamic threshold estimation[15][16], which is an adaptation to changing signal characteristics, for improving false positives detection[16]. To detect the PPG slope the algorithm first applies a differentiation to obtain slope information, then squares it, followed by moving average to smooth the signal, analyzing sample by sample using the same approach for threshold estimation as the ECG peak detector to detect the maximum points [15][16]. For 50% slope calculations, the algorithm then searches back in order to compensate the delay. Having correctly obtained values for 1/Cdx and IHR, we now can estimate BP levels using the equations (1) and (2). Furthermore, for a successful estimation, constants k_s, k_d, k_{IHR}, k_{sys_cal}, and k_{dia_cal} need to be determined. These constants are obtained through a calibration process by using 3 measurements of systolic and diastolic BP, obtained by using an oscillometric device. These values, which have to be acquired for each individual subject, substitute Psi and Pdi in equations (1) and (2) as well as means of 20 computed values for Cdx and IHR obtained before cuff inflation. With these calibration values it is possible to obtain the needed constants.

D. *Online interface and software*

The software interface created for BP estimation depicts ECG and PPG waves, as well as the calculated blood pressure values, estimated in real time, after calibration (Figure 4). The software was written in C# using Microsoft Visual Studio IDE (Integrated Development Environment), applying object-oriented programming best-practices. The interface was designed with a minimalistic approach in order to be graphically appellative and effective, allowing a simple and easy access to all the information and options. It allows connection management, calibration settings, resetting and depicts PPG and ECG signals in real time as well as visualization of estimated BP levels (Figure 4c) both numerically and graphically through a plot designed for that particular purpose. All the gathered and calculated data can be recorded to text files for future analysis.

II. Results

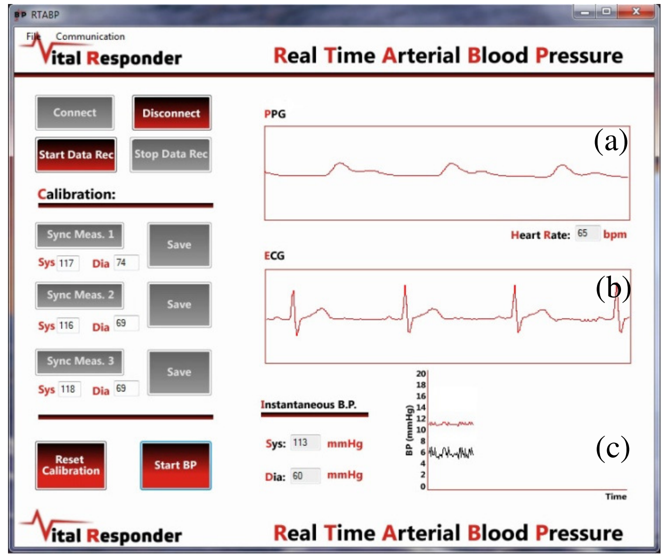

Fig. 4. System interface were (a) PPG wave (b) ECG wave (c) BP estimated values.

The data presented here was acquired in real-time using the developed system described in this paper. As mentioned above we used 3 measurements to calibrate the algorithm (only the 3rd of these is depicted in Figure 5 arrow). After the calibration, we have taken 6, 10 min separated measurements to directly compare them with the BP values estimated by software. The results are presented in tables I and II.

TABLE I. Constants Values ,mean Cdx and IHR values for all the individuals

Subj.	Mean (Cdx)	Mean (IHR)	Ks	Kd	KIHR	Ksys_cal	Kdia_cal
1	5.26	75	-1.22	-1.43	-0.50	138.6	64.0
2	5.77	78	-0.69	-1.00	-0.81	132.9	174.8
3	5.15	80	0.44	0.49	0.51	90.0	15.5
4	4.82	72	-7.63	3.23	-0.08	281.3	-11.9
5	5.62	73	-0.64	-1.67	0.72	134.1	73.1
6	4.74	80	-0.29	0.08	-0.43	110.7	101.3
7	4.37	48	1.17	0.29	-29.20	107.2	201.9
8	5.35	83	0.67	-0.67	0.84	85.3	10.2
9	3.90	82	-2.05	9.63	6.06	156.3	-530
10	5.51	67	0.83	-0.52	0.15	84.02	73.1

The estimated constants, mean values of Cdx and IHR for each individual are presented in Table I. These personalized constants, obtained through the calibration procedure, were used to estimate BP for each individual subject using the method described above. Depicted in figure 5 are the values obtained during a full hour recording. The calculated blood pressure levels are compared directly against the values obtained using a oscillometric device.

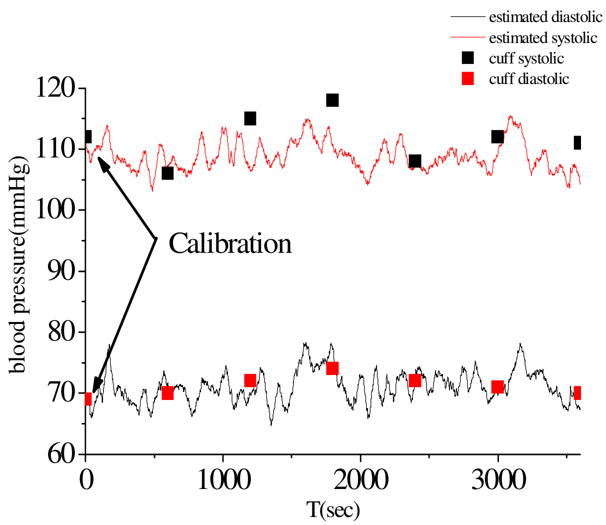

Fig. 5. Measured and estimated systolic and diastolic values obtained from a volunteer (average error of 5.14 mmHg for sys and 2.14 mmHg for dia).

In Table II, we can see the BP values estimated by software and simultaneously obtained by oscillometric method. We can clearly see differences in blood pressure levels between subjects with relative high levels (subjects 7 and 9) and subjects with lower (normal) blood pressure levels. Furthermore we can see that the results obtained through the oscillometric device and the software display some small differences with an average error of 3.5% for systolic and 5.5% for diastolic values, which corresponds to an accuracy of 3.8 mmHg and 3.9 mmHg respectively. These differences can be partially explained by the different methods used to obtain these results. Furthermore this error is within the accuracy reported for automated sphygmomanometer, which is reported as ± 5mmHg [17]. Also, the precision of the PPG slope detection algorithm contributes to some of the in-accuracies found in the results. Furthermore the sampling rate of 75Hz can increased to further improve the accuracy of the system. These modifications will be investigated in future work.

III. DISCUSSION AND FUTURE WORK

From the obtained results we can conclude that this method provides good estimations for BP levels, needing only a prior calibration. The VJ and the PPG ear probe were considered comfortable by all the subjects and can be considered as a good indicator for future tests that aim for prolonged monitoring periods. Alternatives to calibration via electronic sphygmomanometer are being studied because this method is recognized as not being accurate enough to obtain the desired results. Future work will also focus on electrodes embedded into clothing (dry active electrodes have already been tested) [18], flexible electronic modules and its miniaturization as well as new sensing approaches that will result in a more complete system with additional monitoring options. An alternative to the PPG ear probe is also being explored, in order to decrease the sensitivity to induced motion artifacts. A porting of this system to a mobile device is also being explored because this is considered to be the most effective manner to assure the portability of the system while keeping all of its functionality and obtaining a truly wearable device. Although the results were obtained under laboratory conditions we can conclude that this method and its real-time implementation is suitable to be used in field experiments with FR in action in the near future.

ACKNOWLEDGMENT

The authors would like to thank all volunteers for their time and support. This work is supported by an FCT project grant CMU-PT/CPS/0046/2008.

TABLE II. MEASURED AND CALCULATED BLOOD PRESSURE VALUES

Subj	(1)Mean Sys. mmHg	(2)Mean Sys. mmHg	|error| (%)	(1)Mean Dia. mmHg	(2)Mean Dia. mmHg	|error| (%)
1	106±2.4	104±3.2	1.9	62±1.1	60±3.8	3.3
2	107±3.7	109±2.9	1.9	66±4.4	78±8.1	15.4
3	99±2.2	100±3.2	1.0	68±1.4	71±9.3	4.2
4	98±6.7	110±12.7	12.2	61±2.3	61±8.3	0.0
5	112±3.2	110±4.3	1.8	79±3.3	71±7.4	11.3
6	109±3.7	103±0.5	5.5	67±2.5	68±2.2	1.5
7	126±4.8	131±8.2	4.0	73±3.7	69±6.5	5.8
8	106±1.2	104±2.4	1.9	66±2.3	61±4.3	8.2
9	124±2.7	127±2.6	2.4	86±2.0	88±6.0	2.3
10	110±3.4	107±2.8	2.7	67±4.2	69±2.1	2.9
		Average	3.5		Average	5.5

(1)Measured by cuff
(2) Estimated values

REFERENCES

[1] D. A. Alexander and S. Klein, "First Responders after disasters: a review of stress reactions, at-risk, vulnerability, and resilience factors," *Prehosp Disaster Med.*, vol. 24, pp. 87-94, 2009.

[2] D. M. Benedek, C. Fullerton, R. J Ursano., "First Responders: Mental Health Consequences of Natural and Human-Made Disasters for Public Health and Public Safety Workers," *Annual Review of Public Health*, vol. 28, pp. 55-68, 2007

[3] S.N. Kales, E.S. Soteriades, C.A. Chistophi, and D.C. Christiani, "Emergency duties and deaths from heart disease among firefighters in the United States," *N Engl J Med*, vol. 356, pp. 1207 – 1215, March 2007

[4] N. Hjorstkov, D. Rissén, A. K. Blangsted, N. Fallentin, U. Lundberg, K. Søgaard, "The effect of mental stress on heart rate variable and bloodpressure during computer work" *Eur J appl Physiol,* vol.92, pp.84-89, Febr 2004.

[5] P. Armario, R. H. del Rey, M. Martin-Baranera, M. G. Almendros, L. M Geresuela, H, Pardell. "Blood pressure reactivity to mental stress task as a determinant of sustained hypertension after 5 years of follow-up" J of Human Hypertension, vol.17,pp. 181-186, Nov 2002.

[6] D. I. Melvill, E. B. Raftery, "Blood pressure changes during acute mental stress in hypertensive subjects using the Oxford intra-arterial system" J of Phychosomatic Research, Vol25, Issue 6, pp. 487-497, May 2002.

[7] D. B. McCombie, A. T. Reisner, H. H. Asada, "Adaptive blood pressure estimation from wearable PPG sensors using peripheral artery pulse wave velocity measurements and multi-channel blind identification of local arterial dynamics", Engineering in Medicine and Biology Society, 2006. EMBS '06.28th Annual International Conference of the IEEE,New York, NY, Aug. 30- Sept. 3, 2006.

[8] C.C.Y. Poon, Y.T. Zhang, “Cuff-less and Noninvasive Measurements of Arterial Blood Pressure by Pulse Transit Time,”, IEEE-EMBS (Engineering in Medicine and Biology Society), 27th Annual International Conference, Shanghai, China, Jan. 17-18, 2005

[9] P.A. Shaltis, A. T. Reisner, H. H. Asada, "Cuffless Blood Pressure Monitoring Using Hydrostatic Pressure Changes," Biomedical Engineering, IEEE Transactions on , vol.55, no.6, pp.1775-1777, June 2008

[10] J.G.P. Rodrigues, F. Vieira, T.T.V. Vinhoza, J. Barros, J.P.S. Cunha, “A non-intrusive multi-sensor system for characterizing driver behaviour” Intelligent Transportation Systems (ITSC), 2010 13th International IEEE Conference on, pp.1620-1624, 19-22 Sept. 2010

[11] J. P. S. C. et al, "Vital-Jacket®: A wearable wireless vital signs monitor for patients' mobility in Cardiology and Sports,"*4th International Conference on Pervasive Computing Technologies for Healthcare 2010* Campus Garching, Technische Universitat Munchen, Munchen Germany, 2010.

[12] U. Rajendra Acharya*, et al.*, "Heart rate variability: a review," *Medical and Biological Engineering and Computing,* vol. 44, pp. 1031-1051, 2006.

[13] S. O. H. et al, MD, "An Evaluation of a New Continuous Blood Pressure Monitoring System in Critically Ill Patients," *Journal of Clinical Anesthesia,* vol. 12, pp. 509 –518, 2000.

[14] P. Pandian*, et al.*, "Smart Vest: Wearable multi-parameter remote physiological monitoring system," *Medical engineering & physics,* vol. 30, pp. 466-477, 2008.

[15] P. Hamilton, “Open Source ECG Anaysis”, *Computers and Cardiology”*, vol. 29, pp. 101-104, Sept 2002.

[16] P. S. H. a. W. J. Tompkins, "Quantitative investigation of qrs detection rules using the mit/bih arrhythmia database," *IEEE Trans Biomed Eng,* vol. 33, pp. 1157-65, 1986.

[17] K. Soueidan, C. Silu, H. R. Dajani, M. Bolic, V. Groza, "The effect of blood pressure variability on the estimation of the systolic and diastolic pressures," Medical Measurements and Applications Proceedings (MeMeA), 2010 IEEE International Workshop on , vol., no., pp.14-18, April 30 2010-May 1 2010

[18] D. M. D. Ribeiro, L. S. Fu, J. P. S. Cunha, L. D. Carlos, “A Novel Dry Active Bio-signal Electrode based on an Hybrid Organic-inorganic Interface Material,” IEEE J. sensors, accepted for publication.