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# Empirical mode decomposition analysis of HRV data from patients undergoing local anaesthesia (brachial plexus block)

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## Abstract

Spectral analysis of heart rate variability (HRV) is used for the assessment of cardiovascular autonomic control. In this study, a data-driven adaptive technique called empirical mode decomposition (EMD) and the associated Hilbert spectrum has been used to evaluate the effect of local anaesthesia on HRV parameters in a group of 14 patients undergoing axillary brachial plexus block. The normalized amplitude Hilbert spectrum was used to calculate the error index associated with the instantaneous frequency. The amplitude and the frequency values were corrected in the region where the error was higher than twice standard deviation. The intrinsic mode function (IMF) components were assigned to the LF and the HF part of the signal by making use of the centre frequency and the standard deviation spectral extension estimated from the marginal spectrum of the IMF components. The optimal range of the stopping criterion was found to be between 4 and 9 for the HRV data. The statistical analysis showed that the LF/HF ratio decreased within an hour of the application of the brachial plexus block compared to the values at the start of the procedure. These changes were observed in 13 of the 14 patients included in this study.

Keywords: heart rate variability (HRV), local anaesthesia, empirical mode decomposition (EMD)

## 1. Introduction

The study of interbeat variations of the electrocardiograph (ECG) is known as heart rate variability (HRV). As the complex interaction between the sympathetic and parasympathetic nervous system is responsible for the beat to beat regulation of the cardiovascular system

the HRV can be used as a non-invasive technique to assess the autonomic influence on the heart (Task Force of the European Society of Cardiology and the North American Society of Pacing Electrophysiology 1996). In the frequency domain, three frequency bands can be distinguished in the spectrum of short term (2–5 min) HRV signals (Task Force of the European Society of Cardiology and the North American Society of Pacing Electrophysiology 1996). These components are termed as the high-frequency (HF) band (0.15–0.4 Hz), low-frequency band (LF) (0.04–0.15 Hz) and very low-frequency band (VLF) (0.003–0.04 Hz). The HF band is related to the respiration frequency and is also termed as ‘respiratory sinus arrhythmia’ (RSA). The LF band is not only an indicator of sympathetic tone but is also modulated by vagal activity whereas the physiological aspects behind the VLF band are not well established, and therefore, in short-term recording this band should be avoided (Task Force of the European Society of Cardiology and the North American Society of Pacing Electrophysiology 1996). The HRV indices such as the ratio of LF/HF power or the fractional LF power have been used to describe sympathovagal balance (Goldberger 1999).

In the recent past HRV has also been used in the field of anaesthesia with varying success in evaluating the depth of anaesthesia (Fan *et al* 1994, Sleight and Donovan 1999) and also evaluating the effect of different drugs on the cardiovascular system (Tanaka *et al* 2004). In this study HRV signals acquired from a group of patients undergoing local anaesthesia (axillary brachial plexus block) have been analysed to see the effect of local anaesthetic drug on the cardiovascular autonomic control. The inhibitory effect of the Brachial plexus block on the sympathetic and the endothelial activity has been demonstrated in previous studies (Landsverk *et al* 2006, Lehtipalo *et al* 2000, Szili-Torok *et al* 2002) by analysing the laser Doppler flowmetry (LDF) signals. Such information regarding the effect of local anaesthesia on autonomic nervous system could be helpful in anaesthetic management. Since ECG signals are routinely monitored during surgical procedures it will be more beneficial if the effect of local anaesthesia could be observed by using these signals.

The most commonly used methods for the spectral analysis of HRV are the non-parametric (Fourier based) and the parametric (autoregressive modelling) methods. The dependence on linearity and stationarity makes these methods non-optimal for the analysis of HRV data, which depends on time-varying phenomena such as respiration. Also as these methods produce time-average estimates of the power over the entire length of the record it is difficult to analyse transient changes that might be occurring in the HRV signal due to the introduction of anaesthetic drug.

In order to deal with the issues of nonstationarity and nonlinearity, in this study the empirical mode decomposition (EMD) and the associated Hilbert spectrum technique (Huang *et al* 1998) has been applied for the analysis of the signals. Unlike other techniques which use a predefined basis to decompose the data, EMD is a data-driven adaptive method in which the basis is calculated directly from the data. This ability allows EMD to provide a more compact and meaningful representation of the signals especially in the case of non-stationary and nonlinear signals (Huang *et al* 1998).

## **2. Methods**

### *2.1. Subject and protocol*

After obtaining approval from the Local Research Ethics Committee and informed written consent, 14 ASA I and II patients (7 males and 7 females) mean age  $50.6 \pm 20.7$  years

and mean weight  $67 \pm 15.3$  kg undergoing elective general surgery under local anaesthesia were recruited to the study. Patients with known cardiovascular and respiratory problems and those suffering from diabetes were excluded from the study. All the patients were null by mouth 12 h before the operation. In all cases the axillary approach was used for the brachial plexus block. A combination of 30 ml of 1% lidocaine and 29 ml of 0.5% bupivacaine with 1:200 000 part adrenaline was used as the anaesthetic agent. Anaesthesia was applied without the use of a neural simulator. One of the patients included in the study (patient 12) was categorized as anxious by medical staff before the start of the procedure. An AS/3 Anaesthesia Monitor (Datex-Engstrom, Helsinki, Finland) was used to collect lead II ECG signals from the patients. The monitoring started about 30 min before the start of the block and continued for approximately another 30 min after the surgery in the recovery ward. The ECG signal was digitized at 1 kHz sampling frequency as suggested in the literature (Abboud and Barnea 1995, McSharry *et al* 2003) using a PCMCIA 6024E 12-bit data acquisition card (National Instruments Corporation, Austin, TX). The duration of the ECG signals analysed in the study was  $3.75 \pm 0.82$  h.

## 2.2. Data preprocessing

The ECG R-wave peak detection was carried out using the wavelet transform with first derivative of the Gaussian smoothing function as the mother wavelet. The detection was carried out using wavelet scales  $2^m$ ,  $m = 2, 3, 4, 5$  and 6. The algorithm achieved an accuracy of 99.96% and sensitivity of 99.7% in the recorded ECG signals. After the R-wave detection, the *heart timing* signal (Mateo and Laguna 2000) was used for the HRV signal representation and also for the correction of missing and/or ectopic beats. The signals were resampled using cubic spline at a sampling rate of 4 Hz as recommended for HRV studies (Task Force of the European Society of Cardiology and the North American Society of Pacing Electrophysiology 1996). The performance of the *heart timing* representation and the beat correction algorithm has been validated in previous studies (Mateo and Laguna 2003, Shafqat *et al* 2007). The respiration signal was also estimated using the ECG-derived respiration (EDR) technique (Moody *et al* 1986, 1985). The physical influences of respiration result in amplitude variations in the observed ECG signal. In terms of the equivalent dipole model of cardiac electrical activity, respiration induces an apparent modulation in the direction of the mean cardiac electrical axis. The mean cardiac electrical axis often reflects the anatomic position of the heart in the chest. Moody *et al* (1985) presented a simplified technique to estimate the electrical axis direction from two-lead ECG signals. To obtain the direction of the electrical axis the baseline was removed from the ECG signal and then the area of each normal QRS complex was measured in each of the two leads of the ECG signal over a fixed window. The width of the window was estimated during the learning phase of the algorithm by determining the average width of the QRS complexes (starting from the Q-wave and ending with the S-wave) of the ECG signal. Since the window width is fixed, the area is proportional to the mean amplitude of the signal over the course of the window, hence to the projection of the mean cardiac electrical vector on the lead axis. The arctangent of the ratio of the area measure in the two leads gives the angle of the mean axis with respect to one of the lead axes. The axis direction measurement during the QRS complex provides one sample of the EDR signal per cardiac cycle. Cubic spline interpolation can be used to obtain the continuous EDR signal once its sample per cardiac cycle is estimated from the ECG signals. If only one ECG lead is available, the QRS area from that lead can still be used as an approximation to the respiratory signal. After these preprocessing steps the data were ready to be analysed with the EMD technique.

### 2.3. Empirical mode decomposition

In the EMD technique, the signal is first decomposed into a set of simple functions called the intrinsic mode function (IMF). More details about the IMF properties and extraction can be found in the literature (Huang *et al* 1998, 2003). A systematic way for extracting the IMF from a complicated dataset is known as sifting.

As in the EMD process the bases are adaptively generated from the data itself, the properties of the created IMF depends to a large extent on the stopping criterion that is used in the sifting process. In the literature so far, different techniques have been used to define the stopping criterion (Huang *et al* 1999, 1998, Rilling *et al* 2003). In this study the criterion suggested by Huang *et al* (1999) was used and the sifting process was stopped when the number of zero crossings and extrema remains the same for  $S$  successive sifting steps. In order to establish the confidence limit for the parameter  $S$  each dataset was decomposed using different values for the  $S$  parameter ( $S = 2-10, 15$  and  $20$ ) as proposed by Huang *et al* (2003). As in this case the number of IMFs produced by different  $S$  values was not the same, the mean and the standard deviation values were estimated by averaging the corresponding Hilbert spectrum, which has the same number of bins in frequency and temporal space pre-assigned (Huang *et al* 2003). The  $S$  value that gave the smallest squared deviation from the mean was considered optimal and was used as the stopping criterion for the sifting process.

### 2.4. Normalized amplitude Hilbert spectrum and instantaneous frequency error index

In order to validate that the estimation yields a physically meaningful instantaneous frequency for the IMF components, the normalized amplitude Hilbert spectrum (NAHS) introduced by Haung and Long (Haung 2005) was used and the time-dependent error bound was defined as

$$E(t) = [abs(\text{Hilbert transform}(y(t))) - 1]^2 \quad (1)$$

where  $abs$  in (1) is the absolute value of Hilbert transform of  $y(t)$  which represents the normalized IMF component. The normalization of the IMF components is done by using the so-called local amplitude which is obtained by joining all maxima of the IMF component using a cubic spline curve. The instantaneous frequency was considered to be incorrect at positions where the error index was higher than twice the standard deviation of the error. These values were corrected by using a model-based interpolating scheme purposed by Paulo *et al* (2003). After the correction, the marginal spectrum (2) of the IMF components was used to estimate the centre frequency (CF) and standard deviation spectral extension (SDSE) using (3) and (4):

$$h(\omega) = \int_0^T H(\omega, t) dt, \quad \omega = 2\pi f \quad (2)$$

$$CF = \frac{\int_{-\infty}^{\infty} f h(f) df}{\int_{-\infty}^{\infty} h(f) df} \quad (3)$$

$$SDSE = \left( \frac{\int_{-\infty}^{\infty} (f - CF)^2 h(f) df}{\int_{-\infty}^{\infty} h(f) df} \right)^{1/2} \quad (4)$$

where  $H(\omega, t)$  in (2) represents the Hilbert amplitude spectrum.

### 2.5. IMF component assignment to LF and HF bands

The IMF components were assigned to the LF or the HF band of the signal if the CF lay within the band limits and the  $CF \pm SDSE$  value was not more than 20% outside the boundary of that band. This approach of the IMF component assignment was validated with the help of the respiration signal which was estimated with the EDR technique (Moody *et al* 1985). For this purpose the respiration signal was also decomposed with the EMD technique and the IMF components corresponding to the major respiration components were estimated. From the IMF component of the respiration signal the CF and the SDSE were estimated. Using these values the spread of the frequency components in the respiration signal was calculated as  $CF \pm SDSE$ . This value was then compared with the  $CF \pm SDSE$  obtained from the IMF components of the HRV signal in order to obtain the components corresponding to the HF band of the signal. As in this study the respiration signal was estimated using a single lead ECG signal, which could cause a slight error in the estimation of the respiration frequency, a 10% difference was allowed in the  $CF \pm SDSE$  values obtained from the estimated respiration signal and the values obtained from the HF band of the HRV signal.

### 2.6. Statistical test

A non-parametric test (Wilcoxon signed rank test) was used to compare the parameter values estimated from the data obtained from the locally anaesthetized patients. The parameters from each patient were tested individually to check for differences before and after the application of the anaesthetic block. The statistical analysis was carried out using *SigmaStat 2.03* (Systat Software Inc., USA). The significance level was set at  $p < 0.05$  for all tests.

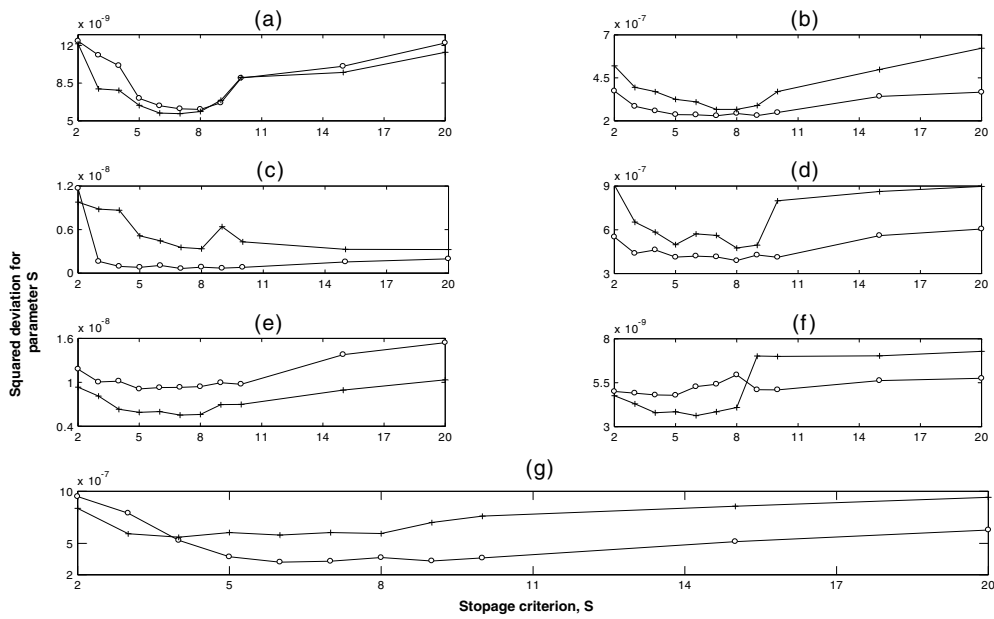
## 3. Results

### 3.1. HRV data from locally anaesthetized patients

In the analysis of the HRV data from locally anaesthetized patients the confidence interval for the EMD decomposition was found as described in section 2.3. The squared deviation results obtained from all 14 patients included in this study are presented in figure 1. The minimum value of squared deviation has occurred between  $S$  values of 4–9 which is quite close to the interval suggested by Huang *et al* (2003), 4–8. In fact only in one of our datasets the minimum has occurred at  $S = 9$ . For the rest of the datasets the minimum has occurred between the values of 4 and 8. Each dataset was decomposed using the optimal  $S$  value.

After obtaining the IMF components from data decomposition, normalization of the IMF components was carried out as described in section 2.4 and the error index associated with the instantaneous frequency was estimated using (1). The correction of instantaneous frequency was carried out with the help of the error index using the method mentioned in section 2.4. After the correction, the IMF components were assigned as the LF or the HF part of the signal by making use of the marginal spectrum and calculating the CF and the SDSE through (3) and (4), respectively. The marginal spectrum was obtained for every 5 min of data and the corresponding section of the IMF component was assigned to a particular band as mentioned in section 2.5.

Figure 2 shows the results obtained by comparing the  $CF \pm SDSE$  values obtained from the HF band of the HRV signal and the values obtained from the estimated respiration signal. Figures 2(a) and (b) show the HF band component obtained from the HRV signal and the major component of the estimated respiration signal, obtained from the EMD decomposition of the respiration signal, in normalized amplitude. The corresponding normalized marginal

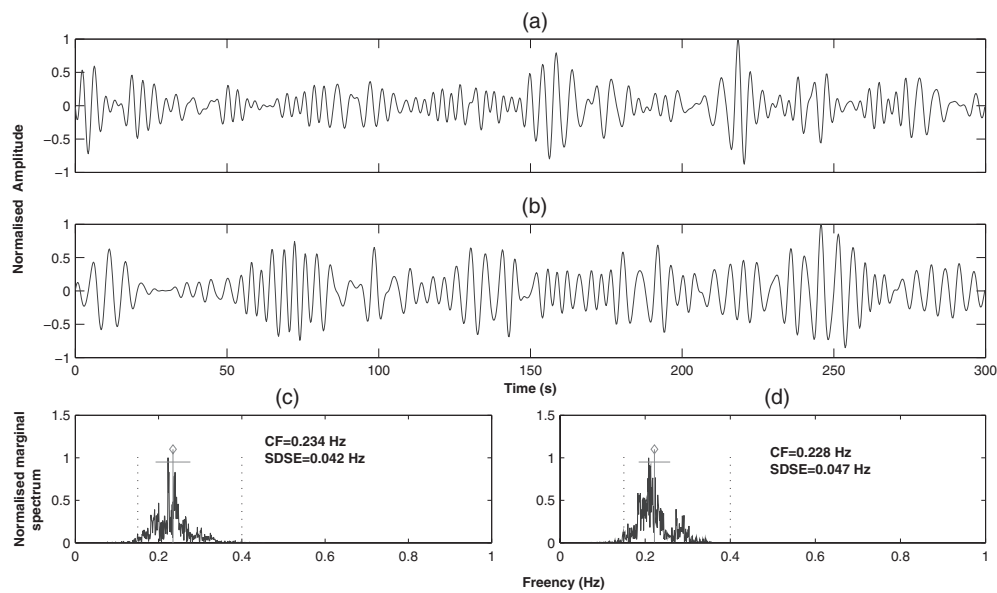


**Figure 1.** Squared deviation obtained for the stoppage criterion  $S$  values of 2–10, 15 and 20 in the case of *heart timing* data obtained from 14 locally anaesthetized patients included in this study. Each plot shows the results obtained from two patients, one with a circle and the other with a plus marker.

spectrum for these two signals are also presented in figures 2(c) and (d), respectively. The CF and the SDSE values shown in figures 2(c) and (d) indicate that the frequency spread ( $CF \pm SDSE$ ) obtained from the estimated respiration signal and the HF band of the HRV signal are within 10% of each other. This result suggests that the technique mentioned in section 2.5 might be suitable to separate IMF components into different frequency bands (HF and LF) of the HRV signal. The 10% difference in the frequency spread ( $CF \pm SDSE$ ) values obtained from the respiration signal and the HF band of the HRV signal was allowed because in this study the respiration was estimated using a single lead ECG signal. By measuring the respiration signal directly from the patient a better estimate of respiration frequency could be obtained which would possibly reduce the difference between these values. The marginal spectrum of the first six IMF components from two different 5 min datasets are presented in figures 3(a) and (b). The spectrum in figure 3(a) indicates that the first two IMF components belong to the HF band and the next three belong to the LF band. The situation is different in the case of figure 3(b) where the first component belongs to the HF band and the second and third component make up the signal in the LF region.

After the assignment of the IMF components into the HF and the LF bands the HRV parameters (LF/HF ratio and power related to the two bands in absolute and normalized units) were estimated. These values were averaged over a period of 1 min. The results presented in figure 4 show changes occurring in the HRV parameters after the application of the anaesthetic block in one of the anaesthetized patients included in this study. The LF/HF ratio (figure 4(b)) values increase after the application of the block (end of grey vertical block in the figure). After this increase the values decreased and remained low for some time compared to the values before the application of the block (data values before the start of the grey vertical block in

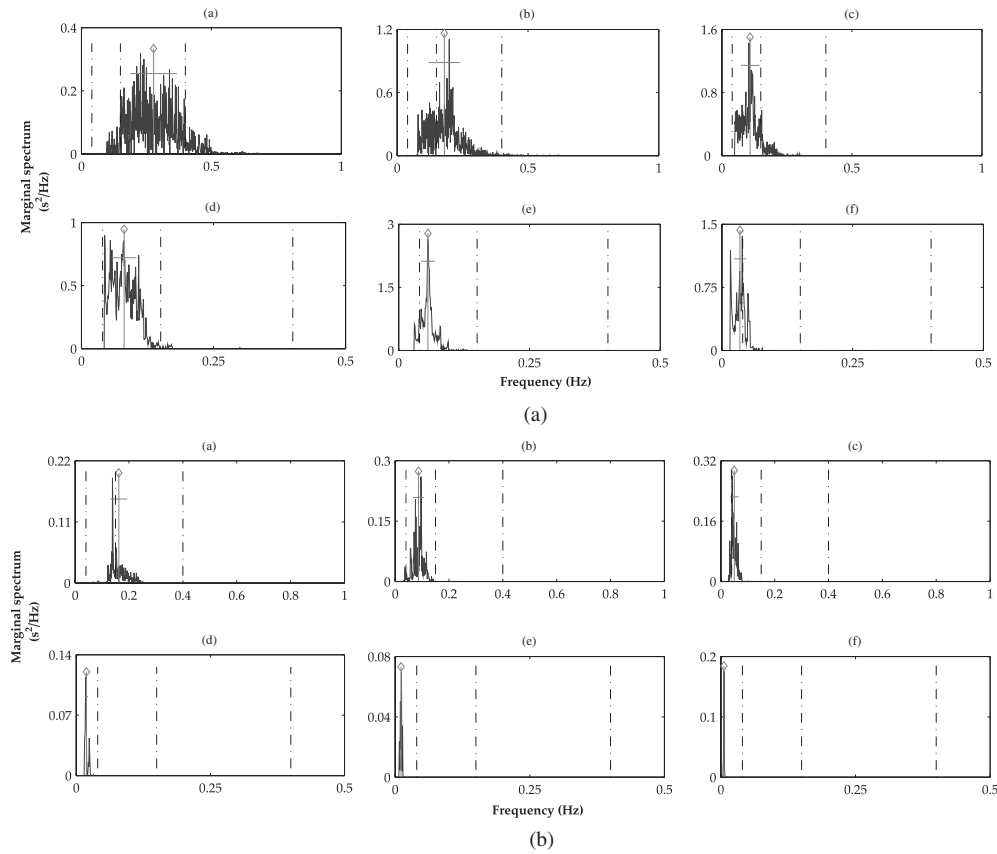




**Figure 2.** Results obtained by comparing the frequency spread ( $CF \pm SDSE$ ) in the estimated respiration signal and the HF band of the HRV signal; (a) normalized HF band component obtained with the EMD decomposition of the HRV signal; (b) normalized respiration component obtained from the EMD decomposition of the estimated respiration signal; (c) normalized marginal spectrum of the signal shown in (a); (d) normalized marginal spectrum of the signal shown in (b). In (c) and (d) the dash-dotted black lines represent the traditional HF rigid *a priori* frequency bands. The diamond mark represents the CF and the solid horizontal (grey) line indicates the  $CF \pm SDSE$ .

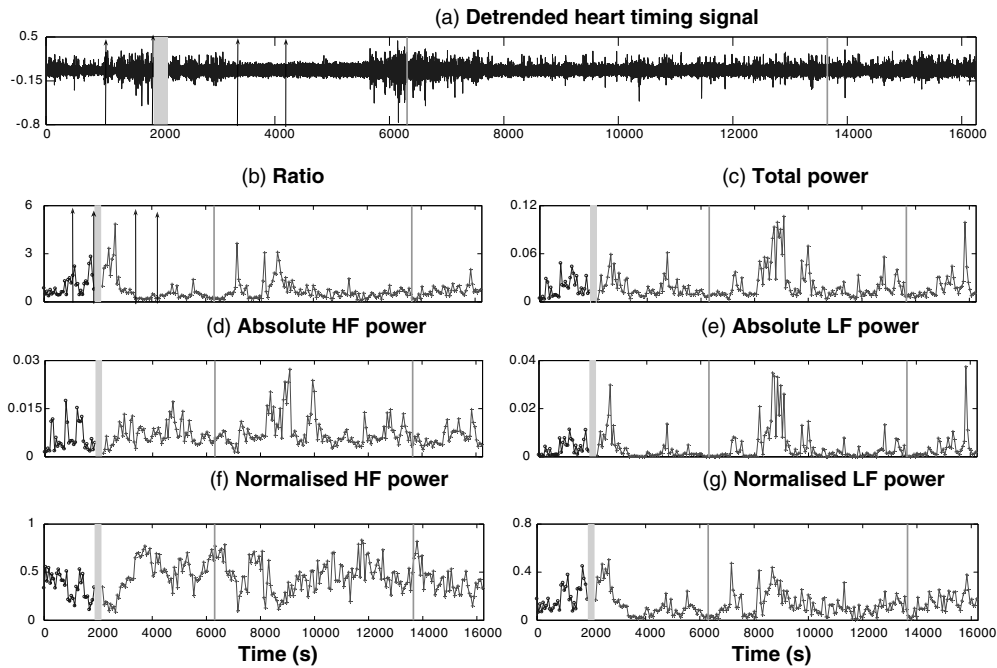
the figure). The increase in the LF/HF ratio values could be due to the presence of a small amount of adrenaline in the anaesthetic drug mixture and may also be due to patient's anxiety. The decrease in the LF/HF ratio values and the less variability shown by the values could be due to the anaesthetic drug indicating a shift in sympathovagal balance towards sympathetic impairment and/or vagal enhancement. The timing of the drop in the ratio value differs from patient to patient, but in each case the drop occurs within an hour of the start of the block. The large variability in the timing of the drop of LF/HF ratio values after the application of the anaesthetic block could be due to several factors such as patients' age, their body mass index and the initial haemodynamic state. As this drop could be considered as an indication of sympathetic blockage, observing the changes in HRV parameters during local anaesthetic procedures could help clinicians to decide about the time requirement for the occurrence of adequate anaesthetic block in individual patients. Estimating this time more accurately might also reduce the need of further anaesthesia during the surgery.

The increase in the HF band amplitude can also be seen by comparing the normalized marginal spectrum shown in figure 5. The 15 min data segments that were used to generate the marginal spectrum before (figure 5(a)) and after (figure 5(c)) the anaesthetic block were the same segments that were used in the statistical analysis. The timing of these segments is indicated in figure 4(b) by a pair of vertical arrows before and after the grey vertical box respectively. The marginal spectrum represents the cumulated amplitude over the entire data span (the 15 min data segments in the case of the results shown in figure 5) in a probabilistic sense. The frequency in the marginal spectrum has a totally different meaning from the Fourier spectral analysis (Huang *et al* 1998). The existence of energy at the frequency,  $\omega$ , means only



**Figure 3.** Parts (a) and (b) of this figure show the marginal spectrum of the first six IMF components (labelled (a)–(f) in each subfigure) for two different 5 min data segments from locally anaesthetized patients. In each plot the solid black line represents the marginal spectrum and the dash-dotted black lines represent the traditional LF and HF rigid *a priori* frequency bands. The diamond mark represents the CF and the solid horizontal (grey) line indicates the  $CF \pm SDSE$ . The CF and the SDSE are calculated using equations (3) and (4).

that, in the whole time span of the data, there is a higher likelihood for such a wave to have appeared locally. If the data exhibit nonstationary or nonlinear characteristics, the marginal spectrum will not be similar to the Fourier spectrum. The lack of a prominent peak in the HF region of the marginal spectrum shown in figure 5(a) means that there is less probability of energy being present in this frequency band (during the 15 min of data segment before the application of anaesthesia) as compared to the energy present in the LF region, which shows a prominent peak around 0.08 Hz. This is also indicated by the results presented in figure 4(b), which shows that the LF/HF ratio is high before the application of anaesthesia when compared to the ratio values in the 15 min of data segment after the application of anaesthesia. During and after the application of the anaesthetic block the LF/HF ratio values dropped, which means that the relative power in the HF band has increased. For this reason the HF peak has also become more prominent in figures 5(b) and (c), indicating a higher probability of the presence of energy in this band. The peak around 0.08 Hz in these two figures still corresponds to the peak in the LF region. In this study, at the start of monitoring approximately 10 min were

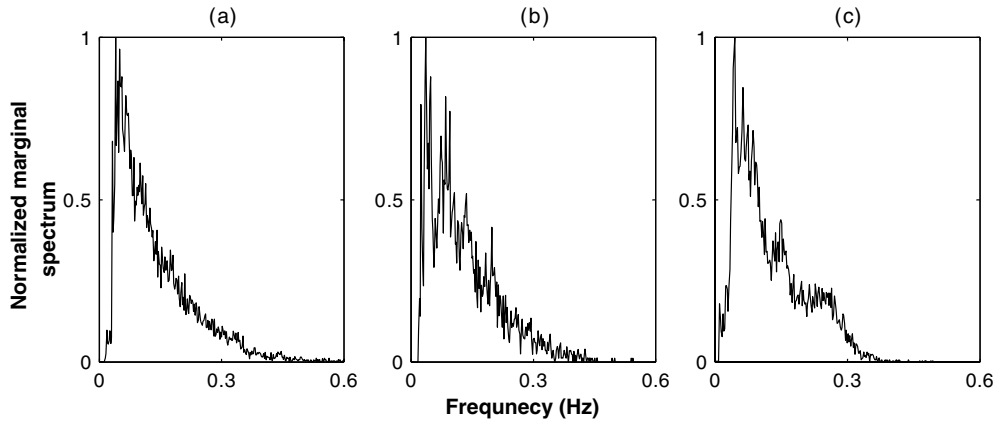


**Figure 4.** Results obtained from the EMD analysis of a patient undergoing a local anaesthetic procedure. Subplot (a) shows the *heart timing* signal from one of the locally anaesthetized patient which was used to estimate the HRV parameters shown in subplots (b) to subplot (g). In each plot the grey vertical block represents the time of the application of the block (anaesthesia) and the dashed grey vertical lines represent the start and the end of the surgery. The pair of vertical arrows in parts (a) and (b) show the data segment before and after the application of the block which was used in the statistical analysis. Data (in subplots (b)–(g)) with circle and plus markers represent the parameter values before and after the block application, respectively. The y-axis unit on subplot (a) is (s) and for the subplots (b, c and d) showing absolute power values is  $s^2 \text{ Hz}^{-1}$ .

allowed for the patients to get comfortable with the experimental setup in supine position so that stable hemodynamic conditions could be attained. After this initial 10 min the maximum duration of stable data that could be obtained from every patient included in the study was approximately 15 min. Therefore, data segments of 15 min were used for comparisons before and after the application of the anaesthetic block. The parameters estimated from all the patients included in this study are presented in the supplementary material available at [stacks.iop.org/PM/32/483/mmedia](https://stacks.iop.org/PM/32/483/mmedia).

### 3.2. Statistical analysis

The parameters related to the power of the signal (i.e. LF/HF ratio, total power (PT), power related to the HF band of the signal ( $\text{HF}_p$ ), the LF band power ( $\text{LF}_p$ ) and the HF and LF normalized power ( $\text{HF}_{p\text{norm}}$ ,  $\text{LF}_{p\text{norm}}$ )) were compared in order to see if their values differ significantly after the introduction of the anaesthetic drugs. The results obtained from the statistical analysis of these parameters are presented in table 1. Also, the  $p$  values for individual cases are presented. The median and percentile values of the parameters before and after the application of the block are presented in the supplementary material available at [stacks.iop.org/PM/32/483/mmedia](https://stacks.iop.org/PM/32/483/mmedia).



**Figure 5.** Normalized marginal spectrum for the data obtained from a locally anaesthetized patient; (a) spectrum of the data before the block; (b) spectrum of the data between the block; (c) spectrum of the data after the block.

**Table 1.** Statistical test (Wilcoxon signed rank test) results obtained from the EMD analysis of data from patients ( $n = 14$ ) undergoing local anaesthesia. From each patient six parameters values ( $LF/HF$  ratio, total power (PT) and high- and low-frequency band power in absolute and normalized units ( $HF_p$ ,  $HF_{pnorm}$ ,  $LF_p$ ,  $LF_{pnorm}$ )) before and after the anaesthetic block were compared. Significance level was defined as  $p < 0.05$ .

Patient no.	LF/HF ratio	PT	$HF_p$	$LF_p$	$HF_{pnorm}$	$LF_{pnorm}$
1	<0.001	0.008	0.008	<0.001	<0.001	<0.001
2	0.026	0.048	0.847	0.018	0.083	0.041
3	<0.001	0.004	<0.001	0.135	<0.001	<0.001
4	<0.001	<0.001	0.847	<0.001	0.002	0.001
5	0.026	0.002	0.001	0.295	0.135	0.041
6	<0.001	0.808	<0.001	0.015	<0.001	<0.001
7	<0.001	0.015	0.208	<0.001	<0.001	<0.001
8	<0.001	0.002	0.041	<0.001	<0.001	<0.001
9	0.978	0.855	0.380	0.806	0.524	0.599
10	<0.001	0.004	0.389	<0.001	<0.001	<0.001
11	<0.001	0.421	<0.001	<0.001	<0.001	<0.001
12	0.005	0.095	0.359	0.022	0.005	0.007
13	<0.001	0.454	0.055	0.107	0.007	0.005
14	0.007	<0.001	0.350	<0.001	0.003	0.003

From the results presented in table 1 it can be seen that the EMD analysis has been able to detect significant changes in the  $LF/HF$  ratio values after the application of the anaesthetic drug in 13 out of the 14 patients included in this study. The normalized power of the two bands (HF and LF) was also affected by the application of anaesthesia. Significant increase was detected in 11 patients for the normalized HF power ( $HF_{pnorm}$ ) and significant decrease was detected in 13 patients for the normalized LF power ( $LF_{pnorm}$ ) after the application of anaesthesia.

#### 4. Discussion

In this study the EMD decomposition technique along with the Hilbert transform was used to obtain the time-frequency distribution of data obtained from locally anaesthetized patients. In order to minimize the various sources of error that are related to EMD analysis, a systematic approach was used. The decomposition into the IMF components was carried out by establishing the confidence limit for the stopping criterion  $S$  and then using the optimal value (the value that gave the minimum squared deviation from the mean (see section 2.3)). The normalized amplitude Hilbert spectrum was used to estimate the error index associated with the instantaneous frequencies of the IMF components. The instantaneous amplitude and frequency values were corrected (see section 2.4) in the region where the error index values were more than twice the standard deviation. After the correction, the IMF components were assigned to the LF and the HF part of the signal by using the CF and SDSE, calculated from the marginal spectrum of the IMF components.

Various studies (Echeverría *et al* 2001, Souza Neto *et al* 2004) can be found in the literature that have used the EMD technique for the analysis of HRV signals; however, in this study a more thorough approach has been used. In this study a new method based on CF and SDSE has been proposed for assigning the IMF components into the HF and the LF bands of the HRV signal. In previous studies (Echeverría *et al* 2001, Souza Neto *et al* 2004) the IMF components were usually split into the HF and the LF bands by manually observing their frequency contents. The results produced in this way could be quite subjective. The use of CF and SDSE for assigning the IMF components into the HF and the LF bands as proposed in this study might be helpful in automating this process to some extent. Also in other studies only HRV signals were analysed with no consideration given to the respiration signal. As shown by the results (see figure 2) in this study, the information obtained by the EMD analysis of the estimated respiration signal was also used to validate the frequency contents related to the HF band of the HRV signal which is an important aspect of this study. It has been shown that under various conditions such as verbal communication and exercise, the effect of respiration could fall outside the traditional HF boundary (0.15–0.4 Hz) (Beda *et al* 2007, Bailon *et al* 2007). Therefore, it is strongly recommended that respiration information should be taken into account during the HRV analysis (Hirsch and Bishop 1981, Hayano *et al* 1994). Even though EMD is an adaptive data-driven technique which automatically splits data into different IMF components in the case where the respiration effect lies outside the traditional boundary of the HF band it might be very difficult to assign IMF components into appropriate (LF and HF) bands without giving any consideration to the respiration signal. Similarly, in order to obtain a reliable estimate of the instantaneous frequency using the Hilbert transform it is important to check the error index as defined in (1) (Haug 2005). Unfortunately, none of the previous studies which have employed the EMD technique for the HRV analysis have used this error check. In this study this error index has been estimated and a model-based interpolating scheme purposed by Paulo *et al* (2003) has been used to correct the values at the positions where the error index was higher than twice the standard deviation of the error. Studies where physiological parameters could be controlled more precisely would allow more accurate validation of this method of the IMF component assignment and also the usefulness of the method described in this study for the correction of the instantaneous frequency and the amplitude. The method of assigning IMF components into the LF and HF band using the CF and SDSE, the use of information from the estimated respiration signal to estimate the HF band of the HRV signal and the validation and correction of instantaneous frequency with the help of the error index are the main features which set this study apart from the previously published work on HRV employing the EMD technique.

As mentioned before, two distinguishable changes were observed in the LF/HF ratio values after the application of local anaesthesia. Adrenaline was considered to be the major factor behind the initial increase in the ratio values, while the drop observed after this transient increase was considered to be due to the effect of local anaesthesia on sympathetic/parasympathetic activity. The effect of adrenaline, present in the local anaesthetic mixture, on the heart rate and blood pressure during axillary brachial plexus block and other procedures involving local anaesthesia has been shown in previous studies (Dogru *et al* 2003, Ueda *et al* 1985, Cioffi *et al* 1985, Ohno *et al* 1988). In all these studies the increase in heart rate and peak concentration of adrenaline was obtained within 10 min of the application of the anaesthetic mixture. However, some conflicting results have also been presented in the literature. For instance, Meechan and Rawlins (1992) observed no significant changes in heart rate and blood pressure immediately after and 10 min following the injection of adrenaline-containing local anaesthesia for minor oral surgery. In other studies (Dogru *et al* 2003, Ueda *et al* 1985, Cioffi *et al* 1985, Ohno *et al* 1988), significant changes in the heart rate were observed approximately 5 min after the anaesthetic injection and the authors reported that the effect was short lived. So it is quite possible that the changes in heart rate values were not significant around and after the 10 min mark. Another important point, that most of the studies mentioned previously, was the fact that even the patient's anxiety due to the anticipation of having minor surgery or having the local anaesthetic injection might produce some adrenaline in the system. However, the adrenaline present in the local anaesthetic mixture was the major cause of increase in the plasma concentration of adrenaline and hence the increase in heart rate.

The effect of the local anaesthetic mixture (bupivacaine and/or lidocaine) used in this study on the autonomic nervous system activity during brachial plexus block has also been studied previously using LDF signals (Landsverk *et al* 2006, Lehtipalo *et al* 2000, Szili-Torok *et al* 2002). Landsverk *et al* (2006) analysed LDF signals from ASA I patients undergoing hand surgery. The block was performed using the transarterial technique and using 2.5 mg ml<sup>-1</sup> of bupivacaine, 10 mg ml<sup>-1</sup> of lidocaine and 6.25 µg ml<sup>-1</sup> of epinephrine. The first recording was made for an interval for 30 min before the application of the block and a second 30 min recording was carried out with a gap of 50 min after the application of anaesthesia. The authors observed significant reduction in relative amplitude in the frequency interval related to the endothelium and the neurogenic activities. The reduction of the endothelium-related oscillations could be a direct effect of the local anaesthetic on the endothelium, as has been described by several investigators (Turan *et al* 2000, Johns 1989, Minamoto *et al* 1997). Another explanation could be that alterations in sympathetic activity during brachial plexus block can have a direct influence on endothelial function (Eisenach *et al* 2002). The overall results from the Landsverk *et al* (2006) study indicate an inhibitory effect of the brachial plexus block on the sympathetic and the endothelial activity.

There are some differences in the results obtained from the LDF signal study and the results reported here. In our study opposite changes were observed in the LF (decrease) and the HF (increase) bands of the HRV signal. However, in the Landsverk *et al* (2006) study, the band approximately equal to the LF band of the HRV did not show any changes while the respiratory related band showed a significant increase. This case would also result in a decrease in the LF/HF ratio indicating a shift of sympathovagal balance towards vagal enhancement. A possible explanation of this difference could be due to the fact that in this study parameters were estimated continuously after the application of the block while Landsverk *et al* (2006) started the post-block measurement of the LDF signal 50 min after the application of the block. Also as in the LDF signal analysis, the changes were observed in the frequency range that nearly approximate the VLF band of the HRV signal; therefore, further analysis of the HRV

data from patients undergoing brachial plexus block could also include this frequency band. This will allow the opportunity to see whether the changes occurring at skin microcirculation level as detected through the LDF signal could have a significant impact on the VLF region of the HRV signal.

Deschamps *et al* (2004) also observed increase in power of the HF band and decrease in the LF/HF ratio values after the application of local anaesthesia, but in this case there were two problems. Firstly, they have analysed data from labouring patients undergoing epidural anaesthesia as compared to patients included in this study which received brachial plexus block for hand surgery. Secondly, they have used wavelet analysis with irregularly sampled signal (tachogram) and have associated the coefficients from specific wavelet decomposition levels to the two (HF and LF) bands of the HRV signal. Estimating HRV parameters in this way could result in error as due to irregular sampling of the data there could be a significant overlap of scales (frequencies) between levels (Milne and Lark).

The studies mentioned in this section provide some evidence in support of the two postulates that were made from the results obtained during this study. Firstly, the presence of adrenaline in the local anaesthesia mixture could cause a transient short-lived increase in the LF/HF ratio which was observed in almost all the patients included in this study. Secondly, the anaesthetic mixture would cause a sympathetic impairment and/or vagal enhancement resulting in a decrease in LF/HF ratio values. The results of this study suggest that changes in the cardiovascular dynamics due to local anaesthesia could be detected through the EMD analysis of HRV signals. The monitoring of parameters such as the LF/HF ratio would help the clinicians in providing a better indication of adequate anaesthetic block. As the time required by different patients to achieve a sufficient level of block might be different, by HRV monitoring the time duration between the application of the block and the surgery could be adjusted to suit individual patient's need. This might also reduce the requirement of additional anaesthesia during surgery.

## 5. Conclusion

The EMD analysis of the HRV data obtained from 14 patients undergoing local anaesthesia (brachial plexus block carried out with axillary approach) revealed that the LF/HF ratio (often used as a marker of sympathovagal balance) values showed a sharp peak almost right after the application of the block and then decreased significantly compared to the values approximately 15 min before the block (see figure 4(b)). Statistical tests (Wilcoxon, signed rank test) carried out to check for significant differences in the parameter values before and after the application of the brachial plexus block showed that the ratio values decreased significantly after the application of the block in 13 out of the 14 patients included in this study (see table 1). By looking at table 1 it can also be seen that the changes in other parameters especially the normalized power related to the two components (LF and HF) of the signal have shown strong correlation with the changes in the ratio values. At this point the cause of not detecting any significant changes in one of the patient included in this study is not clear.

By comparing the results obtained from the EMD analysis to the ones obtained from the traditional parametric analysis (Shafqat *et al* 2007), it can be seen that much better results were obtained using the EMD analysis. The results were better in the sense that the EMD technique was not only able to detect changes in the ratio values before and after the Brachial plexus block in more patients but it also provided better resolution than the parametric analysis which is quite beneficial in the analysis of transient changes occurring during the course of the study.



These results indicate that by appropriate and structured analysis of HRV data it might be possible to detect changes in the sympathovagal balance occurring due to the application of axillary brachial plexus block in patients undergoing minor upper limb surgery. In conclusion these results suggest that during brachial plexus block using a mixture of lidocaine and bupivacaine there is a noticeable and almost consistent change in the sympathovagal balance (LF/HF ratio decreases) which can be detected through HRV analysis.

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