Leak measurements in spontaneously breathing premature newborns by using the flow-through technique

B. FOITZIK, M. SCHMIDT, D. WINDSTETTER, R. R. WAUER, AND G. SCHMALISCH
Department of Pediatrics, Humboldt University, 10098 Berlin, Germany

Foitzik, B., M. Schmidt, D. Windstetter, R. R. Wauer, and G. Schmalisch. Leak measurements in spontaneously breathing premature newborns by using the flow-through technique. J. Appl. Physiol. 85(3): 1187–1193, 1998.—A new method for measuring and correcting air leaks during lung-function testing in infants has been validated in vitro and in vivo by using a flow-through system that measured the inflow and outflow of a face mask. An adjustable leak was quantified by using suction flow to validate the accuracy of leak measurements. To validate the leak correction, the volume of a pump was measured with different air leaks (0–30%). The method developed was tested in 67 infants breathing spontaneously. There was good agreement between measured and simulated leaks (r = 0.998, P < 0.001; 95% limits of agreement were −0.3 and 0.1%, respectively). The volume was generally underestimated because of leaks, and the volume error was up to 94% compared with the maximum error of 5% after leak correction. With continuous leak measurements in vivo, there were <4% actual leaks (median 2.6%), and we did not observe any leaks in >7% of cases. The leak correction improved the accuracy of ventilatory measurements. The monitoring of leaks is helpful for airtight placement of the face mask and for prevention of serious measurement errors caused by leaks.

Lung-function testing; measurement error; differential flow system; numerical correction; infants

Most methods for lung-function testing (LFT) in newborns require ventilatory measurements. Ventilation can be measured by indirect methods (e.g., respiratory inductive plethysmography) and directly by whole body plethysmography or by a pneumotachograph (PT) placed in the airflow. The insertion of a PT in series with a face mask is the simplest method. However, it poses the problem, especially in very-low-birthweight infants, of adding poorly tolerated dead space to the patient. This apparatus dead space can even exceed the infant’s tidal volume. The resulting CO2 rebreathing influences the infant’s ventilatory pattern (10) and makes long-term measurements impossible. Long-term measurements are, however, necessary for infants with an irregular pattern of respiration or with temporary irritation caused by the face mask.

Rigatto and Brady (14) used a flow-through system with one PT in the outflow limb, subtracting a constant background flow electrically from the output of the PT to reach an artificial zero flow line. This arrangement functionally eliminates the apparatus dead space and permits long-term measurements. However, the electrical subtraction doesn’t work correctly whenever a leak occurs. The arrangement of two PTs in a background flow system was first described by Ruttimann et al. (15). By using this flow-through technique (FTT), the breathing flow of patients into the face mask is superimposed on the constant background flow. The tracheal airflow is measured as the relative subtraction from (inspiration) or addition to (expiration) the background flow, respectively. Moreover, the FTT is the only method that provides easy switching and arbitrary conditioning of inhaled gas.

However, by using the FTT, the problem of air leak becomes more obvious than in systems without background flow because even minor leaks with a negligible effect on the measured flow magnitude lead to a significant offset of flow signal (followed by a drift in calculated volume) and make the evaluation of ventilatory signals more difficult. Even if the face masks are properly applied, leaks cannot be avoided, and the resulting errors in volume (and flow) are greater than with conventional equipment. Errors in flow significantly affect the accuracy of most ventilatory parameters as well as the calculation of lung mechanics.

Known methods for detecting leaks during measurement of ventilation in spontaneously breathing infants are quite extensive, e.g., the occlusion test (24) or simultaneous measurements by body plethysmograph investigations (4). In contrast, the possibility of air leak measurement is inherent in the FTT by measurement of in- and outflow of the face mask.

The purposes of this study were, first, to determine the accuracy of air leak measurements with face masks by using the FTT and, second, to clarify whether it is possible to mathematically correct the leak-resulting errors in flow and volume measurements. Furthermore, leak measurements by using the developed algorithms were performed in 67 newborns during routine LFT (to determine how leak measurements and corrections can improve the accuracy of LFT in clinical practice).

Materials and Methods

Subjects.Leaks were measured during routine LFT in 67 spontaneously breathing newborns in our laboratory from August to December 1996 (see also demographic data in Table 1). Thirty-one patients had a pulmonary illness (main diagnosis: bronchopulmonary dysplasia), and the remaining 40 were recovering from respiratory diseases.
Table 1. Demographic and ventilatory data for 67 spontaneously breathing infants

<table>
<thead>
<tr>
<th></th>
<th>Median</th>
<th>Minimum</th>
<th>Maximum</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gestational age, wk</td>
<td>36</td>
<td>26</td>
<td>42</td>
</tr>
<tr>
<td>Age, days</td>
<td>12</td>
<td>0.5</td>
<td>181</td>
</tr>
<tr>
<td>Birth weight, g</td>
<td>2,505</td>
<td>680</td>
<td>4,880</td>
</tr>
<tr>
<td>Actual body weight, g</td>
<td>2,645</td>
<td>1,060</td>
<td>4,600</td>
</tr>
<tr>
<td>Breathing rate, breaths/min</td>
<td>45</td>
<td>25</td>
<td>69</td>
</tr>
<tr>
<td>Measured volume, ml</td>
<td>13.5</td>
<td>8.1</td>
<td>29.1</td>
</tr>
<tr>
<td>Measured minute ventilation, ml/min</td>
<td>597</td>
<td>225</td>
<td>1,355</td>
</tr>
<tr>
<td>Leak-corrected volume, ml</td>
<td>14.1</td>
<td>8.4</td>
<td>30.8</td>
</tr>
<tr>
<td>Leak-corrected minute ventilation, ml/min</td>
<td>630</td>
<td>251</td>
<td>1,365</td>
</tr>
<tr>
<td>Relative error of volume and minute ventilation,* %</td>
<td>3.5</td>
<td>0.0</td>
<td>16.4</td>
</tr>
</tbody>
</table>

*Relative error = \(100 \times \frac{\text{corrected volume} - \text{measured volume}}{\text{corrected volume}}\).

The infants' clothing was comfortable, thus not influencing respiratory efforts. The tests were usually performed 30 min after feeding. Most of the infants were studied during natural, quiet sleep, assessed by behavioral criteria (13). Seven infants were sedated with chloral hydrate immediately before examination (50 mg/kg po). Throughout the whole procedure, the patients were monitored continually by pulse oximetry.

After a period of accommodation (5–20 min), we registered the mean leak values of at least 10 breaths but not more than 30. The number of selected breaths depended on variability in the breathing patterns.

Ethics. The performed lung-function measurements were approved by the local medical ethics committee. Parents were given a full explanation of the tests and equipment used before their consent was obtained.

Leak measurement. We regarded the air leak as nonlinear resistance between the measurement apparatus and the surrounding ambient air (e.g., gap between face and mask) that produces a leak flow dependent on the pressure drop across this resistance. By using the FTT, the air leak was determined by the mean leak flow \(\dot{V}_{\text{leak}}(t)\) escaping through the leak-resistance in relation to the constant background flow (for more details, see APPENDIX)

\[
\text{leak} = \frac{\dot{V}_{\text{leak}}(t)}{\dot{V}_{\text{const}}(t)} = \frac{1}{n \cdot T} \left[ \int_{0}^{nT} \dot{V}_{\text{const}}(t) \, dt - \int_{0}^{nT} \dot{V}_{\text{PT}}(t) \, dt \right]
\]

where \(\dot{V}_{\text{const}}(t)\) is constant background flow measured by \(\dot{V}_{\text{PT}}(t)\) is airflow from face mask measured by \(\dot{V}_{\text{PT}}\), and \(nT\) is \(n\) breaths of duration \(T\).

The mean leak flow \(\dot{V}_{\text{leak}}(t)\) was calculated from the balance of inflowing to outflowing breathing volumes over an integer number \((n)\) of breathing cycles. The delay time of leak calculation depended on the number of breathing cycles (for 5 breaths and frequencies at 60–120 min ranging between 2.5 and 5 s), and the evaluation over a period of five breaths was judged to be appropriate in our clinical practice. Because mean values of leak flow and background flow were used, the leak amount was calculated breath by breath. The principle of continuous leak calculation is shown in Fig. 1.

The numerical leak correction (see APPENDIX) was applied in all in vivo investigations and while the accuracy of volume measurements in vitro was being determined. The actual leak was continuously displayed, but it wasn't set to zero until the examiner had pressed a button once. The leak-resulting errors were generally corrected in the flow signal, and the volume was calculated from this corrected flow.

In vitro. Our model to simulate an air leaks is shown in Fig. 2. The spontaneous breathing of \(\sim 15\) ml with adjustable frequency was achieved with a sinusoidal pump (self-made equipment), approximating the pattern of airflow during quiet breathing. The pump-generated flow and the continuous background flow were fed into a face mask with an air-inflatable rim (size 3, Sherwood) that was placed face-side down onto a flat polyvinylchloride panel to achieve a closed seal.

To test the accuracy in air leak measurements according to Eq. 1, the testing setup shown in Fig. 2A was used. The reference leak flow was a vacuum flow at an adjustable rate from 0 to \(-1,200\) ml/min generated by a central vacuum source and adjusted by a precise rotameter (Aalborg, Monkey, NY). The background flow was kept constant at 5 l/min, and the breathing rate was 60 breaths/min.

Volume data were analyzed to test the accuracy of our leak correction algorithm. To evaluate the volume measurement errors under air leak conditions (for setup, see Fig. 2B), we varied the breathing frequency (30, 60, 90 breaths/min), the background flow (3, 5, 7 l/min), and the air leak. The air leak was simulated by a series combination of a constant screen resistance (180 Pa l\(^{-1}\) s\(^{-1}\)) and an adjustable resistance. The linear resistance was built from a linear PT, and the adjustable resistance was a squeezed tube. The air leak settings were 5, 10, 15, 20, 25, and 30%. This was considered an adequate simulation of genuine leaks that may occur during LFT in spontaneously breathing infants, and the realized leak model was similar to what we can expect between a patient's face and mask.

ATPD (ambient temperature and pressure, dry) conditions were used in both setups (Fig. 2, A and B). Any changes in parameter settings followed the storage of at least 10 breaths of the pump.

The in vitro investigations as well as the investigations in patients were carried out by using our self-made measurement equipment (6) on the basis of the FTT. The data-acquisition and -processing software was written by using LabView for Windows (National Instruments). The flow signals were sampled at 200 Hz with the Multifunction I/O Board AT-MIO 100 (National Instruments) and filtered with analogous fourth-order Bessel filters with a 48-Hz cutoff frequency. The volumes were calculated off-line. Electronic drifts of the transducers were minimized by using specially selected Sensym transducers (SenSym) for differential pressure measurements on the Screen-PTs (Transducer Filter Unit, J. Jaeger). The PTs (Screen-PT for Babies, 180 Pa l\(^{-1}\) s\(^{-1}\), J. Jaeger) were sufficiently linear in the flow range of 0 to \pm 15 l/min, and the deviation from linear characteristic was not greater than 2%. The equipment's offset was compensated for after a 30-min warm-up period by switching off both PTs with solenoid valves for 2 s, measuring the voltage, and subtracting these values from the measured signals until the next offset procedure was done. Both PTs were calibrated simultaneously in a two-point calibration procedure without linearization with room air by using a 100-ml calibration syringe (Hans Rudolph). Temperature and humidity were typed into the measurement software and corrected for calibration and
measurement conditions. The PTs were not heated, either in vitro or in vivo. The accuracy of ventilatory measurements with use of our equipment has been tested previously (6), and we found <3% volume errors when no leak was present up to background flow rates of 7 l/min. In this study, the flow error was not explicitly tested.

Statistics. The measurement procedures in our model were completed twice, and each recorded volume value was the average of 10 measured values. SD and coefficient of variation were also calculated into each series. The results of accuracy in air leak measurements with use of our equipment compared with the adjusted negative leak flow were evaluated by using the method of Bland and Altman (1). Volume measurement errors were calculated as the difference between measured and known volume in relation to the known volume. The significance of factors like background flow and breathing rate was investigated by using ANOVA. Statgraphics software (Manugistics) was used for statistical evaluation. A level of statistical significance of \( P < 0.05 \) was accepted.

RESULTS

Figure 3 shows the comparison of the referenced air leak (adjusted by the high-precision rotameter) and the leak measured by our equipment carried out to validate the accuracy of air leak measurements. As can be seen, there was very good agreement \((r = 0.9998; P < 0.001)\) between both air leak measurements. As shown in Fig. 4, the mean difference amounted to −0.1% and the limits of agreement were −0.3 and 0.1%, respectively.

Errors in volume measurement due to leaks are shown in Fig. 5. The measured, noncorrected volumes were generally underestimated, depending on the breathing rate. For a breathing rate of 30 breaths/min and leaks up to 30%, the volume error rose up to 94%. With increasing breathing rate, the volume error decreased due to higher breathing flow. After leak correction, a slight difference between the pumped volume and the corrected volume remained. An increased leak rate (0–30%) led to a significant \((P < 0.001)\) rise in leak-corrected volume error, but not exceeding 5%.

Background flow had a negligible influence on the accuracy of the corrected volume. For background flow rates of 3, 5, and 7 l/min, the mean volume errors (leaks 0–30%) were 1.4, 1.8, and 2.0%, respectively, for a breathing rate of 60 breaths/min. The ANOVA did not show any significant influence of breathing rate and background flow used in our tests on the accuracy of volume measurements after leak correction. In all in vitro measurements, the coefficient of variation was <0.5%.

![Fig. 1. Sliding time window containing last 5 breaths used during measurements to calculate air leak according to Eqs. 1 and 2. \( \dot{V}_{PT0} \) and \( \dot{V}_{PT1} \), pneumotachographs measuring constant background flow (\( V_{const} \)) and outflow from face mask, respectively.](image1)

![Fig. 2. Equipment to validate accuracy of air leak measurements by using constant suction flow (A) and to validate accuracy of leak-corrected tidal volumes (B). Pump simulates spontaneous breathing. See APPENDIX for definitions of abbreviations.](image2)
By using continuous leak observation for ventilatory measurements in infants, in investigated newborns the median of all the measured air leaks amounted to 2.6%, the 10th percentile was at 1.4%, and the 90th percentile was at 4.3%. The maximum leak value was not greater than 7%. Leaks >5% were observed in only 5 of the 67 patients. Table 2 shows the measured and corrected volumes as well as the absolute and relative volume errors sorted by leaks during measurement of ventilation in 67 spontaneously breathing infants. The greatest error that occurred was 16.4%.

**DISCUSSION**

In spontaneously breathing infants, the problems caused by air leaks in ventilation-measuring equipment, and their significance regarding different methods of LFT, have been described for functional residual capacity (FRC) measurements (7, 21, 22), indirect calorimetry (9, 23), and analysis of breathing loops (3).

In contrast to information on ventilated infants where problems caused by air leaks are well known and described (7–9, 22, 20), detailed information on air leak sizes during spontaneous breathing is very rare in the literature (9, 22, 23). During pneumotachographic measurements of ventilation, exact leak recognition and measurement are difficult to perform. The airtight seal of the measurement setup can be tested by carrying out the occlusion test (24) (provided there is complete equilibration of pressure throughout the respiratory system and the Hering-Breuer reflex is present) or by simultaneous ventilatory measurements with the face-out body plethysmograph (4). Thereby, measurements of ventilation with face mask and of volume movement into the face-out body plethysmograph are possible, and thus the leak can be measured. However, this highly complicated technique is mostly reserved for scientific inquiry, and the dead space problem is still present.

Air leaks during measurement of ventilation by using a differential flow system in spontaneously breathing infants lead to underestimated inspiratory and expiratory flow and volume, because the average of the measured flow in the expiratory limb (which represents the patient’s breathing superimposed on constant background flow) is lower than the measured background flow within the inspiratory limb. This is not valid under the conditions of artificial ventilation, whereby inspiratory actions would be mainly measured within the inspiratory limb and expiratory actions in the expiratory limb, respectively. A leak in the endotracheal tube leads to an overestimation of the inspiratory flow and volume, whereas the expiratory flow and volume are underestimated.

Thus, if respiratory mechanics were measured in spontaneously breathing newborns by using a flow-through system and esophageal manometry, compliance would be underestimated and resistance would seem to be higher as opposed to measurements in artificial ventilation, in which compliance would be high and resistance would be low.

If the flow has an offset, then the calculation of compliance and resistance with the regression method may be imperfect (12). Furthermore, this flow offset can
lead to a time shift between the volume and pressure signals due to false breath recognition and can cause significant errors in calculation of compliance and resistance (25).

The use of FTT offers continuous air leak measurement by applying two fundamental principles: 1) the addition of a known tracer gas concentration to the background flow (gas analysis at the input and output of a patient's face mask) (17) and 2) the pneumotachographic determination of the balance of inhaled to exhaled volume. However, the technical resources required for gas analysis are very high and can only be recommended if a multiple-gas analyzer is readily available (e.g., for alveolar gas analysis or for gas-mixing techniques).

The differential flow system used is more sensitive concerning leaks than are other techniques and, therefore, leak recognition and correction are necessary. Moreover, it is hardly ever possible to achieve a completely airtight measurement arrangement.

Leaks that occur during lung-function measurements are very complex (inconstant, time variant, flow and pressure dependent) and, therefore, cannot be calculated easily. Any calculation of leak by the developed algorithm was merely an approach. The in vivo leak measurements were also influenced by physiological factors and by methodological problems: 1) instability of end-expiratory level due to changes in FRC or brought about by sighing; 2) changes in gas viscosity and real differences in volumes between inspiration and expiration and influences of the gas exchange ratio (R); 3) nonlinearities and differences in slopes of transfer characteristics of both PTs; 4) quality of breath recognition; and 5) time interval of leak measurement.

The influences on the leak measurements of a changing FRC level can only be neglected if a very long time interval of analysis is used (a couple of minutes). However, this is not practical, and therefore no explicit determination of FRC-related leaks is possible. Sighing occurs relatively often (2) and is easy to detect by monitoring ventilation. To exclude any errors caused by sighing, we did not report leaks five breaths before and after a sigh.

Volume measurement errors caused by changed viscosity and volume between inspiration and expiration of breathed gas can also appear as leaks. This volume error, $\delta V$, was calculated as previously described (5); first, the volume was converted to BTPS (body temperature, pressure and saturation) and, second, the change in viscosity due to temperature, $O_2$ and $CO_2$ fractions, and humidity was determined. Because of superimposition of exhaled gas on continuous background flow, the error was decreased even more by factor $k$

$$
\delta V = k \cdot \delta V_{BTPS} = \frac{\dot{V}_{exp}}{\dot{V}_{const} + \dot{V}_{exp}} \cdot \delta V_{BTPS} \tag{2}
$$

The $\dot{V}_{exp}$ is equal to $1/2 \cdot \dot{Z} \cdot PTEF$ (e.g., for sinusoidal signals the peak tidal expiratory flow PTEF is equal to peak tidal inspiratory flow PTIF). Background flow was adjusted to always exceed PTIF (16) so that any rebreathing of exhaled $CO_2$ was avoided; thus $k$ was $\leq 0.41$. The volume error due to $R$ was also reduced by factor $k$. Assuming $R = 0.8$, inspiration of 25°C warm and 50% humidified gas and expiration of BTPS gas with 40 Torr $CO_2$ tension, the volume was numerically corrected according to measured PTIF, background flow, and Eq. 2. Because of the continuous background flow, no water condensation occurred and no PT heating was necessary.

The essential preconditions for precisely measured air leaks are the identical linear characteristics of both PTs over the full operating range (18). Differences in the slope of PT characteristics were taken into account during the calibration procedure. Despite electronic drift compensation and calibration, a slight difference between both PTs remained. Under worse conditions, this difference error can appear as a leak. The volume error after leak correction rose significantly with greater leaks, which could have been by PTs working in different ranges of their characteristics. In a flow-through system, the first PT works in a very small range of its characteristic line, whereas the other PT works in a wide flow range. The characteristics of both PTs are never absolutely linear and identical, and the resulting flow and volume error cannot be compensated for by the two-point calibration used.

Furthermore, the quality of breath recognition is essential for ventilatory measurements. Any shift in time leads to an incorrect assessment of balance of inspired to expired volume and can, misleadingly, appear to be a leak. Several algorithms for breath detec-

### Table 2. Measured leaks and volume errors in ventilatory measurement in 67 spontaneously breathing infants

<table>
<thead>
<tr>
<th>Leaks</th>
<th>0% (n = 2)</th>
<th>1% (n = 6)</th>
<th>2% (n = 23)</th>
<th>3% (n = 21)</th>
<th>4% (n = 10)</th>
<th>&gt;5% (n = 5)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Measured volume, ml</td>
<td>10.4 (9.5–11.4)</td>
<td>11.5 (8.5–24)</td>
<td>13.7 (8.1–29.1)</td>
<td>13.9 (8.4–26.1)</td>
<td>18.2 (9.8–21.3)</td>
<td>11.4 (8.9–13.3)</td>
</tr>
<tr>
<td>Corrected volume, ml</td>
<td>10.5 (9.5–11.5)</td>
<td>11.9 (8.7–24.1)</td>
<td>13.9 (8.4–30.8)</td>
<td>14.3 (9.6–26.8)</td>
<td>19.3 (10.7–22.1)</td>
<td>13.0 (9.8–14.11)</td>
</tr>
<tr>
<td>Absolute error, ml</td>
<td>0.05 (0.0–0.5)</td>
<td>0.4 (0.3–0.9)</td>
<td>0.5 (0.1–1.8)</td>
<td>0.6 (0.3–2.6)</td>
<td>0.8 (0.1–1.0)</td>
<td>1 (0.4–2.3)</td>
</tr>
<tr>
<td>Relative error, %</td>
<td>(0.0–0.4)</td>
<td>(1.5–5.9)</td>
<td>(0.9–12.9)</td>
<td>(1.7–15)</td>
<td>(2.0–9.2)</td>
<td>(3.1–16.4)</td>
</tr>
</tbody>
</table>

Median values are shown, with range in parentheses. $n$, No. of infants. Absolute error = measured volume – measured volume. Relative error = 100 × absolute error/corrected volume.
tion were tested in our laboratory (19), and the one that was most robust regarding disturbances was used.

The time interval \( n \cdot T \) has to correspond to the exact duration of \( n \) breathing cycles; otherwise, volumes of different phases of breaths would be taken into account. The number of breaths used for leak calculation can be chosen arbitrarily. However, the calculation over a very long period of time prolongs the response time of the leak-measuring algorithm enormously, and short-term changes in air leaks are not reflected. On the other hand, if the calculation period selected is too short (e.g., breath by breath), leak measurement will produce widely scattered leak values.

During model investigations, we found, in order of magnitude, comparable differences between the adjusted and the measured leak independent of leak rates (see Fig. 4). The slight bias can be explained by ideal laboratory conditions. However, the errors in the corrected volume were higher and increased with rising leak flow up to 5% (see Fig. 5). The reason for this seeming contradiction to the accurately estimated leaks in the first part of the in vitro studies could be the differences between both models used. The leak flow conditions in both models were different. In the first part of the in vitro studies, a constant suction flow was used to simulate a leak, whereas in the second part the leak was adjusted by a nonlinear resistance that depends on the leak flow. For in vivo investigations, however, the accurate measurement of minor leaks becomes more difficult, and only leaks \( >2\% \) can be interpreted.

We limited leak size to a range between 0 and 30% for the in vitro studies because the equipment we used only detects leaks of \( \leq 35\% \). The measured flow signal is the difference between both PT signals, and, if there are higher leaks, this flow signal no longer crosses the baseline, but the zero crossing is needed to recognize breaths. The in vivo investigations have shown that the 30% limit is not a restriction because only leaks within the range of 0 to 7% were observed.

The algorithm developed for air leak correction is independent of the breathing pattern and the shape of the respiratory signals and can also be applied to infants with continuous positive airway pressure. Therefore, the sinusoidal flow signals used, with negligible breath-to-breath variability, do not have a significant influence on the results of the in vitro investigations.

The practicability of the leak-correction algorithm has been tested in vivo. Clinical experience showed that, after the implementation of our leak-measurement algorithm, more attentive care regarding the correct placement of the face mask was achieved.

Of course, the face mask leaks in measuring ventilation in spontaneously breathing newborns will vary over time. Rapid changes in leaks cannot be recognized by the algorithm because the response time of the algorithm is five breaths long. Slow changes in the leak will be detected, and the examiner can improve the placement of the face mask; otherwise, if improved placement is impossible, the numerical leak correction has to be used. Retrospective analysis of data measured in vivo showed that air leaks decreased because the physician had more experience in the application of this method.

**APPENDIX**

Correction of air leaks by using the FTT is based on the following preconditions: 1) leak resistance is assumed to be constant; 2) the time interval \( n \cdot T \) for leak calculation must correspond to the exact duration of \( n \) breathing cycles; otherwise, volumes of different phases of breaths would be taken into account; and 3) the characteristics of both PTs have to be identical.

The differential flow \( \dot{V}_{diff}(t) \) in the FTT is the difference between both measured inflow and outflow of the face mask. An occurring leak rate affects this \( \dot{V}_{diff}(t) \) by a change in outflow \( \dot{V}_{PT}(t) \), which is the patient's breathing flow \( \dot{V}_{pat}(t) \) superimposed on the constant background flow \( \dot{V}_{const}(t) \) as follows:

\[
\dot{V}_{PT}(t) = \dot{V}_{pat}(t) + \dot{V}_{const}(t) \cdot (1 - \text{leak})
\]

Not only can an offset of the differential flow be observed, but the signal magnitude is also reduced.

\[
\dot{V}_{diff}(t) = \dot{V}_{PT}(t) - \dot{V}_{const}(t) = \dot{V}_{pat}(t) \cdot (1 - \text{leak}) - \dot{V}_{const}(t) \cdot \text{leak}
\]

The faulty \( \dot{V}_{diff}(t) \) can be corrected, first, by addition of offset and, second, by correction of magnitude due to multiplication by the inverse of factor \( (1 - \text{leak}) \) so that the corrected \( \dot{V}_{diff}(t) \) only contains the patient's flow

\[
\dot{V}_{diff}^*(t) = [\dot{V}_{diff}(t) + \dot{V}_{PT}(t) \cdot \text{leak}] \cdot \frac{1}{1 - \text{leak}} = \dot{V}_{pat}(t)
\]

Consequently, the integration of leak-corrected differential flow \( \dot{V}_{diff}(t) \) gives an air leak-corrected volume.

The authors are grateful to S. Schmidt for assistance in collecting data.

This work was supported by the Bundesministerium für Bildung, Wissenschaft, Forschung und Technologie project "Perinatal Lung (01Z29511)" and by Deutsche Forschungsgemeinschaft (Schm 1160/1-2). Address for reprint requests: B. Foitzik, Dept. of Pediatrics (Charité), Schumannstr. 20/21, 10098 Berlin, Germany (E-mail: foitzik@rz.charite.hu-berlin.de).

Received 10 November 1997; accepted in final form 22 April 1998.

**REFERENCES**


5. Foitzik, B., G. Schmalisch, and R. R. Wauer. Effect of physical properties of respiratory gas on pneumotachographic