A METHOD FOR MEASURING TIDAL VOLUME DURING HIGH FREQUENCY JET VENTILATION

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The tidal volume delivered to the lung during high frequency jet ventilation depends on the gas volume emerging from the jet and the volume of gas entrained or spilled into the bias flow [1, 2]. In previous studies, tidal volumes have been measured either by using a whole body plethysmograph [3] or by external spirometry using length transducers around the thorax and abdomen [4]. Although the latter technique is appropriate for routine clinical work, it has to be calibrated for every patient, and recalibrated each time the patient is moved.

We have devised a new method for measuring true tidal volume during jet ventilation, based on measurement of flow in the bias flow tubing. The accuracy of the technique has been evaluated in the laboratory by comparing the measurements obtained by the new method with tidal volumes calculated from pressure changes in a model lung.

METHODS

The model lung

The model lung comprised a copper cylinder of approximately 60-litre volume packed loosely with metal wool to ensure isothermal expansion and contraction of gases. It had a measured static compliance of 55 ml/cm H₂O, corresponding approximately to human respiratory static compliance. The pressure within the cylinder was measured with a Gould pressure transducer type T150 AD and strain gauge amplifier, the signals being recorded on a Bryans "Transcribe" chart recorder. Rapid manual injection of up to 300 ml of air into the model lung produced less than 5% pressure overshoot, indicating that gas compression was virtually isothermal.

The gas system (fig. 1)

Jet pulses were generated by interrupting a 413-kPa oxygen source with a Sperry "Lucifer" solenoid valve. The gas pulses then passed through a needle valve and along a 5-cm length of 3-mm i.d. brass tubing to the jet lumen of a National Catheter Corporation "Hi-Lo Jet" tracheal tube inserted into the model lung. A solid state high pressure transducer (National Semiconductor LX1610) monitored the driving pressure via a side port from the brass tubing. Control of the driving pressure was achieved by adjusting the needle valve. The small volume and thus low compliance of the jet delivery circuit after the solenoid valve provided a nearly square driving pressure–time waveform. The bias flow tubing was a 62-cm length of 2.5-cm i.d. plastic tubing. At one end, a flow of oxygen was introduced via a needle valve while, at the other, a Mercury 300L...
Pneumotachograph head was attached and connected to a Validyne MP45 differential pressure transducer. A side port was fitted to the bias flow tube 12 cm from the pneumotachograph head and this connected to the main lumen of the tracheal tube.

Data collection and calibration

Data were collected from the flow and driving pressure transducers using a BBC Master microprocessor fitted with a locally made fast 12-bit analogue-to-digital converter system based on an HD9410 data acquisition module from Hybrid Systems. The microprocessor generated also the pulses to control the solenoid valve via a simple interface circuit. The control and data manipulation software was written in the department.

The pneumotachograph was calibrated by passing a flow of dry oxygen through it and then through a Parkinson–Cowan dry gas meter. The time taken for one revolution of the dial (25 litre) was timed by the microprocessor using an opto–electrical sensor and the flow calculated from this. The model lung pressure transducer was calibrated directly in terms of volume by introducing 50-ml aliquots of gas into the closed model lung and plotting a calibration graph. The time base for valve timing, sample timing for the analogue-to-digital converters and the integrations was the microprocessor internal clock, which was assumed to be accurate.

Experimental procedure

The model lung was ventilated at 50, 100, 150, 200 and 250 b.p.m., at an I:E ratio of 0.43 and at six to eight different positions of the needle valve at each rate. This produced driving pressures in the range used clinically. The microprocessor logged flow and driving pressure signals, sampling at 512 times per cycle. The pressure signal from the model lung was recorded simultaneously on the chart recorder, to provide a measure of tidal volume independent of the microprocessor. Both bias flow (25 litre min⁻¹) and jet gas were 100% oxygen. The system was recalibrated after each change of ventilatory rate.

Data manipulation

A flow–time tracing constructed from data logged by the microprocessor is shown in figure 2. Area A corresponds to the integral of the difference between the bias flow and the flow recorded by the pneumotachograph up to the point where these values coincide (point x). This area is a measure of the volume of gas entrained by the jet (V_vent). Area B corresponds to the integral of the increase in the measured flow above the bias flow from the point where entrainment ceases (point x) to the point where expiration begins (point y). This is the spilled volume (V_spill). The inflection point, point y, proved difficult to detect using the microprocessor; however, a decrease in the driving pressure signal to 67% of the
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were calculated from the driving pressure waveform and the timing of the valve control signals. The driving pressure signal was averaged over the "on" period to calculate the true mean driving pressure. The true I:E ratio also was calculated from the driving pressure signal.

If the delivered and exhaled tidal volumes are assumed to be the same, the volume of gas delivered by the jet can be calculated from the exhaled, entrained and spilled volumes:

\[ V_t = V_{ent} + V_{del} \]  

(1)

where \( V_t \) = exhaled tidal volume; \( V_{ent} \) = entrained volume; \( V_{del} \) = the unspilled portion of the gas delivered by the jet to the lungs.

Rearranging (1):

\[ V_{del} = V_t - V_{ent} \]  

(2)

The volume delivered by the jet is the sum of the delivered and spilled volumes:

\[ V_{jet} = V_{del} + V_{spill} \]  

(3)

Combining (2) and (3):

\[ V_{jet} = V_t + V_{spill} - V_{ent} \]

**RESULTS**

A graph of the tidal volumes calculated by our new method against tidal volume deduced from pressure changes in the model lung is shown in figure 3. The points all lie close to the line of equality, the correlation coefficient is 0.99 and the linear regression equation is \( y = 1.02x - 2.1 \).

As suggested by Bland and Altman [5], we have...
also plotted the difference of each pair of values against the mean of each pair. This is shown in figure 4. This plot accentuates what the regression equation revealed: that, at low tidal volumes, the values measured by pressure changes are slightly higher than tidal volumes calculated from flow changes. The mean difference is +0.6 ml and the standard deviation is 2.61 ml. This implies that the "limits of agreement" (mean difference and 2 standard deviations), are +5.82 to -4.62 ml. The 95% confidence limits for these estimates using a t test are ±0.9 ml. Thus there is a 95% chance that the "limits of agreement" lie within +5.82 ± 0.9 ml and -4.62 ± 0.9 ml.

DISCUSSION

The results indicate that, compared with tidal volumes deduced from pressure changes within a model lung, there is moderate agreement at low volumes, but an acceptable agreement at higher (and more clinically relevant) tidal volumes. The pressure change method for measuring tidal volume in the model lung was chosen as a standard for comparison as a simple method for determining tidal volumes at high rates of ventilation, totally independent of the measuring system under examination.

We believe our method overcomes many of the problems associated with measuring tidal volumes during high frequency jet ventilation. First, the method works in the presence of any bias flow, and even with no bias flow at all if the risk of entrainment of atmospheric air is acceptable. The only constraint on this is that, at high bias flow rates and low tidal volumes, the changes in measured flow are small compared with the total flow, so errors caused by digitizing and zero drift become more apparent. We believe this is one of the reasons our method differs more from the volumes measured by pressure changes at low tidal volumes. A further small error may occur in clinical use because expired gas has a composition different from that of inspired, and this changes the pneumotachograph calibration slightly. This should be least at high bias flows and low tidal volumes.

Second, the problem of measuring gas loss during inspiration (the spilled volume) is overcome. This problem has been noted [6], but not studied in depth. Our system can measure this volume, and it remains to be seen if this effect is important clinically.

Third, by minimizing the volume of the gas delivery system after the solenoid valve, we have tried to achieve a square wave driving pressure-time profile. The shape of the waveform is important as a determinant of both mean driving pressure and mean gas velocity. The former affects the amount of gas passing down the jet lumen, whilst the latter affects the degree of entrainment [2,7]. The monitoring of driving pressure has other advantages. The delay in opening and closing of solenoid valves may be a significant problem [2], especially at high rates. Using the driving pressure signals these delays can be measured and the valve timing adjusted to achieve the desired true i:E ratio. Also, the point at which

FIG. 4. A graph of the difference between the tidal volume measured by pressure changes in the model lung minus the same volume measured by the new method plotted against the mean of the two values. The dotted lines indicate the "limits of agreement" (2 standard deviations).
inspiration ceases and expiration begins is difficult
to determine, for no flow reversal occurs as in
conventional ventilation [4]. We tried initially to
determine the beginning of expiration by looking
for the inflection point (point y) on the flow–time
tracing, but it became less distinct as the rate of
ventilation increased. However, the inflection
point corresponds to the point where the driving
pressure decreases to about 67% of its peak value.
Because this value changes slightly at different
rates, 67% was chosen as a compromise for all
rates. Errors from this are minimized with a
square wave driving pressure–time profile.

Finally, our system is capable of responding to
very rapid changes. The differential pressure
transducer and pneumotachograph head have a
linear response to 40 Hz, and the driving pressure
transducer is linear to 2000 Hz. Sampling at 512
samples per cycle allows information to be
obtained to a maximum frequency of 256 times
the cycle frequency.

The system as described is not suitable for
clinical use. Safety alarms and cut-outs would be
needed to indicate solenoid failure and to prevent
excessive airway pressure if the expiratory limb
were occluded accidentally.

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