An Automated Bedside Method for Measuring Functional Residual Capacity by N_2 Washout in Mechanically Ventilated Children

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ABSTRACT. Beside measurement of functional residual capacity (FRC) in ventilated children is impractical. Using a simple technique based on open circuit N_2 washout, we measured FRC in ventilated children. The system was evaluated in the laboratory and in patients. Using a mechanical lung, the reproducibility of 200 studies over a range of 100–500 mL at each of four different flow rates (10 determinations at each level) was very high with a mean coefficient of variation of 2.3% (range 0.5–5.1%). Linearity of the integrated N_2 signal for volumes of 100–500 mL washed out at different flow rates was excellent (range 7.4–17.9 L/min), $r = 0.99$. The mean difference between measured and preset mechanical lung volumes was 2.4% (range 0–4.6%). *In vivo*, reproducibility of six to 10 FRC determinations in each of 30 children gave a mean coefficient of variation of 2.7%. Comparison to the conventional Douglas bag collection method showed a high correlation ($r = 0.97$). We conclude that this is an easy, highly reproducible, and accurate method for FRC determination suitable to ventilated infants and children. (*Pediatr Res* 28: 446–450, 1990)

Abbreviations

ETT, endotracheal tube
FRC, functional residual capacity
[N_2], nitrogen concentration
V, volume
V̇, flow
Ve, minute volume of ventilation
PEEP, positive end expiratory pressure
FiO$_2$, fractional inspired oxygen
FiN$_2$, fractional inspired nitrogen

FRC is an important parameter in the respiratory management of ventilated infants and children. Its determination is essential for rational monitoring of mechanical ventilation in infants and children (1). However, its use has been limited because the existing methods for FRC measurements are cumbersome and thus impractical as bedside procedures in this group of patients. Those techniques that use body plethysmography and measure thoracic gas volume cannot be applied in the sick ventilated child, whereas the helium rebreathing methods add compliance to the ventilatory-respiratory system and may interfere with the

ventilatory pressures delivered to the patient (2). Several N_2 washout methods have been developed. The breath-by-breath washout techniques (some of which are also rebreathing methods) require instantaneous and simultaneous measurement of N_2 concentration and air flow for integration and the use of a pneumotachograph (3–5).

Gerhardt *et al.* (6, 7) recently described an open circuit N_2 washout technique for FRC measurements suitable for use in spontaneously breathing, nonventilated small children. The technique is based on continuous analysis of the N_2 concentration washed out from the lungs of a patient who breathes through a system with constant gas flow. However, two requirements of this method are impossible in the mechanically ventilated patient: 1) constant flow of gas during the test, and 2) gas flow during the test that equals that during calibration.

We describe a new automated technique for FRC determination in ventilated infants and children, based on the open circuit N_2 washout principle, suitable for routine bedside FRC monitoring in pediatric intensive care units.

MATERIALS AND METHODS

Theory. The classic method of FRC determination by the N_2 washout is based on the formulas:

$$FRC = \frac{V_{N_2}}{\text{Initial lung } [N_2]} \qquad (1)$$

$$V_{N_2} = \int \dot{V}[N_2] \, dt \qquad (2)$$

In the bag collection technique, the volume of the washed out N_2 (V_{N_2}) is determined from the final mixed N_2 concentration in the bag and the bag volume.

The open circuit N_2 washout technique previously described (6) assumes that the average flow is held constant over the washout period. This is based on the fact that inspiratory and expiratory volumes are equal so that the average washout flow leaving the patient over time remains the same. Therefore, the integrated N_2 concentration can be multiplied by a constant flow value (6):

$$V_{N_2} = \dot{V} \int [N_2] \, dt \qquad (3)$$

This equation holds for both test and calibration (cal) procedures:

$$V_{N_2\,cal} = \dot{V}_{cal} \int [N_2]_{cal} \, dt$$

$$V_{N_2\,test} = \dot{V}_{test} \int [N_2]_{test} \, dt$$

When $\dot{V}_{cal} = \dot{V}_{test}$ as previously reported (6):

$$V_{N_2\,test} = \frac{V_{N_2\,cal}}{\int [N_2]_{cal} \, dt} \int [N_2]_{test} \, dt \qquad (4)$$

and FRC is determined using equation 1. In the ventilated patient, the condition that $\dot{V}_{cal} = \dot{V}_{test}$ is obviously not met. The mechanically ventilated patient may also initiate spontaneous breaths between the ventilator breaths, for instance, when on intermittent mandatory ventilation with different tidal volumes than the ventilator breaths. Moreover, the volume of a ventilator breath does not necessarily equal that of the calibration syringe (the lung compliance differs from the syringe compliance), so $\dot{V}_{cal}$ cannot be regarded as equal to $\dot{V}_{test}$, and based on equation 3, the appropriate equation to be used is:

$$V_{N_2\,test} = \frac{\dot{V}_{test}}{\dot{V}_{cal}} \times \frac{V_{N_2\,cal}}{\int [N_2]_{cal}\,dt} \int [N_2]_{test}\,dt \qquad (5)$$

The ratio $\dot{V}_{test}/\dot{V}_{cal}$ can be derived from the ratio of the $\dot{V}e$:

$$\frac{\dot{V}_{test}}{\dot{V}_{cal}} = \frac{\dot{V}e_{test}}{\dot{V}e_{cal}}$$

and the final equation is:

$$V_{N_2\,test} = \frac{\dot{V}e_{test}}{\dot{V}e_{cal}} \times \frac{V_{N_2\,cal}}{\int [N_2]_{cal}\,dt} \int [N_2]_{test}\,dt \qquad (6)$$

The ratio $V_{N_2\,cal}/\int [N_2]_{cal}\,dt$ is defined during the calibration procedure. If the ratio $\dot{V}e_{test}/\dot{V}e_{cal}$ is also determined, then FRC can be directly calculated from the integrated N_2 concentration.

Method of operation. The experimental system consisted of a second ventilator that delivered 100% O_2 (washout ventilator) and had the same settings as the patient's ventilator (Fig. 1). Both ventilators were connected to the proximal end of the ETT through a slider valve, which was activated at FRC, when the patient was switched to the washout ventilator. The gas leaving that ventilator via the exhalation port was routed through a mixing chamber and the N_2 concentration was then analyzed continuously by a respiratory mass spectrometer (Airspec MGA 2200, Kent, UK) that also measured the $\dot{V}e$ by the argon dilution technique (8). The mixing chamber (Airspec) was originally commercially designed for measurements of O_2 consumption and for flow measurement by the argon dilution technique and, therefore, met the criteria for effective mixing. The chamber size was 600 mL and it contained three serial channels that included baffles and vanes that recirculated air from the later channels back to a previous channel before exiting past the sampling port. This greatly enhanced mixing and provided a rapid equilibration within the chamber. A personal computer (IBM PC)-based signal processing unit (Sensormedics 2600 Pediatric Pulmonary Unit, Anaheim, CA) in line with the mass spectrometer integrated the N_2 concentration signal electronically and provided a real time
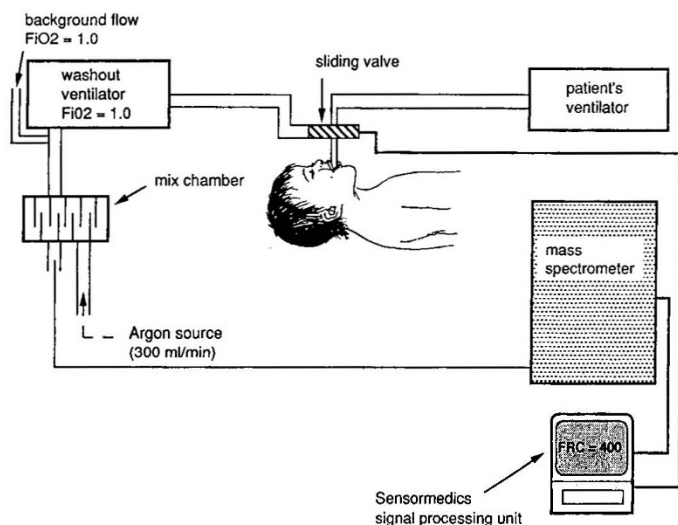

Fig. 1. Schematic diagram of the N_2 washout system for FRC determination in ventilated patients.

display of the N_2 concentration (N_2 washout curve) and the integrated N_2 concentration units on the monitor. The Sensormedics unit also controlled the slider valve and automatically calculated the FRC at the end of the washout period according to the $\dot{V}e$ and the patient's alveolar N_2 concentration measured before the study. The criteria for completion of a washout were that the N_2 concentration on the mass spectrometer readout was stable at zero and that the N_2 washout curve was back to baseline (zero).

The computer system contained a distributed processor design, using a 16-bit computer (IBM) for the graphic presentation and calibration of FRC from the integrated N_2 and volume of calibration. The outboard analog interface was a microprocessor (8085) controlled system with 4K RAM and a 14-bit analog to digital converter. The analog to digital system was capable of spanning ranges of ± 20 V AC. However, its use of a 14-bit digital offset digital/analog converter and a 14-bit span digital/analog converter enabled the system to provide maximum resolution (14 bits) over the limited range of the N_2 signal. The voltage range of the N_2 signal was 0–10 V DC, which is the output of a 12-bit digital/analog converter.

N_2 integration was performed at 250 samples per s, well in excess of the rate of change of the mixed N_2 signal so that no data was lost.

The accuracy of the argon dilution technique was within $\pm 5\%$ when compared with Douglas bag collection (8). The accuracy of the respiratory mass spectrometer was: resolution, up to unity; stability, better than 1% relative over 24 h; linearity, better than 1% over calibrated range; detection, down to 10 ppm. It should be noted that in this technique, the absolute values of each flow (flow during washout and during calibration) are not very important, because the absolute values are not used for FRC determination and the correction factor is based on their ratio only. Thus, even if the flow measurement was off by a certain percent, this cancelled itself as long as the flow measurement was constant. This means that the factors that determined the accuracy were the variability and the linearity of the mass spectrometer; both (better than 1%) were very satisfactory.

Two types of ventilators were tested: *1*) a continuous flow (pressure mode) ventilator (IV-100-B, Sechrist, Anaheim, CA) and *2*) an intermittent flow ventilator (servo-controlled Siemens Elma, Servo-900C, Solna, Sweden) in pressure and volume modes. Whenever the continuous flow on the Sechrist ventilator was relatively high with respect to the absolute amount of N_2 washed out (resulting from low patient $\dot{V}e$, high patient FiO_2, or very low FRC), the N_2 concentration signal was very small because a small amount of N_2 was being mixed with a large amount of washout gas. In that situation, a splitter isolation valve (model no. 8890, Boehringer, Wynnewood, PA) was placed at the ETT end, which directed only the gas exhaled from the patient to the mixing chamber. The resistance of the valve was less than 1 cm H_2O at 10 L/min and the dead space was 3 mL. When the "splitter valve" was used, the ventilator PEEP was set to zero and PEEP was controlled by a PEEP valve attached to the outlet of the splitter valve. Before connecting the PEEP valves, we tested them against a manometer for the range of 0 to 20 cm H_2O. In all situations where the gas reaching the mixing chamber was intermittent rather than continuous, a background flow of 100% oxygen (2–4 L/min) was added to the exhalation port to continuously drive the washed out gas to the mixing chamber. In these cases, the $\dot{V}e$ measured at the mixing chamber was composed of the patient's $\dot{V}e$ and the background flow. This was defined as $\dot{V}e_{(mix)}$, and equation (6) was modified to:

$$V_{N_2\,test} = \frac{\dot{V}e_{test(mix)}}{\dot{V}e_{cal(mix)}} \times \frac{V_{N_2\,cal}}{\int [N_2]_{cal}\,dt} \int [N_2]_{test}\,dt \qquad (7)$$

Calibration was done by a syringe filled with room air attached to the slider valve, in place of the ETT. Two-point calibration was used, at volumes below and above the patient's estimated

FRC. The piston was pushed back by the washout ventilator and advanced by the examiner; the gas from the syringe was analyzed in the same way as above. The ratio of the preset amount of N_2 to the integrated N_2 signal provided the calibration factor.

Evaluation. An error analysis was performed to evaluate the potential errors of the technique and their effects on the FRC measurements. The accuracy of the system was evaluated both in the laboratory (mechanical lung) and on patients.

Mechanical Lung Model. Linearity of integrated N_2 signal. Five preset mechanical lung volumes of 100, 200, 300, 400, and 500 mL were washed out 10 times each at four different total minute ventilation rates (ventilator-patient system flow + background flow) of 7, 11, 14, and 18 L/min (a total of 200 tests). The correlation between the preset volume and the integrated N_2 signal for each of the flow rates was evaluated using linear regression analysis.

Reproducibility. Reproducibility of the system was evaluated by calculating the coefficient of variation of each of the above preset volumes at each of the flow rates ($n = 10$) and for every volume for all four flows ($n = 80$). In addition, a preset volume of 300 mL was measured 72 times at 12 different flow rates using the three experimental setups: Servo ventilator, Sechrist ventilator, and Sechrist ventilator with the Boehringer valve. The coefficient of variation for all these tests was determined.

Accuracy. The difference between the measured volume and the preset volumes was determined. Percent error was defined as: $100 \times$ absolute value of (preset volume − measured volume)/ preset volume.

In Vivo Studies. Reproducibility. Six to 10 determinations of FRC were done in each of 30 ventilated children who had cuffed ETT, and the coefficient of variation was calculated.

Comparison with Douglas bag collection method. During 20 washout procedures, the gas passing the mixing chamber was also collected in a Douglas bag. The mixed N_2 concentration of the bag was measured and the bag volume was determined by evacuating the bag with a metered 3-L syringe. Bag FRC was then calculated by the conventional method (bag volume $\times$ bag N_2 concentration/alveolar N_2 concentration), and the correlation between the two techniques was determined.

Whenever the respiratory rate was high to the degree that the exact point of end exhalation was difficult to determine, we used the expiratory hold option (on the Siemens Servo-900C ventilator) or switched the ventilator mode to continuous positive airway pressure (on the Sechrist) and switched the slider valve only after the clinical determination that the patient was at FRC.

Our study was approved by the Committee of Clinical Research of our institution and informed consent was obtained on behalf of the patients in this study. After completing this study, we conducted a second study based on the technique described here. The patient population for the second study was not the same as the patient population reported here.

RESULTS

Mechanical lung. The linearity of the integrated N_2 signals *versus* mechanical lung volumes containing room air was very high at each of the four different flow rates (Fig. 2). The coefficient of correlation for each of the flow rates ranged between 0.99 and 1.00. The regression lines intersected the x-axis at about 45 mL.

The reproducibility of FRC measurements at five preset volumes and different flow rates was very high when washout was repeated 10 times at each flow rate (coefficient of variation 0.5–5.1%, mean 2.3%) (Table 1). The reproducibility remained very good when the results of each preset volume at the four different flows were combined while the calibration values remained unchanged (Table 1). The coefficient of variation of the 300-mL washouts, using the three different techniques (24 tests with each) was less than 2.0% (Table 2).

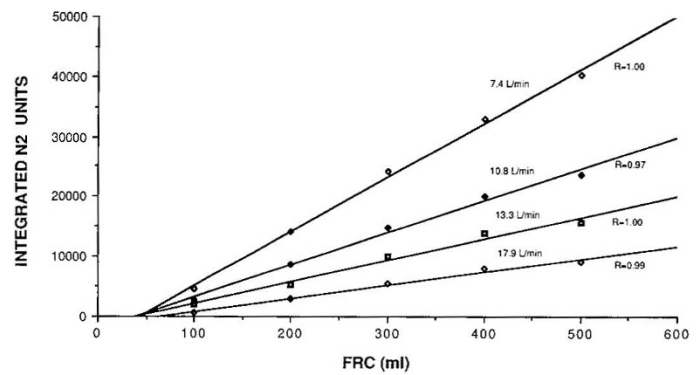The accuracy of the system was excellent at all $\dot{V}e$ and all

Fig. 2. Linearity of integrated N_2 concentration at different flow rates and different preset volumes.

Table 1. *Reproducibility of 10 washouts at five washout volumes at four flow rates**

Flow rate (L/min)	Mechanical lung vol (mL)				
	100	200	300	400	500
7.4					
Mean	97	199	305	400	504
CV	2.4	2.4	0.8	0.8	1.5
SD	2.3	4.9	2.5	3.1	7.4
10.8					
Mean	100	209	311	393	498
CV	1.6	3.5	1.3	1.9	0.5
SD	1.6	7.2	3.9	7.3	2.6
13.3					
Mean	95	194	306	398	498
CV	3.6	3.1	1.7	1.1	1.7
SD	3.4	6.1	5.3	4.5	8.3
17.9					
Mean	102	202	304	402	502
CV	5.1	4.1	4.8	2.8	1.6
SD	5.2	8.2	14	11	8.1
Mean CV	3.2	3.3	2.1	1.6	1.3

* CV, coefficient of variation.

Table 2. *Washout results of preset volume of 300 mL measured by the three different ventilator techniques*

Technique	Measured volume (mL)	Mean	CV*
Siemens volume control	297, 295, 300, 295, 291, 301, 291, 298, 302, 301, 301, 302, 303, 297, 294, 297, 306, 297, 296, 295, 304, 305, 294, 297	298.3	1.4
Sechrist pressure control	292, 295, 310, 311, 291, 301, 299, 299, 303, 297, 303, 292, 305, 293, 291, 297, 300, 300, 297, 297, 302, 290, 296, 295	298.2	1.1
Sechrist, Boehringer valve	311, 307, 300, 296, 290, 305, 309, 305, 296, 311, 303, 310, 298, 299, 300, 305, 301, 290, 299, 302, 298, 299, 300, 309	301.8	2.0

* Coefficient of variation.

volumes tested. In the 200 bench tests, the mean of the absolute percent error was 2.3% (range 0–11%). The maximal overestimation of volume by the system was 10% and maximal underestimation was 11%.

Studies on patients. The mean age of the 30 ventilated children studied was 2.5 (range 3 wk–7 y). The mean body weight was 12.5 kg (range 3.8–28.0 kg). The FiO_2 concentration was 0.4–0.65.

FRC results were reproducible with a coefficient of variation of less than 6.5% (0.7–6.5%) and the mean coefficient of variation per patient being 2.7%.

FRC results by N_2 integration were close to those obtained by the bag collection method. Paired *t* test between the two techniques gave $p = 0.75$ and a highly significant correlation coefficient of $r = 0.97$ (Fig. 3).

The average washout time for FRC measurement was 50 s (range 30–80 s).

Error analysis of the system. The FRC is calculated from

$$FRC = \frac{V_{N_2}}{FiN_2 (t_0) - FiN_2 (t_1)}$$

where $FiN_2 (t_1)$ is practically zero but for the purpose of the error analysis, we take it as above zero.

Errors in V_{N_2}. Obviously, there is a linear function of the error in V_{N_2}, *i.e.* a 1% error in V_{N_2} translates (in the absence of other errors) to a 1% error in FRC (see below).

Errors in FiN_2. Offsets, *e.g.* when the mass spectrometer reads too high by one percentage point, do not count, as the errors cancel out by the subtraction.

When errors in $FiN_2 (t_0)$ and $FiN_2 (t_1)$ are in the opposite direction, the analysis refers to the worst case analysis: the direction of the error depends on which is high and which is low. We calculated the error for washout from 78% N_2 to 2% with a positive error on the $FiN_2 (t_0)$ and a negative error on the $FiN_2 (t_1)$, which gives underestimates for FRC. Reversing the signs just results in the same numbers with opposite signs. The error was less than 3.5%. If the washout starts at only 40%, the problem is worse, but the error is still less than 4.8%. Thus, assuming a 1% error in N_2 concentration, the worst case error in FRC value is under 5%.

Errors in both V_{N_2} and FiN_2 (errors in opposite sense as in worst case analysis above). The worst case is a positive error in V_{N_2}, a negative error in $FiN_2 (t_0)$, and a positive error in $FiN_2 (t_1)$. Again, when the washout is over a small range (*e.g.* 40 to 2%) things get worse. In this case, the combined errors of 1% in V_{N_2} and 1% in FiN_2 give up to a ±7% error in FRC. Realistically, the value $FiN_2 (t_0) - FiN_2 (t_1)$ was probably measured within 0.2%, so the errors translate to less than 2% in FRC even with a 40% N_2 concentration starting point.

Errors in $V_{N_2\,test}$. From equation 6, what is critical is how well the $\dot{V}e$ can be measured by the argon dilution method. Thus, for this part, we assumed errors in $V_{N_2\,cal}$, $\int[N_2]_{cal}$, and $\int[N_2]_{test}$ to be up to 1% each.

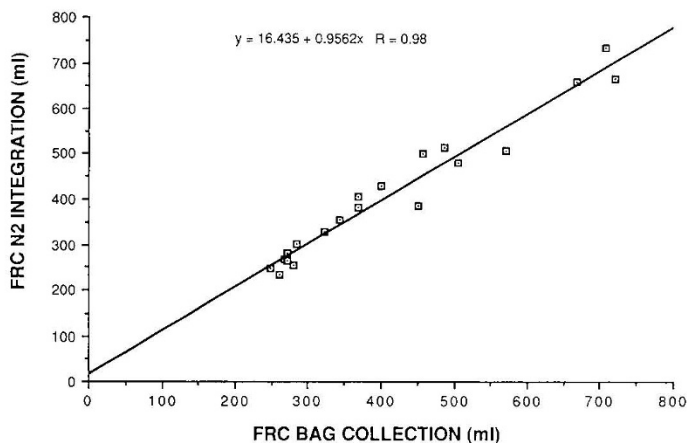*Errors in $\dot{V}e$.* From Davies and Denison (8), the final equation

Fig. 3. Correlation between FRC measurements in 20 patients by both bag collection and N_2 integration techniques.

for $\dot{V}e$ using argon dilution and taking into consideration the contribution of the respired tracer (argon is already present in inspired air by 0.9%) is of the form:

$$\dot{V}e = \frac{\dot{M}_{tr}}{X} - \dot{M}_{tr}$$

where

$$X = Fm_{tr} - FE_{tr} \frac{1 - Fm_{tr}}{1 - FE_{tr}}$$

and where $\dot{M}_{tr}$ is the quantity of tracer gas (argon) per unit time, Fm_{tr} is the fractional concentration of tracer gas in the gas mixture (m), FE_{tr} is the concentration of tracer in the expirate upstream of the injection site, and X is the fractional concentration of tracer in the gas mixture taking into consideration the contribution of the respired tracer.

The argon flow is the same during all studies (0.3 L/min). Worst case analysis: Ventilator/patient + background flow = 18 L/min (highest value in our study). Thus, by volume dilution, the mixed argon concentration is 1.639% (100·0.3/18.3) above the argon concentration in the inspired gas (*i.e.* 1.639 + 0.90 = 2.539%). We assume that the mass spectrometer can resolve the argon with an accuracy of ±0.02% at a given level; thus, $Fm_{tr} = 2.539 ± 0.02%$ and $FE_{tr} = 0.90 ± 0.02%$.

The term $(1 - Fm_{tr})/(1 - FE_{tr})$ is 0.98, so

$$\dot{V}e = \frac{\dot{M}_{tr}}{Fm_{tr} - FE_{tr} (0.98)} - \dot{M}_{tr}$$

Assuming a 1 mL/min error in argon flow, $\dot{V}e$ (mL/min) = 300 ± 1.

$$\dot{V}e = \frac{300 ± 1}{\frac{1.639}{100} ± \frac{0.04}{100}} - 300 ± 1$$

$$= \frac{300 ± 0.3\%}{\frac{1.639}{100} ± 2.4\% \times 0.98} - 300 ± 1$$

$$= 18\ 300 ± 2.7\% - 300 ± 1$$

$$= 18\ 000 ± 2.7\% \quad (±1\ \text{ignored}).$$

So, worst case, the error in $\dot{V}e_{test}$ is about 2.7%. The error calculations at 14, 11, and 7 L/min are 2.2, 1.8, and 1.2%, respectively.

Calculating accuracy of entire system, i.e. root mean squared sum of accuracies of five measured values in equation 6. Assuming similar errors in $\dot{V}e_{test}$ and $\dot{V}e_{cal}$ (although $\dot{V}e_{cal}$ would be normally better determined because of repeated measurements)

$$V_{N_2\,test} = \frac{\dot{V}e_{test} (2.7\%)}{\dot{V}e_{cal} (2.7\%)} \times \frac{V_{N_2\,cal} (1\%)}{\int[N_2]_{cal}\,dt\,(1\%)} \int [N_2]_{test}\,dt\,(1\%)$$

Root mean squared sum = 1.88%, *i.e.* less than 2%.

As mentioned above, a 1% error in V_{N_2} translates into a 1% error in FRC. FiN_2 can be determined at any time to within 0.5% N_2, *i.e.* 78.0 ± 0.5, and our error in FRC determination for a 78 to 2% N_2 washout could be up to 3.3% worst case. For a 40 to 2% N_2 washout, the error could be up to 4.8% worst case. The measurements in the study were indeed made within the limitations of the error analysis.

DISCUSSION

FRC is an important physiologic index in patients with respiratory diseases, especially those with respiratory failure. It is the point of equilibrium between the elastic recoils of the lungs and the chest, serves as an O_2 reservoir during the breathing cycle,

determines the resting length of the respiratory muscles, and may provide the clinician with information regarding overinflation or volume loss. Although measurements of FRC are easily done in the pulmonary function laboratory, they are very difficult to obtain in mechanically ventilated patients, and at the moment, measurements are confined to research studies (9). This statement is especially true in small children. Nevertheless, it is in this group of ventilated patients where FRC measurements are greatly needed because these patients are usually in extreme respiratory system decompensation and their FRC are highly dependent on the various manipulations of positive pressure during assisted ventilation (especially the PEEP).

Our study describes an automated bedside method for FRC determination in ventilated children. Its great advantage is that it does not require a pneumotachograph and pressure transducer and thus avoids the problem of cumbersome calibration procedures used in the past (3). The technique assumes that the flow in the system is constant during the washout period. Although the instantaneous flow rate of the washout circuit changes continuously, the average flow over time remains constant. The volume that is subtracted during inspiration is added back to the system during expiration as long as the inhaled and exhaled gases have similar humidity and temperature, which are obtained by the use of a heated humidifier in the ventilator circuit. Thus, over the entire test period (usually 30–80 s) the volume per time (average flow) remains constant. Results of FRC measurements based on this argument have been shown to be very reproducible in nonventilated, spontaneously breathing small children (6, 7). The situation in the ventilated patient is, however, different because of intervals of no flow in the exhalation washout circuit that occur during inspiration. Therefore, a background flow was used after the exhalation port, resulting in very good mixing as reflected by the high reproducibility and accuracy.

A major problem in applying the technique to ventilated infants and children is that the flow rate during calibration differs from that during the test because the compliance of the calibration apparatus differs from that of the patient. We solved this problem by measuring the $\dot{V}e$ during both the calibration and the test, and correcting the results with the $\dot{V}e$ of the patient. The very high linearity of the N_2 signal and the excellent accuracy of the mechanical lung tests when five preset volumes were measured at four different flow rates after calibration was done at a different $\dot{V}e$ provide an experimental verification for the applicability of this correction procedure. The reason that the regression lines do not intersect the xy axis at 0,0 is that the integrator has a preset digital offset for reasons that are associated with other tests that the system is capable of performing. The offset is, therefore, the same during calibration and test and cancels itself in the equation.

Another technical problem in applying the technique to very small ventilated children is the small amount of N_2 in the lungs, especially in the presence of lung disease with a significant volume loss (*e.g.* pulmonary edema) and a relatively high FiO_2. This may result in a very small N_2 washout curve affecting the resolution and accuracy of the system. This is particularly true when a small amount of N_2 (sometimes as low as 40 mL) is mixed in a high (ventilator circuit) flow (15–24 L/min). In our study, the use of a three-way splitter valve in those situations was found to provide very reproducible and accurate results. By directing only the gas exhaled by the patient to the mixing chamber without the large baseline flow existing in the system, this valve significantly reduces the amount of N_2 free gas in which lung gas is diluted. By using a two-point calibration technique rather than a one-point (with the other being zero), the reproducibility remained very high even in this subgroup as long as the measured FRC value was between the two calibration points.

The main advantages of the technique are that studies can be performed within minutes and can be repeated easily. The calibration procedure is simple and does not have to be repeated between studies or adjusted to the patient's settings. No change is required in the patient's ventilatory settings except for the temporary (30–60 s) change in the FiO_2 during the test and the procedure does not cause any discomfort to the patient. Thus, the system is suitable for use in very sick and unstable children over a wide range of oxygen concentrations.

This technique has some disadvantages. Specific equipment is needed. It requires that there be no leak from the system, and thus cannot be used in children with an air leak around the ETT, or those with pneumothoraces. It cannot be used, of course, in patients who are already on a very high oxygen concentration (close to 100%).

The accuracy of the N_2 integration technique was very high both *in vitro* (mean error 2.3%) and in the 20 patients studied (correlation coefficient 0.97 to bag collection technique). It was reassuring to note that the reproducibility in patients was very close to that obtained in the mechanical lung, because higher reproducibility can be expected in laboratory tests than in a biologic system. One explanation might be that the volume in the mechanical lung could be preset with an accuracy of ± 3 mL, which may cause a technical error of up to 6% (± 3 mL out of 100). This also explains the relatively higher coefficients of variations at lower mechanical lung volumes (Table 1). On the other hand, it emphasizes the accuracy of the technique in studies on patients.

In conclusion, this open circuit, automated N_2 washout technique is a very accurate and easily reproducible method for FRC determination in ventilated patients. Its accuracy holds true also in extreme patient conditions—in small infants ventilated at relatively high oxygen concentrations. Its main advantage is that it is a technically simple bedside procedure that does not interfere with patient care and can provide important data within a short time in the pediatric intensive care environment. This method may prove useful in studies designed to evaluate the clinical significance of FRC measurements or the effects of ventilation parameters (*e.g.* PEEP), and in clinical decision making regarding everyday care.

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