Adaptive Filtering for Ventricular Repolarization Variability Assessment

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Abstract

The purpose of this study was to assess ventricular repolarization variability canceling the influence of heart rate variability using adaptive filtering. Short-term recordings, at supine rest and after head-up tilt, were studied from forty-seven subjects (athletes, cardiac transplant recipient, and normal subjects). RR and RT intervals were extracted to obtain heart rate variability (HRV) and ventricular repolarization variability (VRV) time series respectively. A least mean squares adaptive filter was applied to extract the VRV unrelated to HRV (VRVu). Power spectral densities were estimated from HRV, VRV and VRVu series, and LF, HF were extracted for each power spectrum. Also, percentage power ratio between VRVu and VRV at LF and HF were calculated. Results showed that: i) the heart rate modulation on the ventricular repolarization phenomena showed to be more important in LF band, ii) HF oscillation in VRV is not modulated by the autonomic nervous system.

1. Introduction

It is well known that most of the ventricular repolarization variability is influenced mainly by the heart rate and autonomic nervous system (ANS) tone [1-2]. Heart rate variability (HRV) describes variation of both instantaneous heart rate and RR intervals. Ventricular repolarization variability (VRV) is usually measured by the oscillations between consecutive QT or RT intervals.

Three main spectral components are distinguished in a spectrum calculated from short-term 5-minute recordings of ventricular repolarization time series: very low frequency (VLF), low frequency (LF), and high frequency (HF) components. The distribution of the power and the central frequency of LF and HF are not fixed but may vary in relation to changes in autonomic modulations. The physiological origins these oscillations are explained only partially and the interpretation of the spectral parameters resulting from VRV sequences remains incomplete [2-3].

Porta et al, using dynamic parametric model, concluded that VRV unrelated to HRV has its most important frequency components in the very low frequency band (VLF) [4]. Moreover, a recent work shows that HF rhythm in VRV series seem to be related to the mechanical effect of respiration [5]. But VRV remains a not fully understood issue.

These controversial results led us to use adaptive filter techniques to explore relationships between heart rate and ventricular repolarization phenomena. This procedure allows to cancel the influences of heart rate in the ventricular repolarization variability obtaining a sequence unrelated with HRV (VRVu).

From a methodological point of view, adaptive filtering appears as an interesting tool to explore the HF and LF components in VRV unrelated to HRV series.

Adaptive filter have been recently used to provide accurate heart rate variability indices of sympathetic and parasympathetic activity [6].

The aims of this work were to study the LF and HF components in VRV power spectra and the percentage power ratio unrelated to heart rate variability, to reflect the state of sympathovagal balance.

2. Methods

2.1. System formulation

In this work, an adaptive noise canceling technique was used [7]. Basically, an adaptive noise canceller is a dual-inputs, closed-loop adaptive feedback system, as seen in figure 1.

The system analyses the beat-to-beat series of the HRV series from the RR durations and VRV series from the RT apex (RT max) periods, after subtraction of their mean values. From these two signals, an adaptive process, which involves the automatic adjustment of the tap weight of the filter in accordance with the estimation error, allows to cancel the heart rate variability from the ventricular repolarization variability and obtain the part unrelated to heart rate variability (VRVu)

From figure 1, *HRV(n)* is the heart rate variability at *n*th cardiac beat, *VRV(n)* is the ventricular repolarization variability at *n*th cardiac beat, *HRVy(n)* is the filter output signal, *VRVu(n)* is the error signal and constitute the overall system output.

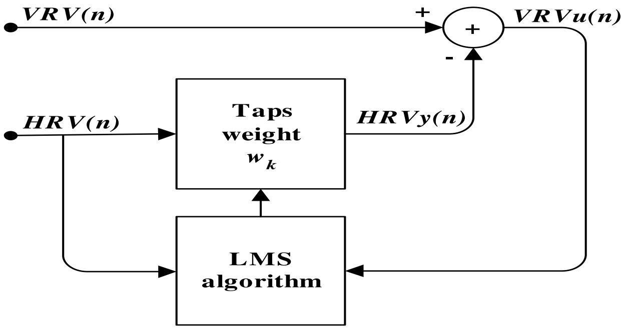

Figure 1. Adaptive HRV canceling technique.

The two inputs of the system are derived from a pair of sensors. The primary sensor receives the *VRVs(n)* signal corrupted by additive noise $HRVn_O(n)$, as shown by:

$$VRV(n) = VRVs(n) + HRVn_0(n) \qquad (1)$$

The signal *VRVs(n)* and the noise $VFCn_0(n)$ are uncorrelated with each other, that is:

$$E[VRVs(n)HRVn_0(n-k)] = 0 \text{, for all } k \qquad (2)$$

The reference sensor receives a noise *HRV(n)* that is uncorrelated with the signal *VRVs(n)* but correlated with the noise $VFCn_0(n)$ in the primary sensor in an unknown way, that is:

$$E[VRVs(n)HRV(n-k)] = 0 \qquad (3)$$

and

$$E[HRVn_0(n)HRV(n-k)] = p(k) \qquad (4)$$

Where, as before, the signal are real valued and *p(k)* is an unknown cross-correlation for lag *k*.

The reference signal *HRV(n)* is processed by an adaptive filter to produce the output signal:

$$HRVy(n) = \sum_{k=0}^{M-1} w_k(n)HRV(n-k) \qquad (5)$$

Where the $w_k(n)$ are the adjustable tap weights of the adaptive filter and *M* the filter order. The filter output *HRVy(n)* is subtracted from the primary signal *VRV(n)*, serving as the "desired" response for the adaptive filter. The error signal is defined by:

$$VRVu(n) = VRV(n) - HRVy(n) \qquad (6)$$

This signal is used to adjust the tap weight of the adaptive filter.

The adaptive filter attempts to minimize the mean-square value of the error signal *VRVu(n).* To adjust the weight, the least mean squares (LMS) algorithm was used. The LMS algorithm is a stochastic gradient algorithm in that it iterates each tap weight of a transversal filter in the direction of the gradient of the squared magnitude of an error signal with respect to the tap weight.

The tap weight adaptation is given by:

$$w(n+1) = w(n) + \mu HRV(n)VRVu^*(n) \qquad (7)$$

Where, μ is the step-size parameter, given by:

$$0 < \mu < \frac{2}{ME\left\{|HRV(n)|^2\right\}} \qquad (8)$$

2.2. Study protocol

In this work, forty-seven subjects were studied as indicated bellow:

Protocol I: An ECG database of twenty-three young healthy man: 12 swimmers and 11 sedentary.

Protocol II: An ECG database of twenty-four males subjects: 16 cardiac transplant recipients and 8 control subjects.

Each subject performed two tests with duration of five minutes each, as indicated below:

- Test 1: free ventilation in supine position (rest).
- Test 2: free ventilation in standing position (tilt).

In total, ninety-four short-term ECG records were analyzed.

HRV sequences were obtained using the Gritzali QRS detector followed by a manual verification of each detected R-wave [8]. VRV series were obtained by applying the algorithm proposed by Vila et al [9]. Artifacts were also manually removed.

The HRV and VRV sequences were linearly interpolated and uniformly sampled at 2 *Hz*. A time-varying autoregressive modeling of the interpolated HRV and VRV sequences was performed by using a 25 seconds sliding window. Power spectral densities were estimated. The LF and HF bands were defined respectively by [0.04-0.15 Hz] and by [0.15-0.4 Hz] as proposed in [10]. For each test, the following parameters were determined: LF_{HRV}, HF_{HRV}, LF_{VRV} and HF_{VRV}.

2.3. Adaptive HRV cancellation

A LMS adaptive filter was used to obtain the VRV unrelated to HRV. The filter order (*M*) and the step-size parameter (μ) was modified experimentally in agreement with equation (8), until observe convergence of the signal for each sequence analyzed.

In order to illustrate in a simple way, the operation of the filter, it was validated using sine waves of frequencies corresponding to LF and HF components.

for each components LF and HF, two percentage power ratio unrelated to HRV was defined, as the relation between the variance from VRVu to VRV, given by equations (9) and (10):

$$\%LFu = \frac{LF_{VRVu}}{LF_{VRV}} x100\% \qquad (9)$$

$$\%HFu = \frac{HF_{VRVu}}{HF_{VRV}} x100\% \qquad (10)$$

From the adaptive filter, the sequence VRVu were obtained, power spectral densities were estimated and the parameters *%LFu* and *%HFu* were determined.

3. Results

The methods were implemented using MATLAB Signal Processing Blockset. Results are presented as mean ± SD. The comparisons were carried out using Wilcoxon tests. The significant threshold was fixed at $p<0.05$.

A LMS adaptive filter ($M=20$) was implemented. The μ values are shown in table 1.

Table 1. Step-size (μ) values

Subject	Test 1	Test 2
Swimmers	6.5e-6±4.9e-6	1.1e-5±1.2e-5
Sedentary	1.1e-5±9.1e-6	1.4e-5±1.1e-5
Cardiac transplant	1.3e-3±1.6e-3	1.5e-3±1.8e-3
Control	3.9e-5±3.3e-5	4.9e-5±3.5e-5

Figure 2 shows HRV, VRV and VRVu sequences and power spectra for a sedentary subject at supine rest.

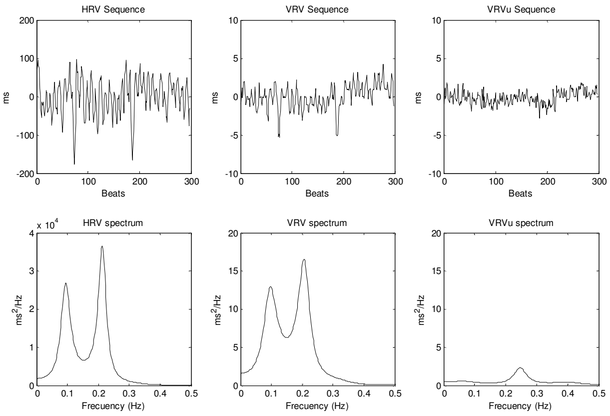

Figure 2. Sequences and power spectra of HRV, VRV and VRVu for a sedentary subject at rest.

Figure 3 shows HRV, VRV and VRVu sequences and power spectra for a cardiac transplant recipient at supine rest. Both VRVu power spectra are negligible compared with VRV power spectra ($p<0.01$) and they are characterized by an important HF peak.

The variance of the spectra (ms^2) at LF and HF bands from VRVu sequences are summarized in table 2 and 3 for protocol I and II, respectively.

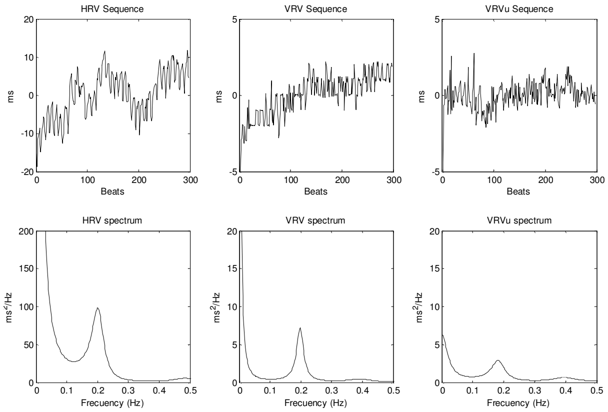

Figure 3. Sequences and power spectra of HRV, VRV and VRVu for a cardiac transplant recipient at rest.

Table 2. Variance (ms^2) for VRVu sequence for protocol I

Subject	Test 1		Test 2	
	LF	HF	LF	HF
Swimmers	3.6±2.1	2.8±3.3	6.3±3.7	3.5±2.6
Sedentary	4.7±2.5	6.1±9.9	8.3±4.8	8.6±6.9

Table 3. Variance (ms^2) for VRVu sequence for protocol II

Subject	Test 1		Test 2	
	LF	HF	LF	HF
Cardiac transplant	1.9±1.1	4.1±3.5	2.7±3.0	3.6±4.9
Control	1.7±1.5	2.1±2.0	3.6±3.5	4.5±5.1

For protocol I, in average the sedentary subjects show a HF component greater than the LF component, however for swimmers, the effect is the opposite, the LF component is significantly greater than the HF component ($p<0.05$), for both test. For protocol II, in average for control subjects the HF component is greater than the LF component and for cardiac transplant recipient the HF component is significantly greater than the LF component ($p<0.05$)

The power ratio between *%LFu* and *%HFu* are shown in figure 4 for both protocols. For protocol I at rest, the *%LFu* and *%HFu* show the same percentage, nearly to 35%. However, at tilt up position, the *%HFu* is significantly greater than the *%LFu* ($p<0.05$). For protocol II, the *%HFu* is greater than the *%LFu* (p<0.05) at rest and at tilt. The *%HFu* for sedentary subjects increase significantly from rest to tilt ($p<0.02$). However, for swimmers, cardiac transplant recipient and control subjetcs no significant differences were found between rest and tilt for *%HFu*.

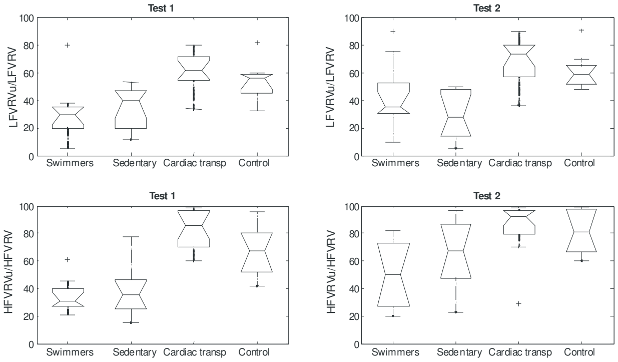

Figure 4. *%LFu* and *%HFu* in test 1 and test 2.

4. Discussion and conclusions

In this study, a novel technique was proposed, based on adaptive filtering, to study the relationship between ventricular repolarization variability and heart rate variability. This strategy made possible to cancel the heart rate modulation on ventricular repolarization phenomena.

In VRVu power spectra, for sedentary, cardiac transplant recipient and control subjects, the HF component is more important than the LF component. In these populations, the heart rate modulation on the ventricular repolarization phenomena showed to be more important in the low frequency band.

The HF rhythm in ventricular repolarization variability are not modulated by the ANS. Respiration could be the most important physiological source of HF component, since in cardiac transplant recipient, ANS efferent nerves are severed during the surgical procedure.

For swimmers, the LF component observed in VRVu power spectra is more important than the HF component, and, in average, the variance of VRVu for sedentary subject is greater than the observed for swimmers. In this case, for swimmers, the VRV is strongly modulated by the heart rate.

Our findings are in agreement with previous studies. Wong et al [11], reported that HF in VRV series has not physiological significance correlated to ANS regulation, Porta et al [4] concluded that VRVu has its most important frequency components in the VLF band.

HF oscillations in VRV series is not yet a fully understood issue. In a clinical context, explains the vagal and sympathetic components are important issues. Further works are necessary to better understand ventricular repolarization phenomena, setp-size (μ) adaptive would be interesting to track the signal and analyze short-term recording at controlled ventilation could be explain the effect of respiration in ventricular repolarización variability. Long-term ECG records is proposed as a step forward to explore VLF component.

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