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Effect of inspiratory flow waveform on work on endotracheal tubes: a model analysis

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Abstract *Objective:* This model analysis aimed to predict the impact of different inspiratory flow waveforms, i. e., constant, sinusoidal, and two linearly decreasing flows, on the resistive work (W_{res}) performed on endotracheal tubes.

Design: Model analysis.

Results: Model analysis predicts that: (i) minimal W_{res} is obtained with the constant flow; (ii) for any given tidal volume/inspiratory duration (V_T/T_I , mean inspiratory flow), W_{res} increases with decreasing tube size; (iii) for any given inspiratory flow waveform, W_{res} increases with increasing V_T/T_I , being lowest with constant flow.

Conclusions: These findings have major clinical implications: at any given ventilator setting, not only the size of the endotracheal tube but

also the inspiratory flow waveform must be taken into account to interpret the values of W_{res} and hence of the total work of breathing.

Key words Flow resistance · Sinusoidal · Square wave and decelerating flow · Mechanical ventilation

Introduction

Most modern microprocessor-equipped ventilators allow selection of different inspiratory flow waveforms to provide options to improve the distribution of the inspired gas and hence pulmonary gas exchange in patients with inhomogeneous lungs and different diseases underlying respiratory failure [1]. Some reports have demonstrated improved gas exchange with decelerating flow [2–5], while others showed no significant effect of different inspiratory flow waves (constant, decelerating, sinusoidal) on arterial blood gases [6, 7]. However, it has been shown that the work performed by the ventilator differed significantly among sinusoidal, decelerating, and constant inspiratory flow patterns, mainly due to

the resistive properties of the narrow bore endotracheal tubes (ETTs) [6]. Although in 1925 Bretschger [8] suggested that constant flow minimized the mechanical work of breathing, it took almost 50 years for this to be confirmed mathematically [9]. In the present model analysis, we predict the effect of various flow waveforms commonly used during mechanical ventilation on the work done on the ETTs.

Inspiratory flow waveforms

Our analysis was based on the four flow patterns illustrated in Fig. 1 – namely, constant (C), sinusoidal (S), and linearly decreasing flow from a maximal initial

Fig. 1 Inspiratory flow patterns used in the present study. Lung inflations of 0.6 l and 0.8 s duration. *C* constant flow, *S* sinusoidal flow, *D*_{1/3} and *D*₀ decelerating flow from a maximal value (*V*'max) to 1/3 *V*'max and zero, respectively

value (*V*'max) to 1/3 *V*'max (*D*_{1/3}) or zero (*D*₀) at the end of inspiration.

ETT resistance

The resistance of endotracheal tube (*R*) can adequately be described by Rohrer's equation [10, 11]:

$$R = K1 + K2 \cdot V' \quad (1)$$

where *V*' is flow, and *K*1 and *K*2 are constants. The values of *K*1 and *K*2 of the three Mallinckrodt ETTs used in the present analysis are listed in Table 1. Values pertain to inspiration.

Resistive inspiratory work

The resistive inspiratory work (*W*res) is given by:

$$Wres = V_T [a \cdot K1(V_T/T_i) + b \cdot K2(V_T/T_i)^2] \quad (2)$$

where *V*_T is tidal volume, *T*_i inspiratory duration, *V*_T/*T*_i mean inspiratory flow, and *a* and *b* are factors which depend on the flow waveform (see appendix). Table 2 provides the values of these shape-dependent constants. Both *a* and *b* increase progressively in the following sequence: *C*, *D*_{1/3}, *S*, *D*₀. With *D*₀, the values of *a* and *b* are, respectively, 33 and 100% higher than with *C*.

Figure 2 depicts the relationship between *W*res and *V*_T/*T*_i for the different flow waveforms and ETTs, computed according to Eq. 2 using the constants in Tables 1 and 2 for a fixed *V*_T of 0.6 l. For a given ETT and

Table 1 Values^a of *K*1 and *K*2 of Eq. 1 for the endotracheal tubes (including connectors) of different internal diameters *i. d.* and standard length

<i>i. d.</i> (mm)	<i>K</i> 1 (cmH ₂ O · s · l ⁻¹)	<i>K</i> 2 (cmH ₂ O · s ² · l ⁻²)	<i>K</i> 1/ <i>K</i> 2 (l · s ⁻¹)
7	0.75	8.75	0.086
8	0.52	4.32	0.120
9	0.31	2.42	0.128

^a Values determined by Navalesi P and Gottfried SB (unpublished observations) with the precautions described by Chang and Mortola [12]

Table 2 Values of constant *a* and *b* in Eq. 2 for various inspiratory flow waveforms (*V*'max flow at the onset of lung inflation)

Flow waveform	<i>a</i>	<i>b</i>
Constant (<i>C</i>)	1.00	1.00
Decelerating to 1/3 <i>V</i> 'max (<i>D</i> _{1/3})	1.08	1.25
Sinusoidal (<i>S</i>)	1.23	1.64
Decelerating to zero flow (<i>D</i> ₀)	1.33	2.00

*V*_T/*T*_i, *W*res increases progressively according to the following sequence: *C*, *D*_{1/3}, *S*, *D*₀, reflecting the corresponding increases in *a* and *b* (Table 2).

Figure 3 (left) depicts the relationship of *W*res with *S*, *D*_{1/3}, and *D*₀, expressed as a fraction of *W*res for constant flow (*W*res, *C*) to *V*_T/*T*_i. For each flow waveform, the ratio increases progressively with augmenting *V*_T/*T*_i, approaching an asymptote which corresponds to *b* (Table 2). The intercepts of the curves in Fig. 3 (left) at zero *V*_T/*T*_i correspond to *a* (Table 2), while the relative increase of the *W*res/*W*res, *C* ratios from *a* to *b* is governed by the following equation:

$$(Wres/Wres, C - a)/(b - a) = V_T/T_i/(K1/K2 + V_T/T_i) \quad (3)$$

This equation shows that, allowing for *a* and *b*, the rate of rise of *W*res/*W*res, *C* with increasing *V*_T/*T*_i is inversely related to the ratio of *K*1/*K*2: that is, the greater this ratio, the slower the rate of the rise of *W*res/*W*res, *C*.

Figure 3 (right) depicts the relationships of *W*res, *S*/*W*res, *C* to *V*_T/*T*_i for the three different ETTs. While all curves have the same intercept at zero *V*_T/*T*_i (Table 2), the increase in the ratio *W*res, *S*/*W*res, *C* with increasing *V*_T/*T*_i is greater with a no. 7 ETT, reflecting the lower *K*1/*K*2 ratio to that of ETTs 8 and 9 (Table 1). Since the *K*1/*K*2 ratio of the latter ETTs was similar (0.12 and 0.128, respectively), the corresponding relationships of *W*res, *S*/*W*res, *C* to *V*_T/*T*_i were essentially the same, as predicted by Eq. 3.

As shown in Fig. 3, over the range of *V*_T/*T*_i commonly used in mechanically ventilated intensive care patients (0.5 to 1 l · s⁻¹), the *W*res/*W*res, *C* ratios are close

Fig. 2 Relationship of inspiratory resistive work W_{res} done on endotracheal tubes of different internal diameters (mm) to mean inspiratory flow V_T/T_i for various flow waveforms. Relationships pertain to a fixed V_T of 0.6 l. C constant flow, S sinusoidal flow, $D_{1/3}$ and D_0 decelerating flow from a maximal value (V'_{max}) to $1/3 V'_{max}$ and zero, respectively

to the maximal values of the constant b . As a result, at $V_T/T_i > 0.5 \text{ l} \cdot \text{s}^{-1}$ the relative increase of W_{res} with S , $D_{1/3}$, and D_0 , with respect to W_{res} , C , is nearly maximal. However, the absolute differences in W_{res} between the

various flow patterns and constant flow increase with increasing V_T/T_i (Fig. 2).

Polese and coworkers [6] measured the inspiratory work of breathing at a fixed V_T (0.6 l) and T_i (1.6 s) in eight patients after open-heart surgery during passive mechanical ventilation with constant, sinusoidal, and decelerating inspiratory flow. In line with our predictions, we found that with sinusoidal and decelerating flow the inspiratory work of breathing was higher than with constant flow, the discrepancy being mainly due to the W_{res} of the ETT.

Fig. 3 Left Relationships of resistive work W_{res} with different flow waveforms, expressed as a fraction of corresponding work with constant flow W_{res} , C to mean inspiratory flow (V_T/T_i). Relationships pertain to a V_T of 0.6 l and endotracheal tube of 8-mm i. d. S sinusoidal flow, $D_{1/3}$ and D_0 decelerating flow from a maximal value (V'_{max}) to $1/3 V'_{max}$ and zero, respectively. The asymptotic values of $W_{res}/W_{res}, C$, corresponding to the constant b (Table 2), are indicated. **Right** Relationships of W_{res} , S/W_{res} , C for endotracheal tubes of different sizes. The intercept of the curves at zero V_T/T_i is indicated by a and the asymptote of infinite V_T/T_i by b

Clinical implications

Measurements of work of breathing (WOB) during mechanical ventilation has been advocated to assess the status and progress of intensive care patients [13–15] and as a weaning parameter [16, 17]. In most clinical trials, WOB includes the resistive work done on the ETT. Since the magnitude of WOB depends on the inspiratory flow pattern, the latter should be allowed for, particularly if ETTs of small size are used. In contrast,

with size 9 the Wres due to ETT is relatively small, independent of the flow pattern (Fig. 2, right). It should be noted, however, that lung inflation with patterns other than constant flow should also increase the resistive work done on the patient (Eq. 2). In fact, Polese et al. [6] found that with constant flow, Wres was lower than with both sinusoidal and decreasing flow pattern.

The effect of inspiratory flow pattern on WOB in normal, spontaneously breathing subjects has been studied by several authors [9, 18, 19]. Lafortuna and colleagues [20] found that in normal subjects the sinusoidal inspiratory flow pattern used during quiet breathing changed into a quasi-constant inspiratory flow profile during voluntary hyperventilation and exercise. This was interpreted as a choice of the respiratory controller of a more economical flow pattern in terms of WOB as ventilation increases [20]. Those results are in agreement with our model predictions. No such analysis is available for spontaneous breathing during weaning. Clearly, in view of its important potential effect on WOB, the inspiratory flow pattern adopted by patients during weaning without ventilatory support needs to be assessed. Under these conditions, a quasi-constant inspiratory flow profile should be beneficial for minimizing WOB in view of the fact that in most intensive care patients flow resistance is increased due to airway obstruction and/or tracheal intubation.

It should be noted that our model analysis was not designed to predict the effects of different inspiratory flow patterns on patient WOB during assisted mechanical ventilation. Recent work by Cinella et al. [21] showed less patient WOB with pressure control (decel-

erating flow) than volume ventilation (square wave flow) because of greater inspiratory muscle inhibition, with an early elevated peak flow in the former case.

Appendix

The flow-time relationships of constant (C), sinusoidal (S), decelerating flow to one third of the initial value ($D_{1/3}$) or to zero (D_0) are given, respectively, by:

$$V'_{(t)} = V_T/T_i \quad (1A)$$

$$V'_{(t)} = V_T/T_i [0.5 \pi \cdot \sin(\pi \cdot t/T_i)] = V_T/T_i \cdot G \quad (2A)$$

$$V'_{(t)} = V_T/T_i [1.5 - t/T_i] = V_T/T_i \cdot G \quad (3A)$$

$$V'_{(t)} = V_T/T_i [2 - 2 t/T_i] = V_T/T_i \cdot G \quad (4A)$$

The time dependence of resistive pressure is given by:

$$\text{Pres}_{(t)} = (K_1 \cdot V_T/T_i \cdot G) + [K_2 \cdot (V_T/T_i)^2 \cdot G^2] \quad (5A)$$

Accordingly, the resistive work is:

$$\text{Wres} = \int_0^{T_i} \text{Pres}_{(t)} \cdot V'_{(t)} \cdot dt \quad (6A)$$

Substitution of $\text{Pres}_{(t)}$ and $V'_{(t)}$ from Eq. 5A and Eqs. 1–4A, respectively, into Eq. 6A yields:

$$\text{Wres} = V_T \cdot K_1 \cdot (V_T/T_i) \int_0^{T_i} G^2 \cdot dt + V_T \cdot K_2 \cdot (V_T/T_i)^2 \int_0^{T_i} G^3 \cdot dt \quad (7A)$$

where the integrals correspond to a and b in Eq. 2, respectively [22].

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