Estimation of instantaneous venous blood saturation using the Photoplethysmograph (PPG) waveform

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Abstract. Non-invasive estimation of regional venous saturation ($SxvO_2$) using a conventional pulse oximeter could provide a means of obtaining clinically relevant information. This study was carried out in order to investigate the hypothesis that $SxvO_2$ could be estimated by utilising the modulations created by positive pressure ventilation in the Photoplethysmograph (PPG) signals. The modulations caused by the mechanical ventilator were extracted from oesophageal PPG signals obtained from twelve patients undergoing cardiothoracic surgery. The signals analysed in this work were acquired in a previous study. For the purpose of this analysis the raw PPG signal was considered to have three major components, AC PPG signal (cardiac related component), a static component or DC PPG signal (created mostly by the absorption of light by surrounding tissue) and the ventilator modulation component. These components were then used to estimate instantaneous arterial blood oxygen saturation (SpO_2) and $SxvO_2$ by utilising time-frequency analysis technique of Smoothed-pseudo Wigner-Ville Distribution (SPWVD). The results showed that there was no significant difference in the traditionally-derived (time-domain) arterial saturation and the instantaneous arterial saturation. However, the instantaneous venous saturation was found to be significantly lower than the estimated time-domain and instantaneous arterial saturation ($P=<0.001$, $n=12$).

Keywords: Photoplethysmograph (PPG), Regional venous saturation, Oxygen saturation, Pulse oximetry, Time-frequency analysis, Wigner distribution

1. Introduction

The pulse oximeter mainly relies on two conditions for the estimation of arterial blood oxygen saturation (SpO_2). Firstly, it requires adequate pulsatile blood flow at the region where the sensor is placed and secondly the difference between the spectral absorbance of the two haemoglobin species: oxyhaemoglobin and deoxyhaemoglobin (Allen; 2007). The pulsatile blood flow allows the spectral absorbance of the tissue to be measured at a maximum and minimum perfusion state. The spectral properties of the blood in motion, and hence its oxygen saturation can then be estimated using the changes in the spectral absorbance profile of the tissue during this transition. Previously, the pulsatile component (AC part of the PPG signal) of the PPG signal was considered solely due to the arterial blood motion however, in previous studies (Shelley et al.; 1993, 2005) it has been shown that motion of venous blood can also contribute to the PPG signal. This contribution in the pulsatile PPG component has been considered as a source of artefact which hinders the estimation of arterial oxygen saturation (Shelley et al.; 1993; Shafqat et al.; 2006). However, by using the two conditions mentioned above with this moving venous component the saturation associated with venous blood can also be estimated. The simultaneous accurate estimation of arterial and venous saturation utilising a conventional pulse oximeter could provide useful information about local oxygen extraction. This information may also provide useful insights in the analysis of tissue perfusion, indicating whether the supply of oxygenated blood is sufficient for the metabolic demands of the tissue in question. Non-invasive real time estimation of venous oxygen could also be clinically useful in monitoring and detecting important events such as early phases of shock.

There have been some attempts to estimate venous saturation non-invasively. Some earlier studies have used an external pressure cuff arrangement to facilitate the pulsatility in the venous blood (Yoxall and Weindling; 1997; Nitzan et al.; 2000; Echiadis et al.; 2007; Schoevers et al.; 2009). Even though the results were encouraging, these studies had some drawbacks. For instance, in some of these studies venous occlusion (required for the estimation of venous oxygen saturation) had to be separated by enough time as required by the tissue to reach equilibrium, and the venous saturation values could only be calculated at discrete time intervals (Yoxall and Weindling; 1997; Nitzan et al.; 2000). Secondly, this method could not be used for an extended period of time as this could lead to potential complications such as venous stasis, venous thrombosis, and interference with intravenous access. Similarly in some other studies, due to the way the pressure cuff was placed on the finger, it was not possible to ensure that each pneumatic pressure pulse would induce repeatable artificial blood pulse with similar pulse volume (Echiadis et al.; 2007; Schoevers et al.; 2009). This could have a significant effect on the performance of the sensor (Mannheimer et al.; 1997). Also, relying on an external source to generate modulations in the PPG signals could possibly restrict the areas which could be used for the estimation of regional venous saturation ($SxvO_2$).

In a more recent study (Walton et al.; 2010), the authors have taken advantage of the fact that the PPG waveform is influenced by both positive pressure ventilation (Natalini et al.; 2006) and peripheral venous pulsations (Wardhan and Shelley; 2009) and have estimated the venous saturation without using any extra source to create venous pulsations. The authors (Walton et al.; 2010) have analysed PPG signals from the oesophagus of ten patients undergoing coronary artery bypass

surgery and postoperative care in the intensive care unit (Kyriacou et al.; 2002b, 2003). They proposed time and frequency domain algorithms for the estimation of arterial and regional venous saturation. The frequency base (Fourier) method for the estimation of arterial and venous saturation as suggested by the authors in (Walton et al.; 2010) was also employed in other studies (Thiele et al.; 2011; Colquhoun et al.; 2012) to estimate venous saturation in the jugular vein. The results from these studies showed that the venous saturation obtained from the PPG signal could provide useful clinical information. However, the method requires further refinement such as obtaining cleaner spectral representation, possibly by decomposing the signal into different components (arterial and venous) before the estimation of arterial and venous oxygen saturation.

In this work a new method for extracting the venous modulations from oesophageal PPG signals is presented. After separating the AC (cardiac) PPG signal, the venous related component and the DC PPG signal were used for the estimation of SpO_2 and $SxvO_2$. Unlike previous studies (Colquhoun et al.; 2012; Thiele et al.; 2011; Walton et al.; 2010), time-frequency analysis technique of Smoothed-pseudo Wigner-Ville Distribution (SPWVD) was used for the estimation of saturation values. This would possibly allow better understanding of changes occurring in saturation values over the analysis period. As the modulations extracted from the PPG signal are caused by respiration/mechanical ventilation this signal could also be used for the estimation of respiratory/ventilatory frequency. In order to validate the accuracy of the respiratory frequencies estimated from the venous modulation signals, these frequencies were compared with the respiratory frequency values estimated from the Electrocardiograph (ECG) signal.

2. Methods

2.1. Subjects

Oesophageal PPG data, collected in previous studies (Kyriacou et al.; 2003; Kyriacou; 2013), from twelve ASA (American Society of Anaesthesiologists) 2–4 anaesthetised patients (aged 26 to 81) undergoing elective cardiothoracic surgery with positive pressure ventilation were analysed in this work. The data were acquired following local research ethics committee approval and patient consent. For each patient, approximately 100 minutes of ECG and PPG data was collected at a sampling rate of 100 Hz. The purpose of the previous studies Kyriacou et al. (2002b, 2003) was to investigate the oesophagus as a viable site for measuring blood oxygenation saturation Kyriacou et al. (2002b).

2.2. Venous modulation extraction from PPG

The venous modulation was extracted from the PPG signal by first detecting the peaks and troughs related to the AC (cardiac) component of the PPG signal. For the detection of the peaks and troughs the signal was filtered using a FIR (Finite Impulse Response) bandpass filter in order to isolate the cardiac component. An adaptive threshold was created by convolving the absolute values of the filter signal with a square window of 80 samples (which is slightly less than the width of the PPG pulse at a sampling rate of 100 Hz). A possible peak was detected in the region where the filtered signal crosses the threshold. A peak was considered a valid peak if it occurred

at least 0.2 s (refractory period) after the last valid peak and its amplitude laid within 0.2 and 3 times the average amplitude which was calculated from the last five correctly detected peaks. The troughs of the signal were detected using a similar technique.

After the peaks and troughs detection the upper and the lower envelope of the signal were constructed. For the upper envelope the peaks of the signal were connected using cubic spline interpolation while troughs were connected using the same technique to obtain the lower envelope. These upper and lower envelopes were then used to estimate the mean envelope, which after separating the DC (slow varying) component was used as the estimate of the ventilator modulation presented in the PPG signal. This ventilator modulation signal was used as the AC component in the estimation of venous saturation.

2.3. Arterial and venous blood oxygen saturation estimation

The estimation of both the arterial and the venous saturation was done in the conventional manner by calculating R (ratio of ratios) as presented in (1). Using the R value, the oxygen saturation was obtained by employing the linear relationship presented in (2) (Allen; 2007). In the case of the arterial saturation the AC_{red} and AC_{ired} part in (1) was calculated using the cardiac component present in the PPG signal while, in the case of venous saturation the ventilator modulation present in both the red and the infrared PPG signals was considered as the AC part. The time domain (traditional) saturation values were estimated by using every two seconds of data.

$$R = \frac{AC_{red}/DC_{red}}{AC_{ired}/DC_{ired}} \tag{1}$$

$$SpO_2 = 110 - 25 \times R \tag{2}$$

For the estimation of instantaneous arterial and venous saturations the time-frequency representation of the signals were obtained using Smoothed-pseudo Wigner-Ville Distribution (SPWVD), a quadratic transform from Cohen's class (Cohen; 1989). The time-frequency analysis was carried out using SPWVD instead of more basic Short-Time Fourier transform method as it provides better temporal and frequency resolution, by allowing independent adjustment of frequency and time window lengths. The SPWVD representation for a discrete sequence $s[n]$ can be expressed as (3) (Richman et al.; 1998; O'Neill et al.; 1999).

$$SPWVD_x[n,k] = \sum_{l=-P+1}^{P-1} h[l] \sum_{m=-Q+1}^{Q-1} g[m] \times r[n-m,l]\, e^{-j2lk\pi/M} \tag{3}$$

Where $r[n,l] = s[n+l]\, s^*[n-l]$ is the instantaneous auto correlation function and M is the length of the signal. In (3) a window $g[m]$ of length $2Q-1$ is used for smoothing in the time direction while smoothing in the frequency direction is carried out using the window $h[l]$ of length $2P-1$. In this study all the signals were analysed in the analytical form. The time and the frequency smoothing was carried out using a 5.05 second Gaussian window and a 10.05 second Hamming window respectively. The time and frequency windows length were chosen empirically (based on prior work (Shafqat et al.; 2009)) to have sufficient frequency resolution and time smoothing so that interference terms, a major cause of error

in quadratic (bilinear) representations, could be reduced. From the time-frequency presentation the instantaneous power of the signal could be calculated as shown in (4).

$$P_{inst}[n] = \sum_{k=k_{min}}^{k=k_{max}} |SPWVD_x[n,k]| \tag{4}$$

Where k_{max} and k_{min} in (4) are the minimum and the maximum frequency values associated with the signal. The power values can then be used to calculate R (see (1)) in similar manner as done in previous studies (Walton et al.; 2010; Thiele et al.; 2011; Colquhoun et al.; 2012) using the Fourier transform. However, in this case the R values and the resulting saturation values will be time varying functions.

2.4. Respiration rate estimation

As the modulations in the PPG signals were induced by ventilator/respiration, after extraction the mean envelope signal (see section 2.2) could also be used to estimate the ventilatory/respiratory frequency. In order to validate the accuracy of the respiratory frequency estimated from the venous modulation signals extracted from the PPG signals, it was compared with the respiratory frequency obtained from the ECG signals. The respiratory signal was estimated from the Electrocardiogram (ECG) signal using an ECG Derived Respiration (EDR) technique (Moody et al.; 1985, 1986). This technique uses the changes in the electrical impedance of the thoracic cavity, caused by respiration cycle, to estimate the respiration signal. The respiration signal derived from the ECG signal was bandpass filtered to restrict the frequency contents of the signals to be between 9 to 30 breaths per minute.

2.5. Statistical test

The statistical analysis was carried out using SigmaStat 2.03 (Systat Software Inc., USA). A non-parametric test (Wilcoxon, signed rank test) was used to check for statistically significant differences in the time domain arterial saturation, instantaneous arterial and instantaneous venous saturation values estimated using the oesophageal PPG data from the twelve cardiothoracic patients. A similar test was also carried out to compare the respiration frequency data obtained from the venous modulations and the ECG signals. The significance level was set at $P<0.05$ for all tests.

3. Results

3.1. Modulation extraction and saturation estimation

The peaks and troughs detection results from the oesophageal PPG signal from one of the patient's data included in this study are shown in Fig. 1. These peaks and troughs were joined, using cubic spine interpolation to obtain the upper and lower envelopes which are also shown in Fig. 1. These upper and lower envelopes were then used to calculate the mean envelopes (ventilator modulation) which are presented in Fig. 1 with thick lines. The results from the infrared PPG signal are shown in Fig. 1(a) while the results from the red PPG signal are shown in Fig. 1(b). The mean envelopes (ventilator modulation) were then subtracted from the raw PPG signals to obtain the arterial AC component. The cardiac and venous (ventilator modulation

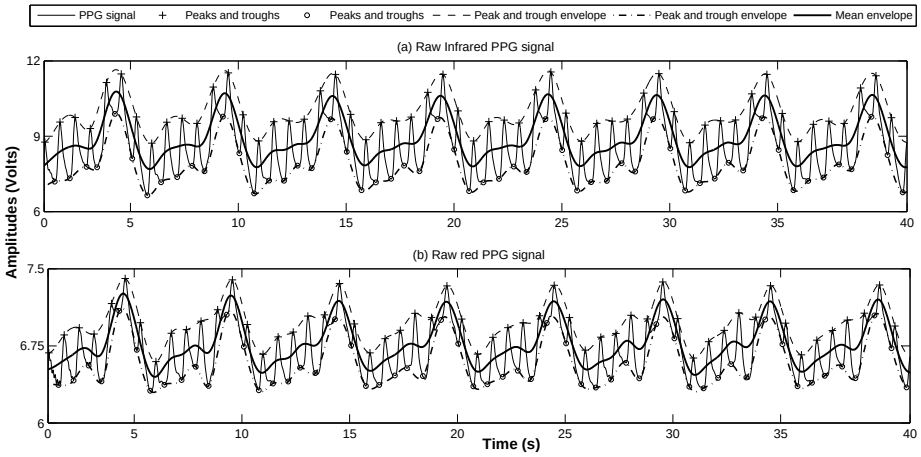

Figure 1. Peak and trough detection and the estimation of the mean envelope (thick line) in one of the data sets analysed in this study; (a) results from the infrared PPG signal; (b) results from the red PPG signal

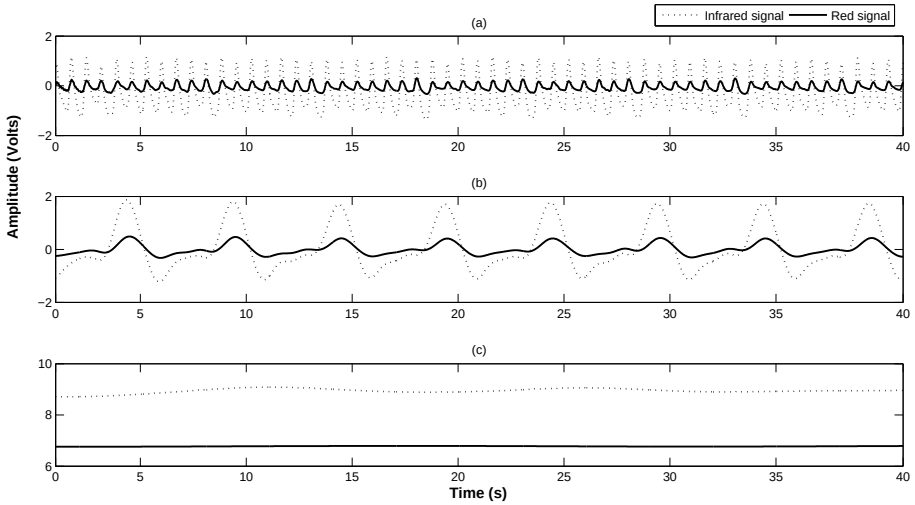

Figure 2. (a) Arterial AC PPG signals obtained after subtracting the venous modulation from the raw signals shown in Fig. 1; (b) AC signal related to the venous component (venous modulation); (c) DC PPG signals. In each part the infrared signal is presented with dotted lines while the red signal is presented with solid lines

signal) AC signals obtained from the raw signal and the corresponding DC PPG signals are shown in Fig. 2. The results shown in Fig. 2 also highlights the fact that subtracting the mean envelope from the raw PPG signal results in a cardiac AC component with no significant DC offset as required for the estimation of oxygen saturation using (1 and 2). Subtracting the upper (peak) or the lower (trough) envelope from the raw PPG signal would results in DC offset, negative offset (upper envelope subtraction) or positive offset (lower envelope subtraction), in the cardiac AC component.

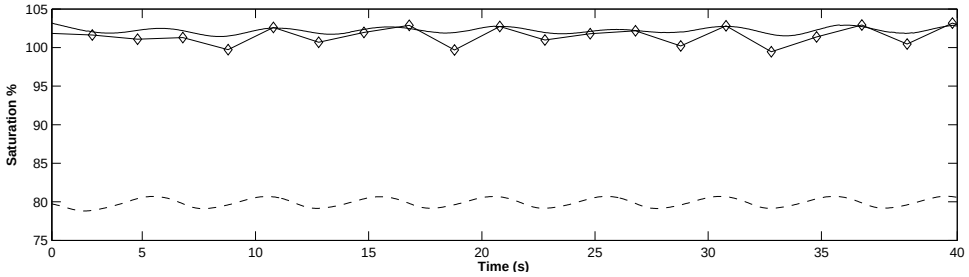

Figure 3. Oxygen saturation values obtained from the signals shown in Fig. 2. Traditional (time domain) arterial saturation is shown in a solid line with diamond markers. The instantaneous arterial saturation is shown as a thin solid line while the instantaneous venous saturation is shown in a dotted line

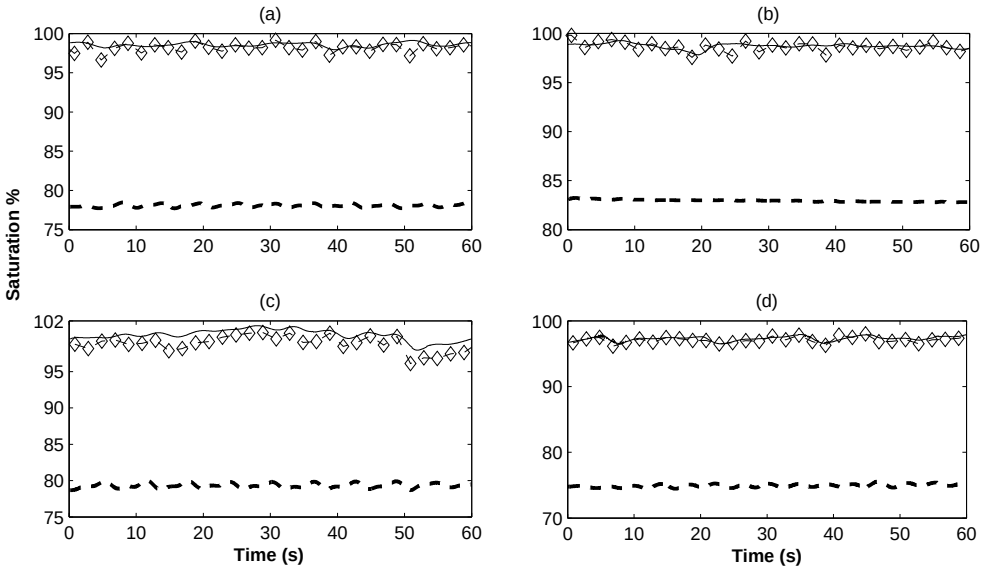

Figure 4. Instantaneous saturation results obtained from the data of four patients included in this study. In each subplot the time domain (traditional) arterial saturation values are shown with the solid line with diamond marker, the instantaneous arterial saturation values are shown in solid line and the instantaneous venous oxygen saturation values are represented by dotted line

The instantaneous arterial and venous saturations were then estimated using the time-frequency method as explained earlier in section 2.3. The traditional (time domain), instantaneous arterial and instantaneous venous oxygen saturations estimated using the signals shown in Fig. 2 are presented in Fig. 3. The time domain values were estimated after every two seconds.

The results presented in Fig. 3 shows that there is a close match between the time domain (traditional) and instantaneous arterial saturation (both values are around 102-103%) while, the instantaneous venous saturation values (around 80%) were lower than both the time domain (traditional) and instantaneous arterial oxygen saturation values. The results for arterial and venous saturation estimation obtained from the data of four of the twelve patients included in this study are presented in Fig. 4.

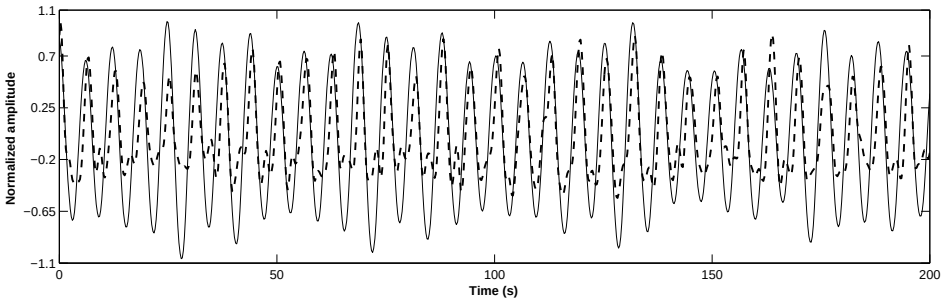

Figure 5. Normalised estimated respiratory signal from the ECG (dotted thick line) and the PPG (solid line). The PPG respiration signal is the mean envelope signal which is estimated as explained in section 2.2

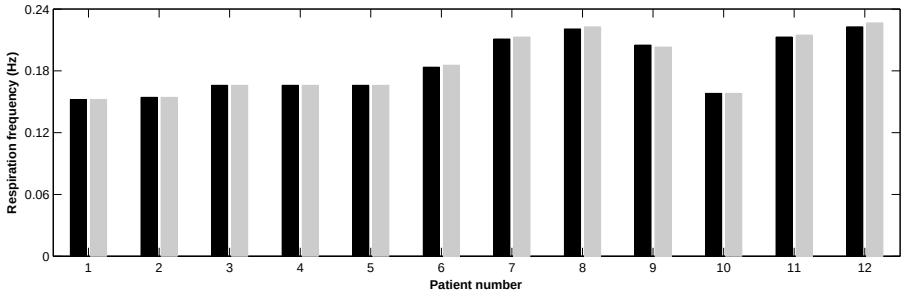

Figure 6. Respiration signal estimated from all twelve patients included in this study. The values obtained from the ECG are shown with black bars while the values obtained from PPG signals are shown with grey bars.

From the results shown in Fig. 4 it can be seen that the instantaneous venous saturation was lower than the time domain (traditional) and instantaneous arterial saturation in all four cases.

3.2. Respiration frequency estimation

The respiratory signals obtained from the ECG and the modulations extracted from the PPG signal from one of the patients included in this study are shown in Fig. 5. The signals in Fig. 5 are normalised with maximum amplitude for display purposes. For each signal the respiratory frequency was defined as the maximum peak in the power spectrum of the signal. The respiratory frequency estimated from the ECG and the PPG modulation signals from all twelve patients included in this study are shown in Fig. 6.

3.3. Statistical test results

To compare the arterial and venous saturation estimated from the PPG signal, a statistical test was carried out using the Wilcoxon Signed Rank Test. For the time domain arterial saturation the median and percentile (25% -75%) values were 101.2 (99.7 - 102.1) while the values for instantaneous arterial and venous saturations were 101.2 (99.5 - 102.4) and 79.3 (78.2 - 81.9) respectively. A graphical

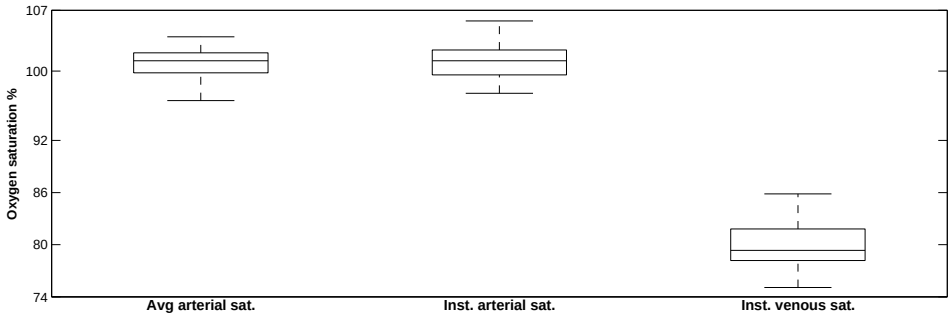

Figure 7. Box-and-whisker plots showing the percentile (25% -75%) and range for the traditional arterial saturation (Avg. arterial sat), instantaneous arterial (Inst. arterial sat.) and instantaneous venous saturation (Inst. Venous sat.) values estimated from the data of twelve patients included in this study

representation of the results is shown in Fig. 7. The results of the statistical test showed that there was no significant difference between time domain and instantaneous arterial saturations (P=0.501, n=12), while the venous saturation was significantly lower compared to both the time domain (traditional) and instantaneous arterial saturations (P=<0.001, n=12).

A similar test was also performed to compare the respiratory frequencies estimated from the modulations extracted from the PPG signal and the respiratory frequency estimated from the ECG signal. The median and percentile values for the ECG estimated respiration frequency were 0.175 (0.160 - 0.212) while the values for the PPG estimation were 0.176 (0.160 - 0.214). The statistical test showed that there was no significant difference between the two respiration frequency estimates (P=0.219, n=12).

4. Discussion

This study was carried out to validate the hypothesis that regional venous saturation could be estimated by utilising the PPG signal. In order to refine the method proposed in the previous study (Walton et al.; 2010), an algorithm was presented for the extraction of respiratory/ventilatory induced modulations from the PPG signal. The extracted modulations and the cardiac related PPG components were then used for the estimation of regional venous and arterial oxygen saturation values respectively (see section 2.3). The PPG data analysed in this work were collected from the oesophagus of twelve patients undergoing cardiothoracic surgery. The results from statistical tests showed that the regional venous saturation values, 79.3% (78.2% - 81.9%), were significantly lower than the time domain arterial saturation values, 101.2% (99.7% - 102.1%) and instantaneous arterial saturation values 101.2% (99.5% - 102.4%). The arterial saturation values estimated in this study were slightly higher than the logical maximum allowed (100%). The reason for this is the fact that the data was collected from a custom made probe which uses different wavelength LEDs (655 nm red and 880 nm infrared) as compared to the LEDs (660 nm red and 940 nm infrared) used in most commercial pulse oximeters. The photodetector also differs from the ones used in the commercial probes, however, it was chosen to minimise the difference in response at the two wavelengths (655 nm red and 880 nm infrared). Due to these differences

in the wavelengths and the sensitivity of the photodetector at the chosen wavelengths when equation (2) which is derived for 660 nm red and 940 nm infrared wavelengths is used for the estimation of oxygen saturation; values slightly higher 100% could occur. Further details and discussion regarding this issue and comparison of arterial oxygen saturation values estimated from this research device with blood gas analysis and Co-oximeter are presented in previous studies (Kyriacou et al.; 2002a, 2003).

The estimated venous saturation values of around 80% are within the expected physiological range of venous saturation. These values could not be considered to be solely due to noise as they are below 85% which is the saturation obtained when the signals at the two wavelengths are identically modulated, as in the case of pure noise. In fact, the instantaneous venous saturation estimated in this study is in close agreement with the venous saturation values obtained in previous studies. For instance, Walton *et al.* (2010) have found venous saturation in the oesophagus to be in the range of 90% to 65% with a median value around 80%. Taitelbaum *et al.* (2000) have reported venous saturation of 86.2% ($\pm$4.1) for the finger and 80.0% ($\pm$8.2) for the hand. Thiele *et al.* (2011) have found the venous saturation mean values of 75%, 80%, and 80% at the antecubital, external jugular, and internal jugular veins, respectively.

There are two major differences in this work and the previous studies (Colquhoun et al.; 2012; Thiele et al.; 2011; Walton et al.; 2010). Firstly, venous related modulations have been separated from the raw PPG signal before the estimation of saturation values. Secondly, instead of using the Fourier method, a time-frequency analysis has been used to estimate instantaneous arterial and venous saturation values. The separation of raw signals into separate components is desirable as the amplitude of the DC PPG signal and sometimes even the modulation caused by ventilator/respiration could be quite large compared to the AC (cardiac) component of the PPG signal. This could cause power leakage in the spectrum of the raw PPG signal. Hence, the value at the cardiac component could be affected by the DC and/or the modulation component which in turns could be affected by the large DC component. This could possibly be the reason for the large standard deviation in the venous saturation values observed by Walton *et al.* (2010). Also, the variation is respiration frequency could adversely affect the readability of the raw signal spectrum as suggested by Thiele *et al.* (2011). Walton *et al.* (2010) have also proposed a time domain method for the estimation of instantaneous saturation values. However, their method suffers from stability issues near the trough of the signal and required subjective thresholding in the vicinity of the unstable values. The results in this study have shown that stable instantaneous saturation values could be estimated with the help of time-frequency analysis technique without any subjective thresholding. By avoiding these instabilities and therefore the need of thresholding the instantaneous saturation values could possibly provide better understanding of any transient changes occurring in the saturation which could provide useful clinical information. In this study Smoothed-pseudo Wigner-Ville Distribution (SPWVD) was preferred over the more basic forms of time-frequency analysis (Short-time Fourier transform) as it allows independent control over the time and frequency resolution. Other time-frequency analysis methods such as wavelets and Hilbert-Huang transform should also be applicable but comparison of such techniques was considered to be beyond the scope of this initial work.

In order to provide further evidence that the respiration/ventilator is the major component of modulations extracted from the PPG signals, the respiratory/ventilatory

frequency calculated from this signal was compared with the respiratory/ventilatory frequency derived from the ECG (see section 2.4). The results showed that there was no significant difference between the calculated values of the respiratory/ventilatory frequency obtained from the PPG modulation and the ECG signal (see Fig 6).

There are some limitations in the work carried out in this study. The extraction of the modulating signal from the PPG relies on the accurate detection of the peaks and troughs in the raw PPG signal. If the algorithm fails to detect peaks and troughs, due to artefacts in the raw signal or overlapping of cardiac frequency and peak respiration frequency, it will not be able to extract an accurate modulating signal. This will have an adverse effect on the accuracy of the estimated saturation values. Also, another limitation of our study was the lack of validation of the estimated venous saturation values against a gold standard, such as blood gas analysis or co-oximetry. These values were not available to us as they were not collected during the clinical trials conducted in previous studies (Kyriacou et al.; 2002b, 2003). However, as the saturation values estimated in this work are within the range of saturation values obtained in previous studies (Nitzan et al.; 2000; Walton et al.; 2010; Thiele et al.; 2011) it can be concluded that the method presented in this study could provide further refinement for the extraction of venous related modulation caused by the ventilatory/respiratory effort from the PPG signal and its possible use for the estimation of regional venous saturation. If accurate estimation of regional venous saturation could be obtained with such a technique then it might be possible to combine the information from several different sites (finger, oesophagus, earlobe, forehead *etc.*) to generate a surrogate for true mixed venous saturation. Also, due to predominate sympathetic innervation, regional venous saturation from extremities such as finger might be helpful in detection of early sign of shock. It is also important to note that the PPG signal also contains a cardio-synchronous venous component but usually this component has a relatively small amplitude compared to the rest of the signal. In a recent study (Yousefi and Nourani; 2015) it has been shown that arterial saturation estimated at different breathing rate are not significantly affected by this cardio-synchronous venous component. This component is also not taken into account in the estimation of SpO_2 by most commercial pulse oximeters. However, PPG signals from certain measuring sites such as forehead have shown to exhibit large amplitude cardio-synchronous venous pulsation which has been reported to cause error is the estimation of SpO_2 (Shelley et al.; 2005). Nevertheless, this issue can be resolved by applying small pressure to the sensor (Shelley et al.; 2005). As the PPG signals analysed in this work did not exhibits large amplitude of this component and the arterial oxygen saturation estimated from these signals has been validated against blood gas analyser and CO-oximeter values this component was not included in further analysis in this study.

5. Conclusion

In this study regional venous saturation was estimated by utilising the respiratory/ventilatory modulations present in the PPG signal. The results showed that regional venous saturation values were significantly lower than the time domain arterial saturation values and instantaneous arterial saturation values.

The analysis method proposed in this study has some advantages over the technique proposed in the previous study (Walton et al.; 2010). Walton *et al* (2010) observed large standard deviation in the estimated venous saturation values compared

to the results obtained in this work. Thiele *et al.* (2011) have also mentioned the possible negative effect respiratory variations could have on the spectrum of the raw signal. In order to avoid these problems in this study the raw PPG signals were decomposed into components (venous related modulation, cardiac and DC PPG) before the frequency domain analysis. This could possibly provide more readable/cleaner spectrum and more accurate estimation of power related to each component. Also, the extracted modulation signal could be analysed further to study the effect of respiration variation on the PPG signal.

In this work is was also shown that by applying time-frequency analysis method stable instantaneous saturation values (arterial and venous) could be obtained without the need of subjective thresholding as required by the method presented in previous work (Walton et al.; 2010). These instantaneous saturation values might be able to detect transient changes that could be clinically significant.

It can be concluded that this study provides further validation of the hypothesis that modulation caused by respiration/ventilator in the PPG signal could be used to estimate the regional venous saturation and proposed a more refine analysis method for the extraction of venous related component and saturation estimation as compared to the previous studies.

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