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Adaptive singular value cancelation of ventricular activity in single-lead atrial fibrillation electrocardiograms

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Abstract

The proper analysis and characterization of atrial fibrillation (AF) from surface electrocardiographic (ECG) recordings requires to cancel out the ventricular activity (VA), which is composed of the QRS complex and the T wave. Historically, for single-lead ECGs, the averaged beat subtraction (ABS) has been the most widely used technique. However, this method is very sensitive to QRST wave variations and, moreover, high-quality cancelation templates may be difficult to obtain when only short length and single-lead recordings are available. In order to overcome these limitations, a new QRST cancelation method based on adaptive singular value cancelation (ASVC) applied to each single beat is proposed. In addition, an exhaustive study about the optimal set of complexes for better cancelation of every beat is also presented for the first time. The whole study has been carried out with both simulated and real AF signals. For simulated AF, the cancelation performance was evaluated making use of a cross-correlation index and the normalized mean square error (nmse) between the estimated and the original atrial activity (AA). For real AF signals, two additional new parameters were proposed. First, the ventricular residue (VR) index estimated the presence of ventricular activity in the extracted AA. Second, the similarity (S) evaluated how the algorithm preserved the AA segments out of the QRST interval. Results indicated that for simulated AF signals, mean correlation, nmse, VR and S values were 0.945 ± 0.024, 0.332 ± 0.073, 1.552 ± 0.386 and 0.986 ± 0.012, respectively, for the ASVC method and 0.866 ± 0.042, 0.424 ± 0.120, 2.161 ± 0.564 and 0.922 ± 0.051 for ABS. In the case of real signals, the mean VR and S values were 1.725 ± 0.826 and 0.983 ± 0.038, respectively, for ASVC and 3.159 ± 1.097 and 0.951 ± 0.049 for ABS. Thus, ASVC provides a more accurate beat-to-beat ventricular QRST representation than traditional techniques. As a consequence, VA cancelation

is optimized and the AA can be extracted more precisely. Finally, the study has proven that optimal VA cancelation is achieved when a number between 20 and 30 complexes is selected following a correlation-based strategy.

Keywords: averaged beat subtraction, electrocardiogram, optimal beats selection, QRST cancelation, singular value decomposition

1. Introduction

The surface electrocardiographic (ECG) recording provides a widely used and noninvasive way to study atrial fibrillation (AF). This arrhythmia is most commonly diagnosed in clinical practice and affects up to 1% of the general population and up to 15% of the population over 80 years (Kannel *et al* 1998). Some advantages of using the surface ECG include the ability to record data for a long period of time, the minimal costs and risks involved for the patient, in comparison with invasive electrophysiological procedures, and the possibility of observing a global activity in the atria and ventricles during AF (Petrutiu *et al* 2006).

The extraction of atrial activity (AA) under the best conditions is crucial in the study of the electrophysiological processes that underlie AF, such as refractory periods, autonomic response, drug effects, etc (Bollmann *et al* 2006). However, due to the reduced amplitude of AA, the cancelation of the signal components associated with ventricular activity (VA), that is, the QRS complex and the T wave, is mandatory (Slocum *et al* 1992). Unfortunately, AA and VA possess overlapped spectral distributions, thus rendering linear filtering solutions unsuccessful (Rieta *et al* 2004).

To date, several methods have been proposed to extract the AA signal from surface ECG recordings. The most powerful techniques are those that exploit the spatial diversity of the multilead ECG (Langley *et al* 2006), such as the method that solves the blind source separation problem (Rieta *et al* 2004) or the spatiotemporal QRST cancelation strategy (Stridh and Sörnmo 2001). However, the performance of these techniques is seriously reduced when the early stages of AF, i.e. paroxysmal AF, have to be analyzed. The reason is that this type of recordings is usually obtained from Holter systems, with no more than two or three available leads, as in this work, which are not sufficient to exploit the ECG spatial information.

For single-lead applications, the most widely used alternative is the averaged beat subtraction (ABS). This method relies on the assumption that the average beat can represent approximately each individual beat. However, QRST morphology is often subject to minor changes caused by respiration, patient movement, etc, and, therefore, QRST residua and noise are often present in the estimated AA or remainder ECG (Petrutiu *et al* 2006). Additionally, in clinical practice the ECG consists of recordings that are 10 s long and, therefore, a high-quality QRST cancelation template may be difficult to obtain (Lemay *et al* 2007). Thereby, this work presents a new cancelation method based on adaptive singular value cancelation (ASVC) of each single beat. The method is able to cancel out the QRST complexes in single-lead AF ECG recordings. The proposed ASVC methodology extracts the basis signal corresponding to the VA component by exploiting the mutual information available in the set of ECG beats under processing.

On the other hand, although the ABS technique has been studied in many previous works and different and complex variants have been proposed (Stridh and Sörnmo 2001, Lemay *et al* 2007), there are still some unsolved aspects within this context. The optimal number

of complexes to generate the cancelation template is one of the pending tasks. Moreover, the decision about which is the concrete subset of complexes, from the available set in the recording under analysis, that will provide optimal cancelation is the second challenging problem of great interest in this context. Thereby, in this work a thorough study about the VA cancelation accuracy, when different complexes are selected to obtain the ventricular cancelation template, is presented for the first time to the scientific community. As a consequence, some recommendations regarding the optimal number of complexes and their localization into the ECG recording will be proposed.

The paper is structured as follows. Section 2 describes the database used, composed of simulated and real AF signals, and the previous preprocessing applied to the analyzed ECG recordings. In section 3, the proposed ASVC method to extract the AA signal, the analyzed parameters to compare ASVC and ABS performances, and the designed analysis to select the most suitable complexes are presented. Section 4 summarizes the results obtained, which are next discussed in section 5. Finally, section 6 presents the concluding remarks.

2. Materials

2.1. Database

The presented methodology was validated using a database composed of 50 1-min simulated AF signals and 50 1-min real ECG recordings. Simulated recordings allowed us to compare the estimated with the original AA, as was known *a priori*. Synthesized AF signals were created from the combination of real AA and VA, which were separately extracted since AA is uncoupled to VA during AF (Slocum *et al* 1985, Stridh and Sörnmo 2001). The AA was obtained from the smooth concatenation of successive TQ segments extracted from real AF recordings. The VA was extracted from normal sinus rhythm ECG recordings, after P-wave cancelation (Castells *et al* 2005). In addition, with the aim of creating simulated AF signals as similar to real AF recordings as possible, some variations in the signal were introduced. The QRST complex amplitude and width and the RR intervals were randomly varied following the specific characteristics of real AF recordings. More concretely, the QRST complex amplitude was randomly reduced or enlarged between 0% and 20% of its original size. On the other hand, the QRST width was randomly varied between 340 and 420 ms making use of different upsampling and downsampling factors. Finally, the TQ intervals were randomly reduced or enlarged in order to obtain 40% of variability in the RR intervals with respect to their mean value. To reduce a TQ interval, consecutive samples were removed. In contrast, in order to expand the TQ interval, a linear combination of the three preceding TQ intervals was introduced in the middle of the considered interval. An example of a synthesized AF signal following this procedure will be shown in figure 4(a) when results are presented.

In addition, to evaluate the suitability of the algorithm to be applied over real scenarios which, of course, was the final purpose of this research, real AF recordings were also used. The signals were obtained from Holter systems of two leads (V_1 and V_{II}), digitized at a sampling frequency of 128 Hz and a resolution of 12 bits. Lead V_1 was the input signal to the QRST cancelation approach, as it is widely considered as the signal with a larger AA content (Petrutiu *et al* 2006).

2.2. Data preprocessing

The recordings were preprocessed in order to reduce noise, nuisance interference and improve later analysis. First, baseline wander was removed by making use of bidirectional high-pass

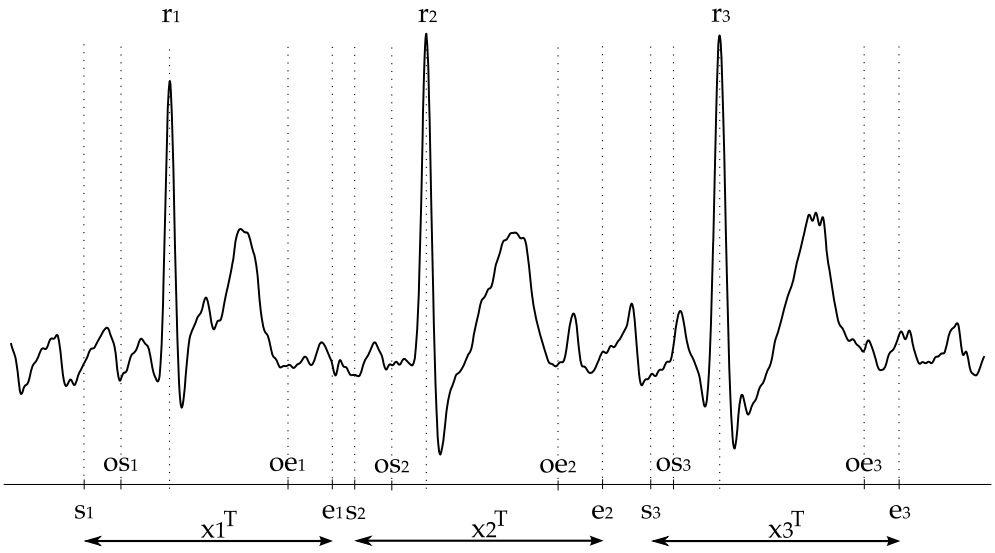

Figure 1. Relevant time instants used by the proposed ASVC algorithm. The points s_i and e_i are the start and end points of the ith QRST complex which is represented by $\mathbf{x}_i$, respectively. The points os_i and oe_i define the zones, at the beginning and the end of the ith QRST complex, respectively, that will be processed to avoid sudden transitions after VA cancelation.

filtering with 0.5 Hz cut-off frequency (Boucheham *et al* 2005, Dotsinsky and Stoyanov 2004). Second, high frequency noise was reduced with an eight-order bidirectional IIR Chebyshev low-pass filtering, whose cut-off frequency was 70 Hz (Sun *et al* 2002, Hamilton *et al* 2000). Finally, powerline interference was removed through adaptive filtering, which preserves the ECG spectral information (Ferdjallah and Barr 1994). In addition, all the signals were upsampled to 1024 Hz in order to improve time alignment accuracy for QRST complex subtraction (Bollmann *et al* 2006), since the variability in the R peak detection was reduced by an average of 40% (Laguna and Sornmo 2000). A cubic splines interpolation method was used because it provided the best resolution (lower than 1 ms) in the R peak detection in comparison with other methods analyzed for the same purpose (Daskalov and Christov 1997).

3. Methods

3.1. Adaptive singular value cancelation of the ventricular activity

The ECG signal presents a high degree of temporal redundancy which could be exploited for the cancellation of VA. Indeed, the QRST waveform usually exhibits a recurrent pattern, although different QRST morphologies as well as minor variations in the QRST waveform may occur (Castells *et al* 2005). In order to take advantage of the aforementioned ergodicity, all the R waves were first detected by making use of the technique of Pan and Tompkins (1985). Next, the ith QRST complex start point was defined as $s_i = r_i - 0.3 \cdot RR_{\min}$, r_i being the R peak wave event and $RR_{\min}$ being the minimum RR interval found in the ECG under analysis. The ith QRST complex end point was selected as $e_i = r_i + 0.7 \cdot RR_{\min}$ (Shkurovich *et al* 1998); see figure 1.

For a compact notation, each QRST complex was assumed to be represented by a column vector of the matrix $\mathbf{X} \in \Re^{L\times N}$:

$$\mathbf{X} = [\mathbf{x}_1, \mathbf{x}_2, \ldots, \mathbf{x}_N], \tag{1}$$

where $\mathbf{x}_i$ contains L samples of the ith complex and N is the number of complexes in the analyzed ECG. Note that all the beats were temporally aligned using its R peak timing. In this case, the singular value decomposition of $\mathbf{X}$ can be expressed as (Golub and van Loan 1989)

$$\mathbf{X} = \mathbf{USV}^{\mathsf{T}}, \tag{2}$$

where $\mathbf{U} \in \Re^{L\times N}$ is a unitary matrix so that $\mathbf{UU}^{\mathsf{T}} = \mathbf{I}$, $\mathbf{I}$ being the identity matrix, $\mathbf{S} \in \Re^{N\times N}$ is a diagonal matrix, and $\mathbf{V} \in \Re^{N\times N}$ fulfills $\mathbf{VV}^{\mathsf{T}} = \mathbf{I}$. The matrix $\mathbf{U}$ contains the N normalized principal components of $\mathbf{X}$, so that the columns of $\mathbf{U} = [\mathbf{u}_1, \ldots, \mathbf{u}_N]$ are the eigenvectors of $\mathbf{X}$, and their cross-correlation coefficients are null. The matrix $\mathbf{S}$ contains the amplitude coefficients corresponding to the N principal components of $\mathbf{X}$. These coefficients are called eigenvalues or singular values and are sorted in a descending order. Thus, the N non-normalized principal components can be obtained as the columns of the matrix $\mathbf{P} = \mathbf{US}$, and can be interpreted as follows.

(i) The first and most significant component is related to the main QRST waveform.
(ii) Subsequently, there are several components related to AA.
(iii) The remaining components correspond to noise in the ECG.

This interpretation is possible because the AA signal (x_{AA}) can be viewed as being uncoupled to VA and noise (Stridh and Sörnmo 2001, Rieta *et al* 2004); thus, the ith observed beat can be modeled as a linear combination of AA ($\mathbf{x}_{i_{AA}}$), VA ($\mathbf{x}_{i_{VA}}$) and noise ($\mathbf{x}_{i_n}$) (Stridh and Sörnmo 2001):

$$\mathbf{x}_i = \mathbf{x}_{i_{\mathrm{AA}}} + \mathbf{x}_{i_{\mathrm{VA}}} + \mathbf{x}_{i_{\mathrm{n}}}. \tag{3}$$

Hence, the first, i.e. the highest variance, principal component of $\mathbf{X}$, which will be called $\mathbf{t}$, can be used as a QRST template to cancel out VA because it contains the mother representation for the QRST complexes in the ECG under analysis. However, since considerable R peak amplitude differences between each individual QRST complex and the template can be found, the $\mathbf{t}$ amplitude was individually adapted to the ith beat as follows:

$$\mathbf{t}_i = \frac{QR_i}{QR_{\mathbf{t}}} \cdot \mathbf{t}, \tag{4}$$

where QR_i and $QR_{\mathbf{t}}$ are the amplitude distances between the Q and R points of the ith complex and the template, respectively.

As a consequence, the AA estimation ($\widehat{\mathbf{X}}_{\mathrm{AA}}$) which is contained in every QRST interval can be obtained as

$$\widehat{\mathbf{X}}_{\mathrm{AA}} = \mathbf{X} - \mathbf{T}, \tag{5}$$

$\mathbf{T}$ being the matrix constituted by the column vectors $\mathbf{t}_i$:

$$\mathbf{T} = [\mathbf{t}_1, \mathbf{t}_2, \ldots, \mathbf{t}_N]. \tag{6}$$

Note that the AA obtained will also contain the noise present during the recording process ($\mathbf{x}_{i_n}$). Anyway, this interferent signal is of notably lower amplitude than the AA during a normal recording. Moreover, a relevant part of it can be removed through the preprocessing methods described in section 2.2.

In the AA signal reconstructed from $\widehat{\mathbf{X}}_{\mathrm{AA}}$ ($\widehat{x}_{\mathrm{AA}}$), sudden transitions may occur at the beginning or the end of each QRST segment; see figure 2(b). They are provoked by the subtraction between each individual complex and the ventricular cancelation template. In

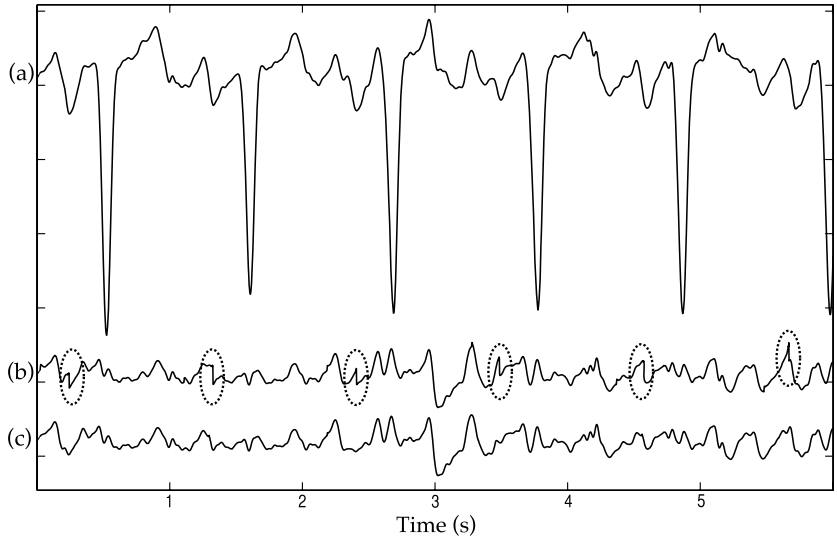

Figure 2. Example of an artificial recording to illustrate the sudden transitions problem. (a) Synthesized ECG ready for ventricular cancelation. (b) AA obtained without the treatment of sudden transitions ($\widehat{\mathbf{X}}_{\mathrm{AA}} = \mathbf{X} - \mathbf{T}$). (c) AA obtained after the treatment of transitions.

order to avoid these transitions and obtain a more accurate AA, two steps were developed. First, an interval of P samples after the start point and before the end point of each QRST complex was defined and the points os_i and oe_i were marked as (see figure 1)

$$os_i = s_i + n_{s_i} \tag{7}$$

$$oe_i = s_i + n_{e_i}, \tag{8}$$

n_{s_i} and n_{e_i} being the temporal instants, along the two defined intervals, in which the absolute minimum differences between the ith QRST complex ($\mathbf{x}_i$) and the $\mathbf{t}_i$ particular template are obtained. By considering these instants, the ventricular template adapted to each beat was modified as follows:

$$\tilde{\mathbf{t}}_i = [\underbrace{0, \ldots, 0}_{n_{s_i}-1}, \mathbf{t}_i(n_{s_i}), \mathbf{t}_i(n_{s_i}+1), \ldots, \mathbf{t}_i(n_{e_i}-1), \mathbf{t}_i(n_{e_i}), \underbrace{0, \ldots, 0}_{L-n_{e_i}}]^{\mathrm{T}}. \tag{9}$$

The new AA was obtained as $\widehat{\mathbf{X}}_{\mathbf{AA}} = \mathbf{X} - \widetilde{\mathbf{T}}$ and, thus, sudden transitions were minimized. Second, in order to completely remove these transitions, a Gaussian window of length equal to $2M$ processed samples was applied to the AA over each transition. The Gaussian window with a maximum amplitude of 1 was generated and divided into two parts. The first M samples formed the signal w_1 whereas the last M samples were w_2. These signals were weighted by the constants k_s and k_e, which were calculated as the amplitude midpoints between the ith complex and $\tilde{\mathbf{t}}_i$ at the instants os_i and oe_i, respectively:

$$k_s = \frac{\widehat{x}_{\mathrm{AA}}(os_i - 1) - \widehat{x}_{\mathrm{AA}}(os_i)}{2} \tag{10}$$

$$k_e = \frac{\widehat{x}_{\mathrm{AA}}(oe_i) - \widehat{x}_{\mathrm{AA}}(oe_i + 1)}{2}. \tag{11}$$

Note that $\widehat{x}_{\mathrm{AA}}$ is the representation of the AA signal obtained from $\widehat{\mathbf{X}}_{\mathrm{AA}}$ after VA cancelation. Finally, the two transitions relative to the ith complex were removed by adding to or subtracting from the AA the signals w_1 and w_2 weighted by the constants k_s and k_e, respectively, as follows:

$$[\widehat{x}_{\mathrm{AA}}(os_i - M), \ldots, \widehat{x}_{\mathrm{AA}}(os_i - 1)]^{\mathsf{T}} = [\widehat{x}_{\mathrm{AA}}(os_i - M), \ldots, \widehat{x}_{\mathrm{AA}}(os_i - 1)]^{\mathsf{T}} - k_s \cdot w_1^{\mathsf{T}} \tag{12}$$

$$[\widehat{x}_{\mathrm{AA}}(os_i), \ldots, \widehat{x}_{\mathrm{AA}}(os_i + M - 1)]^{\mathsf{T}} = [\widehat{x}_{\mathrm{AA}}(os_i), \ldots, \widehat{x}_{\mathrm{AA}}(os_i + M - 1)]^{\mathsf{T}} + k_s \cdot w_2^{\mathsf{T}} \tag{13}$$

$$[\widehat{x}_{\mathrm{AA}}(oe_i - M + 1), \ldots, \widehat{x}_{\mathrm{AA}}(oe_i)]^{\mathsf{T}} = [\widehat{x}_{\mathrm{AA}}(oe_i - M + 1), \ldots, \widehat{x}_{\mathrm{AA}}(oe_i)]^{\mathsf{T}} - k_e \cdot w_1^{\mathsf{T}} \tag{14}$$

$$[\widehat{x}_{\mathrm{AA}}(oe_i + 1), \ldots, \widehat{x}_{\mathrm{AA}}(oe_i + M)]^{\mathsf{T}} = [\widehat{x}_{\mathrm{AA}}(oe_i + 1), \ldots, \widehat{x}_{\mathrm{AA}}(oe_i + M)]^{\mathsf{T}} + k_e \cdot w_2^{\mathsf{T}}. \tag{15}$$

Therefore, after the described process it was possible to obtain an AA signal without sudden transitions, as shown by figure 2(c).

In order to select the best combination for P and M, the similarity (S) between atrial segments outside the QRST was employed. Through this parameter, which will be defined in the following subsection, P and M were selected as 40 and 20 samples (40 and 20 ms approximately), respectively, since they provided optimal results to avoid sudden transitions.

3.2. Performance assessment

The proposed method was thoroughly tested and compared with the previously published ABS strategy, using the quantitative measures of performance that will be next described. The ABS technique was implemented by making use of the details mentioned in the respective literature (Slocum *et al* 1992, Shkurovich *et al* 1998, Holm *et al* 1998, Bollmann *et al* 1998, Xi *et al* 2003).

The AA estimation performance in simulated recordings was computed by comparing the estimated and the original AA in terms of the cross-correlation index (ρ) (Castells *et al* 2005) and mean square error (mse) normalized to the AA root mean square (Lemay *et al* 2007). The cross-correlation index measures the similarities between two signals, and becomes 1 in the case of perfect matching and 0 in the case of completely different and unrelated signals:

$$\rho = \frac{E[x_{\mathrm{AA}}\widehat{x}_{\mathrm{AA}}]}{\sigma\widehat{\sigma}}, \tag{16}$$

where $E[\cdot]$ is the expectation operator and σ and $\widehat{\sigma}$ are the standard deviations of x_{AA} and $\widehat{x}_{\mathrm{AA}}$, respectively. On the other hand, the normalized mean square error (nmse) is defined as

$$\mathrm{nmse} = \sqrt{\frac{\sum_{i=1}^{Q}(x_{\mathrm{AA}}(i) - \widehat{x}_{\mathrm{AA}}(i))^2}{\sum_{i=1}^{Q} x_{\mathrm{AA}}(i)^2}}, \tag{17}$$

Q being the number of samples of AA involved. This parameter can vary between 0 and 1, so that high values indicate notable differences between the original and the estimated AA, and values close to zero are associated with very similar AA signals.

The performance of the proposed method was also tested on real ECGs. Obviously, in these signals, the real AA on the ECG was unknown. As a consequence, the only evidence for a

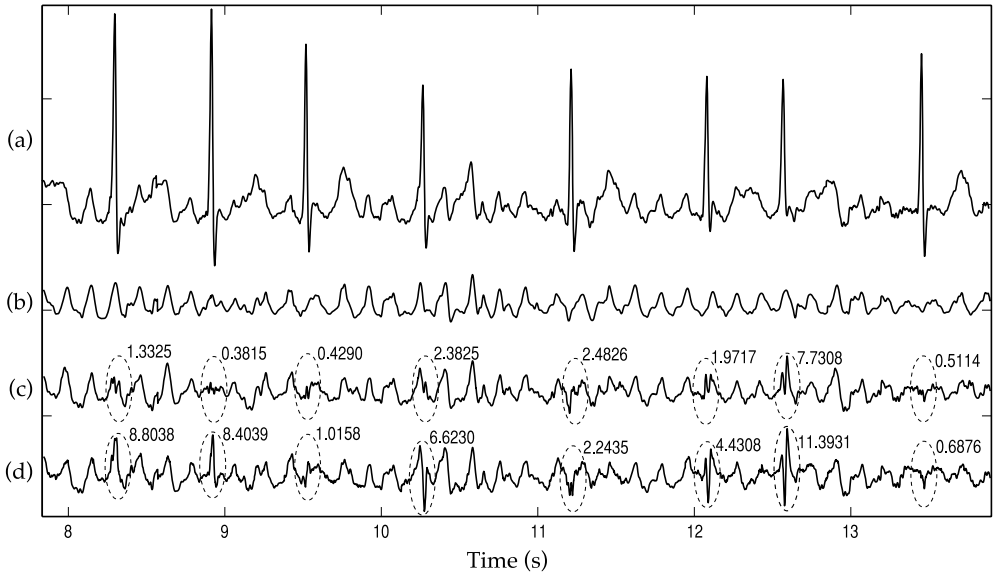

Figure 3. Illustration of the VR index and how it can robustly reflect the defects in VA cancelation. (a) Simulated ECG signal. (b) Original AA added to the ECG. (c) AA signal obtained with ASVC and (d) AA signal obtained with ABS, together with the VR values calculated for each QRS complex.

successful ventricular complex cancelation was the absence of QRST residua. Unfortunately, there does not exist a parameter able to robustly quantify the existence of ventricular residua in the extracted AA. Indeed, each previous work where this aspect has been studied presents different parameters that have been proposed by their authors, as will be later discussed. Thereby, a new robust parameter able to estimate ventricular residua in the extracted AA is next proposed. This index considers the area and amplitude of the QRS interval in the AA to evaluate the ventricular residue (VR). Thus, for the ith complex, VR is defined as

$$\mathrm{VR}_i = \frac{1}{\frac{1}{Q}\sum_{n=1}^{Q} \widehat{x}^2_{\mathrm{AA}}(n)} \sqrt{\sum_{k=r_i-H}^{r_i+H} \widehat{x}^2_{\mathrm{AA}}(k)} \cdot \max_{k=r_i-H,\ldots,r_i+H} (|\widehat{x}_{\mathrm{AA}}(k)|), \tag{18}$$

$2H + 1$ being the number of samples corresponding to the QRS interval. Note that VR was normalized to the AA root mean square in order to obtain comparable values when signals with different amplitude levels are analyzed. To show visually how this parameter behaves, figure 3 plots the AA signals obtained by making use of ASVC and ABS, together with the computed VR values with H equal to 50 samples (50 ms approximately). As can be appreciated, the most important QRS residua present the highest VR values.

On the other hand, in order to evaluate how the AA extraction algorithm preserves the atrial waveforms in those intervals where the AA should remain unchanged, the similarity (S) parameter was used (Rieta and Hornero 2007). This index was evaluated by estimating the cross-correlation coefficient, for each segment free of QRST residua, between the original ECG and the extracted AA. In this respect, a segment free of ventricular residua in the original ECG is defined as the remaining beat time where there is no ventricular activity (Rieta and Hornero 2007). A time interval of 420 ms was considered as the standard VA duration. This interval was wide enough to include all the QRST morphologies. The atrial segments (free

of QRST residua) were composed of the remaining parts between ventricular segments in the ECG.

Time intervals of 10, 30 and 60 s in length were selected from all the ECG recordings, both synthesized and real, described in section 2.1. Next, the ASVC strategy was applied to all of them in order to evaluate the effect of segment length on the method's performance.

3.3. Selection of the optimal set of complexes

To date, numerous authors have introduced or made use of the ABS technique (Slocum *et al* 1992, Shkurovich *et al* 1998, Holm *et al* 1998, Bollmann *et al* 1998, Xi *et al* 2003). In addition, several refinements to improve the extracted AA quality have been proposed (Stridh and Sörnmo 2001, Castells *et al* 2005, Lemay *et al* 2007). However, in these works, the ventricular cancelation template was generated by making use of all the beats available in the ECG recording. Thus, an exhaustive study about the ideal subset of complexes able to provide the most accurate QRST cancelation template has not yet been performed.

In this work, the selection of QRST complexes able to generate the highest quality ventricular template has been thoroughly analyzed by making use of the ASVC method because it reported better results than ABS, as will be shown in section 4.1. However, any other technique based on ventricular cancelation could be used. In this step, the AA signal obtained by two different strategies was evaluated using the parameters described in the preceding section. First, a neighbor-based strategy was applied and each QRST complex was canceled out by making use of a ventricular cancelation template generated with the $N/2$ preceding and $N/2$ subsequent complexes. If there were no $N/2$ complexes before the one under cancelation, more subsequent complexes were selected so that the ventricular template was always generated from N beats. In the opposite situation, more preceding beats were selected. As a second option, a correlation-similarity-based strategy was applied and the N most similar QRST complexes to those under cancelation were used to generate the template. The similarity between QRST complexes was evaluated using the cross-correlation. In both cases, the number of selected complexes was varied between 2 and 60. The upper limit was established because, as will be shown later, the inclusion of a larger number of complexes does not improve results notably. In addition, this is a reasonable upper limit in the number of beats to consider within an ECG of 1 min in length.

4. Results

In this section, a comparative study of ASVC and ABS is first presented. The performance indices obtained are shown together with some examples of the extracted AA signals. Second, the results obtained when a different subset of QRST complexes is selected to generate the ventricular cancelation template are discussed.

4.1. ASVC versus ABS

Both methodologies were first applied to the simulated AF recordings. The cross-correlation and the normalized mean square error between original and estimated AA signals, together with the ventricular residue and the similarity between atrial segments of the ECG and the estimated AA, were used to compare the performance of both methods. Table 1 summarizes the obtained values of these parameters for segments of 10, 30 and 60 s in length, which were selected from each simulated ECG recording. Note that significant statistical differences between ASVC and ABS are reported for all the studied parameters and the analyzed segment

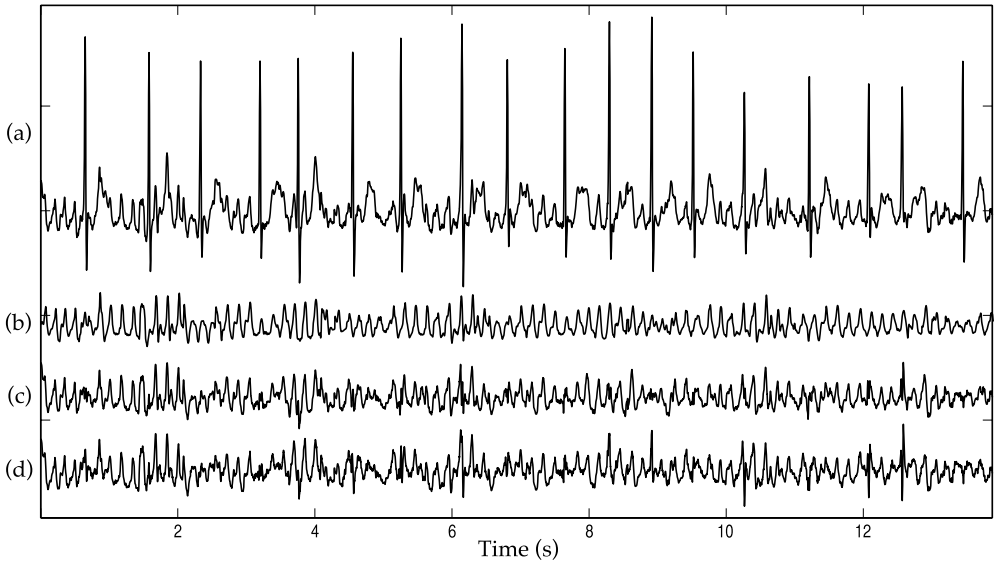

Figure 4. Comparison between VA cancelation strategies for simulated signals. (a) Synthesized original ECG signal under processing. (b) Original AA added to synthesize the ECG. (c) AA signal provided by ASVC. (d) AA signal provided by ABS.

Table 1. Results provided by the comparison between ASVC and ABS obtained for simulated AF recordings of 10, 30 and 60 s in length. Values indicate mean ± standard deviation.

Length (s)	Method	ρ	nmse	VR	S
10	ASVC	0.921 ± 0.041^{a}	0.390 ± 0.106^{a}	1.553 ± 0.458^{b}	0.950 ± 0.033^{b}
	ABS	0.837 ± 0.047^{a}	0.481 ± 0.115^{a}	2.226 ± 0.586^{b}	0.880 ± 0.053^{b}
30	ASVC	0.947 ± 0.023^{a}	0.324 ± 0.073^{a}	1.471 ± 0.479^{b}	0.980 ± 0.015^{b}
	ABS	0.870 ± 0.037^{a}	0.406 ± 0.109^{a}	2.148 ± 0.677^{b}	0.920 ± 0.038^{b}
60	ASVC	0.945 ± 0.024^{a}	0.332 ± 0.073^{a}	1.552 ± 0.386^{b}	0.986 ± 0.012^{b}
	ABS	0.866 ± 0.042^{a}	0.424 ± 0.120^{a}	2.161 ± 0.564^{b}	0.922 ± 0.051^{b}

[a] $0.0001 < p < 0.01$.
[b] $p < 0.0001$.

lengths. The statistical significance (p value) was obtained with the Student t-test because it was corroborated that the obtained parameter values presented normal and homoscedastic distributions by making use of the Jarque–Bera and Levene tests, respectively.

As a graphical summary, figure 4 shows the estimated AA signals corresponding to a typical simulated AF recording when both methods are applied. As can be appreciated, the estimated AA through ASVC matches the original AA with more fidelity than ABS. This fact agrees with the cross-correlation index and the nmse mean values presented in table 1. In addition, it can be observed that the AA extracted by ABS presents QRS residua of larger amplitude, which is coherent with the calculated VR mean value. In addition, another relevant observation is the absence of sudden transitions into the AA segments provided by ASVC. This result justifies the higher similarity values obtained with ASVC. In contrast, the AA obtained with ABS presents notable sudden transitions.

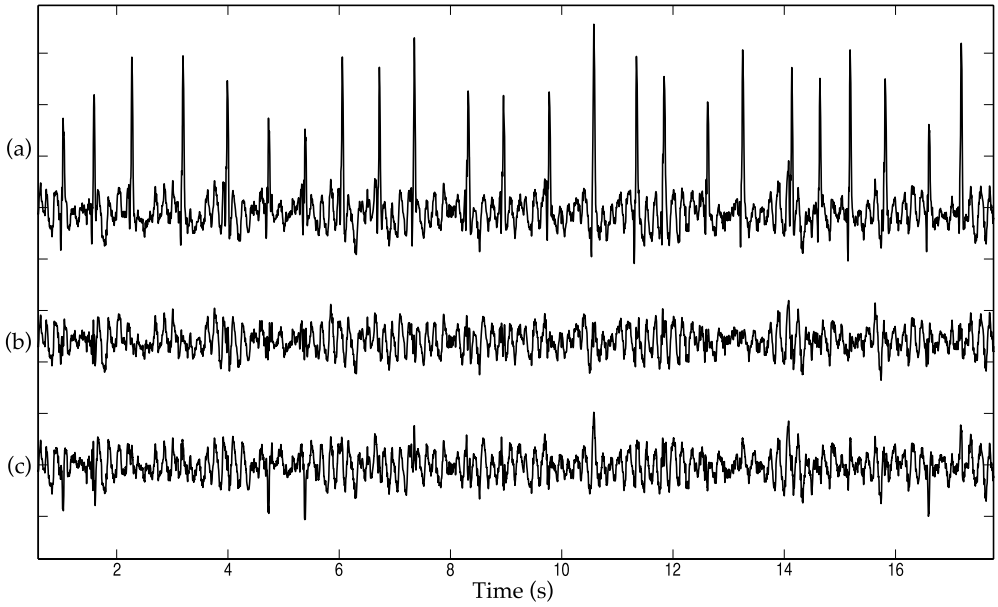

Figure 5. Comparison between VA cancelation strategies for real ECG recordings. (a) Real original ECG signal under analysis. (b) AA signal provided by ASVC. (c) AA signal provided by ABS.

Table 2. Results provided by the comparison between ASVC and ABS obtained for real AF recordings of 10, 30 and 60 s in length. Values indicate mean ± standard deviation.

Length (s)	Method	VR	S
10	ASVC	1.729 ± 0.804^{b}	0.913 ± 0.042^{a}
	ABS	3.186 ± 1.242^{b}	0.864 ± 0.062^{a}
30	ASVC	1.717 ± 0.805^{b}	0.963 ± 0.025^{a}
	ABS	3.111 ± 1.137^{b}	0.916 ± 0.035^{a}
60	ASVC	1.725 ± 0.826^{b}	0.983 ± 0.038^{a}
	ABS	3.159 ± 1.097^{b}	0.951 ± 0.049^{a}

[a] $0.0001 < p < 0.01$.
[b] $p < 0.0001$.

Both methodologies were also applied to real ECGs. Because of the fact that the original AA was previously unknown, the cross-correlation coefficient and normalized mean square error were useless. As a consequence, only the ventricular residue and similarity of atrial segments were computed. The obtained values of these parameters are presented in table 2. In the same way as with synthetic signals, the AA obtained with ASVC presents lower ventricular residue and higher similarity between atrial segments than those obtained with ABS. Moreover, the ventricular residue presents slight variations and the similarity increases considerably when the recording length is increased. In this case, significant differences between ASVC and ABS were also obtained for all the studied parameters and the analyzed segment lengths. An example applied to a real recording is shown in figure 5, which presents the ECG of a patient in AF and the corresponding AA signals provided by ASVC and ABS.

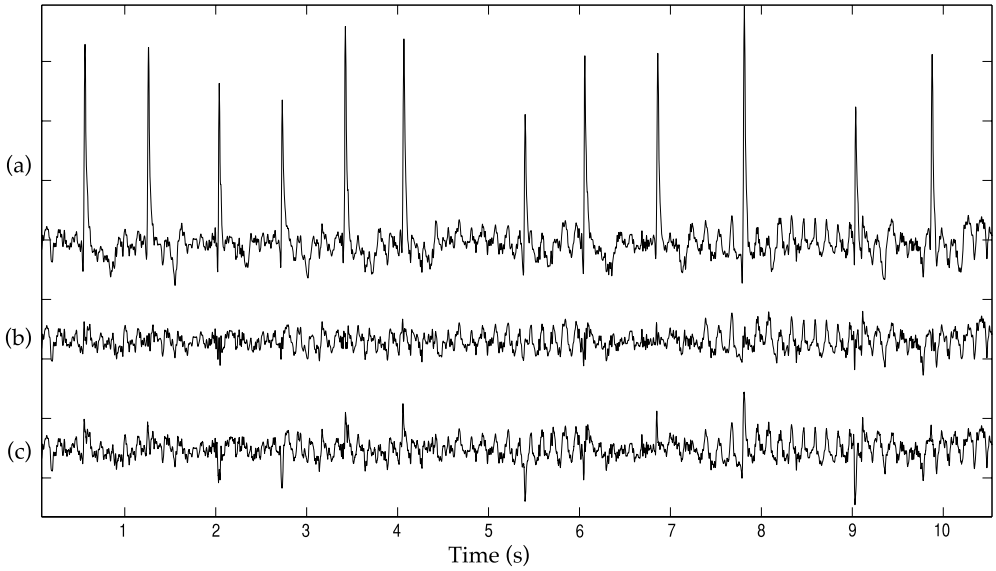

Figure 6. Example of a real ECG in AF with an irregular QRST shape. (a) ECG ready for VA cancelation. (b) AA signal provided by ASVC. (c) AA signal provided by ABS.

Finally, since ASVC considers dynamics in the QRST waveform, a more accurate cancelation template is obtained. As a consequence, it behaves more robustly in those ECGs with variable QRST morphologies. In contrast, the AA estimated by ABS will be highly contaminated by QRST residua; see the example in figure 6. In addition, the ventricular cancelation template obtained with ABS is more sensitive to ectopic beats than that obtained with ASVC. An example is shown in figure 7. As can be observed, the AA obtained with ABS presents more ventricular residua, although the ectopic beat causes a considerable residue in the AA signals provided by both strategies.

4.2. Selection of the optimal subset of complexes

Figure 8 shows mean values for the cross-correlation coefficient, normalized mean square error, ventricular residue and similarity between atrial segments obtained for simulated AF recordings when the complexes to generate the ventricular cancelation template were selected as described in section 3.3. For each number of complexes selected between 2 and 60, two AA signals were obtained by making use of the neighbor-based and correlation-similarity-based strategies and the performance parameters were calculated. In figures 8(a) and (b), it can be appreciated that the cross-correlation coefficient and the normalized mean square error between the original and the estimated AA signal increase and decrease, respectively, when the number of selected complexes also increases. These increments can be observed when the most similar beats as well as the preceding and subsequent complexes to the one under cancelation were selected. However, in the first case higher cross-correlation coefficient values and lower nmse values were obtained.

The variations in ventricular residue with the number of selected complexes are displayed in figure 8(c). It can be observed that higher ventricular residue values were obtained when the preceding and subsequent complexes to the one under cancelation were selected. Moreover,

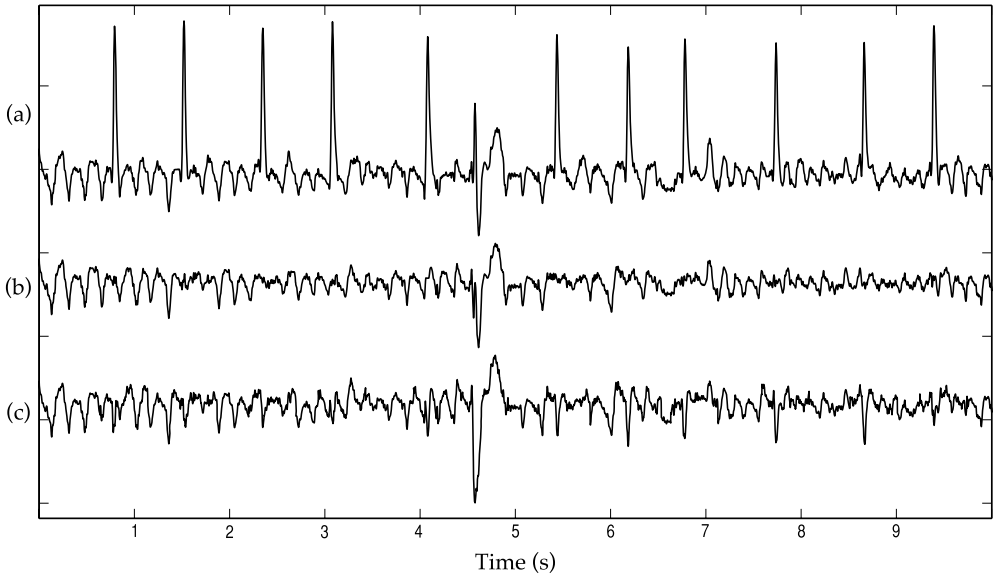

Figure 7. Example of a real ECG in AF with an ectopic beat. (a) ECG ready for cancelation. AA signals estimated with (b) ASVC and (c) ABS.

when the number of complexes was higher than 10, the ventricular residue obtained remained approximately constant. In contrast, when the number of the most similar complexes was increased, the ventricular residue value also increased considerably. Nevertheless, in this last case the ventricular residue values obtained were considerably lower.

On the other hand, the variations in similarity between the atrial segments of the ECG and the estimated AA were coincident with the variation in cross-correlation coefficient, as shown by figure 8(d), although the difference between the two curves was lower than in figure 8(a). Therefore, in both cases as the number of complexes increased, the similarity also increased.

The variations in ventricular residue and similarity as a function of the number of complexes for real ECGs are presented in figure 9. The ventricular residue curves obtained were very similar to those calculated with simulated recordings, also presenting similar values; see figure 9(a). Regarding the similarity of the atrial segments, the calculated curves presented the same tendency as those obtained with simulated AF signals. However, when the number of selected complexes was lower than 20, the curves obtained for the preceding and subsequent complexes and the most similar ones were closer than those presented for simulated signals. Nevertheless, the considerable coincidences between the results obtained with simulated and real AF recordings increase the consistency of the defined indices.

In order to get an automatic way to obtain the optimal number of complexes, ventricular residue must be minimized and similarity between atrial segments must be maximized. Therefore, the best compromise between these parameters could determine the best set of complexes that will provide an optimal QRST cancelation template for long AF recordings. Thus, if the index q is defined as

$$q_i = (1 - \mathrm{S}_i) \cdot \mathrm{VR}_i, \tag{19}$$

where S_i and VR_i are the similarity and ventricular residue, respectively, for a number of complexes equal to i, the minimum value of this index for the tested i values indicates the optimal number of complexes to generate the most accurate ventricular cancelation template.

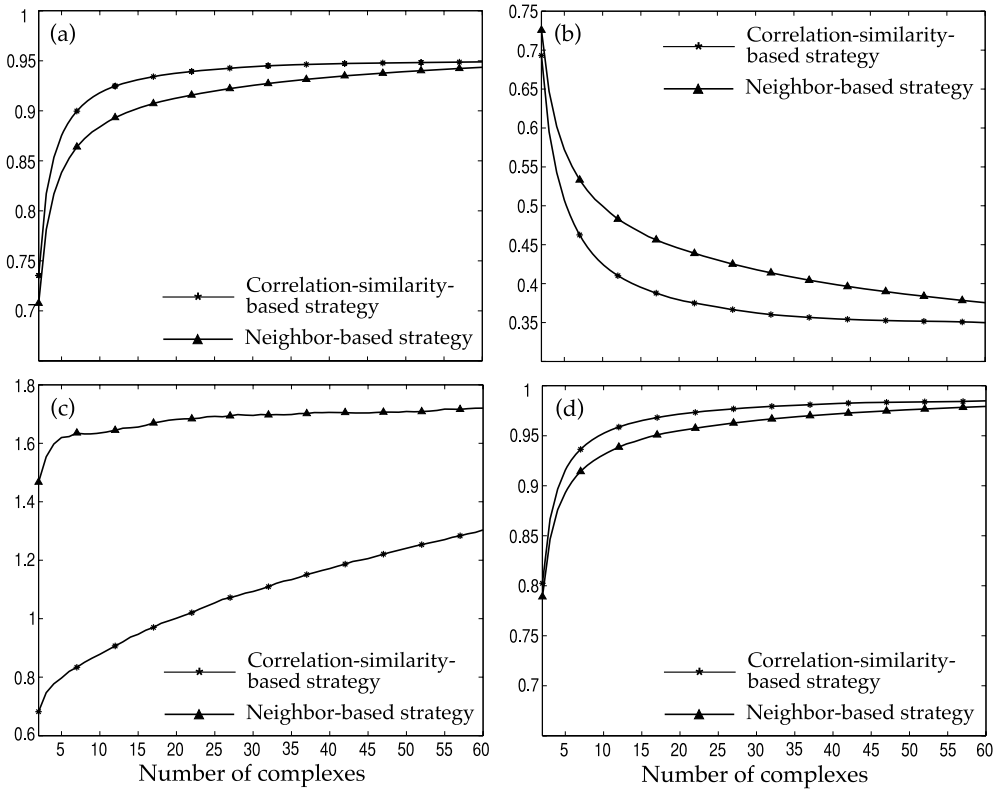

Figure 8. Results for the study of the selection of optimal complexes obtained for simulated AF recordings by comparing the estimated and the original AA. (a) Cross correlation coefficient, (b) normalized mean square error, (c) ventricular residue and (d) similarity of the atrial segments.

For the simulated and real ECG recordings used in this study, the average optimal number of complexes provided by the q index was 30 and 24, respectively. This approach, able to provide the optimal number of complexes, has been added to the ASVC algorithm as an improvement.

5. Discussion

With the aim of providing the purest atrial activity, free of ventricular contamination, for later analysis and study of AF, the cancelation of VA in the ECG has been the subject of investigation during last few years. In fact, this task has been widely considered as an essential step in the development of clinical applications focused on the analysis of AA (Bollmann *et al* 2006). Since the paper by Slocum *et al* (1992), where the ABS technique was used to detect the presence of AF in the ECG, a remarkable number of authors have presented a wide variety of clinical applications associated with the analysis of AA from the ECG in AF after QRST cancelation. In this respect, the AA signal has been used to study the electrophysiological processes that underlie AF, such as the effective refractory period (Wijffels *et al* 1995), autonomic tone response (Zimmermann 2003), atrial activation process (Dibs *et al* 2008), drug effects (Husser *et al* 2005), AF termination prediction (Alcaraz and

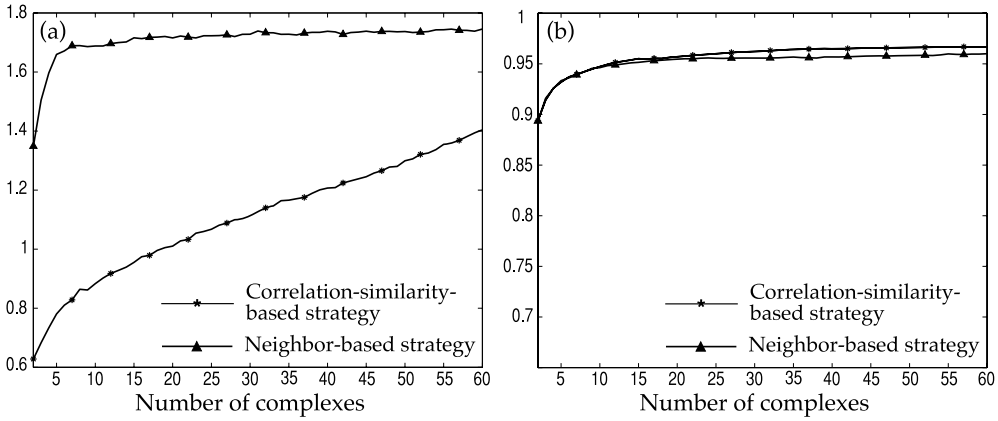

Figure 9. Results for the study of the selection of optimal complexes obtained for real AF recordings by comparing the estimated and the original AA. (a) Ventricular residue and (b) similarity of the atrial segments.

Rieta 2008), cardioversion outcome (Holmqvist *et al* 2006), defibrillation threshold (Bollmann *et al* 2002), AF detection and feature extraction (Stridh *et al* 2006), time–frequency analysis (Husser *et al* 2007), etc.

On the other hand, the ABS method is the most widespread technique for atrial signal extraction when a reduced number of leads are available. However, this method presents an important limitation since it is highly sensitive to variations in QRST morphology. Thereby, this method needs long ECG signals together with regular VA complexes of the same type. In order to overcome this limitation, an alternative technique based on ASVC has been proposed in the present work.

For any ventricular wave cancelation study, the major problem is the validation of the results obtained when real signals are employed. Thereby, ASVC and ABS were validated by making use of simulated and real AF recordings and three performance parameters previously published. Moreover, in order to reliably evaluate the presence of ventricular residua in real recordings, a new parameter, VR, has been proposed. In this way, several previous works have proposed a variety of parameters but, in our opinion, they presented different limitations. Thus, Castells *et al* (2005) used the fibrillatory frequency peak to evaluate their AA extraction method in real ECGs with paroxysmal AF episodes. However, this frequency parameter cannot reflect the existence of particular QRS residua. In Lemay *et al* (2007), the QRS intervals that contained absolute values above the threshold $\xi = m + 2 \cdot \sigma$, m and σ being the median and the standard deviation values of the amplitude distribution of the estimated AA signal, respectively, were identified as ventricular residua. However, we consider that this procedure is unreliable, because fibrillatory waves (f waves) with large amplitudes could be marked as QRS residua, and important ventricular residua with amplitudes lower than the threshold could be omitted. Vasquez *et al* (2001) proposed the beat-to-beat signal-to-noise ratio as the indicator of VA attenuation. The powers for AA and VA were estimated over fixed length windows (140 and 514 ms, respectively) around the R peak. As in the previous cited method, the large amplitude f waves could notably affect this signal-to-noise ratio. Finally, in other highly cited work (Stridh and Sörnmo 2001), only the visual inspection of one example taken from a patient with chronic AF was proposed to support the results based on simulated signals.

In order to overcome the aforementioned limitations with real AF recordings, the parameter VR has been defined and has provided coherent values in all the cases, as has been shown in figure 3; therefore, VR could be considered as a robust index able to estimate the presence of ventricular residua within the extracted AA signal in real ECGs.

The results obtained when ASVC and ABS were compared, which have been presented in tables 1 and 2, showed that ASVC is able to provide a more accurate representation of VA than ABS, thus resulting in higher quality AA extraction for short (<20 s) and long (>20 s) single-lead ECG recordings.

On the other hand, the study about the most suitable subset of complexes proved that the best cancelation template was achieved by selecting the most correlation-based beats similar to those under cancelation, rather than selecting the preceding and subsequent beats, as has been shown in figures 8 and 9. Given that the cross-correlation coefficient between the original and the estimated AA increased with the number of selected complexes (see figure 8(a)), it can be considered that a more accurate VA representation and a better quality AA signal can be obtained when the number of complexes used to generate the QRST cancelation template is increased. However, when this number was higher than 20, the cross-correlation index only increased by 1.11%. Hence, the selection of a high number of complexes is not recommended, because the cross-correlation improvement is notably limited, the computational burden is considerably increased and, finally, the probability of including an ectopic beat is also higher.

Regarding the ventricular residue, it can be observed that this parameter increases linearly when the number of the most similar complexes is also increased; see figures 8(c) and 9(a). When an additional complex is considered to generate the ventricular cancelation template, this selected complex will be the most dissimilar to that under cancelation. This justifies the presence of larger QRS residua. In contrast, when the neighbor-based strategy is used, the QRS residue remains constant when the number of complexes is higher than 10. The high probability of including the same number of similar and dissimilar complexes to those under cancelation could be an explanation of this result.

As figures 8(d) and 9(b) show, the similarity of atrial segments between the ECG and the estimated AA increases as the number of selected complexes also increases. Moreover, it can be appreciated that the similarity curves are very similar to the cross-correlation ones, displayed in figure 8(a). When the number of complexes increases, the presence of AA in the QRST cancelation template is reduced because the common waveform between the AA signals present in the selected complexes is minimized. Thus, the ventricular template part which does not contains VA (the ventricular template width is equal to $RR_{\min}$) will carry f waves with lower amplitude and, therefore, the effect of their subtraction on the atrial segments is minimized. This aspect could justify the outcomes obtained.

The performance parameter values obtained and discussed here allow us to indicate some recommendations about the ASVC method, as follows. (1) A high-quality AA extraction is obtained when the most correlation-based complexes similar to those under cancelation are selected. (2) When only short (<20 s) single-lead ECG recordings are available, all the complexes should be used to generate the QRST cancelation template, because higher values of cross-correlation and similarity between atrial segments and a considerably low QRS residue will be obtained. In contrast, (3) when long (>20 s) single-lead ECG recordings are available, the selection of a number of complexes between 20 and 30 is recommended because a higher number will cause a considerably large QRS residue. In addition, a lower number will provoke poorer AA extraction in terms of the cross-correlation index and the similarity of atrial segments.

As a consequence, the ASVC technique is the first cancelation strategy able to calculate the most suitable number of complexes prior to extracting the AA signal, considering the

specific characteristics of the input ECG recording. In contrast, all the previously proposed ABS strategy variants considered all the ECG complexes to generate the ventricular template (Slocum *et al* 1992, Shkurovich *et al* 1998, Holm *et al* 1998, Bollmann *et al* 1998, Stridh and Sörnmo 2001, Xi *et al* 2003, Castells *et al* 2005, Lemay *et al* 2007).

In comparison to ABS, its proposed variants have provided higher quality AA extraction processes. Thus, Stridh and Sörnmo (2001) proposed the spatiotemporal QRST cancelation method. In this strategy, the average beats of adjacent leads are mathematically combined with the average beat of the analyzed lead in order to suppress the electrical axis alterations and produce optimal cancelation. A modification has been recently introduced by Lemay *et al* (2007) in which the QRS complex and the T wave are separately processed. One important motivation that explains why separate processing is pursued is that the repolarization waveform may be different in nature while the depolarization waveform is not. However, the improvement introduced by these techniques is notably reduced when single-lead ECG recordings only are available, such as in this study, since an individual lead cannot exploit the spatial information of the ECG. Lemay *et al* (2007) proposed other methods that permit cancelation of the ventricular involvement in short episodes of AF, with a minimum length of one complete cardiac cycle. However, this method also needs as many leads as possible to estimate the dominant T wave morphology in each individual beat (van Oosterom 2003, 2004); thereby, its application to single-lead ECG recordings is not recommended. In Castells *et al* (2005), the authors presented a strategy in which the temporal dependence of the AA was exploited, using principal component analysis, to estimate the AA from single-lead ECG recordings. The main features of ventricular and atrial activities are extracted, and several basis signals for each subspace are determined. Finally, the AA is reconstructed back exclusively from the basis signals that formed the atrial subspace. The performance of this method is critically dependent on the algorithm being employed for subspace identification; however, details of this algorithm are not presented. This method was validated using a database of ten simulated and ten real AF recordings, and the mean value of cross-correlation coefficient between the original and the estimated AA signals was only 0.774 ± 0.106, which is lower than that reported by ASVC in this paper.

Moreover, a final advantage of ASVC, in comparison to those other methods that have been described in the preceding paragraphs, is its easy implementation, in which complex mathematics are not required.

6. Conclusions

A method based on adaptive singular value cancelation has been proposed to remove the ventricular involvement in single-lead ECG recordings with high-quality results. Thus, a more accurate QRST cancelation template than those obtained with ABS and its variants was generated, avoiding the inherent limitation of this technique. In addition, ASVC is more robust in ECGs with variable QRST morphology and in the presence of ectopic beats than ABS.

Moreover, a thorough study about the most suitable complexes to obtain the most accurate ventricular cancelation template has been presented for the first time. Two forms of selecting the beats to generate the QRST template were studied, obtaining a higher quality AA extraction when the most correlation-based complexes similar to that under cancelation were selected. Regarding the optimal number of beats to select, the best compromise between ventricular residue and similarity of the atrial segments could be a good approximation. Nevertheless, two interesting recommendations could be indicated: (1) when only short single-lead AF recordings are available, all the ECG beats must be used to obtain the QRST cancelation

template, and (2) with long single-lead ECGs, the best quality AA signal is obtained by selecting a number of complexes between 20 and 30, instead of using all the beats.

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