Effect of inspiratory flow waveforms on the mechanical work of ventilation; a fluid dynamic analysis

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Research

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Abstract

In most intensive care patients, the mechanical work (MW) is increased due to airway obstruction and/or tracheal intubation. Increasing MW is known as a risk factor for ventilator induced lung injury (VILI). Moeover, minimizing of MW is crucial to facilitate weaning process. In this paper MW is compared between three different inspiratory flow waveforms. The fluid dynamic analysis is used to compute the resistive pressure drop and the resistive work. We have compared square, sinusoidal and decelerating flow waveforms under the same tidal volume. The results show that under the constant tidal volume and I:E ratio, for tidal volumes below 1 lit, an square flow profile is beneficial for minimizing MW while a sinusoidal flow profile is prefered for tidal volumes of 1 lit or higher. It is shown that for make a decision about most beneficial flow profile in terms of less MW, both tidal volume and I:E ratio is important. By the way the results suggest to use decelerating flow waveforms with higher 1:E ratio. The qualitative conclusion is that in order to lower the MW especially in patients with obstructive lung diseases, sinusoidal, square and decelerating flow became preferable respectively with increasing I:E ratio. Our study suggest the square and sinusoidal profile for tidal volumes below and equal or over 1 lit, respectively in pointwiew of less MW. This paper also encouraged the engineers to add an option to select sinusoidal flow waveform in VCV mode to lower MW when tidal volume is l lit or higher.

Keywords: Mechanical work, inspiratory flow waveform, resistive pressure drop

Introduction

Traditional strategies to overcome ventilator-associated lung injury (VALI) is limiting tidal volume to protect from overinflation and using PEEP to prevent cyclic alveolar collapse. The concept of mechanical power (MP) as a measure for development of ventilator-induced lung injury (VILI) is a promising idea (1). MP is defined as MW divided by the time and can be derived by multiplication of MW with respiratory rate (RR). MW may be defined as the product of pressure and volume. It can be analyzed by plotting airway pressure against tidal volume. Minimizing imposed MP is crucial for lowering the risk of VILI in patients under mechanical ventilation. Moreover, in order to facilitate weaning process for most intensive care patients, it is desired to reduce MW. From this point of view, we are interested to know which inspiratory flow waveform imposes less MW at a constant RR.

A decelerating flow pattern is suggested for patients with restrictive lung disease such as acute respiratory distress syndrome (ARDS) and acute lung injury, because of reducing the risk of VILI and more even gas distribution (2, 3). But its shortened expiratory time is not suggested for obstructive pulmonary diseases. For these patients with asthma or chronic obstructive pulmonary disease (COPD) a square waveform is suggested (4, 5).

Conflicting results have been obtained, from both animal models and clinical observations, as to the relative effectiveness of different inspiratory flow patterns during mechanical ventilation. The selection of decelerating flow has been claimed to favor better gas exchange when compared with volume-controlled ventilation (VCV) with constant inspiratory flow (6-15). Pressure-controlled ventilation has some advantages (11) for example its resulting square wave pressure waveform provides the maximum inspiratory pressure for the entire inspiratory time favors lung recruitment. This beneficial effect of pressure-controlled ventilation may be useful to overcome atelectasis. However, tidal recruitment increases mechanical stress on the lungs, and this may promote ventilator-induced lung injury especially in preterm infants (16).

Dembinski et al. (17) compared the effect of decelerating, square and a fixed combination of both flow wave forms on the distribution of ventilation and perfusion in an animal model of acute lung injury. They reported that contrary to the hypothesis, square waveform provides a more favorable distribution of ventilation and perfusion, and hence better oxygenation, when compared with decelerating or combined flow waveforms in this model of ALI. Their different conclusion may be come from choosing lower tidal volume for square rather than decelerating or combined flow waveforms.

Roth et al. (18) reported that decelerating inspiratory flow had no beneficial effects on pulmonary gas exchange when compared with square inspiratory flow while an increase in mean airway pressure in Pressure-controlled ventilation may raise the potential risk of ventilator induced lung injury. Some other experimental studies (19, 20) showed that there are no differences between square or decelerating flow waveform in oxygenation. Antonaglia et al. mathematical model(21) showed that both ventilatory modes provided similar gas distribution, but in square flow peak pressures were higher in the sicker compartment respect to decelerating flow. They demonstrated that less pressure variability in pressure controlled ventilation could reduce the potential ventilator induced lung injury.

In this study we are aimed to compare different inspiratory waveform from the viewpoint of the MW. The components of MW include the work needed to overcome resistive and elastic properties of the respiratory system. The elastic work is not dependent to flow details and it is constant used for a given inspiratory volume. So the flow-dependent component of MW is resistive work. Resistive work must be done to overcome the frictional resistance to gas flow that occurs in the airways. Airways beyond generation 10 contribute less than 10% to airway resistance (22). So, the resistive work can be computed considering the first 10 generations of the respiratory tract. The mechanical work is compared between square, decelerating, and sine waveforms of inspiratory flow. The fluid dynamics is analyzed to derive pressure drop in lung airways. The resistive work then computed to find the most favorable flow pattern for ventilated patients in terms of less MW.

Material and methods

Airway pressure during PCV can be described mathematically by Eq. 1 (15). Airway pressure (P) increases exponentially to the pressure support level (PS) with a rising time (t_r) and then remains constant until the termination of the inspiratory phase.

$$P = PS(1 - e^{-t/t_r}) \tag{1}$$

Inspiratory flow is determined by the airway pressure, the airways resistance, and the respiratory time constant τ which is the product of airways resistance R and respiratory system compliance C. PCV has a decelerating flow waveform explained by Eq. 2.

$$f_{decelerating} = \frac{P}{R} e^{-t/\tau} \tag{2}$$

The area of the flow-time curve is the delivered tidal volume V_T which can be computed by Eq. 3.

$$V_T = \int_0^{t_i} f_{decelerating} dt = \int_0^{t_i} \frac{PS(1 - e^{-t/t_r})}{R} e^{-t/\tau} dt = \frac{PS}{R}(\tau - \tau e^{(-t_i/\tau)} + (t_r \tau(-1 + e^{-(1/t_r + 1/\tau)t_i})) / (t_r + \tau)) \tag{3}$$

For comparison, consider other flow waveforms during VCV which has similar V_T. Eqs. 4 and 5 descerib e square and sinusoidal flow waveforms.

$$f_{square} = \frac{V_T}{t_i} \tag{4}$$

$$f_{sinusoidal} = \frac{V_T \pi}{2t_i} \sin(\frac{\pi t}{t_i}) \tag{5}$$

The MW can be found by Eq. 6.

$$MW = \int_0^{t_i} f.p\, dt \tag{6}$$

So, the MW can be compared for different inspiratory flow waveforms.

$$MW_{decelerating} = \int_0^{t_i} f_{decelerating}.P_{decelerating}dt = \int_0^{t_i} \frac{P^2}{R} e^{-t/\tau}dt = \frac{PS^2}{R}\int_0^{t_i} (1-e^{-t/t_r})^2 e^{-t/\tau}dt =$$

$$\frac{PS^2}{R}\left[\begin{array}{l}(\tau e^{t_i(-2/t_r-1/\tau)}(-e^{(2t_i/t_r)}(t_r^{\,2}+3t_r\tau-2\tau^2(e^{(t_i/\tau)}-1))-t_r(t_r+\tau)+2t_r e^{(t_i/t_r)}(t_r+2\tau))) \\ /((t_r+\tau)(t_r+2\tau))\end{array}\right] \quad (7)$$

The airway pressure in VCV mode can be determined by the equation of motion which states that the pressure required to deliver a volume of gas into the lungs is determined by the elastic and resistive properties of the respiratory system.

$$P^{VCV} = R.f + V/C \quad (8)$$

In this equation, f and V are respectively the air flow rate and the delivered tidal volume by the time of t. The MW under VCV mode then can be determined by Eq. 9.

$$MW^{VCV} = \int_0^{t_i} f(R.f + V/C)dt = R\int_0^{t_i} f^2 dt + V_T^{\,2}/(2C) \quad (9)$$

The first term of the above equation is the resistive MW needs to overcome the resistance of the airways and the second term is the elastic MW required to inflate the lung which is not dependent to flow waveform. Substituting Eqs. 4 and 5 into Eq. 9, the MW with square and sinusoidal flow waveforms can be found by Eqs. 10 and 11, respectively.

$$MW_{square} = \int_0^{t_i} f_{square}(R.f_{square} + V_T/C)dt = \frac{RV_T^{\,2}}{t_i} + V_T^{\,2}/(2C) \quad (10)$$

$$MW_{sinusoidal} = \int_0^{t_i} f_{sinusoidal}(R.f_{sinusoidal} + V_T/C)dt = \frac{RV_T^{\,2}\pi^2}{8t_i} + V_T^{\,2}/(2C) \quad (11)$$

Referring back to Eq. 8, it will be seen that we have assumed that resistive pressure is directly related to flow. An assumption which is valid only in laminar flow regime. By the way, the airflow in mechanical ventilation could be as high as 120 lit/min. So, we should use a modified equation for higher flow rates to improve our analysis. Reynolds number is used as a criterion for characterizing the flow regime. When the Reynolds number is below 2300, we can expect the flow to be laminar, but when it is larger, the flow would be in critical or turbulent regime. Since the lung airways are coverd by mucus, the smooth pipe flow is valid for Re>2300 in lung airways. The Blasius equation can be used for friction coefficient in smooth pipes (23). So, the friction coefficient f_D is a function of Re as Eq. 12 where Q is the mean flow in an airway whose diameter and length are d and l, respectively. ρ and μ stand for the kinematic viscosity and density of air, respectively.

$$f_D = \frac{0.3164}{Re^{1/4}};\ Re = \frac{4\rho Q}{\mu \pi d} \tag{12}$$

The turbulent pressure loss in terms of flow rate Q can be expressed by the following equation.

$$\Delta P_{turbulent} = f_D \frac{l}{D}\frac{\rho v^2}{2} = 8 f_D \frac{l}{D}\frac{\rho Q^2}{\pi^2 D^4} = 0.241 \rho^{3/4} Q^{7/4} l \mu^{1/4} D^{-19/4} \tag{13}$$

The delivered flow reachs the trachea, then it divides between downstream airways. So, Re sould be calculated seperately in each airways of the lung to decide about fluid flow regime (laminar or turbulent). Considering Weieble symmetric model (24), the flow after each bifurcation equally divided between airways. Also, the geometrical characteristic of airways in each generation is given by this model. So, Re can be found in each generation of the lung assuming different level of PS or flow delivered settings of ventilator. For comparison three different tidal volumes of 0.5, 1 and 2 lit has been used which consequently generate different flow rates. In this study the respiratory rate of 15 per minute and I:E of 1/4 which corresponds to inpiration time of 1 sec is assumed as a reference condition. The corresponding levels of PS during PCV are 11.8, 23.7 and 47.3 cmH2O to deliver desired tidal volumes of 0.5, 1 and 2 lit.

The elastic work is not dependent to flow details and it is constant for a given inspiratory volume. So the flow-dependent component of MW is resistive work. Resistive work must be done to overcome the frictional resistance to air flow that occurs in the airways. Airways beyond generation 10 contribute less than 10% to airway resistance (22). So, the resistive work can be computed considering the first 10 generations of the respiratory tract (Table 1).

Table 1. First 10 generations of lung airways by Weibel model

Generation	**diameter(mm)**	**length(mm)**
0	18	120
1	12.2	47.6
2	8.3	19
3	5.6	7.6
4	4.5	12.7
5	3.5	10.7
6	2.8	9
7	2.3	7.6
8	1.86	6.4
9	1.54	5.4

Friction coefficient in fully developed laminar flow in an airway can be expressed by Eq. 14.

$$f_D = \frac{64}{Re} \tag{14}$$

However, the flow at the entrance of the airways is developing because the airway length is shorter than the required length to reach fully developed flow (l/d<0.06Re) and the velocity profile after each bifurcation will be disrupted. Considering the pressure drop due to flow development in the entrance region(25), the pressure drop (ΔP) in an airway with developing flow can be explained by Eq. 15.

$$\frac{\Delta P_{\text{laminar developing}}}{0.5\rho U^2} = f_D \frac{1}{D} + \frac{4}{3} \quad \Delta P_{\text{laminar developing}} = \frac{128\mu l Q}{\pi d^4} + \frac{32\rho Q^2}{3\pi^2 d^4} \tag{15}$$

where U is the mean velocity of air which can be calculated by $Q/(\pi d_2/4)$.

The flow and Re in each generation can be calculated for the assumed delivered flow waveform. Then the pressure drop can be computed by Eqs. 13 or 15 according to the flow regime.

In order to compute resistive MW, considereing the time profile of flow rate, the following integral should be computed instead of Eq. 6.

$$MW_{res} = \int_0^{t_i} \Delta P.f \, dt \tag{16}$$

So, The resistive MW in laminar and turbulent flow regime can be expressed by Eqs. 17 and 18, respectively.

$$MW_{res,laminar} = \frac{128\mu l}{\pi d^4} \int_0^{t_i} f^2 dt + \frac{32\rho}{3\pi^2 d^4} \int_0^{t_i} f^3 \, dt \tag{17}$$

$$MW_{res,turbulent} = 0.241\rho^{3/4} \, l\mu^{1/4} \, D^{-19/4} \int_0^{t_i} f^{11/4} \, dt \tag{18}$$

The integrals of flow with different power exponents can be computed mathematically using flow relations expressed by Eqs. 2, 4 and 5 for decelerating, square or sinusoidal flow waveforms, respectively.

For the sake of comparision, a reference condition for a normal lung (R 10 cmH2O/L/s, C 0.05 L/cmH2O) and a fast pressure rising (t_r=0.01 s), with constant respiratory rate of 15/min and I:E ratio of 1/4 has been assumed which correspods to t_i of 1 sec. The three flow waveforms with a constant tidal volume of 0.5 lit are shown in Fig. 1.

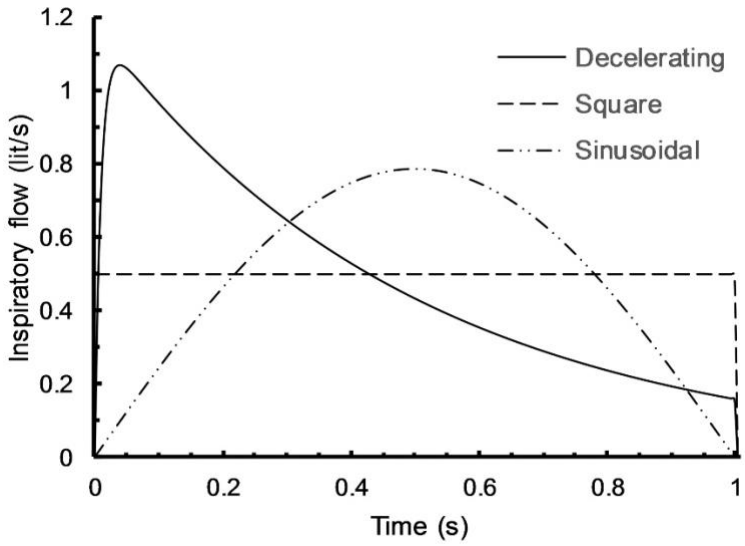

Fig. 1. Three different flow waveforms under tidal volume of 0.5 lit, RR=15/min, I:E=1/4.

We also compare profiles considering different I:E ratios in a constant respiratory rate of 15/min as shown graphyically for tidal volume of 0.5 lit in Fig.2.

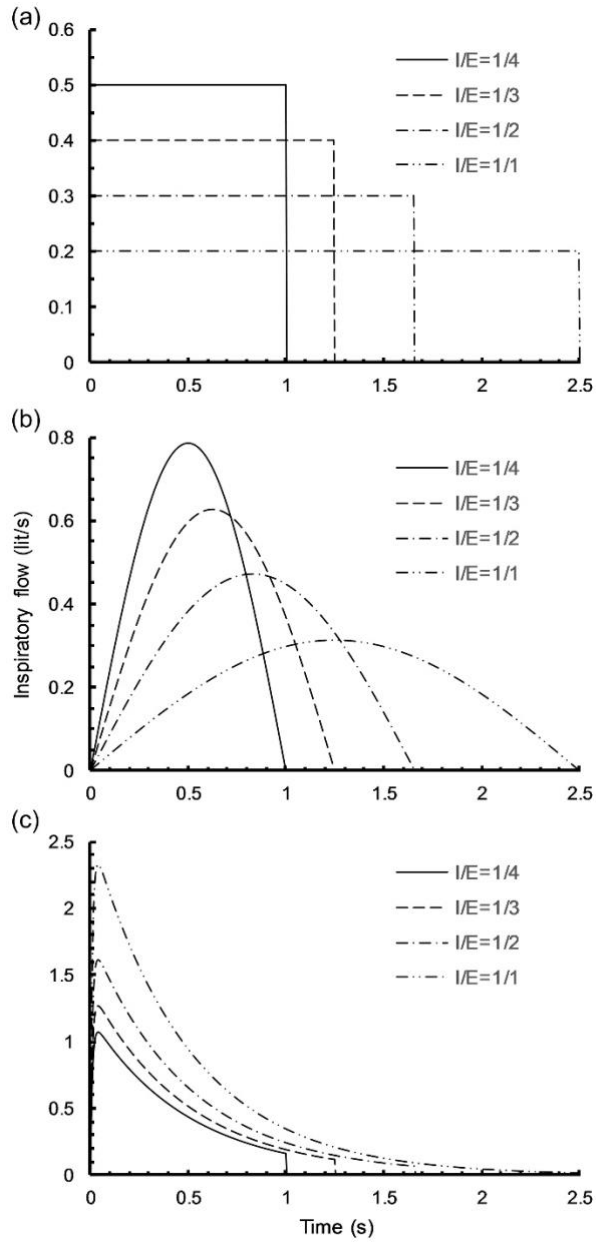

Fig. 2. Effect of changing I:E ratio (1/4, 1/3, 1/2 and 1/1) on inspiratory flow profile under constant tidal volume of 0.5 lit, a) square flow waveform b) sinusoidal flow waveform c) decelerating flow waveform.

Results

Assuming laminar fully developed flow, The MW in the reference condition (RR=15 lit/min, I:E=1/4) can be calculated through Eqs. 7, 10 and 11. The comparison of this results for different inspiratory flow waveforms with two different rise times and three types of lung mechanics (normal, resistive and obstructive) is shown in Table 2.

Table 2. MW of ventilation (cmH2O.lit) with different flow waveforms (decelerating, square and sinusoidal) with 3 types of lung mechanics. Also illustrated 2 levels of pressure support and 2 rise times. Note that the decelerating and square flow waveforms have the largest and the smallest MW, respectively.

Pressure Setting	**Inspiratory Flow Waveform**	**Normal Lung** **R 10 cmH2O/L/s, C 0.05 L/cmH2O**	**Restrictive Lung Dieases** **R 10 cmH2O/L/s, C 0.02 L/cmH2O**	**Obstructive Lung Dieases** **R 20 cmH2O/L/s, C 0.05 L/cmH2O**
t_r 0.01 s, PS 10 cmH2O	Decelerating Square Sinusoidal	4.18 3.57 3.99	1.84 1.25 1.33	3.09 2.90 3.36
t_r 0.1 s, PS 10 cmH2O	Decelerating Square Sinusoidal	3.11 2.44 2.72	1.05 0.61 0.65	2.49 2.20 2.54
t_r 0.01 s, PS 20 cmH2O	Decelerating Square Sinusoidal	16.71 14.28 15.95	7.38 5.01 5.34	12.34 11.61 13.42
t_r 0.1 s, PS 20 cmH2O	Decelerating Square Sinusoidal	12.44 9.74 10.88	4.21 2.44 2.60	9.96 8.79 10.16

As shown in Table 3, both in normal or dieasesd lungs, square and decelerating flow waveforms have the smallest and largest MW, respectively. This results is confirmed for both two levels of pressure support and rise times. On should note that the increase in the MW of ventilation by using faster rise time and higher pressure support setting is due to increasing tidal volume. Moreover, the work values between normal and diseased lungs can not be compared with each other due to different tidal volumes. The above table can be used only for comparing MW under different flow waveforms having the same tidal volume. The previous studies also, suggest the square instead of decelerating flow for patients with obstructive lung dieases to lower MW (4, 5).

Considering flow regimes and developing flow in lung airways, the resistive MW can be computed using Eqs. 17 and 18. The results have been compared for three levels of tidal volumes of 0.5, 1 and 2 lit under assumed reference condition as shown in Table 3.

Table 3. Resistive MW of ventilation (cmH2O.lit) assuming differe flow waveforms (decelerating, square and sinusoidal) with three levels of tidal volumes. For the volumes below 1 lit, the smallest resistive work is obtained by square flow, while the sinusoidal flow has the smallest resistive work for volumes equal or higher than 1 lit.

Inspiratory Flow Waveform	V_T=0.5 lit	V_T=1 lit	V_T=2 lit
Decelerating	0.051	0.154	0.558
Square	0.025	0.076	0.288
Sinusoidal	0.035	0.074	0.135

According to Table 3, decelerating flow waveform has the largest resistive work in all range of volumes similar to the previous results in Table 3 but the smallest MW can be obtained either by square or sinusoidal depending on the tidal volume. For volumes smaller than 1 lit, the resistive work with square flow waveform is smallest, while for volumes of 1 lit or more, the sinusoidal waveform has the smallest resistive work rather than two other flow waveforms. Our computations show that disregarding developing flow would not change the result. So, considering turbulence is important in computation of MW in larger tidal volumes.

The above analysis was under respiratory rate of 15/min and I:E of 1/4 which correspods to t_i of 1 sec. We have considered this same inspiration time for all comparisons but what happens if different inspiration times was used. In order to asnwer this quaestion, we compare the results repeating our analysis considering 1/3, 1/2 and 1/1 for I:E ratio in a constant respiratory rate of 15/min (i.e. t_i of 1.25, 1.66 and 2.5 sec). The results of resistive MW assuming different flow waveforms with different I:E ratios and tidal volumes are shown in Fig. 3.

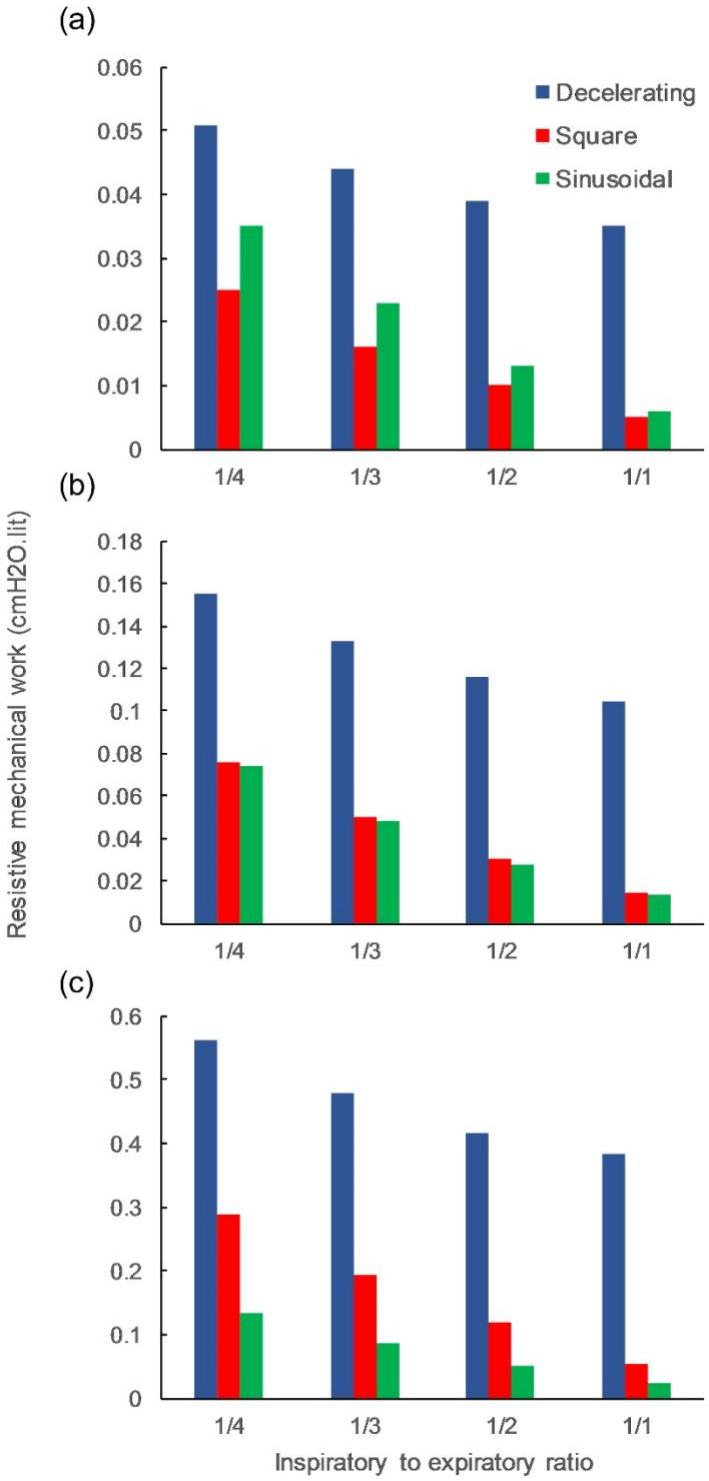

Fig. 3. Resistive MW of ventilation (cmH2O.lit) assuming different flow waveforms (decelerating, square and sinusoidal) with different I:E ratios under different tidal volumes; a) V_T=0.5 lit, b) V_T=1lit, c) V_T=2 lit; Increasing I:E ratio in a same RR would decrease MW for all flow profiles. In a constant I:E ratio, the square profile has the smallest MW for V_T=0.5 lit, while for V_T=1 lit and V_T=2 lit the smallest MW is

acheived by sinusoidal profile. However, under V_T=1 lit, a square profile under I:E=1/2 is better than the sinusoidal one under I:E=1/4 or 1/3 in terms of less MW.

According to Fig. 3 it can be deduced that lower tidal volumes have lower MWs. Moreover, increasing I:E ratio in a constant tidal volume will decrease resistive MW and consequently MW for each flow profile. It suggests that in order to lower MW, if tidal volume is below 1 lit, select square and if it is equal or higher than 1 lit, select sinusoidal profile. On the other hand, if you decide to use decelerating profile for a reason, you should increase I:E ratio as much as possible.

Comprehensive investigation of the MW values in table 5 shows that I:E ratio setting is important to lower MW. For instance under tidal volume of 0.5 lit, a decelerating profile with I:E=1/1 has the same MW as a sinusoidal profile with I:E=1/4. However under this tidal volume and constant I:E, the square profile has the least MW.

Similarly, be carefull in the results presented in the above table, under the same tidal volume of 1 lit, a sinuoidal profile is better than square and decelerating ones for all I:E ratios. However, under this tidal volume, a square profile under I:E=1/2 is better than the sinusoidal one under I:E=1/4 or 1/3. Decelerating profile has the largest MW rather than square and sinusoidal flow waveforms under the same tidal volume of 1 lit or higher.

Discussion

The resistive mechanical work is dependent on flow waveform. Since that linear relation between flow rate and pressure loss is not valid for large Re>2300, a new equation for pressure loss computation is presented. Our result show that different MW is computed assuming different flow regimes. Considering flow regimes based on calculated Re, is essential for accurate estimation of MW. Morover, according to our results inclusion of developing region in our model involved no significant change in MW. Our results show that for all tidal volumes in a reference condition of ventilator settings of RR and I:E ratio, square profile has the smallest MW if we assume laminar flow regime. However, considering flow regimes the results are different for tidal volumes below or over 1lit. For tidal volumes below 1 lit, square is the best profile in pointview of less MW, but for larger tidal volumes sinusoidal profile take an advantage over the two other flow profiles. Decelerating flow profile in all circumstances has the largest MW.

Furthure comparisons with different I:E ratios show that make an explicit decision about most favored flow waveform in terms of less MW strictly depends on the settings of tidal volume and I:E ratio on the mechanical ventilator. According to our findings, if you are forced to use a lower I:E ratio (1/2 or 1/3 or 1/4), you have two options according to your desired tidal volume. For tidal volumes below 1 lit, square and for 1 lit or more, sinusoidal profile is suggested. On the other hand, if you decide to use decelerating profile for a reason, you should increase I:E ratio as much as possible. It should be noted that this results is reported assuming respiratory rate of 15 per minute but other values also would not change the trend of our findings. Moreover, upgrading the conventional ventilator to be able to deliver sinusoidal flow waveform is strongly suggested.

Although our results do not suggest using decelerating flow profile due to its high MW, its beneficial aspects such as less pressure variability are not investigated in this study. However, we are not still sure about the best flow profile in terms of less MW, better gas exchange and reducing potential ventilator induced lung injury.

Conclusion

Due to the fact that in most intensive care patients flow resistance is increased due to airway obstruction and/or tracheal intubation, there are many attempting to minimize MW. Increasing MW is also known a risk factor for VILI. In this paper the MW of ventilation is computed and the effect of inspiratory flow pattern on it was studied. We used fluid dynamic analysis on a realistic model for lung airways and consider the air flow regime during mechanical ventilation. The MW is compared between three flow profiles of square, sinusoidal and decelerating.

Our result show that make an explicit decision about most favored flow waveform in terms of less MW is not possible and it is strictly depends on the settings of tidal volume and I:E ratio on the mechanical ventilator. By the way, increasing I:E ratio in a constant tidal volume will decrease resistive MW for all flow profiles. The qualitative conclusion is that in order to lower the MW especially in patients with obstructive lung diseases, sinusoidal, square and decelerating flow became preferable respectively as I:E ratio will be increased. Despite previous studies which suggest the square inspiratory flow profile for minimizing MW in patients with obstructive lung dieases (26), our study suggest the square profile for tidal volume of 1 lit and sinusoidal for tidal volume of 1 lit or higher. However, if you decide to use decelerating profile for a reason, you should increase I:E ratio as much as possible to minimize MW. This paper also encouraged the engineers to add an option to select sinusoidal flow waveform in VCV mode to lower MW when tidal volume is l lit or higher.

Declarations

- Ethics approval and consent to participate

  Not applicable

- Consent for publication

  Not applicable

- Availability of data and materials

  Not applicable

- Competing interests

The authors declare that they have no competing interests.

- Funding

  Not applicable

- Authors' contributions

  Mojdeh Monjezi perform the fluid analysis and is a major contributor in writing the manuscript. Hamidreza Jamaati designated and co-wrote the manuscript. All authors read and approved the final manuscript

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- Authors' information

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