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Automated, real-time calibration of the respiratory inductance plethysmograph and its application in newborn infants

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Abstract

Respiratory inductive plethysmography (RIP) is widely used in infants, children and adults. The technique is well accepted as it provides important qualitative information on the pattern of breathing, although its ability to record volume accurately was questioned due to calibration uncertainties. Existing calibration methods require two-position calibration, or patient cooperation in performing various breathing manoeuvres, or prolonged calibration paradigms. The disadvantages from calibration difficulties are even more pronounced in infants.

We present a computer system that is capable of performing a single-posture, real-time RIP calibration during natural breathing and is suitable for use in newborns. The calibration algorithm is based on interactive, point-by-point calculations of maximal correlation between airflow at the mouth, \( V_{ao} \), and summed differentiated RIP signals. The quantities are calculated interactively at every sample point, and the process continues until stable results are reached and convergence criteria are met. A graphic user interface was developed to assist in the rapid implementation and ease of use. Validation schemes were evaluated in 33 newborn infants against actual \( V_{ao} \).

Calibration factors were obtained within 21 ± 11 s with a mean correlation coefficient of 0.97 ± 0.03. All RIP-derived values were similar to actual airflow signals, with error values ranging from 0.4 ± 3.0% for respiratory rate to 1.8 ± 7.3% for tidal volume. Calibration was found to be stable and reliable for up to 3.5 h and in changing sleep states.

It is concluded that the new single-posture real-time RIP calibration system is safe and simple to use, and also quick, accurate and stable. The system was found to be suitable for use in newborns during natural breathing while asleep.
Keywords: real time, single posture calibration, correlation and convergence, airflow, pattern of breathing, REM, non-REM sleep

1. Introduction

The assessment of ventilatory function including breathing patterns, respiratory rates, tidal volume, and inspiratory and expiratory flows is possible without physical connection to the airway. Respiratory inductive plethysmography (RIP) is the most widely accepted method for quantitative and qualitative noninvasive respiratory measurements in adults and children. When correctly calibrated, RIP allows the measurements of volume and time components of the breathing cycle as well as the relative contribution of rib cage and abdomen (Cohn et al 1978, Allen et al 1990). In 1967, Konno and Mead demonstrated that movements of the respiratory system could be approximated with two degrees of freedom of motion such that volume changes at the mouth (open system) were equal to the sum of volume changes of the rib cage and abdominal compartments. Occlusion of the oro-nasal orifices (closed system) produced only one degree of freedom of motion, where any changes of the rib cage component were equal and opposite to changes of abdominal component—the isovolume manoeuvre. The majority of the earlier calibration techniques are based on the Konno and Mead model (Konno and Mead 1967) and utilize simultaneous recording of breathing volume by spirometer or pneumotachograph (PNT) (Sackner et al 1980, Duffty et al 1981, Chadha et al 1982).

Untrained subjects, however, often find it difficult to perform the isovolume manoeuvre, and other calibration procedures that do not require this special breathing manoeuvre have been advocated (Stagg et al 1978, Loveridge et al 1983, Warren and Alderson 1985). More recently, Sackner et al (1989) suggested a single-posture method for deriving the proportionality constant between rib cage and abdominal amplifiers of the respiratory inductive plethysmograph. Their qualitative diagnostic calibration (QDC) is based on equations of the isovolume manoeuvre calibration and is carried out during a 5 min period of natural breathing. The proportionality constant approximates the ratio of standard deviations of the non-calibrated changes of abdomen to rib cage volume deflections. The QDC method became very popular, as it requires no subject cooperation utilizing a single-posture calibration scheme. However, recent publications have raised doubts as to its reliability (Sartene et al 1993, Brown et al 1998, Thompson 1999, De Groot et al 2001).

The present work describes a real-time RIP calibration computer system based on a new single-posture calibration model, which is suitable for use in newborn infants. The system is simple, quick and accurate, and requires no patient cooperation nor is it limited to constant volume amplitudes.

2. Theory

Assuming two degrees of freedom (Konno and Mead 1967), changes in lung volume as measured by RIP are equal to the sum of rib cage and abdominal volume displacements:

\[ V_{\text{rip}} = V_{\text{rc}} + V_{\text{ab}} \]  

(1)

each of these volume displacements can be expressed as

\[ V_{\text{rc}} = A R_{\text{rc}} + C_1 \]

\[ V_{\text{ab}} = B R_{\text{ab}} + C_2 \]

where \( R_{\text{rc}} \) and \( R_{\text{ab}} \) are RIP signals for rib cage and abdomen, respectively, \( A \) and \( B \) are channel gains, and \( C_1 \) and \( C_2 \) are constants.
The calibration procedure first determines the relative gain of the rib cage and abdominal RIP signals (equivalent to the slope of the isovolume line), and then relates by linear regression the derivatives of RIP signals to a known airflow signal record.

The first step of the calibration procedure is to maximize the correlation between the airflow signal and the flow signal determined by differentiation of the RIP signals. Since the correlation is a linear operation we can rearrange equation (1):

\[
\frac{1}{A} V_{\text{rip}} = R_{\text{rc}} + K R_{\text{ab}} + \frac{C_1 + C_2}{A}
\]  

where \( K = B / A \) is a constant representing the relative gain of the rib cage and abdomen RIP signals. Differentiating the above equation with respect to time will yield the relationship for flow:

\[
\frac{1}{A} \dot{V}_{\text{rip}} = \dot{R}_{\text{rc}} + K \dot{R}_{\text{ab}}
\]  

denoting: \( x = \dot{R}_{\text{rc}}, y = \dot{R}_{\text{ab}} \) and \( z = \frac{1}{A} \dot{V}_{\text{rip}} \), equation (3) becomes \( z = x + Ky \).

The correlation coefficient \( r(f, z) \) of airflow at the mouth against the RIP-derived flow is defined by

\[
r(f, z) = \frac{r(f, x) + KSr(f, y)}{\sqrt{1 + K^2 S^2 + 2KSr(x, y)}}.
\]

See appendix for detailed derivation.

To find \( K \), which maximizes the correlation between the airflow signal and the signal derived by differentiation of the RIP signals, \( r(f, z) \), equation (4) is differentiated with respect to \( K \) and equated to zero. Then the maximum value for \( r(f, z) \) is given by

\[
K = \frac{r(f, y) - r(f, x)r(x, y)}{S[r(f, x) - r(f, y)r(x, y)]}
\]

The second step of the calibration procedure utilizes the standard expression for linear regression of \( f \) upon \( z = \frac{1}{A} \dot{V}_{\text{rip}} = \dot{R}_{\text{rc}} + K \dot{R}_{\text{ab}} \):

\[
\left[ f - \text{mean}(f) \right] = \frac{r(f, z)S_f}{S_z} [z - \text{mean}(z)]
\]

and, using this line to give values for a RIP signal \( \dot{V}_{\text{rip}} \) (i.e., setting \( \dot{V}_{\text{rip}} = f \)), we get

\[
\dot{V}_{\text{rip}} = \frac{r(f, z)S_f}{S_z} [z - \text{mean}(z)] + \text{mean}(f).
\]

A direct comparison with the original expression for \( \dot{V}_{\text{rip}} \) (equation (1)) yields

\[
A = r(f, z)S_f \frac{S_f}{S_z}
\]

\[
B = AK = r(f, z)S_f K \frac{S_f}{S_z}
\]

\[
C = \text{mean}(f) - r(f, z)S_f \frac{S_f}{S_z} \text{mean}(z).
\]

All the quantities in these expressions have already been calculated. The quantity \( S_z = S(x + Ky) \), as was shown above, can be determined from \( S_z^2 = S_z^2 + K^2 S_y^2 + 2KS_z r(x, y) \).

3. Implementation

The following is a detailed description of the computer software program for implementing the calibration algorithm in real time. Amplified RIP and pneumotachograph signals were sampled and then processed by the RIP calibration program to yield simultaneous flow signals (\( \dot{V}_{\text{rip}} \)), comprised of \( \dot{R}_{\text{rc}} \) and \( \dot{R}_{\text{ab}} \), respectively.
Figure 1. The Respitrace system. The two winding coils within the elastic bands are activated by a low current oscillator (20 mV at 300 kHz) to produce electromagnetic fields whose inductances vary with body circumference. Output voltages are demodulated and presented in digital form.

3.1. Hardware

The subjects were fitted with elastic bands around their rib cage and abdomen as shown in figure 1. These bands served as sensors providing analogue voltage output signals corresponding to changes in waveform amplitudes during respiration. The output signals were demodulated by the Respitrace® microprocessor module (Respitrace®, Ambulatory Monitoring Inc., Ardsley, NY, USA) and passed through additional conditioning and amplification (medium gain amplifier 8802A, Hewlett-Packard Co., Waltham, MA, USA). For the short period of airflow recording that is necessary for calibrating the RIP, the subject breathes through a soft-rim, tightly sealed face mask (Laerdal Medical, Stavanger, Norway) and a heated pneumotachograph (Fleisch), connected to a differential pressure transducer (±2 cm H₂O, MP-45, Validyne Engineering Corp., Northridge, CA). Amplified RIP and \( V_{ao} \) signals were sampled at rates of up to 100 Hz by an analogue-to-digital converter on a data acquisition board (DAS-802, Keithley MetaByte, Taunton, MA; figure 2). The signals were then processed by the RIP calibration program to yield simultaneous flow signals. No further calibration of the system is necessary, provided that there are no major changes in the subject’s posture.

3.2. Software

A real-time Microsoft Windows application program written in C (Borland® C++ compiler v4.52) was developed. The program performs data acquisition, conditioning (digital filtering and differentiation) and calibration algorithm implementation. The acquired signals and calibration results are displayed on the computer monitor in real time for data acquisition and calibration process monitoring. The Matlab® High-Performance Numeric Computation and Visualization Software was used as an aid for data acquisition. A graphic user interface (GUI) was developed for user convenience and simplicity of use. The self-explanatory GUI provides the user with complete control over each procedural step.
3.3. Online visualization and calculating scheme

The system simultaneously acquires three analogue signals: pneumotachograph flow \( F \) and two RIP signals, \( R_{rc} \) and \( R_{ab} \), which are displayed on the computer screen in real time while calibration algorithm calculations take place (as described in section 2). The analysis yields the gains ratio \( K \) and \( r(f, z) \) (equations (4) and (5)) as well as gains \( A \), \( B \) and the dc shift \( C \) (equation (8)). \( \dot{V}_{rip} \) is then reconstructed from equation (3). The program calculates these quantities interactively in real time over a region which increases in length by one sample at a time and produces consecutive values of \( K \) and \( r(f, z) \). The process continues until stable convergence criteria are reached.

Prior to implementing the calibration procedure, the data had been filtered by low-pass digital filters to reduce noise and artefacts, and RIP signals had also been integrated at every sample point. For the low-pass filter, we used a moving average (median) filter \( F(n) \), \( N = 10 \), which we chose for its convenience in digital implementation and low time delay (a critical feature in real-time applications):

\[
F(n) = \begin{cases} 
0 & n \geq N \\
\frac{1}{N} \sum_{k=0}^{N-1} x(n-k) & \text{otherwise}
\end{cases}
\]

Following low-pass digital filtering, the RIP signals were numerically differentiated by the least-squares rather than the interpolation polynomials method since it is less affected by random errors during empirical data handling (Korn 1961). This yielded simultaneous flow signals, \( \dot{V}_{rip} \), comprised of \( R_{rc} \) and \( R_{ab} \).

3.4. Recurrence relationships

The calibration procedure requires many statistical manipulations which are time consuming. Application of the following recurrence relationships allows for greatly reduced computer
computation time. Define $D = \bar{x}_{j-1} - x_j$, i.e., the difference between the mean value for $x$ calculated up to the $(j - 1)$th point and the value of $x$ at point $j$. Then

$$\bar{x}_j = \bar{x}_{j-1} - \frac{D}{j}.$$  \hspace{1cm} (9)

Thus, the mean value after sampling the $j$th point can be determined from the difference between the mean value for the $(j - 1)$ points and the quantity $x_j$. Recurrence relationships expressing a value at point $j$ as a function of the $(j - 1)$th value and quantity $D$ could be readily derived for variance and covariance:

$$S_j^2 = \frac{1}{j} \left( S_{j-1}^2 + \frac{D^2}{j} \right)$$

$$\text{cov}_j(x, y) = \frac{1}{j} \left( \text{cov}_{j-1}(x, y) + \frac{D_x D_y}{j} \right)$$

where

$$D_x = \bar{x}_{j-1} - x_j$$

$$D_y = \bar{y}_{j-1} - y_j.$$  

Hence, at each iteration step, i.e., at each sample point, we can compute variance and covariance directly from the values of the previous step.

3.5. Breath detection

A simple and accurate algorithm for the recognition of respiratory pattern was used for the subsequent computation of various respiratory indices. The respiratory cycle was considered as consisting of two respiratory phases, inspiration and expiration. An integral is a piecewise monotonic function increasing during inspiration and decreasing during expiration. Hence, maximum inspiratory and expiratory flows (local minima or maxima) are identified during flow integral increases or decreases, respectively. In practice, this criterion is checked on sequential points for which monotony of the integral function exists. The number of points in this window is chosen arbitrarily, depending on sampling frequency and respiratory rate, so as to eliminate various artefacts but not to fail in detecting short inspirations or expirations.

After all local maximal and minimal locations (designated as 1 and $-1$) were identified, an index matrix was built. An example of a flow signal and its index matrix is presented in figure 3. The template matrix $[-1, 1, -1, 1]$ represents a respiratory cycle in a symbolic form and respiratory cycle locations are identified every time this template coincides with the index matrix. After each respiratory cycle is located and defined, respiratory indices such as tidal volume, inspiratory and expiratory times, and maximum inspiratory and expiratory flows can be easily computed.

3.6. Validation

All respiratory parameters were determined for each breath and mean, SD and range were calculated over the full 1 min epoch for each infant. For validation of our calibration model, we determined and compared the values pertaining to the amplitude-tidal volume ($V_T$), the frequency of respiration ($f$), the time period ($T_{tot}$), the inspiratory and expiratory times ($T_i, T_e$), and the duty cycle ($T_i/T_{tot}$). Three validation schemes were performed:

1. Following the calibration procedure, a 1 min epoch of quiet breathing was recorded and analysed. RIP derived flows and volumes were compared to simultaneous recordings of actual flow at the mouth, $\dot{V}_{ao}$, and to derived volume, $V_{ao}$. Examples of such correlations are presented in figures 4 and 5 for two infants with the best and worst correlation coefficients. In figure 4, a 1 min recording of RIP derived flows, $\dot{V}_{rip}$ (1500 sampled values), are plotted against actual flow rates at the mouth, $\dot{V}_{ao}$. In figure 5, 10 s of $\dot{V}_{rip}$ tracings are superimposed over $\dot{V}_{ao}$.  

Automated Respitrace calibration for use in infants

Figure 3. Respiratory pattern analysis. Example of flow signal and its ‘index matrix’ used for respiratory cycle recognition. This template represents a respiratory cycle in symbolic form, since every time this template coincides with the index matrix, a respiratory cycle location is found. Consequent concurrences with the index matrix yields the consequent respiratory cycles location.

(2) Further validation was performed by checking the accuracy and stability of the present calibration scheme in 1 min sessions recorded 30 ± 40 min (range 5–217 min) after calibration in 24 infants.

(3) To add another degree of freedom for our validation, the RIP calibration scheme was tested in 11 infants in whom a single calibration epoch was used for calculations of recordings that were obtained during both REM and non-REM sleep states.

(4) Finally, the quality of our algorithm was tested by comparing the results of the present study against the results obtained by running the algorithm previously suggested by Stagg et al (1978).

3.7. Subjects

Thirty three (19 males/14 females) healthy, full-term newborns were tested. All had an uneventful delivery at a mean ± SD gestational age of 40.1 ± 1.3 weeks (range 37–44) and a mean weight of 3332 ± 355 g (range 2640–3970). The infants were tested 24.3 ± 10.0 h after delivery (range 2–52) when their mean weight was 3235 ± 334 g (range 2640–3890). They were all asleep in the supine position at 30–60 min post feeding. Sleep phases (i.e., REM versus non-REM) were confirmed in all by behavioural criteria (Prechtl 1974). The institutional ethics committee approved the study and maternal consent was obtained in all cases.

4. Results

The calibration model was tested against a 1 min calibration record and the results were compared. As can be seen from table 1, the mean convergence time for the short calibration
model (21 ± 11 s, range 8.8–60 s) was significantly shorter than the full, 1 min model (p < 0.001). Only six studies required over 30 s for convergence and only one required the full 60 s. This was achieved with no loss of accuracy, with values of $K$ and of cross-correlation coefficients, $r(f, z)$, being similar. Goodness of fit of the model was judged to be satisfactory, with a mean ± SD correlation coefficient, $r(f, z)$, of 0.97 ± 0.03 and with an $r(f, z) < 0.9$ in only one study. Flow signals derived from the best and worst correlation coefficients are presented in figure 4.

To judge the reliability and applicability of the calibration model, different respiratory indices (tidal volume and timing events) were calculated using the breath detection algorithm.
described above. These indices were derived from actual airflow signals measured at the mouth and were compared to flow signals reconstructed from the RIP signals (table 2). The differences between the two determinations were minute for all derived parameters with error values (%) ranging from 0.4 ± 3.0 for respiratory rate to 1.8 ± 7.3 for tidal volume, $V_T$. None of the differences were statistically significant except for $T_{tot}$ ($p = 0.04$), but this was judged to be physiologically negligible. Further validation was done in 24 infants in 1 min sessions recorded 30 ± 40 min (range 5–217) after calibration was performed (table 3). As can be seen, mean ± SD tidal volume ($V_T$) and minute ventilation ($V_E$) measured by RIP were comparable to those measured at the mouth with no loss of calibration stability or accuracy.
Table 1. Mean ± SD values of cross-correlation coefficients, $r_{xz}$, for the calibration procedures in 33 subjects. Two calibration procedures have been employed: the short calibration was performed on the shortest possible duration (see text for further details), and the 1 min calibration was performed on a full minute epoch.

<table>
<thead>
<tr>
<th></th>
<th>Short calibration</th>
<th>1 min calibration</th>
</tr>
</thead>
<tbody>
<tr>
<td>$K$</td>
<td>$r_{xz}$</td>
<td>Time (s)</td>
</tr>
<tr>
<td>Mean ± SD</td>
<td>1.07 ± 0.66</td>
<td>0.967 ± 0.026</td>
</tr>
<tr>
<td>Range</td>
<td>−0.11–3.02</td>
<td>0.849–0.996</td>
</tr>
</tbody>
</table>

Table 2. Mean ± SD (range) values of tidal volume ($V_T$), respiratory rate ($f$), respiratory period ($T_{tot}$), inspiratory and expiratory timing ($T_i, T_e$), and duty cycle ($T_i/T_{tot}$). RIP derived values are compared to those measured at the mouth, AO. Individual results were calculated and averaged over a 1 min period.

<table>
<thead>
<tr>
<th></th>
<th>AO</th>
<th>RIP</th>
<th>$p$</th>
</tr>
</thead>
<tbody>
<tr>
<td>$V_T$ (ml)</td>
<td>17.0 ± 3.95 (10.4–26.2)</td>
<td>16.7 ± 4.0 (10.6–27.6)</td>
<td>0.150</td>
</tr>
<tr>
<td>$f$ (min$^{-1}$)</td>
<td>47.7 ± 13.0 (20.0–83.0)</td>
<td>47.6 ± 13.6 (19.0–86.0)</td>
<td>0.768</td>
</tr>
<tr>
<td>$T_{tot}$ (s)</td>
<td>1.27 ± 0.27 (0.71–1.67)</td>
<td>1.26 ± 0.27 (0.69–1.60)</td>
<td>0.044</td>
</tr>
<tr>
<td>$T_i$ (s)</td>
<td>0.58 ± 0.12 (0.37–0.84)</td>
<td>0.57 ± 0.12 (0.36–0.84)</td>
<td>0.654</td>
</tr>
<tr>
<td>$T_e$ (s)</td>
<td>0.69 ± 0.17 (0.33–1.00)</td>
<td>0.68 ± 0.18 (0.33–0.98)</td>
<td>0.502</td>
</tr>
<tr>
<td>$T_i/T_{tot}$ (%)</td>
<td>45.5 ± 4.1 (33.5–57.5)</td>
<td>45.6 ± 4.6 (38.3–60.4)</td>
<td>0.899</td>
</tr>
</tbody>
</table>

Table 3. Validation performed on 1 min sessions recorded 30 ± 40 min (range 5–217) after calibration in 24 infants. Mean ± SD and range values of tidal volume ($V_T$) and minute ventilation ($\dot{V}_E$) are measured by RIP, are compared here to those measured at the mouth, AO. Mean ± SD values of the differences (%) are also presented. See text for further details.

<table>
<thead>
<tr>
<th></th>
<th>AO</th>
<th>RIP</th>
<th>Difference (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>$V_T$ (ml)</td>
<td>Mean ± SD</td>
<td>18.3 ± 4.4</td>
<td>17.9 ± 4.0</td>
</tr>
<tr>
<td></td>
<td>Range</td>
<td>11.1–27.0</td>
<td>11.9–27.9</td>
</tr>
<tr>
<td>$\dot{V}_E$ (ml min$^{-1}$)</td>
<td>Mean ± SD</td>
<td>877 ± 153</td>
<td>891 ± 212</td>
</tr>
<tr>
<td></td>
<td>Range</td>
<td>644–1238</td>
<td>521–1296</td>
</tr>
</tbody>
</table>

Mean (±SD) $r_{xz}$ value for the 11 infants in whom recordings were available during both sleep states were 0.959 ± 0.041 in NREM and 0.957 ± 0.027 in REM sleep ($p = 0.81$). Likewise, the time for convergence was similar (23.0 ± 16.1 and 19.3 ± 9.7; $p = 0.55$). It can be further seen (table 4) that percentage errors between RIP-derived variables and those derived from actual flow recording were of the same order of magnitude.

In order to further test the present method, the same respiratory parameters described above were calculated by the calibration scheme suggested by Stagg et al (1978) from a full minute recording in all 33 infants. For brevity, only $V_T$ and minute ventilation ($\dot{V}_E$) are presented in table 5. As can be seen, the method of Stagg et al yielded significantly lower $r_{xz}$ values ($p < 0.001$), being less than 0.9 in 13 infants compared to only in one in the present study. As a result, the differences in respiratory parameters obtained from that analysis were greater than those obtained by the present method, but due to large variability did not reach significant levels. For example, differences in $V_T$ greater than 20% were found in five of the infants compared to none in the present study. The discrepancies were even more pronounced...
Table 4. Comparison of the differences between values determined from RIP versus actual flow measurements for 11 infants in whom measurements were made during non-REM and REM sleep. Differences are presented in absolute values and as a percent of pneumotachograph-derived values. All details are similar to those in table 3.

<table>
<thead>
<tr>
<th></th>
<th>Non-REM</th>
<th>REM (n = 13)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Difference (%)</td>
<td>Difference (%)</td>
</tr>
<tr>
<td>$V_T$ (ml)</td>
<td>Mean ± SD</td>
<td>0.5 ± 1.4</td>
</tr>
<tr>
<td></td>
<td>Range</td>
<td>−3.0–3.9</td>
</tr>
<tr>
<td>$\dot{V}_E$ (ml min$^{-1}$)</td>
<td>Mean ± SD</td>
<td>15.9 ± 58.5</td>
</tr>
<tr>
<td></td>
<td>Range</td>
<td>−115–142</td>
</tr>
</tbody>
</table>

Table 5. Mean ± SD and range values of various breathing parameters in 33 infants as measured at the mouth, AO, and by RIP (see previous tables for definitions). These were obtained using either the algorithm suggested by Stagg et al (1978) or that of the present study. In each case, proportionality constants were determined by the short calibration procedure and used for comparison during 1 min sampling period. See text for further details.

<table>
<thead>
<tr>
<th></th>
<th>Present analysis</th>
<th>Stagg et al (1978)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Difference (%)</td>
<td>Difference (%)</td>
</tr>
<tr>
<td>$r_f$</td>
<td>0.967 ± 0.026</td>
<td>0.873 ± 0.119</td>
</tr>
<tr>
<td></td>
<td>0.849–0.996</td>
<td>0.526–0.995</td>
</tr>
<tr>
<td>$V_T$ (ml)</td>
<td>16.9 ± 3.9</td>
<td>16.4 ± 4.1</td>
</tr>
<tr>
<td></td>
<td>10.4–26.2</td>
<td>10.6–27.6</td>
</tr>
<tr>
<td>$\dot{V}_E$ (ml min$^{-1}$)</td>
<td>825 ± 164</td>
<td>808 ± 180</td>
</tr>
<tr>
<td></td>
<td>570–1145</td>
<td>482–1157</td>
</tr>
</tbody>
</table>

for $\dot{V}_E$ where differences greater than 20% were found in 13 infants compared to only two in the present study.

5. Discussion

We present a computer system that is capable of performing a single-posture real-time RIP calibration during natural breathing and is suitable for use in infants. The calibration procedure was tested on 33 healthy newborns during natural sleep, and stable calibrations were reached within a matter of seconds. Goodness of fit of the present technique was judged to be satisfactory with a high cross-correlation coefficient ($r_f > 0.9$ in all but one study). All RIP-derived respiratory indices were similar to those derived from actual airflow signals with a maximal error value of less than 2% for $V_T$. This goal was achieved with the rapid calibration scheme which was found to be as good as the full 1 min calibration model, stable for up to 3.5 h, and insensitive to changing sleep states. It is noted, however, that this stability was only demonstrated in healthy infants who even in REM sleep may have little chest wall to abdominal asynchrony. It would, therefore, be important to assess the calibration technique in very preterm infants or those with significant lung disease in whom asynchrony is known to be a much bigger problem.

The errors in $V_T$ (RIP) values reported by us deviated by only 0.3 ± 1.3 ml. Percentage wise, these were similar to or even better than values of previous studies on trained adult
populations (Sackner et al 1980, Chadha et al 1982, Zimmerman et al 1983, Stromberg et al 1993). Thus, it was reassuring to find that we could keep the errors to the same level as in adults in whom these discrepancies amounted to 100 ml or greater.

The single-posture calibration is obviously the method of choice in non-cooperating subjects such as infants. Stagg et al (1978) suggested a single-posture calibration procedure using magnetometers, in which the calibration factors were determined by linear regression analysis of RIP signals to integrated flow signals at the mouth during 56–160 quiet breaths recorded in four normal adults and one with diaphragmatic paralysis. Loveridge et al (1983) employed a similar calibration technique and showed a good agreement in eight normal adults. They also showed the technique to be stable for up to 60 min provided the subjects stayed in the same posture (seated) and did not change their pattern of breathing. The present calibration procedure is an extension of the model suggested by Stagg et al (1978). As can be seen in table 5, the method of Stagg et al yielded significantly lower $r_{x}$ values ($p < 0.001$), being <0.9 in 13 infants compared to only one in the present study. As a result, the differences in respiratory parameters obtained from that analysis were greater than those obtained by the present method, but due to large variability did not reach significant levels. These findings, which could not be attributed to the goodness of fit of the calibration scheme, can be explained by the fact that Stagg et al correlated volume signals, whereas the present algorithm optimized for flow. It is our belief that the reason for the large discrepancy in $V_{T}$ (>100% in one infant) was due to wide fluctuations observed in end-expiratory levels which are not accounted for when using volume correlations, and it is possible that the agreement reported by Stagg et al in their original report was due to the fact that they studied normal, highly trained adults exhibiting very little end-expiratory level variability. It is interesting to note that Hudgel et al (1984) were successful in reaching a fair agreement between actual measures of $\dot{V}_{AO}$ and $\dot{V}_{RIP}$, but could not reasonably detect induced changes in end-expiratory levels with errors as high as 30.7%. Since the present calibration scheme, which is based on flow and not volume recordings, is unaffected by end-expiratory level fluctuations, the present method seems most suitable for investigations in infants.

More recently, Sackner et al (1989) introduced the qualitative diagnostic calibration method, which is based on up to 5 min sampling of quiet breathing without the need for simultaneous measurements of actual volume changes at the mouth. They found their method to be as effective as the isovolume manoeuvre with significant errors averaging 3.4%. The same authors later tested their method in newborns and found it to be as effective, but they only reported data of mean $V_{T}$ values obtained from sets of ten breath composites (Adams et al 1993a, 1993b). Their results were encouraging in that the mean difference in $V_{T}$ amounted to 0.5 ml just as in the present study, but the variability of the individual differences in $V_{T}$ was much higher (SD of 8.5 versus 1.3 ml). However, their calibration procedure lasted 5 min and was satisfactory only in regularly breathing normal adults. In their hands, the degree of error increased during irregular breathing patterns or during shorter calibration periods, reaching a level of discrepancy greater than 20% in roughly one-quarter of the cases for a 1 min calibration procedure. These concerns and further doubt as to its reliability were also raised in recent publications (Sartene et al 1993, Brown et al 1998, Thompson 1999, De Groot et al 2001).

A short calibration procedure has obvious advantages, especially in newborns. It is important for other reasons which may affect the quality of calibration. The breathing pattern of infants may be affected by the instrumentation itself and its dead space (Tabachnik et al 1981). Furthermore, it is rather impractical to maintain a face mask and pneumotachograph closely sealed on an infant for long periods. Finally, infants change their sleep state as well as their pattern of breathing rapidly, and this may render impractical some of the previously

It is recognized that the calibration and validation procedures presented here were tested during quiet, REM and non-REM sleep states but not in awake infants or in other body postures as has been done in previous studies (Duffty et al 1981, Adams et al 1993b, Revow et al 1987). It is further noted that the present system has not been assessed in preterm infants or those with significant lung disease in whom asynchrony is expected to be much more pronounced. Based on the results and on the above discussion, we feel confident that the present calibration scheme, which relies on cross-correlation of flow signals, is insensitive to changes in end-expiratory levels or in patterns of breathing normally observed in infants and newborns as they move rapidly from one sleep state to another. Following changes in body posture, changes in rib cage to abdominal proportions are substantial, and it remains to be determined whether the present calibration scheme is stable under wider changes.

In conclusion, we present an automated, computer-based system that is capable of performing a single-posture real-time RIP calibration. The calibration procedure is based on an algorithm that determines calibration coefficients for RIP signals by maximizing the correlation between their summed derivatives and airflow at the mouth. These are calculated interactively in real time at every sample point and the process continues until stable values are reached and convergence criteria met. A GUI was developed to assist in the rapid implementation and ease of use of the above model. The calibration scheme is safe and simple to use, and it is quick, accurate and stable. The system was found to be suitable for use in newborns during natural breathing while asleep.

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Appendix

Equation (3) in section 2 has the linear form

\[ z = x + Ky. \]  \hspace{1cm} (A.1)

The correlation coefficient \( r(f, z) \) for \( f \) (airflow at the mouth) against \( z \) (RIP derived flow) is commonly defined as

\[ r(f, z) = \frac{\text{cov}(f, z)}{S(f)S(z)} \]  \hspace{1cm} (A.2)

where \( \text{cov} \) is the covariance and \( S \) is the standard deviation (=\( \sigma^2 \); variance).

Using equation (A.1), \( \text{cov}(f, z) = \text{cov}(f, x + Ky) \):

\[ r(f, z) = \frac{\text{cov}(f, x + Ky)}{S(f)S(x + Ky)}. \]  \hspace{1cm} (A.3)

Let us denote \( r(f, x) \) as correlation coefficients for \( f \) against \( \dot{R}_{rc} \), \( r(f, y) \) as correlation coefficients for \( f \) against \( \dot{R}_{ab} \), \( f(x, y) \) as correlation coefficients for \( \dot{R}_{rc} \) against \( \dot{R}_{ab} \) and \( S_x, S_y \) as standard deviations of \( \dot{R}_{ab} \) and \( \dot{R}_{rc} \), respectively.

Since

\[ \text{cov}(f, z) = \text{cov}(f, x + Ky) = \text{cov}(f, x) + K \text{cov}(f, y) \]
then from equation (A.2) we get
\[ \text{cov}(f, z) = \text{cov}(f, x) + K \text{cov}(f, y) = S_y S_x r(f, x) + K S_y S_x r(f, y). \] (A.4)

Similarly
\[ S_x^2 (x + K y) = S_x^2 + K^2 S_y^2 + 2K \text{cov}(x, y) \]
\[ = S_x^2 + K^2 S_y^2 + 2K S_y S_x r(x, y). \] (A.5)

Inserting these relationships into equation (A.3) and rearranging the order, we get
\[ r(f, z) = r(f, x) + K S_x r(f, y) \frac{r(f, x) + K S_x r(f, y)}{\sqrt{1 + K^2 S_y^2 + 2K S_x S_y r(x, y)}}. \] (A.6)

where \( S = S_y / S_x \).

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