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WATERSON, CHARLES KENT

GAS DISPERSION AND TRANSPORT WITH HIGH FREQUENCY JET VENTILATION

THE UNIVERSITY OF ARIZONA M.S. 1984

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GAS DISPERSION AND TRANSPORT
WITH HIGH FREQUENCY JET VENTILATION

by

Charles Kent Waterson

A Thesis Submitted to the Faculty of the

DEPARTMENT OF ELECTRICAL AND COMPUTER ENGINEERING

In Partial Fulfillment of the Requirements
For the Degree of

MASTER OF SCIENCE
WITH A MAJOR IN ELECTRICAL ENGINEERING

In the Graduate College

THE UNIVERSITY OF ARIZONA

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This thsis has been approved on the date shown below:

Kenneth C Mylrea   July 23/84

K. C. MYLREA, Ph.D.   Date
Associate Professor
Electrical and Computer Engineering

PREFACE

The research detailed in this thesis was performed at the Arizona Health Sciences Center in the Department of Anesthesiology of the College of Medicine. It was sponsored in part by Healthdyne, Inc., Marietta, Georgia.

The author wishes to acknowledge the assistance given by the Division of Animal Resources and those who gave advice and support in conducting the animal investigations described here.

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ABSTRACT

Three studies have been performed to investigate the effects of operational variables (ventilator frequency, percent inspiratory time, airway pressure) used with High Frequency Jet Ventilation (HFJV). In the first study, ventilator frequencies above 300 cycles/min and percent inspiratory times above 30% were associated with increased arterial carbon dioxide tensions. Higher peak airway pressures gave lower carbon dioxide tensions. In the second study, nitrogen washout times in a mechanical analog lung model were measured for the same HFJV operational variables. Highest calculated gas dispersion coefficients, indicating maximal mixing and most rapid washouts, were achieved at 150 cycles/min and were associated with higher peak airway pressures and airway pressure excursions (difference between peak and end-expiratory pressure). In the third study, airway pressure excursion was independently controlled and found to have a greater effect on arterial blood gas tensions than did either mean or end-expiratory pressures.

INTRODUCTION

The purpose of breathing, whether as a spontaneous act or supported by artificial means, is to exchange respiratory gases with the blood. There are times, however, when a patient cannot breath on his own. When this occurs, some assistance must be given to temporarily assume the ventilatory process. Although many devices for providing artificial respiration have been developed through the years -- from fireplace bellows to the iron lung -- none are perfect replacements for natural breathing.

The primary problem with most means of artificial respiration is that they depend on a positive (above atmospheric) pressure to drive gases into the lung. This is exactly opposite of the normal breathing process in which expansion of the chest and downward deflection of the diaphragm cause gas to move into the lungs by transiently producing a negative (sub-atmospheric) pressure. This transient negative pressure is also transmitted to other organs and tissues in the thorax and aids them in their functioning. For example, when the pressure in the chest becomes less than that surrounding the rest of the body, blood is pushed into the central veins leading to the heart. This enhancement of venous return results in better filling of the heart and an increase in stroke volume and cardiac output.

When a positive pressure is produced in the chest by positive-pressure ventilation, the increased pressure has detrimental effects. These effects include decrease in venous return, and therefore cardiac output (Cournand, et al, 1948; Braunwald, et al, 1957; Morgan, et al, 1969). Use of positive-pressure ventilation can also lead to tissue damage (barotrauma) in the lung. This barotrauma may range from minor airway or alveolar membrane disruptions and over-distensions to pneumothorax, pneumomediastinum, pneumopericardium, interstitial emphysema, and bronchopulmonary dysplasia (Mushin et al., 1980).).

All positive-pressure mechnical ventilation exposes a patient to some risk. The iron lung, and its successor, the Cuirass ventilator, probably provide the closest approximation of natural physiology by supporting ventilation through application of a negative pressure to the chest and thorax. However, these ventilators are usually impractical and always difficult to apply.

The search continues for a convenient, practical means of providing ventilatory support with minimal risk to the patient. It is for this reason that there has been so much interest in the new techniques recently advanced under the blanket term "High Frequency Ventilation". Although equipment, methodologies, and operational variables are different from investigator to investigator, there is a common goal: to provide adequate exchange of respiratory gases with minimal negative homeostatic or pathologic impact.

Background of HFV Development

High frequency ventilation (HFV) is a name given to several approaches for providing ventilatory support at rates higher than those experienced with normal spontaneous breathing or conventional positive pressure mechanical ventilation (CMV). These techniques generally employ tidal volumes approaching or smaller than the anatomical dead space of the airways and lungs. The pressures in the airways and lung are also usually less with HFV than with CMV.

Since the tidal volumes associated with HFV are less than the dead space, the physiology of gas exchange with HFV is presumed to be different than that with CMV. When spontaneously breathing, the volume of gas taken in with each inhalation is enough to fill the upper airway (mouth, nose, pharynx, etc.), the conducting airways (trachea, bronchi, bronchioles), and the alveolar region of the lung. This gas is moved as a plug or bolus by convective transport due to the pressure difference between the lung and the atmosphere. In the alveolar region, the inspired gas mixes by molecular diffusion with gas that has reached equilibrium with the blood flowing through the pulmonary capillaries (Comroe, 1965). This is also the gas exchange process presumed to be present with conventional positive pressure ventilation. However, if the volume of gas inhaled is less than the dead space, it will not reach the alveolar region of the lung by convection. If gas exchange with the alveolus is to take place under these conditions, then some mechanism of gas transport other than plug-flow convection or molecular diffusion must be responsible.

There are several different methods by which the small, rapid gas pulsations associated with High Frequency Ventilation can be delivered to the airway. Each method has a different impetus driving its development. High Frequency Oscillatory Ventilation (HFOV) was pioneered by physiologists looking for a means of enhancing the gas exchange produced by introducing a steady flow of oxygen into the trachea via a small catheter. This method, called "diffusion respiration" could maintain adequate oxygenation, but carbon dioxide would accumulate in the blood. When the investigators applied pressure oscillations to the airway at high rates, they found that they could remove CO_2 (Lunkenheimer et al., 1972). High Frequency Positive Pressure Ventilation (HFPPV) was developed by researchers searching for a means of ventilating animals without causing hemodynamic artifacts during cardiovascular experiments. This technique involved an attempt to maintain the same minute alveolar ventilation by increasing rate and decreasing tidal volume (Sjostrand, 1977). The main differences between CMV and HFPPV are the rate, tidal volume, and the lack of an expiratory valve which closes during inspiration to prevent loss of gas through the expiratory limb of the breathing circuit. High Frequency Jet Ventilation (HFJV) grew out of the use of catheters to ventilate during laryngoscopies and bronchoscopies. It involves the delivery of a small pulse of gas to the airway through a narrow lumen or catheter. The gas comes from a high pressure source and is periodically pulsed by some sort of valve. Again, there is no expiratory valve and HFJV can be used without an endotracheal tube or breathing circuit (Quan et al., 1983; Klain and Smith, 1977).

The first of these efforts, High Frequency Oscillatory Ventilation, had its roots in the description of augmented dispersion. Emerson, applying knowledge of the enhanced dispersion and augmentation of effective diffusivity which takes place in a pulsating column of gas patented a device in the late 1950's for "vibrating the airways" (Emerson, 1959). This device could probably be considered to be the original High Frequency Oscillatory Ventilator.

Meanwhile, anesthesiologists caring for patients undergoing anesthesia for bronchoscopy and laryngoscopy were attempting to devise techniques by which they could provide airway control and ventilation while not competing with the surgeon for the operative field. External appliances such as the Cuirass ventilator (Green and Coleman, 1955), a modification of the iron lung technique, made it possible to ventilate the patient; but not without a considerable amount of difficulty and complexity of equipment. Sanders devised attachments for the bronchoscope which allowed him to ventilate through the same tool that the surgeon was using in the airway (Sanders, 1967). His device was extremely simple, consisting of a regulator and a toggle valve attached to a high pressure gas source and connected at the distal end to a channel in the bronchoscope. This "Sanders Jet Technique" was soon adopted for laryngoscopies as well, with various attachments made for operating laryngoscopes. Further adaptation of the technique, including cricothyroid puncture for percutaneous transtracheal jet ventilation (Klain and Smith, 1977), were soon developed. While these techniques were used at normal ventilatory rates, they encompassed many of the attributes of High Frequency Jet Ventilation (HFJV) including entrainment, passive exhalation, no expiratory valve, and control of inspiratory gas delivery by the interruption of a high pressure gas source.

Concurrent with these developments in respiratory support and anesthesia management, physiologists involved in hemodynamic studies in animals were searching for a means of providing respiratory support without the ventilatory artifact that positive-pressure ventilation, or even spontaneous breathing, introduces into cardiovascular measurements. Sjostrand and associates developed the idea of reducing tidal volume and increasing ventilatory rate to maintain a minute volume which would provide adequate gas exchange while lessening artifact (Sjostrand, 1977). In order to accomplish this, they felt that they had to reduce the dead space of the animal and ventilator circuit as much as possible. This included reduction of the internal compressible volume of the ventilator system. Through a series of technical developments they were able to bring about the technique called High Frequency Positive Pressure Ventilation (HFPPV) using rates up to 100 cycles/ min. Their clinical efforts concentrated on the techniques for bronchoscopy and laryngoscopy and their device adaptations were somewhat similar to those already developed for the automation of the Sanders technique. But there was one major difference in approach. The Sanders injectors were designed to induce a certain amount of entrainment and used this effect to dilute the oxygen to a pre-determined value via the venturi effect. Sjostrand concentrated on eliminating entrainment. This was accomplished by the use of what they called the "pneumatic valve principle." This approach allowed for the introduction of a high flow of gas to the proximal end of the endotracheal tube, thereby bypassing the compressible volume and dead space of the ventilator circuit. However, no attempt was made to exploit the flow characteristics of a jet.

The next step was tha use of the jet injector at higher than normal ventilatory rates. Although automatic controllers had already been developed for the Sanders injectors previously discussed, Klain and Smith designed a new jet controller based on fluidics which could deliver the jet pulsations to the airway at rates of up to 600 cycles/min (Klain and Smith, 1976). They used soft, narrow catheters to deliver the jet pulsations to the airway. At times, these catheters were inserted percutaneously through the cricothyroid membrane into the trachea. For routine clinical use, a suction catheter was inserted into the trachea without the use of an endotracheal tube. A variation on this technique used a catheter contained within an endotracheal tube to deliver the pulses to near its distal tip. Another method involved an adapter which placed the jet near the proximal end of the endotracheal tube (Carlon et al., 1981). Whether the jet was introduced percutaneously, at the distal tip of the endotracheal tube, or via a proximal adapter, there was no expiratory valve, and hence no way to directly control tidal volume. In addition, the proximal injector promotes significant entrainment of gas from the ventilator circuit. This effect exists to a lesser extent with the more distally introduced jets.

While development of High Frequency Positive Pressure Ventilation and High Frequency Jet Ventilation were proceeding, other physiologists were exploring High Frequency Oscillatory Ventilation. This technique grew out of interest in "diffusion respiration" in which a steady flow of oxygen is delivered to the airways of an otherwise apneic subject. While this technique was capable of providing adequate oxygenation for extended periods of time, carbon dioxide was not removed and would accumulate in the blood. Lunkenheimer, in a manner similar to that patented earlier by Emerson, found that by superimposing rapid pressure oscillations on the airway, carbon dioxide could be washed out of the lung while oxygen diffused in (Lunkenheimer et al., 1972). Since these pressure pulsations were sinusoidal oscillations imposed by a loudspeaker driver or piston, the name High Frequency Oscillatory Ventilation was used.

Characteristics of High Frequency Ventilation Devices

The various devices and techniques for delivering High Frequency Ventilation were developed for several different applications and by several different investigators. For this reason, there are many differences in how the ventilatory pulsations are derived, controlled, and delivered to the airway. In order to classify these different techniques, it is helpful to define whether a device has active or passive inspiration, active or passive exhalation, is open or closed, allows entrainment or not, where the gas is delivered to the airway, the magnitude of the delivered tidal volume in relation to the anatomic and equipment dead space, and what the effective frequency range is for adequate ventilatory support.

High Frequency Positive Pressure Ventilation (HFPPV) is an open system with active inspiration and passive exhalation. This means that there is no expiratory valve and gas can escape during the inspiratory phase. This is unlike a conventional ventilator where an expiratory valve closes during inspiration to allow pressurization of the airways until a set delivered volume or pressure is reached. Inspiration with HFPPV occurs because gas is supplied to the airway more rapidly than it escapes through the open expiratory limb. This causes pressure to rise above atmospheric and pushes gas into the lungs. Expiration takes place due to the elastic recoil of the lungs when the rapid gas input to the airway is stopped. The inspiratory gas is delivered to the proximal endotracheal tube and is introduced in such a way as to preclude entrainment. Typical HFPPV ventilatory rates range from 60 to 200 cycles/min, but are usually below 100 cycles/min. A variation of HFPPV is sometimes employed using a conventional ventilator (with an expiratory valve) operated at high rates and small tidal volumes. However, when this is done, there is a significantly greater compressible ventilator circit volume and it is not as effective as using a ventilator designed for HFPPV. Since HFPPV is an open system, with no valves in the expiratory limb, a patient can breath spontaneously while it is being used by inhaling through the open breathing circuit. Tidal volumes are approximately equal to the dead space volume, perhaps just slightly larger.

High Frequency Oscillatory Ventilation (HFOV) uses a closed or semi-open system. The oscillator may be either a diaphragm, such as a large loudspeaker, or a piston. It provides for both active inspiration and active exhalation by driving gas into the airway on its forward stroke, and sucking it out on its backstroke. Usually, some form of fresh gas is introduced between the oscillator and the airway. This gas provides the oxygen to move into the lung and washes out the carbon dioxide removed from the lung. Systems with a simple fresh gas flow which is flushed through the ventilatory circuit and exhausted through a length of tubing can be considered to be semi-open. They allow for spontaneous breathing by being able to inhale gas back in through the exhaust tubing. This feature comes at the expense of the loss of some gas during inspiration. However, other systems use valves on the fresh gas flow to impose a high impedance which prevents loss of inspiratory pressure, but also precludes spontaneous breathing. In either case, a flow must be maintained through the system. The tidal volumes with HFOV are generally much less than the anatomic dead space and may approach the compressible volume of the ventilator itself. HFOV has been used at rates from 120 cycles/min to 2400 cycles /min, but is usually used at approximately 900 cycles /min.

High Frequency Jet Ventilation (HFJV) is accomplished by delivering pulses from a high velocity jet into the airway. It is an open system without an expiratory valve. Usually the jet is delivered to one of two sites. In one case, the jet is delivered to the proximal end of the endotracheal tube by use of an adapter which positions the jet in the center of the tube a short distance below the ventilator circuit connection. This placement promotes significant entrainment of gas from the ventilator circuit, but probably allows much of the jet's velocity to be dissipated before it reaches the airways of the lung. In the second case, the jet is delivered near the distal tip of the endotracheal tube, usually through a port extruded in the wall of the tube itself. Figure 1 schematically shows a jet ventilator and an endotracheal tube with a jet catheter and airway pressure monitoring lumen in place. Although there is still entrainment, it is less than with the proximal jet. However, the jet appears to maintain its high velocity further into the lung. Inspiration is active, driven by the introduction of gas into the airway more rapidly than it is escaping. Expiration is passive, depending on the elastic recoil of the lungs and chest wall. Tidal volumes are difficult to measure with the jet because of how and where it is introduced, but estimates place the tidal volume at or below the anatomic dead space. Because the jet is delivered through a narrow catheter at high pressures and flows, the compressible volume of the ventilator is negligible. Also, when the jet is delivered to the distal tip of the endotracheal tube, additional dead space is by-passed. The effective frequencies for HFJV range from about 60 cycles/min to 900 cycles/min, with it usually applied between 100 cycles/min and 300 cycles/min.

The particular High Frequency Jet Ventilation system used in the investigations described here delivers the jet pulsations to the distal endotracheal tube via a lumen extruded in its wall. An anesthesia circle

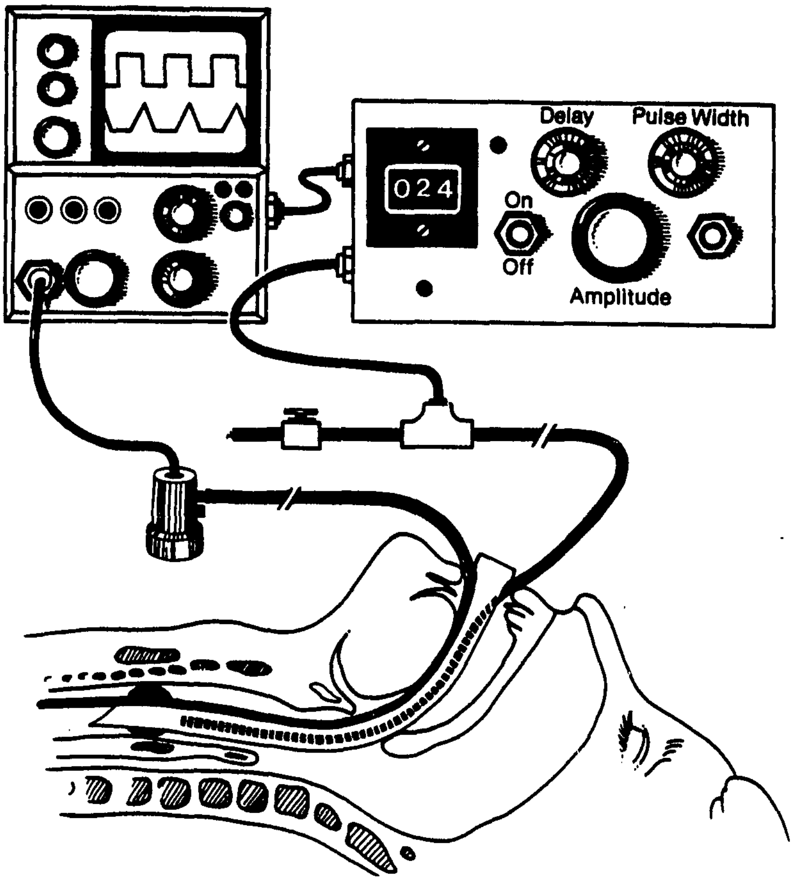


Figure 1

HIGH FREQUENCY JET VENTILATION

SCHEMATIC OF DEVICE FOR DELIVERING HIGH FREQUENCY JET VENTILATION THROUGH ENDOTRACHEAL TUBE LUMEN breathing circuit was used to provide a reservoir of entrainment gas of known and controlled composition. The same circuit was used with an anesthesia ventilator for CMV comparisons.

Gas Exchange With High Frequency Jet Ventilation

Because the tidal volumes and rates are different from those used with conventional positive pressure ventilation, there has been significant interest in gas exchange with the lung during High Frequency Ventilation. This interest includes curiosity about intra-pulmonary gas transport mechanisms and the effects of the operational variables of the ventilators themselves on gas exchange efficiency.

With conventional volume ventilation, an attempt is made to provide tidal volumes and rates similar to those provided by spontaneous ventilation. This has led to operational rules-of-thumb based on patient size, age, and weight which allow a clinician to estimate appropriate initial settings for the ventilator. He then adjusts settings to fine-tune to the physiologic needs of the patient. Under the conventional rules, minute ventilation (the product of rate times tidal volume) is assumed to govern carbon dioxide removal. Oxygenation is controlled by adjustment of inspired oxygen concentration and by mean airway pressure. These rules are based on assumptions about gas transport in the lung which state that the tidal volume must be larger than the anatomic dead space in order for gas to be exchanged with the alveolar units. This tidal volume is moved by convection in the central and distal conducting airways, gradually losing velocity as the cross-sectional area of the lung exponentially increases. Velocity approaches zero near the alveolar level. Mixing with alveolar gas and exchange across the alveolar membranes into the pulmonary capillaries occurs by molecular diffusion, driven only by the concentration gradients of the various gases. The role of mean airway pressure is to hold more of the alveolar areas of the lung open. This increases the avilable diffusion area. Increased oxygen concentration speeds the diffusion process because of a steeper concentration gradient. Increasing minute ventilation increases the convective exchange within the lung. This aids diffusion of carbon dioxide from the terminal airways and alveolar regions.

However, with High Frequency Ventilation the tidal volume is approximately equal to or smaller than the anatomical dead space made up by the conducting airways. Therefore, the rate of gas exchange by other processes must be increased to compensate for decreased convection. Furthermore, the role of ventilator settings on gas exchange is not well definded with High Frequency Ventilation. Hence the clinician has no rules-of-thumb to govern initial operational settings or subsequent adjustments.

This situation has led to a renewed interest in intra-pulmonary gas transport phenomena. Several hypotheses have been advanced in an attempt to explain the effectiveness of High Frequency Ventilation. Some of these deal with convective transport and turbulent mixing, such as the concepts of coaxial flow (Haselton and Scherer, 1980) and axial dispersion (Scherer et al., 1975; Ultman and Thomas, 1979). Others are based on augmentation of the diffusion process by enhanced effective diffusivity (Solway et al., 1984). All of these theories deal with some mechanism of enhanced mixing of gas in the airways due to flow.

Scope of Present Investigation

The research detailed in this thesis is an attempt to explore the effects of certain operational variables and possible gas transport mechanisms contributing to gas exchange effectiveness with High Frequency Jet Ventilation. The primary emphasis is on identifying rules for adjusting operational variables to obtain a desired physiologic result. Investigation of gas transport phenomena is a secondary goal and is undertaken as an aid in attempting to understand the effects of the operational variables on gas exchange.

The operational variables studied were ventilatory frequency, percent inspiratory time, and airway pressures. Percent inspiratory time is defined as the ratio of the time the jet is on during inspiration divided by the time for a complete ventilatory cycle. Airway pressures were controlled by setting jet drive pressure. Since they are a result of frequency, percent inspiratory time, and drive pressure, airway pressures are not entirely independent variables. However, they were controlled as such in these experiments. It is reasonable to do this because airway pressure is easily measured and can be used as a basis of comparison from delivery system to delivery system. Drive pressure cannot be used this way because the resultant delivery to the airway depends on sizes and lengths of connecting tubes and jet catheters. In order to define the effects of operational variables with High Frequency Jet Ventilation, three investigations were conducted.

The first study investigated the effects of varying ventilator settings (frequency, percent inspiratory time, peak airway pressure) on the resultant arterial carbon dioxide tension ($PaCO_2$)of dogs. In this study, $PaCO_2$ was used as an indicator of gas exchange effectiveness. This study is the second chapter.

The second investigation examined the effects of ventilator settings on washout times and concentration gradients within a mechanical analog lung model. This second experiment used ventilator settings similar to those used in the first animal experiment. The purpose of the study was to try to identify the possible mechanism of intra-pulmonary gas transport with HFJV and how that mechanism was affected by variations frequency, percent inspiratory time, and peak airway pressure. This study is presented in the third chapter.

The last investigation was designed to more fully explore the effects of airway pressures on gas exchange in animals. The results of the first and second studies had suggested that gas exchange efficiency, as indicated by arterial carbon dioxide tensions in animals and by washout times in the model, was dependent more on airway pressure excursion (defined as the difference between peak and end-expiratory airway pressure) than on peak airway pressure. Airway pressure excursion was independently controlled at a single frequency and percent inspiratory time combined with a pre-determined mean or end-expiratory pressure. This experimental design allowed the study of each airway pressure variable (mean and end-expiratory pressure, airway pressure excursion) separately for its effect on gas exchange with the lung. This study is described in the fourth chapter.

HIGH FREQUENCY JET VENTILATION: EFFECTS OF OPERATIONAL VARIABLES ON ARTERIAL CARBON DIOXIDE TENSION

High Frequency Jet Ventilation (HFJV) has been shown by several researchers to be capable of providing adequate respiratory support (Carlon and Howland, 1983; Smith and Babinski, 1983; Quan et al, 1983). However, as mentioned in the introduction, there are differences in gas transport characteristics between HFJV and conventional postitive pressure ventilation. Thus, the usual guidelines for adjusting ventilator settings are not applicable. For this reason, an investigation into the effects of the variables which can be set by the user of a high frequency jet ventilator (HFJV) was performed. The relative efficiency of various combinations of operational settings has been determined by comparing the arterial carbon dioxide tension ($PaCO_2$) obtained with each combination. $PaCO_2$ is assumed to be controlled by the rate of alveolar ventilation, as described earlier, and so should be an indicator of the relative efficiency of each ventilator setting in promoting gas exchange with the lung.

The variables that the HFJV user has direct control over are frequency and percent inspiratory time.

Percent inspiratory time (I%) is defined as the amount of time the jet is on divided by the time of the total ventilatory cycle. The operator can also control the drive pressure of the gas supplied to the jet. However, variations in sizes of jet catheters and locations of introduction of the jet into the airway will result in differences in resultant gas flow and airway pressures. For this reason, it was decided to measure and control peak airway pressure as one of the operational variables rather than drive pressure.

Methods

Six mongrel dogs (8-30 kg body weight) were anesthetized with pentobarbitol (30 mg per kg) and intubated with an 8 mm inside diameter Hi-Lo jet endotracheal tube (NCC division of Mallinkrodt, Argyle, NY). The endotracheal tube had a jet ventilation lumen extruded in the side wall with a jet area to exhaust lumen area ratio of approximately 1:11. The jet lumen opened into the endotracheal tube near its distal tip. The proximal end of the endotracheal tube was connected to a standard circle absorber anesthesia breathing circuit with a 3 liter reservoir bag (Model 20, Ohio Medical Products, Madison, WI). The inspired oxygen concentration was controlled to 40% through an oxygen-air blender supplying both the ventilator and the anesthesia circuit. The breathing circuit supplied gas for entrainment. Jet ventilation was provided by a prototype High Frequency Jet Ventilator capable of delivering jet pulses at rates from 80 to 900 cycles/min with variable percent inspiratory time and drive pressure.

Six other dogs of similar size were also anesthetized and intubated with standard 8 mm endotracheal tubes. These six control animals were connected to a conventional positive pressure ventilator (Bird Mark 7, Bird Medical, Palm Springs, CA).

All animals were cannulated in a femoral artery for the measurement of arterial blood pressure and for the removal of samples to determine blood gas tensions. An intravenous catheter was placed to provide maintenance fluids (crystaloids, 5ml/kg/hr) and drugs. The animals were paralyzed with pancuronium bromide (0.1 mg/kg).

Airway pressures on the HFJV animals were measured 6 cm beyond the distal tip of the endotracheal tube via a 3mm ID polyethylene catheter attached to the outside of the endotracheal tube. This catheter was connected to a pressure transducer (PM5ETC, Gould-Statham Instruments, Inc., Hato Rey, Puerto Rico) and recorder (8805B, Hewlett-Packard, Waltham, MA) which had previously been checked for adequate frequency response (natural frequency = 25 Hz; damping coeffecient = 0.13).

Baseline ventilator settings on the HFJV animals were determined by adjusting peak airway pressure at a rate of 150 cycles/min and 30% inspiratory time until normocarbia was achieved, defined as arterial carbon dioxide tension of 40 + or - 5 torr. The conventionally ventilated animals were started at a tidal volume of approximately 25 ml/kg and rate of 10 breaths per minute. Tidal volume and/or rate was then adjusted to maintain normocarbia.

After baseline values for peak airway pressure at 150 cycles/min and 30% inspiratory time were established, one of 15 combinations of frequency, I%, and peak airway pressure was randomly set. The animal was ventilated for 10 minutes to allow blood gas tensions to stabilize. A blood gas sample was then drawn and the next setting established.

The 15 operational variable combinations used were chosen to test each ventilator control as an independent variable. To do this, only one control was varied from baseline at a time. Percent inspiratory time was kept at 30% and peak airway pressure was maintained constant by adjustment of drive pressure while frequency settings of 100, 150, 200, 300, 450, 600, and 900 cycles per minute were tried. With frequency set at 150 cycles/min and peak airway pressure maintained by adjustment of drive pressure, percent inspiratory times were varied to 10, 20, 30, 40, and 50%. Finally, with a frequency of 150 cycles/min and 30% inspiratory time, peak airway pressure was varied to 5, 10, 15, and 20 cmH_2O.

Data was analyzed by linear regression.

Results

Plots of arterial carbon dioxide as a function of the operational variables frequency, percent inspiratory time, and peak airway pressure are shown in figures 2, 3, and 4 respectively. In general, increasing frequency was associated with increasing arterial carbon dioxide. Increasing percent inspiratory time was also associated with increasing arterial carbon dioxide. However, a minimum was noted at 20% inspiratory time. While not statistically significant, there appears to be an optimum percent inspiratory time at 20 to 30%.

Peak airway pressure showed a strong and statistically significant ($p<0.05$) correlation with arterial carbon dioxide with lower $PaCO_2$ at higher peak airway pressures. Although the study was not designed to investigate them as independent variables, positive

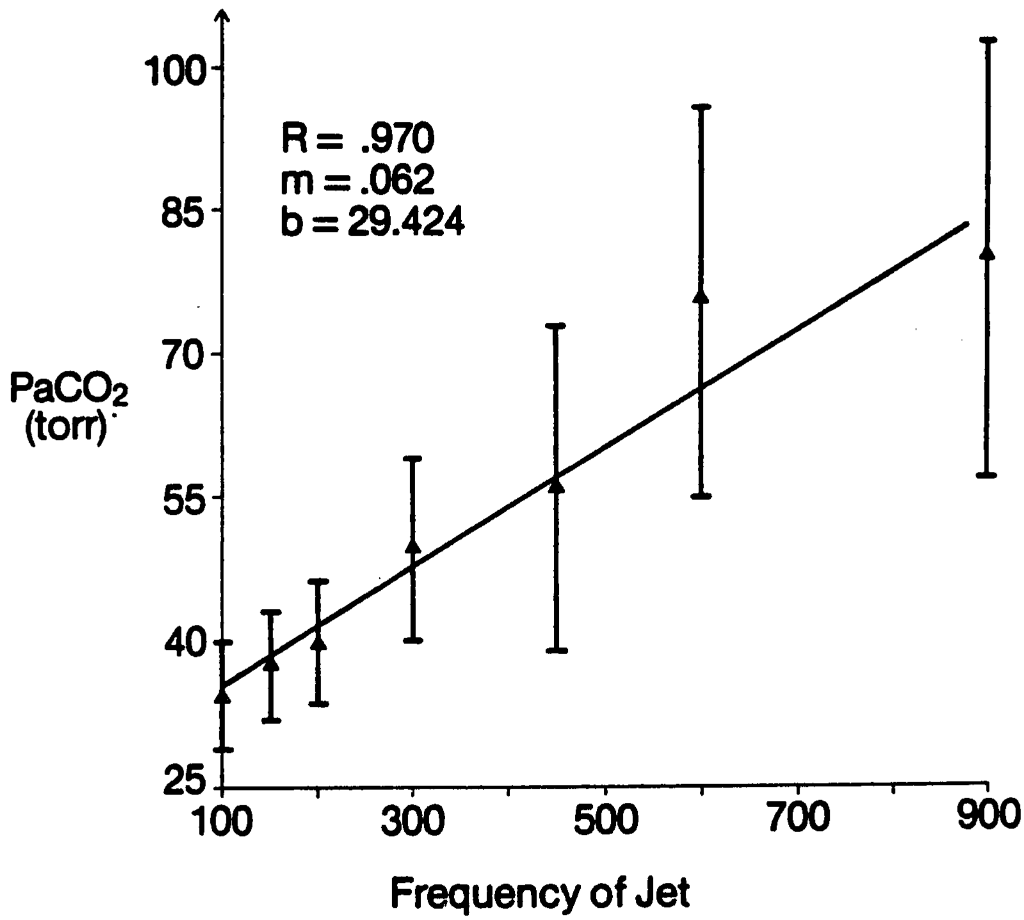


Figure 2

Jet Frequency vs $PaCO_2$

Arterial carbon dioxide tension shown as a function of ventilator frequency (cycles/min). Values shown are mean $\pm$ standard deviation for six dogs at a constant percent inspiratory time (30%) and peak airway pressure.

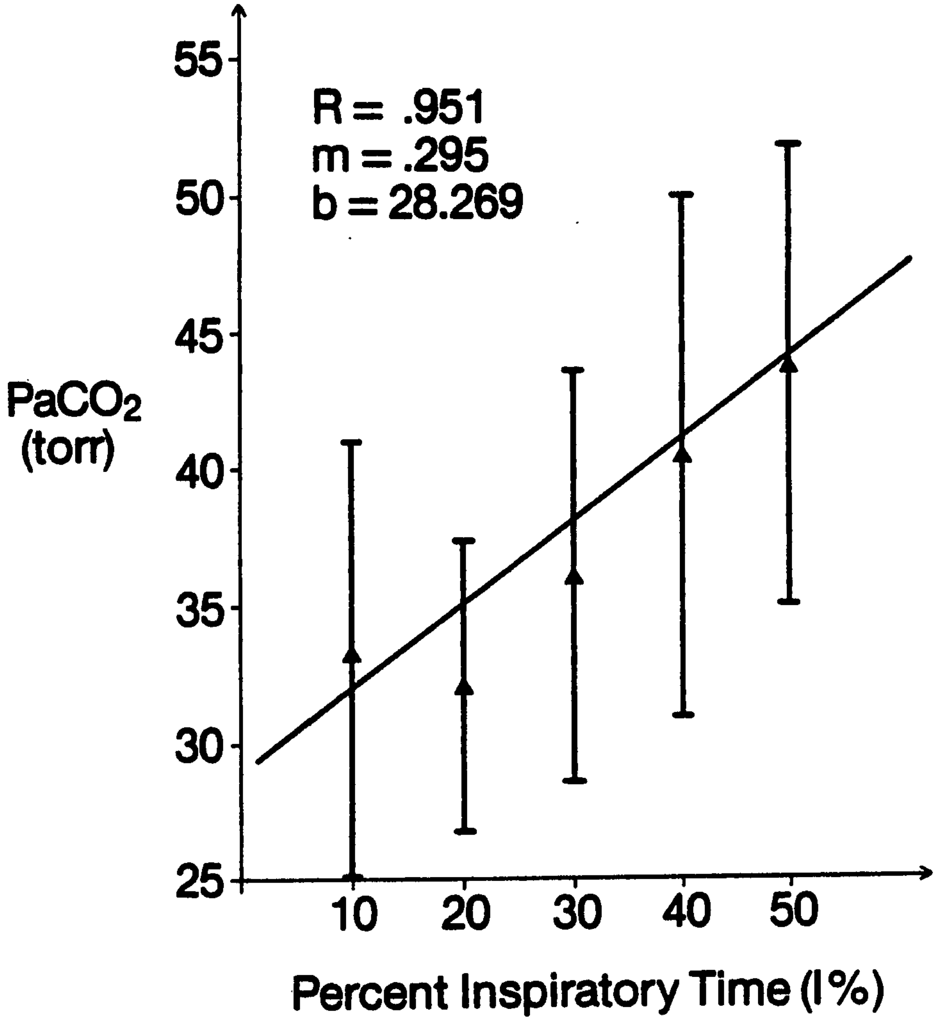


Figure 3

Percent Inspiratory Time vs $PaCO_2$

Arterial carbon dioxide tension shown as a function of percent inspiratory time at a constant ventilator frequency (150 cycles/min) and peak airway pressure. Values shown are mean $\pm$ standard deviation for six dogs.

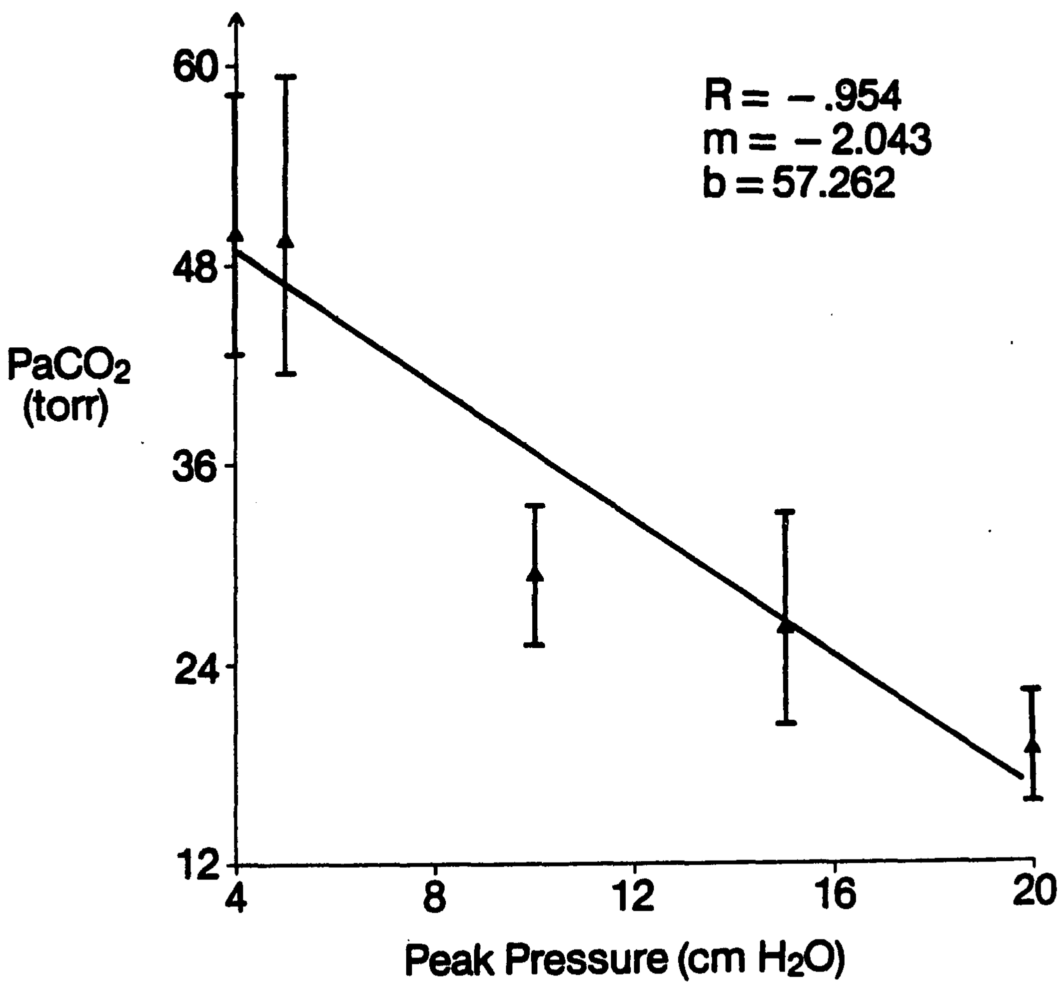


Figure 4

Peak Airway Pressure vs $PaCO_2$

Arterial carbon dioxide tension shown as a function of peak airway pressure at a constant ventilator frequency (150 cycles/min) and percent inspiratory time (30%). Values shown are mean $\pm$ standard deviation for six dogs.

end-expiratory pressure (PEEP) and airway pressure excursion (difference between peak and end-expiratory pressure) also showed correlations with $PaCO_2$. Arterial carbon dioxide tension as a function of the difference between peak and end-expiratory airway pressure is shown in figure 5, indicating a decreasing $PaCO_2$ with increasing airway pressure excursion. The effect of PEEP on $PaCO_2$ is shown in figure 6, indicating higher arterial carbon dioxide tensions at low values of PEEP.

Discussion

HFJV seems to depend for its effectiveness on a balance of the properties of the gas delivery system with the mechanical characteristics of the airways and chest. How the operational variables of frequency, percent inspiratory time, and drive pressure interract with this mechanical balance is not entirely clear. Using arterial carbon dioxide tension as an indicator of gas exchange effectiveness, an attempt has been made to identify the affects of the operational variables under direct and continuous control of the user.

For dogs with healthy lungs, it appears that there is no advantage to using frequencies higher than about 300 cycles/min to remove carbon dioxide from the lung. For the settings tested, the jet was more effective at the lower frequencies, around 100 to 150

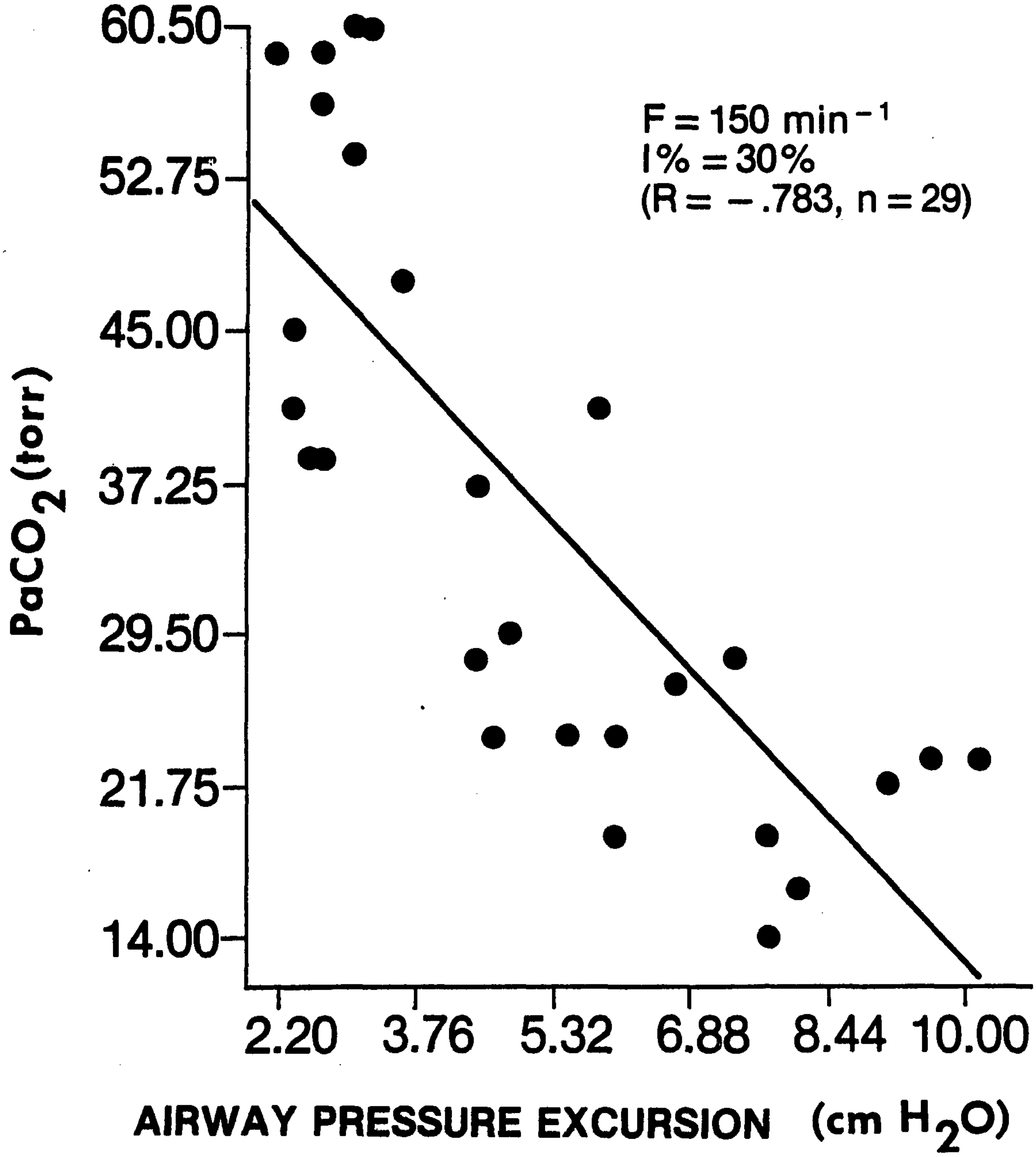


Figure 5

Airway Pressure Excursion vs $PaCO_2$

Arterial carbon dioxide tension shown as a function of airway pressure excursion (difference between peak and end-expiratory pressure).

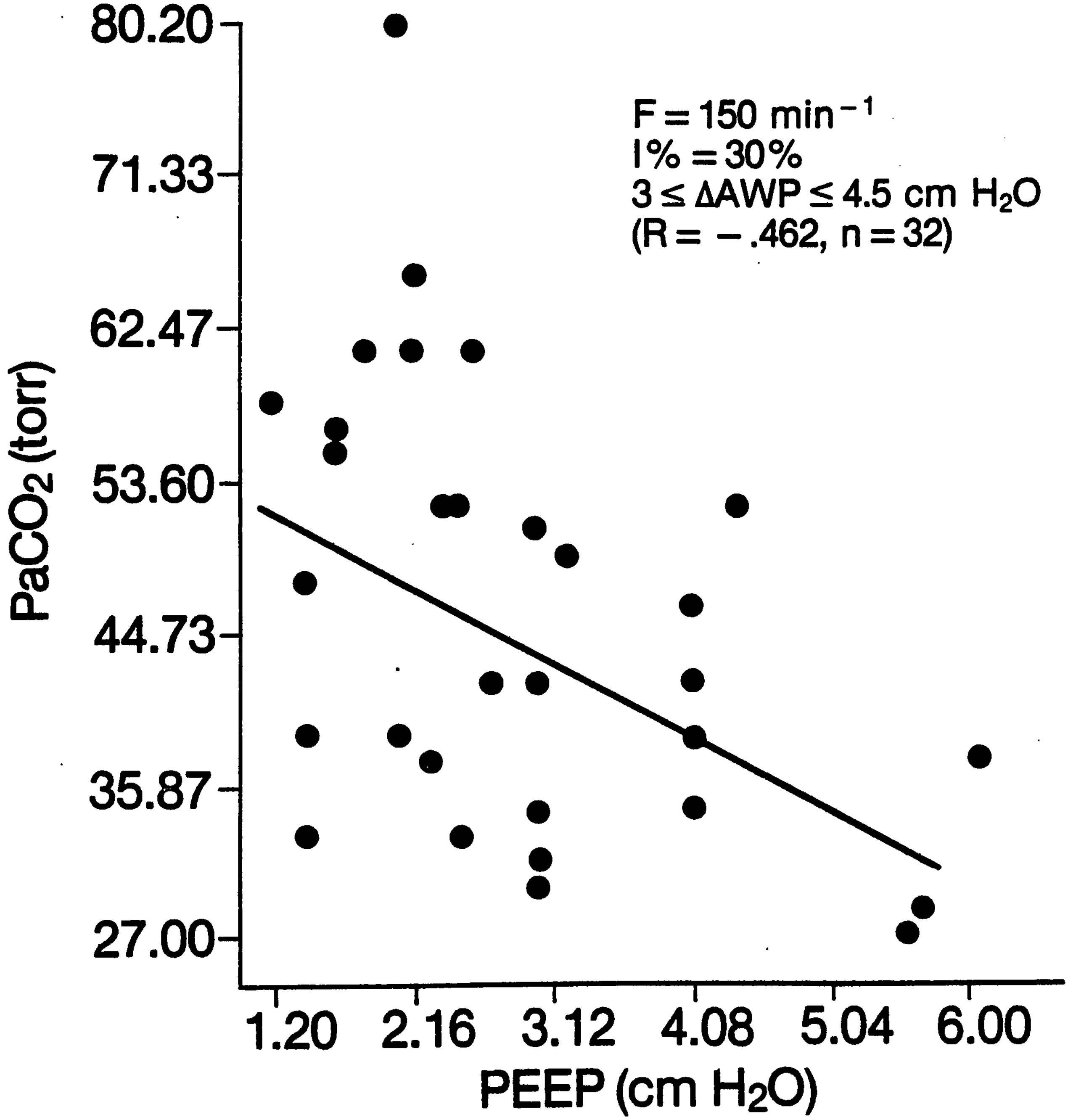


Figure 6

PEEP vs. $PaCO_2$

Arterial carbon dioxide tension shown as a function of PEEP (positive end-expiratory pressure).

cycles/min. However, this might in part be affected by the choice of 30% inspiratory time for all frequencies. At higher frequencies, the chest may not have sufficient time to empty with each breath. This exhalation is a passive process dependent on the elastic recoil of the chest and the time constants of the airways. Higher rates might have been possible with shorter inspiratory times. However, this would have required higher drive pressures in order to get sufficient gas into the lung during inspiration.

It appears that the optimum percent inspiratory time at 150 cycles/min, based on arterial carbon dioxide tensions, is 20 to 30% for the same reason. In this range of percent inspiratory times, a balance is apparently achieved between the input and the exhaust. The former is driven by the high pressure, high velocity jet through a narrow lumen while the latter is driven by lower airway pressures through the larger diameter exhaust path.

The variable which appears to have the strongest affect on gas exchange is airway pressure. Although peak airway pressure was the variable controlled in the study, it was noted that airway pressure excursions also exhibited a strong correlation with arterial carbon dioxide tensions. It appears that the reason for a correlation between larger airway pressure excursions and a lower arterial carbon dioxide tension is that airway pressure excursions are associated with volume excursions of the lung, indicating larger tidal volumes of gas exchange. However, it was also noted that some small amount of PEEP, approximately 3 cmH_2O, was necessary for effective gas exchange. This PEEP was probably necessary to keep airways open at frequencies too high and tidal volumes too low to allow them to respond with each ventilator cycle. Further investigation to separate the effects of peak, mean, PEEP, and airway pressure excursion was indicated and the results of that investigation are discussed in the fourth chapter.

For our setup, in healthy lung dogs, it appears that an initial setting of 150 cycles/min, 20 or 30% inspiratory time, a PEEP of about 3 cmH_2O, and an airway pressure excursion of approximately 4 cmH_2O will provide adequate gas exchange. To lower carbon dioxide tensions, the first attempt should be to increase airway pressure excursion by adjustment of drive pressure and percent inspiratory time. Frequency can also be used to control gas exchange, but changes in frequency must be compensated for by changes in percent inspiratory time and drive pressure to again obtain optimal airway pressure excursion.

MECHANICAL ANALOG MODEL FOR STUDY OF GAS EXCHANGE WITH HIGH FREQUENCY JET VENTILATION

The study of the effect of operational variables on arterial carbon dioxide tensions described in th previous chapter indicated what manipulations of those variables might be used to achieve a desired carbon dioxide value. However, the data from that study does not suggest what the mechanisms of gas transport may be with HFJV. Because of the difficulties inherent in studying intra-pulmonary gas exchange in an animal, models are frequently useful for developing insight into the mechanisms of gas transport within the lung. A model allows direct instrumentation for pressure and gas concentration measurements which would be difficult or impossible in vivo. Also, because conditions are more controlled, it is possible to study cause and effect relationships which may be obscured by random physiological variations in living systems. For these reasons, a model with certain specific mechanical properties was developed as an analog for the lung. This model was used to test the effects of various ventilator settings on gas exchange during High Frequency Jet Ventilation (HFJV).

Materials and Methods

Model Description

While an attempt was made to give the model the same basic mechanics as the lung, the geometry of the model was chosen to allow for studies of concentration gradients, rather than to strictly resemble the airways of the lung. This approach is justified for two reasons.

First, the geometry of the lung is extremely difficult to accurately model. It has non-symmetric branching and distributed compliance in the airways and parenchyma which make its geometry volume and pressure dependent. It also has an exponentially increasing number of branches and resultant increase in total cross-sectional area (Comroe, 1980). Previous models have approached this problem by modeling only the first few generations of branching and then terminating the model in a bulk compliance, or leaving it open (Scherer et al., 1975; Scrimshire, 1979; Jebria, 1984). These models are also often constructed as symmetrically branching tubes and capillaries. The present model departs from this approach by modeling the lung as a porous medium. This allows simulation of more generations of branching and the resistance gradations found in the lung.

The second reason for a geometry in the model different than that in the lung, is that gas dispersion (mixing of gases due to both molecular diffusion and turbulence from flow) in more than one dimension is much more complex to analyze. Dispersion is dependent on the random motion of individual molecules and its mathematical model is statistical in nature (Saffman, 1959a; Saffman, 1959b). However, in a psuedo-one-dimensional model, distance in one direction, concentration,and time are more easily analyzed (Scotter et al., 1967). While the absolute value of rates of dispersion may be expected to be different in a model with a limited cross-sectional area and an exagerrated dispersion path length, the relative increases or decreases due to different types of excitation are expected to be consistent between the model and the lung.

Having made these assumptions about geometry of the model and having a desire to match the mechanics of the lung, a model was built based on a gas dispersion apparatus used in studies of vapor transfer with the soil (Scotter, 1967). The model, shown in figure 7, consists of a plexiglass tube 5 feet in length with a 2 inch inside diameter. Ports for measuring gas concentration were located at the midpoint, and 8.5

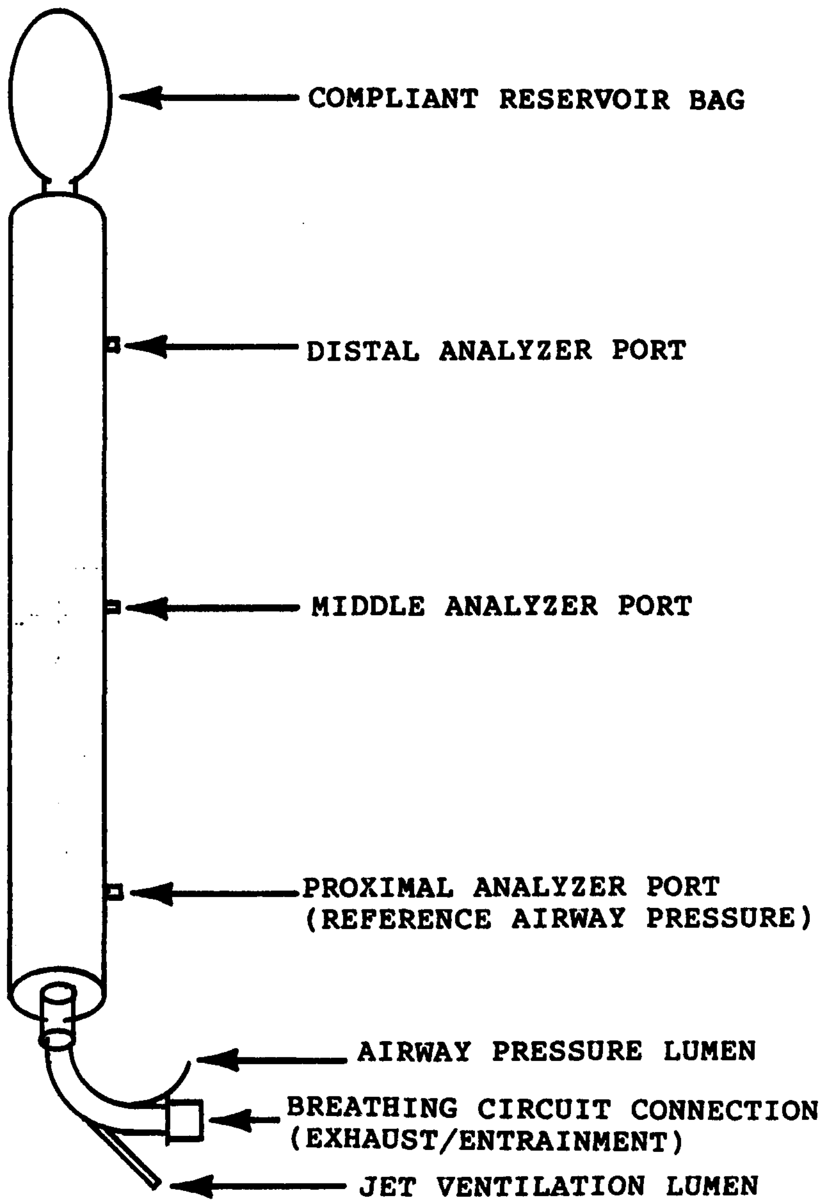


Figure 7

MECHANICAL ANALOG MODEL

APPARATUS FOR MEASURING NITROGEN WASHOUT TIMES WITH HFJV inches from each end. The bottom end of the tube had an opening which received an endotracheal tube. The top end had an opening to which a 1 liter anesthesia reservoir bag was attached. This bag served as a compliance volume. The port closest to the bottom was also used to measure pressures while establishing ventilator settings.

Resistance gradations were provided by packing the tube with various sizes of glass beads, marbles, and glass wool. These packings also served to simulate the various sizes of passages and non-symmetric branching within the lung. However, the total cross-sectional area of the model was constant, unlike that of the lung. The resistances chosen for the various regions of the lung were based on values obtained in dogs (Macklem et al., 1969). Table 1 shows how the resistances in the model compare to those of the dog.

Polarographic oxygen analyzers (IL404, Instumentation Laboratory, Lexington, MA) were used to measure oxygen concentration in the model at each of the three ports simultaneously. The concentrations were recorded at 15 second intervals by a laboratory computer (MINC, Digital Equipment Company, Maynard,MA) which stored the values and plotted them on a printer. Pressures were measured by strain-gage transducers

TABLE 1

Resistance Gradations of Dog Lung and Mechanical Analog Model

	Resistance to Flow ($cm\ H_2O$\liter\sec.)			
	Proximal Airways	Middle Airways	Distal Airways	Total Lung
Dog (60%-80% Vital Capacity)	0.49	0.14	0.47	1.1
Dog (25%-50% Vital Capacity)	0.47	1.08	0.35	1.9
Tube Model	0.40	1.00	1.30	2.7

(PM6TC Gould-Statham Instruments, Inc., Hato Rey, Puerto Rico) and recorded on a strip chart recorder (Model 2200, Gould, Instruments Division, Cleveland, OH).

Procedure

Before each data run, the tube was flushed with nitrogen. This established a known initial condition. The reservoir bag was removed, emptied, and replaced. The oxygen analyzers were removed from the tube and calibrated by checking their output to the computer in room air (21%) and pure oxygen (100%). The middle and distal analyzer ports were plugged. A pressure transducer was connected to the proximal. Another transducer was connected to the monitoring line in the Hi-Lo Jet endotracheal tube (NCC division of Mallinkrodt, Argyle, New York). Using nitrogen in both the jet ventilator and the anesthesia circuit, the ventilator settings were established and measurements made for airway pressures and exhaust flows. Exhaust flows were measured at the breathing circuit pop-off valve by an ultrasonic spirometer (VM90, Bear Medical Systems, Inc., Riverside, CA). The reference airway pressure values were always obtained from the model's proximal analyzer port rather than the endotracheal tube (ET) port because of jet flow artifact in the latter reading. However, the ET port was occasionally checked during a washout run to confirm that ventilator settings had not changed.

After calibration and measurements of pressure and exhaust flow, and while the tube was still being supplied with nitrogen, the oxygen analyzer probes were reinserted in the monitoring ports and the computer program started. At time zero, the jet ventilator driving gas and the gas supplied to the anesthesia circuit was changed to oxygen. At 15 second intervals thereafter, readings were made and recorded by the computer from each of the three oxygen analyzers. The washout was continued until the oxygen concentration in the model at the most distal analyzer (top of tube) read greater than 90%.

A total of 10 different combinations of ventilator operational variables were tested for washout times. Eight of these combinations utilized High Frequency Jet Ventilation. The combinations used tested percent inspiratory times (I%) of 10, 30, and 50% at 150 cycles/min and 8 cmH_2O peak airway pressure; frequencies of 80, 150, 300, and 900 cycles/min at 30% inspiratory time and 8 cmH_2O peak airway pressure; and peak airway pressures of 5, 8, and 19 cmH_2O at 150 cycles/min and 30% inspiratory time.

In all cases, the pop-off valve of the breathing circuit was left open, providing a slight positive end-expiratory pressure (PEEP). A constant 2 lpm bias flow was provided to the circuit ot provide entrainment gas. The exhaust flow recorded includes both this 2 lpm and the gas supplied by the jet which exhasted through the open pop-off-valve.

Two control runs were performed for comparison to the HFJV data. In one case, ventilation was provided by a conventional anesthesia ventilator at a rate of 10 cycles/min and a tidal volume of 400 ml per breath. The minute flow to the anesthesia circuit in this case was increased to 5 lpm to avoid re-breathing. In the second case, only a 2 lpm CPAP (continuous positive airway pressure) flow was used without any ventilator cycling. This condition was used to estimate the rate of washout occuring due to diffusion only.

Data Analysis

The primary dependent variable measured in each washout experiment was the time necessary to reach 63% oxygen concentration at each location in the tube. This time value was used to calculate a dispersion coefficient after the method presented by Scotter and associates (Scotter, et al, 1967). In order to use this model, a porosity for each location must be assumed.

Estimates for porosity of each packing material had previously been determined by measuring the amount of water needed to fill a beaker containing the packing material. This established the approximate ratio of void space to total volume occupied by the material. Values determined by this method for the 4mm glass beads approximately agreed with the porosity given by Scotter and associates for the same material. The porosity for the lower (proximal) tube is estimated at 0.42, the middle tube at 0.37, and the whole tube (distal port) at 0.50.

Using these values for porosity and the times and initial conditions from the washout experiments, we can solve equation (1), derived by Scotter and associates, for the dispersion coefficient.

$$(1) \qquad D = \frac{x^{2} s}{4\, t\, [\, \mathrm{erf}^{-1} (1-C/C_{1})^{2}\,]}$$

C= Oxygen Concentration at distance x and time t.

Cl= Oxygen Concentration of ventilatory gas = 100%

t= time to reach concentration C at distance x.

x= distance to analyzer port from distal tip of endotracheal tube.

s= porosity of model up to distance x.

erf= the error function

The larger the calculated dispersion coefficient, the more augmentation of diffusion is being provided by the ventilator settings. The calculated dispersion coefficients for 63% washout were plotted against the independent variables of frequency, inspiratory percent time, and airway pressure excursion (difference between peak and end-expiratory airway pressure).

Results

Washout times for each set of ventilator settings are shown in table 2. Included are 8 different combinations of HFJV conditions and the data from the conventional and CPAP runs. The calculated dispersion coefficients are also shown.

Dispersion coefficient is plotted as a function of airway pressure excursion at a frequency of 150 cycles/min and 30% Inspiratory time in Figure 8. The dispersion increases with increasing airway pressure excursion at all locations. However, the proximal dispersion coefficient was much greater than the middle or distal at an excursion of 4 cm H_2O. All three coefficients were much greater at an excursion of 9.4 cm H_2O than at 1.8 or 4.0 cm H_2O.

The effect of frequency on dispersion coefficient at a constant 30% inspiratory time and 8 cmH_2O peak airway pressure is shown in figure 9. A maximum value of dispersion for the proximal port is reached at

TABLE 2

Summary Data for Tube Model Nitrogen Washouts

RATE	I%	DRIVE (psi)	AIRWAY PRESSURE PEAK (cm H_2O)	MEAN	PEEP	ΔP	FLOW EXHAUST (lpm)	TIME TO 63% PROX (seconds)	MID	DISTAL	DISP. COEF. PROX (cm^2/sec)	MID	DISTAL
150	30	10	8.0	5.8	4.0	4.0	6.96	12	645	2520	35.0	7.2	7.3
150	30	24	19.0	14.0	9.6	9.4	19.5	10	180	645	42.0	26.0	29.0
150	30	4	5.0	4.0	3.2	1.8	3.61	245	5115	12840	1.7	0.9	1.4
150	10	22	8.0	4.9	3.0	5.0	6.83	25	660	2115	17.0	7.0	8.7
150	50	6	8.0	6.4	4.4	3.6	11.0	125	1080	3630	3.4	4.3	5.1
900	30	8	8.0	6.7	6.0	2.0	12.2	195	1050	2760	2.2	4.4	6.7
300	30	10	8.0	6.9	6.0	2.0	11.1	90	990	4320	4.7	4.7	4.3
80	30	8	8.0	5.6	3.6	4.4	8.17	80	525	1785	5.2	8.8	10.4
10*	20	--	19,5	5.5	1.2	17.3	5.0	90	1050	1980	4.7	4.4	9.3
--**	---	--	3.0	3.0	3.0	0	2.0	420	5825	11700	1.0	0.8	1.6

* Conventional positive pressure ventilation, tidal volume = 400 ml.

** CPAP flow only with no ventilator cycling.

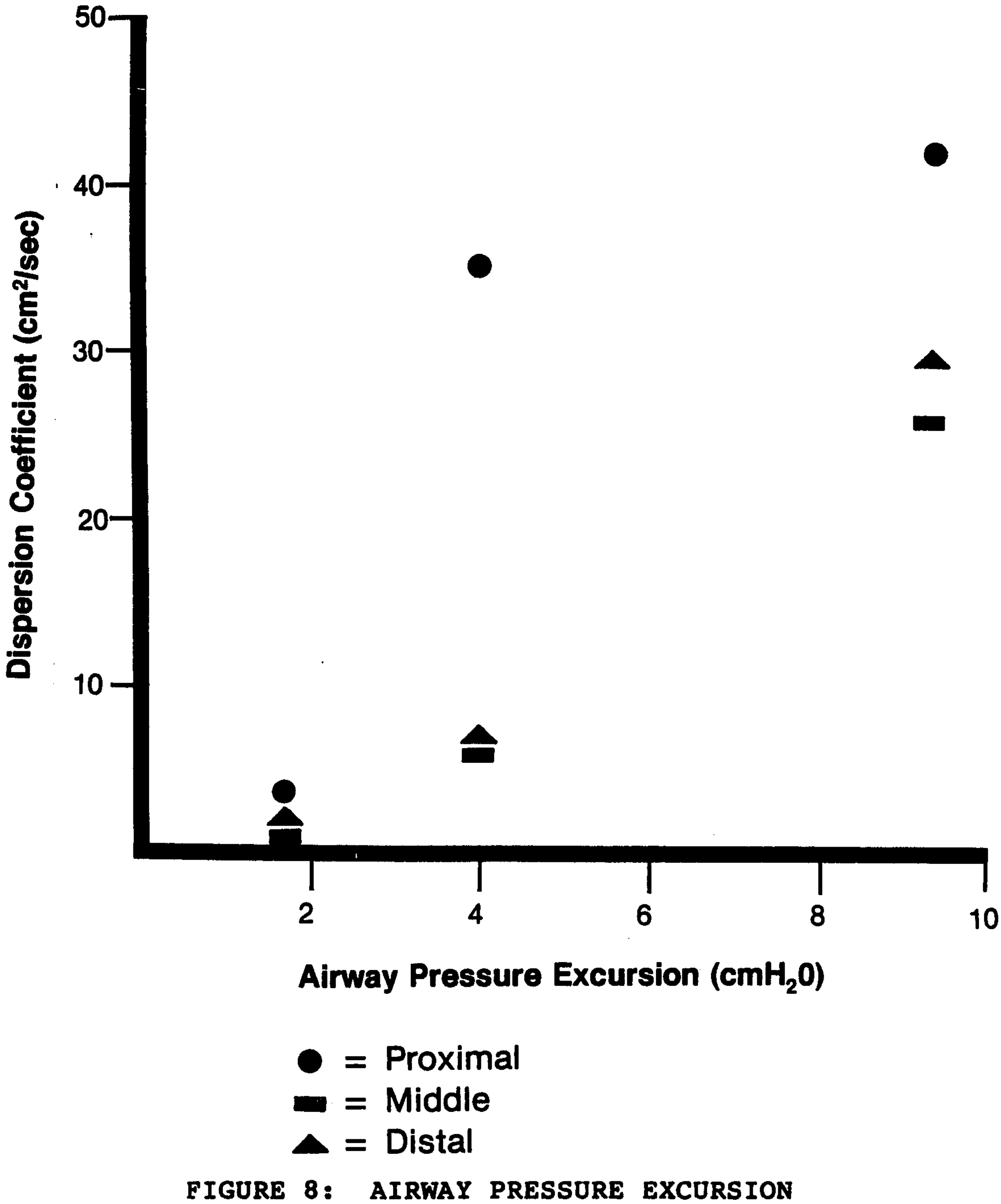


FIGURE 8: AIRWAY PRESSURE EXCURSION
vs
DISPERSION COEFFICIENT

Dispersion coefficient is shown as a function of airway pressure excursion (difference between peak and end-expiratory pressure) at each location in the model.

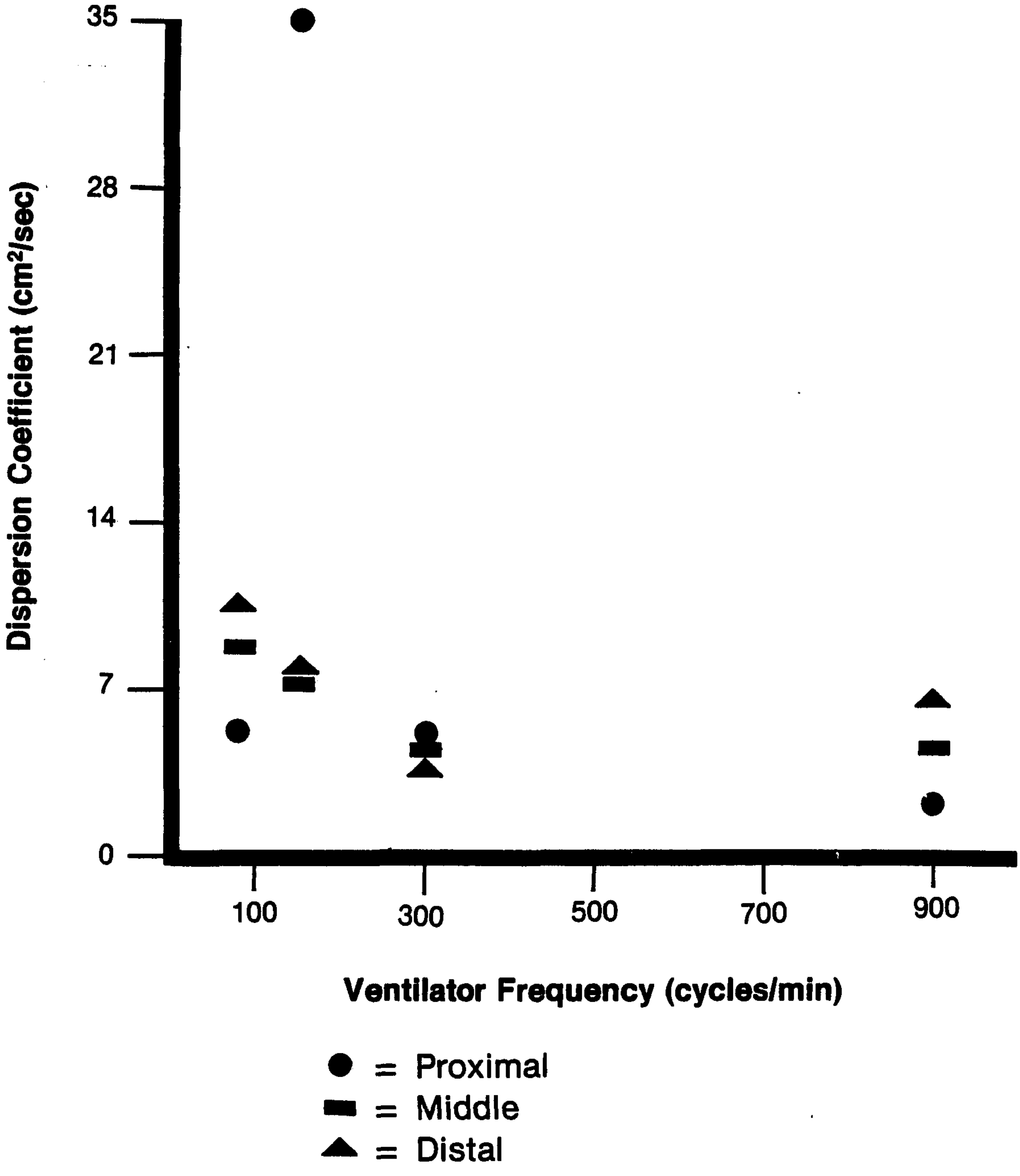


FIGURE 9: VENTILATOR FREQUENCY
vs
DISPERSION COEFFICIENT

Dispersion coefficient is shown as a function of ventilator frequency at each location in the model. Peak airway pressure and percent inspiratory time were held constant.

150 cycles per minute. The coefficients at other frequencies were not greatly different. Of note, however, is the larger coefficient for the distal than proximal port at 80 cycles/min and at 900 cycles/min.

The effect of percent inspiratory time on dispersion coefficient at 150 cycles/min and 8.0 cmH_2O peak airway pressure is shown in figure 10. The maximum value is reached at 30% for the proximal port, but 10% is as effective at the middle and distal ports. All three values are lower at 50% inspiratory time.

Discussion

With HFJV the operator has control over frequency, inspiratory percent time, and drive pressure. Through adjustment of these variables, desired peak, mean, and end-expiratory airway pressures and airway pressure excursion may be obtained. However, in order to provide adequate ventilatory support, manipulation of these variables must result in gas exchange with the lung. The purpose of this study was to examine how adjustments of frequency, inspiratory percent time, and airway pressures related to the efficiency of gas exchange during HFJV.

Because of the open character of HFJV, operating without an expiratory valve, the actual tidal volume exchanged with the lung in each ventilatory cycle is

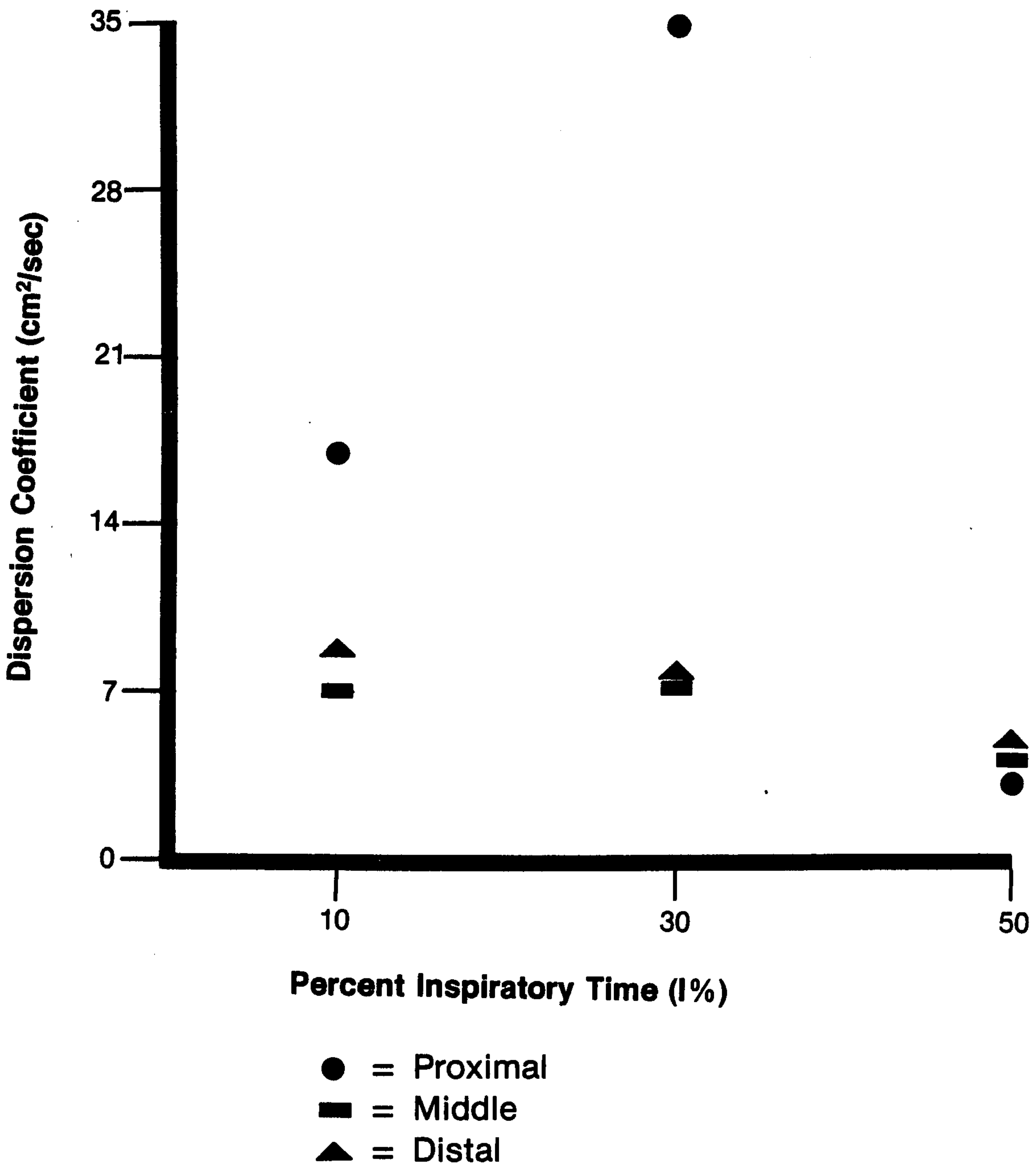


FIGURE 10: PERCENT INSPIRATORY TIME
vs
DISPERSION COEFFICIENT

Dispersion coefficient is shown as a function of ventilator percent inspiratory time at each location in the model. Peak airway pressure and ventilator frequency were held constant.

dependent on interractions between the delivery system and the mechanics of the lung. If there is a significant resistance to flow in the airways or the lung is stiff and will not easily expand, then back pressure will cause gases delivered to the airway by the jet to simply escape out the exhaust pathway instead of entering the lung. Likewise, as a volume of gas goes into the lung during inspiration it causes the pressure in the lung to increase and can begin to push jet gas out the exhaust path while the jet is still on. In either case, entrainment would also cease. During expiration, gases must be expelled by the passive recoil of the chest wall and lung.

Similarly, inspiratory and expiratory flows must be matched to allow for mass transport to balance. If gas is put into the lung more rapidly than it can be exhaled, it will accumulate causing an increase in lung volume and pressure. However, sufficient time must also be allowed during inspiration or less gas will enter the lung than can be exhaled. This balance is dependent on the time constant of the lung. Settings which provide the proper balance of inspiratory flow and expiratory time should lead to maximal airway pressure excursion. Although not a primary goal of this investigation, this was confirmed by the airway pressures obtained from the model. At the same frequency, 150 cycles/min, significantly different drive pressures were necessary at different inspiratory percent times to obtain the same peak airway pressure (See table 3). These also gave different airway pressure excursions, with the excursion increasing with decreasing inspiratory time due to more time for passive exhalation. The highest excursion, at 150 cycles/min and 10%, also gave a quicker washout of the distal tube than either 30 or 50% using the same peak airway pressure.

The situation with frequency is more complex. With varying frequency, not only must the amount of gas exchanged per cycle be considered, but so must the number of times this volume of gas is exchanged per minute. This can be seen by noting that although the airway pressure excursion is slightly higher at 80 cycles/min than at 150 cycles/min, the washout is quicker at 150. However, at either 300 cycles/min or 900 cycles/min, there is apparently insufficient time for gas to be exhaled causing an increase in PEEP and a decrease in pressure excursion. This in turn leads to a loss of gas exchange efficiency and lower dispersion coefficients.

This dependence on a match between the mechanical characteristics of the lung and the operating characteristics of the ventilator make it important to identify those operational variables which can be used to match the ventilator to the lung and define what the effect of manipulating those operational variables will be upon the efficiency of gas exchange.

The results of this investigation indicate that airway pressure excursion shows the strongest relationship to increased gas exchange efficiency as indicated by shorter washout times and higher calculated dispersion coefficients. Frequency and percent inspiratory time indicated weaker effects. If airway pressure is analagous to tidal volume with conventional ventilation, which seems reasonable, then its affect on gas exchange is as expected.

Also of interest, however, are the differences in calculated dispersion coefficients from different areas of the model. With the settings at 150 cycles/min, the larger dispersion coefficients were present in the proximal tube when larger airway pressure excursions were used. At higher frequencies and lower pressure excursion amplitudes, dispersion coefficients were calculated to be higher more distally in the model.

One possible explanation for this might be that large pressure excursions are accompanied by more back-and-forth gas movement and therefore more mixing.

However, there must also be a frequency component to this effect since the conventional ventilation at a rate of 10 cycles/min and HFJV at 80 cycles/min did not show the high proximal dispersion coefficients. This hypothesis would be consistent with the theories of Solway and associates who have advanced a model of intra-pulmonary gas transport based on augmentation of diffusivity (Solway et al., 1984). This augmentation is related to the root-mean-square velocity of the gas in the airways. However, without measurements of tidal volume or velocity or flow visualization, the applicability of this theory cannot be proved.

Another possible explanation depends on the steepness of concentration gradients and convective transport. The more rapid, larger excursions offer more convective transport to the proximal region of the model, providing a quick washout of that area and steepening the concentration gradient between the proximal area and the more distal areas of the model. This would in turn increase the rate of diffusive exchange between the proximal and distal airways. The extremely high dispersion coefficients obtained with some settings would be indicative of convective transport rather than just an augmentation of diffusive mixing. This hypothesis is also supported by the high dispersion coefficients obtained for the middle and distal regions with an airway pressure excursion of 9.4 cmH_2O at 150 cycles/min and 30% inspiratory time. At this apparently high exchange rate, it appears that the whole tube was involved in convective transport to some extent.

At first glance it is perplexing to have the distal dispersion coefficients at times be higher than the middle or proximal values. However, there are several possible reasons for this. First, the porosity value used may be a poor estimate. The lower and middle tube are packed with glass beads and the ratio of open space to total volume is easily measured. However, the upper tube contains glass wool. Since it can be packed and expanded, it is not as consistent in its porosity and not easily measured. Also, the porosity value used to calculate the distal dispersion coefficient is an average for the whole tube. Second, the distal port is near the 1 liter reservoir bag. Since this bag is compliant, there is gas movement into and out of it during each ventilator cycle, perhaps leading to increased mixing in the upper tube. Third, there was a tendancy to have increasing values of dispersion as time was increased. Since it takes longer for the distal tube to reach 63% washout than it does for the proximal and middle tube, this time dependency may have biased the estimates for the distal tube. This time effect also was part of the reason for choosing 63% rather than 90% washout for measurement. I am not sure why there should be a time factor to the dispersion coefficient. It is possible that there was a small leak in the bag or distal tube cap which was not noticed by pressure testing. However, due to the rapid washout of the proximal tube with most settings, it was not possible to choose an arbitrary time, such as 15 minutes, and calculate the dispersion coefficient from the measured concentration at that time. This approach would mean trying to calculate a coefficient for the proximal tube based on 96% washout and one for the distal tube based on 2%. This is outside of the accuracy limits of the oxygen analyzers.

The results of this study confirm that when ventilator settings are matched to the mechanical properties of the system being ventilated, airway pressure excursion and gas exchange efficiency, as evidenced by washout time and dispersion coefficient, are increased.

HIGH FREQUENCY JET VENTILATION: EFFECTS OF AIRWAY PRESSURE ON GAS EXCHANGE

Although the mechanism of intra-pulmonary gas transport with High Frequency Jet Ventilation (HFJV) is not fully understood, airway pressure has been demonstrated to be an important variable for controlling and monitoring its effectiveness (Sladen et al., 1983; Lendinez et al., 1983). Its role as an operational variable affecting efficiency of gas exchange has also been suggested by the results of the experiments reported in chapters 2 and 3 of this thesis.

During HFJV there are three reasons for monitoring airway pressure. First is safety and vigilance. Because of the high flows of gas and the high driving pressures with which they are propelled with HFJV, airway pressure can rapidly reach dangerous levels. Monitoring airway pressure allows flow to the jet to be interrupted if pressure rises due to an exhaust pathway occlusion. Second is confirmation of device performance. Malfunctions of the jet's timing device, flow interruption valve, loss of drive gas pressure, or airway disconnect would be indicated by changes in airway pressure. Third is to indirectly reflect the physiologic effect of HFJV in promoting gas exchange with the lung.

Although the necessity of monitoring airway pressure during HFJV for reasons of safety and confirmation of device performance are self-evident, the use of airway pressure as a predictor of gas exchange effectiveness is still at issue. The effects of driving pressure, jet cannula size and position, and jet flow pulse waveshape on resultant airway pressures and ventilation have been previously explored (Carlon et al., 1983; Colgan et al., 1983; Calkins et al., 1982), but the utility of airway pressure as a control variable for influencing gas exchange has not been directly investigated. The previous investigation into the effects of operational variables reported in chapter 1 used peak airway pressure as an independent variable. However, a correlation was also observed between airway pressure excursion (the difference between peak and end-expiratory pressure) and arterial carbon dioxide tension. Therefore, it is necessary to determine which airway pressure variable (peak, mean, end-expiratory, airway pressure excursion) is the most useful predictor of gas exchange during HFJV.

In order to confirm the usefulness of airway pressure as a predictive variable for monitoring and controlling the efficiency of gas exchange with HFJV, the following study was performed. In it, the relationship between independently controlled positive end-expiratory pressure (PEEP), mean airway pressure ($\overline{AWP}$), or airway pressure excursion (ΔP) and resultant arterial carbon dioxide ($PaCO_2$) and oxygen (PaO_2) tensions.

Methods

Six mongrel dogs with healthy lungs were anesthetized with sodium pentobarbital (30 mg/kg body weight). The trachea of each animal was intubated with a Hi-Lo jet endotracheal tube (NCC division of Mallinkrodt, Argyle, NY). The jet delivery lumen (approximatley 2.5 mm ID) was located in the wall of the endotracheal tube. The tube had an inside diameter of 8 mm and a jet lumen to exhaust lumen area ratio of about 1:11. The jet port was coincident with the distal tip of the endotracheal tube. The additional airway pressure monitoring lumen extruded in the wall of this tube was not used for airway pressure measurement. Instead, airway pressures were obtained from a 1 mm ID polyethylene catheter affixed to the external wall of the endotracheal tube and extending

6 cm beyond its distal tip. This catheter was connected to a pressure transducer and recorder previously tested for adequate frequency response (PM5ETC, Gould-Statham Instruments Inc., Hato Rey, Puerto Rico; 8805B, Hewlett-Packard, Waltham, MA; natural frequency = 25 Hz, damping coeffecient = 0.13) Airway pressures obtained from this external catheter were used to determine the settings of the high frequency jet ventilator (Model 300 HFV, Healthdyne Inc., Marietta, GA). Entrainment gases and control of expiratory pressure were provided by a circle absorber anesthesia breathing circuit (Model 20, Ohio Medical Products, Madison, WI). Both the jet and the entrainment circuit were supplied with blended air and oxygen at an oxygen concentration of 40%.

A catheter was placed in the femoral artery for monitoring of arterial blood pressure and sampling of blood for analysis of arterial blood gas tensions. A forepaw vein was cannulated for administration of maintenance fluids (crystalloids, 5 ml/kg/hr) and drugs. The dogs were paralyzed with pancuronium bromide (0.1 mg/kg) and ventilated at a frequency of 150 cycles/min and percent inspiratory time of 30%.

Airway pressure was controlled by setting drive pressure to achieve an airway pressure excursion (Δ P) of 2, 4, or 6 cm H_2O in combination with either an independently set PEEP level (0, 5, 10, or 15 cm H_2O) or an independently set mean airway pressure (6 or 10 cm H_2O). PEEP or mean airway pressure was controlled by adjusting the expiratory resistance of the pop-off valve on the breathing circuit.

Summary data are expressed as mean +/- standard deviation (SD). Two-tailed t-tests with Bonferroni's correction (Zar, 1974) were used to determine statistical significance at $p<0.05$ for comparisons of ventilator setting groups.

Results

The effect of changes in airway pressures on arterial blood gas tensions are summarized in table 3 and figures 11 through 14. There was a statistically significant downward trend in $PaCO_2$ with increasing airway pressure excursions (ΔP) as shown in figure 11. When a mean airway pressure (AWP) of 6 cm H_2O was compared to a mean airway pressure of 10 cm H_2O at each airway pressure excursion, there was no significant change in $PaCO_2$. There was no statistical difference in arterial oxygen tension (PaO_2) between either different mean airway pressures at the same ΔP or between different ΔP's at the same mean airway pressure. This is shown in figure 12.

TABLE 3

Arterial Blood Gas Tensions at Varying Combinations of PEEP and Airway Pressure Excursions (ΔP)

PEEP (cm H_2O)	ΔP (cm H_2O)	$PaCO_2$(mean $\pm$ S.D.) (torr)	PaO_2(mean $\pm$ S.D.) (torr)
0	2	52 $\pm$ 16	150 $\pm$ 34
0	4	35 $\pm$ 7	207 $\pm$ 13*
0	6	28 $\pm$ 9*	212 $\pm$ 20*
5	2	55 $\pm$ 16	173 $\pm$ 33
5	4	35 $\pm$ 19	207 $\pm$ 10
5	6	27 $\pm$ 18*	219 $\pm$ 19*
10	2	53 $\pm$ 11	184 $\pm$ 22@
10	4	34 $\pm$ 7*	206 $\pm$ 17
10	6	29 $\pm$ 9*	219 $\pm$ 11*
15	2	75 $\pm$ 25	127 $\pm$ 33@
15	4	47 $\pm$ 13	197 $\pm$ 19*
15	6	39 $\pm$ 16*	206 $\pm$ 21*

* - Statistically different than ΔP = 2 cm H_2O at same PEEP.
@ - Statistically different at ΔP = 2 cm H_2O between 10 cm H_2O and 15 cm H_2O PEEP.

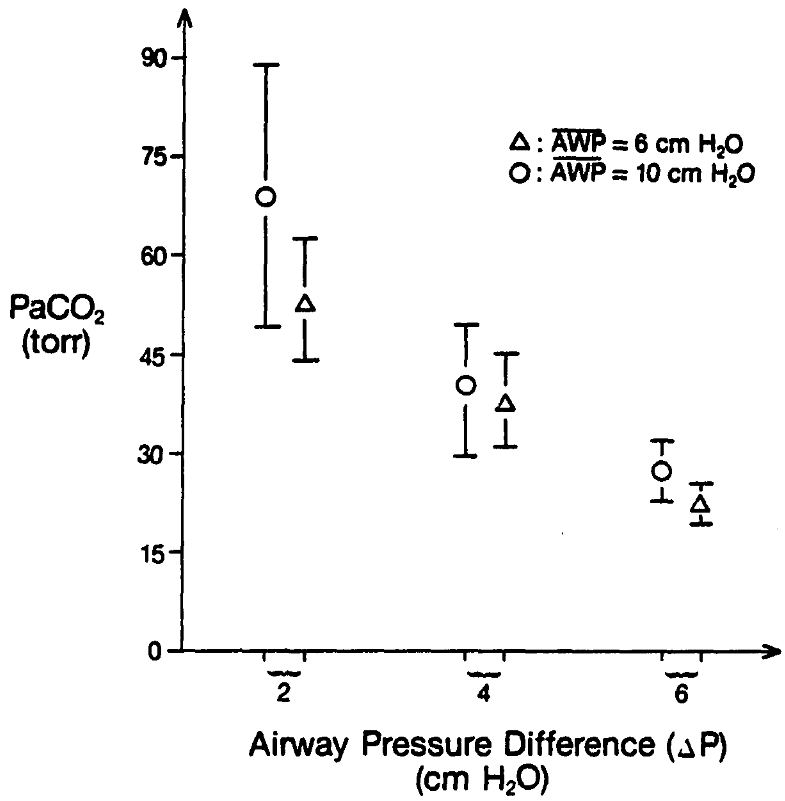


FIGURE 11: ARTERIAL CARBON DIOXIDE TENSION
vs
AIRWAY PRESSURE EXCURSION

Arterial carbon dioxide tension is shown at different airway pressure excursions (Airway Pressure Difference between peak and end-expiratory pressures) and different mean airway pressures ($\overline{AWP}$).

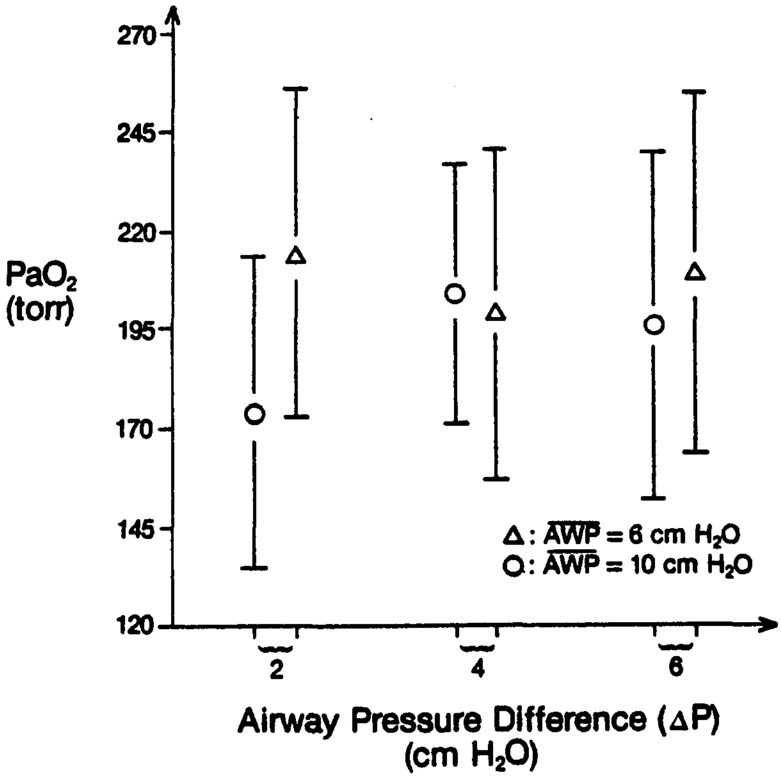
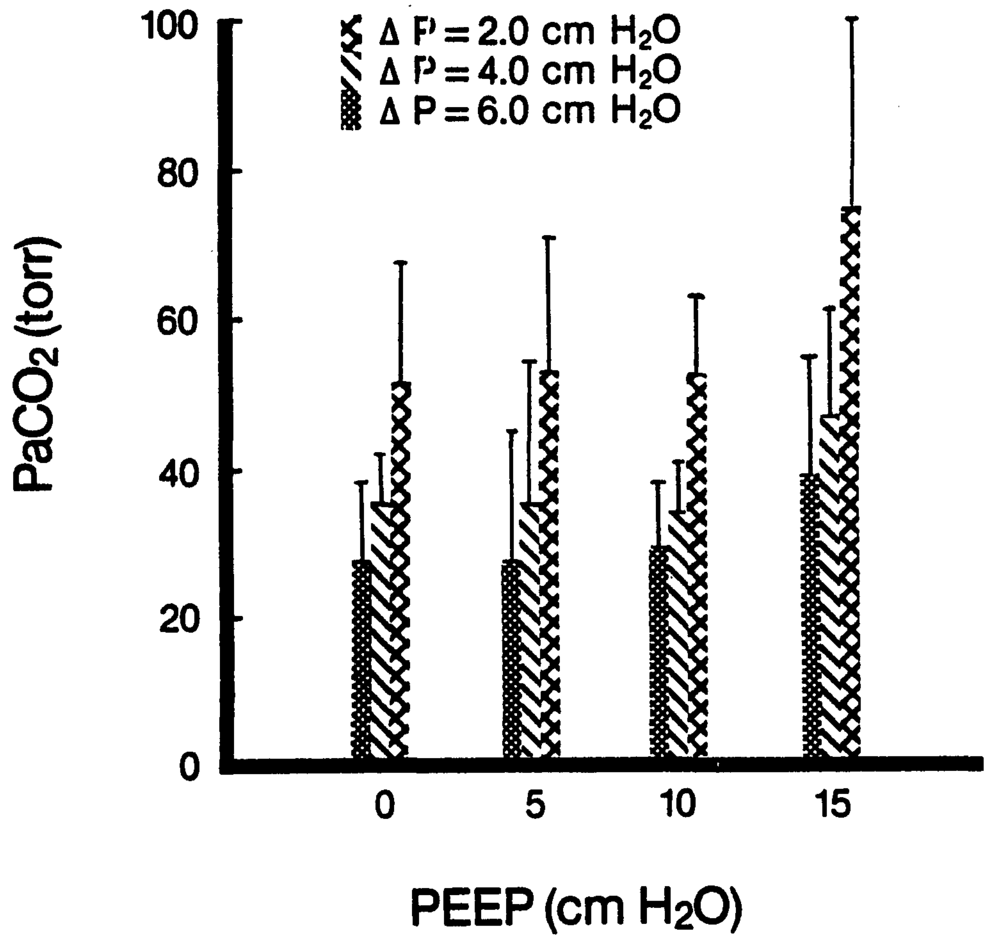


FIGURE 12: ARTERIAL OXYGEN TENSION
vs
AIRWAY PRESSURE EXCURSION

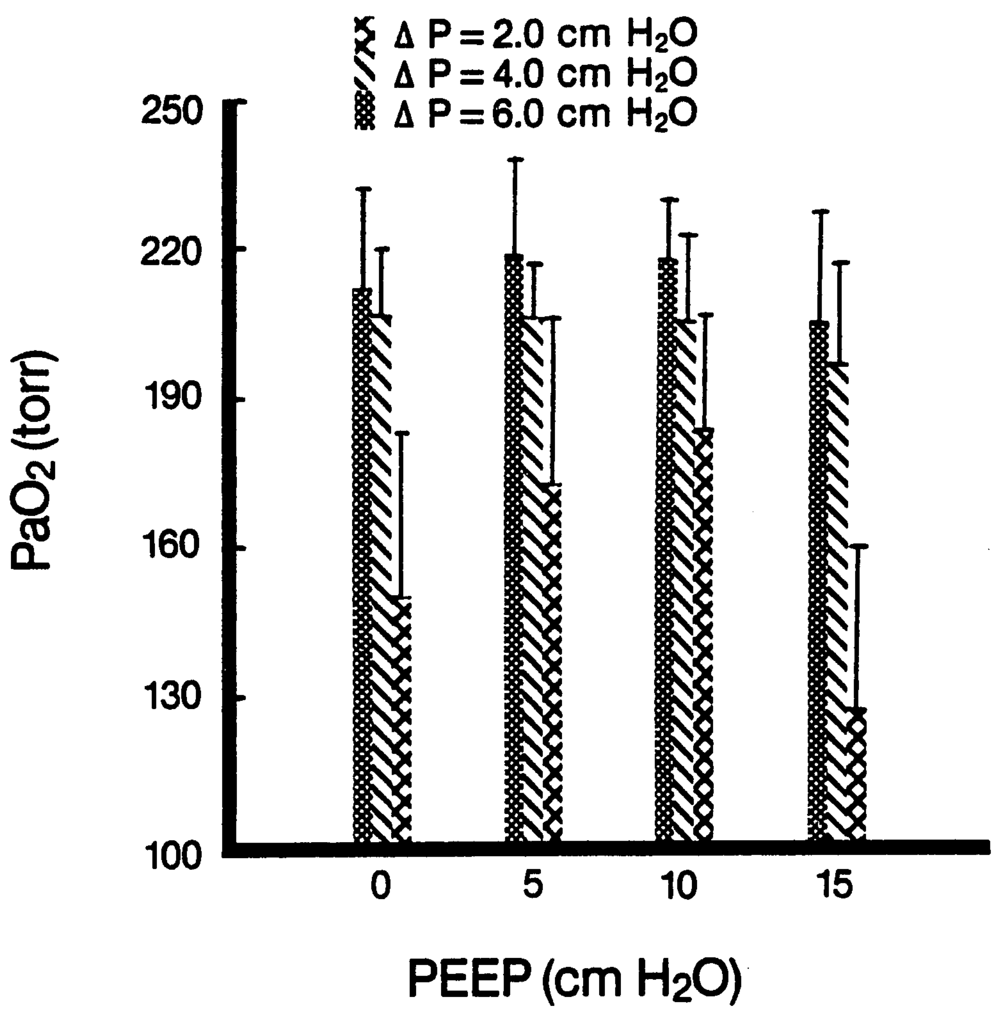
Arterial oxygent tension is shown at different airway pressure excursions (Airway Pressure Difference between peak and end-expiratory pressures) and different mean airway pressures ($\overline{AWP}$)



Note: $f = 150\ \text{min}^{-1}$ I% = 30 $FiO_2 = .40$

FIGURE 13: ARTERIAL CARBON DIOXIDE TENSION
vs
POSITIVE END EXPIRATORY PRESSURE

Arterial carbon dioxide is shown at combinations of PEEP (positive-end-expiratory pressure) and airway pressure excursion (ΔP, difference between peak and end-expiratory pressure).



Note: $f = 150\ \text{min}^{-1}$ I% = 30 $FiO_2 = .40$

FIGURE 14: ARTERIAL OXYGEN TENSION
vs
POSITIVE END EXPIRATORY PRESSURE

Arterial oxygen tension is shown at combinations of PEEP (positive-end-expiratory pressure) and airway pressure excursion (ΔP, difference between peak and end-expiratory pressure).

The effects of varying levels of PEEP at airway pressure excursions (ΔP) of 2, 4, and 6 cm H_2O are shown in figure 13, depicting $PaCO_2$ vs PEEP; figure 14, depicting PaO_2 vs PEEP; and in table 3. Increasing ΔP results in a lower $PaCO_2$. Increasing PEEP does not significantly affect $PaCO_2$. As shown in figure 14, the PaO_2 does not change significantly with varying PEEP at airway pressure excursions of 4 or 6 cm H_2O. However, at ΔP = 2 cm H_2O, there is a slight increase in oxygenation with increasing levels of PEEP up to 10 cm H_2O. At 15 cm H_2O PEEP, there is a statistically significant drop in oxygenation.

A statistically significant change in PaO_2 or $PaCO_2$, or both, was associated with increasing airway pressure excursion from 2 to 4 cm H_2O at some PEEP levels. This change was significant at all PEEP levels for increases of airway pressure excursion from 2 to 6 cm H_2O. There were no statistically significant differences between ΔP's of 4 and 6 cm H_2O for either oxygen or carbon dioxide at any PEEP level.

Discussion

In the study described in chapter 2, the effect of peak airway pressure during HFJV was correlated with arterial carbon dioxide tensions at ventilator frequency of 150 cycles/min and 30% inspiratory time.

That study showed a relationship between increasing peak airway pressure and lower arterial carbon dioxide tensions. These results were confirmed by comparing washout times in a mechanical analog lung model in chapter 3. However, PEEP, AWP, and airway pressure excursions were not independently controlled in these previous studies. Analysis of data from these studies suggested a correlation between gas exchange efficiency and airway pressure excursion. The study described here confirms the strength of that relationship and separates the effects of mean or PEEP from the effects of airway pressure excursion in healthy lungs.

The observation that increasing airway pressure excursion correlates with more effective ventilation and oxygenation may be explained by the relationsip between airway pressure and lung volume. Because of the elastic properties of the lung and chest, a larger airway pressure excursion (ΔP) should be associated with a larger tidal volume.

Unfortunately, tidal volumes are not easily measured during HFJV. Although an attempt to collect the total exhaled gas over a measured time period and then divide by the number of ventilatory cycles has been used to estimate the tidal volume, this technique ignores the varying effects of entrainment on gas exchange volume with HFJV (Carlon et al., 1983; Colgan et al., 1983). Variables which could potentially affect entrainment include the location of the jet within the airway, the relative amounts of resistance proximal and distal to the jet, the flow pulse characteristics of the jet itself, the velocity of gas escaping the jet, the size of the jet cannula, and the duration of inspiration. An accurate measurement of tidal volume should include both the amount delivered by the jet and the amount of entrainment, if any. Additionally, these two measurements must be synchronized to assure that the entrainment gas has not stalled and reversed its flow while the jet is still delivering to the airway. This stalling phenomena can be shown to occur with both the Hi-Lo jet tubes and the proximal injectors at long inspiratory times (Unpublished observations of the author). It is difficult to predict the amount of stalling because of the same multitude of variables affecting entrainment efficiency.

Because of difficulty in measuring tidal volume during HFJV, an alternate variable is needed which will closely correlate with efficiency of alveolar gas exchange. The results of this study suggest that airway pressure excursion could be used as that monitoring and control variable.

The effects of mean airway pressure and PEEP are not so clear. Normally, better oxygenation is expected to correlate with higher mean airway pressure or PEEP. However, this study was performed in animals with healthy lungs where maximal recruitment of the lung may have occurred at relatively low PEEP or mean pressures. Further increases in airway pressure may have over distended already open gas exchange units increasing anatomic dead space (Sladen et al., 1983; Dueck et al., 1977; Simon et al., 1982)

Further information would be necessary to determine what airway pressure excursion would be appropriate for the initial trial of HFJV in a patient. However, this study indicates that this is an important variable to aid in contolling gas exchange efficiency.

SUMMARY

Three investigations have been presented in this thesis. They have the common goal of attempting to identify what variables contribute to the effectiveness of gas exchange with High Frequency Jet Ventilation. In the first study, arterial carbon dioxide tension in animals was used as the indicator of ventilatory efficiency. In the second study, nitrogen washout time in a mechanical analog model of the lung was used as a measure of gas exchange efficiency. In both of these studies, the operational variables of frequency, percent inspiratory time, and peak airway pressure were studied as independent variables.

The results of these first two studies indicated that gas exchange might be better correlated to another operational variable, airway pressure excursion, so a third study was designed. This third investigation attempted to separate the effects of peak, mean, and end-expiratory pressure from those of airway pressure excursion. As in the first study, the indicator of gas exchange efficiency was arterial carbon dioxide tension. However, since oxygenation is also an important part of respiration, the effects of

airway pressures on arterial oxygen tensions were also investigated.

In whole, this series of experiments indicates the importance of airway pressure excursion as a monitoring and control variable in HFJV. Its relationship to CO_2 removal in the lung appears to be much stronger than any other easily controlled or measured HFJV variable. In addition, it seems to offer an analog from which to infer the amount of volume exchange with each ventilator cycle. This is important in attempting to establish the proper mechanical balance between the HFJV system and the lung.

Of secondary interest are the effects of various operational variables on dispersion coeffecients in the mechanical analog model. The high coeffecients present with large airway pressure excursions seem to indicate that there is substantial convective transport into at least the middle region of the model. This suggests that a rapid convective washout of the central airways may contribute to the effectiveness of HFJV in man.

The model also served to somewhat clarify the frequency component of HFJV effectiveness. Large pressure excursions at lower rates did not provide the same washout rapidity as the same excursions at higher rates. However, when rates above 300 were used, the mechanical time constant of the tube could no longer respond fast enough to allow large pressure excursions and effectiveness was lost. This seems consistent with the data on frequency versus $PaCO_2$ in animals.

For future investigations with HFJV, the effects of the operational variables have now been sufficiently defined to allow an educated guess about initial settings and which direction to move them to obtain a desired result. However, there are still large gaps in understanding about mechanisms of intrapulmonary gas transport with HFJV. Until these mechanisms are better understood, it will be difficult to fully exploit the advantages of high-rate, low-volume ventilation.

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