Measurement of Functional Residual Capacity during High-Frequency Oscillatory Ventilation (HFOV) by Argon Washout Method without Interruption of HFOV*

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A modified indicator gas washout method was developed to measure functional residual capacity (FRC) during high-frequency oscillatory ventilation (HFOV) without interruption of HFOV. A hot-wire flowmeter and medical gas analyzer measured the flow rate and argon concentration, respectively, at the expiratory end of the respiratory circuit. Upstream of the hot-wire flowmeter, two heat-and-moisture exchangers for resistance and a rubber balloon for capacitance were placed to convert the oscillating expiratory flow to an almost continuous flow. This made it possible to measure FRC during HFOV without interrupting HFOV. To measure the volume of the entire respiratory circuit, a 10 percent argon in 90 percent oxygen gas mixture was initially used as a bias flow, and after equilibration, the test gas was switched to 100 percent oxygen. By electrical integration of the product of the expiratory flow rate and argon concentration, the total amount of argon equilibrated in the entire respiratory circuit was calculated. The volume of the circuit was calculated by dividing the total amount of argon by the initial argon concentration. Functional residual capacity plus the volume of the respiratory circuit was similarly calculated and the difference was estimated as FRC. The accuracy and reproducibility of our method were evaluated by using a one-compartment lung model. There was a high correlation between the volume setting of the model lung and the estimated FRC. This method can be used to estimate FRC in a one-compartment lung model during HFOV, and it is potentially useful in clinical situations.

![FRC = functional residual capacity; HFOV = high-frequency oscillatory ventilation; C = capacitance; R = resistance; PEEP = positive end-expiratory pressure; Vd = volume of dead space of circuit; HME = heat-and-moisture exchanger](image-url)

High frequency oscillatory ventilation (HFOV) is an accepted respiratory modality in treating respiratory failure, especially in neonates and infants. However, the physiologic effects of HFOV on pulmonary function have not been thoroughly clarified because of the difficulty of measuring physiologic parameters during HFOV. Functional residual capacity (FRC) is one of these parameters. Most reports on FRC during HFOV are related to the relative changes in FRC determined by using impedance plethysmography, body plethysmography, or a closed spirometer system. Absolute FRC can be measured with either a conventional body plethysmography method or an indicator dilution method, but both require interruption of HFOV.

In this study, we developed a new method to measure FRC of a model lung during HFOV without interruption of HFOV with an inert gas washout method.

**MATERIALS AND METHODS**

Figure 1 shows the experimental setup of our method. We selected an indicator gas washout method with argon gas. An HFO ventillator (Hummingbird BMO-20N, Senko Medical Instruments, Tokyo, Japan) creates a sinusoidal pressure change by means of a sealed piston pump driven by a linear magnetic motor. The air-tight respiratory circuit was made of low-compliant plastic tubing (internal diameter, 10 mm). To detect air leaks in the connections, the circuit was immersed in water. At the expiratory end of the circuit, upstream of the sampling points for both flow rate and argon concentration, a compliant air-filled rubber balloon (20 ml/cm H2O) for capacitance (C) and two heat-and-moisture exchangers (HME) (Thermo-vent, Portex) for resistance (R) were interposed, with the C between the two Rs. The combination of C and R constituted a low-pass filter. This modified low-pass filter was connected to the positive end-expiratory pressure (PEEP) device of the HFO ventilator that was used to regulate mean airway pressure. To evaluate the effect of C and R, we compared the flow rate and argon concentration at the sampling point with the addition of C and R to the values obtained without them. The signals were recorded by a thermal recorder (WS-641G, Nihon-Kohden, Japan) via a preamplifier (RMP-8004, Nihon-Kohden, Japan). To estimate the effect of C and R on the ventilatory efficiency of HFOV, we examined the

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When the respiratory circuit, upstream of the low pass filter of the high-frequency oscillatory (HFO) ventilator, a rubber balloon and two heat-and-moisture exchangers (HME) were placed. Downstream of these devices, a hot-wire flowmeter and a medical gas analyzer measured, respectively, the flow rate and argon concentration. The signals were then fed to a microcomputer, and the product of these signals and its integration were calculated to measure the volume of argon washed out. Continuous bias flow was changed by a crossover solenoid valve from a 10 percent argon in 90 percent oxygen mixture to 100 percent oxygen.

A hot-wire flowmeter (RM-300, Minato Medical Science, Osaka, Japan) and a medical gas analyzer (Mass spectrometer, Perkin-Elmer 1100) measured expiratory flow rate and argon concentration every 12 ms downstream of C and R, respectively. These signals were fed to a microcomputer (RM-300, Minato Medical Science, Osaka, Japan) and the product of these signals and its integration were automatically calculated. Initially, a 10 percent argon in 90 percent oxygen gas mixture was used as the inspired bias flow. When the argon concentration at the sampling point of the expiratory terminal had become sufficiently stable, a crossover solenoid valve switched the bias flow to 100 percent oxygen to wash out the argon gas in the entire circuit. Simultaneously, the microcomputer started to integrate the product of the flow rate and argon concentration, which corresponded to the amount of argon gas washed out from the system. Argon concentration at the sampling point became nearly zero within seven minutes. Thus, both argon concentration and flow rate were sampled for seven minutes to calculate the volume of argon washed out. The amount of argon washed out divided by the initial argon concentration was considered the estimated volume.

To evaluate the accuracy of our method, we measured the volume of a syringe with a known capacity. First, the measurement was done with the T-adapter closed. The calculated value represented the volume of dead space of the circuit (Vd) from the crossover solenoid valve to the sampling point. A syringe was then connected to the T-adapter and the volume setting of the syringe was changed from 20 to 40, 60, and 100 ml. The volume of the circuit, including that of the syringe for each volume setting, was measured in the

![Diagram](attachment:image.png)

**Figure 1.** Diagram of our system. An oscillatory pump driven by a linear motor delivers sinusoidal oscillations to a model lung. The volume was changed from 0 to 20, 40, 60, and 100 ml. At the expiratory end of the respiratory circuit, upstream of the low pass filter of the high-frequency oscillatory (HFO) ventilator, a rubber balloon and two heat-and-moisture exchangers (HME) were placed. Downstream of these devices, a hot-wire flowmeter and a medical gas analyzer measured, respectively, the flow rate and argon concentration. The signals were then fed to a microcomputer, and the product of these signals and its integration were calculated to measure the volume of argon washed out. Continuous bias flow was changed by a crossover solenoid valve from a 10 percent argon in 90 percent oxygen mixture to 100 percent oxygen.

![Graph](attachment:graph.png)

**Figure 2.** Panel a displays a recording of pressure swings at the airway opening during high-frequency oscillatory ventilation (HFOV) without the addition of capacitance (C) or resistance (R). Panel b shows the same recording with the addition of C and R.
same manner. This volume minus Vd was determined as the estimated FRC of the syringe. Each experiment was performed in triplicate.

The settings of the HFO ventilator were as follows: mean airway pressure of 3 cm H2O at the airway opening; respiratory frequency of 15 Hz (900 cycle/min); piston stroke volume of 15 ml; and continuous bias flow rate of 3 L/min.

The correlations between the volume settings of the syringe and the estimated FRC were evaluated with a regression analysis. Reproducibility was checked by plotting the difference between the two values (measured volume—setting volume) against the setting values. The coefficient of variation, ie, SD/m was also determined. A p value less than 0.05 was considered statistically significant.

RESULTS

Figure 2 shows the pressure swing at the airway opening with or without C and R. The amplitude and sinusoidal pattern of the pressure swing did not change with the addition of C and R. This indicates that C and R did not affect the ventilatory pattern during the measurement of FRC. Figure 3 shows the flow rate at the sampling point of the expiratory terminal with or without C and R. Figure 4 shows the argon concentration at the same point. Without the addition of C and R, severe oscillations of flow direction and argon concentration were observed. These oscillations disappeared with the application of C and R.

Figure 5 shows the significant correlation between the volume setting of the syringe and the estimated FRC (Y = 0.9996X - 0.288, r = 1.00, p < 0.001). The Vd measured in our system was 95 ± 0 ml (mean ± SD). Figure 6 shows the plotting of the two values (measured volume—volume setting) and volume setting. The median coefficient of variation was 0.41 percent (range, 0 to 0.92 percent).

DISCUSSION

High-frequency oscillatory ventilation delivered by
a piston device with moving small volumes at high frequency (10 to 30 Hz) can maintain adequate alveolar gas exchange. However, the physiologic bases for gas exchange during HFOV have not been thoroughly investigated. Most studies stress the importance of an increase in FRC during HFOV for alveolar gas exchange. Saari et al. showed lung volume increases in humans during HFOV. Simon et al. reported on gas trapping or auto-PEEP in dogs during HFOV. If gas trapping occurs, FRC rather than mean airway pressure is a good monitoring parameter.

Some authors did not measure the absolute value of FRC, but they did measure a relative increase in lung volume by using impedance plethysmography, body plethysmography, or a closed spirometer system. Rouby et al. used a differential linear transformer to measure changes in lung volume in humans during high frequency jet ventilation. Impedance plethysmography, employing two belts around the chest and abdomen, measures their cross-sectional areas by changes in the impedance of the belts. This device, however, needs calibration through delivery of known volumes of gas to the respiratory system of each patient, and thus requires interruption of HFOV with muscle relaxant. A differential linear transformer involves the same problems as impedance plethysmography. Changes in lung volume above FRC can be measured by body plethysmography and closed spirometry, but both require clamping the HFOV line to allow the respiratory system to return to its baseline volume.

To measure FRC directly, pressure body plethysmography and an indicator dilution or washout method are used, but both necessitate interruption of HFOV followed by switching over to manual pumping or conventional mechanical ventilation. These procedures are therefore not practical.

The measurement of FRC during HFOV by conventional tracer gas washout methods is not feasible unless HFOV is stopped, because neither flow rate nor gas concentration in the bidirectional rapid flow can be measured accurately with, respectively, a flowmeter and a gas analyzer. Additionally, it is difficult to correct the differences in the time lags and time constants between the flow and concentration signals. To overcome these problems, we converted the oscillatory flow to an almost continuous flow by using a modified low-pass filter consisting of a very compliant rubber balloon interposed between two HMEs, located upstream of the flowmeter and medical gas analyzer. As a result, the oscillatory flow pattern was eliminated, so that the flow became almost constant, although the pressure swings at the airway opening were maintained. Moreover, their fluctuations at the sampling point were slow enough for the differences in the time lags and time constants between the flow and concentration signals to become negligible. Because flow oscillation at the sampling point can be eliminated with our method, other commercial flowmeters can be used.

A combination of only one HME and a rubber balloon, however, is not adequate to decrease the oscillations in our setup. A greater number of HMEs and rubber balloons would be more effective as a filtering device, but they would increase the volume of dead space. It is desirable that the rubber balloon be very compliant and that the dead volume of the entire respiratory circuit be as small as possible. Our experimental setup does not include a humidifier, although it is possible to perform the measurement with a humidifier. In clinical situations, a bypass circuit around the humidifier may be helpful in decreasing the dead volume during FRC measurement.

The dead volume of the entire respiratory circuit (Vd), which was calculated with the T-adapter closed, was 95 ± 0 ml in triplicate measurements. Because the lung volume of infants or neonates during HFOV ranges from 20 to 100 ml, the Vd of our circuit is not so large compared with the FRC. The Vd might be influenced, however, by mean airway pressure because of the compliance of the rubber balloon. Therefore, although we did not plot a calibration curve of Vd vs mean airway pressure in this experiment, this calibration curve should be determined with varying the mean airway pressure before a patient is placed on HFOV.

High correlation with a small coefficient of variation for each measurement between the measured FRC of the model lung and the volume setting proved the theoretical accuracy of our method. This system requires neither the interruption of HFOV nor changes in the settings of HFOV. This report on in vitro measurement of FRC during HFOV without interruption of HFOV, may be applicable to clinical use.
Conceivably, this method has some problems inherent in inert gas washout methods to determine lung volumes. For example, if ventilation heterogeneity increases, washout takes a longer time and becomes inadequate. This might lead to underestimation of measured FRC.

In summary, we developed a new method to measure FRC during HFOV without interruption of HFOV. The validity of our method was confirmed with the aid of a one-compartment model lung. This method may be used to estimate FRC in clinical situations.

REFERENCES