Extraction and Analysis of Short-Time Excursions in RR-interval Time Series

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Abstract

RR-interval time series often present impulses corresponding to short-duration increases or decreases in the heart rate, most probably due to bursts in autonomic activity. The time-domain heartrate variability index pNN50 is obviously linked to these spikes. As linear filtering is not appropriate for the extraction of these spikes, we developed a nonlinear algorithm for this task. In this study we demonstrate the potential of this approach to assess caffeine-induced changes in the autonomic tone, and discuss the potential of this approach.

1. Introduction

Even after correcting for possible ectopic beats [1], observation of RR-interval (RRI) time series often reveals the presence of downward and upward impulses corresponding to short-term increases and decreases in the heart rate. Downward and upward impulses are predominantly present when the subject is respectively in a supine or standing position. In Figure 1 an RRI signal (regularly resampled at 4 Hz) from a healthy subject lying for the first 360 s and then standing up is displayed. The impulses as well as their inversion in polarity are clearly visible (two of them highlighted by the red impulses).

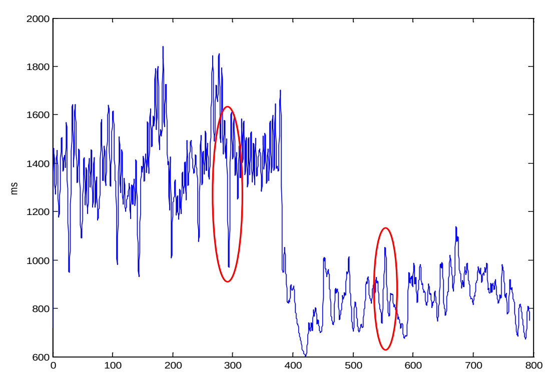

Figure 1. RRI from a healthy subject. Change from supine to standing at time 360 s. Two impulses are highlighted by the red ellipses.

This phenomenon can be observed in several studies published in the field of heart rate variability (HRV), figure 5, p. 361 of the classical reference [2] is an example. Yet, those impulses have rarely (never?) been the main focus of previous studies. Actually, several HRV parameters proposed in the relevant literature are more or less directly connected to these impulses. The most classical one [3] is the pNN*x*, i.e. the mean number of times per hour that successive RRI differ by more than *x* ms, the most common value being $x = 50$. Also, the SD1 parameter corresponding to the minor axis of the ellipse fitted to the Poincaré plot of an RRI time series measures the short-term variability of the latter [4]. An extension of this approach, the complex correlation measure (CCM) aims at quantifying the point-to-point variation of the RRI rather than the global short-term variability [5].

However, direct extraction of these impulses is a difficult problem due to their limited time extent and asymmetry, and due to respiratory sinus arrhythmia (RSA). In [6], an encoding of RRI using a bank of optimized linear filters was proposed, with short filters corresponding to fast regulation mechanisms. These filters being bandpass ones, only short-term *oscillations* can be retrieved. The same problem arises with the nonlinear approach described in [7], which proposes a sparse joint decomposition on a pair of wavelet bases with high and low Q-factors. Again, only short- and long-term oscillations can be separated. Extraction of asymmetric impulses is not possible using these methods.

Hence we propose in this paper to use a simple nonlinear scheme initially designed to enhance fast change, such as QRS complexes in an electrocardiogram signal [8]. We first present this scheme and its adaptation to the problem at hand. Next we illustrate on a small database how the extracted impulse signals can be used to assess drug-induced changes in autonomic activity.

2. Method

2.1. Principle of the extraction method

The principle of the so called relative-energy (RE) algorithm [8] is quite simple. Since an impulse is characterized by a local surge in signal energy, the idea is to measure the ratio between a short-term variance estimate and a long-term variance estimate on a sample-by-sample basis. The larger this ratio is, the more impulsive the local signal is. Quite naturally, the short-term and long-term variances are estimated respectively on two windows with odd lengths 2S+1 and 2L+1, with L >> S centred on successive signal samples. The resulting RE and the original signal are then multiplied sample-by-sample to yield an impulse-enhanced signal.

This approach does not conserve amplitudes, i.e. the extracted impulses are known only up to a multiplicative coefficient. While not relevant in a detection context, this raises issues if impulse extraction is sought for. An empirical solution consists in scaling the resulting impulse signal so that, when subtracted from the original signal, the resulting signal has minimum skewness, i.e., the asymmetry induced by the impulses is minimized.

Also, extraction is improved by applying the RE scheme in additional iterations (by experience two iterations is usually enough) to the successive impulse signals, in order to remove spurious impulses. Of course skewness minimization should still be performed using the original signal.

Figure 2 presents an example on a 2000-sample signal generated as follows: a sinusoid with normalized frequency 0.05 was created, and for five half-periods the amplitude was multiplied by a factor of two or three. This signal presents some similarity with an RRI one, as the sinusoid corresponds to an RSA at a breathing frequency of 0.2 Hz for a standard sampling frequency of 4 Hz, and the impulses are similar to those in figure 1. The lengths of the short and long windows were 2S+1=7 and 2L+1=61samples. Figure 2, from top to bottom, displays the original signal, the true impulse signal (original signal minus sinusoid), the first estimated impulse signal, and the estimated impulse signal after two iterations. The error-to-signal variance ratio between the true impulse signal and the estimated one is 0.027. The relative amplitude errors for the smaller impulses are about 0.001, but they amount to 0.09 for the two largest impulses.

2.2. Adaptation to RRI signals

The RE extraction scheme can of course be applied on the raw RRI time series. In order to make time and frequency interpretations possible, we decided to perform this analysis on cubic-spline regularly-resampled RRI signals using the standard 4 Hz sampling frequency.

Components present in most RRIs, and that would obviously impair the performance of the RE scheme, are the positive mean value and ultra-low frequency (ULF, <0.01 Hz) activity. To remove them we use singular spectrum analysis (SSA) [9] with a window length of 80. The SSA component with the smallest frequency (containing both the mean value and the ULF) is subtracted from the signal. Figure 3 illustrates the impulse extraction process on a recording from a healthy subject in supine position. The top graph displays the raw regularly-resampled RRI signal. The middle graph displays the RRI signal after mean/ULF subtraction. The bottom graph displays the extracted impulse signal, two RE iterations, window lengths 2S+1=7 and 2L+1=61. The impulses are obviously mainly negative ones, which is, as mentioned in the Introduction, a common feature for RRIs in the supine position.

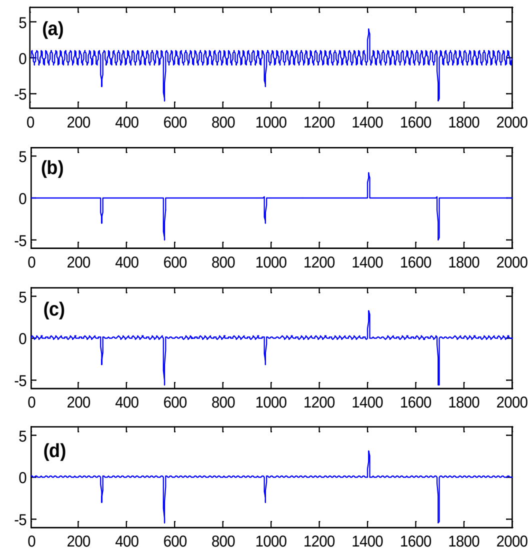

Figure 2. (a) Original signal. (b) True impulse signal. (c) Estimated impulse signal, first iteration. (d) Estimated impulse signal, second iteration.

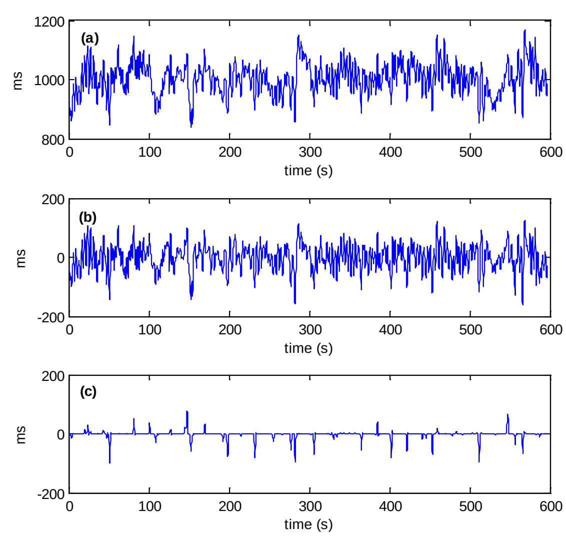

Figure 3. (a) Original RRI. (b) RRI after subtraction of mean/ULF. (c) Estimated impulse signal, two iterations.

3. Materials

Data were recorded from 15 young healthy subjects. The protocol followed the Declaration of Helsinki and was approved by the Ethical Committee of Lausanne University. All participants provided oral and written informed consent prior to participation.

Airflow was monitored breath-by-breath (Medgraphics, CPX, St. Paul, MN, USA) at the mouth (Pitot tube). 3-lead ECG was monitored using an analogue amplifier. The ECG and airflow were acquired simultaneously at 1000 Hz, using an analogue-to-digital converter (PowerLab 16/30, ADInstruments, Bella Vista, Australia) and recorded with commercially available software (LabChart v.7.2 ADInstruments, Bella Vista, Australia). The participants were asked to abstain from caffeine, heavy exercise and alcohol for 12h. The subjects underwent recording first in control condition and then under the influence of caffeine (6 mg/kg) administrated by pills. Each recording session consisted of 10 minutes spontaneous breathing (SB), 10 minutes breathing at 9 breaths-per-minute (brpm) and 10 minutes breathing at 12 brpm in a randomized order. To ensure correct cadence of breathing, the subjects were instructed to follow continuously a metronome at 9 and 12 brpm. Baseline recordings were performed in all three breathing modes before the subjects ingested the caffeine. Caffeine recordings started 45 minutes after the ingestion. RR-intervals were extracted from the ECG and regularly resampled at 4 Hz after compensation of ectopic beats.

4. Results

The impulse signals were extracted from all the RRI ones, with two RE iterations and window lengths 2S+1=7 and 2L+1=61samples. We characterized the asymmetry in the impulse signals by their mean. For comparison purposes, we computed the power in the LF band (0.04 – 0.15 Hz), its version normalized by the total power nLF, and the ratio LF/HF between the power in the LF band and that in the HF band (0.15 – 0.4 Hz). As caffeine elicits a sympathetic reaction, one should expect a decrease in the mean of impulse signals (i.e. more negative impulses), and an increase in LF, nLF, and LF/HF, from baseline to caffeine conditions. In Table 1, the numbers of subjects out of 15 for whom the impulse mean decreased, and the LF, nLF, LF/HF increased, are listed for the three respiration modes. One can observe that the impulse mean decreases in most subjects, especially in the spontaneous breathing mode, while increases in the LF-related values are far less predominant. The somewhat paradoxical LF/HF decrease in 14 subjects for the respiration rate of 9 brpm may be explained by the fact that this rate corresponds to a frequency of 0.15 Hz, i.e. the upper and lower bound respectively for the LF and HF bands. The RSA at the same frequency has a major (and in this case negative) impact on the LF/HF ratio. Table 2 displays the asymptotic *p*-values for the two-sample Kolmogorov-Smirnov test, that confirm the significant (marginally at 12 brpm) changes in impulse mean induced by caffeine.

Table 1. Number of subjects out of 15 with a change from baseline to condition coherent with caffeine-induced sympathetic activation.

respiration	SB	9 brpm	12 brpm
Imp. mean	14	12	11
LF	9	8	7
nLF	7	8	4
LF/HF	9	1	7

Table 2. Asymptotic *p*-values for the two-sample Kolmogorov-Smirnov test between baseline and caffeine condition.

respiration	SB	9 brpm	12 brpm
Imp. mean	0.001	0.017	0.31
LF	0.89	0.99	0.89
nLF	0.89	0.59	0.89
LF/HF	0.89	0.052	0.99

5. Discussion

In this paper we intend to draw the attention of the HRV community on the impulses that can be often observed in RRI signals. We propose an empirical method, the nonlinear nature of which makes approximate impulse extraction possible. There is interest in elucidating the origin of these impulses, but this requires obviously microneurographic recordings of sympathetic and vagal activities. Let us only mention that, in most RRI signals in our study, the average time interval between large negative (due to supine position) impulses was between 40 and 50 seconds.

In terms of applications, extraction of the impulse signal presents interesting aspects that have been illustrated in our experiments. Specifically, although these impulses represent but a small fraction of the total HRV power, they seem to be representative of the sympatho-vagal balance, and are have the advantage to be insensitive to RSA fluctuations and independent of the LF and HF bands. In the present study, we used only a very simple parameter, namely impulse signal mean value, but more elaborate measures can be used, for instance by processing positive and negative impulses separately.

A salient point in our experiments is the poor sensitivity of frequency-based HRV parameters to caffeine-based sympathetic activation, especially at a respiration rate of 9 brpm. These parameters are widely used in bio-psychological works, in stress assessment for instance [10]. However, a factor that is often overlooked is respiration. Of course in most subjects the average respiration (and thus RSA) frequency is above 0.15 Hz, but it may intermittently cross this boundary. Also, especially in supine position, subjects such a athletes may have a respiration frequency below 0.15 Hz. Impulse extraction, that is not influenced by respiration, does not present this problem.

Subtracting the extracted impulse signal from the original RRI one may also be useful prior to bandpass filtering of the latter. Indeed, the impulses may contaminate the filter output due to their wideband nature. Figure 4 illustrates this point. The upper plot is an RRI signal regularly resampled at 4 Hz, acquired from a normal subject in supine position with an imposed respiration frequency of 0.225 Hz. A zero-phase bandpass filter centred on this frequency was used to extract the RSA, both on the raw RRI signal and after impulse signal subtraction. The bottom plot shows the instantaneous frequency (IF) estimates of the two RSAs obtained through the Hilbert transform (HT). The thin and thick lines are the RSA IF estimates respectively without and with impulse subtraction. As is well known, the HT-based instantaneous frequency estimation gives sensible results for narrowband signals only. Visibly, the IF estimate for the original RRI signal contains aberrant values, especially around times 25 s and 180 s for this reason. The IF estimate after impulse subtraction is more coherent.

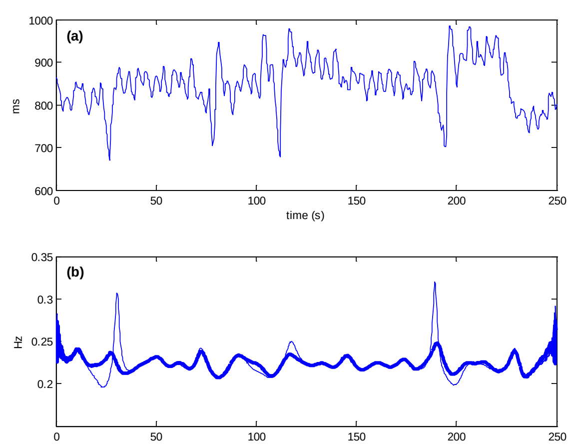

Figure 4. (a) Original RRI signal. (b) Thin line: RSA IF estimate using the original RRI signal. Thick line: RSA IF estimate after impulse subtraction.

6. Conclusion

Impulses are often present in RRI signals, and correspond to short-term accelerations and decelerations of the heart rate. In this paper we present a nonlinear scheme to extract these impulses from an RRI signal, and demonstrate that features drawn from the extracted impulse signals are of interest to assess sympathetic activation. In the near future, we plan to explore possible clinical uses of these impulses.

References

[1] Citi L, Brown EN, Barbieri R. A real-time automated point-process method for the detection and correction of erroneous and ectopic heartbeats. IEEE Trans Biomed Eng 2012;59:2828-37.

[2] Camm A, Malik M, Bigger J, Breithardt G, Cerutti S, Cohen R, Coumel P. Heart rate variability: Standards of measurement, physiological interpretation and clinical use. Task Force of the European Society of Cardiology and the North American Society of Pacing and Electrophysiology. Circulation 1996;93:1043-65.

[3] Mietus JE, Peng CK, Henry I, Goldsmith RL, Goldberger AL. The pNNx files: re-examining a widely used heart rate variability measure. Heart 2002;88:378-80.

[4] Tulppo MP, Makikallio TH, Takala TES, Seppanen T, Huikuri HV. Quantitative beat-to-beat analysis of heart rate dynamics during exercise. Am J Physiol 1996;271:H244–H252.

[5] Karmakar CK, Khandoker AH, Voss A, Palaniswami M. Sensitivity of temporal heart rate variability in Poincaré plot to changes in parasympathetic nervous system activity. BioMed Eng OnLine 2011;10:17.

[6] Lucena F, Barros AK, Principe JC, Ohnishi N. Statistical Coding and Decoding of Heartbeat Intervals. PLoS ONE 2011;6:e20227.

[7] Selesnick IW. Resonance-based signal decomposition: A new sparsity-enabled signal analysis method. Signal Process 2011;91:2793-809.

[8] Yazdani S, Vesin JM. A novel preprocessing tool to enhance ECG R-wave extraction. Computers in Cardiology 2016;43.

[9] Harris TJ, Yuan H. Filtering and frequency interpretations of Singular Spectrum Analysis. Physica D 2010;239:1958-67.

[10] Choi J, Ahmed B, Gutierrez-Osuna R. Development and evaluation of an ambulatory stress monitor based on wearable sensors. IEEE Trans Inform Tech Biomedicine 2012;16:279-86.

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