TIME-FREQUENCY ANALYSIS OF HEART MURMURS IN CHILDREN

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1 Introduction

Examination of heart sounds (Phono-Cardiology) facilitates diagnosis of defects which manifest themselves as irregularities in the heart sound. Simply by listening to such sounds a trained physician can diagnose a variety of heart complaints, *e.g.* Atrial Septal Defects (ASDs) or colloquially "holes in the heart". Such defects may not manifest themselves on ECGs. Any abnormality in a heart sound is generically termed a murmur.

It is common practice for GPs to examine young children to check for the presence of a heart murmur. However, heart murmurs are relatively common in paediatric patients, the vast majority of these murmurs are non-pathological and the murmur will vanish, with no ill effects, over time. The problem arises in that GPs can detect the presence of a murmur, but cannot distinguish an "innocent" murmur from a pathological one. This classification task can be performed by specialist physicians simply listening to heart sounds [3]. The goal of this work is to develop computer based methods capable of performing the task of the trained physician, *i.e.* the classification of heart murmurs.

Heart sounds are clearly non-stationary signals and hence the natural analysis methods are those of time-frequency and/or time-scale. In this application there is no evidence to suggest that the analysis technique would benefit from a multi-resolution type analysis, so we concentrate on time-frequency, rather than time-scale, methods. The study of heart sounds via time-frequency analysis has been undertaken by other authors [*e.g.* 5].

2 Heart Sounds

Classically the sounds made by a healthy heart are conceived as being a nearly periodic signal consisting of four components [1]. These four parts are referred to as the first, second, third and fourth heart sounds. The first two heart sounds give rise to the familiar 'lub-dup' beating sound of the heart and tend to dominate the Phono-Cardio-Graphic (PCG) signals. The first heart sound is caused by the closure of the mitral and tricuspid valves. The second heart sound is due to the closure of the aortic and pulmonary valves. The third and fourth heart sounds are generally of much smaller amplitude and we shall overlook them for present.

The heart murmurs which interest us occur as additional components in the PCG signal, most often arising in the interval between the first and second heart sound. The murmur signal is often of a much smaller amplitude than either of the heart sounds. Further when measuring heart sounds there are often several competing sources of acoustic noise, probably the loudest is the sound of breathing. The result is that, whilst a trained physician can detect and classify a murmur, they are not apparent to others, (*e.g.* the first two authors).

Each beat is separated by an interval of the order of 1s, with each heart sound having a duration of roughly 50 ms. The interval between beats varies, even in a patient at rest, because of respiration. Similarly the exact nature of each beat varies from beat to beat. The result is a signal which is non-periodic, even though it has a

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repetitive character. Some form of cyclo-stationary model is appealing but a simple frequency modulated (FM) model is inappropriate, since if the interval between beats changes there is no evidence that the individual heart sounds vary, as an FM model would require.

Many murmurs are described as "whooshing" sounds and are believed to be derived from flow noise. We propose a model for a heart beat which may be expressed as follows:

$$b(t) = h_1(t) + m(t-\tau_m) + h_2(t-\tau_s) \tag{1}$$

where $b(t)$ is the sound due to a single beat, $m(t)$ is the heart murmur, and $h_1(t)$ and $h_2(t)$ are the first and second heart sounds. All the components of the model are assumed to have their epoch at $t=0$, and the delays τ_m and τ_s position the murmur and second heart sound appropriately. A deterministic model for $h_1(t)$ and $h_2(t)$ is assumed. This implies that the first and second heart sounds are perfectly repetitive, an assumption which is not fully justifiable. The murmur $m(t)$ is assumed to be an amplitude modulated random process, such that

$$m(t) = a(t) \times w(t) \tag{2}$$

where $a(t)$ is deterministic and $w(t)$ random. If the model described in (1) is averaged over many realisations, assuming the delays are fixed, then the average heart beat will not contain the murmur, *i.e.*

$$E[\ b(t)\ ] = h_1(t) + h_2(t-\tau_s) \tag{3}$$

Subtracting (3) from (1) then $b(t)-E[b(t)] = m(t-\tau_m)$. The fact that the data set is finite and the inevitable presence of modelling errors means that complete suppression of the heart sound will not occur, in practice one only achieves a relative enhancement of the murmur.

The measured heart sound, $x(t)$, can be expressed as :

$$x(t) = \sum_k b_k(t - T_k) + n(t) \tag{4}$$

where T_k is the epoch of the k^{th} beat (whose time history is denoted as $b_k(t)$) and $n(t)$ is an additive noise. To reflect the fact that the delays τ_m and τ_s are beat dependent we re-label them $\tau_{m,k}$ and $\tau_{s,k}$.

3 Methodology

The initial stage of the processing scheme is to segment the heart sound into individual beats and then within each to identify the individual heart sounds. If the Signal to Noise Ratio (SNR) is sufficiently high then the individual heart sounds can be located by examining a local estimate of the signal energy (if the SNR is too low for this relatively simple process to succeed then the SNR will almost certainly defeat any attempts to locate or classify a murmur). Once the sounds have been located a simple logic based routine pairs them into whole beats, whereupon they are segmented. With this set of beats one has to identify the first and second heart sounds, this task is most effectively performed with reference to a simultaneous ECG signal. The first heart sounds approximately coincide with the QRS complex in the ECG signal. This segmentation process results in two data sets, one containing all the first heart sounds and the second containing the second heart sound.

The methodology discussed here processes each heart sound individually and not each beat as implied by (1). This is because the variability in the timing between the first and second heart sound is large and by processing each heart sound individually we side-step this problem.

The segmented heart sounds are then aligned (using cross-correlations) and averaged to yield a mean heart sound. Once the mean heart sound is available then extraneous beats can be identified; these outliers may result from periods when the SNR is low, or isolated failures in the segmentation routine. The offending beats are then removed from the data set and the mean heart sound re-computed.

The two sets of heart sounds are then analysed using time-frequency methods. The poor resolution of the spectrogram makes it an unsatisfactory tool for this application. To obtain greater resolution we adopt Pseudo-Wigner-Ville distribution (PWVD) [2], which in continuous time we write as

$$W_x(f,t) = \int g(\tau)x(t-\tau/2)x(t+\tau/2)^* e^{-2\pi i f\tau}\, d\tau \qquad (5)$$

where * represents complex conjugation and $g(t)$ is a windowing function, in this work a Gaussian windowing function is used . It is well understood that the PWVD has good resolution in the time-frequency plane but suffers from problems due to the cross-terms it introduces [2]. Since the PWVD is a quadratic distribution, the law of superposition of signals does not hold; more specifically if $\bar{x}(t)$ is the mean of $x(t)$, then the $W_{\bar{x}}(f,t)$ is not the mean of $W_x(f,t)$. Indeed according to our model (1) if one were to form the PWVD of the average beat then the murmur would fail to appear, whilst in general the murmur would appear in the average of the PWVDs. A strategy for estimating the PWVD of the murmurs alone is to calculate the averaged PWVD of the residual signals, which we denote $R(f,t)$, such that

$$R(f,t) = \overline{W}_r(f,t) \qquad (6)$$

where

$$r(t) = x(t) - \bar{x}(t)$$

This process has an interesting interpretation in the context of cyclo-stationarity [4].

4 Results

Data was gathered using an electronic stethoscope (Bosch Est 40) from a selection of patients attending clinics at the SGH. All the PWVDs displayed in this section are plotted using a logarithmic grey scale, with darker shades indicating larger values. In each case the negative values in the plot are set to zero and 1% of the peak value of PWVD added to all points prior to taking logarithms.

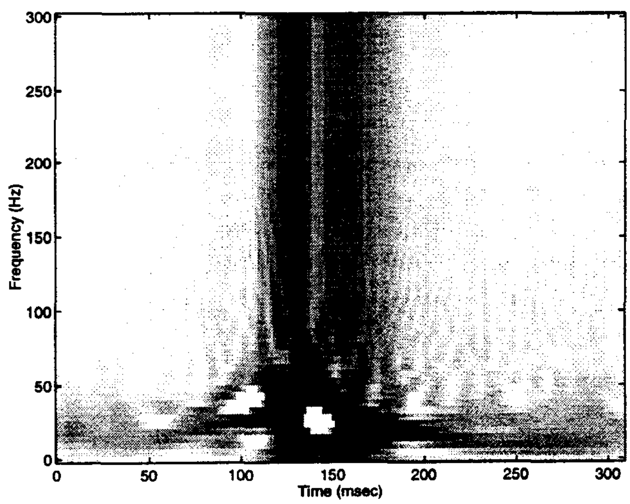

Figure 1 : Averaged PWVD for a normal first heart sound

Figure 1 shows the averaged PWVD of the first heart sound for a healthy patient. This shows a characteristic upward sweep component in conjunction with a later more rapid down sweeping component. Such structure has been noted by other authors [5].

We now illustrate the power of the method by examining the results from a patient diagnosed as having an innocent pulmonary flow murmur. Such murmurs are thought to result from normal blood flow in the pulmonary artery. In this case the recording has a good SNR and the murmur is unusually loud. It should be emphasised that in general most murmurs do not produce such clear results. A total of 50s of data, sampled at 600Hz was used. Figure 2a shows the averaged PWVD for the first heart sound. The murmur appears in this plot as the downward sweeping shadow starting at 160 Hz about 220 ms into the plot.

Figure 2b shows the result of calculating $R(f,t)$ for this same data set. The shadow is now enhanced relative to the residue of the first heart sound, making both the detection and classification of the murmur easier.

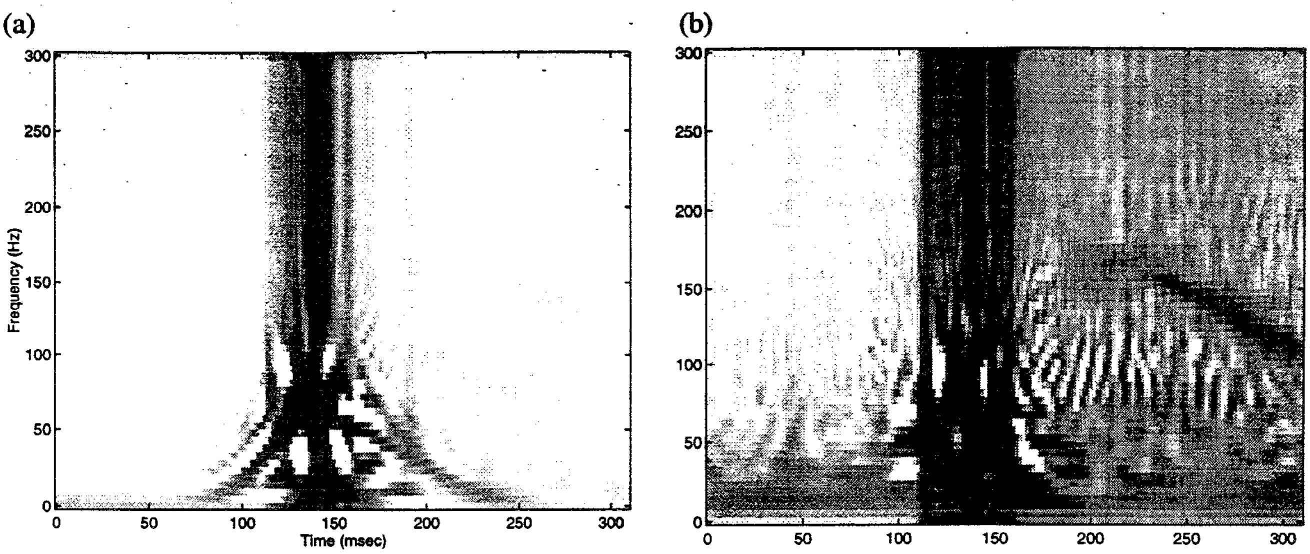

Figure 2 : (*a*) Averaged PWVD of the first heart sound with a pulmonary flow murmur (*b*) $R(f,t)$ of the same signal.

5 References

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