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# A Novel Algorithm for the Heart Rate Variability Analysis of Short-Term Recordings: Polar Representation of Respiratory Sinus Arrhythmia

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A new method for the analysis of heart rate variability in short-term recordings is presented which consists of an analysis of the respiratory sinus arrhythmia in the time domain by means of a polar representation. Its main advantage is that it is applicable in experiments in which the respiration of the subject is not controlled. The algorithm is applied to data recorded on two astronauts during the Euromir-95 space mission. Statistical hypothesis tests demonstrate that the presence of a mouthpiece induces an increase of the respiratory sinus arrhythmia amplitude. 

## INTRODUCTION

Conventional heart rate variability (HRV) analyses on short-term recordings (2–5 min) of the R–R interval (RRI) time series have been widely used to assess regulation mechanisms of the heart rate by the autonomic nervous system (*1–3*). In the time domain, the standard deviation of normal to normal RRI and the root mean square of the successive RRI difference methods can be used to estimate the HRV (*4*). Whereas the first method only reflects the overall HRV, the latter only takes into account the short-term component of HRV (*4*). Moreover these two methods cannot distinguish between the various physiological regulations of the cardiac rhythm and are not easily interpretable (*4*). For this reason the use of frequency domain methods for the analysis and interpretation for HRV of short-term recordings is recommended by the task force of the European Society of Cardiology and the North American Society of Pacing and Electrophysiology (*4*). Although frequency domain methods are readily interpretable in terms of physiological regulating mechanisms their applicability is restricted to certain conditions (*5*), and limited and cautious interpretation of HRV spectra is required (*6*). For instance, stationary RRI series longer than 2 min are required to allow the Fourier spectral analysis (*4, 7*). It is also well recognized that, among other physiological inputs, the influence of respiration should be taken into account in order to allow accurate HRV power spectra interpretation (*4, 8*).



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In this paper we present a new time domain method for the analysis of HRV which aims at estimating the respiratory sinus arrhythmia (RSA) on very short recordings ($\sim$ 1 min) without the need for a controlled respiration. Our algorithm quantitatively estimates the amplitude and phase of the RSA and provides the possibility of statistical testing of these parameters. The presented method is compared to the peak valley (PV) technique which is a time domain quantification procedure for estimating the RSA on a breath-by-breath basis (*9*). We will show results obtained with data from the Euromir-95 space mission launched in August 1995 and landing February 29, 1996. These data are a good example of a situation our algorithm is particularly designed for, i.e., a study with few subjects in an experimental situation where only a minimal number of parameters are controlled. In particular, we will demonstrate that normal breathing through a mouthpiece amplifies the RSA.

## DATA AND METHODS

### *Data*

Physiological signals were recorded on two astronauts (A1, A2) at different dates before, during, and after the six-month Euromir-95 space mission. An experimental session consisted of the measurement of respiration and electrocardiogram (ECG), among other signals, for about 3 min, during which a subject was asked to breath as normally as possible in a resting position (sitting on ground or floating in space approximately in the same position as on ground). The heart rate was extracted from the ECG during acquisition of the data. The respiration signal was recorded with a respiratory inductive plethysmograph (RIP). This apparatus is made of two wires sewed with a zigzag pattern around the ribcage and the abdomen in a suit tailored for each astronaut. The two signals are proportional to the variations of cross sections. After calibration, the sum of the ribcage and the abdomen signals from the RIP was used as a measure for the lung volume variations (*10*). In addition, the respiration signal was also measured by means of a flow meter that was placed in a mouthpiece. In the following we will analyze in particular the influence of the presence (+MP) or the absence (−MP) of this mouthpiece on the RSA. All signals were sampled at 100 Hz. Table 1 presents a summary of the

TABLE 1

Summary of Experimental Sessions

| | Subject A1 | | | Subject A2 | | |
|---|---|---|---|---|---|---|
| | *N* | +MP (s) | −MP (s) | *N* | +MP (s) | −MP (s) |
| Preflight | 4 | 150 (27) | 255 (73) | 4 | 227 (106) | 194 (61) |
| In-flight | 11 | 229 (66) | 199 (80) | 7 | 257 (42) | 254 (36) |
| Postflight | 5 | 202 (67) | 179 (59) | 4 | 179 (75) | 199 (88) |

*Note.* Number of recordings (*N*) with their mean duration (in seconds) and standard deviation for each subject with (+MP) and without mouthpiece (−MP).

number, length, and conditions for all experimental sessions. Note that respiration was not controlled during the experiments.

*Algorithm*

Figure 1 shows an example of the normalization of the time axis of the tachogram. Let $t^{RR}$ be the location in time of the detected R waves and $t^{insp}$ and $t^{expi}$ the time of onset of the inspiration and expiration, respectively. The later parameters are identified by a threshold detection on the respiration signal and its derivative. The accuracy of this detection is controlled visually. Hence, for each breath cycle $i$ starting at $t_i^{insp}$,

$$t_i^{exp\,i} = t_i^{insp} + \Delta_i^{insp} \qquad [1]$$
$$t_{i+1}^{insp} = t_i^{exp\,i} + \Delta_i^{exp\,i},$$

where $\Delta_i^{insp}$ and $\Delta_i^{exp\,i}$ are respectively the duration of the $i$th inspiration and expiration. R waves occurring in each breath cycle are sorted according to whether they occur during the inspiration or expiration. Let $TRR_{k,i}^{insp}$ ($TRR_{k,i}^{expi}$) be the delay

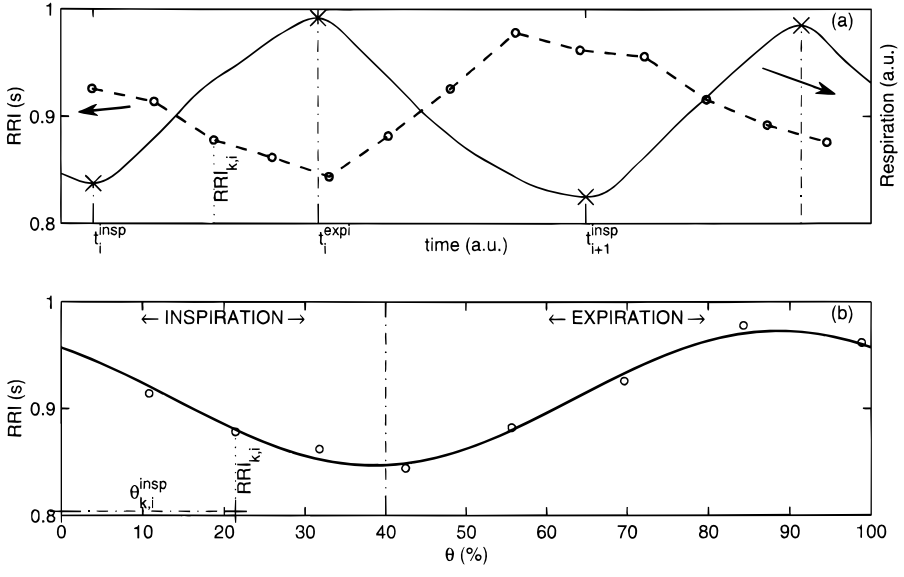


**FIG. 1.** Normalization of the RRI with respect to the breath cycle. (a) RRI (dashed line with circles) and respiration signal (full line) as a function of time, $t$. (b) RRI (circles) as function of their phase in breath cycle $\theta$ where inspiration = 40% of the total breath. The model fit (Eq. [4]) has the following parameter values for this example: $R$ = 0.910 s (±0.005), $\rho_c$ = 0.063 s (±0.007), $\theta_c$ = −11.5% (±1.9); all errors are 95% confidence intervals.

between the onset of the *i*th inspiration (expiration) and the *k*th R wave:

$$\mathrm{TRR}_{k,i}^{\mathrm{insp}} = t_k^{\mathrm{RR}} - t_i^{\mathrm{insp}} \qquad [2]$$
$$\mathrm{TRR}_{k,i}^{\mathrm{exp}\ i} = t_k^{\mathrm{RR}} - t_i^{\mathrm{exp}\ i}$$

We assume that the inspiration occurs during the first 40% of the entire respiration and that, hence, the expiration lasts the remaining 60% of the breath cycle. This assumption is based on the observation that the ratio of the inspiration to the total respiration cycle is given by $\Delta_i^{\mathrm{insp}}/(\Delta_i^{\mathrm{insp}} + \Delta_i^{\mathrm{exp}\ i}) = 0.400 \pm 0.003$. Hence, we define

$$\theta_{k,i}^{\mathrm{insp}} = \frac{\mathrm{TRR}_{k,i}^{\mathrm{insp}}}{\Delta_i^{\mathrm{insp}}} \cdot 0.4 \qquad [3]$$
$$\theta_{k,i}^{\mathrm{exp}\ i} = 0.4 + \frac{\mathrm{TRR}_{k,i}^{\mathrm{exp}\ i}}{\Delta_i^{\mathrm{exp}\ i}} \cdot 0.6,$$

where $\theta_{k,i}^{\mathrm{insp}}$ and $\theta_{k,i}^{\mathrm{exp}\ i}$ represent the relative phases of $\mathrm{RRI}_{k,i}$ with respect to the breath cycle. The correlation of the cardiac cycle with the respiration appears as an oscillation of the RRI during the respiration period (Fig. 1b), which corresponds to the well-known RSA. The Levenberg-Marquard method for minimizing the least mean squares cost function is used to fit the oscillation of the RRI with the model

$$\mathrm{RRI}_{k,i} = R + \rho_c \cdot \cos\left[\frac{2\pi}{100} \cdot (\theta_{k,i} - \theta_c)\right], \qquad [4]$$

where $R$ is the mean RRI, and $\rho_c$ and $\theta_c$ are, respectively, the amplitude and phase of the cosine. The use of a cosine curve in the model corresponds to the analysis of the respiratory peak in the traditional Fourier spectral analysis.

As all respiration periods are pooled together there is no reason to differentiate between $\theta_c = 0\%$ and $\theta_c = 100\%$, which represent the same physiological situation: the functional residual capacity of the subject, i.e., the point where the lungs are at their lowest volume during normal breathing. Hence, we can apply a polar coordinate representation where the radial coordinate $\rho$ is the RRI and the angular coordinate $\theta$ is the phase in breath cycle. In this polar representation of the RSA (Fig. 2) the cosine-fitting curve appears as a circle of radius $R$ and center $c$ with coordinates $(\rho_c, \theta_c)$. Hence, a large RSA appears as a large off-centering of the fitted circle.

*Hypothesis Testing*

Hypothesis testing is used in order to identify statistically significant differences between sets of data. The three parameters of the fit are computed for each set of

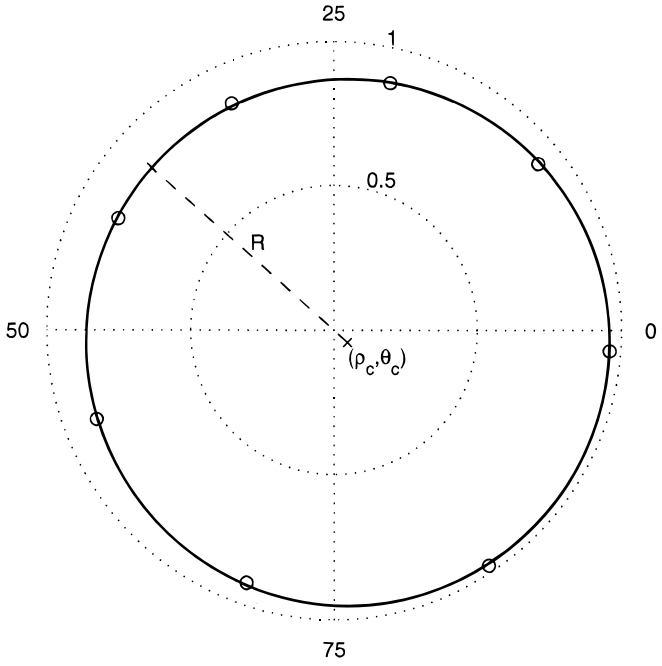


**FIG. 2.** Polar representation of the RSA. The fitting curve of Fig. 1b appears here as a circle of radius $R = 0.910$ s ($\pm 0.005$) and center $c$ of coordinate $\rho_c = 0.063$ s ($\pm 0.007$) and $\theta_c = -11.5\%$ ($\pm 1.9$); all errors are 95% confidence intervals. The radius points in the direction of the shortest RRI. The inspiration runs counterclockwise from 0 to 40% and the expiration from 40 to 100%.

data, and the null hypothesis that the two sets can be fitted with the same parameters is tested.

First a Student's t-test is used to evaluate the differences between the mean heart rate (parameter $R$) in each set. Removing this mean and fitting a reduced model with only $\rho_c$ and $\theta_c$ as parameters, we test the hypothesis that there are no differences between these parameters of the two sets of data. A comparison of nonlinear regression procedure is used to test this null hypothesis and to compute the significance level for the observed differences between the two data sets. The procedure is fully described by Ratkowsky (*11*) and can be summarized as follows:

1. Fitting the model separately to each data set the residual sum of squares S1 and S2 for each set is computed.
2. The model is fitted simultaneously to both data sets, and S3 the residual sum of squares for the pooled data is computed.
3. The variance ratio VR is defined as:

$$\mathrm{VR} = \frac{(\mathrm{S3} - (\mathrm{S1} + \mathrm{S2}))/2}{(\mathrm{S1} + \mathrm{S2})/(n1 + n2 - 4)}, \qquad [5]$$

where $n1$ and $n2$ are respectively the number of data points in both data sets. We

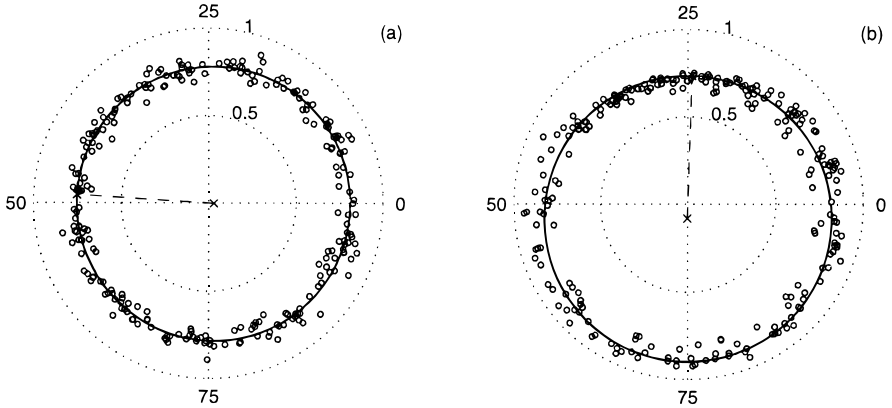


**FIG. 3.** Polar representation of the RSA for subject A1 on ground, 7 days after landing, for the test without mouthpiece (−MP, a) and with (+MP, b). Values for the parameters of the fitting curves are presented in Table 1. The cross (x) indicates the position of the center of the model circle, and the radius points in the direction of the shortest RRI.

can test whether VR is approximately distributed as an F-distribution with 2 (number of parameters in the reduced model) and ($n1 + n2 - 4$) degrees of freedom, which should be the case if the null hypothesis is true.

## RESULTS AND DISCUSSION

To demonstrate our method we apply this algorithm to data where traditional spectral analysis would yield not easily interpretable results. Indeed, neither the frequency nor the tidal breathing volume was controlled during the experiments. Hence, the breathing frequency was quite variable and on many occasions under the 0.15-Hz limit frequency which is of concern in the interpretation of the spectral analysis of HRV (*4*). Figure 3 shows an example of an analysis on data recorded on subject A1 at 7 days after landing. The results are summarized in Table 2. It is seen that the three parameters are significantly different for both experiments.

For comparison the PV technique (*9*) was applied to all the data. This method consists of computing, for each breath cycle, the difference between the shortest

TABLE 2

Hypothesis Testing on +MP and −MP Experiment for Subject A1, 7 Days after Landing

| Parameters (units) | +MP (95% CI) | −MP (95% CI) | $p$ |
|---|---|---|---|
| $R$ (ms) | 817 (3) | 782 (2) | <0.01 |
| $\rho_c$ (ms) | 81.7 (8.6) | 29.1 (6.8) | <0.001 |
| $\theta_c$(%) | −25.4 (1.7) | −0.754 (3.68) | <0.001 |

RRI during the inspiration and the longest RRI during the expiration. The average RSA amplitude (RSAPV) is computed as the mean PV across the total number of breaths. Figure 4 presents $\rho_c$ as a function of RSAPV. The adjusted squared correlation coefficient, $R^2_{\text{adj}} = 0.7352$, indicates a fairly good agreement between the two methods. Note that the polar representation which is presented here takes into account every heartbeat, whereas the PV method only considers two RRI per respiration. Furthermore, the PV method does not provide any phase information on the RSA. In conclusion $\rho_c$ can be considered as a measure for the RSA.

Figures 5 and 6 present the parameters of the model for each recording session respectively for subject A1 and A2, with and without mouthpiece. Sessions where a significant difference is found at a significance level of $p < 0.05$ are indicated with an asterisk. When a significant difference is found, the significantly different parameters are identified as those where the 95% confidence interval (also shown) does not overlap. Table 3 summarizes the results of the statistical testing.

The mean heart rate is clearly affected by the presence of the mouthpiece (80% for A1 and 40% for A2). With regard to the nonlinear estimated parameters $\rho$ and $\theta$, it is noticeable that in around 50% of experiments, for both subjects, the amplitude $\rho$ in the $+$MP is significantly larger than in the $-$MP experiments but that we

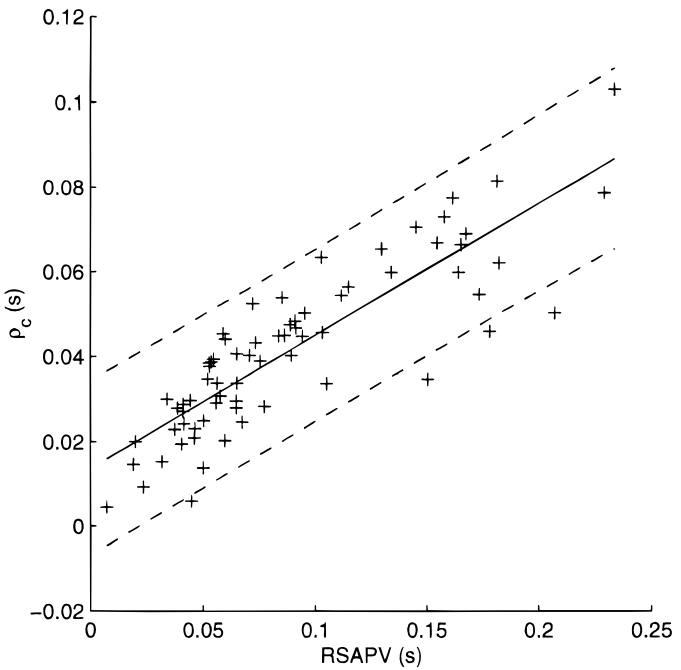


**FIG. 4.** The polar representation ($\rho_c$) as a function of the peak valley RSA estimate (RSAPV) for each recording session (total number = 70; see Table 1). The linear regression (solid line) yielded slope = 0.31 ($\pm$ 0.022), intercept = 0.014 ($\pm$ 0.0023), and adjusted squared correlation coefficient $R^2_{\text{adj}} = 0.7352$.

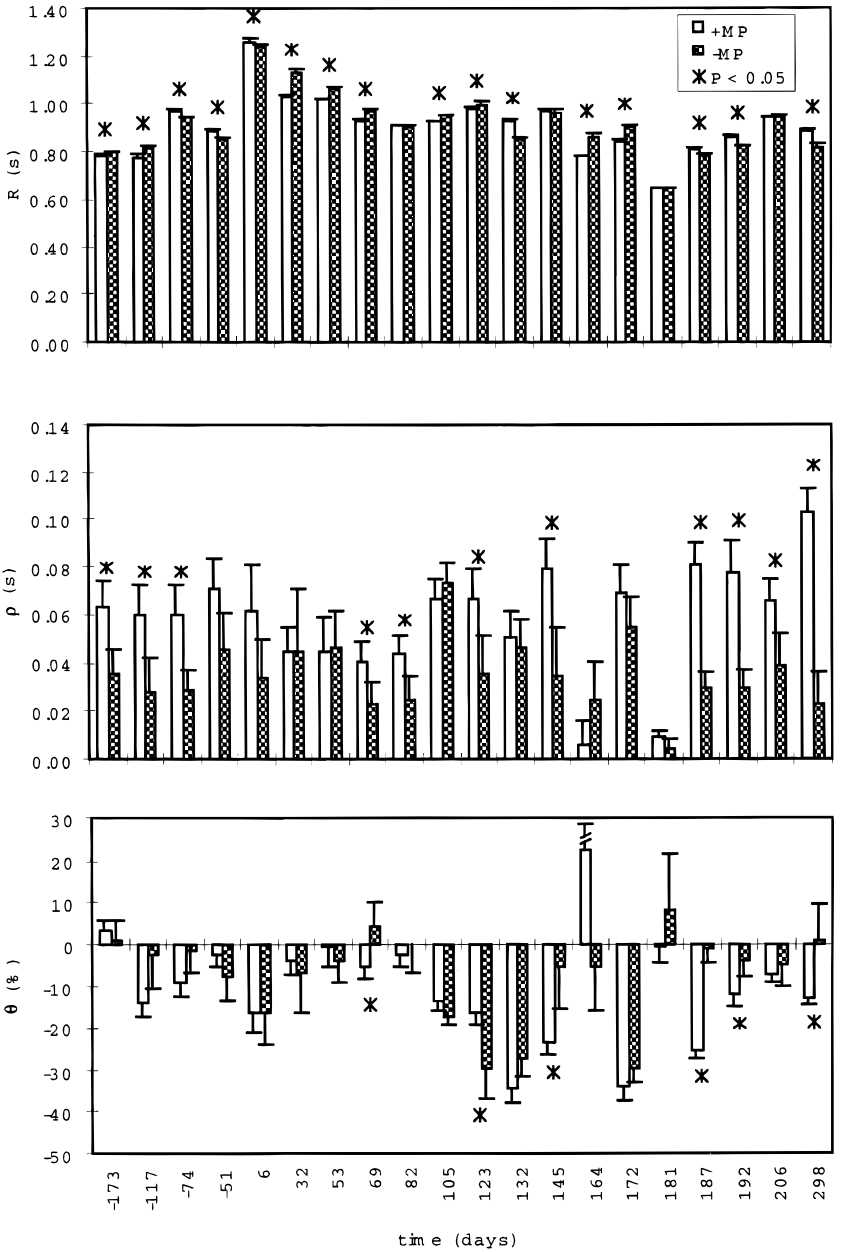


**FIG. 5.** Parameters of the model curve for subject A1 for each session of the mission and for +MP (white bars) and −MP (gray bars) experiments. Error bars are 95% confidence intervals. Asterisks indicates significant differences observed between +MP and −MP for the considered parameters at $p < 0.05$. (Top) $R$ (s) vs mission day (launch day = 0 and landing = 180). (Middle) $\rho$ (s) vs mission day. (Bottom) $\theta$ (%) vs mission day. On day 164 the error bar for $\theta$ is not presented on scale. Its value ($\pm 216\%$) is due to a very small amplitude of the cosine curve fit.

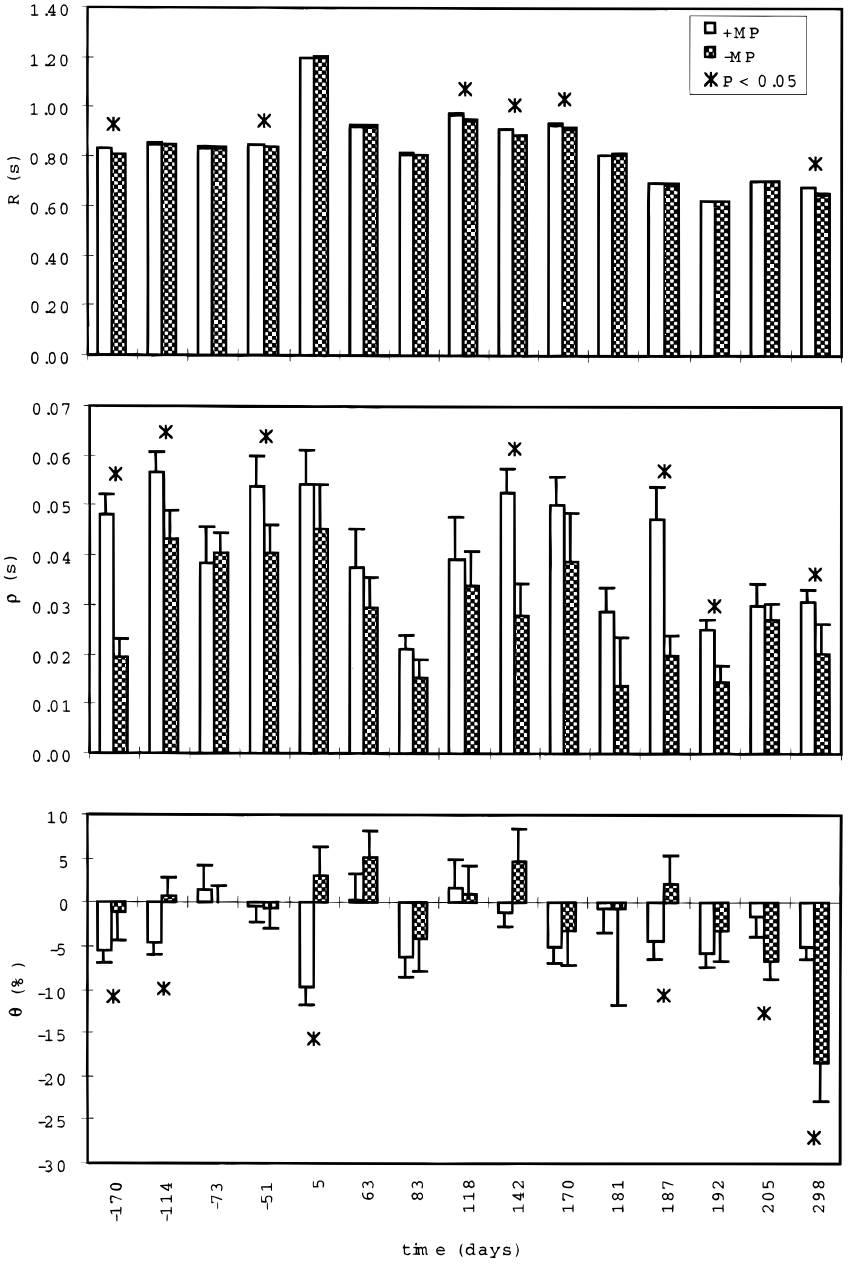


**FIG. 6.** Parameters of the model curve for subject A2 for each session of the mission and for +MP (white bars) and −MP (gray bars) experiments. Error bars are 95% confidence intervals. Asterisks indicates significant differences observed between +MP and −MP for the considered parameters at $p < 0.05$. (Top) $R$ (s) vs mission day (launch day = 0 and landing = 180). (Middle) $\rho$ (s) vs mission day. (Bottom) $\theta$ (%) vs mission day.

TABLE 3

Statistical Hypothesis Testing between +MP and −MP Experiments (Number of Significantly Different ($p < 0.05$)/ Total Number of Sessions)

| Parameters | Subject A1 (%) | Subject A2 (%) |
|---|---|---|
| $R$ | 80 | 40 |
| $\rho$ | 55 | 47 |
| $\theta$ | 30 | 47 |

cannot identify such a clear relation for the phase. Such results, which are impossible to obtain with the traditional spectral analysis, suggest that the presence of the mouthpiece amplify the RSA. This amplification can be due to a reflex mechanism originated in the brain or to a physical hemodynamic effect engendered by the respiration. The origin of this influence is beyond the scope of the present paper but our results suggest that the presence of a mouth-piece in HRV studies should be taken into account.

## CONCLUSION

We presented an algorithm designed for the analysis of the RSA in situations where the traditional spectral analysis is not applicable. Our time domain analysis consists of the normalization of the time axis with respect to the respiration cycle. Hence, this method is independent of the breathing frequency which is important for studies where voluntary control of breathing is not possible (e.g., during sleep) or not desired (e.g., normal breathing experiments). The RSA is analyzed through a nonlinear regression that provides the possibility of computing statistical hypothesis testing. Hence, intrasubject statistical testing of the evolution of RSA on short or very short (less than 1 min) recordings of the ECG and respiration are feasible. The amplitude of the RSA derived from the presented algorithm is in agreement with RSA computed with the time domain PV method.

The algorithm was used to compare the differences between normal breathing experiments with and without mouthpiece on the Euromir-95 mission. It is shown that, for the two subjects, breathing with a mouthpiece significantly increases the amplitude of the RSA and that the phase of this modulation of the RRI at the breathing frequency seems to be unaffected.

The advantage of our approach is a wider applicability to studies where respiration is not controlled and the possibility of computing significance level for the differences in parameter estimation between two different situations, even in short or very short recordings. In addition, information is obtained on the mean phase of the RSA with respect to the respiration which is not available from the more traditional methods.

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