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An Efficient Thresholding Technique For Segmentation of Phonocardiographic Signals

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***Abstract*—Segmentation and exact timing information of PCG(Phonocardiographic) Signals and its components S1, Systolic period, S2 and Diastolic period in order with time of great importance for accurate diagnosis. In this work, we propose a novel algorithm for the segmentation of Phonocardiographic Signals into its constituent components having both normal and abnormal pathological conditions. The algorithm introduces an adaptive thresholding technique not only to segment but also to estimate the time duration of different cardiac events. Experimental testing was performed on 950 cycles taken from 13 normal, 27 abnormal and 18 fetal cases. The obtained results show 100.00% accuracy in normal, 95.95% accuracy in abnormal signals and 96.7% accuracy in fetal leading to overall 97% accuracy.**

***Keywords*—Efficient Thresholding, Segmentation, Phonocardiographic Signals, S1 sound, S2 sound.**

I. INTRODUCTION

The heart signal(PCG) analysis facilitates accurate diagnosis of several cardiovascular diseases. Various mechanical and electrical events occurring during the functioning of the heart produce a definite pattern of signals and any deviation from the pattern clearly states the presence of underlying heart pathology. Noninvasive procedures such as electrocardiogram (ECG) or Phonocardiogram (PCG) not only provide us useful information about the movement of heart but can be used in some cases where invasion consists of high risk to the patient. In heart sound analysis by auscultation, the observer listens and analyses the heart sound separately. This depends highly on the skill and experience of the observer so analyzing by any computerized technique is desirable. For any automatic analysis, the heart sounds are required to be segmented into its main constituent components i.e., the first heart sound (S1), the Systolic period, the second heart sound (S2) and the diastolic period.

Several earlier attempts to segment the heart sound (PCG) signals reported in literature were either dependent on ECG or/and carotid signals [1],[2]. Acquiring ECG or carotid pulse requires additional hardware which may not be available in some cases. Another major challenge of using these techniquesis due to timing between electrical and mechanical activities of heart which may vary from patient to patient due to different pathological states [3]. In recent years, researchers proposed techniques for segmentation using PCG signal solely. Zhou et al. [4] proposed an algorithm for segmentation using normalized average Shannon energy (NASN). Although using Shannon Energy gives us a smooth envelope of the signal, but the location of each component is not very accurate. Haiyan et al. [5] and Liang et al. [6] introduced a segmentation algorithm using Multiresolution Analysis of Wavelet Transform and wavelet decomposition and reconstruction. Its major disadvantage was the large amount of computation making it too tedious for detection. Xingming et al. [7] presented a method for detecting heart sound based on mathematical morphology. Although its reported accuracy is very high for the normal heart sound, but it is difficult to find a suitable structural elements.

In this paper we present a novel algorithm that uses only the heart sound signal for its segmentation, without ECG as a reference signal.In our proposed algorithm, we segment the first heart sound (S1), the systolic period, the second heart sound (S2) and the diastolic period in sequence in time after predicting the location of the S1 and S2 components and using an efficient thresholding technique. After segmentation, we calculate the duration of S1, S2, systolic and diastolic periods. This information can be used in accurate diagnosis of cardiac anomalies.

The remaining part of this paper is organized as follows. Section 2 describes the acquisition procedure of our database. Section 3 explains our thresholding algorithm. Section 4 consists of discussion about the experimental results whereas concluding remarks are presented in Section 5.

II. DATABASE ACQUISITION

The heart sound were recorded using a wireless data acquisition system based on bluetooth [8]. The signals were sampled at a sampling frequency of 8000Hz and with a 16-bit accuracy. Total 58 recording (including 13 normal, 27 abnormal and 18 fetal) were acquired from people ranging from 16-40 age groups and from abdomen of pregnant women having pregnancy period less than or equal to 8 months. The whole process was carried in presence of an experienced cardiologist. The abnormal signal consists of recording from heart patients suffering from ventricular septal defect, pulmonary valve stenosis, Mitral Regurgitation, Aortic Stenosis and Tricuspid Regurgitation.

III. METHODOLOGY

Heart sounds are very weak acoustic signals. During the process of acquisition, heart sound signals are vulnerable to external acoustic signals and electrical noise interference, in particular, the friction caused by breathing or body movement of subjects[9]. The signals were denoised using wavelet based subband dependent thresholding algorithm which applies signal dependent optimum thresholding values to all the subbands of the signal [10]. The proposed work for segmentation of Phonocardiographic signals consists of three major steps and is described in Figure 1.

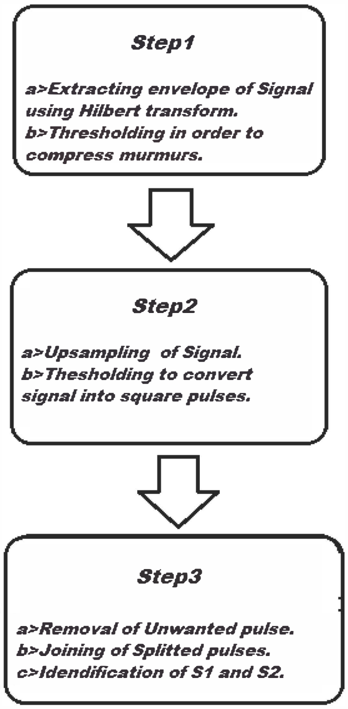

Fig. 1. Block Diagram of Proposed Algorithm.

A. Basic Processing

Outline of a signal is known as*envelope* of the signal and is of great importance in heart sound analysis. In this paper, we use Hilbert transform $H(y(t))$ for the purpose of Envelope extraction of a signal $y(t)$, given by:

$$H(y(t)) = \frac{1}{\pi} \int_{-\infty}^{+\infty} \frac{x(\tau)}{t-\tau} d\tau \tag{1}$$

The Hilbert transform is a linear operator which takes a function and produces a function with the same domain. It provides a concrete means for realizing the harmonic conjugate of a given function or Fourier series. Hilbert transform basically phase shifts a signal by $\pi/2$ radian, Thus a complete complex sequence is given by the equation:

$$Z(x) = y(t) + jH(Y(t)) \tag{2}$$

$$Y(X) = |(Z(x))| \tag{3}$$

Modulus of Z(x) would gives the envelope of the signal (3). The obtained output is passed through a low pass filter to remove ringing and to smoothen the envelope.
In order to compress murmurs and the noise from the envelope, the result obtained above is thresholded using following criteria:

$$Y(x) = \begin{cases} y(x) * 5, & y(x) > M \\ y(x)/5, & y(x) < M \end{cases} \tag{4}$$

where

$$M = mean(Y(x)) * b$$

$$b\epsilon(0,1)$$

Here b is a constant called 'Accuracy factor', selected depending upon nature and purpose of analysis. As b increases, accuracy of getting less noise and murmurs during segmentation increases and as it decreases, accuracy of exact location of S1 and S2 is increased because it sets the basic thresholding level. The output obtained above is then normalized as per equation:

$$Y_n(x) = \frac{y(x)}{max(|y|)} \tag{5}$$

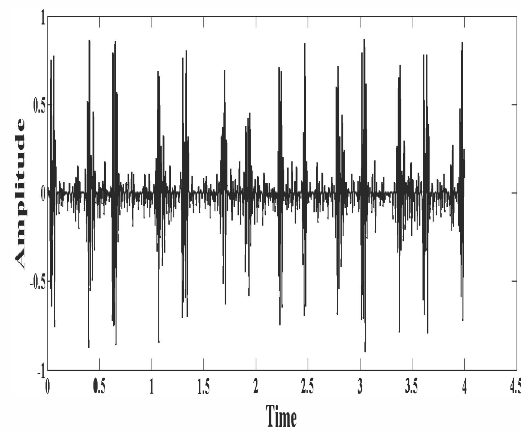

Fig. 2. Amplitude v/s time plot of the signal.

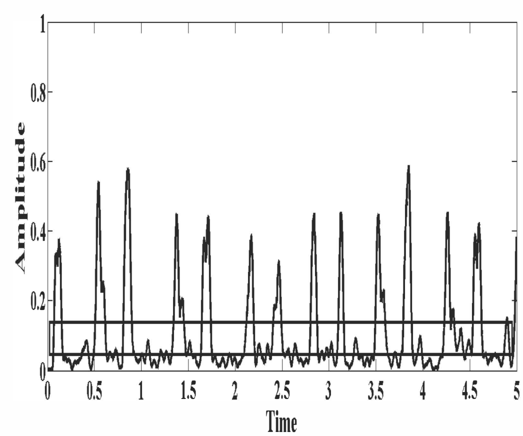

Fig. 3. Extracted envelope of the signal. Square box shows murmurs and noise present.

B. Conversion to Square Pulse

The signal obtained above is upsampled to a sampling frequency of 10,000 Hz making the signal suitable for further

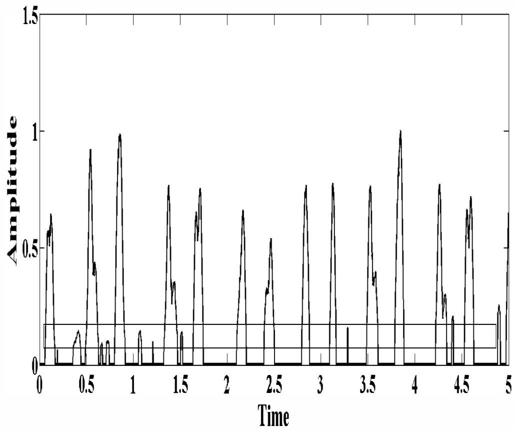

Fig. 4. The normalized thresholded envelope of the signal (see in box the effect of normalized Thresholding).

processing. The upsampled signal obtained is then thresholded according to equation:

$$S(x) = \begin{cases} 1, & Y(x) > T \\ 0, & Y(x) < T \end{cases} \tag{6}$$

Where, T is the thresholding level which can be calculated using the modified standard deviation equation:

$$T = \log(27)[\frac{1}{n}\sum_{i=1}^{n}(x - X)^2]^{\frac{1}{2}} \tag{7}$$

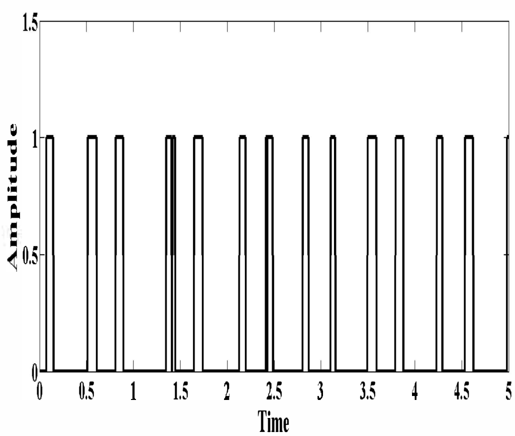

Fig. 5. Square pulses extracted from normalized envelope.

C. Segmentation of Signal

This step involves mainly 3 steps:

- $Removal\ of\ unwanted\ pulse:$
  The time interval of heart sounds S1 and S2 lies between the range 70-150 ms and 60-120 ms respectively[12]. The separation between two consecutive peaks is 100 ms in general [13]. With the help of these known facts, we have removed the pulses generated by noise and murmurs.
- $Joining\ of\ split\ in\ heart\ sound:$
  Taking in account the fact that in general S1 and S2 sounds aere in range of 70-150 ms and 60-120 ms, we introduce here a variable window of size 160 ms starting from each S1 and S2 pulses which joins all the square pulses lying in it.
- $Identification\ of\ S1\ and\ S2:$
  The identification of S1 and S2 is done by using the fact that systolic time duration is always less than the diastolic time duration [14]. Using the information that between any S2 pulses, there must be a S1 pulse and vice versa, any 3 consecutive pulses can be chosen for identification of S1 and S2.

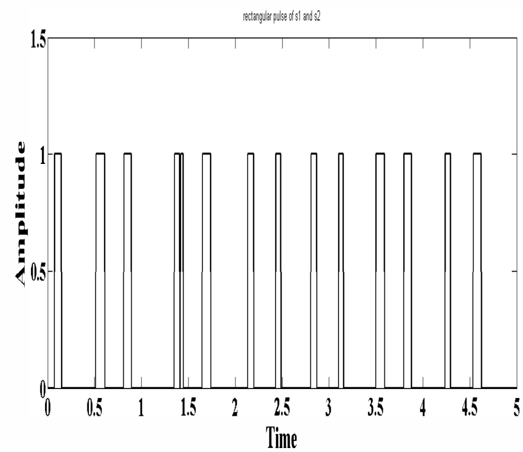

Fig. 6. Square pulse after removal of unwanted pulse from the signal

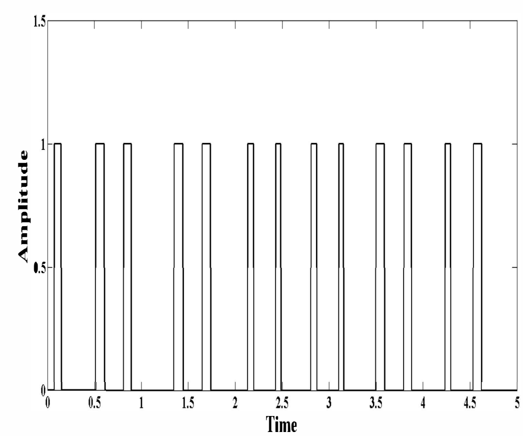

Fig. 7. Final obtained pulse

IV. Results

From the identified S1 and S2 pulses, the duration for S1 and S2 can be easily calculated by measuring the width of the pulse. For the Systolic period, we find the difference between the starting point of S2 pulse the end point of previous S1 pulse and for Diastolic period we find the difference between the Starting point of S1 pulse the end point of previous S2 pulse. The acquired signals are divided into three categories namely, Normal, Abnormal and Fetal. The normal consists of the signal acquired from healthy persons, abnormal signals consists of the PCG signals acquired from persons

TABLE I. PERFORMANCE OF ALGORITHM IN SEGMENTATION (RESULTS IN PERCENTAGE)

Nature of Signal	Number of Samples	cycles	Cycle detected(in %)
Normal	13	152	100
Fetal	18	402	96.70
Abnormal	27	396	95.95
Total	**58**	**950**	**97.0**

TABLE II. PERFORMANCE OF ALGORITHM IN AVERAGE TIME CALCULATION(IN MS)

Nature of Signal	S1	Systolic Period	S2	Diastolic Period
Normal	75.9	22.22	70.3	37.50
Abnormal	81.2	18.23	74.4	17.86
Fetal	150	40	220	30

suffering from different cardiovasular diseases as described in Section 2. The fetal category consists of signals acquired from abdomen of pregnant women having duration of 7-8 months of pregnancy.

Result are shown in two tables where Table I describes the segmentation results whereas Table II describes the timing information of the tested signals.

The given algorithm is then compared with NASN [4], Multiresolution Analysis using wavelet transform [5] and Morphology [7] based segmentation algorithm. Our Proprsed algorithm performs better than NASN and Morphology based segmentation algorithm when tested on abnormal and fetal Signals. Considering the complexity involved in Multiresolution Analysis using Wavlet tranforms our algorithm performs well. The performance is recorded in Table III as the percentage of heart sound components segmented when tested on the dateset recorded.

V. CONCLUSION

This work presents a novel algorithm for segmentation of S1 and S2 heart sound solely from Phonocardiographic signal as a source. The literature shows that earlier attempts for segmentation of heart sound signal require either too much calculation or don't shows good results when applied to abnormal pathological conditions due to the presence of strong systolic murmurs. The developed algorithm is based on adaptive thresholding technique showing an overall accuracy of 97%.

VI. ACKNOWLEDGMENT

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TABLE III. COMPARISION OF PERFORMANCE BY OUR AND OTHER PRE-EXISTING ALGORITHM.

Nature of Signal	NASN	MRWT	MORPHOLOGY	OUR
Normal	96.1%	98.7%	100%	100%
Abnormal	91.2%	95.3%	94.8%	95.95%
Fetal	92.4%	96.1%	95.1%	96.7%

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