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PSD modifications of FHRV due to interpolation and CTG storage rate

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ABSTRACT

Cardiotocographic data provide physicians information about foetal development and permit to assess conditions such as foetal distress. An incorrect evaluation of the foetal status can be of course very dangerous. To improve interpretation of cardiotocographic recordings, great interest has been dedicated to foetal heart rate variability spectral analysis. It is worth reminding, however, that foetal heart rate is intrinsically an uneven series, so in order to produce an evenly sampled series a zero-order, linear or cubic spline interpolation can be employed. This is not suitable for frequency analyses because interpolation introduces alterations in the foetal heart rate power spectrum. In particular, interpolation process can produce alterations of the power spectral density that, for example, affects the estimation of the sympatho-vagal balance (computed as low-frequency/high-frequency ratio), which represents an important clinical parameter.

In order to estimate the frequency spectrum alterations of the foetal heart rate variability signal due to interpolation and cardiotocographic storage rates, in this work, we simulated uneven foetal heart rate series with set characteristics, their evenly spaced versions (with different orders of interpolation and storage rates) and computed the sympatho-vagal balance values by power spectral density. For power spectral density estimation, we chose the Lomb method, as suggested by other authors to study the uneven heart rate series in adults. Summarising, the obtained results show that the evaluation of SVB values on the evenly spaced FHR series provides its overestimation due to the interpolation process and to the storage rate. However, cubic spline interpolation produces more robust and accurate results.

1. Introduction

Cardiotocography (CTG) is one of the most diffused, non-invasive pre-natal diagnostic techniques, in clinical practice, to monitor foetal health, both in ante partum and in intra partum period. In some countries, it is a medical report with legal value. In CTG monitoring, foetal heart rate (FHR) and uterine contractions (UC) are simultaneously recorded by means of two probes placed on the maternal abdomen (a US Doppler probe for FHR signal and a pressure transducer for UC signal) [1,2]. It is important to underline that CTG is an indirect method to record FHR, and so it does not provide the same information carried by direct ECG. However, some authors state that recent US monitors are capable of calculating the heart rate enough precisely and reliably [3,4].

Nowadays, cardiotocographic data still represent, for physicians, an important source of information about foetal development and health. Moreover, it permits to early assess conditions such as foetal distress. However, interpretation of CTG recordings still lacks of complete objectivity and reproducibility, so more detailed information about the foetal status are searched and several analysis methodologies have been proposed in recent years. For these reasons, despite possible limitations in accuracy, we decided to study US-FHR signals.

Great interest has been dedicated to the variability of the FHR around its baseline, named FHR variability (FHRV). In particular, attention has been paid to FHRV spectral analysis, which could be a base for a more powerful, detailed and objective analysis [5–8]. Moreover, the power spectral density (PSD) of FHR seems to be the method that best recovers all the information present in the heart rate (HR) series [9].

Before any FHR signal processing, it is worth mentioning that FHR signals are calculated on an instantaneous basis (each FHR value is computed as inverse of the time between two consecutive R waves), so that FHR values are available only when new heart beats occur. So, FHR is intrinsically an uneven series and in order to perform classical spectral analysis additional signal processing is needed but it can strongly modify the resulting spectral information. The Fourier transform is widely used because it is a very straightforward method and makes it easier to compare results obtained in different laboratories [4]. Using fast Fourier transform

Abbreviations: bpm, beat per minute; CTG, cardiotocography; FFT, fast Fourier transform; FHR, foetal heart rate; FHRV, foetal heart rate variability; HR, heart rate; PSD, power spectral density; SD, standard deviation; SVB, sympatho-vagal balance; UC, uterine contractions.

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(FFT), the series should be interpolated and the resulting signal should be evenly sampled. Zero-order, linear or cubic spline interpolation can be employed. For example, to provide in output evenly sampled series, some commercial cardiotocographs (e.g. HP-135x) use a zero-order interpolation, that is each sample is held constant until the next heart beat occurs [10]. In particular, FHR and UC signals are internally recorded at a storage rate of 4 Hz (corresponding to a sampling interval of 250 ms). This process provides FHR data at fixed sampling time instants, by delaying some samples and adding some duplicates (in the case of missed or undetected heart beats). This is an efficient solution for computing classical parameters, such as baseline, accelerations and decelerations, but it is not suitable for frequency analyses because interpolation introduces alterations in the FHR power spectrum [9]. In the literature the introduced error in the evaluation of PSD has not been widely documented [11,12]. However, it is known that the interpolation process produces possible artifacts and an attenuation of the high-frequency components of the PSD that, for example, affects the estimation of the sympatho-vagal balance (computed as low-frequency/high-frequency power ratio), which represents an important clinical parameter [13,14].

Furthermore, some commercial software, employed for semi-automatic CTG analysis, in order to reduce required space memory and computational time, record CTG data at a lower storage rate (please see [15] for definition).

In order to evaluate possible errors in frequency analysis of FHRV signals due to the employed signal processing and in particular, to estimate the frequency spectrum alterations due to interpolation and CTG storage rates, in this work we simulated uneven FHR series (in the following let us call, without distinction, series or signal) and their evenly spaced versions, as proposed in previous works [1,15], and estimated sympatho-vagal balance (SVB) values by PSD. We considered do not analyse real signals since the real spectral characteristics are not known. To estimate PSD several methods there exist and the choice depends on the particular application (often an empirical approach is followed for method choice) [16]. Among most commonly employed methods to evaluated PSD, parametric and non-parametric, we can mention Short-time Fourier transform, Auto Regressive methods, Fast Recursive least square algorithms [6,8,16] and wavelet transform [17]. We decided to estimate the PSD for unevenly sampled FHR signals by means of Lomb method, extensively employed for uneven series in different research fields, and suggested for HR signals in adult subjects by Laguna et al. [9,18–20], since no explicit data replacement is made and PSD is calculated from only the known values. In order to compare obtained results, despite of high computational time, we adopted Lomb method also for PSD estimation of even FHR series.

This paper presents, in particular, a series of experiments to quantify the errors in FHRV spectral estimation due to zero-order, linear and cubic spline interpolations and sampling rates normally used to obtain typical CTG storage rates (2 and 4 Hz).

2. Materials and methods

2.1. Simulation of FHR signals

Synthetic FHR signals were artificially generated, via software, using a slightly modified version of a method proposed for adults by other groups [21,22] and already employed in previous works of the authors [1,15]. Following that procedure, an artificial R-R tachogram with specific power spectrum characteristics was generated. Considering that in FHRV different relationships between the LF and the HF bands are present, the following model parameters were adapted to resemble real foetal cases. LF and HF bands of the FHRV power spectrum were considered to lie between 0.04 and 0.2 Hz, and 0.2–1 Hz, respectively. LF/HF power ratio (considered a measure of the SVB) was fixed to 5 and Standard Deviation (SD) of HF band to 0.03. FHR mean value was initially set at 140 beat per minute (bpm), within the physiological range of 120–160 bpm; other values were chosen in the range 100–180 bpm. In addition, SD of FHR signal was heuristically set at 2 (for more details of the algorithm, please refer to the previous publication [1]). Finally, to obtain signals resembling other physiological conditions, accelerations were simulated by using Gaussian-like signal tracts (with SD heuristically chosen equal to 0.2); we adopted the classical definition: accelerations are transient increases of the FHR from the baseline of at least 15 bpm for at least 15 s [23,24]. For each FHR series, three accelerations were simulated.

All simulated FHR signals had duration of 25 min (see Fig. 1a and b for examples of simulated FHR signals).

Another algorithm previously developed by the authors [15] was employed to provide evenly sampled series of the simulated FHR signals. Artificial signals were interpolated by means of zero-order, linear and cubic spline interpolation and then sampled at 4 Hz (storage rate usually employed by the HP cardiotocographs) and decimated at 2 Hz (as done by some commercial software [25]).

2.2. Signal organization and processing

For each fixed FHR mean value, 350 FHR signals were generated and grouped in sub-sets of 50 signals; in particular, 50 uneven FHR signals (for simplicity, here also named series at 0 Hz) and 50 even FHR signals for each kind of interpolation (zero-order, linear and cubic spline) and storage rate (2 Hz and 4 Hz).

After that, to all these FHR series, 3 accelerations were added, as described in Section 2.1.

Regarding evaluation of PSD alterations, according to the literature [21], we considered SVB values as an index of eventual estimate differences. In particular, with regard to SVB computing, in order to highlight its common trend, average and standard deviation of all the SVB values were computed for each FHR signals sub-set. SVB values were computed for the same FHR signals without and with accelerations in order to evaluate the influence of these FHR alterations (accelerations) on SVB evaluation.

Furthermore, as in the real recordings the FHR mean value is not a priori known, for each kind of interpolation and storage rate, we then grouped the signals differently: we obtained sub-sets of 250 synthetic signals merging the sub-sets corresponding to different FHR mean values. The differences between SVB values evaluated on these sub-sets and ones estimated on the subset of 250 uneven FHR synthetic signals were computed.

2.3. The Lomb method

To estimate PSD of FHRV signals, we employed Lomb method [9,26].

The Lomb method is based on the minimization of the squared differences between the projection of the signal onto the basis function and the signal under study. Let $x(t)$ be the continuous signal under study and $b_i(t)$ an orthogonal basis set that defines the transform. The coefficients $c(i)$ that represent $x(t)$ in the transform domain are:

$$c(i) = \int_{-\infty}^{+\infty} x(t) b_i(t)\, dt \tag{1}$$

and these coefficients $c(i)$ are those which minimize the squared error $e(c_i)$ defined as [9]:

$$e(c_i) = \int_{-\infty}^{+\infty} (x(t) - c(i) b_i(t))^2\, dt \tag{2}$$