Time-frequency Analysis of Atrial Fibrillation Comparing Morphology-clustering Based QRS-T Cancellation with Blind Source Separation in Multi-lead Surface ECG Recordings

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Abstract

To separate the atrial (AA) from the ventricular (VA) electrical activity in surface ECG recordings of atrial fibrillation (AF), various methods have been proposed, such as QRS-T cancellation by beat-averaged template subtraction, and blind source separation (BSS). Although QRS-T cancellation is computationally more efficient than BSS, and allows the preservation of spatial information, it is sensitive to morphology changes, which produce large residuals in AA, biasing the frequency analysis.

Aim of this study was: (i) to propose an improved approach to VA cancellation based on k-means morphology clustering (MC); (ii) to validate its ability to estimate AF dominant frequency (DF) on a standard database with intra-cardiac and surface ECG recordings (IAFDB, Physionet.org); (iii) to compare the temporal evolution of the spectral content of MC-estimated AA (MC-AA) with the one obtained from a reference BSS method based on Independent component analysis (ICA) and second-order blind identification (SOBI), in 14 body surface potential map (BSPM) recordings.

QRS-T amplitude in MC-AA was significantly lower ($p<0.001$) than in ECG (in closest BSPM channel to V1).

The validation on IAFDB showed no significant difference in DF estimation ($p=0.546$) in 17 recordings. Also no significant difference in DF estimation ($p=0.208$) with respect to the reference BSS method was observed.

The proposed QRS-T cancellation method effectively suppresses VA and accurately estimates DF compared to an established BSS method.

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1. Introduction

Atrial fibrillation (AF) is a common cardiac arrhythmia affecting between 2% and 10% of people over 50 years of age [1]. AF is a major cause of morbidity and mortality in the elderly population where the risk of stroke is five times higher [2]. For this reason an increasing clinical research interest has been devoted to AF in recent years [3-4].

Methods reported in literature to cancel VA from the ECG involve direct suppression of the QRS complex and the T-wave by subtracting a fixed template obtained by averaging consecutive beats (BA). It relies on the fact that AF is uncoupled to the ventricular activity, thus subtracting an averaged QRS-T segment from the ECG produces a residual signal which closely represents AF [2,4-5]. Unfortunately, since average beat subtraction is performed on individual leads, this process is sensitive to alterations in the electrical axis, which produces large QRS-related residuals in AA [4] which in turn affect its spectral content. Another approach to AA extraction is based on the hypothesis that AA and VA originate from statistically separable sources [2, 4], which may be treated by blind source separation (BSS) methods, when multi-lead surface ECG recordings are available. Independent component analysis (ICA) has successfully been adopted [3-4] and improved with second order blind identification (SOBI) [6], as well as principal component analysis (PCA) [2].

A recording length of at least 10 s is required for adequate computation of the average beat in QRS-T cancellation, whereas the recording length can be shorter in BSS [4]. However, BSS only allows the derivation of a global atrial signal with contributions from all leads, which limits its spatial resolution [4]. On the other hand, QRS-T cancellation is computationally more efficient than BSS, and allows the preservation of spatial information from individual leads.

The aim of this study is threefold: (*i*) to propose an improved approach to VA cancellation based on a *k-means* morphology clustering criterion; (*ii*) to validate its ability to estimate DF on a standard database with simultaneous intra-cardiac and surface ECG recordings (*Intracardiac Atrial Fibrillation Database*, Physionet.org); (*iii*) to compare the temporal evolution of the spectral content of MC estimated AA (MC-AA) with

the one obtained from a reference BSS method based on ICA-SOBI (ICA-AA), in 14 body surface potential map (BSPM) recordings.

2. Methods

2.1. Data acquisition

Fourteen recordings of 67-lead surface ECG (64-thoracic, three bipolar limb leads) from fourteen male patients with AF (age 60±9 years) were considered for this study, each 3 minutes in duration.

The surface ECG data were recorded by Biosemi ActiveTwo™ (Biosemi, Amsterdam, NL) at a sampling rate of 2048 samples/s, 24-bit/sample.

2.2. ECG processing

The ECG signal was processed offline in MATLAB® (The Mathworks, Natick, Massachusetts, USA).

The ECG was band-pass filtered (3-dB pass-band: 0.5-100 Hz) and then down-sampled to 250 samples/s for beat detection. An established discrete-wavelet transform based method [7] was used for beat detection.

2.3. QRS-T clustering

For the *i*th detected beat, a QRS-T window was defined as:

$$\begin{aligned} T_1(i) &= QRS_{PK}(i) - \Delta_{80} \\ QTc_{REF} &= \Delta_{440} \\ T_2(i) &= T_1(i) + QTc_{REF} \cdot \sqrt{RR_{MEAN}} \\ W_{QRST}(i) &= \{T_1(i), T_2(i)\} \end{aligned} \quad (1)$$

where $QRS_{PK}(i)$ is the fiducial point (i.e. dominant peak) of the *i*th beat, Δ_{80} is a time delta of 80 ms, QTc_{REF} is an empirical reference value of QTc (Bazett) set to 440 ms, and RR_{MEAN} is the mean RR interval duration (in units of seconds) of the entire recording.

A QRS-T collection matrix *B* can be constructed, whose columns are the ECG samples of $W_{QRST}(i)$, i=1,...,M, where *M* is the number of detected beats.

Rows of *B* are treated as *N* observations of an *M*-dimensional variable. *k-means* clustering is performed using correlation distance: $d_{CORR}(x_n, x_m) = 1 - \rho(x_n, x_m)$ as dissimilarity criterion. Collected beats are grouped according to the dissimilarity metrics to form the columns of the starting guess matrix to initialize the *k-means* clustering process.

2.4. Blind source separation

An extensive description of the ICA-SOBI approach to BSS of AA is given in [6].

In the present study a subset of 14 channels was chosen from the anterior BSPM map as sensor variables for ICA-SOBI, including the top-central portion of the right-anterior torso, as shown in Fig. 1. This choice includes CH15 (closest to V1 in 12-lead ECG) where the atrial activity is prominent. ICA (JADE implementation) and subsequent SOBI were then performed, as suggested in [6]. Spectral concentration (SC) was adopted to select the source representing AA (source with highest SC):

$$\begin{aligned} SC(S_k) &= \frac{\sum_{f=0.82\cdot f_P}^{1.17\cdot f_P} \Gamma_{S_k}(f_i)}{\sum_{f=0}^{F_S/2} \Gamma_{S_k}(f_i)} \\ f_P &= \arg \max_{f \in (3.5-10Hz)} (\Gamma(f_i)) \end{aligned} \quad (6)$$

where S_k is the *k*th estimated source, Γ_{Sk} is the power spectral density (PSD) of the *k*th source, f_P is the peak frequency in the frequency band of AF: 3.5 – 10 Hz, namely DF.

2.5. Time-frequency analysis of AA

The AA signal (MC-AA and ICA-AA) was divided into consecutive non-overlapping 20 s windows and PSD (Welch periodogram, Hamming window, 50% overlap, frequency resolution: 0.24 Hz) was computed for each.

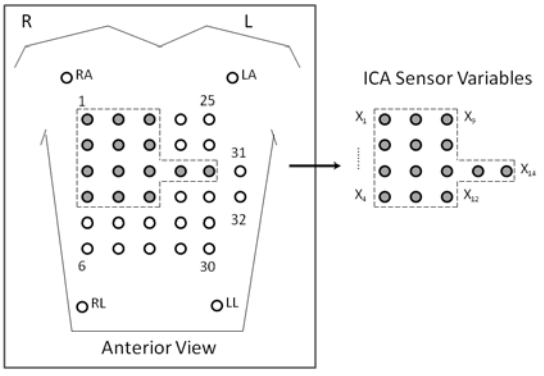

Figure 1. 32-channel anterior view of BSPM. Dotted region indicates chest electrodes used for ICA.

2.6. Validation on standard database

The *Intracardiac Atrial Fibrillation Database* (IAFDB) from Physionet (www.physionet.org) was adopted for validation of MC estimation of DF.

Recordings whose intracardiac signals would allow reliable peak-detection were considered (n=17). Each recording was divided into consecutive non-overlapping 20 s windows. For each, MC-AA was extracted from lead V1 and DF estimated as the dominant peak frequency of PSD. The actual atrial fibrillatory rate (AAFR) was calculated as the inverse of the mean interval between consecutive peaks within the 20 s window.

3. Results

3.1. VA cancellation

Figure 2 shows MC compared to BA in VA cancellation.

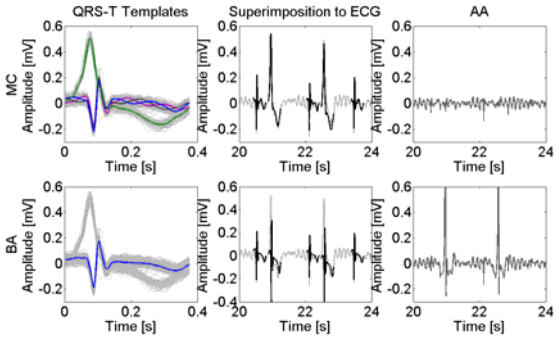

Figure 2. MC (top panels) compared to BA in VA cancellation

Table 1 shows comparison between MC and BA in QRS-T cancellation (Median(inter-quartile range) is reported). In all cases for which multiple templates were generated (rows with #Cluster>1) there was a significant improvement (Wilcoxon rank-sum test (WRT), α=0.05) in QRS-T cancellation in MC with respect to BA.

Table 1. VA cancellation: comparison of MC and BA.

		AA QRS-T Amplitude [μV]		
#Rec	#Cluster	MC	BA	*p value*
r1	17	224(98)	307(113)	*p<0.001*
r2	1	250(185)	244(180)	*N.S.*
r3	4	617(432)	708(120)	*p<0.001*
r4	1	192(86)	195(84)	*N.S.*
r5	1	274(183)	276(190)	*N.S.*
r6	5	141(123)	193(1167)	*p<0.001*
r7	11	424(492)	424(489)	*N.S.*
r8	11	161(60)	196(70)	*p<0.001*
r9	11	280(115)	349(160)	*p<0.001*
r10	1	224(199)	226(194)	*N.S.*
r11	1	326(265)	327(266)	*N.S.*
r12	1	262(226)	267(233)	*N.S.*
r13	3	302(111)	335(102)	*p<0.001*
r14	5	543(333)	610(507)	*p<0.01*

N.S. = not significant

Table 2 shows cumulative data analysis of QRS-T amplitude in MC-AA with respect to ECG in CH15. MC significantly reduces QRS-T amplitude with respect to ECG, though the ECG amplitude in the TQ segment remains significantly higher than that of MC-AA in the QRS-T, indicating that MC does not "over-cancel".

Table 2. QRS-T cancellation: cumulative data analysis

Amplitude [μV]				
QRS-T MC	QRS-T ECG	*p value*	TQ ECG	*p value*
260(223)	1239(1096)	*p<0.001*	194(188)	*p<0.001*

3.2. DF estimation

DF estimation difference (Δ_{DF}) is shown in Table 3 (Median(inter-quartile range) is reported). Statistical analysis (WRT, α=0.05) is presented for the two datasets: standard database (IAFDB) and experimental data (14 BSPM recordings). To compare DF estimation to ICA, MC-AA from the lead maximizing SC was selected. In both datasets the median value of Δ_{DF} is below the frequency resolution (0.24 Hz) used in PSD computation.

Table 3. DF Estimation. Median(inter-quartile range)

	Δ_{DF} [Hz]	
	MC vs. AAFR	*p value*
IAFDB	0.03(0.37)	*N.S.(p=0.546)*
	MC vs. ICA	
14 Recordings	0.00(0.49)	*N.S.(p=0.208)*

N.S. = not significant

3.3. Time-frequency analysis

Figure 3 shows an example of PSD evolution for recordings r1 through r4.

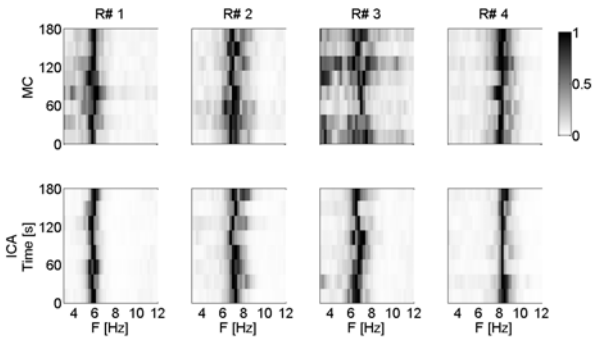

Figure 3. Normalized PSD temporal evolution of MC-AA (top panels) and ICA-AA.

Figure 4 shows an example of the spatio-temporal evolution of PSD for recording r. SC is higher (DF most

prominent) in the right side of the torso.

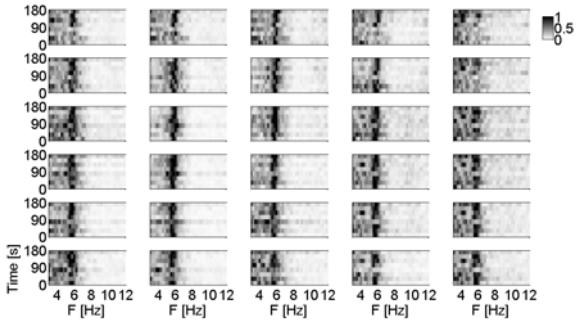

Figure 4. Spatio-temporal evolution of normalized PSD for recording r1, for the first 30 BSPM channels (following the order displayed in Figure 1).

Table 4 shows statistical analysis (WRT, α=0.05) of SC for the two methods (Median(inter-quartile range) reported). SC in ICA is significantly different (higher) from MC in 7 of 14 recordings, which according to [6] may indicate higher performance (reliability) in the extraction of AA.

Table 4. SC: comparison of MC and ICA.

	SC [%]		
#Rec	MC	ICA	*p value*
1	57.5(6.4)	80.9(4.7)	*p<0.01*
2	58.8(3.9)	70.5(8.0)	*p<0.01*
3	61.7(9.0)	67.8(11.2)	*N.S.*
4	30.4(8.3)	61.6(5.9)	*p<0.005*
5	63.7(6.3)	60.7(16.4)	*N.S.*
6	53.2(7.9)	60.5(6.9)	*N.S.*
7	40.1(4.5)	56.0(23.7)	*N.S.*
8	48.7(9.0)	55.3(9.3)	*N.S.*
9	31.2(9.4)	51.9(6.2)	*p<0.01*
10	34.3(4.4)	43.5(6.2)	*p<0.01*
11	29.6(9.9)	38.0(6.6)	*p<0.01*
12	30.2(9.1)	33.0(3.6)	*N.S.*
13	20.3(3.3)	32.0(4.6)	*p<0.01*
14	28.4(4.3)	29.3(6.2)	*N.S.*
Average	39.5(27.2)	54.4(26.6)	*p<0.001*

4. Discussion and conclusions

A multi-template clustering based approach to VA cancellation (MC) was presented. The proposed method showed significant improvement (p<0.001) in VA cancellation with respect to "beat-averaging" approach in BSPM recordings for which multiple templates were generated (no significant difference in the others).

The validation on a standard database (IAFDB, Physionet) showed no significant difference in DF estimation (p=0.546) in 17 recordings.

The proposed method also showed no significant difference in DF estimation with respect to ICA in BSPM recordings (p=0.208) in spite of significantly lower SC (p<0.001). However, replication on a larger dataset (including multifocal ectopies) is required to confirm these results.

VA cancellation by MC allows spatial localization of time-frequency distribution of DF, as shown in Figure 4 where frontal-right channels exhibit a marked DF pattern. This degree of spatial resolution may only be achieved by single-lead source cancellation methods, which on the other hand suffer from VA residuals, as shown by significantly lower SC with respect to BSS methods.

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