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Computerized wheeze detection in young infants: comparison of signals from tracheal and chest wall sensors

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Abstract
Computerized wheeze detection is an established method for objective assessment of respiratory sounds. In infants, this method has been used to detect subclinical airway obstruction and to monitor treatment effects. The optimal location for the acoustic sensors, however, is unknown. The aim of this study was to evaluate the quality of respiratory sound recordings in young infants, and to determine whether the position of the sensor affected computerized wheeze detection. Respiratory sounds were recorded over the left lateral chest wall and the trachea in 112 sleeping infants (median postmenstrual age: 49 weeks) on 129 test occasions using an automatic wheeze detection device (PulmoTrack®). Each recording lasted 10 min and the recordings were stored. A trained clinician retrospectively evaluated the recordings to determine sound quality and disturbances. The wheeze rates of all undisturbed tracheal and chest wall signals were compared using Bland–Altman plots. Comparison of wheeze rates measured over the trachea and the chest wall indicated strong correlation ($r \geq 0.93$, $p < 0.001$), with a bias of 1% or less and limits of agreement of within 3% for the inspiratory wheeze rate and within 6% for the expiratory wheeze rate. However, sounds from the chest wall were more often affected by disturbances than sounds from the trachea (23% versus 6%, $p < 0.001$). The study suggests that in young infants, a better quality of lung sound recordings can be obtained with the tracheal sensor.

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Keywords: respiratory sounds, phonopneumography, sensor placement, computerized wheeze detection, respiratory function testing, infant

(Some figures may appear in colour only in the online journal)

1. Introduction

Computerized respiratory sound analysis is an objective method for diagnostics and monitoring that was developed to overcome the shortcomings of stethoscope-based auscultation (Guntupalli et al 2008), such as high inter-observer variability (Elphick et al 2004, Prodhan et al 2008), lack of standardization in the nomenclature of respiratory sounds (Elphick et al 2000, Pasterkamp et al 2016), and varying quality of stethoscopes (Loudon and Murphy 1984, Brooks and Thomas 1995). Devices for computerized respiratory sound analysis digitally record a patient’s breathing sounds, analyze the sound signal (Kulkas et al 2010, Emoto et al 2014), and detect relevant normal and adventitious sounds such as crackles or wheezes (Gurung et al 2011).

Automatic wheeze detection can be performed at any time and allows a continuous monitoring, namely overnight and in several conditions especially in children with asthma (Boner et al 2010, Eising et al 2014). Furthermore, it has been used to assess treatment effects in infants with bronchiolitis (Sánchez et al 2005, Faber et al 2015). Recent research has reported a correlation between automatically detected wheeze and conventional parameters of lung function testing (LFT) in a high risk population of neonates and young infants (Fischer et al 2016, Schmalisch et al 2015). Wheezes are continuous musical sounds with one or more tonal components (Pasterkamp et al 1997, Ellington et al 2012), that are usually louder than the underlying breath sounds. However, it can be difficult to assess respiratory sounds in neonates and infants (Pasterkamp et al 1983, Tal et al 1991). Previous studies of wheezing in older children and adults exclusively measured respiratory sounds from the trachea (Boner et al 2010), other studies used only chest wall sensors (Blowes et al 1995, Elphick et al 2004) while still others recorded lung sounds on both locations (Bentur et al 2004, Eising et al 2014). Computerized lung sound measurements in neonates, however, require a reduction of the number of sensors due to the smaller thoracic surface. Especially in preterm infants, the use of a single respiratory sound sensor would be most convenient, because these patients often have other monitoring sensors attached to the thorax. To our knowledge, no previous study has evaluated the effect of the sensor position on the quality of sound recordings in this age group. Therefore, it is important to know which sensor location achieves better results.

We hypothesized that the quality of the recorded respiratory sounds would be different for tracheal and chest wall sensors, which may cause differences in the measured wheeze rate. Therefore, the aim of this study was to evaluate the quality of respiratory sound recordings in young infants, and to determine whether the position of the sensor affected computerized wheeze detection.

2. Methods

2.1. Subjects

This retrospective study compared respiratory sound signals that were recorded simultaneously over the trachea and chest wall in a convenience sample of 112 infants at Charité University Medical Center, Berlin in a dedicated LFT unit. The respiratory sounds were
recorded as a part of LFT in our routine follow-up program of infants who required intensive care (Schmalisch et al 2013). Evaluation and publication of the recorded patient data was approved by our Institutional Data Safety Committee. All parents provided written informed consent before respiratory sound analysis and LFT. Ninety-five infants (85%) were tested once and seventeen (15%) twice, so there were 129 test occasions. The indications for LFT were bronchopulmonary dysplasia \( (n = 44) \), respiratory distress syndrome \( (n = 33) \), congenital diaphragmatic hernia \( (n = 11) \), respiratory maladaptation \( (n = 5) \), double aortic arch anomalies \( (n = 3) \), congenital cystic adenomatoid malformation \( (n = 2) \), tracheomalacia \( (n = 2) \) and others \( (n = 10) \). In this study, we only included infants (patients less than one year of age), as special problems with sound recording have been described in this age group (Pasterkamp et al 1983, Tal et al 1991). The digital recordings were stored for subsequent evaluation.

2.2. Computerized wheeze detection and sound recordings

Before the LFT, body weight was measured to the nearest 10 g (Seca, Hamburg, Germany) and body length (crown to heel) and chest circumference was measured to the nearest 5 mm. Respiratory sounds were recorded by the PulmoTrack® Model 2020 (Karmel Sonic Ltd, Israel). This device uses the signal of two phonopneumographic contact sensors, one fixed in the left axillary line (chest wall (CW) sensor) and the other over the manubrium of the sternum (tracheal (TR) sensor), to quantify the relative duration of inspiratory and expiratory wheezing. Another air-coupled microphone was placed next to each infant to record ambient noises and improve the signal-to-noise ratio. A respiration belt was strapped around each infant’s chest to detect the inspiratory and expiratory phases of the breathing cycle (figure 1). A detailed description of the measurement protocol was presented previously (Puder et al 2014). Briefly, respiratory sounds were recorded for 10 min in sleeping infants who were clinically stable and free of respiratory infections during the 3 previous weeks. Sleep was induced by oral chloral hydrate \( (50 \text{ mg} \cdot \text{kg}^{-1}) \) given 15–30 min before starting the recording, because sedation is necessary for the subsequent LFT (Schmalisch et al 2013). Respiratory sound recordings were performed directly before LFT was started. The children remained sedated during the whole lung function testing. All infants remained stable during sound recording and LFT, none showed clinical signs of respiratory distress. Recording took place in a quiet room with background noise kept to a minimum. The recording was repeated if there were technical problems or considerable ambient noise. The PulmoTrack® uses a fast Fourier transform-based algorithm to detect wheezing. The PulmoTrack® calculates the relative inspiratory and expiratory wheeze rate as:

\[
\text{Inspiratory wheeze rate } (%) = 100 \cdot \frac{T_{\text{w in}}}{T_{\text{in}}}
\]

\[
\text{Expiratory wheeze rate } (%) = 100 \cdot \frac{T_{\text{w ex}}}{T_{\text{ex}}}
\]

where \( T_{\text{w in/ex}} \) is the breathing time with wheeze during inspiration/expiration and \( T_{\text{in/ex}} \) is the total inspiratory/expiratory breathing time over a period of 1 min.

2.3. Quality analysis of the sound recordings

The quality of the 129 sound recordings was analyzed in January and February 2016. The sound signals were acoustically evaluated by a medical doctor who had special expertise in the interpretation of infant respiratory sounds. This investigator listened to the recorded sounds
and assessed their quality, but was blinded to the spectrograms and the results of the computerized wheeze detection. Moreover, the investigator categorized each recording as ‘normal’ or ‘disturbed’, and further classified the ‘disturbed’ recordings as group 1 (permanent interfering signals with constant frequency spectra) or group 2 (weak breathing and/or other interfering noises that overlaid the respiratory sounds). Only interfering sounds that were present most of the time were defined as disturbances. Physiological sounds that did not overlay the lung sounds (e.g. heart sounds, peristaltic sounds) and temporary sound artifacts due to infant movements or occasional crying were not classified as disturbances.

2.4. Statistical methods

Patient characteristics and lung sound data are reported as rates (%) or as medians and interquartile ranges (IQRs). Incidences were compared using Fisher’s exact test or a Chi-squared test as appropriate. Paired continuous variables were compared by the Wilcoxon rank-sum test; and independent variables by the Mann–Whitney test. Wheeze rates that were measured over the trachea and chest wall were compared using the statistical approach described by Bland and Altman (1986). Intraclass correlation coefficients with 95% CI (relative reliability of both measurements) were calculated to assess the consistency and reproducibility of the wheeze rates measured by the CW and TR sensor. All statistical analyses were performed using Statgraphics Centurion® software (Version 16.0, Statpoint Inc., Herndon, VA, USA) and MEDCALC (Version 9.1.0.1, MedCalc Software, Mariakerke Belgium). A p-value less than 0.05 was defined as significant.
Table 1. General characteristics of neonates (top) and characteristics on the day of the respiratory sound recordings (bottom).

<table>
<thead>
<tr>
<th></th>
<th>Median (IQR) or number (%)</th>
</tr>
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<tbody>
<tr>
<td><strong>Neonatal period (N = 112)</strong></td>
<td></td>
</tr>
<tr>
<td>Male</td>
<td>64 (57%)</td>
</tr>
<tr>
<td>Gestational age (weeks)</td>
<td>28 (26–33)</td>
</tr>
<tr>
<td>Birth weight (g)</td>
<td>1052.5 (770–1950)</td>
</tr>
<tr>
<td>Surfactant administration(a)</td>
<td>68/107 (64%)</td>
</tr>
<tr>
<td>Mechanical ventilation(a) &gt;24 h</td>
<td>53/110 (48%)</td>
</tr>
<tr>
<td><strong>At day of measurement (N = 129)</strong></td>
<td></td>
</tr>
<tr>
<td>Age (days)</td>
<td>144 (100–203)</td>
</tr>
<tr>
<td>Postmenstrual age (weeks)</td>
<td>49 (44–60)</td>
</tr>
<tr>
<td>Body length (cm)</td>
<td>58.0 (54.0–64.0)</td>
</tr>
<tr>
<td>Body weight (g)</td>
<td>4940 (4000–6370)</td>
</tr>
<tr>
<td>Chest circumference (cm)</td>
<td>36.5 (34.5–39.5)</td>
</tr>
</tbody>
</table>

\(a\) Numbers reduced due to incomplete data of outpatients examined by LFT.

Table 2. Respiratory sound recordings with and without disturbances, \(n\) (%).

<table>
<thead>
<tr>
<th></th>
<th>Without disturbance</th>
<th>With disturbance</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Group 1</td>
<td>Group 2</td>
</tr>
<tr>
<td>Tracheal sensor (TR)</td>
<td>121 (94%)</td>
<td>3 (2%)</td>
</tr>
<tr>
<td>Lateral chest wall sensor (CW)</td>
<td>99 (77%)</td>
<td>11 (8%)</td>
</tr>
</tbody>
</table>

3. Results

3.1. Study population

Table 1 shows the characteristics of the 112 enrolled infants. Almost half of the infants (48%) required mechanical ventilation during the first days of life, and 43 (38%) were extremely low gestational age newborns (ELGANs), born before 28 weeks of gestation.

3.2. Subjective respiratory sound assessment.

The subjective assessment of the respiratory sound recordings showed that sounds from the CW sensor were more often affected by disturbances than those sounds from the TR sensor (table 2). In particular, disturbed respiratory sounds were present in 8 (6%) recordings from the TR sensor and in 30 (23%) recordings from the CW sensor \(p < 0.001\). Moreover, it became apparent during the subjective sound assessment that the lung sounds recorded by the CW sensor mostly had a lower volume than those recorded by the TR sensor. Undisturbed recordings from both sensors were obtained in 97 (75%) cases. Infants with and without disturbances had no significant differences in chronological age, postmenstrual age, or anthropometric measures at the day of measurement as well as in the main diagnoses (data not shown).

3.3. Assessment of signal disturbances

Table 2 shows the incidence and classification of the disturbances in the CW and TR sensor signals. In total, 14 (37%) of 38 disturbed recordings were classified in group 1 due to
the loud, continuous interfering signals that were similar to an acoustical feedback. Figure 2 shows a typical spectrogram from a patient in this group. In addition 24 (63%) of the disturbed sound signals were classified in group 2 due to the interference by white noise, cracking sounds, and whistling. These sound artifacts overlaid the respiratory sounds and made them difficult to identify, especially if respiratory sounds were low in volume. Figure 3 shows an example of a spectrogram with differences in the clearness of the respiratory sounds between the CW and the TR sensor signals.

**Figure 2.** Spectrogram shows permanent interfering signals with constant frequency spectra from the chest wall sensor, but a normal breathing pattern from the tracheal sensor.

**Figure 3.** Spectrograms showing differences in the clarity of the respiratory sound from the tracheal and chest wall sensors.
3.4. Comparison of wheeze detection between CW and TR sensors

The intraclass correlation coefficient (95% CI) between both sensors was 0.971 (0.953–0.981) for the inspiratory wheezing rate and 0.965 (0.942–0.978) for the expiratory wheezing rate. Analysis of the 97 undisturbed recordings of both sensors indicated a strong correlation between the wheeze rates measured over the trachea and chest wall for inspiratory wheezing (figure 4) and expiratory wheezing (figure 5). For both inspiratory and expiratory
wheezing the coefficients of determination were greater than 88%. Bland–Altman plots showed that the individual within-subject differences in wheeze rates were randomly distributed. There was a small but statistically significant bias for inspiratory and expiratory wheezing, with a slightly higher measured wheeze rate for the TR sensor (mean bias (95% CI) inspiratory wheezing 0.48 (0.17–0.8)%), \( p = 0.003 \); expiratory wheezing 1.0 (0.42–1.65)%), \( p = 0.001 \). The limits of agreement were ±3% for inspiratory wheezing and ±6% for expiratory wheezing.

**Figure 5.** Relationship between the expiratory wheeze rate measured over the chest wall and over the trachea (same presentation as figure 4).
4. Discussion

The present study showed differences in the quality of respiratory sound recordings between CW and TR sensors in a convenience sample of young infants. Disturbances were significantly more common in the CW signal than in the TR signal. When undisturbed recordings from both sensors were available, there were strong correlations between the inspiratory and expiratory wheeze rates measured with the different sensors, and high intraclass correlation coefficients of $>0.95$.

The biases in the inspiratory and expiratory wheeze rates were statistically significant, but were within the range of the digital resolution of the PulmoTrack® ($\pm1\%)$ and were not relevant from a clinical point of view. The Bland–Altman plots demonstrated relatively small limits of agreement ($\pm3\%$ for inspiratory wheeze rate and $\pm6\%$ for expiratory wheeze rate) when both signals were undisturbed. These differences in wheeze rates from the different sensors can be considered as random errors which reflect technical difficulties in wheeze detection from the lung sound spectrograms (Pasterkamp et al 1997). Furthermore, the limits of agreement for the wheeze rates were higher for expiration than for inspiration. This might be because the wheezing time is significantly longer during expiration than inspiration (Eising et al 2014) and this could cause a wider scattering.

The significantly higher incidence of disturbed recordings from the CW sensor might be because lung sounds from the chest wall have a lower volume than those from the trachea, in accordance with previous studies of adults and children (International Lung Sounds Association 1982, Hidalgo et al 1991). Previous studies of children with asthma indicated that wheezing sounds are better heard over the trachea than over the chest wall (Takezawa et al 1980, International Lung Sounds Association 1982, Fenton et al 1985). This finding agrees with our results in young infants. The lower volume of the chest wall signal is apparently due to the filtering effect of the lung parenchyma, and a better transmission of sounds through the airways is presumably responsible for the higher volume over the trachea (Wodicka et al 1989, Meslier et al 1995). From a technical point of view, the low-volume sounds recorded over the chest wall have a lower signal-to-noise ratio, and are therefore more likely to be affected by interfering background noises than the louder sounds over the trachea.

We found no correlation between disturbances in the sound recordings and demographic or anthropometric measures of the infants, indicating that the equipment used provided reliable sound signals in the whole study population. Because our study was retrospective, it was not possible to determine if the disturbances reflected technical, environmental or physiological interferences. Nevertheless, a careful preparation of the test set-up (i.e. accurate fixation of the sensors), may help to improve recording quality.

When there is a significant difference between the wheeze rates recorded from both sensors, this suggests a technical disturbance in one of the sensors. Therefore, it is advisable to use both sensors and compare the wheeze rates as a quality control measure. Another way to detect disturbances is by visual inspection of the spectrogram. Especially the permanent interfering signals with a constant frequency spectra are easy to detect (figure 2).

This study has several strengths and limitations. The main strengths include the use of a relative large sample size and the same equipment and protocol for all patients. The equipment we used has recently been validated for young infants (Puder et al 2014, Fischer et al 2015). One of the limitations of this study is that the retrospective nature of the evaluation did not allow investigation of the causes of disturbances in the lung sounds. Furthermore, we only assessed two fixed sensor locations, so we cannot make any conclusions about the quality of other sensor placements. A further limitation of our study is that only one observer has assessed the quality of the respiratory sounds. Moreover, the indications for infant LFT were
too heterogeneous and the subgroups with disturbed sound recordings were too small to analyze in detail the relationship between sound quality and clinical diagnoses. Finally, respiratory sounds were recorded in sedated, sleeping infants in a relatively quiet lung function unit, so these results may not be applicable for computerized wheeze detection in awake patients or in noisier clinical settings.

5. Conclusion

The present study of computerized wheeze detection in young infants suggests that a better quality of respiratory sound recordings can be obtained with the TR sensor, because the TR sensor was less often affected by disturbances than the CW sensor. The use of both sensors, however, has the potential to serve as a quality control of the measured wheeze rate. Further studies should seek to improve the computerized wheeze detection by combining the signals of both sensors.

Acknowledgments

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Conflict of interest statement

None of the authors has a financial relationship with any commercial entity that has an interest in the subject of this manuscript.

Authors’ contribution

LP, HF and GS had primary responsibility for study design, data analysis, and writing of the manuscript. SW measured all lung sounds and GS performed statistical analysis. LP assessed all of the sound recordings. All authors read and approved the final manuscript.

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