High Blood Press. 1994; 3: 23-29

Methods in heart rate variability analysis: May the ventricular or the pulse rhythm be used as a substitute for the atrial rhythm?

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ABSTRACT. The cardiac atrial, ventricular, and pulse rhythms differ on account of beat-to-beat variations in atrioventricular conduction time, pre-ejection period, and pulse wave velocity. We investigated the practical consequences of this physiological discordance for heart rate variability analysis. Measurements were done in seven young volunteers in the supine and standing posture. The three rhythms were derived from a simultaneously recorded esophageal electrogram, an electrocardiogram, and a continuous peripheral blood pressure. The atrial rhythm was the gold standard. Ventricular low-frequency (0.07-0.14 Hz) and high-frequency (0.14-0.40 Hz) spectral powers were within 10% from the standard in 94.6% and 87.5% of the spectra. Pulse low- and high-frequency powers were within 10% from the standard in 91.1% and 26.8% of the spectra. The mean of all computed pulse high-frequency powers was 136% of the standard. We conclude that the ventricular rhythm is an acceptable substitute for the atrial rhythm. The pulse rhythm should be used with caution.

Key words: Heart rate variability, atrial rhythm, ventricular rhythm, pulse rhythm, non-invasive continuous blood pressure.

INTRODUCTION

The autonomic nervous system is one of the major modulators of cardiac performance. In the presence of coronary artery disease or myocardial abnormalities, autonomic influences may induce function disorders, such as ischemia and arrhythmias (1). Likewise, cardiac autonomic control influences the risk of sudden cardiac death (2). Therefore, assessment of efferent cardiac autonomic activity, preferably by methods which are noninvasive and suitable for widespread use, is of great clinical relevance. Heart rate variability analysis may serve this purpose.

Under conditions of rest or light exercise, the heart is commonly under concurring sympathetic and vagal influence. In such states, the heart rate results from combined sympathetic cardioacceleration and vagal cardiodeceleration. Multiple combinations of sympathetic and vagal tones may lead to a similar heart rate (3). Therefore, given a specific heart rate value, the underlying sympathetic and vagal tones remain unknown. Spectral analysis of heart rate variability allows for some degree of separation of the sympathetic and vagal influences, because low-frequency (0.07-0.14 Hz) fluctuations in the cardiac rhythm are both sympathetically and vagally mediated,

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Received June 2, 1993; accepted in revised form October 28, 1993.

whereas high-frequency (>0.14 Hz) fluctuations are solely vagally mediated (4-8).
Ideally, the rhythm generated by the sinoatrial node should be used for heart rate variability analysis, since the sinus node is directly modulated by the efferent cardiac autonomic outflow. However, analysis of the sinus rhythm is too cumbersome to be of much practical use since recording of the sinus rhythm requires an atrial electrogram, or an automated, computerized method for the reliable detection of P-waves in the long-term ECG. Hence, in clinical practice, the cardiac rhythm is derived from either the QRS complexes in the ECG or, in cases in which continuous blood pressure measurement is performed, from the blood pressure pulsations.
The time lags between the firings of the sinoatrial node, the onsets of the ventricular depolarizations, and the onsets of the peripheral pressure pulsations, fluctuate with rate and with autonomic tone (9-13). Hence, there is a physiological discordance between the atrial, the ventricular, and the pulse rhythm. As a consequence, heart rate variability indexes are dependent on the source from which they are derived. We investigated the practical impact of this discordance on 2 elementary indexes in heart rate variability analysis: the low-frequency ("10-second rhythm") and high-frequency ("respiratory sinus arrhythmia") spectral powers. Measurements were done in young normal subjects in the supine and standing posture. As low- and high-frequency power values differ between individuals and between supine rest and orthostatic stress (6), this protocol ensures a variety of low- and high-frequency power values.

METHODS

Seven apparently healthy volunteers (3 male, 4 female, ages 20-39 years, mean 25 years) were studied during a one-hour session. The subjects were freely breathing; they were not allowed to speak, and were watching an indifferent video tape. A session consisted of a 20-minute supine period, followed by 10-minute episodes of alternate standing and lying. Throughout the sessions an esophageal electrogram [pill-electrode as described by Arzbaecher (14)], a 3-lead electrocardiogram, and a continuous non-invasive finger blood pressure [Finapres device (15)] were recorded on a digital instrumentation recorder (Earth Data EDR-8000).
The esophageal electrogram, one ECG lead, and the blood pressure were subsequently played back at half speed, and recorded on a Holter tape which was analyzed on a Marquette Series 8000 Holter Analyzer. By virtue of the reduced playback speed, the Holter analyzer accepted the relatively high-frequency esophageal complexes, the QRS complexes, as well as the blood pressure pulsations as "beats". This procedure entails an effective time quantization of 1/256 s or ≈ 4 ms (normally, for real time signals, the sample distance of the Holter analyzer is 1/128 s or ≈ 8 ms). The atrial rhythm, the ventricular rhythm, and the peripheral pulse rhythm were then separately analyzed by single-channel analyses of the same Holter tape.
The three time series constituting the atrial, ventricular and peripheral pulse rhythms were established with the Marquette Holter Analyzer as follows. After initial automated analysis, an extensive interactive review and editing session started, in which three of the authors participated. This procedure aimed at the determination of reliable onsets of the esophageal complexes, the QRS complexes and the peripheral pressure pulsations, respectively. Onsets were established by moving the vertical cursor from the automatically found fiducial point to the sample best corresponding with the first visible deflection of the signal from the baseline (esophageal electrogram, ECG), or from the downsloping blood pressure during diastole. In case of disagreement, a majority decision was taken. In this way, the edit-and-review-team interactively handled all beats of all studied subjects, for the esophageal electrogram, the ECG, as well as for the blood pressure signal.
Heart rate variability was determined in eight stationary 4-minute episodes adjacent to the postural transitions (Fig. 1). After each transi-

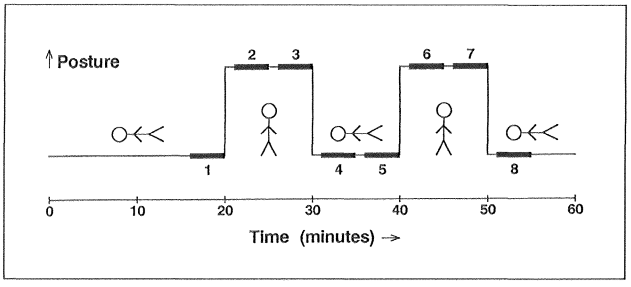

Fig. 1 - Study protocol. A one-hour session consisted of a 20-minute supine period, followed by 10-minute periods during which a subject was either lying or standing. The 8 bars represent stationary 4-minute episodes near the postural transitions that were used for analysis.

tion a one-minute period, needed for heart rate adaptation (16), was discarded from the analysis. Episode stationarity was ensured by division of the episode in 12 segments of equal length, and by applying reverse arrangements tests (17) at the 10% level (18) to the segment means and variances.
Low-frequency (0.07-0.14 Hz) and high-frequency (0.14-0.40 Hz) powers were computed from the Fast Fourier transforms of the inter-beat intervals, after normalization of the inter-beat intervals by dividing them by the mean interval (19), linear trend removal (20), 10% cosine tapering (18), and zero padding. Resulting data consisted of 56 heart rate values (7 subjects, each 8 episodes), and of 56 low-frequency and 56 high-frequency spectral powers for each rhythm variant, of which half were supine, and half were standing values.
The atrial rhythm, being closest to the rhythm of the sinoatrial node, served as gold standard. Comparison of spectral power values X, computed from either the ventricular or pulse rhythm, and the reference spectral powers A, computed from the atrial rhythm, was done by calculation of a relative deviation: (X-A)/A.
Due to decision uncertainties during the above described interactive determination of the onsets of the esophageal complexes, the QRS complexes and the peripheral pressure pulsations, respectively, some inevitable artifactual variability was introduced into the atrial, the ventricular, and the pulse rhythms. Due to the intense visual contact with the processed signals, the edit-and-review team could make a fair assessment of the uncertainty in the decisions taken. In the esophageal electrogram, there was only occasionally disagreement about the choice between two consecutive samples. Hence, the

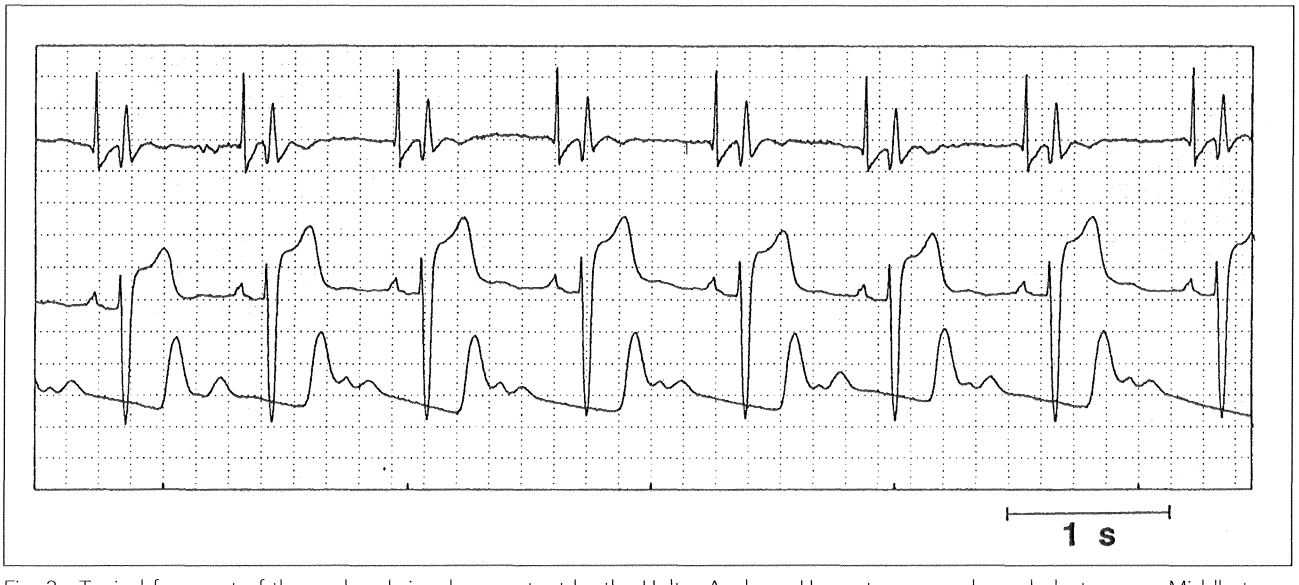

Fig. 2 - Typical fragment of the analyzed signals, as output by the Holter Analyzer. Upper trace: esophageal electrogram. Middle trace: electrocardiogram (Z-lead). Lower trace: finger blood pressure. Arbitrarily units along the y-axes.

average uncertainty in the onsets of the esophageal complexes was less than one sample distance (one sample distance = 1/256 s ≈ 4 ms), corresponding to a (rounded) measurement noise variance of 5 ms^2. Similarly reasoning, the estimated measurement noise variances in the onsets of the QRS complexes in the ECG and in the onsets of the blood pressure pulsations were 10 and 20 ms^2, respectively. The variance of the atrial-ventricular and atrial-pulse lags due to measurement noise would then be 15 and 25 ms^2, respectively. As this study addresses the physiological variability of these time lags, the total, summed physiological and artifactual variances should be considerably more than 15 and 25 ms^2. In all episodes, deviates were computed by subtracting the mean episode atrial-ventricular or atrial-pulse lag from each atrial-ventricular or atrial-pulse lag in the episode. Then, total atrial-ventricular and total atrial-pulse variances were computed from the pooled atrial-ventricular and atrial-pulse deviates.

RESULTS

High-quality signals could be obtained in all subjects. Figure 2 shows a typical recording, depicting the esophageal electrogram, the electrocardiogram and the finger blood pressure.

The mean ± SEM supine and standing heart rate values were 63.9±1.4 and 84.1±2.0 beats-per-minute, respectively. The mean ± SEM supine and standing lags between the atrial and associated ventricular events were 137.8±0.3 and 125.3±0.4 ms. The mean ± SEM supine and standing lags between the atrial and associated pulse events were 377.8±0.4 and 383.1±0.3 ms.

The measured total (artifactual plus physiological) atrial-ventricular and atrial-pulse lag variances were 35.2 and 82.9 ms^2, respectively.

Table 1 presents the statistics of the low- and high-frequency power values as computed from the 3 rhythm variants. There were large ranges for both the low- and the high-frequency powers. There were minor differences between the means, standard deviations, and ranges as computed from the atrial, the ventricular, or the pulse rhythm.

Table 2 lists the statistics of the relative deviations of the rhythm variants. Taking a 10% relative deviation as just acceptable, it appears that no rhythm variant complies completely. All mean relative deviations, except for the high-frequency ventricular power, differ significantly ($p<0.01$) from zero. Although statistically significant, these differences, except for the high-frequency pulse power, are small in absolute terms, and therefore not physiologically relevant in practice.

Figure 3 shows the relative deviations of the high-frequency ventricular and pulse powers as a function of the corresponding atrial power. The largest relative deviations of the high-frequency pulse power occur at the smallest

Table 1 - Mean 0.07–0.14 Hz and 0.14 – 0.40 Hz powers.

Rhythm Variant	0.07- 0.14 Hz Power ($\cdot 10^{-3}$)		0.14 - 0.40 Hz Power ($\cdot 10^{-3}$)	
	Mean±SD	Range	Mean±SD	Range
A	1.7±1.6	0.1–8.5	1.5±2.3	0.1–11.2
V	1.6±1.5	0.1–8.3	1.4±2.2	0.1–10.1
P	1.7±1.5	0.1–8.4	1.6±2.2	0.2–10.1

Mean±SD 0.07-0.14 Hz and 0.14-0.40 Hz power values for all rhythm variants. A = Atrial rhythm; V = Ventricular rhythm; P = Pulse rhythm.

Table 2 - Relative deviations of spectral powers.

Spectral Band	Rhythm Variant	Relative deviation Mean±SD	Range	% between -0.10 and +0.10
Low	V	-0.02±0.04*	-0.15 – 0.02	94.6
	P	0.03±0.05*	-0.12 – 0.13	91.1
High	V	0.01±0.07	-0.19 – 0.25	87.5
	P	0.36±0.37*	-0.10 – 1.85	26.8

Statistics of the relative deviations of the low-frequency (0.07-0.14 Hz) and high-frequency (0.14-0.40 Hz) powers computed from the ventricular and the pulse rhythm. Asterisks mark means differing significantly ($p<0.01$) from zero. V = Ventricular rhythm; P = Pulse rhythm.

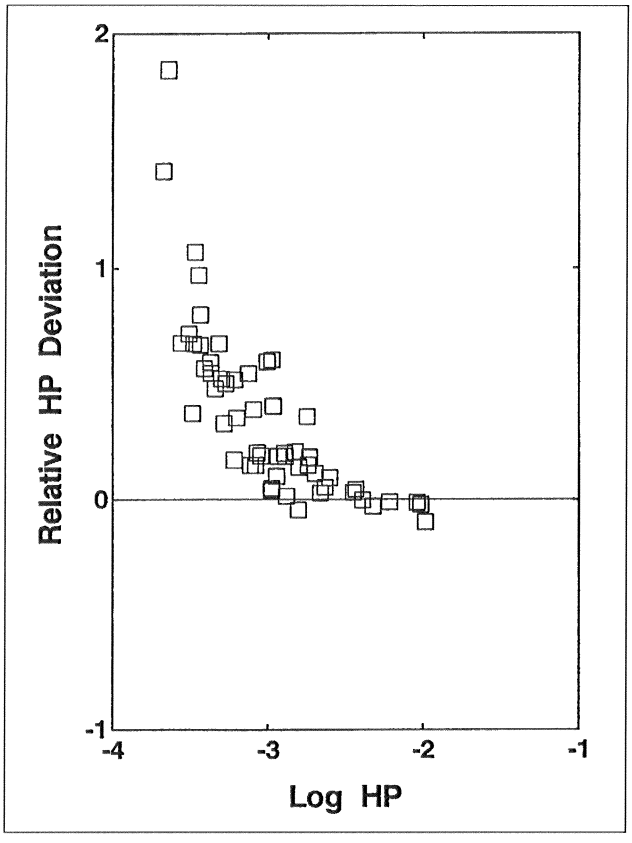

Fig. 3 - Relative deviations of the high-frequency powers of the pulse rhythm (ordinate), related to the logarithm of the corresponding reference atrial powers (abscissa). The logarithmic scale was chosen because of the large power range. HP = 0.14–0.40 Hz power.

atrial power values. Deviations become prominent at power values <0.001.

DISCUSSION

In general, differential measurements have to be done with high precision, because of the importance of the signal to noise ratio. In this study, high-quality signals were recorded under laboratory conditions. With thorough editing and reviewing procedures, we tried to determine as reliably as possible the onsets of the cardiac events. The estimated resulting measurement noise explained 15 of the 35.2 ms^2, and 25 of the 82.9 ms^2 total variance in the atrial-ventricular and atrial-pulse lags, respectively. Hence, mainly physiological variability has been measured. In routine Holter electrocardiography, where no interactive beat-to-beat timing adjustment is done, there will possibly be a higher measurement noise. A possible explanation for the characteristic behavior of the data in Figure 3 is a constant small amount of extra measurement noise in the pulse rhythm (random measurement noise is likely to have similar effects as low frequent sampling (21), and will consequently introduce extra high-frequency power).

Our study demonstrates that physiological plus, to a lesser extent, artifactual corruption of the atrial rhythm does not lead to relevant changes in the means of the spectral powers, except for an average increase of 36% in the high-frequency pulse rhythm power. However, single spectra can differ more (deviations up to 185% were measured).

The ventricular rhythm is an acceptable substitute for the atrial rhythm (Table 2). Naturally, atrioventricular conductance disturbances resulting in variable PQ-intervals or isolated P-waves (Mobitz type I and II atrioventricular blocks) hamper any reasonable interpretation of the ventricular rhythm. Furthermore, it is self-evident that supraventricular and ventricular ectopics do not reflect efferent cardiac autonomic outflow to the sinus node. In episodes with (supra)ventricular ectopy the normal QRS-rhythm must therefore first be reconstructed (e.g. by interpolation methods (22, 23) to facilitate estimation of spectral heart rate variability.

In hypertensive patients the change in pulse wave velocity per unit change of blood pressure will be greater than in persons with normal blood pressure values (9). For this kind of patients the high-frequency pulse rhythm powers may deviate even more from the atrial standard. Heart rate variability is reduced in disease and old age (6, 24, 25). Due to the constant amount of measurement noise, the relative deviation of the pulse rhythm may then increase to unacceptable levels (Fig. 3). Therefore, the use of the pulse rhythm is not recommended in such cases. Naturally, with a more proximal recorded blood pressure signal, the physiological discordance of a thus obtained rhythm will be less.

In conclusion, our study indicates that the ventricular rhythm is an acceptable substitute for the atrial rhythm and may be used to study efferent cardiac autonomic activity. Our study also indicates that the pulse rhythm is of use in heart rate variability analysis, but it requires cautious interpretation. It is not recommended to use the pulse rhythm in diseased and older patients, and in patients with hypertension.

ACKNOWLEDGEMENTS

Support by Netherlands Heart Foundation grant 89.110 is gratefully acknowledged. We thank the Foundation "Fysiologic", Zeist, and Marquette Electronics Inc. for supporting our research in part.

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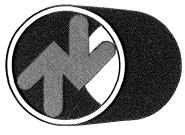

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