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# Filtering of Chest Compression Artefacts in the Electrocardiogram

**Jekova I.[*], Krasteva V., Dotsinsky I.**

*Centre of Biomedical Engineering*
*Bulgarian Academy of Sciences*
*105 Acad. G. Bonchev Str., 1113 Sofia, Bulgaria*
*E-mail: irena@clbme.bas.bg*

**Summary:** Long interruptions of cardiopulmonary resuscitation (CPR) in case of a sudden cardiac arrest result in higher failure rate of resuscitation. The current work concerns the filtering of the chest compression (CC) artefacts during CPR, which is essential for the CPR continuation during electrocardiogram (ECG) analysis by automated external defibrillators (AEDs). We have studied two possible approaches – one based on high-pass filter (HPF), and another using band-stop filter (BSF) with adjustable cut-off frequency. The purpose is to improve the quality of the signal provided to the ECG analysis module, aiming at a reliable decision to Stop CC if VF is present or to Continue CC for all other rhythms, including asystole (ASYS) or 'normal' rhythms with ventricular complexes (NR). The two filters are tested with artificially constructed ECG+CC signals, as well as with real ECGs recorded during CPR. The HPF passes the high-frequency components of the QRS complexes and effectively suppresses CC artefacts. This allows correct recognition of NR and ASYS. However, HPF suppresses the VF amplitude thus compromising the VF detection sensitivity. The BSF is favorable for detection of NR and VF but presents problems for ASYS detection because there are often attending residual high-frequency components belonging to the CC artefacts.

**Keywords:** ECG, AED, CPR, ventricular fibrillation, chest compression artefact, band-stop filter, high-pass filter, adjustable cut-off frequency

## 1. INTRODUCTION

Public access defibrillation (PAD) programs recommend the use of automated external defibrillators (AED) for early treatment of out-of-hospital cardiac arrests (OHCA) advising 2 minutes of uninterrupted cardiopulmonary resuscitation (CPR), without a check for termination of ventricular fibrillation (VF) or a check for signs of life or a pulse [7]. The chest compressions (CC) during CPR induce large artefact components into the electrocardiogram (ECG) acquired via the defibrillation pads [6]. The superposition of ECG and CC artefacts results in accuracy reduction of AED shock advisory

[*] *Corresponding author*

systems [1, 2, 5, 8]. Therefore, the current practice recommends CPR interruption when it is necessary to assess the rhythm [3] thus providing noise-free ECGs as required for a reliable shock/no-shock decision in AEDs [9]. However, long interruptions of CPR result in higher failure rate of resuscitation [4]. Therefore, a tool is needed which makes it possible to continue performing CPR while ECG data is collected and analysed by a shock advisory algorithm.

This study is related to the novel feature required for the AEDs to analyse the rhythm even during the CPR aiming to recognize VF and to advise early defibrillation. The current work concerns the CC artefact filtering by two possible approaches – one based on high-pass filter (HPF), and another using band-stop filter (BSF) with adjustable cut-off frequency.

## 2. MATERIALS AND METHODS

*Shock advisory module in presence of CPR*

The process for reliable VF detection in presence of CPR is composed by several modules presented in Fig. 1.

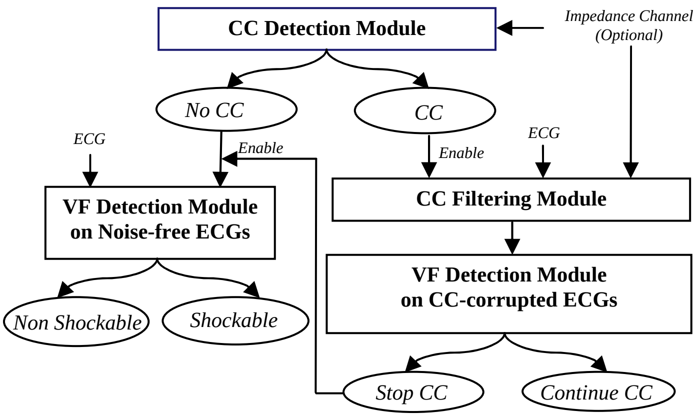


Fig. 1 Diagram of the shock advisory module in presence of CPR

The *CC Detection Module* is typically based on analysis of the impedance signal and provides information whether the rescuer is applying mechanical CC. If no CCs are present, the *VF Detection Module on Noise-free ECGs* is activated to determine with high accuracy whether a shock should be delivered to the patient. If CCs are present, the *CC Filtering Module* must be activated to improve the signal quality to such extent that the following *VF Detection Module on CC-corrupted ECGs* to be able to take a decision to *Stop CC* if VF is present or to *Continue CC* for all other rhythms, including asystole (ASYS) or 'normal' rhythms with ventricular complexes (NR). In the first case the rhythm is again fed to the *VF Detection Module on Noise-free ECGs* before shock application.

*ECG data*

The study is carried out on ECG recordings of OHCA interventions with AEDs (FredEasy, Schiller Medical SAS, France) collected by the emergency medical service in the region of Nancy (July 2006 – January 2007). They are assembled in 2 subsets:

- Subset 1: Pure ECG signals of ASYS, NR, slow ventricular tachycardias (VT) and VF. These signals are taken during the AED analysis periods on clean ECG.
- Subset 2: CC contaminated ECG episodes, taken just before the AED analysis periods. The recognition of the ECG rhythm under CC is inherited from the adjacent AED analysis on clean ECGs with the assumption that the ECG rhythm does not change during the last 10 seconds of CC. The Subset 2 contains CC-contaminated ASYS (pure CC artefacts), NRs, VTs and VFs.

*CPR artefact filtering*

*Method 1: High-pass filter (HPF)*

Linear-phase high-pass filters have been designed by minimax approximation method. Filters with different characteristics (e.g. edge frequencies and number of coefficients) have been studied and the filter with edges at 4 Hz and 8 Hz shows the best performance. Its amplitude-frequency response is presented in Fig. 2.

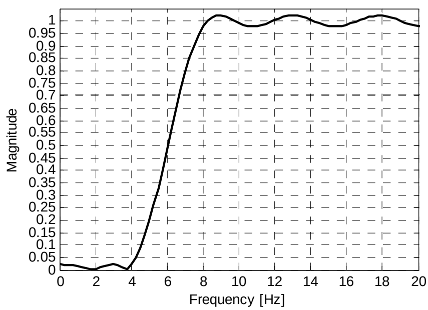


Fig. 2 Amplitude-frequency response of the designed filter HPF

*Method 2: Band-stop filter with adjustable cut-off frequency (BSF)*
Another approach to suppress the CC artefacts in ECG signals is by using a band-stop filter. Following the approach of Irusta et al [8] we apply filter with the following equation:

$$H(z) = \prod_{k=1}^{N} \frac{z^2 - 2\cos(2\pi k f_0) z + 1}{z^2 - 2*(1 - \frac{\mu_k}{2})\cos(2\pi k f_0) z + (1 - \mu_k)}$$

$$\mu_k \frac{BW_k * \pi}{f_s}$$

where $f_0$ is the fundamental frequency of the CC artefacts, $k$ is the number of filtered harmonics, $f_s$ is the sampling frequency and $BW$ is the width of the stop-band. This is equivalent to a cascade of $N$ band-stop filters centered in the harmonics of the fundamental frequency $kf_0$.

To adjust the BSF to the fundamental frequency of the CC artefact, it is considered to be of quazi-sinusoidal waveform in the impedance channel and the mean CC period is obtained from the equation:

$$T_0 = 2\pi \frac{\sum_{i=1}^{m} |CC_i|}{\sum_{i=1}^{m} |CC_i - CC_{i-1}|}$$

where $CC_i$ are the impedance signal samples and $m$ is the number of samples between the beginning and the end of the CC episode [10].

We have studied band-stop filters with different $k$ and $BW$. The filter with $k$=5 and $BW$=1.2 Hz has proved to be the most suitable since it suppresses the CC artefacts without influencing the high-frequency components (above 10 Hz) of the QRS complexes. Its amplitude-frequency response is presented in Fig. 3.

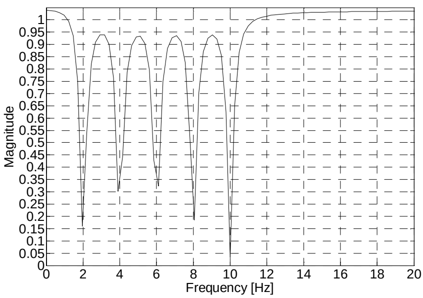


Fig. 3 Amplitude-frequency response of the filter BSF for $f_0$=2 Hz

## 3. RESULTS

Aiming to compare the results achieved with the HPF and BSF we have tested both methods on the same signals.

*Results with artificially constructed ECG+CC*

The first test is on artificially constructed signals of noise-free ECGs superimposed with CC artefacts. We calculate the error of both HPF and BSF as the difference between the pure ECG and the output of each filter. The operation of the two filters on NR, ASYS, VF and VT signals is illustrated with the examples in Fig. 4-7, where:

- *ECG* – a noise-free ECG from Subset 1.
- *CC* – a pure CC artefact on asystole from Subset 2.
- *ECG+CC* – the noise free ECGs superimposed with the pure CC artefact. This mixed signal is the input of the filter.
- *IMP* – the impedance signal, used for calculation of $f_0$ for BSF.
- *HPF*– the output of the HPF (black/top trace).
- *BSF* – the output of the BSF (black/top trace).

*Error* - the difference between the noise-free ECG and the output of each filter (red/bottom traces in HPF and BSF).

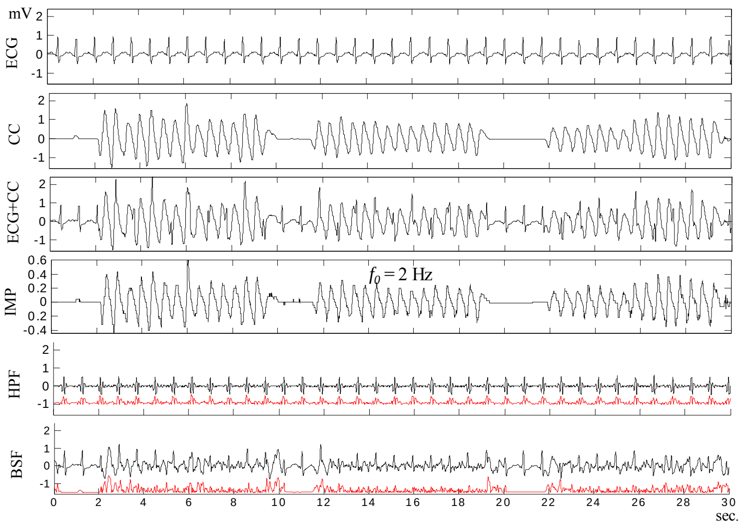


Fig. 4 Example of HPF and BSF filtering of NR mixed with CC. The fundamental frequency of the CC artefact is measured to be $f_0 \approx 2$Hz

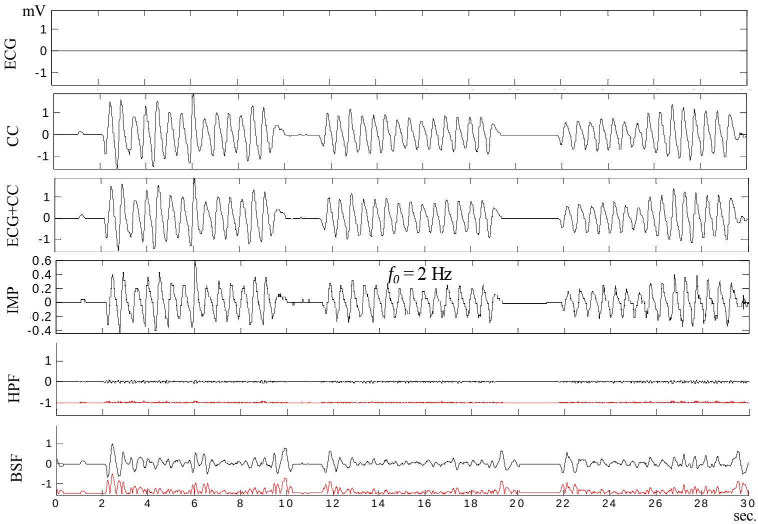


Fig. 5 Example of HPF and BSF filtering of ASYS mixed with CC (measured $f_0 \approx 2$Hz)

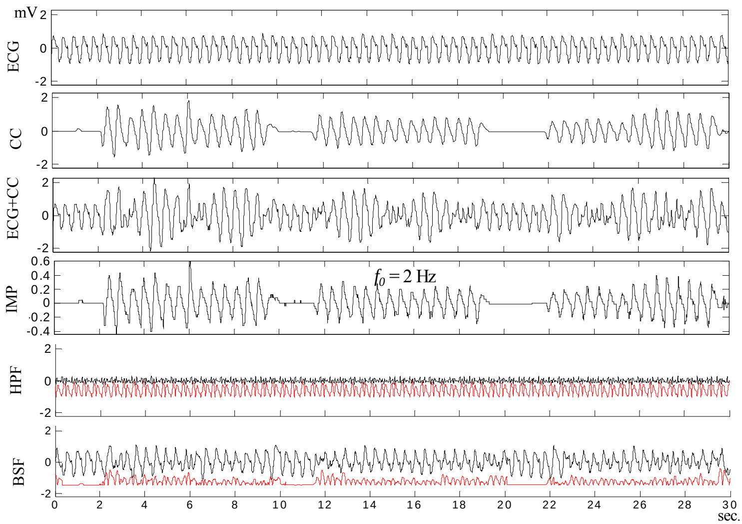


Fig. 6 Example of HPF and BSF filtering of slow VT mixed with CC (measured $f_0 \approx 2$Hz)

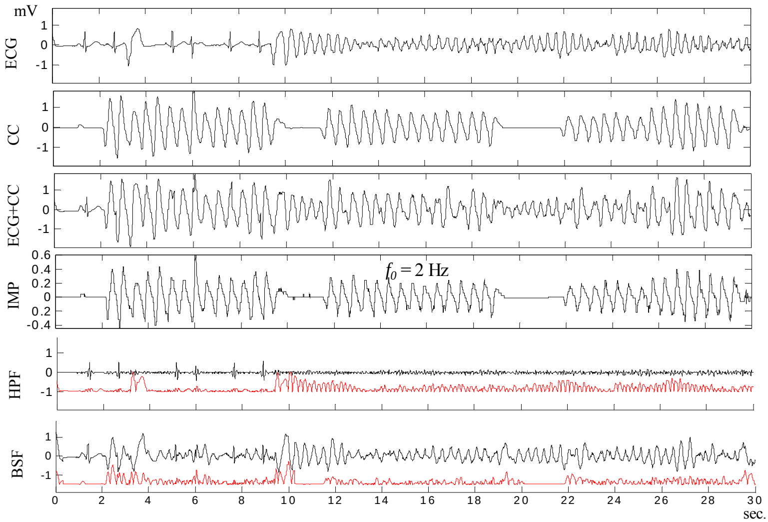


Fig. 7 Example of HPF and BSF filtering of VF mixed with CC (measured $f_0 \approx 2$Hz)

*Results with real CC-contaminated ECG recordings*

At the second step, the performance of HPF and BSF filters is tested with real ECG recordings acquired during CPR (Subset 2). Examples showing the CC-corrupted ECG *(ECG+CC)*, *IMP*, *HPF* and *BSF* signals are illustrated in Fig. 8-10.

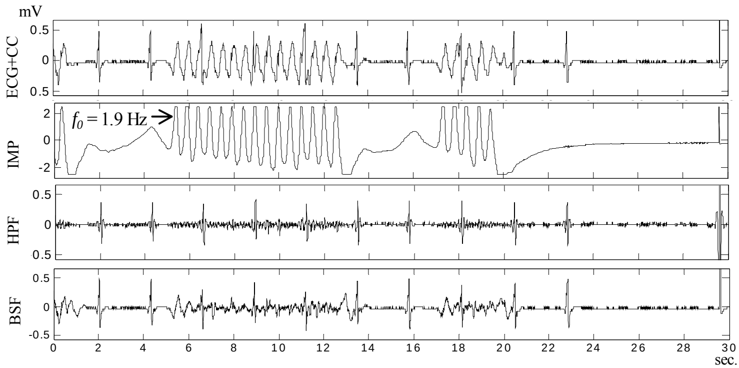


Fig. 8 Example of HPF and BSF filtering of real CC-contaminated NR. The fundamental frequency of the CC artefact is measured to be $f_0 \approx 1.9$Hz

The different examples aim to illustrate different arrhythmias corrupted by various CC artefacts as observed during real OHCA interventions.

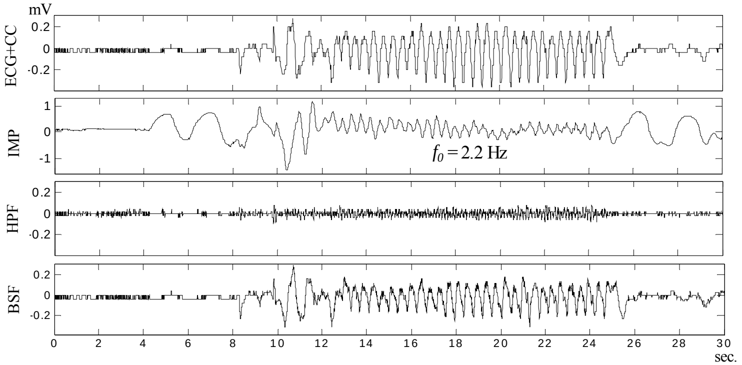


Fig. 9 Example of HPF and BSF filtering of real CC-contaminated ASYS. The fundamental frequency of the CC artefact is measured to be $f_0 \approx 2.2$Hz

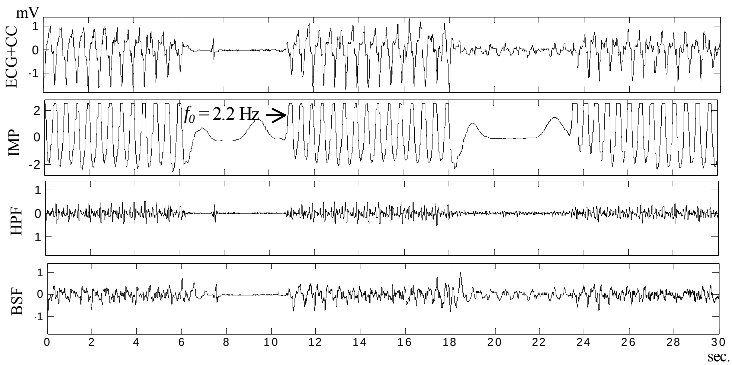


Fig. 10 Example of HPF and BSF filtering of real CC-contaminated ECG just at the moment of transition between NR and VF (between 12$^{th}$ and 18$^{th}$ second). The fundamental frequency of the CC artefact is measured to be $f_0 \approx 2.2$Hz

## 4. DISCUSSION AND CONCLUSIONS

The results with artificially constructed ECG+CC and real CC-contaminated ECG recordings do not differ in general.

HPF shows adequate performance for CC-corrupted NR and ASYS. HPF passes the high-frequency components of the QRS complexes (Fig. 4,8) and effectively suppresses the CC artefacts (Fig. 4, 5, 8, 9) so that HPF output signal is adequate to be correctly classified as NR or ASYS, respectively. A disadvantage is that HPF suppresses the VF amplitude to a critical level near to the asystole threshold (Fig. 7, 10), as well as destroys the waveform of slow VTs (Fig. 6) to resemble VF. Another disadvantage of HPF is due to the constant cut-off frequency setting, leading to HPF failure to sufficiently suppress the high-frequency CC components when present (Fig. 10).

BSF is a better solution for filtering of CC-corrupted NR, VT and VF signals. BSF preserves to large extent the waveforms and the amplitudes of the QRS complexes in NR (Fig. 4, 8), slow VTs (Fig. 6), as well as the VF waves (Fig. 7, 10). Although BSF uses additional impedance channel to measure the fundamental CC frequency and thus to adjust the comb of band-stop frequencies, it has still problems with the suppression of all CC artefact components. This especially impedes the correct detection of ASYS, because the BSF output resembles VF (Fig. 9).

The developed two methods for CC artefacts filtering show different advantages and disadvantages when applied on different ECG arrhythmias and the optimal solution of this problem is still a point of investigations. Considering the results obtained in this study we can conclude that simple high-pass filtering or band-pass filtering adjustable to the central frequency of CC could not provide enough qualitative ECG for reliable shockable/non-shockable rhythm discrimination. A possible solution of this problem could be a complex algorithm combining features of different filter outputs, or even features from the input CC-corrupted ECG itself.

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