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EFFECT OF CYCLING ON THE HEART RATE VARIABILITY : A SIMULATION STUDY

O. Meste, G. Blain, S. Bermon

Equipe SIGNAL

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Laboratoire d'Informatique de Signaux et Systèmes de Sophia Antipolis - UNSA-CNRS
2000, rte.des Lucioles – Les Algorithmes – Bât Euclide B – B.P. 121 – 06903 Sophia-Antipolis Cedex – France
Tél.: 33 (0)4 92 94 27 01 – Fax: 33 (0)4 92 94 28 98 – www.i3s.unice.fr
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1 Processing background

Spectral analysis techniques, such as Fourier transform or autoregressive modeling, have been the most extensively used methods to quantify HRV [1]. Although under stationary conditions the quantification of the spectral domain is a simple task, during dynamic exercise condition, the variation of the spectrum frequencies and of their relative amplitude makes it more challenging. In [3], we showed that using a time-varying approach, both time-varying frequencies and amplitudes can be estimated from the R-R intervals series recorded during maximal and graded exercise. In [4], the application of this signal processing method evidenced a linear relationship between the amplitude of respiratory sinus arrhythmia (RSA) component in HRV and ventilation during intense exercise. In this application, the amplitude measurement was based on a time-varying filtering. We can get rid of the filter design difficulty by computing directly from the short-time Fourier transform (STFT) the quantity :

$$R(k) = \sqrt{\frac{1}{K} \sum_{f=f_{obs}(k)-\delta}^{f_{obs}(k)+\delta} |M(k,f)|^2} \tag{1}$$

with $f_{obs}(k)$ the time-varying frequency of interest. $M(k,f)$ is the STFT of the R-R intervals variability as defined in [3] :

$$M(k,f) = \sum_u m(u)h(u-k)e^{-j2\pi\frac{\ell}{K}u} \tag{2}$$

$$\text{with } -K/2 \leq \ell \leq K/2-1 \text{ integer and } f = \ell/K$$

The analysis window $h(u)$ is energy normalized. In our application, aiming to quantify a pedaling frequency component (PFC) in HRV, the value of the parameter δ was set $\pm$ 0.1Hz around the imposed pedaling frequency. Among the available time-frequency representation, the STFT get the advantageous to be linear and thus allow a direct implementation of a time-varying filter [3]. Although in this application only the envelop is of interest, additional information can be obtained from the filtered signal. It is well known that the STFT has a limited time-frequency resolution. Not only this resolution is sufficient in our application, but also the STFT has the advantageous to be less sensitive to interferences such as cross-term interference, that are known to distort the analysis process [5]. If other time-varying approaches, such as the Wavelet, are better adapted to signals that contains transient events at different scales, the STFT is fully adapted to time-varying spectral lines quantification [5].

Moreover, although the frequency of interest, i.e. the pedaling frequency f_p, is maintained constant in the continuous time domain in our experimental setting, this frequency becomes time-varying when extracted from in the heart period signal such as :

$$f_{obs}(k) = po(k)f_p \tag{3}$$

where $po(k)$ is the trend (or instantaneous mean heart period) of the heart period. Because of workload increase, the quantity $po(k)$ will vary with time during a maximal and graded exercise test and so forth for the observable pedaling frequency.

It should be mentioned that the shanon criteria apply for this observation because the continuous pedaling signal is sampled at the R-wave occurrence with a time-varying sampling period ($\approx po(k)$). This criteria imposes that the sampling frequency (the inverse of the sampling period) should be twice the maximum frequency f_{max} of the continuous observation. If this condition is not fulfilled, the spectrum that lies in the frequency band greater than half the sampling frequency will be duplicated in a frequency band lower than half the sampling frequency. This effect will mislead the interpretation and quantification of the spectrum. For instance, with a pedaling cadence of 70 rpm the spectrum will be free of aliasing when $po(k)$ (the mean heart period) is lower than 430 ms (heart rate higher than $\pm$ 140 bpm). For a pedaling rate of 90 rpm, the limit will be more drastic since $po(k)$ should be lower than 330 ms (heart rate higher than $\pm$ 180 bpm). In case of aliasing, care should be taken in the interpretation of results when the duplicated frequency is superimposed to another one, such as the RSA component.

2 Analysis of simulated heart period signals

In the following simulation, we chose a pedaling frequency equal to 90rpm and used cardiac and ventilatory characteristics of regular trained cyclists. The analysis was conducted by using different scenarios in order to check the validity of the method and reject any bias suspicion :

- A typical signal is synthesized in order to show what is the output of the proposed method
- A noisy pedaling interference is processed only in order to check the agreement between the estimated quantity and the synthesized one
- A combination of respiration and pedaling interference is used to measure the interaction of the two interferences in the estimation process
- A monte-carlo simulation which includes uncorrelated and stationary noise only is proposed to evaluate the possible bias introduced by the envelop estimation.

2.1 Simulation one

The pedaling frequency signal has been added to the respiration signal (related to RSA) in order to produce a synthetic ECG [3]. A R wave detector has been applied to the ECG. A representation of the heart period signal or tachogram (intervals between successive R waves) is given in figure (1). In figure (2), the corresponding variability signal $m(k)$ and the trend $po(k)$ are given. Note that for this pedaling frequency (90rpm) the aliasing effect should disappear when the trend is below 330ms as explained in the previous section. In figure 3, the respiratory and pedaling components are clearly visible in this representation of the amplitude of the STFT. The two white dashed lines correspond to the frequency bounds used for the envelop estimation (1). The component at the normalized frequency equal to 0.15 is due to the respiration and the second more visible component is related to the pedaling frequency. Because the Shannon constrain is not fully verified, the first part (k from 0 to 1200) is an aliased spectrum that is superimposed to the respiration component at around $k = 300$. The interval [1200-end] contains non-aliased component because in that case the heart rate is 2 times higher (corresponds to a heart period lower than 330 ms) than the pedaling frequency (see section 1). Using (1), the amplitude of the pedaling component is obtained from the STFT. Assuming that the PFM model is valid [3], the corrected result is plotted in figure 4 where the dynamic behavior of PFC is clearly exhibited. Note that the overlapping of PFC and RSA resulted in a significant amplitude increase at approximately $k = 300$.

2.2 Simulation two

Noise was added to the synthesized ECG, which contains only the pedaling frequency modulation. The STFT of the corresponding tachogram is given in fig. (5) and the estimated envelop

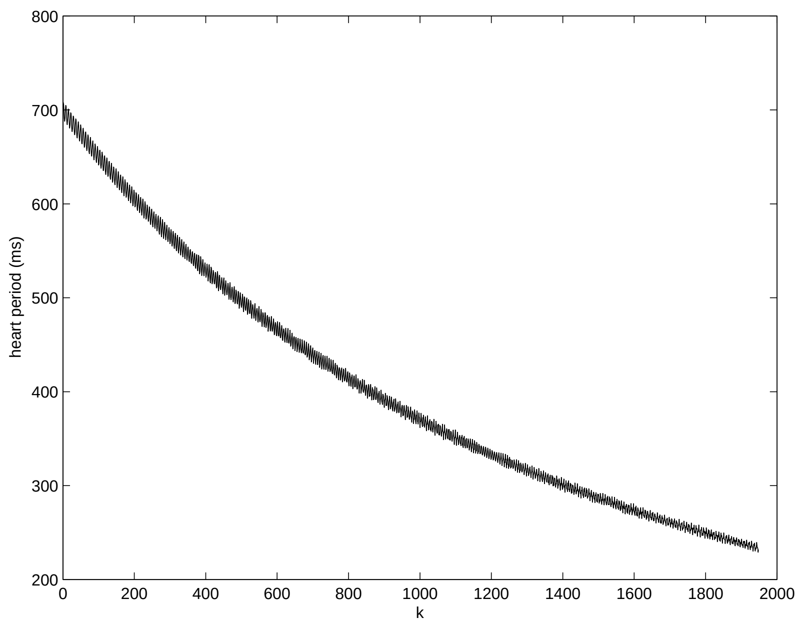

FIG. 1 – Heart period from the simulated ECG

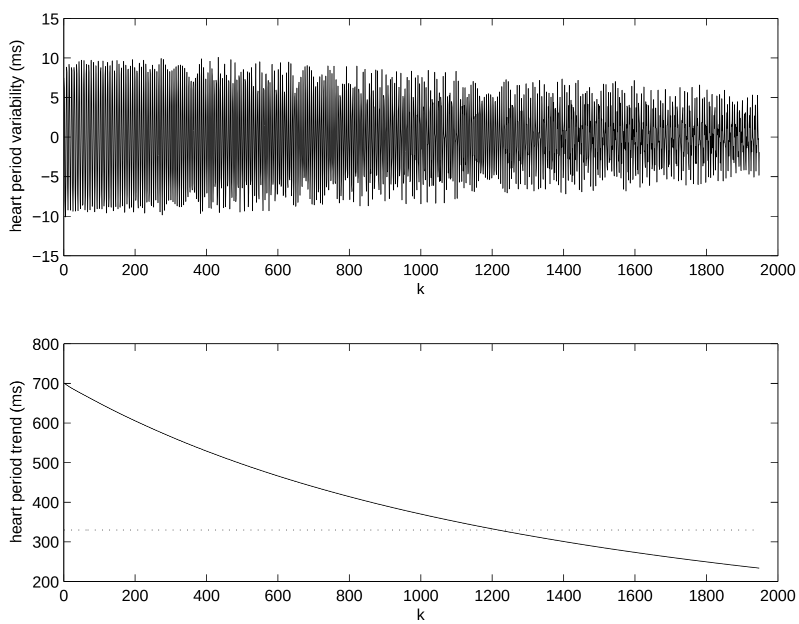

FIG. 2 – The heart period variability signal that is going to be time-varying filtered (upper panel) and the heart period trend (lower panel). The dotted line in the lower panel corresponds to the heart period limit above which aliasing of the PFC component occurs

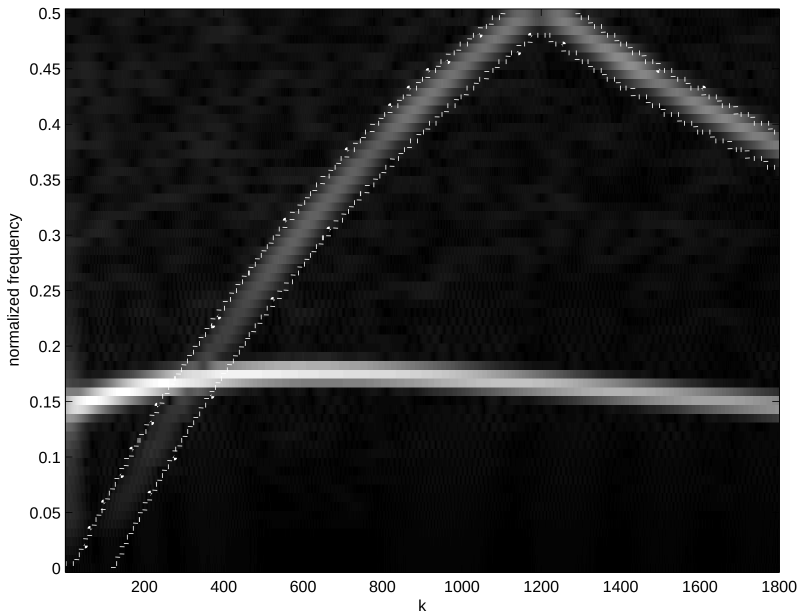

FIG. 3 – Amplitude of the short time fourier transform on the complete heart period signal

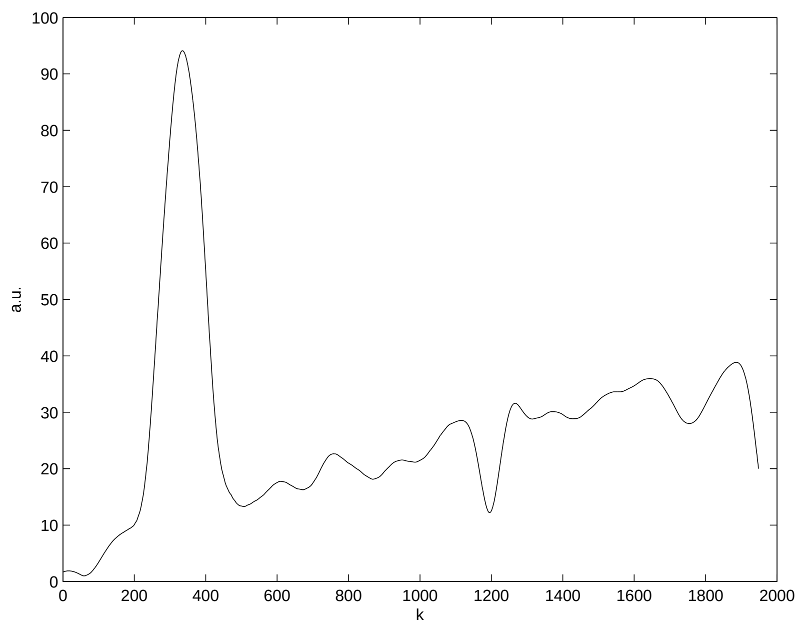

FIG. 4 – The quantified PFC from the short time fourier transform

(1) in fig. (6). The good agreement of this estimation and the real amplitude is clearly visible in fig. (6). The visible gap at approximately k=1200 is due to the aliasing effect (see fig. (5) at the same k index).

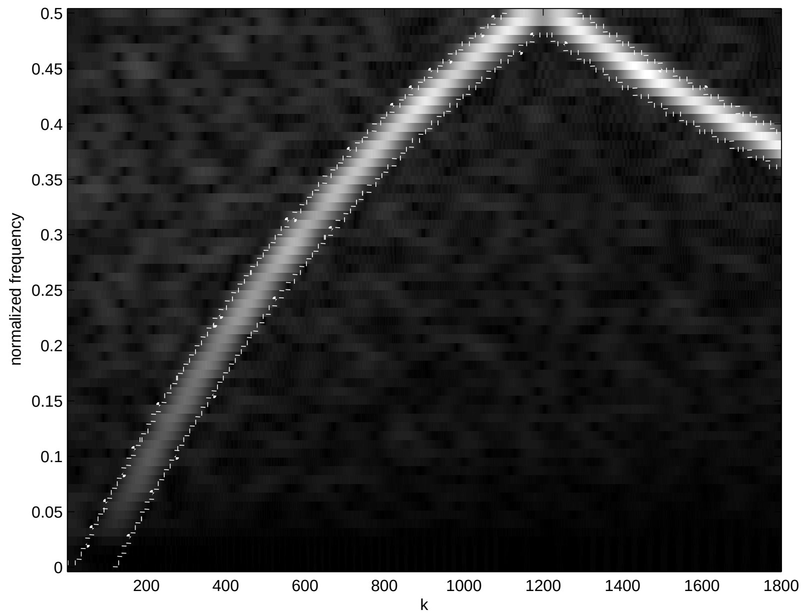

FIG. 5 – Amplitude of the short time fourier transform on the heart period signal that contains only PFC.

2.3 Simulation three

We have added to the previous pedaling interference a respiration effect that is decreasing and higher in amplitude. The STFT of the corresponding tachogram is given in fig. (7). The estimated pedaling envelop is illustrated superimposed to the simulated tachogram in fig. (8), and compared to the previous simulation analysis in fig. (9). Here again the agreement between the simulated signal and extracted signals is clearly visible except for the overestimation at approximately k=325, which is due to the PFC and RSA overlapping. Note that because the noises added to the simulated ECGs are different in simulation one, two, and three, estimated envelops are slightly different.

2.4 Simulation four

A monte-carlo simulation has been conducted over 100 trials. For each trial a noisy ECG has been synthesized without pedaling interferences. Note that for this simulation the respiration modulation and the noise effect have been multiplied by ten. The corresponding envelop has been estimated supposing that the pedaling interference is present. One STFT result is given in fig. (10) where the respiration component is visible. In fig. (11), the average of the 100 estimated data is shown. Importantly, this simulation study showed that the mean envelop remained flat and did not increase over time (except at k=325), which is consistent with the characteristics of an uncorrelated and stationary noise. The spike in amplitude at approximately k=325 is due to the respiration component that lies in the expected pedaling frequency bounds (see white dashed lines in fig. (10)). In the simulation the added noise has been chosen. This explain the flatness of the result.

In conclusion of this simulation study, and in accordance with the positive visual inspection of the STFT we performed prior to apply the envelop estimation, our results confirm that the

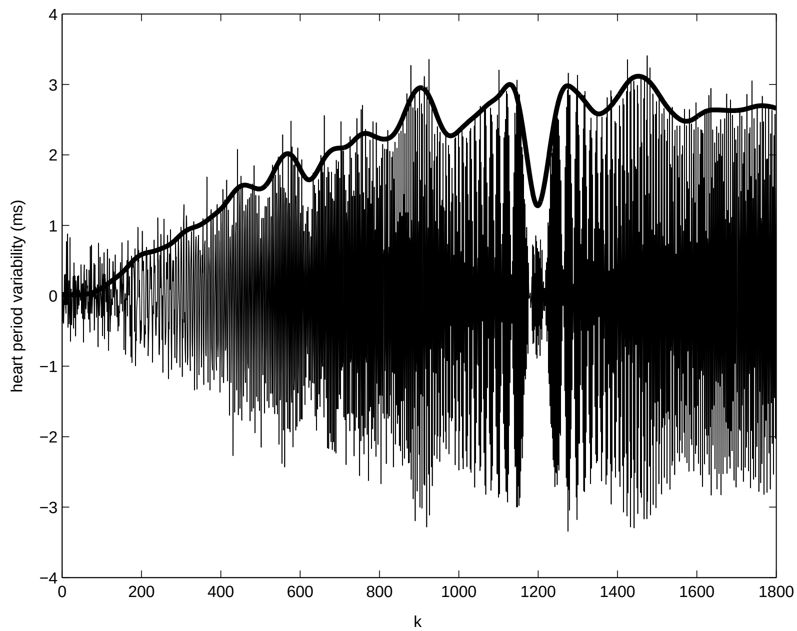

FIG. 6 – The heart period variability that contains only the pedaling component and its estimated envelop from the STFT (thick line)

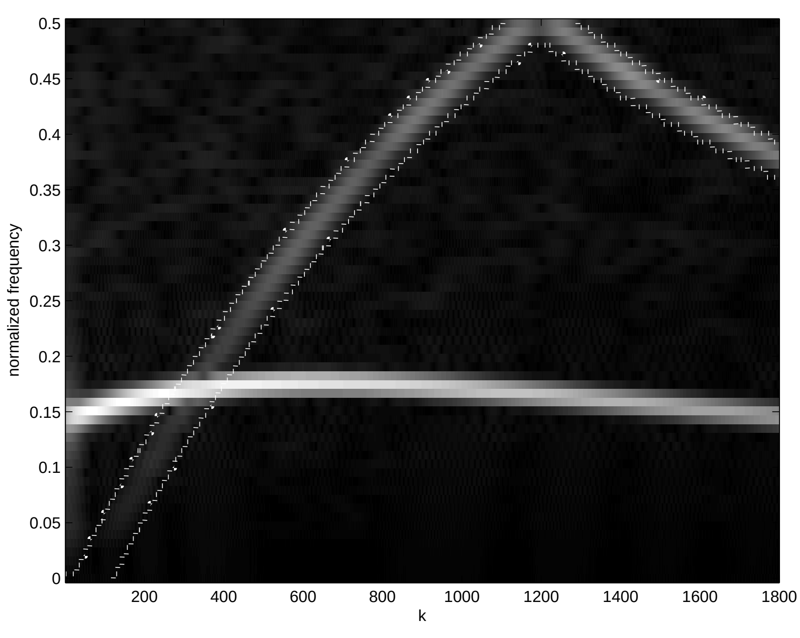

FIG. 7 – Amplitude of the short time fourier transform on the complete heart period signal that contains both the pedaling and respiratory components

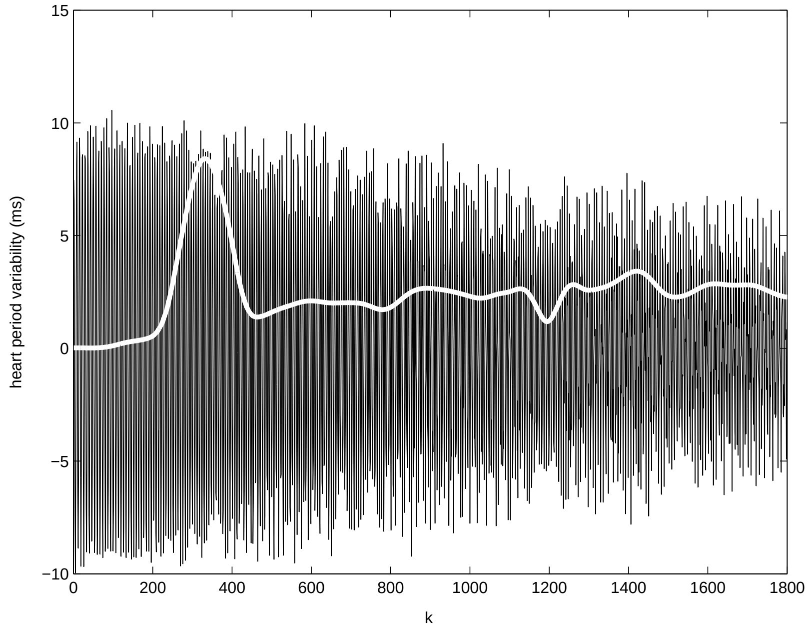

FIG. 8 – The heart period variability that contains the pedaling and respiratory components superimposed to the estimated pedaling envelop from the STFT (white line)

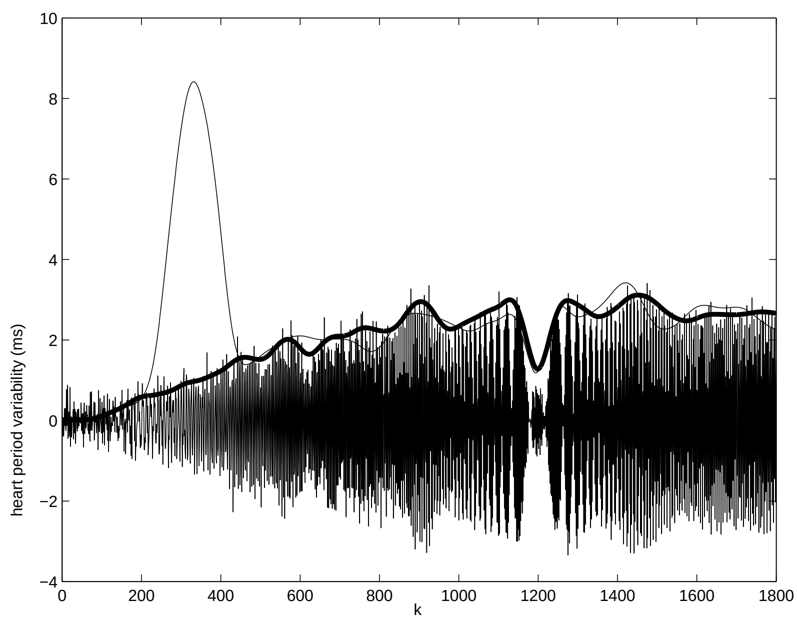

FIG. 9 – The embedded pedaling signal in fig. 8 and the estimated envelop of the pedaling signal from fig. 6 (pedaling component only)(thick line) and from fig. 8 (pedaling plus respiration components)(thin line). Note that the noise realizations are different

dynamic trend of the pedaling envelop observed in R-R intervals series recorded in healthy men during dynamic exercise is unlikely due to a bias in the estimation process or method of analysis.

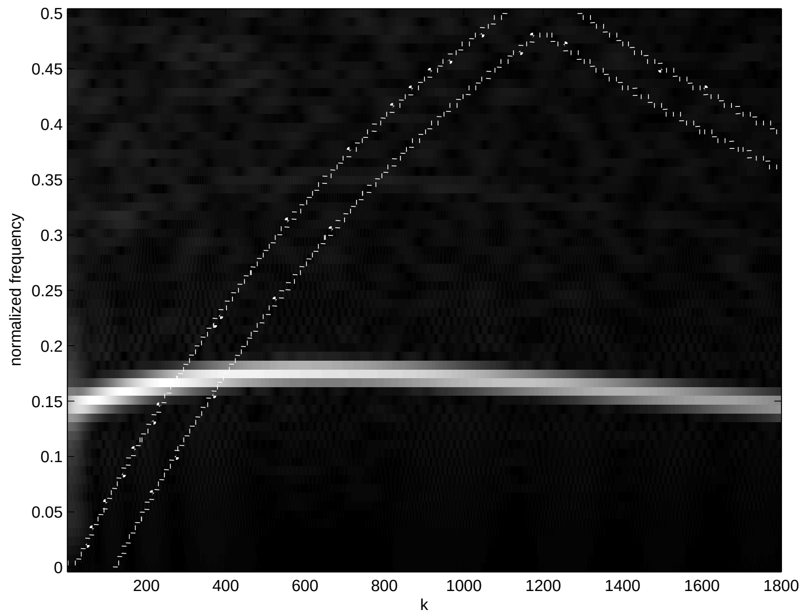

FIG. 10 – Amplitude of the short time fourier transform of one realization of the noisy respiration (pedaling signal not added)

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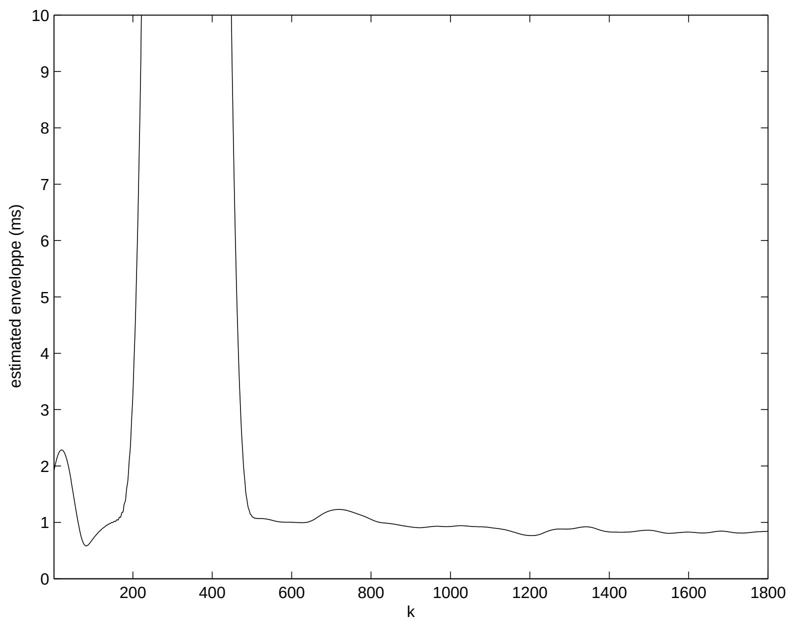

FIG. 11 – Average of the estimated envelop of the pedaling signal from the 100 noisy respiration signals (pedaling signal not added)