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Estimation of Respiratory Flow by Means of Normal Lung Sound

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Abstract. The respiratory flow is a good indicator of sleep-related breathing disorders. Common praxis is to use a pneumotachograph as the golden standard for flow measurement. However, it does not have to be necessarily the best possible test device for long-term condition, because the device is very uncomfortable and rarely suitable for measurement during sleep. A computer-based method to determine the respiratory flow, called ThorAKUSTIK, yielded a highly positive correlation between the calculated flow out of the tracheal breath sound and a measured flow signal. In order to avoid noise interference due to a breath-mask or a pneumotachograph, in this study we applied the ThorAKUSTIK-method to lung sound which was measured at the back of 18 subjects and investigated the correlation between the calculated flow and the measured flow by a pneumotachograph. The new method showed a highly positive correlation (r = 0.89 and 0.90). Additionally we examined the use of an accelerometer signal to distinguish between inspiration and expiration. In this case we got high correlation coefficients of r = 0.87 and 0.88 between the calculated and measured airflow as well.

Keywords: Respiratory sounds, lung sounds, respiratory airflow, new method

1. Introduction

1.1. Respiratory airflow

Especially in polysomnography, the continuous detection of respiratory airflow is an important method to detect sleep-related respiratory disorders, e.g. apnea. Nowadays the pneumotachograph is regarded as the golden standard of flow measurement. However, especially in sleep medicine due to its size the spirometer is disruptive and uncomfortable for the patient. Previous studies examined alternative methods and yielded good results with an acoustic flow estimation based on tracheal lung sound as well as on normal lung sound [1-4]. The advantage of an acoustical calculation is that the used microphones are small and less bothering for the patient in contrast to a spirometer. The acoustical estimation is able to monitor the airflow independently and

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regardless of the patient's active collaboration. Thus, this method enables an application in long-term monitoring, even overnight, and can improve the patient's benefit immensely.

1.2. ThorAKUSTIK-method

The procedure is based on a frequency analysis. Initially the sound signal is transformed into the frequency domain, by using the fast Fourier transform (FFT). Consequently the frequency intensities are assorted ascending and a predetermined percentile is used to estimate the airflow. In previous studies the new ThorAKUSTIK-method was already used to calculate the respiratory airflow from tracheal breath sound and it yielded high linear correlation coefficients to the airflow which was simultaneously measured by a spirometer [5, 6].

Recent results motivated us to extend this method and to examine it by using normal lung sound which was recorded dorsal. With the formerly monitored tracheal breath sounds the distinction between the processes of inspiration and expiration was carried out by using the spirometer signal. Our aim is to replace the spirometer by new techniques and therefore we additionally investigated a breath phase distinction by using an accelerometer which is integrated in the recorder and attached to the patient's chest.

2. Methods

2.1. Data

Data of this study were recorded from 18 subjects (14 men, 4 women) between the ages of 22 and 51 years and a BMI between 19 and 34 kg/m². The average information is shown in Table 1.

Table 1. Information of the subjects: average age, height and body mass index (BMI)

	Age (Year)	Height (m)	BMI (kg/m²)
Average	29,2	1,79	24,1
Std.	± 6,3	± 0,10	± 3,8

The lung sound was recorded by two microphones which were attached to the subjects' backs placed over the right and left lung. To find the optimal position of the microphones the lung was auscultated beforehand. As reference the respiratory airflow was additionally recorded by a pneumotachograph, which was connected to a full face mask. In order to investigate the option of breath phase-detection due to the breathing movement, an accelerometer was integrated into the recording device which was

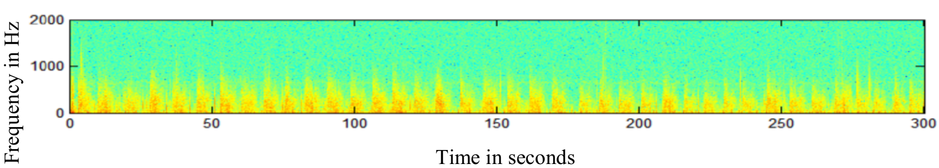

Figure 1. Spectrogram of sound signal; complete five minutes.

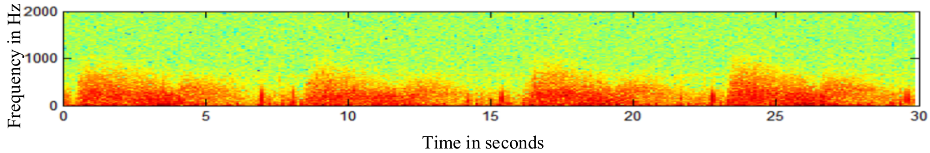

Figure 2. Spectrogram of sound signal; 30 sec.-extraction.

attached on the subjects' chests. The recordings were conducted in supine position for 5 minutes. During the measurement, the signals of the two microphones, as well as the flow signal and the three axes measured by the accelerometer were acquired and recorded simultaneously and saved as binary files.

2.2. Acoustical Flow Estimation

The data were recorded with a sampling rate of 5512 Hz in order to cover the complete frequency range of lung sound which reaches from below 100 Hz up to 2000 Hz and above [7]. According to the Nyquist-Shannon sampling theorem the sampling rate has to be at least twice the highest frequency component. For the flow calculation the sound signal was windowed into 1024-sample blocks with 50% overlap. In each block we computed the frequency spectrum using the fast Fourier transform. Typical spectrograms are illustrated in Figure 1 and Figure 2. In further frequency analysis, the ThorAKUSTIK-method kept the relevant frequency components and calculated a qualitative acoustically determined flow curve.

2.3. Phase Detection

To examine a method to replace the pneumotachograph for the breath phase detection, we investigated to what extent it is possible to detect in- and expiration with an accelerometer which was attached on the subject's chest (Figure 3). For the phase detection the signals of all three axes coming from the accelerator were initially smoothed. In order to find distinguishable characteristics between inspiration and expiration the gradient of each signal was compared to the flow signal of the pneumotachograph. Afterwards the determined phases were subsequently assigned to the calculated flow signal so that the recorded values during expiration became negative, similar to the spirometer signal. As a reference, the distinction between the two phases was additionally conducted by use of the spirometer signal. Therefore the

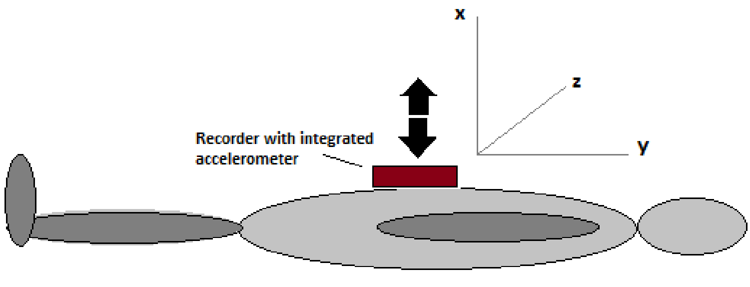

Figure 3. Axes of accelerometer signal

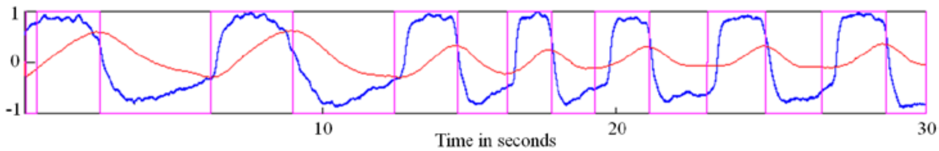

Figure 4. Extraction of flow signal (blue), x-orientation of accelerometer signal (red) and determined breath phases (magenta)

inspiration was determined at a flow of 0.1 l/s and above and the expiration respectively at flow rates of minimum -0.1 l/s of the spirometer signal. The correlation coefficients were calculated for both the calculated signal with distinction by accelerometer and the calculated signal with distinction by spirometer signal and finally compared to each other.

3. Results

We decided that the respiratory phases can be found best in the gradient of the x-orientation signal of the accelerometer, whereby a positive gradient indicates the inspiration and respectively a negative gradient the expiration. In this case, the x-orientation represents the movement of the thorax in ventral-dorsal direction and vice versa as it is illustrated in Figure 3.

In Figure 4 a flow signal with corresponding accelerometer signal and an additional signal which indicates the different phases are illustrated. As a result this method gives a practicable distinction between inspiration and expiration phases.

Subsequently we calculated the Pearson correlation coefficient r and the corresponding standard deviation (Std.) of the measured flow with the spirometer correlated with the acoustically calculated flow for the right and left lung side. To compare the breathing phase distinction, according to accelerometer signal with the distinction by spirometer signal, we calculated the mentioned values for both estimated flow rate methods separately. As average correlation coefficients we got $r = 0.90$ (right side) and $r = 0.89$ (left side) ($p<0.001$) for calculated flow signal without accelerometer and $r = 0.88$ (right side) and $r = 0.87$ (left side) ($p<0.001$) for estimated flow signal whereby the breath phases were determined by the accelerometer signal. Figure 5 illustrates a 30 sec.-extraction of the airflow signal which was measured by the pneumotachograph and the calculated flow signal.

Table 2 shows the average values for r and their standard deviation Std.

The boxplot in Figure 6 illustrates the correlation coefficients’ distribution for each calculation method.

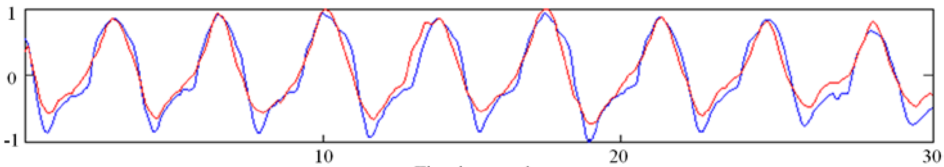

Figure 5. Qualitative comparison of measured airflow (blue) and calculated airflow with accelerometer (red); 30 sec.-extraction.

Table 2. Average results for correlation coefficient r and standard deviation Std. between measured flow signal and calculated signal (with and without accelerometer signal); level of significance (p-value) < 0.001

	with accelerometer		**without accelerometer**	
	left side	**right side**	**left side**	**right side**
r	0.87	0.88	0.89	0.90
Std.	± 0.07	± 0.08	± 0.12	± 0.13

4. Discussion

The results of the present study show that with the recently introduced ThorAKUSTIK-method in general it will be possible to calculate a qualitative respiratory flow from lung sound signals with relative high correlation coefficients of at least 0.87. Compared to our former study in which we used the tracheal lung sound to calculate the airflow and yielded correlation coefficients between 0.89 and 0.92 we could achieve similar high correlation coefficients [6]. Soufflet investigated eight different methods of flow estimation and yielded correlation coefficients between 0.55 and 0.70 [3]. In this relation, our results show that the ThorAKUSTIK-method achieves comparatively high correlations. The calculated values during the expiration are lower than during the inspiration probably due to the quieter respiratory sound while expiration [8]. In order to improve the correlation of the expiratory flow values we should consider preprocessing the expiratory flow signal more properly in further studies.

Furthermore, the breath phase detection with an accelerometer signal yielded good results in comparison to the distinction via the spirometer signal and the coefficient for both methods are almost equal. As a consequence, the breath phase distinction by an accelerometer can adequately replace the former distinction by the spirometer which was one of our goals. Nevertheless, this method needs to be investigated more precisely in further studies, especially the drift between the real change of in- and expiration and the change in the accelerometer signal. As one can see in Figure 4 the borders between inspiration and expiration which are determined by the accelerometer signal (magenta square-wave signal) do not exactly agree with the thresholds (± 0.1l/s) of the spirometer signal which we determined for the distinction between in- and expiration. Consequently the values of the airflow determined by the accelerometer may have especially at these thresholds a different sign which reduces the correlation coefficient. This could be the reason why the correlation coefficients are a bit lower for the calculated flow with accelerometer signal as without this additional signal. Furthermore, in this study the recordings were conducted only in supine position and so we have to test similar algorithms in different positions to validate the general use of the method.

During a long-term application patients certainly change lying position and in this case it may be expected, that a single orientation vector will not be sufficient to detect the breath phases adequately. Consequently a combination of different orientations will be necessary.

We are also planning to evaluate the lung sound method in further studies under CPAP- and BiPAP-ventilation, to ensure a very local and specific determination of the respiratory flow among the pulmonary area.

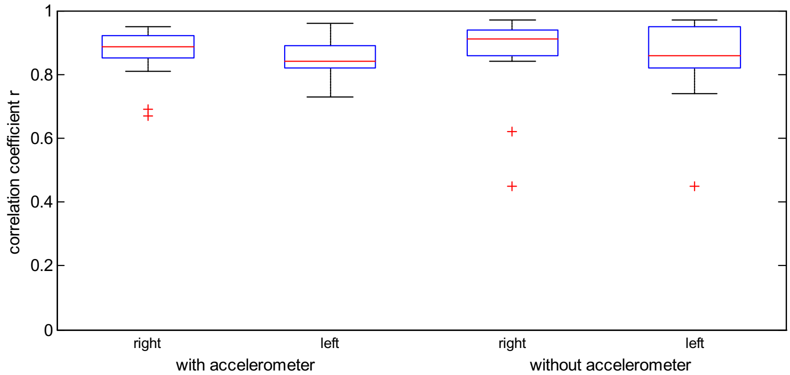

Figure 6. Boxplot of correlation coefficients for the two different calculation methods and left and right lung side

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