A Sub-Band Energy Tracking Algorithm for Heart Sound Segmentation

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Abstract

The objective of this paper is to present an algorithm for automatic segmentation of the heart sound. The algorithm utilises an Auto Regressive (AR) model to estimate the Power Spectral Density (PSD) of the signal as well as the energy in certain frequency bands for consecutive overlapping frames. The starting and end points of each event are then calculated by filtering the tracking level using a morphological transform and estimating the boundary of its dominant peaks. The algorithm was tested for 960 cycles of heart sound recorded from all four popular auscaltatory areas of 30 patients. Results indicate the capability of this algorithm to isolate desired events in subjects with various pathological conditions.

1. Introduction

It is widely accepted that many pathological conditions cause aberration in heart sound much before they are observed in other comparable techniques such as ECG analysis and due to this fact considerable research efforts have been devoted for finding and documenting the diagnostic features of heart sound by either analyzing the template or frequency characteristics of heart sound components. The externally recorded heart sound, (PCG), is a composite acoustic signal generated by deterministic events such as valve opening and closure as well as non deterministic events such as turbulence produced by blood flow. Any aberration in the structures that generate different components of the heart sound is likely to manifest its signature in the associated components. Considering this fact, segmentation of the heart sound into isolated events is a primary requirement for any attempt towards analyzing the heart sound for diagnostic purposes. Being a composite signal and normally superimposed by various background noises including ambient, instrumental and extra-cardiac noise such as respiratory noise makes the automatic segmentation of the PCG a difficult task. There has been limited research on PCG segmentation. Amongst them the methods proposed by Iwata et.al [1], Lehner et.al[2] and Branek et.al[3], are few examples. Iwata et.al developed a detection algorithm for the first and second heart sound(S_1 and S_2) based on the frequency domain characteristics of PCG evaluated by Linear Prediction method. Lehner et.al. proposed an algorithm for PCG segmentation using the ECG and carotid pulse as a reference. In their algorithm the beginning of S_1 is estimated by using the onset of the R wave in ECG and the beginning of S_2 by using the Dicrotic Notch in Carotid pulse. In Branek et. al. algorithm major events in the cardiac cycle are detected by analyzing the PCG envelope found by either low pass filtering or Hilbert transform. Some of the existing methods for PCG segmentation use some assumed knowledge about the PCG either in the time or frequency domain. This makes them sensitive to recording quality and the overall frequency response of the entire system. On the other hand some of these methods fail to perform properly due to following reasons:

-The dominant peaks in PCG envelope are not well correlated with cardiac events when they are buried in strong murmur.

-The timing between electrical and mechanical (E-M) activities in a cardiac cycle will not be exactly constant amongst different patients because of a variety of pathological conditions. These conditions maybe either electrical or mechanical in nature. Bundle branch block, valvular abnormalities caused by severe aortic or mitral disease, hypertrophic left ventricle caused by myopathy and high blood pressure as a result of valve disease are some of the possible reasons for the lack of synchronizm in E-M timing of cardiac activities.

As a part of our major effort towards developing a multi-channel PCG analyzer for non-invasive detection of heart diseases, a method for PCG segmentation has been developed. This algorithm

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which is based on tracking the energy in a certain frequency band, uses the relative duration of S_1 and S_2 regarding the entire cardiac cycle to discriminate the energy peaks belonging to S_1 and S_2 from other events. Since this is a relative measure, the algorithm is not sensitive to recording condition and abnormalities in PCG.

2. Data Acquisition

Heart sounds were recorded from four auscultatory areas namely Aortic Area (AA), Mitral Area (MA), Pulmonary Area (PA) and Left Sternal Edge (LSE) simultaneously using four HP21050A contact sensors having a flat frequency response from .02 to 2000 Hz. ECG signal from lead II of a three lead standard was also recorded as a timing reference to isolate each cardiac cycle. All the channels were digitized using a 12 bit multi-channel A/D converter with a sampling rate of 3000 Hz. The digitized signal *y(n)*, then was used as the input to the segmentation algorithm.

3. Algorithm

In a cardiac cycle, S_1 and S_2 are the two major sounds. The first heart sound, S_1, caused by ventricular contraction which coincides with the onset of the R wave in ECG (Figure 1). Following the systolic pause the second heart sound, S_2 is caused by closure of semilunar valves just after the T wave in ECG.

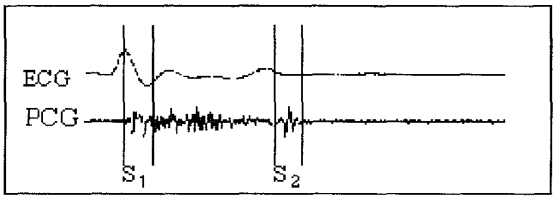

Figure 1. ECG and PCG for a patient with Systolic murmur.

As a primary step in segmentation algorithm a QRS detection algorithm is used to isolate the recorded heart sound into individual cycles. Each frame of heart sound starting 50 ms. before the onset of the R wave and ending 50 ms. before the onset of the next R wave is considered as a PCG cycle and is subjected to the following processing stages for segmentation.

3.1. Estimation of the frequency band for energy tracking

The next stage is to segment each cardiac cycle into S_1, S_2, Systolic and diastolic parts. The boundaries of S_1 and S_2 are determined by tracking the energy in a certain frequency band which represent the -3dB boundaries of the dominant peaks in S_1 and/or S_2 spectrums.

Several parameters including the structure of transmission media between heart and chest wall and pathological condition of heart valves affect the location of dominant frequencies of S_1 and S_2. Therefore it would be undesirable to perform the energy tracking procedure by a pre-defined frequency band for all cases. This frequency band must be first evaluated for each PCG cycle individually. Previous research indicates that first and second heart sounds contain lower frequency components compared to sounds produced by systolic and diastolic murmurs. On the other hand since both S_1 and S_2 are generated through similar mechanisms, the peak frequencies in their PSD are very close[4].The small difference in main frequency peaks of S_1 and S_2 usually is not detectable from the PSD of a complete cycle of heart sound. Limited frequency resolution in PSD estimation techniques results in merging of close frequency components and therefore they appear as a single peak in the spectrum.

Considering the above fact it is therefore reasonable to associate the first peak of the PSD for a complete cycle with S_1 and S_2. It also suggests that the energy tracking should be performed for a frequency band rather than a single frequency as in [1].

To estimate the dominant frequency peak corresponding to S_1 and S_2 and its -3dB boundary, an AR model of order 15 driven by white noise process was used to model each PCG cycle. The output of this model, $\hat{y}(n)$ can be represented as a linear combination of previous output as the following:

$$\hat{y}(n) = -\sum_{k=1}^{p} a_k y(n-k) + u(n) \qquad (1)$$

where:

$\hat{y}(n)$ is the output of AR filter (estimated signal)

p is the model order

$u(n)$ is the driving input (a white noise process with zero mean an variance of σ^2) and

a_k is the model coefficient.

To calculate the model coefficients and the variance of driving white noise Modified Covariance method is used. Having the AR coefficients and the variance of the driving noise, the power spectral density, *PSD(f)* of the signal can be calculated as follows:

$$PSD(f) = \frac{T\sigma^2}{\left|1+\sum_{k=1}^{p} a_k e^{-j2k\pi f}\right|^2} \quad (2)$$

where:

f is the frequency in Hz and T is the sampling interval in sec.

The -3dB boundary of the first dominant peak in the signal spectrum obtained by this method is the frequency band for energy tracking.

3.2. Sub-Band Energy Tracking

To track the energy in desired frequency band obtained from the previous stage each cycle of PCG is divided into N consecutive overlapping frames. The length of each frame is chosen to be 20 ms and it is shifted by 5 ms each time. Each frame, k, is modelled with an AR model of order 8 and the energy in desired band, $E(k)$,*is* calculated by:

$$E(k) = \sum_{f=f_1}^{f_2} \frac{T\sigma^2}{\left|1+\sum_{k=1}^{p} a_k e^{-j2k\pi f}\right|^2} \quad (3)$$

where:

k is the frame number and f_1 , f_2 are the -3dB boundaries of the first peak found in previous stage.

Figure 2 shows the tracking curve, for the same subject in Figure 1. Dominant peaks in the tracking curve provide a clear indication of S_1 and S_2 locations and therefore the systolic and diastolic segments. The notch in the first energy lobe also indicates the split in two component of S_1.(M_1 and T_1.)

In some cases due to partially occluded coronary arteries, the peak frequency of systolic or diastolic murmurs are very close to that of S_1 and S_2. In such cases the tracking curve shows a large peak in systolic or diastolic area. As a result a simple decision rule (e.g. the first two dominant peaks) does not provide an indication of S_1 and S_2 locations. To overcome this problem a peak selection technique based on the morphology of peaks in the tracking curve is used.

3.3. Peak Selection

To accentuate the peaks belonging to S_1 and S_2

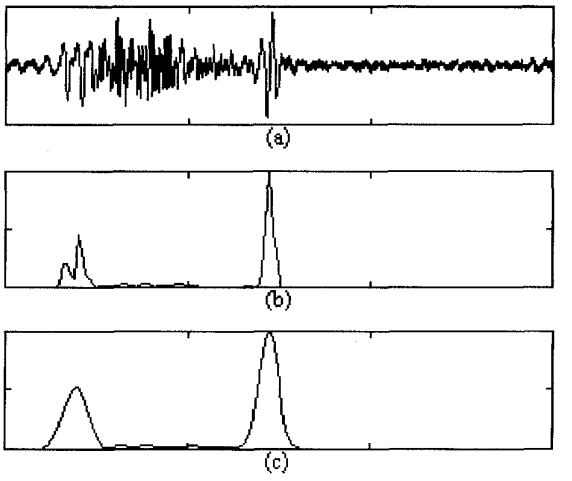

Figure 2. (a) PCG for the same subject in Figure 1. (b) the Energy Tracking curve,(splitting in two component of S_1 (T_1,M_1) is also detected in tracking curve) (c) Tracking curve after smoothing.

and attenuate the peaks belonging to other events we consider the fact that S_1 and S_2 are of shorter duration compared to systolic and diastolic murmurs. As a result the tracking curve shows sharper peaks for S_1 and S_2 compared to murmurs.

The tracking curve $E(k)$ is normalized to its maximum amplitude and then smoothed using a FIR filter of order 8. For the smoothed tracking curve, $E_s(k)$, a peak detection algorithm selects all the peaks above a threshold of -25dB. We refer to the location of these peaks as p_1, p_2,....., p_q. From the smoothed tracking curve, $E_s(k)$, a new set is constructed using the following transformation around the location of each peak, p_q

$$E_{sm}(k) = \begin{cases} E_S(k) - 0.5\left(E_S(p_q - l) + E_S(p_q + l)\right) \\ \quad \text{for} \quad p_q - l \le k \le p_q + l \\ 0 \qquad\qquad\qquad \text{otherwise} \end{cases} \quad (4)$$

where:

$E_{sm}(k)$ is the output of this transform and l is a constant.

This transform has a different effect on sharp and broad peaks. It constructs a new set using the shape of the input, l points around each peak. As a result of this transform sharp peaks will have a larger amplitude and wide peaks will be suppressed(i.e. peaks belonging to S_1 and S_2 versus peaks belonging to systolic and diastolic murmurs). Having found the peaks belonging to S_1 and S_2, the boundaries of these events are found from the smoothed tracking curve

considering a threshold of 5% for each peak. Figure 3(b) shows the smoothed tracking curve for a subject with mild systolic murmur. In this case the energy lobe in the systolic area is of greater amplitude than the energy lobe corresponding to S_2. figure 3(c) shows the tracking curve after the above transform. The wide peak is suppressed and sharp peaks are highlighted.

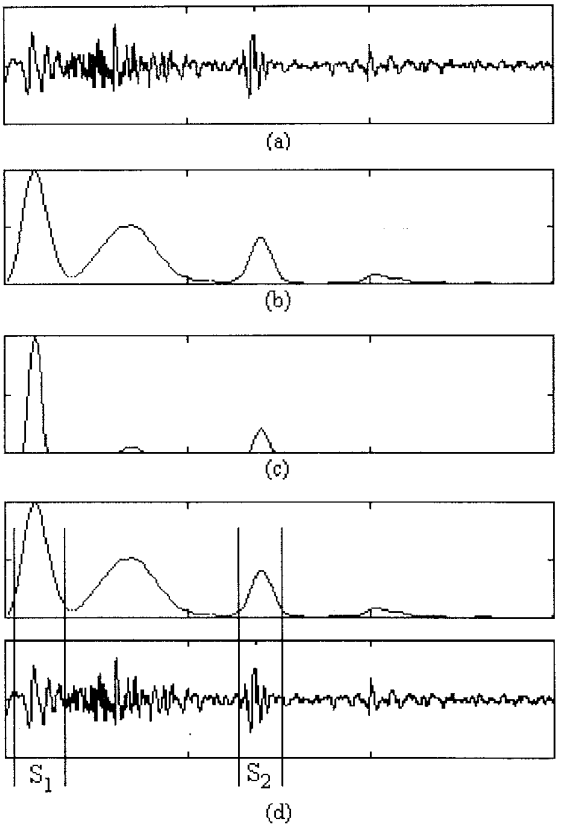

Figure 3.(a) PCG for a subject with systolic murmur. (b) $E_s(k)$, energy tracking curve. (c) $E_{sm}(k)$, peaks belong to S_1 and S_2 are emphasized after the morphological transform with l=3 (d) Segmentation to S_1, S_s, Systolic And Diastolic parts

Having found the boundaries of systolic and diastolic areas, these segments can be processed by the same technique to detect the secondary heart sounds in these areas.(i.e. S_3, S_4, ...).

4. Results and Discussion

An algorithm for segmentation of heart sound based on tracking the energy in a certain frequency band with the following features has been developed:

-The input to this algorithm has no limitation in frequency bandwidth(i.e. low, high frequency PCG).

-The algorithm does not use the E-M timing in the cardiac cycle.

-There is no absolute assumption regarding the frequencies of different components of heart sound.

-The algorithm is capable of detecting the possible splitting of the two components of S_1 and S_2 (i.e. M_1,T_1 in S_1 and A_2,P_2 in S_2)

To assess the merit of this algorithm, it was tested for 960 cycles of PCG from a group of 30 patients with a variety of pathological conditions. Comparison between the result from this algorithm and those identified by experienced cardiologist shows a great degree of agreement. However to express this agreement statistically, more subjective segmentation carried out by experts is being investigated.

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