Estimation of respiratory volumes from the photoplethysmographic signal. Part I: experimental results

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Abstract—*To evaluate the possibility of respiratory-volume measurement using photoplethysmography (PPG), PPG signals from 16 normal volunteers are collected, and the respiratory-induced intensity variations (RIIV) are digitally extracted. The RIIV signals are studied while respiratory volume is varied. Furthermore, respiratory rate, body posture and type of respiration are varied. A Fleisch pneumotachograph is used as the inspired volume reference. The RIIV and pneumotachograph signals are compared, and a statistical analysis is performed (linear regression and t-tests). The key idea is that the amplitude of the RIIV signal is related to the respiratory volume. The conclusion from the measurements is that there exists a relationship between the amplitude of the RIIV signal and the respiratory volume ($R=0.842$, $s=0.428$, $p<0.005$). Absolute measurements of the respiratory volume are not possible from the RIIV signal with the present set-up. The RIIV signal also seems to be affected by respiratory rate and type. More knowledge about respiratory parameters and improved sensor and filter design are required to make absolute measurements of volumes possible.*

Keywords—*Photoplethysmography, Non-invasive measurements, Respiratory volume, Clinical monitoring*

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1 Introduction

HEART AND respiratory rates are of great importance when critically ill adults and newborn infants are monitored. Clinical monitoring of heart rate is generally performed by counting QRS complexes of the ECG per time unit or by using a pulse oximeter. The respiratory rate is usually measured by timing a number of breaths with a stopwatch, or is monitored through the use of techniques such as transthoracic impedance plethysmography (ALLISON *et al.*, 1964; ASHUTOSH *et al.*, 1974; HAMILTON *et al.*, 1967) or capnography (LENTZ and HEIPERTZ, 1991; ROY *et al.*, 1991). Thermal (CYNA *et al.*, 1991) and acoustic (HENNEBERG *et al.*, 1992; HÖK, 1991; HÖK *et al.*, 1993) techniques have also been developed for this purpose.

Recently, respiratory and heart rate monitoring using photoplethysmography (PPG) has been suggested (LINDBERG *et al.*, 1992; UGNELL, 1995). The accuracy of this method was shown to be comparable with the earlier, more established methods (VEGFORS *et al.*, 1993).

PPG was first described by HERTZMAN and SPEALMAN (1937) and is an optical technique based on variations in absorption, refraction and scattering of light within an illuminated volume of blood and tissue. Light reflected from the skin surface is modulated by changes in blood volume and perfusion (see CHALLONER (1979) for a review).

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The possibility of undisturbed, long-term measurement and other advantages, such as non-invasiveness and reproducibility, have made photoplethysmography attractive from a monitoring point of view. The DC part of the signal has been studied by, among others, DAVIS and BAKER (1969) for estimating total blood volumes. The heart-synchronous part of the signal (AC) has frequently been suggested as a measure of skin perfusion (ELDRUP-JORGENSEN *et al.*, 1966; DORLAS and NIJBOER, 1985) but also as a way to monitor heart rate (ELINGS, 1959; LINDBERG *et al.*, 1992; UGNELL, 1995).

In addition to heart-synchronous variations, the photoplethysmographic signal contains respiratory-induced intensity variations (RIIV). This modulation arises from respiratory-induced variations in venous return to the heart, caused by the alterations in intrathoracic pressure (UGNELL, 1995). A part of the respiratory-related fluctuations in perfusion also originates from the autonomous control of the peripheral vessels, synchronous with respiration (SHEPERD and VANHOUTTE, 1975; BRECHER, 1956).

Many authors have noticed these variations. Early works (TRAUBE, 1865; HERING, 1869; MAYER, 1876) report on slow variations in blood pressure. These waves generally have a frequency lower than the respiratory rate (PENAZ, 1978). However, there are certain difficulties with this definition, as slower waves sometimes entrain the respiratory frequency. This problem is most obvious when the frequency of the respiration is low (AHMED *et al.*, 1982). The appearance of these variations in the PPG signal has been studied by, among others, FOSTER *et al.* (1945) and SARA and SHANKS (1978).

Power spectra from the PPG signal contain distinct peaks associated with heart and respiratory rates, respectively, but also contributions from slower waves (LINDBERG *et al.*, 1992).

With suitable filters, the RIIV signal can be extracted from the complicated total PPG signal (UGNELL, 1995). From this signal, the respiratory rate can be calculated and monitored by applying an appropriate frequency-calculating algorithm.

An even more qualitative and reliable monitoring method would be achieved if ventilatory volumes could be estimated in addition to the rate of respiration. This would provide the possibility of calculating the ventilation and thereby detecting undesirable ventilatory trends.

Methods used today for respiratory-volume registration are mainly spirometry, body plethysmography, pneumotachography and impedance plethysmography. These methods are either not suitable for long-term monitoring in clinical use, or are inadequate in other ways. Therefore the possibility of obtaining ventilatory volume information from the PPG signal is of great interest. Our previous results have shown that a relationship exists between the amplitude of the RIIV signal and the volume of inspired air (JOHANSSON and ÖBERG, 1996). The aim of this work is to investigate this relationship and the applicability of PPG for respiratory-volume measurement.

2 Method

The measurement set-up (Fig. 1) consisted of two major parts: the photoplethysmographic part, providing PPG signals from which the RIIV signals could be extracted, and the pneumotachographic part, providing air-flow data for reference respiratory-volume calculations. The amplitude of each RIIV pulse (volts), associated with each breath, could be compared with the corresponding volume of inspired air (litres), as calculated from the pneumotachograph data. Fig. 1 also shows an acoustic and a visual control system used for guidance of the subject (see below).

2.1 *Subjects*

The experiments were performed on 16 healthy subjects, eight male and eight female, in the age range of 21–35 years. No restrictions regarding pre-experimental activities were made. Each subject was placed comfortably in a semi-supine position, with the measured forearm resting approximately at heart level. A resting period of at least 5–10 min occurred before the start of the measurements. The importance of relaxation was stressed during the whole measuring period to minimise psycho-physiological effects and involuntary variations in thoracic tension.

2.2 *Pneumotachography*

A pneumotachograph with a Fleisch tube* was chosen as the volume reference method. A heating element kept the Fleisch tube at a constant temperature (37 °C). The pneumotachograph signal was sampled at 24 Hz† after having passed a noise-reduction filter‡. The samples were then stored in a PC§. An oscilloscope¶ was used for real-time monitoring of the signal.

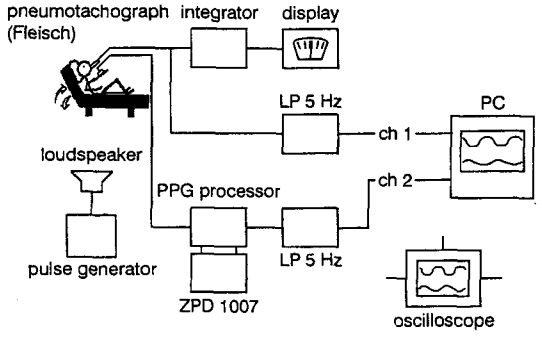

Fig. 1 *Experiment set-up*

The system was calibrated with the use of a calibration syringe containing an air volume of 3000 ml**. Before each measurement period, five slow strokes were made with the syringe through the pneumotachograph. The sum of the samples was calculated for each stroke. A mean value of the five strokes was then used as a reference value corresponding to a volume of 3000 ml.

The pneumotachograph output was assumed to be linearly related to the volume air-flow (SULLIVAN *et al.*, 1984). A linearity test was performed for further assurance of a linear relationship. Different flows were created by using the calibration syringe at different speeds. The pneumotachograph signals were analysed after sampling and storage in the computer. The mean value of the samples and the time for the syringe volume to pass the pneumotachograph were calculated. The results of the measurements are presented in Fig. 2. From these recordings, it can be concluded that the pneumotachograph signal responds linearly to passing flow in the considered flow interval.

2.3 *Photoplethysmography*

Infrared light of 880 nm from an LED†† was used as the light source. A solar cell‡‡, linear in the range of interest, served as detector for remitted light (UGNELL, 1995). Both components were mounted in a light-weight probe, serving as the optical sensor.

The sensor was placed ventrally on the left forearm using double-adhesive tape. The signal from the detector was processed in a PPG processor, containing parts for amplification and offset-balancing (LINDBERG *et al.*, 1992). The previously described noise-reduction filter, hardware and sampling frequency were used for storage of the PPG signals (see Section 2.2). An external bandpass filter (0.13–0.48 Hz, modified Bessel, lowpass 0.6 Hz (UGNELL, 1995)) was implemented in a digital signal processor§§ to extract the RIIV signals for real-time presentation on the oscilloscope.

2.4 *Measuring procedure*

When possible, the sensor was positioned above a visually identified vein on the forearm. The probe was covered by a dark cloth to minimise the influence of background light. A nose clip was used to prevent air flowing through the nose. The subjects were asked to breathe with preset volumes. The actual volumes, obtained through analogous integration, were displayed to the subjects. A more accurate calculation of the volume was performed in the computer (see below). As the amplitude characteristic of the respiratory filter is non-linear, there was a need to keep a constant respiratory rate. To do this, an acoustic signal was generated using a pulse generator and a

*Hugo Sachs Elektronik KG, Germany
† AD-card: AT-MIO-16L-9, National Instruments, USA
‡ KEMO analogue filter, Type VBF 8, low-pass 5 Hz, UK
§386SX, IBT, Malaysia
¶Gould, Model 740, UK
**Gould, Model M20, USA
††SFH 484-2, Siemens, USA
‡‡BPY63P, Siemens, USA
§§ZPD 1007, Burr-Brown, USA

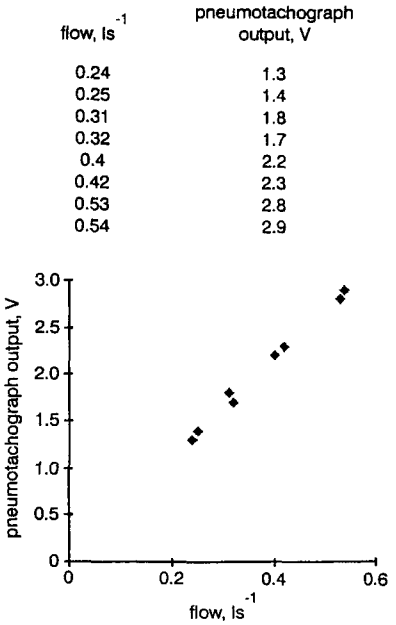

Fig. 2 *Pneumotachograph linearity test*

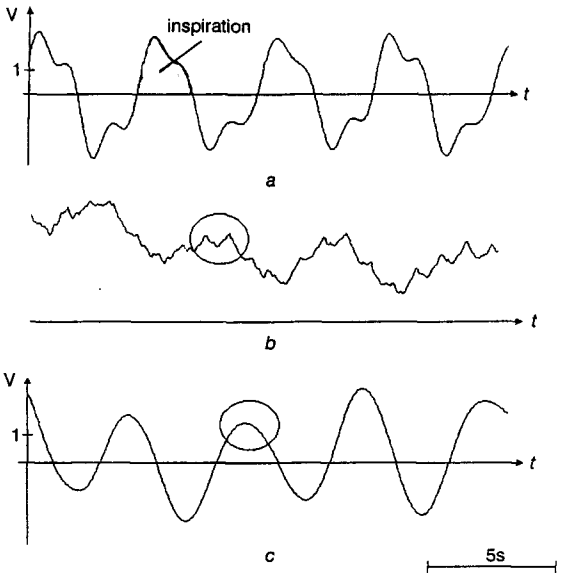

Fig. 3 *Typical signals: (a) pneumotachograph, (b) PPG and (c) RIIV. Inspiratory flow is shaded, and related peaks are marked in PPG and RIIV signal (note time delays)*

loudspeaker. During the measurements, the subjects were asked to keep to the requested pace by starting each inspiration at the generated sounds.

During the main measuring period, alternations in respiratory volume were performed. The subjects were requested to breathe with volumes in the range 0.2–2 litres (0.2, 0.5, 1.0, 1.5 and 2.0 litres). A specific volume was kept for 3–5 breaths before being changed. Before the start of each measurement, the subjects were allowed to practise breathing for some time. Volumes and paces were varied during the training period.

To investigate the influence of other parameters, three complementary measurements were made. Variations were made (one at a time) in: respiratory rate (10, 15 and 20 breaths min^{-1}), posture (supine, semi-supine and sitting) and type of respiration (mainly diaphragmatic, normal and mainly thoracic musculature in use). A respiratory rate of 15 breaths min^{-1}, a semi-supine position and normal breathing were used as 'default' values. The three measuring periods lasted approximately 2 min each.

2.5 *Signal processing*

After the measurements had been completed, the stored PPG and pneumotachograph signals were processed using LabWindows*. The RIIV signals were extracted using the previously mentioned digital filter. A typical RIIV signal and corresponding PPG and pneumotachograph signals are presented in Fig. 3.

When switching between inspiratory volumes (during the main measuring sequence), the rapid changes sometimes produced transition effects in the signals. To minimise the influence of these phenomena, disturbed pulses representing breaths after each rapid volume shift were eliminated. Except for these pulses, each pulse in the extracted RIIV signal was identified as a breath, and the maximum value (relative to the baseline) was recorded. The volume of each breath was determined by calculating the area of the positive part (representing inspiratory volume) of the pneumotachograph pulse. This area was then quantified by comparison with the corresponding calibration signal.

2.6 *Statistics*

For each individual, the correlation between the RIIV amplitudes and the reference volumes was investigated using linear regression. To make a comparison between the individuals possible, the measurements were normalised. To do this, each RIIV pulse amplitude was divided by the mean amplitude for the corresponding individual. A summarised regression was then performed. Two regression models, one linear and one second order, were used. The complementary measurements were evaluated using *t*-tests after corresponding normalisations. For the statistical analysis, STATISTICA software* was used.

3 Results

Fig. 4 shows actual pneumotachograph signals (broken lines) and corresponding RIIV signals (solid lines). As seen in Fig. 4*a*, there is a time delay between the two signals that can be explained by the transport delay in the respiratory and circulatory systems in the human body, together with the filter delay.

Fig. 4*b* shows that increased flows/volumes result in corresponding increases in the amplitude of the RIIV signal.

The RIIV signal from all subjects showed a significant dependence on respiratory volume ($p<0.05$, $R_{mean}=0.89$, $n_{mean}=17.4$). Fig. 5 shows the relationship between the inspired volume, in litres, and the corresponding amplitude of the RIIV signal for the whole material (after normalisation).

To reduce the residual sum of squares, a second-order regression model was also used. Both regression lines are included in Fig. 5, and the analysis is summarised in Table 1.

Table 2 presents an analysis summary of the complementary measurements. The numbers presented are mean RIIV amplitudes. A higher degree of thoracic respiration and a

* v 2.3a with Advanced Analysis Library

*StatSoft v5.1

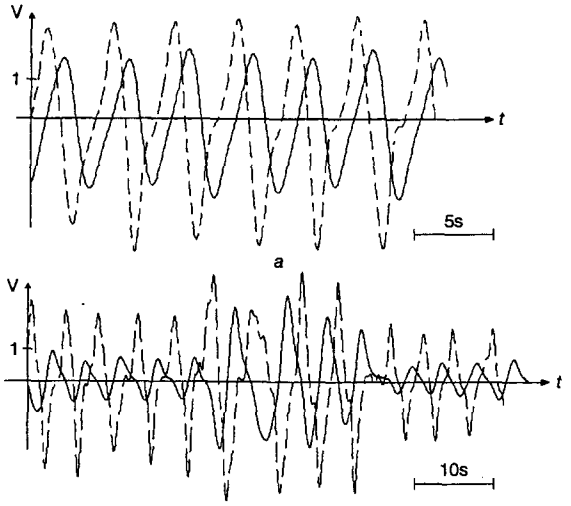

Fig. 4 *Actual recordings of (- - -) pneumotachograph and (—) RIIV signals. (a) Normal respiration at fixed volume and rate (0.8 litres at 15 breaths min^{-1}); (b) normal respiration with fixed respiratory rate (15 breaths min^{-1}) and shifts in respiratory volume from 0.5 to 1.5 and from 1.5 to 0.3 litres*

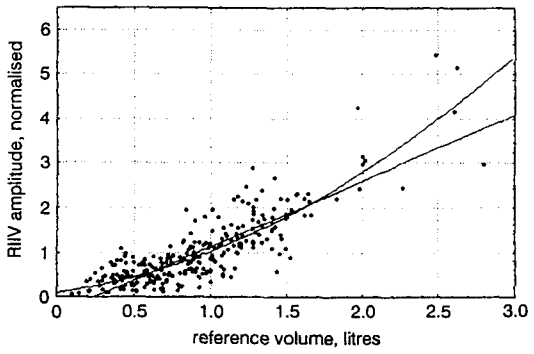

Fig. 5 *RIIV amplitudes (normalised) as function of reference volume of collected breaths (litres) for subjects. Linear and second-order regression lines are included*

higher respiratory rate seem to give rise to greater RIIV amplitudes. No significant dependence on posture could be identified.

4 Discussion

4.1 *Ventilatory monitoring*

The advantages of a respiratory-volume monitoring function are obvious, especially if it can be performed without access to the patient's airways. In many cases, the respiratory rate may be inadequate or less reliable in assessing the ventilation of the patient. For instance, during an episode of apnoea, a normal respiratory rate may be recorded in spite of respiratory volumes being too small. In other situations, respiratory volume changes can be used to predict conditions of insufficient ventilation. The possibility of detecting undesired ventilatory trends from the RIIV signal is therefore of great interest in connection with intensive care.

Non-invasiveness is an important property of techniques in long-term use and can be realised with reflection photoplethysmography. Optical techniques have previously been developed for measurement of parameters such as heart rate, respiratory rate and blood oxygenation. Ventilation monitoring can therefore easily be combined with measurements of this kind. Such combined measurements would make it possible to design monitoring equipment that can be adapted for many situations. For example, pulse oximeters could be made to respond faster and more reliably by adding a ventilatory-trend function.

The estimation of respiratory volumes from optical perfusion measurements of the skin may not seem to be the most straightforward technique. However, this method eliminates invasive procedures and airway interference. The utilisation of the RIIV signal requires a deep understanding of the physiology of respiration and the interaction of light with tissue.

Table 1 Summary of regression analysis for both models

	Model	
	linear ($Y = B_0 + B_1 \cdot X + \varepsilon_1$)	second-order ($Y = B_0 + B_1 \cdot X + B_2 X^2 + \varepsilon_2$)
R	0.842	0.862
s	0.428	0.404
p	<0.005	<0.005
B_0	−0.348	0.092
B_1	1.478	0.525
B_2	—	0.415

$\varepsilon_1 \sim N(0, \sigma_1)$; $\varepsilon_2 \sim N(0, \sigma_2)$; $n = 278$

Table 2 Summary of t-tests of complementary measurements. Numbers presented are mean RIIV amplitudes for 16 subjects

Parameter			
Respiratory type	normal	abdominal	rib cage
	0.924	0.776	1.300
		***	
	**		
Posture	semi-supine	supine	sitting
	0.917	0.999	1.084
Respiratory rate	15	10	20
	0.290	0.318	0.392
		**	
	***		

$^*p<0.05$; $^{**}p<0.025$; $^{***}p<0.01$

4.2 *Interpretation of respiratory waves*

The PPG signal is complex and contains several sources of slow oscillations. These seem to interfere with the respiratory part of the PPG signal. The oscillations are due to different phenomena such as vasomotion and regulation of blood pressure and body temperature. The influence of these components on the RIIV signal may explain some of the scattering of the data in Fig. 5. Improvements can be made in the design of the skin sensor as well as in the digital filters used for extraction of the RIIV signal from the PPG signal.

We believe that a more detailed study of the following conditions can lead to a better understanding of the volume dependence of the signal:

(*a*) Individual respiratory differences affect the signal. The rate of abdominal respiratory activity compared with thoracic respiratory activity needs to be better understood.
(*b*) Variations in posture and effects of body movements also seem to affect the RIIV signal. These phenomena are probably caused by hydrostatic effects on the blood vessels, together with a change in the distribution of blood in the body. The way

the PPG signal is affected by movement seems to vary greatly between individuals. More knowledge of sensor design in relation to movement is therefore required.
(*c*) Respiratory rate seems to affect the RIIV signal in a non-linear way. Models that compensate this phenomenon should be possible to design, but more information is required.
(*d*) In addition to the physiological disturbances, problems may arise concerning extremely low or high respiratory rates, owing to the finite limits of the filter extracting the respiratory signal. This problem can probably be solved by using respiratory-frequency adaptive filters.

Some of these conditions have been studied in detail in a theoretical model, described in a companion paper (JOHANSSON and ÖBERG, 1998).

When males and females in the material are compared, there is a tendency towards a difference in the dependence of respiratory type and rate. In the present study, the number of subjects in each group is too few for a more detailed analysis. This will be studied further in the future.

4.3 *Signal processing and statistics*

The main purpose of this paper is to investigate the relationship between the respiratory volume and the RIIV signal. Whether this information is extracted in the best way possible using the amplitude of the RIIV signal is not obvious. Therefore attempts at relating the integral of the RIIV samples of each pulse to the respiratory volume were performed, as this seemed to be a good approach. In neither of the measuring situations could a significant difference compared with the amplitude approach be found. This correlates well with previous results (CHALLONER, 1979).

Another approach could be to use the peak-to-peak value of the RIIV signal instead of the amplitude. This would facilitate the signal processing and the need to suppress the DC component in the signal. Amplitude measurements were chosen in this study, as these were assumed to suppress the physiological variations in the shape of the RIIV signal.

If the amplitude or peak-to-peak values are used, a model has to be designed to translate the RIIV signal information to volume information. Whether the best model to use is one of the two models presented in this study, an exponential relationship, or an empirically determined relationship, remains to be investigated.

The choice of a regression approach for the analysis of the volume measurements was straightforward owing to the test conditions. Analysis of variance (ANOVA) was considered for the complementary measurements, but was not regarded as necessary. Both linear regression and *t*-tests assume normally distributed data, which can easily be confirmed.

5 Conclusions

A relationship exists between the amplitude of the RIIV signal and the respiratory volume. Absolute measurements are not possible with the present set-up, but trends in ventilation are possible to monitor, and this is an important parameter in intensive care.

More knowledge about respiratory parameters and improved sensor and filter design are required to improve the quality of absolute measurements of volumes.

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Author's biography

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