Spectral analysis of atrial signals directly from surface ECG exploiting compressed spectrum

P Bonizzi, O Meste, V Zarzoso

Laboratoire I3S, University of Nice - Sophia Antipolis, France

Abstract

Atrial fibrillation dominant frequency (AFDF) has been demonstrated to provide useful information on the characteristics of atrial fibrillation. The present work points forward a new method for AFDF estimation directly from a single-lead ECG recording, exploiting the concept of compressed spectrum as spectral estimator. The main purpose is to verify if ventricular activity subtraction might be avoided when the detection of the AFDF is the main goal. The greatest advantage of the proposed method is that it does not require any detection of the QT interval, making it very low cost and simple. Comparison of the estimated AFDFs with those obtained from the analysis of the corresponding atrial signals extracted using a Bayesian spatio-temporal QRST-cancellation technique showed that the two estimates were very similar. We conclude that the estimation of the AFDF can be pursued directly from a single-lead ECG recording, without any kind of QRST-subtraction or QT-detection.

1. Introduction

Atrial fibrillation (AF) represents the most common sustained cardiac arrhythmia in adults. It consists of a misfunction of the atrium characterized by a modification of the normal atrial activity (AA) pattern on the electrocardiogram (ECG) signal.

The accurate extraction of the AA signal from the ECG of AF is of great interest for subsequent analysis, since it has been documented to provide significant information on the properties of AF episodes [1]. In particular, a good estimate of the AA signal is important for an accurate analysis of the temporal evolution of the spectral content of the AA signal. This analysis is justified by the evident correlation between the spontaneous termination of the episode and the decreasing trend of the AF dominant frequency (AFDF) [2]. Still, AFDF is a feature of clinical relevance to assess drug treatment and to predict the outcome of cardioversion or ablation therapies.

Generally, AFDF characterization is obtained through spectral analysis of the remainder AA signal extracted from ECG recordings during AF. This extraction requires the cancellation of the signal components associated with ventricular activity (VA), that is, the QRS-T complex. However, a lot of facts hinder this operation. In particular, the much lower amplitude of the AA signal compared to the ventricular one and the spectral overlapping of the two phenomena [3].

Therefore, the remainder AA signal derivation needs advanced signal processing techniques, characterized by a high computational cost. There exist in the literature two different families of methods applied to cancel out VA in the ECG. The first involves spatio-temporal QRS-T cancellation methods [4, 5], while the second involves all the methods based on the blind source separation (BSS) approach [3, 6].

However, the detection of the AFDF is often the main goal in clinical practise, one can wonder whether VA cancellation could be avoided performing a spectral analysis directly on the original ECG recording. Different authors have already made an effort in this direction, generally performing different kinds of spectral analysis on those intervals in the ECG where VA is absent (i.e., the TQ intervals). Rosenbaum & Cohen [7] averaged periodograms computed on all the several TQ intervals at disposal. One limitation of this approach is the low spectral resolution. To overcome this problem, Sassi *et al.* [8] observed that the successions of TQ intervals, obtained by removing the ECG portions affected by QRS-T waves, could be treated as a time-discontinuous (or unevenly sampled) signal. In that way, the problem can be handled as a missing data problem in a long time series, and they explored two different techniques to detect the AFDF from single-lead analysis of ECG recorded during AF: the Lomb periodogram and the Iterative Singular Spectrum Analysis. They showed that both algorithms were able to provide reliable estimates of AFDF, suggesting that AFDF estimation is feasible without applying QRS-T cancellation. However, common limitations of these two papers are the need to isolate ECG segments containing QRS-T waves (QT intervals). This procedure involves a certain computational cost, and does not solve the problem of vanishing

TQ interval at high heart rates, so that this kind of ECG recordings cannot be handled by these methods.

At this time, it is useful to consider that AF can be properly supposed to contain some harmonicity, inversely dependent from its level of disorganization and criticality. Several works attest the importance of exploiting this harmonicity as a contribution to AF description (e.g., see [9]). Generally, the harmonicity is considered to be negligible after the second harmonic. Starting from these considerations, the method proposed in this contribution exploits the concept of compressed spectral analysis, whose principle is to accumulate the harmonics of the searched fundamental frequency by contracting the frequency axis of the signal spectrum. Appropriately defined for the problem under analysis, this approach allows the evaluation of the AFDF directly from a single-lead ECG recording, without needing QT interval removal and missing data approaches.

2. Methods

2.1. Data and preprocessing

A dataset composed of 22 recordings (all presenting AF) was employed to apply the proposed idea. All signals were recorded and digitized at a sampling rate of 1 KHz. All the segments employed in this analysis were recorded using a standard 12-lead system. Pre-processing was done by applying a zero-phase high pass filter with a -3 dB cut off frequency at 0.5 Hz to remove physiologically irrelevant low frequency signal variations ($<$1 Hz) [10]. A zero-phase notch filter at 50 Hz was implemented to suppress power line noise [11].

2.2. Spectral compression

The principle underlying spectral compression is to accumulate the harmonics of the searched fundamental frequency of the signal spectrum [12]. Compression consists in applying a transformation with an integer scaling factor to the frequency axis (a contraction) and then summing compressed spectra values. The expected goal of this method is that all the harmonics' amplitudes add only to the fundamental frequency, not to other frequencies of the selected range, rising the fundamental frequency peak in the spectrum. Compressed spectrum (CS) is defined as a power spectral density (PSD), as follows

$$CS(w_0) = \sum_{k=1}^{n(w_0)} P_k(w_0) \qquad (1)$$

where $P_k(w_0) = P(kw_0)$ is the power spectral density compressed of a factor k, $n(w_0)$ is the number of harmonics exploited and $k = 1$ corresponds to the fundamental. To illustrate the method we consider a simulated AF signal

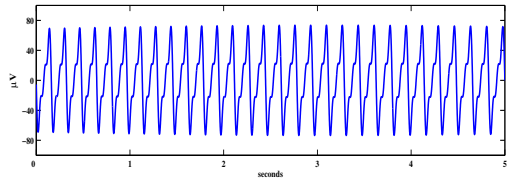

Figure 1. Simulated atrial fibrillation signal.

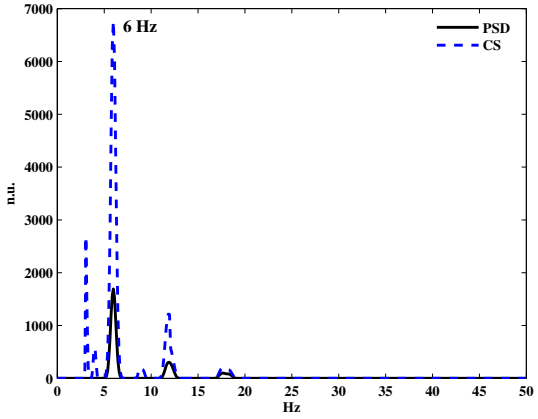

Figure 2. Power spectral density (PSD, thick line) and compressed spectrum (CS, dashed line) of the simulated AF signal in Fig. 1; n.u., normalized units.

obtained as described in [4], with the setup of parameter defined as follows

Parameter	Value
Frequency, f_0	6 Hz
Frequency variation, Δf	0.2 Hz
Frequency variation frequency, f_f	0.1 Hz
Harmonics, M	3
Amplitude, a_l	75 μV
Amplitude variation, Δa_l	25 μV
Amplitude variation frequency, ω_a	0.08 Hz

The resulting simulated AF signal is shown in Fig.1. The PSD and the CS of this signal (calculated in the range $[3, 12]$ Hz) are shown in Fig. 2. As we can notice, the fundamental peak at 6 Hz is rised up by the CS (dashed line) in respect to its harmonics and to its equivalent in the PSD plot (thick line).

2.3. Compressed spectrum and AF dominant frequency assessment

To exploit the concept of CS for the evaluation of the AFDF directly from a single-lead ECG recording (named

$y(t)$), a suitable pre-processing has to be done, as summarized in the following steps.

• An amplitude threshold is applied on the ECG signal to cut QRS complexes'peaks, in order to reduce the energy contribution of these components in the spectrum. The threshold is fixed to the value of $\xi = \pm 2\mu_{|y|}$, where $\mu_{|y|}$ is the mean value of the rectified signal $\mid y(t) \mid$.

• A zero-phase band-pass filter with a -3 dB high-pass cut off frequency at 3 Hz and -3 dB low-pass cut off frequency at 60 Hz is applied, taking into account the $[3, 12]$ Hz AF frequency range of interest, and their first harmonics.

• The power spectral density (PSD) of the signal is measured using a Welch's method, with a 8192 points FFT, 4096 sample size Hamming window and 50% overlapping.

• The CS is computed using eq. (1), where $P_k(\omega_0)$ is the PSD of signal $y(t)$. Taking into account the $[3, 12]$ Hz AF frequency range of interest where AFDF is supposed to be located, eq. (1) is evaluated only for the frequencies belonging to this range. CS is calculated till the second harmonic ($n(w_0) = 3$, see [9]).

Fig. 3 shows an example of the CS of an ECG during AF, obtained directly from V1-lead analysis, following the aforementioned steps ($\mathbf{CS_{\tilde{y}}}$, dashed-dotted line), compared to V1-lead PSD ($\mathbf{PSD_y}$, continuous thin line), V1-lead PSD after QRS complex thresholding ($\mathbf{PSD_{\tilde{y}}}$, dashed line), CS of V1-lead without QRS thresholding ($\mathbf{CS_y}$, dotted line) and to the PSD of the AA signal estimate obtained from the same recording as reference ($\mathbf{PSD_{Ref}}$, continuous thick line). It is clear how the AFDF peak is risen by CS, and the usefulness of the QRS thresholding before CS computation, even if thresholding is not able by itself to produce a good estimate of AFDF (see also the table of the results in section 3.2).

3. Results

3.1. Reference AA signal for reference AFDF computation

Like in [8], to assess the performance of the proposed method on real ECG data, a reference value for the AFDF was needed. To this goal, the spatio-temporal QRST-cancellation method proposed by Meste *et al.* [5] was chosen to obtain a reference AA signal. The PSD of this signal was computed using the Welch's method (window lenght 8192 points, with 50% overlapping, as described in Section 2.3). The AFDF was searched in the range $[3, 12]$ Hz. For each ECG recording, the reference AFDF was estimated from the remainder V1-lead, this lead being considered as the most suitable to capture information on AF.

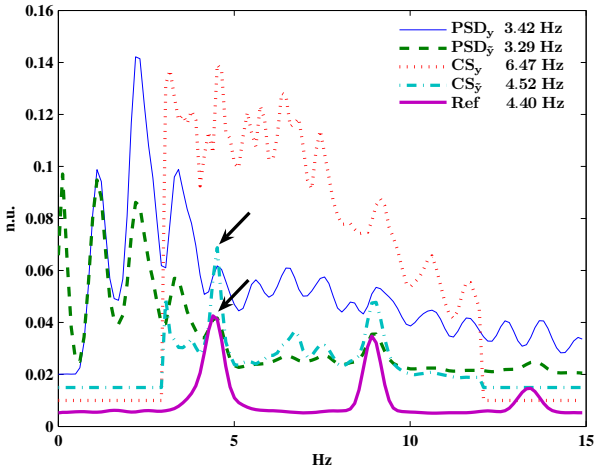

Figure 3. Example of compressed spectrum ($\mathbf{CS_{\tilde{y}}}$, dashed-dotted line) obtained directly from the analysis of V1-lead compared to other methods as described in section 2.3. AFDF peak is highlighted by the arrows for $\mathbf{CS_{\tilde{y}}}$ and the PSD of the reference signal ($\mathbf{PSD_{Ref}}$, continuous thick line). An offset has been applied to all the plots in the figure to make it more clear; n.u., normalized units.

3.2. Performance evaluation

The proposed method was applied to a dataset of 22 ECG recordings, all presenting AF episodes. Its performance is evaluated taking the normilized mean square error (NMSE) between the estimated AFDFs and the reference values computed as described in Section 3.1. Results are given in terms of mean estimated AFDF ($\mu_{\mathbf{AFDF}}$), mean absolute difference (MAD) and NMSE, for all the methods introduced in Fig.3, as summarized in the following table (ns: no sense)

	$\mu_{\mathbf{AFDF}}$ (Hz)	MAD (Hz)	NMSE (%)
$\mathbf{PSD_y}$	5.67 ± 1.65	0.91 ± 1.20	5.84
$\mathbf{PSD_{\tilde{y}}}$	5.62 ± 1.15	0.58 ± 1.05	3.71
$\mathbf{CS_y}$	6.35 ± 0.95	0.47 ± 0.63	1.58
$\mathbf{CS_{\tilde{y}}}$	6.20 ± 1.02	0.24 ± 0.50	0.78
$\mathbf{PSD_{Ref}}$	6.01 ± 1.10	ns	ns

The discrepancy with the reference method was generally ≤ 0.2 Hz. Fig. 4 shows the box-and-whisker plot for all the methods presented in the table.

4. Discussion and conclusions

This work proposes a new method for the assessment of the AFDF directly from a single-lead ECG recording during AF, pointing out the possibility to estimate AFDF avoiding any VA cancellation. The method exploits the

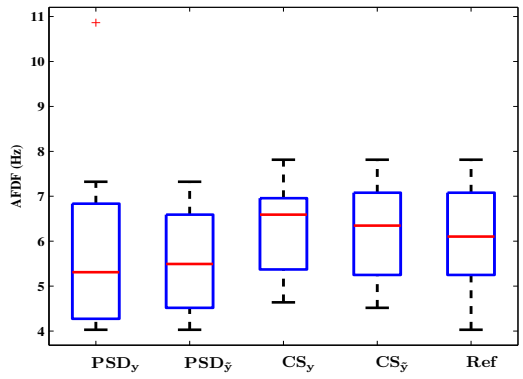

Figure 4. Box-and-whisker plot of the AFDFs estimated directly from V1-lead using CS ($\mathbf{CS}_{\tilde{\mathbf{y}}}$), from the reference signals estimated using a bayesian spatio-temporale cancellation method ($\mathbf{Ref}$), and from the other presented methods, for completeness.

concept of CS and makes the fundamental frequency of a signal standing out in its spectrum taking into account the harmonic contribution to the fundamental estimate.

The results showed that when CS is applied to a single-lead ECG recording during AF, after a suitable pre-processing, it can provide reliable estimates of the AFDF. This is attested by the small NMSE values found between the estimated AFDF and the reference one, confirming a negligible difference between the two sets. This important result underlines very interesting aspects. The most important one is that AFDF estimate can be performed directly on a single-lead basis, without needing neither a QRST cancellation nor a QT interval detection and removal, making the algorithm very cost efficient. Together with the simplicity of the technique, this makes the algorithm suitable for real-time applications.

Future work requires to test the method on high heart rate ECGs, not included in the present dataset. This is not expected to be a limitation in the proposed method, since it does not rely on TQ interval, and it should not be affected by the problem of vanishing TQ interval at high heart rates. Moreover, performance of the method should also be tested for ECG signals sampled at low frequency.

Acknowledgements

The authors would like to express their gratitude to Leif Sörnmo, Francisco Castells and to the Hôpital Pasteur de Nice for providing the real data. The work of Pietro Bonizzi is supported by the EU by a Marie-Curie Fellowship (EST-SIGNAL program: http://est-signal.i3s.unice.fr) under contract No MEST-CT-2005-021175.

References

[1] Bollmann A, Husser D, Mainardi L, Lombardi F, Langley P, Murray A, Rieta JJ, Millet J, Bertil Olsson S, Stridh M, Sornmo L. Analysis of surface electrocardiograms in atrial fibrillation: techniques, research, and clinical applications. Europace 2006;8:911–926.

[2] Petrutiul, S. and Sahakian, V. and Ng, J. and Swiryn, S. Analysis of the surface electrocardiogram to predict termination of atrial fibrillation: The 2004 computers in cardiology/physionet challenge. Proc Computers in Cardiology 2004;31:105–108.

[3] Rieta JJ, Castells F, Sánchez C, Zarzoso V, Millet J. Atrial activity extraction for atrial fibrillation analysis using blind source separation. IEEE Trans on Biomed Eng July 2004; 51, No. 7:1176–86.

[4] Stridh M, Sörnmo L. Spatiotemporal QRST cancellation techniques for analysis of atrial fibrillation. IEEE Trans Biomed Eng January 2001;48:105–111.

[5] Meste O, Serfaty N. QRST cancellation using bayesian estimation for the auricular fibrillation analysis. In Engineering in Medicine and Biology. 2005; .

[6] Castells F, Rieta JJ, Millet J, Zarzoso V. Spatiotemporal blind source separation approach to atrial activity estimation in atrial tachyarrhythmias. IEEE Transactions on Biomedical Engineering February 2005;52(2):258–267.

[7] Rosenbaum, D.S. and Cohen, R.J. Frequency bases measures of atrial fibrillation in man. In Engineering in Medicine and Biology Society Proceedings of the Twelfth Annual International Conference of the IEEE 1990;582–583.

[8] Sassi, R. and Corino, V.D.A. and Mainardi, L.T. Analysis of Surface Atrial Signals Using Spectral Methods for Time Series with Missing Data. In Computers in Cardiology 2007; 34:153–156.

[9] Stridh, M. and Sörnmo, L. and Meurling, C. J. and Olsson, S.B. Sequential Characterization of Atrial Tachyarrhythmias Based on ECG Time-Frequency Analysis. IEEE Trans on Biomed Eng 2004;51:100–114.

[10] Sörnmo L, Laguna P. Bioelectrical Signal Processing in Cardiac and Neurological Applications. Elsevier Academic Press, 2005.

[11] Mitra SK. Digital Signal Processing: A computer-based approach, *2nd ed.* McGraw-Hill, 2001.

[12] Khaddoumi, B. Analyse et modelisation d'electrocardiogrammes dans le cas de pathologies ventriculaires. Ph.D. thesis, Universit de Nice-Sophia Antipolis, 2005.

Address for correspondence:

Pietro Bonizzi, Ph.D. Student
Laboratoire I3S - BIOMED UNSA - CNRS
2000, Route des Lucioles Les Algorithmes - bât. Euclide B
B.P. 121
06903 Sophia Antipolis - Cedex, France
E-mail: bonizzi@i3s.unice.fr