ROBUST ESTIMATION OF FETAL HEART RATE VARIABILITY USING DOPPLER ULTRASOUND

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ABSTRACT

Heart rate variability (HRV) provides important information about the development of the cardiovascular system in fetuses. This paper presents a new measure of fetal HRV that can be estimated using Doppler ultrasound techniques. This measure employs the multiple signal characterization (MUSIC) algorithm which is a high-resolution method for estimating the frequencies of sinusoidal signals embedded in white noise from short-duration measurements. We show that the product of the square-root of the estimated signal-to-noise ratio (SNR) and the variance of the frequency estimates is independent of the noise level in the signal. Since variations in the angle of incidence of the Doppler ultrasound beam effectively changes the input SNR, this measure of HRV is robust to the input noise as well as the angle of incidence. Presented analysis results validate the robustness properties and the usefulness of the HRV measure.

1. INTRODUCTION

Eight out of thousand live-born infants have some form of heart defect, making it the single most common class of congenital abnormalities. Identification of these cases during early pregnancy reduces risks by timely treatment and/or planned delivery at tertiary care centers. Ultrasonography is a safe, non-invasive, and cost-effective tool for monitoring fetal cardiovascular system through imaging and blood flow velocity measurements. Short-term temporal and spectral variability of fetal heart rate can be used for the assessment of cardiovascular development in fetuses during early human pregnancy [1]. Much of the prior work [2] in characterizing the spectral dynamics of the fetal heart rate from Doppler ultrasound blood flow measurements has been done using the beat-to-beat variability estimated from fast Fourier transform techniques. There are two fundamental problems associated with such techniques. First, the estimated beat-to-beat distances may vary with maternal breathing, fetal movements and noise present in the blood velocity waveforms. The second major drawback of current methods is that blood velocity measurements using Doppler ultrasound techniques and the associated HRV measurements are impacted by the variations of the angle of incidence of the ultrasound beam as well as the measurement noise. Consequently, HRV estimates obtained using current techniques are not reliable.

This paper presents a solution to both the problems described above. The rest of the paper is organized as follows. We first demonstrate that blood velocity waveforms can be modeled as a single sinusoid embedded in white noise over short intervals of time with the frequency of the sinusoid corresponding to the heart rate. We then use the MUSIC algorithm [4] to estimate the frequency and amplitude of the signal as well as the variance of the noise component in the signal. We describe the robust normalized measure of HRV in Section 2. Experimental results verifying the robustness properties of the measure is provided in Section 3 using simulated and real Doppler measurements. The conclusions are given in Section 4.

2. HRV ANALYSIS

2.1. Signal Modeling

Figure 1 depicts a portion of an umbilical arterial blood velocity waveform acquired from an eighteen-week old fetus. The temporal characteristics of the waveform suggest that this signal can be modeled as a pure sinusoid embedded in white noise over a short duration of time. Such inference can be further justified when we consider the autocovariance function of the signal. Figure 2 depicts the unbiased sample autocovariance function of the signal in Figure 1. Since the estimated covariance function appears to match the sinusoidal model, we modeled the fetal blood velocity waveforms as a pure sinusoid embedded in noise over short intervals. Based on this observation, we studied the variations in

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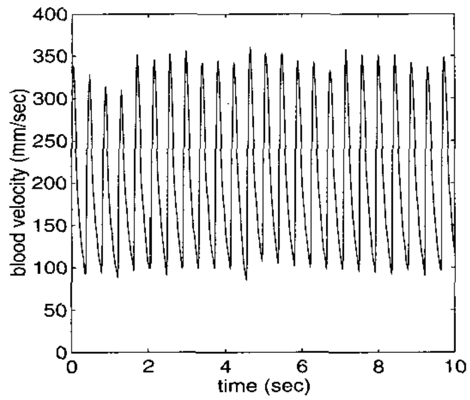
Fig. 1: chart


Fig. 1. Umbilical arterial blood velocity waveform of a fetus at 18 weeks of gestation and its autocovariance function.

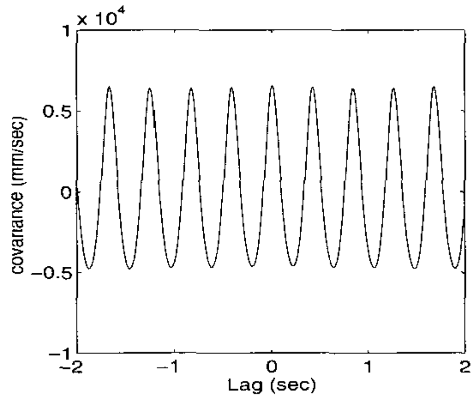


Fig. 2. Sample autocovariance function of the signal in Figure 1.

the characteristics of the velocity waveforms using the estimated frequency and amplitude values over time using the MUSIC algorithm. The amplitude and the frequency values evaluated by the algorithm correspond to the instantaneous peak blood velocity and heart rate values, respectively. The HRV can then be estimated by the temporal variability of these estimates over a period of time.

2.2. A robust heart rate variability measure

The instantaneous blood flow velocity using Doppler ultrasound is given by

$$\nu = \frac{cf}{2f_o \cos\theta}, \qquad (1)$$

where c is the velocity of ultrasound in the medium, f is the Doppler shift, f_o is the ultrasound frequency, and θ is the angle of incidence. Blood velocity waveform estimates are affected by variations in the angle of incidence and the environmental noise. We assume that the instantaneous peak velocity, frequency and the angle of incidence do not change significantly over short intervals of time corresponding to two to three heart beats. Consequently,we can model the appropriately sampled discrete-time blood velocity signal during i^{th} short interval at time n using the equation

$$\nu(i,n) = \alpha_i \cos(2\pi f_{\nu,i} n + \phi_i)\cos\theta_i + \xi(n), \qquad (2)$$

where $f_{\nu,i}$, α_i and ϕ_i are the instantaneous frequency, the amplitude and the phase values of the velocity waveform, θ_i is the instantaneous value of the angle of incidence and $\xi(n)$ is an independent, identically distributed (IID) noise process with zero mean value and variance σ_ξ^2. Similarly, we assume that the statistics of the noise process $\xi(n)$ changes slowly compared with the variations in the frequency and the velocity of the waveform.

Let, $\hat{\alpha}_i$, $\hat{f}_i$ and $\hat{\sigma}^2_{\xi,i}$ denote the estimated amplitude, frequency and the noise variance, respectively for the i^{th} block. We note that $\hat{\alpha}_i$ is an estimate of $\alpha_i \cos\theta_i$ in (2). Then, an estimate of the signal-to-noise ratio (SNR) for the signal in the i^{th} block is given by

$$SNR_i = \frac{\hat{\alpha}_i^2}{2\sigma^2_{\xi,i}}. \qquad (3)$$

From the above discussion, it is clear that the SNR is directly proportional to $\cos^2\theta_i$ when all the other parameters are fixed. For a complex-valued sinusoid embedded in white noise, Kay [3] has empirically shown that the mean-square error (MSE) in estimating the frequency using MUSIC is inversely proportional to the SNR of the input signal. This result can be extended to the case of real sinusoids to show that the MSE is inversely proportional to the square-root of the SNR. Therefore, we can eliminate the dependence of the MSE on the angle of incidence by defining a normalized HRV measure as

$$NHRV = \left[\frac{1}{N}\sum_{i=1}^{N} \|\hat{f}_i - \mu_f\|^2 \sqrt{SNR_i}\right]^{1/2}, \qquad (4)$$

where N denotes the number of segments in the waveform and μ_f is the mean heart rate computed over the whole waveform. In contrast, an estimate of the HRV given by

$$HRV = \left[\frac{1}{N}\sum_{i=1}^{N} \|\hat{f}_i - \mu_f\|^2\right]^{1/2} \qquad (5)$$

depends significantly on the measurement noise as well as the angle of incidence. Therefore, this unnormalized measure cannot provide reliable information on the cardiovascular system.

In our implementation of the normalized HRV, we estimated the parameters of the sinusoidal model from the autocovariance function. Since the harmonic frequencies of the heart rate are of relatively low strength, they are attenuated in the autocovariance sequence due to the squaring effect. Therefore, the autocovariance sequence provides more accurate estimates of the fundamental frequency or the instantaneous heart rate value of the waveform.

3. EXPERIMENTAL RESULTS

In this section, we verify the properties of the normalized HRV measure in two ways. The first set of experiments involved simulated Doppler signals. We demonstrate that the normalized HRV shows little dependence on the measurement noise level and the angle of incidence. The second set of experiments involved analyzing fetal umbilical arterial blood velocity waveforms.

3.1. Robustness to Angle of Incidence and Measurement Noise

A synthetic signal modeled as in (2) was generated with f_ν = 2.5 Hz, α_i = 10 cm/sec and a sampling frequency of 100 samples/second. The angles of incidence was varied from -80 to +80 degrees in steps of 2 degrees. One thousand independent experiments were performed at each angle θ and the experiments were repeated for noise variance values σ_ξ^2 = 0.001, 0.01, and 0.1. Each experiment employed 110 samples. The MUSIC algorithm was applied to estimate the instantaneous frequencies of the signal.

Figures 3 and 4 display the HRV and the normalized HRV estimated using (5) and (4) respectively, as functions of the angle of incidence for the three different noise levels. We can see from these results that the normalized HRV measure is relatively independent of the angle of incidence as well as the noise variances. This indicates that we may be able to use the normalized HRV measure to assess fetal cardiovascular function.

3.2. Evolution of Fetal HRV with the Gestational Age

In normal fetuses, heart beat change patterns with the gestational age can be explained using physiological considerations. By 8-10 weeks of the gestational age myofibrils appears in large numbers improving the contractility of the myocardium and hence decreasing the heart rate [1]. The autonomic nervous system develops first with the parasympathetic nervous system which matures around 15 weeks of gestation. The HRV remains approximately constant from 10 to 15 weeks and increases from 15 to 20 weeks of gestation. The increased heart rate variability with fetal age is due to the maturation of the parasympathetic limb of the autonomic nervous system around the 15 weeks of fetal life [5].

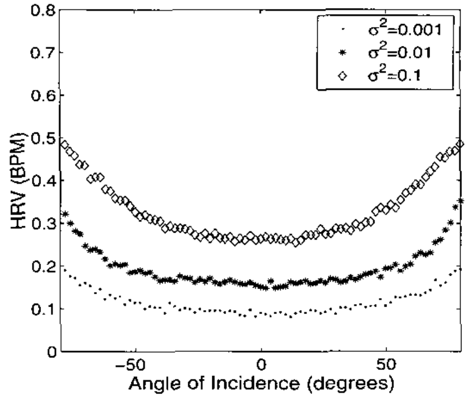


Fig. 3. Distribution of the HRV with the angle of incidence.

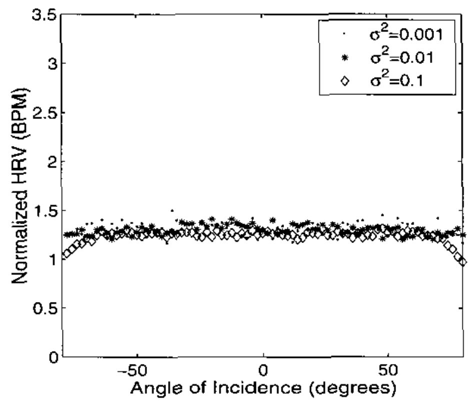


Fig. 4. Distribution of the normalized HRV with the angle of incidence.

In what follows, we show using ultrasound blood flow velocity waveforms collected from 108 fetuses between 10 and 20 weeks of gestational age that the evolution of the normalized HRV measure is similar to the behavior predicted based on the physiological considerations.

3.2.1. Data Acquisition

The fetal ultrasound measurements were collected at the Department of Obstetric and Gynecology, Academic Hospital Rotterdam-Dijkzigt, Rotterdam, The Netherlands. There were 108 women with normal singleton pregnancy between 10 and 20 weeks of gestation. The angle of incidence was kept below 20° and the flow velocity waveforms were obtained from the free floating loop of the umbilical artery. All Doppler studies were performed for 18 to 45 seconds with the women in the semi-recumbent position and during fetal apnea. Doppler waveforms were stored in sVHS tapes and flow velocity waveforms were reconstructed offline using

LabVIEW (National Instruments) software. The sampling frequency of the reconstructed waveform was 93.75 samples/sec.

The average short-term or beat-to-beat variability of the fetal heart rate is 2-3 beats per minute (BPM) [6]. Therefore, we assume that the instantaneous peak velocity, heart rate and the angle of incidence of the blood velocity waveform do not change significantly over short intervals of time corresponding to two to three heart beats. In this work, we estimated the instantaneous heart rate values using blocks of 110 samples. It corresponds to approximately three heart beats. The input to the MUSIC algorithm consisted of the autocovariance function of each block for the lags in the range [-35, 35]. The algorithm then estimated the autocovariance of this data for lags in the range [0,10], and from that estimated the frequency and the amplitude of a single real sinusoid in the model along with the variance of the noise component.

3.2.2. *Heart Rate Variability*

The variability of the heart rate estimates were characterized using the normalized root-mean-square deviation of the estimated mean heart rate over each waveform, calculated using (4). Figure 5 depicts the behavior of the observed heart rate variability with gestational age. The solid line (P_{50})

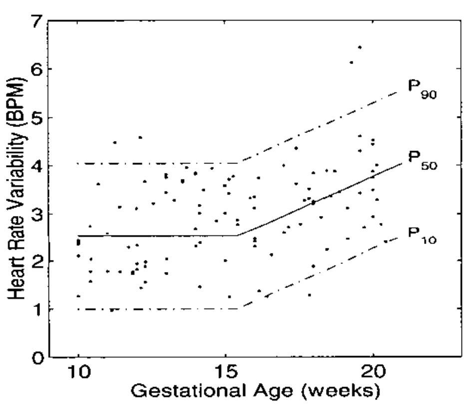


Fig. 5. Heart rare variability as a function of the gestational age.

shows the piecewise-linear least-squares fit of the HRV with the gestational age and is given by

$$\Delta_f(\omega) = \begin{cases} 2.5188 & ; 10.0 \leq \omega \leq 15.4 \\ 0.2710\omega - 1.6546 & ; 15.4 \leq \omega \leq 20.0. \end{cases} \tag{6}$$

In the first segment from week 10 to 15, variability was constant. In the second segment from week 15 to 20, heart rate variability gradually increased. The estimated root-mean-square deviation of the data from this model is 0.93 BPM. The 90% confidence intervals (P_{10} and P_{90}) of the estimation are marked by the dotted lines.

The estimated heart rate variability evolves with the gestational age as expected according to the physiological considerations. This, in conjunction with its robustness properties, makes makes us believe that the normalized HRV measure is more useful for assessing fetal cardiovascular function than its unnormalized counterparts.

4. CONCLUSIONS

Reliable estimation of frequency components of nonstationary data is difficult with Fourier transform-based algorithms when data lengths are short. The HRV measures based on the Doppler ultrasound measurements have been handicapped in the past because of their dependence on the measurement noise and the angle of incidence of the ultrasound beam. The normalized HRV measure based on frequency estimation using the MUSIC algorithm is robust to variations in the angle of incidence of the ultrasound beam as well as the measurement noise. This is a significant advantage of our approach over the current state-of-the-art in HRV measurement, and we believe that the normalized HRV is an excellent candidate for assessing fetal cardiovascular health.

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