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Detection and Adaptive Cancellation of Heart Sound Interference in Tracheal Sounds

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Abstract—— ***Heart sound is often a serious source of noise in the analysis of respiratory sounds. The interference of heart sound is important when breath sounds are analyzed, since the intensity of heart sound is 1 to 10 times greater than the intensity of breath sound. Methods to reduce heart sound interference in breath sound analysis had been suggested. However, the majority of those methods require acquisition of an additional signal and knowledge of the spectral content of heart sound. In this paper it has been proposed an algorithm for the detection and adaptive filtering of heart sounds using only the sound signal and, minimizing distortion in the respiratory sound. The main components of the power spectral density (PSD) of the filtered tracheal sound are comparable with the PSD of the original signal. The experimental results shown reduction of the heart sound interference when the adaptive filter is applied. This kind of filter could facilitate the analysis and a better interpretation of the respiratory sound during spontaneous ventilation.***

I. Introduction

Heart sound is an important source of interference in the analysis of respiratory sounds. It is detectable at all sensor location including the anterior neck. If it is not used any filter, heart sound is the main component of the acoustic signal, with an intensity ratio of 10:1 or greater at some pickup locations [1, 2, 3]. Due to high intensity of heart sound, the amplifiers and the A/D converter can easily saturate, leading to a truncated signal with artefacts. Heart sounds are particularly difficult to manage in children, where the ratio between heart and respiratory sounds is still greater.

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Several methods have been used to minimize the effect of heart sounds in the analysis of respiratory sounds. Selection of pickup sites where the heart sounds are less intense is useful, but may lead to the loss of potentially important data.

Increasing respiratory flow level increases the intensity of breath sounds without altering the intensity of heart sound. This method is useful with cooperative patients who accept to change their breathing pattern upon request, whereas it will be more complex with infants, debilitated patients or patients with severe airway diseases.

It is important to note that other types of interference could be present in the respiratory sound records: vocal sounds, muscle sound produced during muscle contraction and ambient noise. Although they can interfere in breath sounds records, their contribution is not relevant in well controlled protocol develop in respiratory laboratory. Thus, these types of noise are not investigated in this paper.

When there is no frequency overlapping between the interesting signal and the noise, the classical method used to remove or reduce the noise in the signal consists in the use of a linear and time-invariant filter (high pass, low pass or band pass).

Nevertheless, when the frequency content of the interference is overlapped with that of the analyzed biomedical signal, it is necessary to design a specific filter that removes the noise and preserves the signal. In the ideal case, where the signal and the source of interference are stationary and where it is known a good approximation of the noise waveform, it is possible to design an optimum filter with constant coefficients. Nevertheless, when the interference is not stationary, it is desirable to implement a filter that adapts their coefficients in order to cancel the interference, without altering the main signal.

The adaptive cancellation of interferences requires the use of a primary input, that contains the signal contaminated by the interference, and another reference input, that contains a signal highly correlated with the noise contained in the primary input but uncorrelated with the signal to study. With this method, the reference signal is adaptively filtered and removed from the primary input to obtain an estimation of the studied signal.

In this paper, an adaptive filter that performs reduction of heart sound interference in the respiratory tracheal sound is presented. The reference signal is estimated from the respiratory signal, using a detector that identifies the cardiac activity sound.

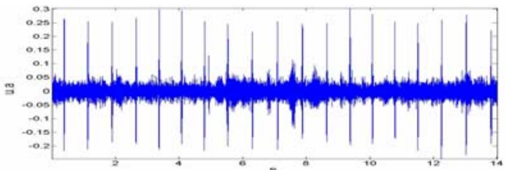

Fig.1. Segment of tracheal sound with heart sound

II. METHODOLGY

A. Signals and Population

Respiratory sounds were recorded, non-invasively, from the trachea, using a piezoelectric PPG sensor (Technion University, Haifa, Israel) [4, 5].

The signal was amplified and filtered with a 4 poles Butterworth analog filter with a bandwidth between 70 and 2000 Hz, (Krohn-Hite 3916B, England). Signal was next digitized with a sampling rate of 5000 Hz. The resolution of the ADC used for acquisition was 12 bits.

The Fig. 1 shows an example of a respiratory sound signal with heart sound noise. It can be observed that the cardiac signal interferes with the respiratory sound.

The tracheal sound signal database utilized in this paper is composed of 26 records of subjects with asthma (15 male and 11 female) with different levels of obstruction, according to a previous spirometry. Those signals were recorded during 120 seconds during spontaneous ventilation in the *Department de Pneumologia, Hospital Universitari Germans Trias i Pujol (Badalona, Spain).* For every patient, two records were acquired: at baseline (PRE) and 20 minutes after inhalation of a bronchodilator drug (POST). Of these 26 records, 18 patients presented high level of heart sound.

B. Description of the algorithm

This section shows the structure of the algorithm used for the reduction of heart sounds in respiratory sound signals. The algorithm has two main parts. The first part consists in the generation of the reference signal (nb_k), that must be highly correlated with the heart sound interference. The second part corresponds to the description of the adaptive filter structure.

1. Generation of the reference signal

The main contribution of the proposed adaptive filtering method is that the reference signal is generated from the original respiratory sound signal (d_k). The estimation of the reference signal consist in the detection of the cardiac activity present in the respiratory sound signal, the selection of heart sound template, and then the generation of the reference signal nb_k , as a train of heart sounds.

The algorithm for the cardiac activity detection is based on the convolution between a cardiac beat sound (CBS) pattern (manually selected) and the original respiratory signal, filtered in the frequency band where the heart sound activity is present [7, 8].

Fig. 2 shows the algorithm used to generate the reference signal. As the main cardiac frequencies are bounded in the range 4 -150Hz [6, 8, 9], the original signal of respiratory sound (Fig. 3A) was filtered with a 4th order Butterworth filter with this frequency range. The frequency band of the filter was chosen so that the energy of the cardiac beat was predominant in the signal and so that a great part of the respiratory sound was eliminated (Fig. 3B).

Then, the filtered signal is convolved with a CBS pattern (Fig. 3C). The next step consists in a non-linear transformation: squared convolution (Fig. 3D).

This transformation makes the signal positive and allows its integration. The local maxima of this signal determine the fiducial point for the detection of the cardiac beats. The detection of those local maxima is realized by means of two moving average windows on the convolved signal (with lengths of 0.1s and 1s, respectively) (Fig. 3E).

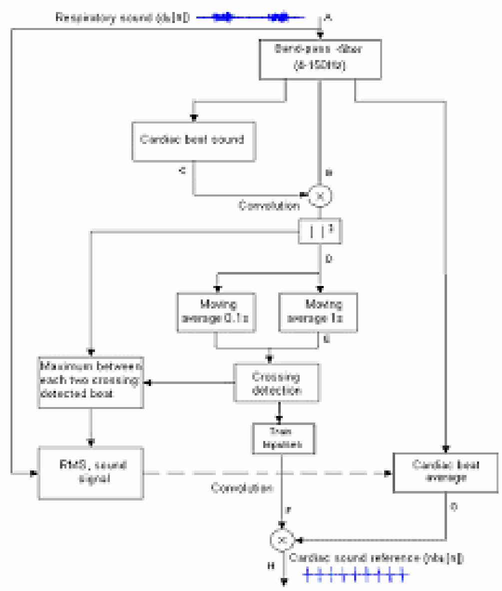

Fig. 2. Algorithm for detection and estimation of cardiac activity (Sequence A to H correspond to signals of fig. 3)

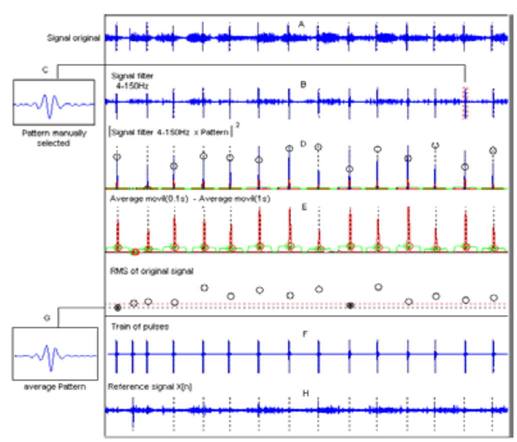

Fig. 3. Example of cardiac sound estimation

A train of unitary impulses is generated from those local maxima, and thus each detected cardiac beat matches an impulse (Fig. 3F). Firstly, the fiducial points of the cardiac beats contained in the respiratory signal are detected. Next, the RMS value is calculated from the respiratory signal (without filtering), within a window located between 0.3 and 0.1s before the detection, and another window located between 0.1 and 0.3s after the detection. The cardiac beats whose value RMS does not overcome a fixed threshold are used to generate an average cardiac beat (Fig. 3G).

Finally, the unitary train of impulses is convolved with the average cardiac sound template, to generate an estimated cardiac sound signal (Fig. 3H). This signal (nb_k) is considered as the reference input of the adaptive filter.

The Fig. 4 shows the correspondence between the cardiac beat sound pattern (manually selected) and the average cardiac beat. This last estimated beat has been used to obtain the cardiac sound reference signal of the adaptive filter.

2. Adaptive filter

The reference input generated in the previous section is an estimation of the cardiac sound interference in the respiratory sound signal. Nevertheless, cancellation of interferences by a simple difference between the respiratory signal and the reference input has the disadvantage that it introduces noise when it appears a false positive or a false negative in the heart sound detection algorithm. Besides, this kind of direct elimination does not allow a right adjustment to beat to beat variation of the cardiac sound signal. For these reasons, the use of an adaptive filter is more suitable.

The Fig. 5 shows the basic diagram of the adaptive filter, as it is described in [5]. It includes two inputs: the primary input $d_k = S_k + na_k$ corresponds to the signal, from which is wanted to remove the interference na_k, and the reference input (nb_k) correlated with the interference na_k and uncorrelated with the signal S_k [7, 9].

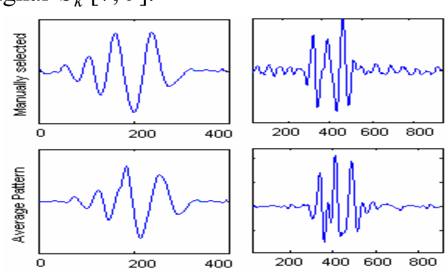

Fig. 4. Two examples of cardiac beat sound pattern

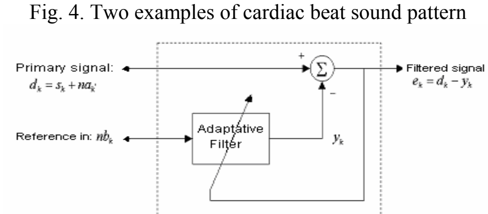

Fig. 5. Adaptive cancellation of interference

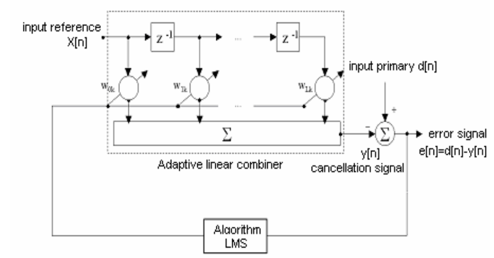

Fig. 6. Scheme of adaptive filter.

The output signal of the filter (error signal) is calculated as $e_k = d_k - y_k$, where y_k is the output of the adaptive filter. The reference input is processed by the adaptive filter which automatically fits its impulse response, depending on the error signal e_k. Then the output signal y_k is generated, and reduces the interference na_k from the primary signal d_k.

It can be noticed that to minimize the power of the error signal $\left(E\left[e_k^2\right]\right)$ is equivalent to minimize the power of $na_k - y_k$. Consequently, to minimize e_k will allow us to fit the output of the filter to the interference component of the primary signal na_k:

$$e_k^2 = (d_k - y_k)^2 = (S_k + na_k - y_k)^2$$
$$S_k^2 + (na_k - y_k)^2 + 2S_k(na_k - y_k)$$
$$E\left[e_k^2\right] = E\left[S_k^2\right] + E\left[(na_k - y_k)^2\right] + 2E\left[S_k(na_k - y_k)\right]$$

$E\left[S_k(na_k - y_k)\right] = 0$, since that S_k is uncorrelated with na_k and y_k. Thus, if e_k is minimized, y_k will be the best estimation of na_k. The LMS algorithm was used in the proposed adaptive filter, to minimize the error signal e_k.

3. Selection of parameters of the adaptive cancellation

The adaptive filter for noise cancellation (Fig. 6) requires selection of two parameters: the number of weights for the linear combiner (L) and adjustment constant (μ) for the LMS algorithm. The choice of the constant μ is a compromise between the complete cancellation of the cardiac sound activity and the preservation of the respiratory signal. In this study, the following values of the adaptive filter were selected: μ = 100 and L=50.

III. RESULTS

This filter was applied to all the signals of the database with heart sounds. Fig. 7a shows an example of the original signal (d_k). Fig. 7b shows the cardiac sound signal (reference signal nb_k), estimated by the proposed method (section II.B.1). Finally, Fig. 7c shows the output signal of the adaptive filter (free of heart sound). It can be observed a reduction of the heart sound contained in the original signal.

Comparisons between the power spectral density (PSD) (estimated by the method of Burg) of the original and filtered signals are shown in the Fig. 8. These results correspond to the same signals shown previously (Fig. 7). It can be observed that the low frequency information of the PSD in the original signal has been reduced after applying the filter.

The adaptive schemes proposed by Gnitecki et al, [10], Kompis and Russi [8], and Iyer et al [11] required the acquisition of additional cardiac signals. In contrast, the adaptive filter proposed in this paper uses only the respiratory sound signal.

In order to perform a qualitative analysis of the heart sound interference reduction, the PSD of the original and filtered respiratory sound signals were averaged within four frequency bands: 20 to 40Hz, 40 to 70Hz, 70 to 150Hz and 150 to 300 Hz [10].

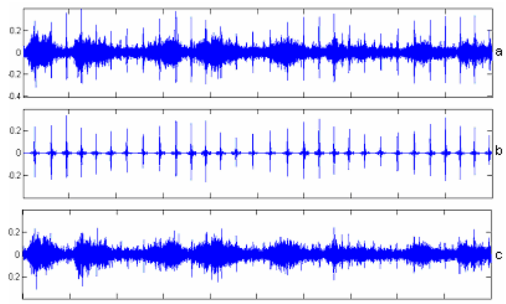

Fig. 7. Examples of signals: a) Input signal with high cardiac sound. b) Reference signal c) Output of adaptive filter

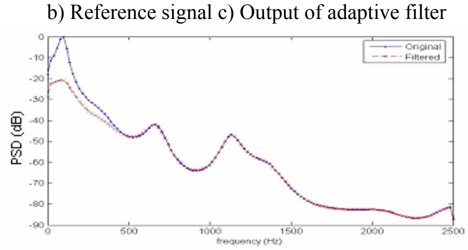

Fig. 8. Examples of PSD of original (blue line) and filtered signal (red line)

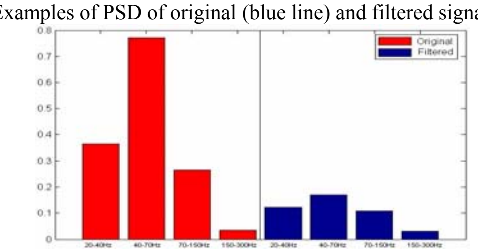

Fig. 9. Average PSD calculated for original and filtered signal

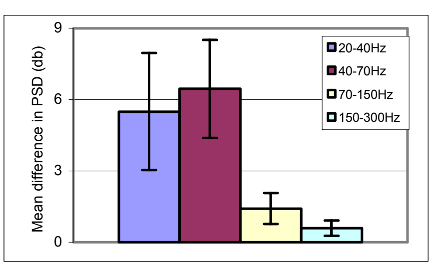

Fig. 10. Differences between PSD of original and filtered signal

The Fig. 9 shows an example of the results obtained in these frequency bands for the signals shown in Fig 8. It can be seen a remarkable difference between the average power of the segment with and without heart sounds. The Fig. 10 shows the mean difference obtained in the PSD (dB) in all the signals of the database that presented heart sound activity (18 patients). It can be observed that the main difference is in the first two bands.

IV. Conclusion

This work presents a method for detection and cancellation of cardiac sounds in respiratory tracheal sounds, using an adaptive filter. The proposed method uses only the sound signal, and estimates the cardiac activity from the respiratory signal itself, and does not use any external signal of reference. The results indicated that this adaptive filter reduces the contribution of the heart sound, including in the cases with low level of respiratory flow. Further investigation should be performed in order to develop an automatic criterion to select the parameters of the adaptive filter, μ and L.

Therefore, this filter may improve the signal analysis and interpretation of the respiratory sound during spontaneous respiration, in the cases where cardiac sound is a relevant noise component.

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