A Variable Step Size LMS Algorithm for the Suppression of the CPR Artefact from a VF Signal

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Abstract

Artefacts created by thoracic compressions during Cardiopulmonary Resuscitation (CPR) prevent the proper classification of the cardiac rhythm by an Automatic External Defibrillator (AED), making a pause in CPR necessary for a correct rhythm analysis.

Previously proposed adaptive filtering methods have produced satisfactory CPR cancellation results but involve complex out of hospital intervention scenarios. A new adaptive method requiring minimal modifications of the basic operation of an AED is proposed for the suppression of the CPR artefact from a Ventricular Fibrillation (VF) rhythm.

A model of the CPR artefact, based on the instants of the thoracic compressions, is used as the reference signal of a variable step size Least Mean Squares (LMS) algorithm. Emphasis is put on building a reference signal highly correlated to the CPR artefact and on the fine tunning of the LMS algorithm. Satisfactory results have been obtained for the average increase in signal to noise ratio (SNR) and the sensitivity of a commercially available AED.

1. Introduction

Artefacts introduced by thoracic compressions during CPR severely condition the shock/noshock decision algorithms of AEDs. Current CPR guidelines require that chest compressions and ventilation must be discontinued during automatic rhythm analysis. The duration of these hands-off periods affects the defibrillation success rate, infact AEDs are most effective if they are programmed to secure minimal hands-off periods before the delivery of the electric shock [1]. CPR artefact suppression can minimize these hands-off periods, thereby increasing the likelihood of resuscitation success.

Initial studies on CPR suppression were conducted on porcine models. Pigs show a clear spectral separation of the VF rhythm and the CPR artefact, a fixed coefficient high-pass filter is therefore a suitable solution [2]. In humans the VF rhythm and the CPR artefact show a significant overlap of spectral components requiring more elaborate solutions. Adaptive filtering schemes based on two to four reference signals have been shown to produce satisfactory results [3, 4] but present a complex out of hospital AED intervention scenario.

The filtering method introduced in the present work requires minor modifications to the current configuration of AEDs. A single reference signal constructed from the instants corresponding to the thoracic compressions is used. The frequency obtained from those instants is strongly correlated to the frequency of the induced CPR artefact and is the basis to model the reference input signal of a modified LMS type of algorithm.

2. Materials and methods

2.1. VF and CPR record databases

Two databases, corresponding to VF and CPR records, have been used in the development of the proposed CPR cancellation algorithm. The VF database is composed of 191 records of variable duration (5-20s), with a mean duration of 15s, and is compliant with the American Heart Association's (AHA) recommendations for AED rhythm recognition algorithms. All registers have been classified as VF episodes by cardiologists of the Basurto (Bilbao, Spain) hospital.

The CPR database is composed of 17 records of variable duration (15-35s) obtained by basic life support personnel during emergency interventions on patients in asystole. This small number of records is sufficient to appreciate the important variability in: the frequency of the thoracic compressions, the duration of the pauses and the waveform. These differences are clearly observable in the two examples shown in figure 1.

All records have been resampled at a 250 samples per second rate and preprocessed to eliminate DC interferences and base line drifts.

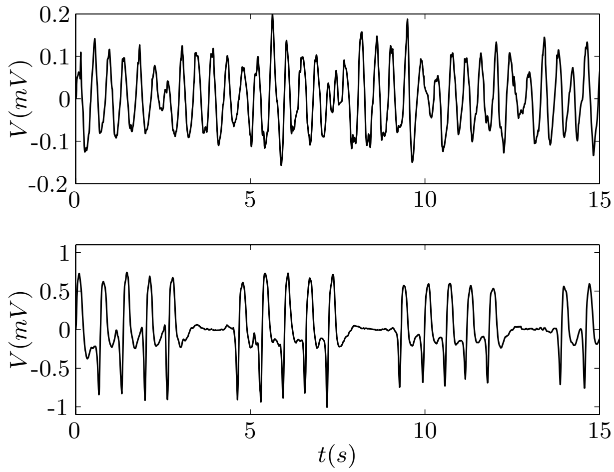

Figure 1. CPR artefact examples

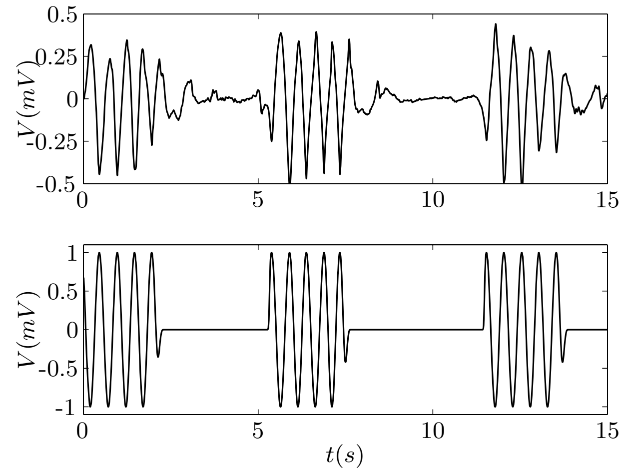

Figure 2. The $(s_{Iref})_1$ reference signal

2.2. The reference signal

Minor modifications of the basic operation of an AED are needed to record the thoracic compression instants. This information is the basis of a CPR artefact model used as a reference signal by the adaptive filtering method. The modeled artefact must account for the observed variability in frequency, pauses and waveform of the CPR records.

A sinusoidal of variable phase and amplitude is used to model the artefact. Since no assumptions are made as to the depth of the compressions, the amplitude ($A(n)$) is held constant during thoracic compressions and is made 0 for ventilation intervals, where the CPR artefact is negligible. Smooth transition periods are defined to avoid abrupt amplitude discontinuities. During the interval corresponding to two consecutive compressions the frequency of the artefact is assumed constant. The phase ($\phi(n)$) is therefore calculated as a piece-wise linear function where the positions of two consecutive peaks are separated by 2π radians. Figure 2 shows an example of an artefact and its corresponding reference signal.

Important harmonic components in the CPR records produce the observed waveform variability. Following the simple model proposed for the reference signal a set of N harmonically related phase and quadrature reference signals are generated as:

$$\begin{aligned} (s_{Iref})_l &= A(n)\cos(l\cdot\phi(n)) \\ (s_{Qref})_l &= A(n)\sin(l\cdot\phi(n)) \end{aligned} \qquad l = 1,..,N$$

2.3. Multiple harmonic VSS-LMS

Figure 3 shows the proposed CPR cancellation algorithm. The input signal is composed of the VF record and the CPR artefact which is assumed to be additive noise. The a_{SNR} coefficient controls the level of the interference and is used to adjust the input signal to noise ratio, SNR_i as:

$$a_{SNR} = \sqrt{\frac{P_{s_{VF}}}{P_{s_{CPR}} \cdot 10^{\frac{SNR_i}{10}}}}$$

Adaptive techniques are used to estimate the interference ($\hat{s}_{CPR}$) which is then subtracted from the input signal to produce the estimated VF signal ($\hat{s}_{VF}$). This output can be interpreted as the original VF signal (s_{VF}) plus a residual noise or error (e); the output signal to noise ratio is easily obtained:

$$SNR_o = 10\log\left(\frac{P_{s_{VF}}}{P_e}\right)$$

CPR artefact waveform variability is accounted for by the use of N harmonically related reference signals. As the harmonic number is increased the contribution to the interference decreases, the amplitudes of the reference signals are therefore scaled by the inverse of the harmonic number. In vector notation $\mathbf{S}_{Iref}$ and $\mathbf{S}_{Qref}$ represent the phase and cuadrature components of all harmonic reference signals pondered by the harmonic number:

$$\begin{aligned} \mathbf{S}_{Iref} &= [s_{Iref\,1}, \tfrac{1}{2}s_{Iref\,2}, .., \tfrac{1}{N}s_{Iref\,N}] \\ \mathbf{S}_{Qref} &= [s_{Qref\,1}, \tfrac{1}{2}s_{Qref\,2}, .., \tfrac{1}{N}s_{Qref\,N}] \end{aligned}$$

Two variable filter coefficients, weights, are used per harmonic, a_l for the phase component and b_l for the quadrature component. A vector arrangement of the filter weights at time n ($\mathbf{a}(n)$, $\mathbf{b}(n)$), is then used to calculate the estimated CPR:

$$\hat{s}_{CPR}(n) = \mathbf{S}_{Iref}\mathbf{a}^T + \mathbf{S}_{Qref}\mathbf{b}^T$$

These coefficients are continuously adjusted according to the difference between the input signal and the output estimated by the filter, $\hat{s}_{VF}$. A simple and widely used weight adjustment process, based on the method of steepest descend, is the Widrow-Hoff LMS algorithm. Better convergence and stability characteristics can be obtained using an extension proposed by Kwong et al. [5], the variable step size LMS (VSS-LMS). The adjustment equations are:

$$\mathbf{a}(n+1) = \mathbf{a}(n) + 2\mu(n)\mathbf{S}_{Iref}(n)\hat{s}_{VF}(n)$$
$$\mathbf{b}(n+1) = \mathbf{b}(n) + 2\mu(n)\mathbf{S}_{Qref}(n)\hat{s}_{VF}(n)$$

where the step size, μ, controls the stability and rate of convergence of the algorithm. The step size is held constant in the LMS algorithm but is variable in the VSS-LMS extension where the following update equations are used:

$$\mu(n+1) = \alpha\mu(n) + \gamma\frac{\hat{s}^2_{VF}(n)}{P_{s_{in}}}$$
$$0 \leq \mu(n+1) \leq \mu_{\max}$$

Two terms control the step size adjustment, an exponential forgetting factor α and a second term involving the error in the input signal estimation, $\hat{s}_{VF}$, scaled by the γ coefficient. Differences in input power levels require a normalization of the error term by $P_{s_{in}}$. All the calculated step size values are finally bounded by $\mu_{\max}$ to ensure convergence.

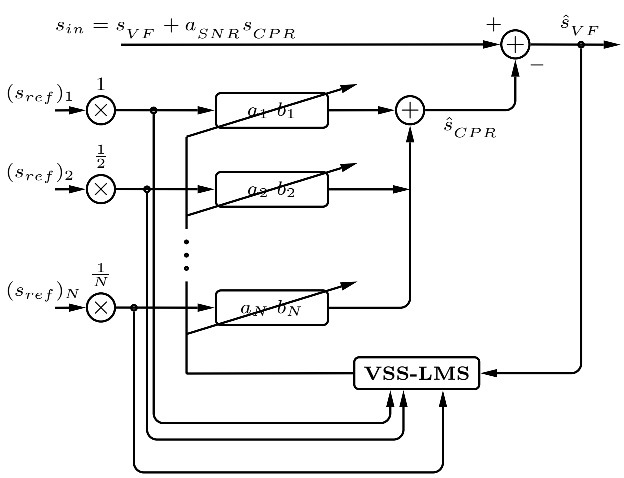

Figure 3. Multiple harmonic VSS-LMS

2.4. Algorithm parameters

CPR cancellation algorithms produce an estimation of the VF signal. A measure of the goodness of the estimation can easily be computed by comparing the output and input SNRs, which in dBs reads:

$$\Delta_{SNR}(dB) = SNR_o(dB) - SNR_i(dB)$$

This figure of merit is used to obtain the optimum values of the VSS-LMS algorithm parameters. Using all the CPR and VF records an average value, $\overline{\Delta_{SNR}}$, of the figure of merit is computed as a function of the different parameters. Variations in the step size adjustment coefficients, α and γ, do not show a significant influence and values close to the ones proposed by Kwong et al. produce good results ($\alpha = 0.975$, $\gamma = 0.001$).

Figure 4 represents the average value of the figure of merit as a function of $\mu_{\max}$ for a single harmonic in the reference signal and $SNR_i = 0dB$. The optimum value corresponds $\mu_{max} = 0.0075$ but no significant variation occurs in the $\mu_{\max} = 0.0075 \pm 0.003$ interval.

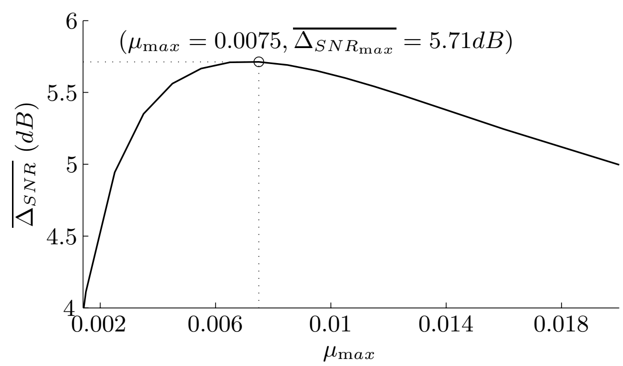

Figure 4. Selection of $\mu_{\max}$; $N = 1$, $SNR_i = 0dB$

The dependency on the number of harmonics is represented in figure 5 for the different input corruption levels considered ($-10, -6, -3, 0, 3\ dB$). A higher number of harmonics gives a better estimate since the CPR artefact model is improved, but involves a higher computational cost (every harmonic requires two additional filter coefficients). No significant advantage is obtained for $N > 2$ making $N = 2$ an adequate choice.

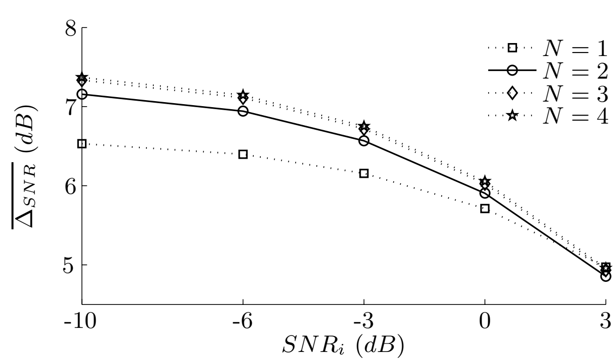

Figure 5. Selection of N, $\mu_{\max} = 0.0075$

3. Results

Once the algorithm parameters are tunned to sensible values the algorithm is fully tested. The mean values of the figure of merit provide a clear picture of the effectiveness of the CPR suppression algorithm. The values obtained correspond to the highlighted curve in figure 5; the average increase obtained is 7.16, 6.94, 6.57, 5.91 and $4.86dB$ with a standard deviation under $1.53dB$ for all scales.

To properly characterize a CPR cancellation algorithm its performance in an AED intervention scenario must be evaluated. An offline matlab version of a rhythm analysis algorithm, deployed in a commercially available AED (Reanibex-200), has been used for this purpose. Sensitivity values obtained before and after filtering are represented in figure 6, the theoretical limit of 99.0% corresponds to the artefact free results. For $SNR_i \geq -3dB$ the sensitivity figures obtained exceed the AHA recommended value of 90%; sensitivity values of 90.4%, 94.0% and 95.7% have been obtained for $SNR_i = -3$, 0, and $3dB$.

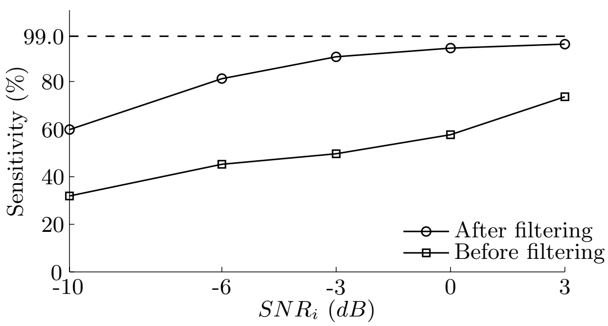

Figure 6. Sensitivity values

A detailed representation of the results reveals an important dependence on the CPR artefact. As shown in figure 7 differences in sensitivity of up to 10.5% are observed between CPR8 and CPR17 for $SNR_i = -3dB$. The maximum difference decreases to 6.3% and 3.7% for $SNR_i = 0$ and $3dB$ respectively, with sensitivity results above 90% for all CPR records.

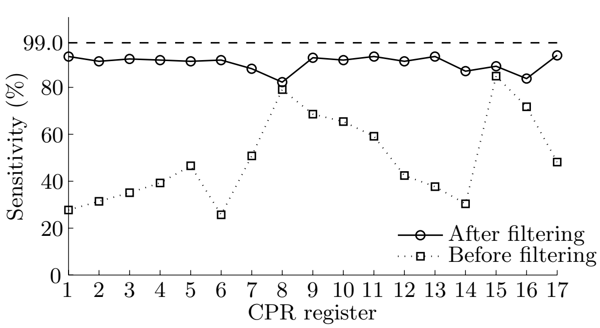

Figure 7. Detailed sensitivity values, $SNR_i = -3dB$

4. Discussion and conclusions

A new adaptive filtering methodology has been proposed for the cancellation of the CPR artefact from a VF signal. In contrast with previous adaptive models the present scheme requires minimal modifications of current out of hospital AED intervention scenarios, which justifies its development.

Promising results have been obtained for both the average value of the figure of merit and the sensitivity of the rhythm analysis algorithm of a commercially available AED. Despite the values falling below previously reported figures [4] a better model for the CPR artefact, through the optimal selection of the scaling factors of the reference signals, should provide improved results.

Further studies should comprise the performance of the algorithm in the presence of other shockable rhythms, particularly fast ventricular tachycardias (VT), and non-shockable rhythms (specificity figures). The VF signal database is ample enough to produce reliable results but the small number of CPR records indicates the need for a confirmation of these results on a wider database.

Acknowledgements

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References

[1] Yu T, Weil M, Tang W, et al. Adverse outcomes of interrupted precordial compression during automated defibrillation. Circulation 2002;106:368–372.

[2] Strohmenger HU, Lindner KH, Keller A, et al. Spectral analysis of ventricular fibrillation and closed-chest cardiopulmonary resuscitation. Resuscitation 1996;33:155–161.

[3] Aase S, Eftestøl T, Husøy, et al. CPR artefact removal from human ECG using optimal multichannel filtering. IEEE Trans Biomed Eng 2000;47(11):1440–1449.

[4] Eilevstjønn J, Eftestøl T, Aase S, et al. Feasibility of shock advice analysis during CPR through removal of CPR artefacts from the human ECG. Resuscitation 2004;61:131–141.

[5] Kwong R, Johnston E. A Variable Step Size LMS Algorithm. IEEE Trans on Signal Processing 1992;40:1633–1642.

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