Relationship of inspiratory and expiratory times to upper airway resistance during pulsatile needle cricothyrotomy ventilation with generic delivery circuit

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Background. Narrow-bore cricothyrotomy retains a clinical role, due to the availability of its component equipment in acute clinical environments, ease of assembly, and operator preference. However, due to infrequent use, there is a need to model this for research and teaching. We present mathematical and laboratory models.

Methods. Using electrical analogy, we mathematically modelled a generic cannula cricothyrotomy circuit, relating inspiratory and expiratory times to the upper airway resistance ($R_u$). We constructed a laboratory model to support our mathematical model. The simulated lung is a smooth-bore tube on a tilting table. The upper airway is simulated by 20 G cannulae. Inspiratory and expiratory times for the water meniscus to travel a preset distance (corresponding to tidal volume) were measured and plotted against the number of cannula.

Results. From the mathematical model, inspiratory time increases hyperbolically with decreasing $R_u$, such that there is a minimum $R_u$ beyond which most of the fresh gas flow leaks out without inflating the chest. Conversely, as $R_u$ increases, inspiratory time decreases to a plateau. Expiratory time is limited by respiratory factors at low $R_u$ and by the resistance of the transtracheal expiratory pathway at high $R_u$, producing a sigmoid-shaped expiratory curve. The experimental results seem consistent with these predictions, although direct theory–experiment mapping is problematic because of the difficulty in assigning a single value to the dynamically changing upper airway resistance.

Conclusions. We can exploit the contrasting changes in inspiratory and expiratory times with the upper airway resistance to optimize conditions for emergent cannula cricothyrotomy ventilation.

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Emergency cricothyrotomy is the final step of failed-intubation–failed-ventilation algorithms.1 2 Cricothyrotomy retains an importance out of proportion to its infrequent clinical usage, because it provides a ready subglottic airway as backup for more commonly used supraglottic airway techniques.3 Cricothyrotomy may use narrow-bore cannulae, wide-bore cannulae, or surgical airways.3–5 The appropriate methods of ventilation would appear to be jet ventilation for narrow-bore cannulae3–7 (except in complete airway obstruction)3 4 and more conventional bag and valve devices for wide-bore and surgical airways.3–5

Narrow-bore cricothyrotomy (NBC) tends to be preferred by many anaesthetists8 9 because it is less traumatic8 9 and quicker to place.1 However, studies have demonstrated problems with locating a jet ventilator, especially outside the operating theatres.10 Even when located, jet ventilators may be unfamiliar to operators. Therefore, we have a dilemma that anaesthetists generally favour using a narrow-bore device, but optimal ventilatory support is unavailable or unfamiliar.

We wished to study the situation of ventilating through a narrow-bore cannula without jet ventilation. We
developed a mathematical model, supported by a laboratory model, of NBC ventilation using equipment and devices available in most acute clinical environments. The laboratory model also functions as a teaching model, as practice is essential for the effective use of any difficult airway technique but especially the rarely performed cricothyrotomy.591 1

Methods

Cannula cricothyrotomy circuit
We based our models on a generic cannula cricothyrotomy circuit. This consists of a 14 G i.v. (or transtracheal) cannula connected to a wall oxygen source via a three-way stopcock and non-distensible green tubing (Fig. 1). The stopcock’s tap is opened to all three limbs, with the thumb occluding the expiratory port for inspiration. This system has previously been described12 – 14 and its components are available in acute clinical environments.

For our study, we used a Terumo VersatusTM; 14 G cannula. The connecting tubing is manufactured by UhsTM (reference UN880), has an internal diameter of 3 mm, and comes in lengths of 30 cm. The stopcock is manufactured by BD ConnectaTM (reference 394601).

To determine the individual resistances of the individual limbs of the stopcock for use in the mathematical analogue (Fig. 2), we put the stopcock into different configurations from that used clinically. First, the unconnected (expiratory) port was closed off, leaving the patent pathway from the fresh gas flow to the cannula. Next, the tap was rotated so that the patent pathway ran from the fresh gas flow to the atmosphere and the cannula limb closed off. The gas flow was measured by substituting the wall oxygen source with a Fabius GSTM; anaesthetic machine, which allowed us to digitally measure the fresh gas flow to three significant places. The driving pressure was measured by placing an invasive pressure sensor immediately upstream of the stopcock, with readout on a Philips Intellivue MP70TM; Anaesthesia monitor. The pressure–flow characteristics of each limb of the stopcock (with cannula attached) were measured for fresh gas flows of 3–12 litre min\(^{-1}\) of O\(_2\) 100%. Polynomial curves were then fitted to the respective data points using Excel®.

The resulting values for resistance include the resistances of the necessary connecting segment, such as the limb connecting to the fresh gas flow.

Mathematical model
The lung model consists of a capacitor to represent respiratory compliance, and linear resistors \(R_u\) and \(R_p\) to represent upper and lower airway resistances, respectively (Fig. 2). By upper airway, we mean respiratory anatomy from the point of entry of the transtracheal cannula to the atmosphere. We expect there to be failure to ventilate for a few minutes by the time cannula cricothyrotomy is instituted. The resulting lung collapse is expected to lead to low compliance and high respiratory resistance. In this paper, we took \(R_p\) to be 6.0 cm H\(_2\)O litre\(^{-1}\) s, which is the upper limit of normal.15 We set respiratory compliance to 10–20 ml cm H\(_2\)O\(^{-1}\) to accommodate laboratory limitations (see Laboratory model).

Voltages, currents, and resistances were related using Ohmic analogy. Equations were derived to relate dependent variables such as inspiratory time and expiratory time to independent variables of \(R_u\) (Appendix I). The inspiratory equations were then analytically solved to give inspiratory time—this was defined as the time needed to deliver a preset inspiratory volume of 100 ml. Expiratory time was defined as the time to empty 95% of the inspiratory volume—95% being chosen because 100% of exponential washout may be unrealistically long. The expiratory equations were analytically solved to give the gas flow through the cannula. This was then numerically integrated in Excel® at 1 ms intervals to give expiratory time. Lastly, the shapes of the inspiratory and expiratory graphs were analytically predicted.

The following assumptions were made: (i) at the limited tidal volumes generated by NBC, respiratory compliance, and resistance do not exhibit large fluctuations and may be approximated to single values for modelling purposes; and (ii) air entrainment is not significant where fresh gas flow is much lower than high-flow jet ventilation16—this is expected to be especially the case with patients who cannot be adequately ventilated supraglottically.17
Laboratory model

The mathematical predictions were compared against a bench model. Respiratory compliance was modelled using smoothbore tubing as a U-shaped manometer (Fig. 3). This rested on a tilting table in a reverse Trendelenberg position. Water was poured in until both descending and ascending limbs are half-filled. Compliance was calculated as the volume enclosed by a length of tubing (area of tubing multiplied by that length), divided by twice the vertical elevation of the tubing across that length (because in a U-shaped manometer, the depression of the fluid level on one side elevates the fluid level on the other side, creating a pressure difference across the two arms that is twice the vertical elevation of one arm). A smaller tube diameter allows greater measurement precision because there is greater gradation from a longer tubing length per unit volume. This, however, comes at the expense of less ability to accommodate large volume changes. For our tubing diameter of 19 mm, we deemed a compliance value of 10–20 ml cm H₂O⁻¹ to be most suitable. The horizontal level for the vertical elevation was determined by a fluid-level ruler.

The upper airway model was constructed from a series of capnography adaptors connected in series (Fig. 1). The adaptors were numbered 1 through 3, with 1 being the adaptor connected to the descending limb of the simulated lung and 3 the adaptor furthermost. Adaptor 3 was closed by a circuit cap. Adaptors 1 through 3 had injectable caps over their capnography ports. It is important that the diameters of the internal orifices of these capnography ports were larger than the venous cannulae inserted through the bungs—otherwise the ports become the limiting resistance. Venous cannulae (20 and 14 G) were inserted through the injectable caps of adaptors 2 and 3, simulating upper airway resistance. The transtracheal venous cannula was inserted through the bung of adaptor 1. We did not model lower airway resistance as we did not expect it to be a limiting factor in the typical obstructed airway.

One limb of the simulated lung was attached to the upper airway model, whereas the other was left open to atmosphere. The fresh gas flow was varied between 3 and 6 litre min⁻¹. Occluding the stopcock forced water down one limb and up the other. We did not measure the dynamic pressure changes in the circuit.

The inspiratory volume was defined as 100 ml. Expiratory time was measured as the time taken for the
inspiratory volume to empty from end-inspiration. Inspiratory and expiratory times for the water meniscus to travel a preset distance (corresponding to tidal volume) were measured through time-stamped video records. A range of compliances, fresh gas flows, and upper airway resistances were studied—these parameters were selected for practical reasons, namely, to keep the tidal volume within the descending limb of the simulated lung.

Results

Cannula cricothyrotomy circuit

The gas flow through the stopcock to the atmosphere is unidirectional, since the pressure in the cricothyrotomy circuit should always be positive with fresh gas flow. There was therefore no need to plot data points for negative flows and a quadratic curve could be generated. The equation fitted to this is $4.00 \times 10^{-4} I^2 + 1.37 \times 10^{-2} I c$, where $I$ is the gas flow through the cannula.

In contrast, the flow might be bidirectional across the transtracheal cannula, depending on the pressure differential between the tracheal pressure and the stopcock. Therefore, the data points for the cannula were mirrored vertically and horizontally to give a symmetrical cubic curve with an inflection point at origin. The equation fitted to this is $7.00 \times 10^{-6} I^3 + 3.00 \times 10^{-19} I^2 + 1.61 \times 10^{-1} I c$, where $I$ is the gas flow through the cannula.

The values were fed into our mathematical model. The pressures generated did not reach pressure valve release threshold.

Mathematical model

Inspiratory time is related to the inspiratory volume, $R_u$, the fresh gas flow, and respiratory characteristics [equation (2)]. The graph of inspiratory time against $R_u$ is a hyperbolic curve (Fig. 4). This means that at very high $R_u$, there is negligible air leak with almost all the fresh gas flow entering the lungs, resulting in a short inspiratory time. As $R_u$ decreases, air leak and therefore, inspiratory time increases. There then comes a point where $R_u$ is so low that the desired inspiratory volume cannot be delivered, despite prolonged inspiratory times. Changing the fresh gas flow changes the inspiratory time, and also the expiratory time against $R_u$ becomes limiting and the expiratory curves vary linearly with the $R_u$. At high $R_u$, there is little expiratory flow through the upper airways. Instead, expiration occurs primarily through the alternate expiratory pathway and the resistance of the cannula and stopcock becomes limiting. Therefore, expiratory time increases to a plateau, the value of which is determined by the alternate expiratory resistance. Now the alternate expiratory resistance depends not only on the inherent resistance but also on the fresh gas flow, which, during expiration, competes with the pulmonary gas flow for use of the alternate expiratory pathway at high $R_u$. Thus, the expiratory curves diverge at high airway pressures on the basis of the fresh gas flow.

The total respiratory time is the sum of the inspiratory and expiratory times and because of their contrasting patterns of change with upper airway resistance, the total respiratory time when graphed against upper airway resistance has the shape of a tick mark (Fig. 4), with the total respiratory time determined primarily by the inspiratory time at low $R_u$ and by the expiratory time at high $R_u$. The shortest respiratory time (and therefore quickest respiratory cycle) occurs at the point where the inspiratory and expiratory graphs intersect. This may be considered ‘optimal’ for NBC ventilation. To put the values for $R_u$ into context, the calculated resistance for the unobstructed upper airway in man is $0.5 - 2 \times 10^{-3}$ cm H$_2$O ml$^{-1}$ s$^{-1}$.18 19

Laboratory model

Table 1 shows that the inspiratory time decreases and expiratory time increases with increasing upper airway
inspiratory and expiratory times with upper airway resistance is evident. The contrasting pattern of changes of inspiratory and expiratory times with upper airway resistance is evident.

<table>
<thead>
<tr>
<th>Conditions</th>
<th>Resistance</th>
<th>( T_{in} ) (s)</th>
<th>( T_{exp} ) (s)</th>
<th>( T_{total} ) (s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cp 20 3 litre min (^{-1})</td>
<td>0xG20</td>
<td>3.75 (0.13)</td>
<td>19.68 (1.26)</td>
<td>23.67 (1.28)</td>
</tr>
<tr>
<td></td>
<td>1xG20</td>
<td>4.47 (0.16)</td>
<td>17.70 (1.22)</td>
<td>22.30 (1.30)</td>
</tr>
<tr>
<td></td>
<td>2xG20</td>
<td>5.11 (0.01)</td>
<td>13.28 (0.72)</td>
<td>18.61 (0.71)</td>
</tr>
<tr>
<td>Cp 10 3 litre min (^{-1})</td>
<td>0xG20</td>
<td>5.97 (0.24)</td>
<td>7.71 (0.40)</td>
<td>13.83 (0.55)</td>
</tr>
<tr>
<td></td>
<td>1xG20</td>
<td>8.82 (0.10)</td>
<td>5.81 (0.20)</td>
<td>14.88 (0.15)</td>
</tr>
<tr>
<td></td>
<td>2xG20</td>
<td>11.25 (0.33)</td>
<td>8.27 (0.08)</td>
<td>20.52 (0.33)</td>
</tr>
<tr>
<td>Cp 10 6 litre min (^{-1})</td>
<td>0xG20</td>
<td>2.04 (0.05)</td>
<td>7.97 (0.31)</td>
<td>10.59 (0.23)</td>
</tr>
<tr>
<td></td>
<td>1xG20</td>
<td>2.30 (0.09)</td>
<td>5.50 (0.08)</td>
<td>8.27 (0.08)</td>
</tr>
<tr>
<td></td>
<td>2xG20</td>
<td>2.77 (0.18)</td>
<td>4.58 (0.13)</td>
<td>7.66 (0.16)</td>
</tr>
</tbody>
</table>

Table 2: Quantitative comparison of predicted and observed inspiratory and expiratory times for simulated complete upper airway obstruction. Two values for predicted expiratory time are shown, one for 95% emptying and the other for 99% emptying. This demonstrates the effect of exponential wash-out.

<table>
<thead>
<tr>
<th>Compliance (litre min (^{-1}))</th>
<th>Fresh gas flow (litre min (^{-1}))</th>
<th>Inspiratory time (s)</th>
<th>Expiratory time (s)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Predicted</td>
<td>Observed</td>
<td>Predicted</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>95%, 99%</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Observed</td>
</tr>
<tr>
<td>20</td>
<td>3</td>
<td>2</td>
<td>3.75</td>
</tr>
<tr>
<td>10</td>
<td>3</td>
<td>2</td>
<td>5.97</td>
</tr>
<tr>
<td>6</td>
<td>1</td>
<td>1</td>
<td>2.04</td>
</tr>
</tbody>
</table>

Discussion

We, together with other authors, note that NBC without jet ventilation is not really suited for high-minute volume ventilation. Our cricothyrotomy circuit could only deliver low tidal volumes effectively—larger tidal volumes require higher pressures and longer inspiratory and expiratory times. This is ameliorated by the fact that cricothyrotomy bypasses the supraglottic anatomical dead space (usually 70 ml). Therefore, 100 ml infraglottic tidal volume is the equivalent of 170 ml supraglottic tidal volume. Nevertheless, NBC without jet ventilation is more suitable for short-term insufflation or low-minute volume ventilation with permissive hypercapnia.

However, oxygenation only requires low-minute volume ventilation. It is carbon dioxide removal that requires high-minute ventilation. Our main aim of emergency oxygenation could therefore be achieved with low tidal volumes, even apnoea at low barotrauma risk. Indeed, low-minute volume ventilation may be the safer initial step, since high-pressure source ventilation can generate rapid positive pressure changes distal to the transtracheal cannula from the gas injection and negative pressures proximal to the transtracheal cannula from air entrainment. The former risks barotrauma, whereas the latter risks tissue deformation and damage exacerbating the situation. Thus, NBC without jet ventilation may well suffice as an emergency first step while successor plans and devices are formulated and set up.

This paper focuses on pulsatile NBC ventilation because observing the inspiratory and expiratory times yields vital information about the upper airway resistance that will guide steps towards improving NBC ventilation (see Clinical application), setting up for insufflation, and the choice of successor airway support devices and techniques—the last two are beyond the scope of this paper.

Generic cricothyrotomy circuit

We studied a generic cannula cricothyrotomy circuit primarily because the clinical availability of its components allows model standardization for research and teaching. There are additional considerations.

The generic circuit differs from a transtracheal jet ventilation system in two respects. First, there is the possibility of an alternate gas pathway through the transtracheal cannula to the atmosphere, resulting in longer inspiratory and shorter expiratory times, and also a different expiratory time–resistance profile, compared with transtracheal jet ventilation. Secondly, the generic circuit delivers up to 12 litre min \(^{-1}\) (the maximum flow is flow-limited by the flowmeter, not pressure-limited by cannula resistance) resulting in maximal gas flows and pressures that are lower than the jet ventilator. This leads to less air entrainment effects compared with the jet ventilator. The lower maximal flows also mean that the main fresh gas port of anaesthetic machines is a sufficient replacement for the wall flowmeter in our experiment.

Dedicated narrow-bore trans-tracheal cannulae are easier to insert and more resistant to kinking compared with the i.v. cannula. However, their clinical impact is as yet unremarkable, limiting their widespread adoption and standardization. The i.v. cannula is ubiquitous and standardized, thus more suitable for teaching and research. Dedicated cannulae often permit the attachment of a bag ventilator circuit. However, manual inflations are variable in both time and force, making this a difficult scenario to model. Further, bag ventilator circuits tend to be low-pressure circuits and thus, do not provide effective ventilation through a narrow transtracheal cannula. When a tubing-stopcock generic oxygen delivery circuit is
attached, the principles derived from studying the generic cannula are applicable.

We chose to use an improvised manometer for a simulated lung, rather than using the more accurate pneumotachometer, because of its ability to visualize tidal volume. Visualization is more important than accuracy for teaching purposes but also helpful for this project because this is an analogy for clinical observation of chest wall movement during NBC ventilation. The smoothbore tubing we used could be obtained from hardware stores. Alternatively, circle tubing may be used, although corrugations introduce further (minor) measurement errors through the variability of tubing diameter and trapping of bubbles within the corrugations.

Limitations of our models
As with all models, our models simplify a real-life system to a manageable theoretical system. Perhaps, the chief simplification is representing $R_u$ as a single constant value. We know this was not realistic, as the upper airway is a soft structure and has compliance. Thus, its resistance is dynamically variable as follows: high upper airway resistance leads to high pressure, which distends the upper airway and lowers the resistance. The converse reaction occurs for low $R_u$. However, even if we should simplistically consider the upper airway to be a rigid structure, resistance remains dynamically variable, as there would be a mixture of laminar, transitional, and turbulent flows, the proportions of which (and resultant resistance) vary depending on the instantaneous gas flow and the geometry of the gas channel. As case in point, it was difficult to identify a single value for the combined resistance of the expiratory cannulae (which are fairly rigid structures) in our bench model and we were obliged to use surrogate markers such as the number of cannulae in the simulated upper airway. However, a more realistic and complex representation would make the model too unwieldy, if not impossible to implement and validate.

Another simplification is the use of Ohmic analogy, which cannot easily accommodate mechanical factors, such as inertia. Inertia resulted in overshoot of the water meniscus past the inspiratory and expiratory endpoints, leading to measurement errors. Even where there was no obvious overshoot, inertia affected the speed at which the water meniscus returned to the inspiratory start point. For example, where the simulated lung was more compliant, exponential wash-out was more gradual, leading to better agreement with the theoretical value for more complete emptying (Table 2). In contrast, where the simulated compliance was decreased, the inertial effects seemed more pronounced, tending to critical damping or overshoot, leading to better agreement with the theoretical value for less complete emptying (Table 2). Future approximation of the inertance in our experimental models to human chest inertance (which typically ranges between 1 and 1.2 Pa litre s$^{-1}$) may improve clinical relevance, albeit at the expense of complexity.

Other unmodelled factors, such as upper airway compliance, may contribute to the discrepancies between predicted and observed inspiratory times seen in Table 2. For instance, at complete airway obstruction, a fresh gas flow of 3 litre min$^{-1}$ (50 ml s$^{-1}$) should deliver a 100 ml tidal volume in 2 s irrespective of respiratory compliance, as there was no alternative expiratory route. However, our experiments showed that inspiratory time was greater than predicted. Other investigators seemed to have observed similar discrepancies. Using a generic delivery circuit similar to ours at a fresh gas flow of 12 litre min$^{-1}$ (200 ml s$^{-1}$), Ryder and colleagues took 4.0 s to deliver a 500 ml inspiratory volume to a simulated lung with complete glottic obstruction—theoretically, it should have taken 2.5 s. This may be partly explained by the compliance of the simulated upper airway, which absorbs some of the inspiratory volume injected. Therefore, incorporating an upper airway compliance similar to that of the human upper airway (measured to be around 5–10 ml cm H$_2$O$^{-1}$) into our models may help with future quantitative validation. The higher initial costs of overcoming friction and changing momentum might also be contributory factors.

Observer error affected the experimental timing of the inspiratory and expiratory phases. This introduced errors in the measurement of the tilt of the simulated lung and in the observation of the meniscus past the inspiratory and expiratory endpoints. We attempted to reduce the latter error through the use of time-stamp video-recording. However, video clarity varied with video resolution, lighting conditions, and unavoidable signal noise. This was most significant with the expiratory endpoint, as exponential wash-out meant that the meniscus slows down dramatically towards end-expiration and every pixel difference is important. Further, the shape of the meniscus was often subtly different on return, making definitive determination of the expiratory endpoint difficult even with video playback. Observer error was therefore unavoidable.

Clinical application
Although work remains to be done with respect to the quantitative validation of our mathematical model, it is qualitatively robust, that is, the contrasting behaviour of the inspiratory and expiratory times with upper airway resistance is both predicted and repeatedly observed. This relationship may be applied clinically, by explaining the ventilatory problems associated with NBC in terms of upper airway resistance and fresh gas flow.

The inability to generate an adequate inspiratory volume with NBC may result from too low an upper airway resistance or too short an inspiratory time for that resistance.
This explains why different cannulae have been observed to have different effectiveness under varying upper airway conditions. Conversely, barotrauma may result from air trapping due to excessively high fresh gas flows or excessively long inspiratory times, for the upper airway resistance. This agrees with the observation by Dworkin and colleagues that the same lung and ventilation factors that promote inflation tend also to promote air trapping. (Dworkin and colleagues note that the diameter of the trachea relative to the breathing tube or cannula, i.e. the effective tracheal diameter, is a factor in air trapping. This is not relevant in NBC, since the cannula diameter is much smaller than tracheal diameter. Instead, we have concatenated all the resistances proximal to the transtracheal cannula into a single upper airway resistance variable.)

Optimal upper airway resistance should therefore be achieved for maximal clinical effect and minimal clinical risk. As we had noted in the Results section, this is that at which inspiratory and expiratory times are comparable in duration and neither prolonged. Practically, we suggest that the inspiratory time may be measured from the point of rest to when the chest wall just begins to rise and expiratory time measured from the end of inspiration (as defined immediately previously) to when the chest wall just stops deflating, in view of the low tidal volumes generated with NBC without jet ventilation. Once the resistance is estimated, larger tidal volumes may be delivered, if possible.

If either inspiratory or expiratory time is prolonged, we can exploit the contrasting patterns of changes in the inspiratory and expiratory times of cannula cricothyrotomy ventilation with upper airway resistance to diagnose and manage the problem. Thus, if the inspiratory time is disproportionately prolonged, the upper airway resistance may be too low. This may be remedied by relaxing jaw thrust, applying occluded airway devices, or increasing the fresh gas flow if using a generic cricothyrotomy circuit as described. (It is probably preferable to apply airway manipulation with care, lest the transtracheal cannula be dislodged and more problems result. We suggest that a better alternative when using the generic circuit is to change the fresh gas flow.) However, if the upper airway resistance is so low that we need to apply counter-measures, then the problem hindering supraglottic ventilation may well have resolved (e.g. laryngeal oedema resolving after epinephrine) and supraglottic ventilation should probably be re-attempted. Conversely, if the expiratory time is disproportionately prolonged, the upper airway resistance or the alternate expiratory resistance may be too high and the reverse manoeuvres may be applied. If both inspiratory and expiratory times are excessively prolonged, there may be obstacles along the inspiratory route, such as cannula kinking or bronchospasm. Adjusting the cannula position or transtracheal injection of epinephrine may be indicated.

In conclusion, our models demonstrate that the inspiratory graph is a hyperbolic curve rising with decreasing upper airway resistance and the expiratory graph is a sigmoid curve rising with increasing upper airway resistance. We can exploit the contrasting patterns of change of inspiratory and expiratory times with upper airway resistance to improve emergency short-term narrow-bore cannula cricothyrotomy ventilation. We agree with other authors in not recommending inspiratory:expiratory ratios fixed by rote. Instead, clinical flexibility and skill are indicated. This goes back to the need for training, in no small part for which we designed our models.

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Appendix I

Mathematical model of our electrical analogy of cannula cricothyrotomy ventilation

**Inspiratory phase**

Calculating pressures from the intratracheal point (Fig. 2), we have:

\[
\frac{dQ_{\text{p}}}{dt}R_p + Q_{\text{p}}E_R = \left( I_t - \frac{dQ_{\text{p}}}{dt} \right)R_u
\]

where \(Q_{\text{p}}\) is the instantaneous volume in the lung, \(dQ_{\text{p}}\) the incremental volume the lung is increasing by, \(I_t\) the fresh gas flow, \(E_R\) the respiratory elastance (the reciprocal of respiratory compliance), and \(R_u\) the pulmonary resistance.

Rearranging and integrating, we have:

\[
\int_{Q_{\text{ins}}}^{Q_{\text{ins}} + \Delta Q_{\text{ins}}} \frac{dQ_{\text{p}}}{I_tR_u - Q_{\text{p}}E_R} = \int_0^{t_{\text{ins}}} \frac{dr}{R_p + R_u}
\]

where \(Q_{\text{ins}}\) is the end-inspiratory lung volume and \(Q_{\text{exh}}\) the end-expiratory lung volume.
Cricothyroid needle ventilation

This gives:

\[ t_{\text{ins}} = -\frac{R_p + R_u}{E_R} \ln \frac{I_t R_u - Q_{\text{ins}} E_R}{I_t R_u - Q_{\text{exh}} E_R} \quad (2) \]

This provides the mathematical basis for the inspiratory graph of Figure 4.

Analysing for the shape of the graph, the logarithmic part of equation (2) should be greater than zero. Thus, \( I_t R_u - Q_{\text{ins}} E_R > 0 \). Thus, \( R_u \to (Q_{\text{ins}} E_R)/I_t \), \( t_{\text{ins}} \to \infty \), that is, it is not possible to inflate the lung to the desired tidal volume when the upper airway resistance is too low—the threshold being dependent on the inspiratory volume, the respiratory elastance, and the fresh gas flow. Conversely, as \( R_u \to \infty \), a re-written equation (1) simplifies from \( (dQ_p/dt)/(R_p/R_u) + 1 = I_t - (Q_p E_R)/R_u \) to \( dQ_p/dt = I_t \), thus giving \( t_{\text{ins}} = (Q_{\text{ins}} - Q_{\text{exh}})/I_t \). In other words, virtually all the fresh gas flow enters the lungs and the inspiratory time decreases to a constant that is dependent on the end-inspiratory volume, the end-expiratory volume, and the fresh gas flow. To sum, the inspiratory curve is a hyperbolic curve with a vertical asymptote at \( R_u = (Q_{\text{ins}} E_R)/I_t \) and a horizontal asymptote at \( t_{\text{ins}} = (Q_{\text{ins}} - Q_{\text{exh}})/I_t \).

**Expiratory phase**

Calculating pressures along the expiratory path from the lung via the upper airway:

\[ Q_p E_p = \frac{dQ_p}{dt} R_p + \left( \frac{dQ_p}{dt} - I_c \right) R_u \]

where \( I_c \) is the gas flow from the trachea through the cannula to the three-way tap.

Rearranging, we have:

\[ \frac{dQ_p}{dt} = \frac{I_t R_u + Q_p E_R}{R_p + R_u} \quad (3) \]

Calculating pressures along the parallel routes of the upper airway via the transtracheal cannula and the alternate expiratory pathway via the three-way tap, we have: \( V_t + V_c = V_u \), where \( V_t \), \( V_c \), and \( V_u \) are the pressure across the tap, cannula, and upper airway, respectively.

From experiments, we know: \( V_c = a I_c^3 + b I_c^2 + c I_c \), where \( I_c \) is the gas flow from the trachea away from the cannula and \( V_t = d(I_t - I_c)^2 + e(I_t - I_c) \).

Solving the two simultaneous equations, we have:

\[ a I_c^3 + (b + d) I_c^2 + (c + 2d I_t + e + R_u) I_c \]

\[ = \frac{dQ_p}{dt} R_u - dI_c^2 - eI_t \quad (4) \]

Substituting equation (3) into equation (4) and rearranging, we have:

\[ a I_c^3 + (b + d) I_c^2 + \left( c + 2d I_t + e + R_u - \frac{R_u^2}{R_p + R_u} \right) I_c \]

\[ = \frac{Q_p E_R R_u}{R_p + R_u} - dI_c^2 - eI_t \quad (5) \]

This is in the form of: \( w t_1 + x t_2 + y t = z \). This is cumbersome to solve manually. We therefore used numerical computing software, which gave three possible solutions (two of which are identical). The correct solution was selected by substituting known values. Inspecting the ohmic diagram of expiration, it becomes apparent that without active forces, there can be no passive forces. Therefore, at a fresh gas flow of zero and a lung volume of zero, there can be no flow through the cannula. The solution that yielded the correct value for cannula flow under these conditions is:

\[ I_c = \left( \frac{z}{2w} - \left( \frac{x}{3w} \right)^3 + \frac{xy}{6w^2} + \left( \frac{y}{3w} \right)^3 + \frac{z}{2w} \right) \]

\[ - \left( \frac{x^3}{27w^4} + \frac{y^2}{108w^4} \right) \left( \frac{z}{6w^3} \right)^{1/3} \]

\[ \left( \frac{y}{3w} \right)^3 + \left( \frac{z}{2w} \right)^2 - \left( \frac{x^2}{27w^4} - \frac{y^2}{108w^4} + \frac{z^2}{6w^3} \right)^{1/3} \quad (6) \]

Values for the variables \( w, x, y, \) and \( z \) are obtained by substituting values of their component variables and the value of \( I_c \) obtained for the instantaneous \( Q_p \). This is substituted back into 3 and the incremental \( Q_{\text{exh}} \) obtained by multiplying \( I_c \) by a time-slice interval of 0.001 \( \text{s} \), \( t_{\text{exh}} \) is the time needed to reach \( Q_{\text{exh}} \) equal to the defined fraction of \( Q_{\text{ins}} \).

Analysing for the shape of the expiratory curve, from equation (3), as \( R_u \to 0 \), \( dQ_p/dt \to Q_p E_R / R_p \). Integrating, we have:

\[ \int_{Q_{\text{ins}}}^{Q_{\text{exh}}} \left( \frac{dQ_p}{Q_p} \right) \to \int_{t_{\text{exh}}}^{t_{\text{ins}}} \frac{E_R}{R_p} dt \]

which gives \( t_{\text{exh}} \to R_p / E_R \ln Q_{\text{ins}} / Q_{\text{exh}} \). Thus, the expiratory time decreases with upper airway resistance towards a constant value that is dependent on pulmonary characteristics and independent of the upper airway resistance. Conversely, as \( R_u \to \infty \), equation (3) becomes: \( dQ_p/dt \to I_c \) and equation (5) becomes:
\[ aI_t^2 + (b + d)I_t^2 + (c + 2dl_t + e)I_t = Q_{exh} - dI_t^2 - eI_t. \]

Thus, expiratory flow becomes a constant that is dependent on the lung volume, the fresh gas flow, and the resistance of the alternate expiratory pathway through the cricothyrotomy cannula. For a constant expiratory volume, \( t_{exh} \) becomes a plateau. The expiratory curve is therefore a sigmoid curve, with expiratory time limited by pulmonary characteristics at low upper airway resistances, and by the tidal volume, the fresh gas flow, and the resistance of the alternate expiratory pathway at high upper airway resistances.

**End-expiratory volume**

Solving for \( t_{ins} \) and \( t_{exh} \) requires \( Q_{exh} \), which is taken to be the point where there is no further expiratory pulmonary flow. In this case, pressure is maintained by fresh gas flow exiting from the upper airway and the three-way tap.

Equation (3) becomes:

\[ Q_{exh} = -I_cR_u \quad (7) \]

\( I_c \) is negative because the gas flow is from the three-way tap to the trachea.

Since the gas flow is unidirectional (from source to upper airway) through the transtracheal cannula, we can simplify the pressure–flow relationship (using the same raw data) across the cannula to a quadratic equation:

\[ V_c = b'I_c^2 + c'I_c \]

Equation (5) becomes:

\[
(b' - d)I_c^2 + (c' + 2dl_t + e - R_u)I_c - dI_t^2 - eI_t = 0, \\
- (c' + 2dl_t + e - R_u) \\
I_c = \frac{\pm \sqrt{(c' + 2dl_t + e - R_u)^2 - 4(b' - d)(dI_t^2 + eI_t)}}{2(b' - d)}
\]

Since \( I_c \) is unidirectional towards the trachea, we accept only the positive root of equation (8). This is then substituted into equation (7) to yield \( Q_{exh} \).

**References**