Discrimination of Normal and At-Risk Populations from Fetal Heart Rate Variability

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Abstract

Using clinically measured intrapartum cardiotocography (CTG) data, the objective of this study was to compare the discrimination of fetal heart rate variability in the low frequency (LF, 30-150 mHz) movement frequency (MF, 150-500 mHz) and high frequency (HF, 500-1000 mHz) bands for two at-risk groups: fetuses experiencing either neonatal depression or metabolic acidosis. These parameters had statistically significant differences when comparing normal fetus and these indexed groups.

1. Introduction

Labour and delivery is routinely monitored electronically with sensors that measure and record maternal uterine pressure (UP) and fetal heart rate (FHR), a procedure referred to as cardiotocography (CTG). The objective of this monitoring is to detect the fetus at substantial risk of hypoxic injury so that intervention can prevent its occurrence.

In this study, we examined higher frequency FHR components ($>$ 30mHz), often referred to as fetal heart rate variability (fHRV). fHRV changes in labour and delivery due to uterine contraction. We wanted to determine whether the baseline or resting fHRV that occurs between uterine contraction could be used to distinguish high-risk fetuses (i.e., those with either neonatal depression, ND or metabolic acidosis, MA) from the normal fetuses that respond well to the insults of labour and delivery.

A preprocessing step in this study reduced the influence of maternal heart rate (MHR) interference on the FHR signal and the associated fHRV estimates. With this cleaner signal, we constructed short-term (1 min) autoregressive (AR) models at one second increments as in [1]. An important feature of these models is that we can construct them despite considerable missing data (a common problem with clinical CTG) in the analysis interval. Using the AR models, estimates of the power spectral density (PSD) were computed and the spectrum was integrated over low frequency (LF, 30-150 mHz), movement frequency (MF, 150-500 mHz) and high frequency (HF, 500-1000 mHz) bands to obtain three instantaneous components of fHRV. A new feature of this study was a smoothing step which low-pass filtered the instantaneous components. Then using overlapping 20 min epochs, the quiescent components of each band were computed from the 5^{th} percentile of their probability distribution functions.

2. Data

We used CTG from singleton, term pregnancies having no known congenital malformations, with $\geq$3 hours of tracing just prior to delivery. 5320 of the cases were normal while 10 experienced ND (Apgar 5 minute score ≤ 4) and 99 had developed MA (umbilical cord base deficit $\leq$ 12 mmol/L). The data come from two US hospitals, one that did routine umbilical cord blood gas measurements and another that routinely performed only the Apgar measurements. One of the hospitals also routinely recorded maternal heart rate in addition to FHR.

3. Methods

3.1. Preprocessing

The CTG data was recorded in a clinical setting, so it was subject to specific types of noise. The loss of sensor contact can temporarily interrupt the UP or FHR signals, and interference from the (much lower) maternal heart rate can corrupt the FHR. These both appeared in the signal as a sharp drop to much lower amplitude followed by a sharp signal restoration. We preprocessed the data to bridge these interruptions with linear interpolation. The FHR was then detrended by a high-pass filter with cutoff frequency 30 mHz, corresponding to the lower limit of the LF band of fetal HRV.

A portion of the data had maternal heart rate (MHR) consistently recorded in addition to the FHR. When the fetal monitor receives both signals, it attempts to detect MHR interference with the FHR and flags these occur-

rences. However, it was apparent from visual inspection of the tracings in our data that some interference had remained undetected. Furthermore, for many FHR-MHR pairs, interference was associated with a low heart rate difference ΔHR=FHR-MHR, while ΔHR was much larger for the more usual case of distinct heart rates.

Therefore to detect this residual interference, we modelled the probability density function (pdf) of ΔHR for individual recordings of MHR and FHR by a mixture of Gaussians estimated by expectation maximization (EM). The underlying assumption was that each Gaussian corresponded to a partitioning of ΔHR into interference and non-interference samples. Final interference detection was done by thresholding ΔHR where the posterior densities of the two conditions were equal. Interference sections detected this way were removed from subsequent HRV analysis.

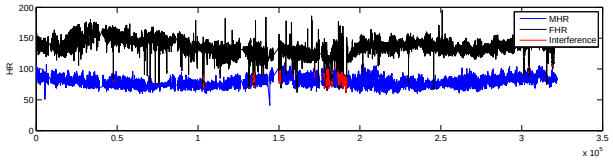
We found that random initialization of the Gaussians often led to local minima. Therefore we used a different initialization method that incorporated some a-priori knowledge about the ΔHR signal characteristics. We created a preliminary ΔHR partition with a hard threshold arbitrarily chosen from visual inspection (10 beats per minute,bpm). Then if interference were present, pair-specific model parameters were obtained by EM iteration which tended to converge to a mixture of two Gaussians, one with near-zero mean (using the above 10 bpm as an upper limit) and another with much larger mean. When a near zero-mean Gaussian was not selected at this stage, we attempted to model ΔHR once again, but with a mixture of three Gaussians. In this case EM was initialized using the two Gaussians found at the first stage and a third with zero mean and small variance (16 bpm^2). Detection was not attempted if a near zero-mean Gaussian was not found at this subsequent step, or if the difference in the two lowest means was below a threshold (chosen from inspection as 20 bpm) or if EM did not converge.

Fig. 1 shows a typical detection for one FHR-MHR pair. Overall, the vast majority of cases did converge within the constraints. Two Gaussian models were sufficient for 3689 of the tracings, three Gaussian models were required for 1330 cases, and 606 cases failed. We intend to refine the more ad-hoc aspects of this pre-processing detection step with a more principled Bayesian approach to initialization, constraint/threshold selection and model selection.

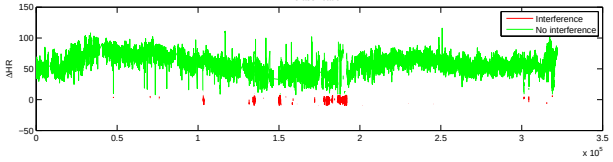
3.2. Power spectral density

We first estimated the fHRV using an autoregressive model of the CTG FHR signal to estimate the power spectral density (PSD), as described in [1]. The PSD was integrated over LF, MF and HF bands to obtain three instantaneous components of fHRV sampled at 1 s intervals. We then low-pass filtered the instantaneous components using a FIR-filter with a 0.04 Hz cutoff frequency to focus on the slower changes in fHRV and attenuate higher frequency noise in the estimates. Finally, using overlapping 20 min epochs, the quiescent (baseline or resting) component of each band was computed from the 5^{th} percentile of their probability distribution functions. This quiescent fHRV generally corresponds to periods of lower fHRV that occur between uterine contractions. We compared these estimates for normal cases and the two indexed groups.

(a)FHR, MHR



(b)ΔHR=FHR-MHR



(c)pdf(ΔHR)

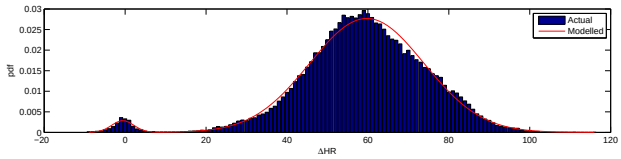


Figure 1. Detection of MHR interference on the acquired FHR signal (a) Distinct FHR (black) and MHR (blue) signals and segments of MHR interference (red). (b) Difference signal ΔHR=FHR-MHR in non-interference (green) and interference (red) segments. (c) Actual (blue histogram) and modelled (red line) ΔHR probability density functions. The units of time on the horizontal axis in (a) and (b) are 1 s samples. The units of heart rates are beats per minute (bpm).

4. Results

Fig. 2(a) shows that in the LF and MF bands, the ND group had consistently lower fHRV over time compared to the normal group, with 11/18 epochs showing statistically significant differences in the LF band and 13/18 epochs in the MF band. The HF band was also lower for the ND group, but not significantly so.

However, Fig. 2(b) shows that the metabolic acidosis group had higher fHRV than normal, especially in the last 90 min of labour where 5 epochs in both the LF and MF

bands and 3 epochs in the HF band had significant differences. These results are consistent with conventional clinical measures of variability [2].

The new preprocessing steps enhanced the quality of the signal sufficiently to improve discrimination. The smoothing step improved the discrimination with one more epoch being significant in each of the three bands for the MA comparison and where one (LF) and three (MF) more epochs were significantly different in the ND comparison. As well, in the MA comparison where maternal interference detection was included, discrimination was improved by one significant epoch in each of the LF and MF bands (these intermediate results not shown in the figures).

5. Conclusions

Our fHRV estimates have identified two at-risk fetal populations with very different characteristics: elevated (MA) or reduced (ND) fHRV. These sub-populations may have very different etiologies reflecting different neural mechanisms. These parameters are therefore useful discriminants of fetal state with promising potential for automated clinical decision-making.

References

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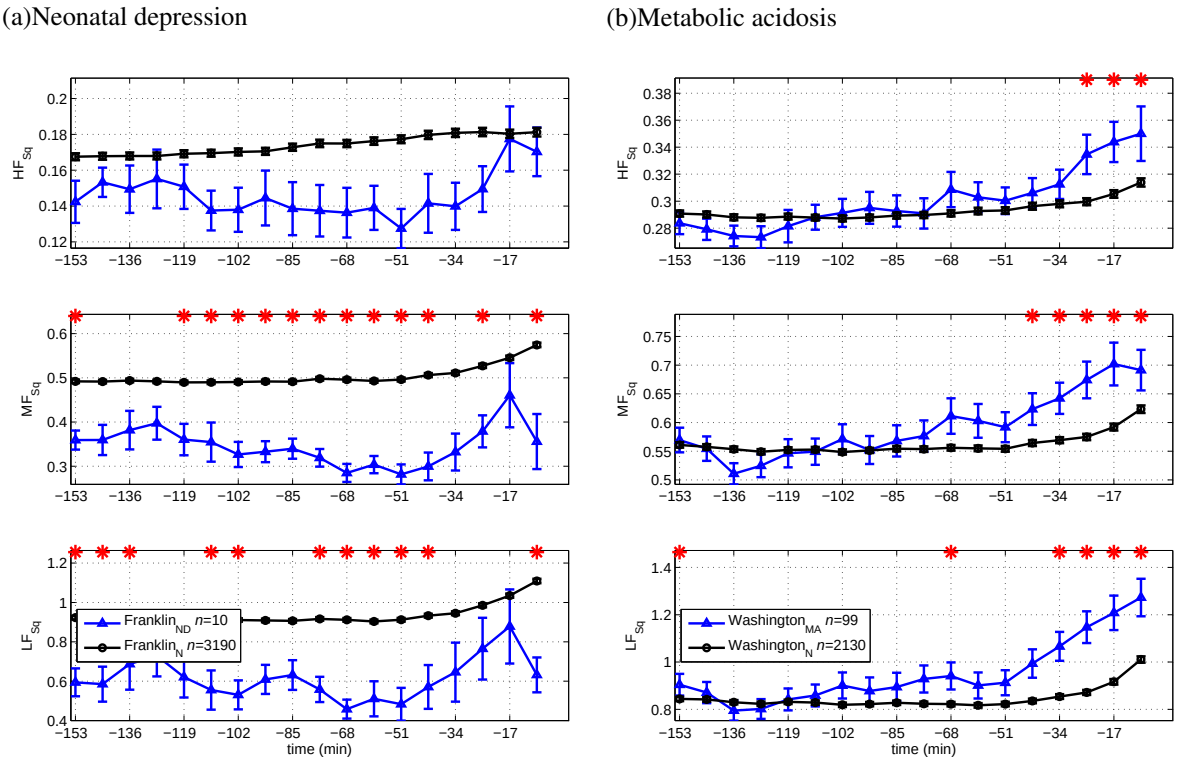

Figure 2. Quiescent fetal heart rate variability for low frequency (LF) and movement frequency (MF) and high frequency (HF) bands comparing normal fetuses to (a) fetuses experiencing neonatal depression (Apgar 5 minute score $\leq$ 4) (b) fetuses experiencing metabolic acidosis (umbilical cord base deficit $\leq$ 12 mmol/L). In each comparison, the means are plotted with bars indicating standard error and the red asterisks indicating statistically significant differences between normal (black circles) and indexed group (blue triangles) cases at that epoch (p <0.05, Kolmogorov-Smirnov distribution test). The time before delivery is indicated on the horizontal axis.