



Validity of the polar S810 heart rate monitor to measure R-R intervals at rest.

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Title page

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Title: Validity of the Polar S810 heart rate monitor to measure RR intervals at rest

Running title: Validity of the Polar S810 heart rate monitor

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ABSTRACT

Purpose: To compare R-R intervals and the subsequent analysis of heart rate variability (HRV) obtained from the Polar S810 heart rate monitor (HRM) (Polar Electro Oy, Kempele, Finland) with an ECG (Physiotrace, Estaris, Lille, France) during an orthostatic test. **Methods:** Eighteen healthy men (Age: 27.1 ± 1.9 years ; Height: 1.82 ± 0.06 m ; Mass 77.1 ± 7.7 kg) performed a passive orthostatic test during which R-R intervals were simultaneously recorded with the HRM and the ECG recorder. The two signals were synchronized and corrected before a time domain analysis, a fast fourier transform (FFT) and a poincaré plot analysis. Bias and limits of agreement (LoA), effect size (ES) and correlation coefficients were calculated. **Results:** R-R intervals were significantly different in the supine and upright position between corrected signal of the two devices ($p < 0.05$, ES = 0.000 and 0.006 respectively). The bias \pm LoA were 0.9 ± 12 ms. HRV parameters derived from both signals in both positions were not different ($p > 0.05$) and well correlated ($r > 0.97$, $p < 0.05$), except RMSSD and SD1 in upright position ($p < 0.05$, ES = 0.052 and 0.057; $r = 0.99$ and 0.98 respectively). **Conclusion:** Narrow LoA, good correlations and small effect sizes support the validity of the Polar S810 HRM to measure RR intervals and make the subsequent HRV analysis in supine position. Caution must be taken in upright position for the parameters sensitive to the short term variability (i.e. RMSSD and SD1).

Keywords: RR intervals, time domain analysis, frequency domain analysis, poincaré graph analysis

Paragraph number 1

INTRODUCTION

The measurement of the heart rate variability (HRV) has become a common tool in the clinical domain, since it appears sensitive to both physiological (12, 20) and psychological (5, 6) disorders. In sports medicine, it is generally used to assess both adaptation (11, 20, 21) or maladaptation to endurance training (7, 14). As such, it could become a common tool in the follow-up of elite athletes.

Paragraph number 2

The measurement of HRV usually requires a high quality Electrocardiogram (ECG) with a sampling rate upper than 250 Hz and an accurate algorithm to detect the QRS complex (18). Over recent years a number of ambulatory ECG recorders or Holter monitors that satisfy these requirements have been developed, permitting the use out of laboratory. However, the cost and the complexity of this equipment made the HRV analysis difficult in the physical training field conditions, particularly in the monthly follow-up of elite or subelite squads.

Paragraph number 3

For coaches and physicians, the development of wireless heart rate monitoring (HRM) with elastic electrode belt allowing the detection of R-R intervals with a resolution of 1 ms (9) represents a very interesting alternative to classical fix or ambulatory ECG for coaches and physicians. It remains to determine the accuracy of this device before using it in regular basis. Kingsley et al. (8) reported a good accuracy of the Polar S810 HRM (Polar Electro Oy, Kempele, Finland) when compared with an ambulatory ECG during exercise at low intensity. Comparison in supine position are lacking when it is recommended condition to detect overreaching (7). Moreover, the impact of differences between the signals on heart rate variability parameters, is not known, both in the time and frequency domain.

Paragraph number 4

Thus, the purpose of this study was twofold: 1) to compare raw data obtained in a supine and upright position from the Polar S810 HRM and an ECG 2) to compare the HRV parameters derived from both signals in the time and frequency domain.

Paragraph number 5

METHODS

Subjects

Eighteen active men (Age: 27.1 ± 1.9 years; Height: 1.82 ± 0.06 m ; Mass 77.1 ± 7.7 kg)with no smoking history and no known cardiovascular disease gave their written informed consent to participate in the study. All of the subjects were submitted to an inclusion protocol before the start of the study. This consisted in an information session about the nature, the potential risks involved and the benefits of the study, followed by a complete medical screening when the subjects were interested in participating to the study. The protocol has been reviewed and approved by the Consultative Committee for the Protection of Human Subjects in Biomedical Research of the Nord – Pas de Calais (France) before the start of the study.

Paragraph number 6

Experimental design

Two weeks after the inclusion visit, the subjects reported to the laboratory within 2 hours of waking (between 6:00 and 10:00 am). The subjects were asked to abstain from caffeine-containing foods and beverages on the day prior to the test. Prior to the 17 minutes recording, the skin of the subject was cleaned and prepared for the attachment of surface electrodes (Blue Sensor, Medicotest Ltd, Ølstykke, Denmark). The electrodes of the ECG were placed in such a way not to prevent the installation of the HRM elastic electrode belt (T61, Polar Electro Oy, Kempele, Finland) with conductive gel being applied to the chest of each subject as described by the manufacturer.

Paragraph number 7

The subject rested comfortably for at least 10 minutes in a supine position and 7 min in a upright position in a quiet and semi dark laboratory room, maintained at a temperature of 19-21°C. The subjects matched their breathing frequency to an auditory metronome set at 0.20 Hz (12 breaths.min⁻¹). No attempt was made to control the tidal volume.

Paragraph number 8

Data acquisition

R-R intervals were recorded simultaneously with a Polar S810 HRM with a 1 ms precision and an 2 leads ECG recorder (Physiotrace Estaris, Lille, France) at a sampling frequency of 1000 Hz. R-wave peaks were detected automatically in the ECG series using a detection algorithm supplied by the manufacturer

(Estaris, Lille, France). Following the recordings and storage of the raw ECG data, each “R” wave peak was verified and validated using a vertical mark on the ECG that indicated the detection of an “R” wave. If the detection was incorrect, “R” wave peak was determined manually, by replacing the vertical marks on the correct “R” wave peak. Subsequently, R-R intervals were exported under the ASCII format. The HRM signal was transferred to the Polar precision performance Softwear (release 3.00; Polar Electro Oy, Kempele, Finland) and R-R intervals were exported under ASCII format.

Paragraph number 9

Data analysis

R-R interval comparison

Time coordinates and R-R intervals from both systems were synchronized using temporal “event” markers recorded prior to the test. Raw R-R intervals from both acquisition systems were edited and compared to discriminate error due to the HRM acquisition or to a non sinus beat. Non sinus beats, which were presents in both signals, were replaced by interpolated data derived from adjacent normal RR intervals. An error due to the HRM acquisition was considered when the difference between ECG and HRM interval exceeded 20 ms (10). Then the HRM interval was labelled anomalous and later assigned to one of five identified error categories (10). A type 1 error was defined as a single point of discrepancy, either positive or negative between the ECG and HRM R-R interval. A type 2 error was considered to be when a long interval was immediately followed by a short interval and the magnitude of the difference between the two ECG and HRM R-R interval were very similar. An inverse error was defined as a type 3 error. A type 4 error was defined when the HRM R-R interval was equivalent to two or three ECG R-R intervals. Finally there was a type 5 error which occurred when the HRM detected two or more short R-R intervals whereas the ECG detected one interval. Generally the addition of these short intervals corresponded to the ECG interval.

Paragraph number 10

To conserve time synchrony between the two data series, and to allow the comparison between the ECG and the uncorrected HRM data, an ECG R-R interval of 0 ms was inserted when a type 5 error was present. On the contrary, a Polar R-R interval of 0 ms was inserted when a type 4 error was present. The correction algorithm for HRM data was the following. When a type 1 error was present, the RR interval

was replaced by interpolated value from the two adjacent RR intervals. When a type 2 or type 3 errors were present, the two uncorrected R-R intervals were averaged. When a type 4 error was present, the R-R interval was divided by two or three according to the number of “R” waves undetected. Finally, when a type 5 error occurred, R-R intervals were added to obtain the same value than the ECG. Once noisy complexes were replaced, the signal was considered to be normal, and to provide Normal-to-Normal (NN) intervals.

Paragraph number 11

Time domain analysis

A corresponding segment of 256 s was selected within the last 300 s last of the supine and upright recordings. The mean NN interval, the standard deviation of all NN intervals (SDNN), the root mean square of differences of successive NN intervals (RMSSD) and the proportion of differences between adjacent NN intervals of more than 50 ms (pNN50) were computed.

Paragraph number 12

Frequency domain analysis

The same segments of 256 s were resampled at 2 Hz and detrended for subsequent analysis. As recommended by the Task Force (18), spectral analysis was performed with a Fast Fourier Transform (FFT) to quantify the power spectral density of the very low frequency (VLF; 0.00 to 0.04 Hz) the low frequency (LF; 0.04 to 0.15 Hz) and the high frequency (HF; 0.15 to 0.40 Hz) bands. Additional calculations included LF+HF, LF and HF expressed in normalized unit (i.e. in a percentage of LF+HF) and the ratio LF/HF.

Paragraph number 13

Quantitative beat-to-beat analysis

The Poincaré plot is a scattergram in which each NN interval is plotted as a function of the previous one. The Poincaré Plot provides both a qualitative and a quantitative analysis of HRV. The shape of the plot can be used to classify the signal into one of various classes (13, 22), but also to fit an ellipse, which enables us to quantify the parameters SD1 and SD2. SD1 represents the dispersion of the points perpendicular to the line of identity, and it is thought to be an index of the instantaneous beat-to-beat

variability of the data. SD2 represents the dispersion of the points along the line of identity, and represents the slow variability of heart rate (3, 22)

Paragraph number 14

Statistical analysis

Standard statistical methods were used for the calculation of means and standard deviations. Normal Gaussian distribution and homogeneity of variance were verified by the Shapiro-Wilk and the Levenne Tests, respectively. Homoscedasticity was checked with a modified Levenne Test. A paired *t*-test, or when appropriate a Wilcoxon matched pairs test, were used to detect the presence of a systematic difference in R-R interval or HRV indices calculated from both systems. Effect Size (ES), which represents the ratio of the mean difference over the pooled variance (19), was used to estimate the magnitude of the difference. As proposed by Cohen (4), the difference was considered small when $ES \leq 0.2$, moderate when $ES \leq 0.5$, and large when $ES > 0.8$. Relative reliability, defined as the degree to which individuals maintain their position in a sample with repeated measurements (1), was assessed by the Pearson's product moment correlation coefficient, or when appropriate by the Spearman rank order correlation. Finally, Bland and Altman plots of all measures from both systems were constructed and the 95% limits of agreement (LoA) were computed. As recommended by Bland and Altman (2), data were log-transformed prior to the calculation of the LoA when heteroscedasticity was present. Statistical significance was set at $p = 0.05$ level for all analysis. All calculations were made with Statistica (Release 6.0, Statsoft, Tulsa, USA).

Paragraph number 15

RESULTS

The total number of R-R intervals detected was 11353 and 9878 in supine and upright position respectively for the ECG and 11335 and 9857 for the HRM. The amount and the type of error are described in table 1. The *t*-test revealed that uncorrected and corrected R-R intervals were different from ECG R-R intervals in supine position ($p < 0.05$, effect size = 0.025 and 0.000 respectively). Figure 1 and 2 represent Bland and Altman plots for combined ECG and uncorrected R-R intervals and the ECG and corrected R-R intervals. The correlation were 0.88 and 0.99 for the uncorrected and the corrected HRM R-R intervals with the ECG in supine position respectively ($p < 0.001$). In upright position, coefficients of

correlation with ECG R-R intervals were 0.88 and 0.99 for uncorrected and corrected HRM data respectively ($p < 0.001$). There were no significant differences for time domain, FFT and Poincaré plot parameters obtained from the two signals except for RMSSD, SD1 in upright position ($p < 0.05$). The correlation of HRM with ECG parameters as well as the coefficient of variation, the bias, the 95% interval of confidence for the bias and the magnitude of the difference are presented in tables 2 and 3.

Paragraph number 16

DISCUSSION

The purpose of this study was to compare raw data and the HRV parameters derived from a Polar S810 HRM (Polar Electro Oy, Kempele, Finland) and a 2 leads ECG recorder (Physiotrace Estaris, Lille, France). The present results provide consistent measurement of heart rate variability from R-R intervals derived from the HRM in healthy subject during an orthostatic test.

Paragraph number 17

The error rate in detection of “R”waves for Polar and ECG was 0.40%. This is in accordance with previous studies which reported a rate of 0.32 to 2.8% (8, 16). The most common error occurring in the uncorrected HRM signal was a type 4 error. It represented 75 and 56% of the total errors in supine and upright position respectively. The origin of this error is not known but a lack of contact between the skin and the elastic electrode belt could cause a decrease in R-wave amplitude and the inability to detect it. The type 5 errors was the second commonly error. It seems that these errors result from multiple triggering during a single cardiac contraction. These errors may have been caused by the HRM registering a T-wave and/or a P-wave as a R-wave (10).

Paragraph number 18

In this study we observed a significant difference between the uncorrected HRM and the ECG R-R intervals in supine and upright position. The decrease of limits of agreement and the increase of the correlation coefficient after the correction demonstrates that correction protocols were successful when applied to the current data. Nevertheless the difference observed between the corrected HRM and ECG R-R intervals remained significant. The large number of observation in supine and upright position ($n = 11353$ and 9878 , respectively) may produce this significant statistical difference since the magnitude of the difference (i.e. the effect size) was very small ($ES = 0.000$ and 0.006 respectively). As already

reported by Kingsley et al. (8), the bias was less than 1 ms in the current study. However, our limits of agreement were wider than those reported by Kingsley et al. (8) in resting condition (LoA: -5.2 to 5.89 ms, $p < 0.05$). This difference may be explained by the method of correction, since Kingsley et al. (9) excluded artefacts and non sinus beats from the signal, while they were corrected in our study. Nevertheless, the very small magnitude of the difference together with a very good correlation between the HRM and ECG data ($r > 0.99$, $p < 0.001$) suggests that the HRM is a valid tool to measure RR intervals.

Paragraph number 19

As reported by Radespiel-Troger et al.(15), we found a good correlation between time domain parameters estimated from HRM and ECG signals ($r > 0.97$, $p < 0.05$). There was no significant differences between parameters estimates, excepted for RMSSD in upright position, A possible explanation for this difference is that RMSSD reflects the short term variability of the signal (18). It is therefore more sensitive to light variations in the R-R interval duration between the HRM and the ECG. Nevertheless, the correlation coefficient for this parameter between the two acquisition systems is good ($r = 0.99$, $p < 0.05$) and the magnitude of the difference is small (effect size < 0.052). In a general manner, we can note that this measurement error If we consider that 3 weeks of intensive training induce a significant increase of RMSSD (from 22.10 ± 22.33 ms to 32.76 ± 49.95 ms), which corresponds to an effect size of 0.38 (14), we note that the measurement error by the HRM is reasonably good.

Paragraph number 20

The calculated SD1 and SD2 for the HRM and the ECG signals were similar in supine position. In standing position, SD1 estimated from the HRM signal was significantly lower than SD1 obtained from the ECG signal (table 3). SD1 represents the standard deviation of instantaneous beat-to-beat variability (22). Then, as RMSSD, SD1 is more sensitive to the light variations in the successive R-R intervals duration between the two acquisition systems. Nevertheless, a good correlation ($r > 0.99$, $p < 0.05$), together with narrow LoA (table 3) support the validity of the HRM to realize a Poincaré plot analysis.

Paragraph number 21

In the frequency domain, the VLF, LF and HF components were almost identical (tables 2 and 3). The observed differences for these parameters were not statistically significant ($p > 0.05$), whatever the

position. The LoAs were in accordance with Kinglsey et al. (8), who reported values lower than 8 ms² for LF and HF versus 10 ms² in this study. Indeed, the magnitude of difference lower than 0.2 for all frequency parameters confirmed this tight difference (4). Again, the measurement error is largely acceptable when compared with the effect size of 0.80 reported in a meta-analysis by Sandercock et al. (17) for HF after training in sedentary men.

Paragraph number 22

In conclusion, narrow LoA, good correlations and small effects size support the use of the Polar S810 HRM (Polar Electro Oy, Kempele, Finland) to measure HRV in supine position. Caution must be taken in upright position for the parameters sensitive to the short term variability (i.e. RMSSD and SD1). Nevertheless the light differences obtained with the Polar S810 are negligible when compared with training or overtraining effects on HRV parameters. Moreover, the use of the same device during HRV studies may allow avoiding this difference.

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