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Computational fluid dynamic modelling of the effect of ventilation mode and tracheal tube position on air flow in the large airways

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Running title: Modelling of air flow through tracheal tubes

Summary

We have used computational fluid dynamic modelling to study the effects of tracheal tube size and position on regional gas flow in the large airways. Using a three-dimensional mathematical model, we simulated flow with and without a tracheal tube, replicating both physiological and artificial breathing. Ventilation through a tracheal tube increased proportional flow to the left lung from 39.5% with no tube to 43.1-47.2%, depending on tube position. Ventilation mode and tube distance from the carina had no effect on flow. Lateral displacement and deflection of the tube increased ventilation to the ipsilateral lung, e.g. when deflected 10° to the left of centre, flow to the left lung increased from 43.8% to 53.7%. Because of the small diameter of a tracheal tube relative to the trachea, gas exits a tube at high velocity such that regional ventilation may be affected by changes in the position and angle of the tube.

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Introduction

When using a tracheal tube (TT), the incidence of endobronchial intubation varies from 5% in intensive care to 28% at cardiac arrests, and though this complication may be regarded as innocuous, it is in fact associated with many adverse events [1]. Clinical detection of endobronchial intubation is difficult. Most of the tests used depend on observation, palpation or auscultation of the chest, but combinations of multiple tests are required to achieve reasonable sensitivity and specificity. A case report of asynchronous ventilation of the right and left lungs during general anaesthesia (GA) with a TT suggested that a Coanda effect can occur in the airway, resulting in preferential ventilation of one lung [2]. This suggestion that unequal ventilation of the lungs may occur even when the TT is above the carina raises interesting questions regarding regional ventilation during GA, and calls into question the use of any test that relies on air movement into the lungs to detect endobronchial intubation.

Measurement of flow to different regions of lung is difficult in-vivo. Functional magnetic resonance imaging can generate detailed images of regional ventilation [3], but cost and ethical constraints prevent this being easy to perform in an anaesthetised and intubated patient. A technique for automatically recording breath sounds from an array of 36 acoustic sensors under the patient, vibration response imaging [4], has been used during anaesthesia, but resolution is low, the technique being just able to differentiate between one- and two-lung ventilation. More recently, a method of measuring regional ventilation using electrical impedance tomography has been developed which has better spatial resolution, producing images with lung regions of 2-5 cm in size [5]. Both techniques have been used in the clinical setting to detect endobronchial intubation [4, 6], and electrical impedance tomography is now being used to guide ventilation in lung-injured patients [7]. The low spatial and temporal resolutions of these methods make them inadequate for detailed study of the influence of TT position on air flow. However, more significant barriers to studying this problem are the practical and ethical challenges of placing and using a TT in different positions, particularly because of the risks of endobronchial intubation. To circumvent all these challenges we have used computational fluid dynamic modelling of air flow to investigate regional lung ventilation with various TT positions and ventilation modes.

Methods

A three-dimensional mathematical model of a human airway was provided by ANSYS UK (Milton Park, Abingdon, Oxon). The model was originally produced by mesh generation algorithms that convert 3-D medical images into a geometric model [8], a technique that has been subsequently developed into 'virtual bronchoscopy' [9]. A geometric model consists of a mesh, that is, a large but finite number of polyhedra that represent the boundary and interior of the lung geometry, as illustrated in Fig. 1. The model used for this study represents the airway from the start of the trachea to 21 segmental bronchi.

The commercial computational fluid dynamic package ANSYS CFX was used to compute the flow through the model using a variety of tube positions and flow patterns. Computational fluid dynamic modelling solves the mathematical equations describing fluid flow, based on conservation of mass, momentum and energy within each of the finite polyhedra of the mesh. In order to model positioning of a TT at different distances from the carina, it was necessary to modify the original geometry of the model supplied by ANSYS UK. Such modifications were limited by the fact that the software used to generate the original geometry is no longer available. The objectives were achieved by using the currently available ANSYS ICEM mesh generation software to truncate the geometry at desired distances from the carina.

Flow of air entering the airway from the TT was then approximated by a fixed circular entry region located at the central position (see below) and as required for different TT positions. Due to this approximation, we were not able to take into account effects due to an angled bevel or a Murphy's eye at the TT tip. Inlet velocity values were computed in order to achieve the desired volume flow rates as described below. At a fixed volume flow rate, inlet velocity is inversely proportional to the inlet area. The cross-sectional area of the trachea in the model is 6.40 cm^2, while the cross sectional areas of 8 mm and 7 mm internal diameter TTs are 0.50 cm^2 and 0.38 cm^2. Hence, in order to achieve the same volume flow rates, the air velocities exiting a 8 mm and 7 mm TT are 13- and 17-times greater respectively, compared with the trachea without a TT. Flow rates to the five main lobar bronchi were calculated and presented as a percentage of the total flow.

Three ventilation modes were initially studied: spontaneous respiration with no TT in place, and the inspiratory phase of artificial ventilation via a TT with either constant-flow volume-controlled ventilation (VCV) or pressure-controlled ventilation (PCV). The same respiratory parameters were used for all three: FiO_2 0.21 in nitrogen, respiratory rate of 12 breaths per minute, minute ventilation of 6 $l.min^{-1}$, and inspiration:expiration ratio of 1:2 with the expiratory phase equally divided between expiration and an expiratory pause. These parameters produce a tidal volume of 500 ml and an inspiratory time of 1.67 s. For spontaneous respiration, the inspiratory pattern was set to follow a sinusoidal shape which is similar to that seen in awake subjects [10]. The peak inspiratory flow with these settings was 28 $l.min^{-1}$. For VCV, a constant inspiratory flow of 18 $l.min^{-1}$ generated the required tidal volume. For PCV, the inspiratory flow pattern was assumed to be an exponential decline from the initial peak flow, with a time constant (tau) of 0.5 s based on normal values for a supine anaesthetised paralysed patient (total resistance 1 $kPa.l^{-1}.s^{-1}$ × total dynamic compliance 0.5 $l.kPa^{-1}$) [11]. With these parameters, the initial inspiratory flow rate was 60 $l.min^{-1}$.

Flow patterns were first modelled with no TT in situ, with the spontaneous respiration flow pattern and with a constant flow of 30 $l.min^{-1}$ (based on the approximate peak inspiratory flow). Volume-controlled ventilation and PCV patterns were then modelled with the tip of a 8 mm TT in the centre of the trachea and 2 cm proximal to the carina. As the inlet cross section of the trachea is not a perfect circle, we calculated the centre as the geometric centroid of the inlet cross section.

Because we found no difference in flow distribution to different lung regions between the sinusoidal spontaneous respiration pattern and steady inflow, for the comparison of tracheal tube positions only steady inflow at 30 $l.min^{-1}$ was used. This approach meant that a shorter number of simulations were required. It was not possible to obtain perfectly steady state outflows, even when using a constant flow at the inlet, due to small transient instabilities in the solution. The total flow rates out of the five main outlet branches were monitored as a function of iteration number, and the solution was assumed to have achieved a quasi-steady state when these oscillated with small amplitude about fixed mean values. This is shown in Fig. 2, which indicates that an initial transient settles quickly to a quasi-steady state. Average values of outlet flow rates were computed from the variable values, starting from time step number 300 which was sufficiently large for quasi-steady outlet flow rates to be achieved.

We studied four different factors affecting flow from the TT:

1. TT size of either 8 or 7 mm internal diameter.

2. Longitudinal position of the tracheal tube: aligned along the central axis of the trachea at 0.5 1, 2 and 4 cm proximal to the carina. The central position was defined at the level of the carina.

3. Lateral displacement from the central axis of the 8 mm TT to right or left by 5 mm or 7 mm.

4. Deflection of the TT from the central axis by 10° to left, right, anterior and posterior. This was modelled only for the 8 mm TT in the normal position 2 cm from the carina in the centre of the trachea. The choice of 10° was based on an estimate of the angle formed by the natural curvature of a TT with its cuff inflated inside the barrel of a 20 ml syringe which is approximately the same size as the trachea.

Results

Ventilation through a TT increased flow to the left lung compared with no TT present, mostly as a result of increased proportional flow to the left lower lobe, and this effect appeared to be greater with a smaller TT (Table 1). Artificial ventilation mode and distance from the carina of a centrally placed TT had little effect on proportional flows to each of the five lung lobes (Table 1). Deflection of the TT away from the midline by only 10° affected proportional flow to each lung, particularly when deflected to the left or right rather than anterior or posterior (Table 1). Lateral displacement of the TT increased proportional flow to the ipsilateral lung, also mostly due to increased flow to the respective lower lobe. Increasing the displacement from the midline from 5 mm (Fig. 3) to 7 mm (Fig. 4) further increased the flow disparity between the two lungs.

Discussion

Our computational fluid dynamic model has provided novel data on the pattern of airflow emerging from a TT, including being able to observe the effect of different inspiratory airflow patterns and various tube sizes and positions. The model has limitations that prevent its findings from being applied to in-vivo situations:

1. It is representative of only a single human airway, and the size and geometry of the airway in the region of the carina is known to be variable. In the individual on whom the model is based, the right lung architecture is slightly atypical with a large lower lobe bronchus relative to the middle and upper lobes (see Fig. 1).

2. We have used assumed patterns for ventilatory flow rates to allow them to be mathematically generated. Actual flow patterns in spontaneously breathing and artificially ventilated patients will be variable, but the patterns and flow rates chosen for our study represent a typical patient receiving standard care.

3. The tip of the tube used in our model is flat, i.e. has no bevel and no Murphy's eye. It is possible that the tube tip shape might alter the flow pattern of the gas emerging from it, but considering the high velocity of gas exiting the tube we think this is unlikely to be a significant factor.

4. We have assumed that the TT is straight for the last few centimetres before the gas exits the tube and therefore that any secondary cross-flow currents induced as gas flows through the TT bend in the oropharynx will have decayed sufficiently quickly to be of minor significance by the time the gas exits the tube.

5. Our model does not take into account differences in regional lung compliance that may affect flow distribution to different lung regions. Regional distribution of

ventilation is usually greater in dependent lung regions and influenced by body position, breath size, inspiratory flow rate and lung pathology [11]. However, in a supine and healthy patient breathing normal tidal volumes, variation in regional ventilation is small [12], and influenced more by gravity within each lobe rather than by relative ventilation to the five lung lobes considered in this study. Thus, we believe our results are applicable to healthy patients under general anaesthesia, but are unlikely to be representative of regional ventilation in patients with lung diseases which may increase ventilation heterogeneity at a lobar level, such as acute lung injury or emphysema.

Validation of our computational fluid dynamic model in-vivo is not currently possible. As already described, currently available techniques for assessing regional lung ventilation in a patient whose trachea is intubated are limited, particularly at the spatial resolution needed to determine flow to individual lung lobes. Furthermore, controlling or ascertaining the position of a TT within the trachea is difficult - flexible bronchoscopy may allow the distance from the carina to be measured, but fluoroscopy in two planes or computerised tomography would be needed to determine its 3D-position.

The lack of effect of ventilation mode was reassuring. Furthermore, Fig. 5 shows that, despite the different flow patterns, there was no obvious effect of time on the relative flow to different lobes. After less than about 50ms, flow to the five lobes is in an established pattern that remains the same until the end of expiration. Physiological differences between these modes in-vivo are small, with no firm agreement in the literature regarding which is the preferred technique in healthy lungs. In injured lungs, pressure-controlled mechanical ventilation of the lungs is preferable, but mostly as a way of limiting inspiratory pressures to reduce pulmonary barotrauma, rather than because of any known physiological benefit. Our data has not been able to contribute to the volume- vs. pressure-controlled mechanical ventilation debate.

Our data confirms what would be intuitively expected of flows through a TT. Passing the same gas flow through an 8 mm TT compared with the trachea which has a diameter of >2 cm inevitably results in a large increase in gas velocity. Compared with when no TT is in place, this results in a high-velocity jet of gas exiting the end of the TT. It is unsurprising that the position and direction of this jet can influence where the gas is distributed. As the gas hits the carina or airway wall, the laminar flow pattern is immediately disturbed, and gas swirls around in the large airways before passing further down into the airway branches (Fig. 1). Despite this abnormal flow pattern, the ultimate distribution of gas to the five lobes remains similar to that seen with no TT in place, though there is a small but consistent increase in ventilation of the left lung in all situations where a TT is present, which is possibly more pronounced with a smaller TT. This observation holds true irrespective of how far a centrally-placed TT is from the carina. The same cannot be said for TT positions away from the centre, with flow distribution being preferentially distributed to the ipsilateral lung. For tubes at the left side of the airway, closer proximity to the carina seems to result in a greater degree of unilateral ventilation, particularly with the 7 mm displacement. The flow pattern in this situation is chaotic, with the gas which eventually ends up on the right side taking a circuitous route via the left main bronchus and trachea. Similar flow patterns are seen with angular deflection of the TT. As may be expected, anterior or posterior deflection results in similar distribution to the left and right lungs as via a TT in other situations. However, angulation to either side causes increased gas flow to the ipsilateral lung with chaotic flow patterns similar to those seen with TT displacement. In summary, the high velocity jet of gas leaving the TT means that the direction in which the TT is pointing in the airway becomes crucial, and any position other than in the centre of, and aligned longitudinally with, the trachea will cause gas distribution between right and left lungs to be abnormal.

Our results indicate that the position of the TT in the trachea may affect regional distribution of ventilation. Pulmonary ventilation-perfusion relationships are known to be abnormal during general anaesthesia, with an increase in areas with both high and low ratios contributing to impaired pulmonary exchange of both oxygen and carbon dioxide [11]. Our results offer a new potential mechanism for the abnormal ventilation distribution seen with artificial ventilation. Furthermore, the influence of tube position on ventilation-perfusion relationships may also, at least in part, explain why the effects of mechanical ventilation of the lungs on gas exchange in patients is so variable [13]. In patients with more abnormal ventilation-perfusion relationships, such as those with lung disease or obesity, the small variations in regional ventilation suggested by our study could be crucial to maintaining acceptable gas exchange.

Our data also has implications for TT design. The larger the TT diameter, the less pronounced will be the increased velocity of gas flow within, so larger TTs should provide more uniform ventilation, but their use has implications for ease of tracheal intubation and trauma to the airway. In theory, the best way to reduce gas velocity in a tube is a gradual (rather than step) increase in tube diameter such as an expanding funnel shape like the bell on a trumpet. This is unlikely to ever be feasible in practice for a TT as this would make inserting and removing the tube impossible without a complex mechanical system for changing the TT shape once it is beyond the vocal cords.

In clinical anaesthesia, the only aspect of TT position that has received attention is its distance from the carina. In 1963, Bamforth wrote that a TT should not be passed more than 3 cm beyond the vocal cords to avoid endobronchial intubation [14]. Subsequent work has recommended that the TT tip should be 4 cm from the carina, but also noted that the tips of 58% of TTs placed for routine surgery were actually less than 3 cm from the carina [15]. Another study, this time in critically ill patients requiring mechanical ventilation of the lungs, showed a mean (SD) distance from the TT tip to the carina of 3.3 (1.6) cm [16]. It is also known that TT position is not constant, with neck flexion and extension moving the tip by 2 cm in either direction [17]. Thus, it is likely that the TT positions used in our study all occur commonly during clinical practice in both anaesthesia and critical care. It is therefore mildly reassuring that the distance of the tube from the carina appears have only a small influence on regional ventilation, provided it is still central within the trachea and not in either main bronchus. It is, however, a concern that no clinical studies have ever considered whether the TT tip is centrally placed in the trachea, which we have found to be an important factor. When inflated outside the body, many TT tube cuffs inflate eccentrically, so it is likely that the TT tube will often be pushed against one side of the trachea, in a random fashion, and that this may adversely affect regional ventilation.

Our model, despite having limitations, has allowed us to obtain novel and potentially clinically important information on how regional ventilation is influenced by the TT position within the trachea. Further studies of TT behaviour using computational fluid dynamics are required to investigate ways in which TTs could be made to behave more physiologically, and so hopefully improve a piece of anaesthetic equipment that has remained essentially unchanged and rarely investigated for almost 100 years.

Acknowledgements

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Tables

Table 1 Effect of inspiratory flow pattern and tracheal tube (TT) position on distribution of gas to the left and right lungs and their lobes, for sinusoidal flow (simulating spontaneous breathing), volume-controlled mechanical ventilation (VCV) and pressure-controlled mechanical ventilation (PCV). Values are proportion of total flow.

		Left lung			Right lung		
	Total	**Upper**	**Lower**	**Total**	**Upper**	**Middle**	**Lower**
No TT: Constant flow	39.5%	11.9%	27.6%	60.4%	3.7%	7.5%	49.2%
Sinusoidal flow	39.6%	12.4%	27.6%	60.3%	3.8%	7.6%	48.9%
Ventilation mode: VCV	43.2%	12.3%	30.9%	56.8%	3.6%	7.3%	45.9%
PCV	43.1%	12.3%	30.8%	56.8%	3.7%	6.9%	46.2%
TT size * 8.0 mm	43.8%	12.3%	31.5%	56.2%	3.9%	7.3%	45.0%
7.0 mm	47.2%	13.0%	34.2%	52.7%	3.7%	6.7%	42.3%
Distance from carina:							
4.0 cm	43.8%	2.3%	31.5%	56.2%	3.9%	7.3%	45.0%
2.0 cm	43.8%	12.3%	9.5%	56.2%	3.9%	7.3%	45.0%
1.0 cm	43.3%	13.3%	30.0%	56.7%	3.9%	7.5%	45.3%
0.5 cm	44.5%	13.0%	31.5%	55.5%	3.8%	7.2%	44.5%

Deflection 10°*	Right	35.4%	10.6%	24.8%	64.4%	4.0%	8.5%	51.9%
	Left	53.7%	14.8%	38.9%	45.5%	3.2%	6.2%	36.1%
	Anterior	44.9%	13.0%	31.9%	55.0%	3.3%	7.4%	44.3%
	Posterior	41.5%	12.3%	29.2%	58.5%	4.0%	7.4%	47.1%

*= TT tip central, 2 cm from carina.

Figures

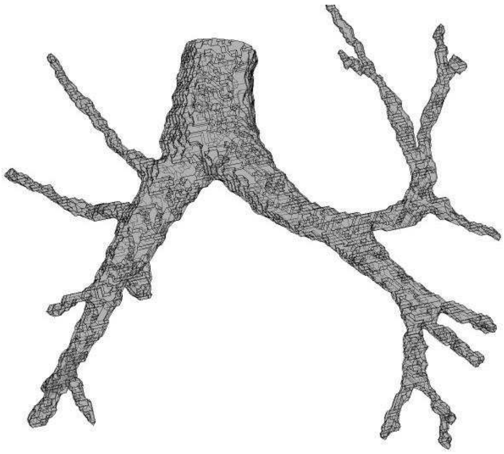


Figure 1 Computer-generated drawing of the 3D-mathematical model of the airway used for the study, showing the overall structure being made up of a large but finite number of complex polyhedra. The coloured lines are streamlines of gas flow, each colour corresponding to the lung lobe which the gas finally enters. The TT is 8 mm diameter, centrally placed within the trachea 4 cm away from the carina, and the gas flow rate is a constant 30 $l.min^{-1}$ corresponding to normal peak inspiratory flow.

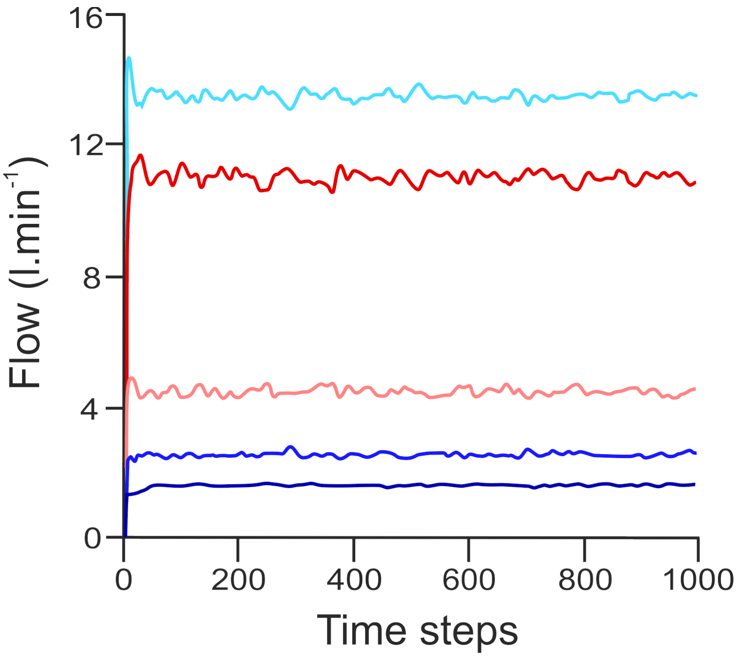


Figure 2 Flow rate of gas to five lobes of the lungs with a constant flow of 30 $l.min^{-1}$ through an 8 mm tracheal tube placed in the midline.

Left lower lobe ; left upper lobe ; right lower lobe ; right middle lobe ; right upper lobe .

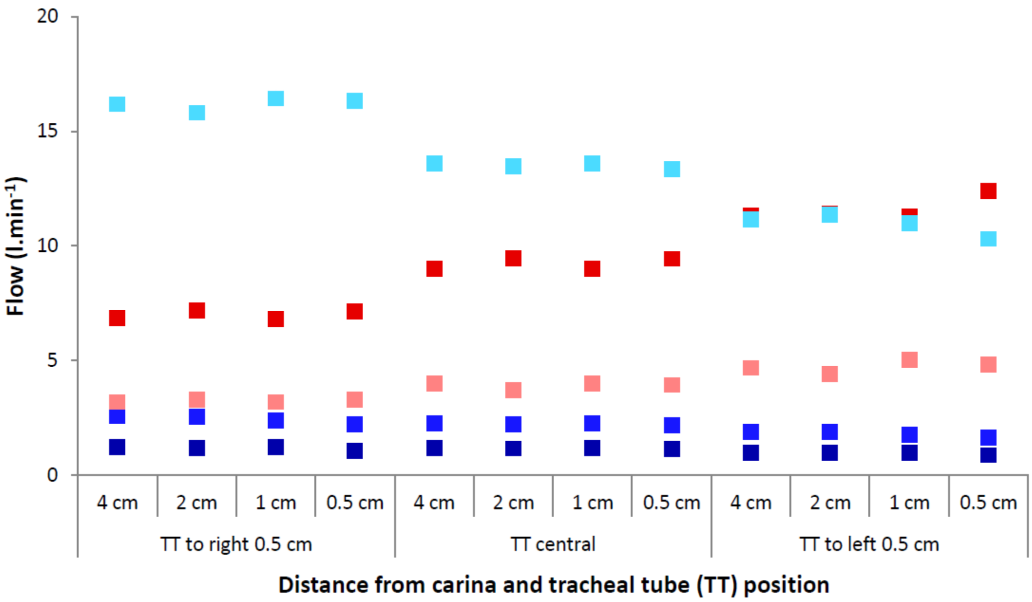


Figure 3 Flow rate of gas to five lobes of the lung with an 8 mm tracheal tube placed 5 mm away from the midline and at four distances from the carina (constant total flow of 30 $l.min^{-1}$).

Left lower lobe ; left upper lobe ; right lower lobe ; right middle lobe ;

right upper lobe .

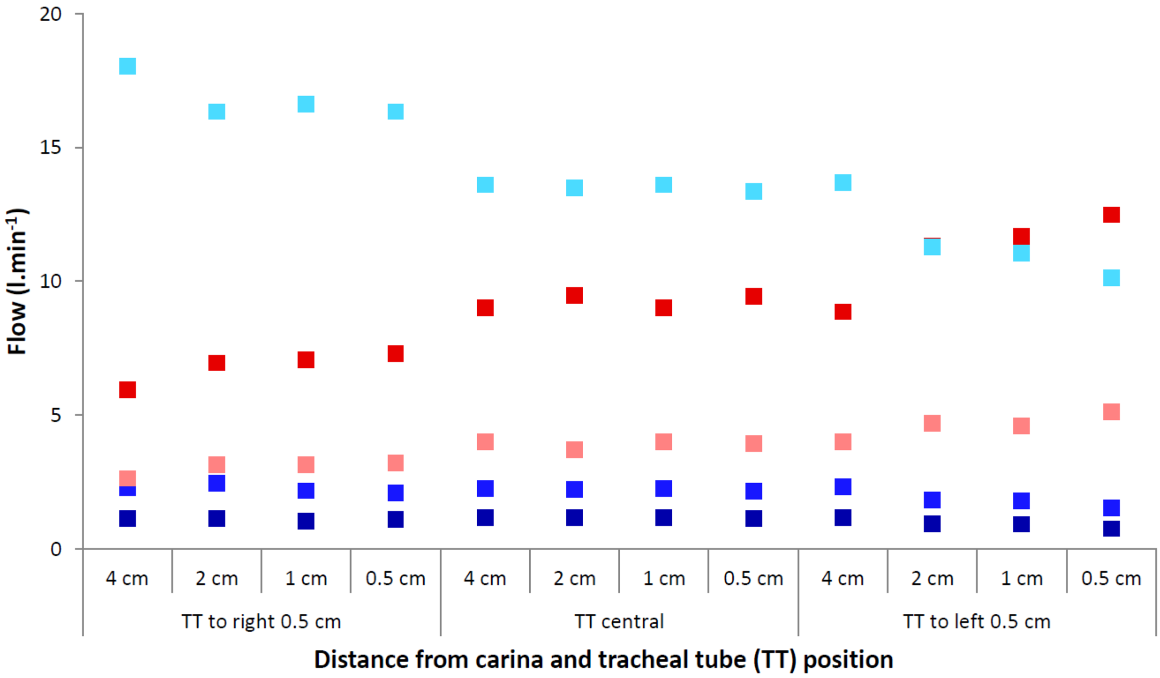
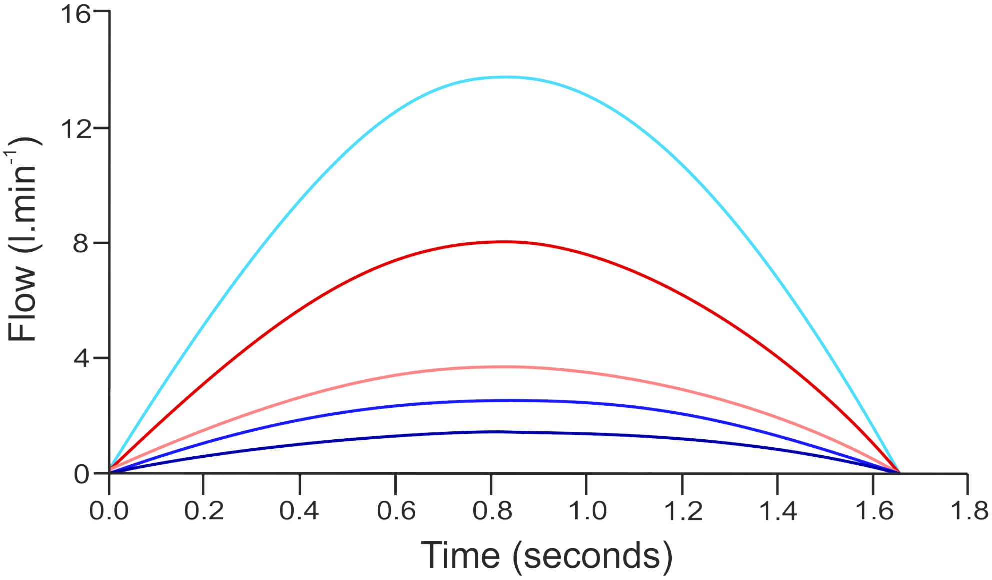


Figure 4 Flow rate of gas to five lobes of the lung with an 8 mm tracheal tube placed 7 mm away from the midline and at four distances from the carina (constant total flow of 30 $l.min^{-1}$).

Left lower lobe ; left upper lobe ; right lower lobe ; right middle lobe ;

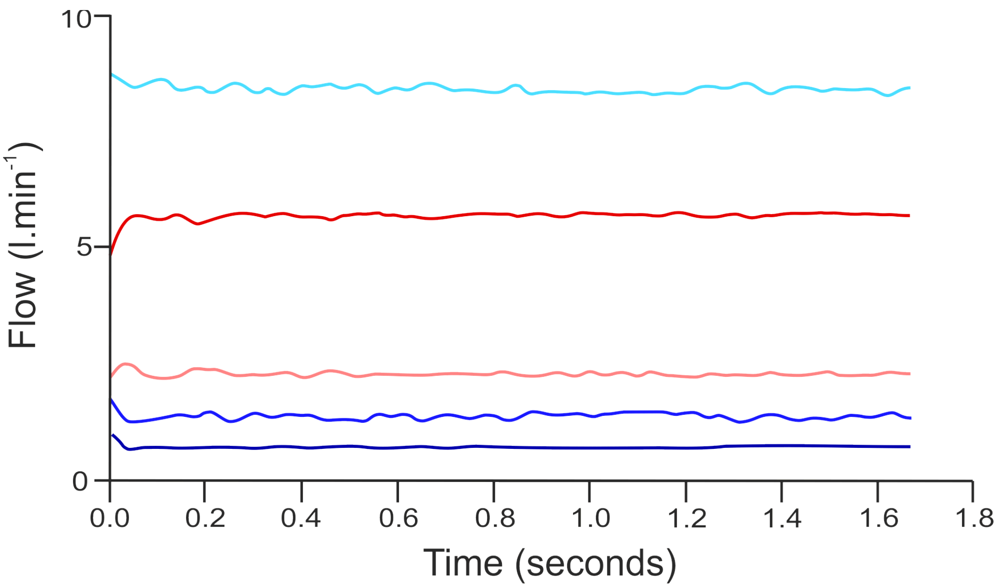
right upper lobe .

A



Flow (l.min^{-1})
16
12
8
4
0
0.0
0.2
0.4
0.6
0.8
1.0
1.2
1.4
1.6
1.8
Time (seconds)

B



Flow (l.min^{-1})
10
5
0
0.0
0.2
0.4
0.6
0.8
1.0
1.2
1.4
1.6
1.8
Time (seconds)

C

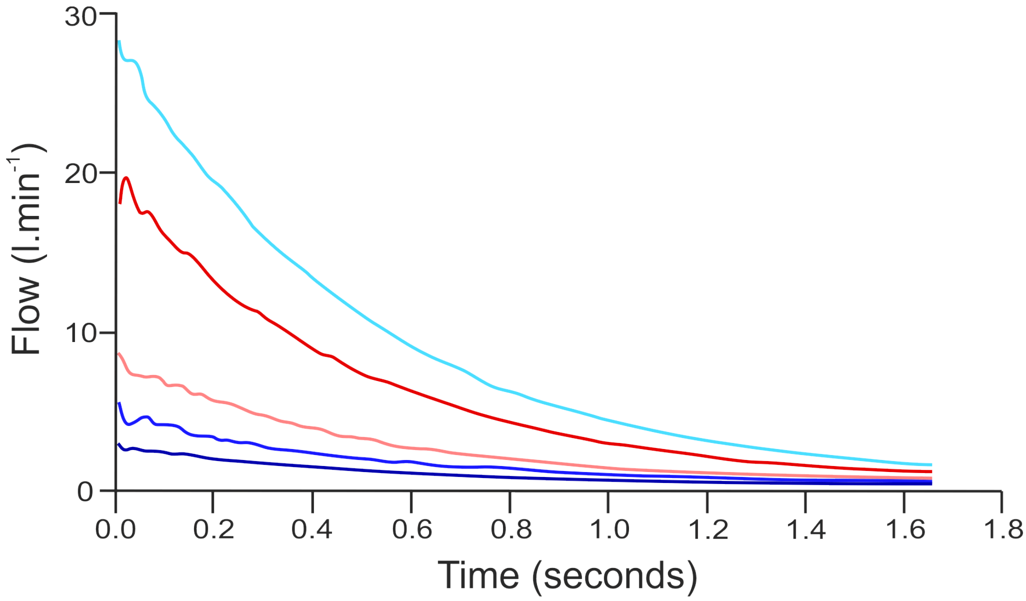


Figure 5 Flow rate to five lobes of the lung relative to time with the three different ventilation modes studied; in each case, tidal volume and total flow rate are as described in the text and the models used an 8mm TT in the centre of the trachea 2cm proximal to the carina:

Panel A. Sinusoidal (normal) ventilation

Panel B. Volume controlled (constant flow) mechanical ventilation

Panel C. Pressure controlled mechanical ventilation.

Left lower lobe ; left upper lobe ; right lower lobe ; right middle lobe ; right upper lobe .