Implementation and Evaluation of an Adaptive Method for Reduce the Respiration Influence on Heart Rate Variability

Raymundo Cassani, Student Member, IEEE[1], Juan Carlos Sanchez Member IEEE[1], Raul Martinez[2]
[1]Department of Graduate Studies and Research, ESIME Culhuacan-IPN, Mexico D.F., Mexico.
[2]Department of Electromechanic Instrumentation of the National Institute of Cardiology, Mexico D.F., Mexico.
Phone +52 (55) 5624 2058 E-mail: raymundo.cassani@ieee.org

Abstract —**In this paper it is described the implementation and evaluation of an adaptive method that has as aim to cancel the influence of the respiratory signal over the Heart Rate Variability (HRV) signal in order to enhance the power estimation of its spectral components. The method consists in an Adaptive Noise Cancellation (ANC) structure that uses a Finite Impulse Response (FIR) filter together with the Normalized Least Mean Squares (NLMS) adaptation algorithm. Respiration and electrocardiogram (ECG) signals were obtained simultaneously using 240Hz sampling frequency. After data acquisition, tachogram was derived from ECG signal to obtain its HRV signal; then ANC filtering is applied, reducing variations due to respiration from HRV signal. This method was evaluated for spontaneous and for two controlled respiration frequencies. 6-minutes registers were taken form 10 people during the 3 different scenarios giving a total of 30 registers. Power Spectral Density (PSD) was estimated from the HRV signal before and after filtering and compared. At the results, it is observed that frequency components related to respiration are cancelled in the HRVs PSD, reaching an improved estimation of the control exerted by the Autonomic Nervous System (ANS) over the heart rate.**

Keywords —**Adaptive Filtering, Adaptive Noise Canceller, Autonomic Nervous System, ECG, HRV, NLMS, Respiration, RSA.**

I. Introduction

The Autonomic Nervous System (ANS) is the component of the peripheral nervous system that is responsible of adapting the performance and functions of internal organs, including heart, for varying internal and external changes [1]. The ANS is subdivided into two parts, sympathetic system and parasympathetic (or vagal) system; those subsystems have effects considered antagonistic on the cardiac function. While the sympathetic increases the function of the heart (rate, contractility, relaxation and conduction speed), the vagal reduces it. [2]

Nowadays, the main disease that affects the ANS is diabetes Mellitus, [3] and [4]; in addition since the last three decades it has been observed an important relationship between the balance on the sympathetic-vagal tone (a measure of the ANS condition) and the cardiovascular mortality [5], this situation has led to the development of some methods that quantitatively evaluate the ANS.

The analysis of the Heart Rate Variability (HRV) and its Power Spectral Density (PSD) are auxiliary elements for the estimation of the ANS status [6].

Heart Rate (HR) is defined as the number of beats during a time period (usually 1 minute), and it is controlled by the Sinus Node which in turn is controlled by the ANS through its subsystems [2]. This regulation on the HR is produced beat by beat and has as objective to keep the homeostasis [7]. The instantaneous HR is calculated as function of the RR interval (RR_i) that is the time interval between two consecutive R peaks.

HRV is a measure of the change of the instantaneous HR for each beat (Fig 1). Because RR_i is not a constant, HRV is a non-uniformly sampled signal [8].

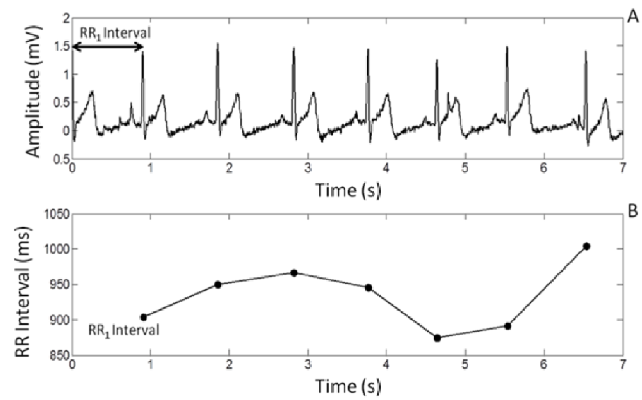

Fig. 1. (A) ECG signal. (B) HRV signal, where markers are located at the occurrence time of the R peaks.

HRV changes synchronously with the respiration (for every single breath); this phenomenon is known as Respiratory Sinus Arrhythmia (RSA) and has its origin mainly in breathing mechanics. During the inspiration, HRV is incremented, and it is reduced in the expiration, [9] and [10]. Respiration influence is the most important on the HRV signal [11].

A method used for reduce the influence of respiration on HRV controls the Respiration Frequency (RF) at frequencies out of the interest band [12]. Other methods make use of metronome-breathing at a fixed frequency that allows to underestimate the influence of the respiration on the HRV measurement [13] and [14]. Methods that control

978-1-4673-4900-0/13/$31.00 ©2013 IEEE

the RF change the natural behavior of the ANS functions, modifying the sympathetic-vagal balance. Moreover these methods need a conscious control on breathing; therefore, they cannot be used in sleep or coma periods.

When signals are contaminated and degraded by background noise whose spectrum overlaps the signal's one, the approach of conventional linear filters with fixed coefficients would lead to unacceptable distortion of the desired signal. That is the reason why it is more appropriate to employ an adaptive filter [15].

Using the Adaptive Noise Canceller (ANC) structure, it is possible to cancel the respiration influence over the HRV signal even with spontaneous respiration.

II. METHODOLOGY

Based on physiological fundaments presented in [1], [2] and [6]-[10]; a linear model for the influence of the respiration on the HRV signal through the RSA is proposed (Fig. 2).

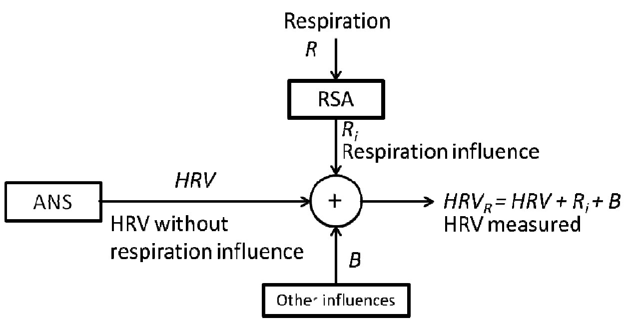

Fig.2. Linear model for the respiration influence on the HRVsignal.

The system utilized is divided in two stages: Conditioning & Acquisition and Processing [16] as shown in Fig. 3.

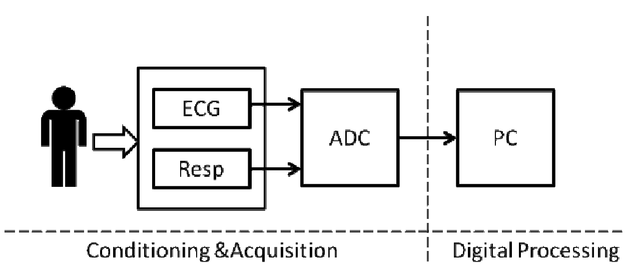

Fig.3. Block diagram of the system.

A. *Signal Conditioning & Acquisition.*

ECG signal was recorded with a one-lead electrocardiograph, Lead I was used. Respiration signal was detected by pressure changes inside a reference balloon which is harnessed to the subject's thorax; this technique is effective for recording nose and mouth breathing. [16]

ECG and respiration signals were acquired simultaneously at 240 samples per second, during 6 minutes [5] through a Data Acquisition (DAQ) board NI PXI 6251. Digitalized signals were saved as Comma Separated Values (CSV) files, for later processing.

B. *Digital Signal Processing.*

Signal processing was carried out offline using the CSV files obtained in the previous stage. The Figure 4 shows the block diagram for the Digital Signal Processing stage.

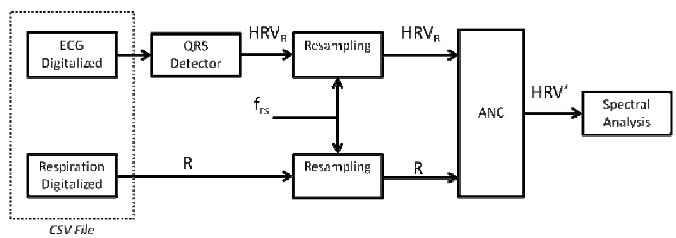

Fig. 4. Digital Signal Processing stage.

The QRS Detector has as input an ECG signal, and, as output, a series with the occurrence time of the R peaks ($\Theta_1, \Theta_2, \ldots$). After this automatic QRS detection, it was performed a visual examination to find and remove possible false negatives and false positives. With this information the HRV_R signal was calculated using (1) [17].

$$RR\ (i) = \theta_{i+1} - \theta_i \tag{1}$$

From Fig. 2; HRV_R is the sum of HRV, the respiration influence (R_i), and other influences (B). Because HRV_R is not a uniformly sampled signal, it was required to be resampled using a frequency of 5 Hz (f_{rs} in Fig. 4). Parallel to this process, respiration signal (R) which was sampled at 240Hz, was decimated to obtain R signal with the same sample frequency than HRV_R.

The Figure 5 shows the ANC structure derived from the linear model proposed.

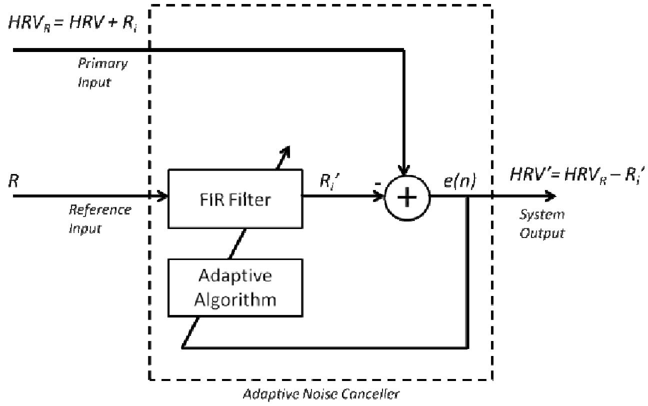

Fig. 5. Adaptive Noise Canceller structure.

The NLMS algorithm was used to adjust the coefficients of the FIR filter sample to sample, in order to minimize the mean square error [15]. The upcoming coefficient weight vector is calculated using (2)

$$\mathbf{W}(n+1) = \mathbf{W}(n) + \frac{\mu}{\alpha + \mathrm{R}^{\mathrm{T}}(n)\mathrm{R}(n)}\mathrm{R}(n)e(n) \qquad (2)$$

where $\mathbf{W}(n+1)$ is the new weight vector at the *(n+1)th* sampling instant; term μ is the step size and defines the stability and convergence rate of the adaptive filter. The expression $\mathbf{W}(n)$ and $\mathrm{R}(n)$ are the weight vector and the delayed input vector (respiration signal) at the *nth* sampling instant. The product $\mathrm{R}^{\mathrm{T}}(n)\mathrm{R}(n)$ denotes the power of the delayed input vector and α is a constant that has as function avoid extremely high values in the convergence factor. Finally *e(k)* is the instantaneous local error given by (3)

$$e(n) = HRV'(n) = HRV_R(n) - \mathbf{W}^{\mathrm{T}}(n)\mathrm{R}(n) \qquad (3)$$

where the term $\mathbf{W}^{\mathrm{T}}(n)R(n)$ is the filter output, which is named as R_i' and it is the estimate of R_i . Therefore *HRV'* is the approximation of *HRV*. The value for the step size μ was heuristically for each record. Delay and correlation between *R(n)* and $HRV_R(n)$ were used to choose the filter length. The order of the vectors and filter was set on N=300.

After filtering, Power Spectral Density (PSD) was derived from HRV_R and *HRV'* during the last 5 minutes of each record and compared. In addition Low Frequency (LF) power and High Frequency (HF) power were calculated. LF range comprehends from 0.04 to 0.15Hz, and HF band is defined from 0.15 to 0.4Hz [5].

III. Results

The methodology was implemented using MATLAB, and it was evaluated by the following scenarios:

1) Controlled RF inside the HF band, RF = 0.25Hz.
2) Controlled RF beyond HF band, RF = 0.6Hz.
3) No controlled RF.

RF control was guided by a visual indicator [16]. Results shown from Fig. 6 to Fig. 8 correspond to a selected subject during the 3 scenarios. Figures are in dB using as reference unit $1ms^2/Hz$

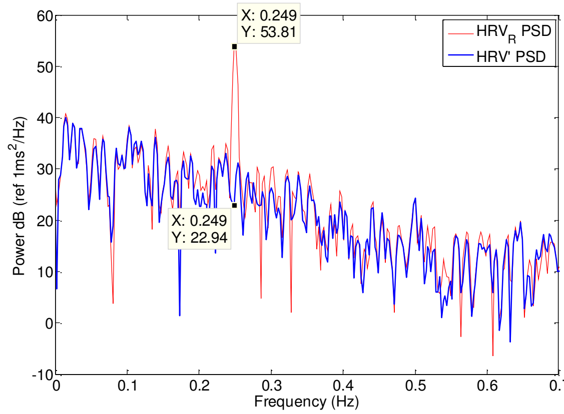

Fig. 6. PSD for HRV_R (red) and PSD for HRV' (blue) Scenario 1.

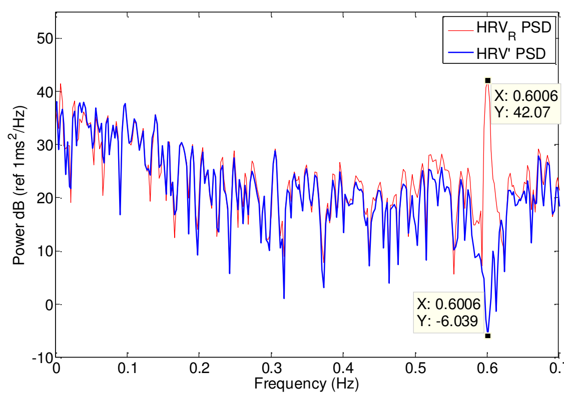

Fig. 7. PSD for HRV_R (red) and PSD for HRV' (blue). Scenario 2.

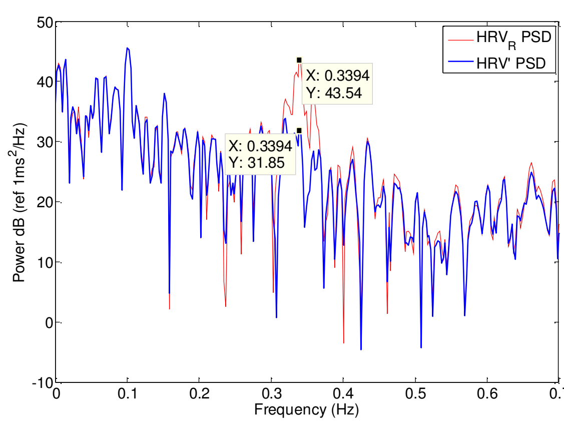

Fig. 8. PSD for HRV_R (red) and PSD for HRV' (blue). Scenario 3.

In Fig 6 it is visible an attenuation of -30.9 dB in the frequency of interest (0.25Hz). Similarly, in Fig. 7 it is appreciable a change of -48.1 dB at 0.6Hz. In Fig. 8 the cancellation is over a range of frequencies. For scenario 3 (Fig. 8), Respiration Band (RB) was defined as the respiration bandwidth (and it is obtained from the Respiration PSD). BR for the selected subject under scenario 3 (Fig. 8) was from 0.30Hz to 0.38Hz. BR was different for every subject. Table I shows a comparison between the attenuations in LF, HF and the frequency components of interest for each scenario, 0.25Hz, 0.60Hz and BR respectively.

TABLE I
COMPARISON BETWEEN SCENARIOS.

	Scenario 1	Scenario 2	Scenario 3
Frequency components of interest	0.25Hz	0.60Hz	BR
LF change (mean)	-0.22 dB	0.02 dB	-0.86 dB
HF change (mean)	-5.75 dB	-0.71 dB	-3.27 dB
Change in Frequency components of interest (mean)	-26.55 dB	-16.91 dB	-5.63 dB

In scenario 1, just HF power is affected because RF is inside this band; during scenario 2, LF and HF do not suffer important attenuation because RF is beyond these bands. In scenario 3 it can be seen that HF is attenuated, because for most of the subjects their BR share frequency components with HF.

IV. CONCLUSION

Other method to reduce the respiration influence from the HRV signal is the Independent Component Analysis (ICA), which is presented in [18]. This method has an inherent disadvantage, the scaling problem, scale the filtered signal would lead to a miscalculation of its PSD. A method based in Multiscale Principal Component Analysis (MSPCA) is presented in [19], within this work the results are obtained during fixed respiration rates.

The evaluation of the adaptive method proposed has had a successful performance reducing the respiration influence on the HRV signal during fixed and spontaneous respiration rates without affecting other frequency components.

This methodology could be implemented in medical protocols that evaluate the ANS on patients who suffer from different neuropathologies.

REFERENCES

[1] *Handbook of Clinical Neurology Vol 74(30) The Autonomic Nervous System Part I*. O. Appenzeller. 1999 Elsevier Science B.V.

[2] Klabunde R. "*Cardiovascular Physiology Concepts*". 2nd Edition. LWW, 2011 http://www.cvphysiology.com/index.html Aug 2012

[3] The DANA Foundation. https://www.dana.org/news/brainhealth/detail.aspx?id=9780 Sept 2012

[4] Department of Neurology, Mayo Clinic. http://mayoresearch.mayo.edu/mayo/research/neurology/autonomic_nerve.cfm Sept 2012.

[5] *Task Force of the European Society of Cardiology and the North American Society of Pacing and Electrophysiology. Heart rate variability: standards of measurement, physiological interpretation, and clinical use*. Circulation 93: 1043-1065. 1996.

[6] S. Akselrod, D. Gordon, F. A. Ubel, D. C. Shannon, A. C.Berger, and R. J. Cohen. *"Power spectrum analysis of heart rate fluctuation: a quantitative probe of beat-to-beat cardiovascular control"*. Science 213:220–222 (1981).

[7] Biocom Technologies. *"Heart Rate Variability Basics"* http://www.biocomtech.com/hrv-science/heart-rate-variability-basics Oct 2012.

[8] R. D. Berger, S. Akserod, D. Gordon, R. J. Cohen, *"An Efficient Algorithm for Spectral Analysis of Heart Rate Variability"*, *Biomedical Engineering, IEEE Transactions on*, vol.BME-33, no.9, pp.900-904, Sept 1986.

[9] J. A. Hirsch and B. Bishop, *"Respiratory sinus arrhythmia in humans: How breathing pattern modulates heart rate", Am. J. Physiol.*, vol. 241, pp. H620–9, Oct. 1981.

[10] Womack B.F *"The analysis of Respiratory Sinus Arrhythmia Using Spectral Analysis and Digital Filtering"*. IEEE Transactions on Bio-Medical, 1971.

[11] Jongyoon Choi; Gutierrez-Osuna, R.; , *"Removal of Respiratory Influences From Heart Rate Variability in Stress Monitoring,"* Sensors Journal, IEEE , vol.11, no.11, pp.2649-2656, Nov. 2011

[12] Joost F et all *"The importance of high-frequency paced breathing in spectral baroreflex sensitivity assessment"* J of Hypertension, vol. 18, no. 11 pp1635-1644

[13] R. Martinez-Memije, B. Estañol, O. Infante, E. Suaste. *"Asociación de la variabilidad de la frecuencia cardiaca y de la variabilidad de áreas pupilares en sujetos sanos, con respiración controlada" Congreso Latinoamericano de Ingeniería Biomédica* 2007.

[14] R. Martinez-Memije, B. Estañol, O. Infante, E. Suaste. *"Asociación de la variabilidad de la frecuencia cardiaca y de la variabilidad de áreas pupilares en diabéticos con neuropatía, con respiración controlada" XXXI Congreso Nacional de Ingenieria Biomedica* 2008.

[15] E. C. Ifeachor, and B. W. Jervis, *"Digital Signal Processing. A Practical Approach."* Great Britain. 1993 Addison-Wesley.

[16] Cassani, R.; Mejia, P.; Tavares, J.A.; Sanchez, J.C.; Martinez, R.; *"Adaptive filtering for respiration influence reduction on Heart Rate Variability"* CCE, 2011 8th International Conference on , vol., no., pp.1-5, 26-28 Oct. 2011

[17] L. Sörnmo and P. Laguna, *"Bioelectrical Signal Processing in Cardiac and Neurological Applications"*, Elservier Academic Press, 2005.

[18] S Tiinanen, M Tulppo and T Seppanen, *"RSA Component Extration fron Heart Rate Signal by Independent Component Analysis"*, Computers in Cardiology, 2009 , vol., no., pp.161-164, 13-16 Sept. 2009

[19] Widjaja D., Van Diest I., Van Huffel S., *"Extraction of Direct Respiratory Influences from the Tachogram using Multiscale Principal Component Analysis"*, in Proc. of the 7the International Workshop on Biosignal Interpretation (BSI2012), Como, Italy, Jul. 2012.