Damping of the dynamic pressure amplitude in the ventilatory circuit during high-frequency oscillatory ventilation

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Abstract

The study deals with experimental measuring of attenuation of dynamic pressure during high-frequency oscillatory ventilation. The experimental circuit consists of high-frequency oscillatory ventilator Sensormedics 3100 B, patient circuit and lung model 5600i. Different values of the flow resistance and alveolar compliance were modelled during the experimental measurement. The ventilatory parameters of the oscillatory ventilator were constant during whole measurement. We confirm a decrease of the amplitude of the pressure swings in the alveolar space during high-frequency oscillatory ventilation. Dependence on the mechanical properties of the respiratory system was found.

1 Introduction

Artificial lung ventilation (ALV) is used in many patients when spontaneous breathing is insufficient or intentionally inhibited. ALV is the most efficient method for treatment of acute respiratory failure. However, there are still strong adverse effects of ALV upon patient’s respiratory system [1]. The adverse effects caused by ALV are known as ventilator induced lung injury. Current trend is to minimize the adverse effects of ALV in the respiratory care and new ventilatory regimens are introduced in order to ensure protective ventilation.

One of these new techniques is high-frequency oscillatory ventilation (HFOV). Frequency of HFOV is higher compared with conventional artificial lung ventilation (CV) and the frequency rate is in range 3–25 Hz. Higher ventilatory frequency allows use of significantly smaller tidal volumes, which are comparable to anatomical dead space or even smaller. Also the pressure swings (∆P) are decreased in the distal parts of the ventilatory circuit during HFOV in comparison with CV. The pressure swings are superimposed on the mean airway pressure (MAP) that inflates the lung. Use of MAP and smaller pressure swings during HFOV prevent the lungs from overdistension, barotrauma and volutrauma [2]. The difference in the frequency, pressure and tidal volume represent the most significant difference between HFOV and CV.

It was shown in animal experiments that amplitude of the pressure swings is damped during HFOV. The alveolar pressure was measured by a capsule that was placed into the lung. The amplitude of the pressure swings was rapidly decreasing in the airways and only about 10 % of the input pressure was present in the alveolar space. The results were affected by the weight of the capsule and the placement of the capsule in the alveolar space [3, 4].

It was showed that MAP is stable with different ventilator parameters and remains on the same level in all parts of the respiratory system [5].

The attenuation of the pressure swings was also studied by the mathematical model [6]. The simulations conducted on the model confirm the significant decrease of the amplitude of the pressure swings during HFOV. Less than 10 % of the input pressure amplitude is present in the alveolar space. The result of simulation conducted on the mathematical model can be seen in Image 1. Ratio of amplitude of the pressure swings for all generations of the bronchial tree to the pressure amplitude in the airway opening is simulated.

Image 1 Amplitude of the pressure swings for CV (0,25 Hz) and HFOV (5 Hz) in the bronchial tree [6].

The simulations conducted on the mathematical model of the respiratory system with endotracheal tube shows that intrapulmonary parameters depend on the pulmonary mechanics [7].

ALV is used in many patients that suffer by pulmonary disease. Diseases change the pulmonary mechanics and it directly affects the efficiency of ALV. The difference in the parameters of the respiratory mechanics between pulmonary and extrapulmonary form of acute respiratory distress syndrome (ARDS) was described [8, 9].

The efficiency of HFOV was studied for primary (ARDS<sub>p</sub>) and extrapulmonary (ARDS<sub>exp</sub>) form of ARDS. It was shown that efficiency of HFOV is dependent on the mechanical properties of the respiratory system. The benefit

The adverse effects caused by ALV are known as...
of the patients with ARDS_{exp} was significantly higher compared with ARDS_{e} after HFOV introduction [10].

It is commonly accepted that diseases like chronic obstructive pulmonary disease, asthma and other obstructive diseases are contraindicative for HFOV [11]. The aim of the study is to experimentally confirm attenuation of the amplitude of the pressure swings during HFOV and to study the effect of mechanical properties of the respiratory system upon the attenuation of the amplitude of the pressure swings.

### 2 Methods

The experimental circuit consists of high-frequency oscillatory ventilator Sensormedics 3100 B (ViaSys Healthcare, USA) with its patient circuit and lung model 5600i (Michigan Instruments, USA). The probes for measuring of pressure were placed in different places of the ventilatory circuit according to the Image 2.

![](Image 2) Scheme of the ventilatory circuit. $R_p$ represents the flow resistance.

Probe 1 measures pressure amplitude in the inspiratory part of the patient circuit. Probe 2 measures pressure amplitude behind the $R_p$, representing the resistance of the endotracheal tube and airways. The alveolar pressure is measured by probe 3 using the external ports of the model that are connected with the bellows representing alveolar space. The pressure is measured by a special device developed to measure three pressures simultaneously in range 0-6.5 kPa. The device that is used for measuring of the pressures consists of sensors 26PC01 (Honeywell, USA) and multifunction data acquisition device Ni USB-6009 (Texas Instruments, USA). Each of the three canals was individually calibrated before experiment. Sampling rate of the device was 1 kHz. Detail description of the system for measuring of the pressure is described in [10]. LabVIEW SignalExpress was used for analyzing the signal. The peak to peak amplitude of the signal was analyzed and the attenuation of the pressure swings was computed for all combinations of the mechanical properties.

The initial setting of the ventilator was: $MAP = 10 \, \text{cmH}_2\text{O}$, bias flow $Q = 40 \, \text{L/min}$, $I:E = 1:2$ and $\Delta P = 10 \, \text{cmH}_2\text{O}$ and ventilatory frequency $f = 5 \, \text{Hz}$. These parameters were held constant during the experimental measurement. The ventilator was directly connected to the model 5600i without any endotracheal tube.

Two bellows form the alveolar space at model 5600i and the compliance of these bellows can be modified by external spring. Alveolar compliance was changed during the experiment in range between 0.1 L/cmH$_2$O and 0.025 L/cmH$_2$O.

The flow resistance was defined by the external elements that can be included in the ventilatory circuit. We used the resistances $R_p = 0, 5, 20$ and 50 cmH$_2$O/L/s. External resistances were supplied with model 5600i.

### 3 Results

Images 3-6 depict time trends of the pressures that were measured in the ventilatory circuit during the simulation of different pulmonary mechanics. Image 3 depicts the attenuation of the pressures swings in the ventilatory circuit with flow resistance $R_p = 0 \, \text{cmH}_2\text{O}/\text{L/s}$ and normal alveolar compliance $C_l = 0.1 \, \text{L/cmH}_2\text{O}$. The average value of peak to peak pressure amplitude in probe 1 was $\Delta P_1 = 1.53$ kPa and the average peak to peak pressure amplitude in probe 3 was $\Delta P_3 = 0.48$ kPa. The attenuation of $\Delta P$ between probes $P_3$ and $P_1$ was 68.68%.

![](Image 3) Attenuation of the pressure swings in the ventilatory circuit. $R_p = 0 \, \text{cmH}_2\text{O}/\text{L/s}$, $C_l = 0.1 \, \text{L/cmH}_2\text{O}$.

Image 4 depicts the attenuation of the pressure swings with flow resistance $R_p = 0 \, \text{cmH}_2\text{O}/\text{L/s}$ and decreased alveolar compliance $C_l = 0.025 \, \text{L/cmH}_2\text{O}$. The average value of the peak to peak pressure amplitude in probe 1 was $\Delta P_1 = 1.64$ kPa and the average peak to peak pressure amplitude in
probe 3 was $\Delta P_j = 0,46$ kPa. The attenuation of $\Delta P$ between probes $P_3$ and $P_1$ was 71,62%.

![Image 4](https://example.com/image4.png)

**Image 4** Attenuation of the pressure swings in the ventilator circuit. $R_p = 0$ cmH$_2$O/L/s, $C_L = 0,025$ L/cmH$_2$O.

Image 5 depicts the attenuation of the pressure swings for flow resistance $R_p = 50$ cmH$_2$O/L/s and alveolar compliance $C_L = 0,1$ L/cmH$_2$O. The average value of the peak to peak pressure amplitude in probe 1 was $\Delta P_1 = 1,22$ kPa and the average peak to peak pressure amplitude in probe 3 was $\Delta P_3 = 0,18$ kPa. The attenuation of $\Delta P$ between probes $P_3$ and $P_1$ was 85,12%.

![Image 5](https://example.com/image5.png)

**Image 5** Attenuation of the pressure swings in the ventilator circuit. $R_p = 50$ cmH$_2$O/L/s, $C_L = 0,1$ L/cmH$_2$O.

![Image 6](https://example.com/image6.png)

**Image 6** Attenuation of the pressure swings in the ventilator circuit. $R_p = 50$ cmH$_2$O/L/s, $C_L = 0,025$ L/cmH$_2$O.

Image 6 depicts the attenuation of the pressure swings for flow resistance $R_p = 50$ cmH$_2$O/L/s and alveolar compliance $C_L = 0,025$ L/cmH$_2$O. The average value of the peak to peak pressure amplitude in probe 1 was $\Delta P_1 = 1,21$ kPa and the average peak to peak pressure amplitude in probe 3 was $\Delta P_3 = 0,17$ kPa. The attenuation of $\Delta P$ between probes $P_3$ and $P_1$ was 85,83%.

All results measured during the experiment are summarized in Table 1.

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<th>$C_L$ [L/cmH$_2$O]</th>
<th>$\Delta P_j$ [kPa]</th>
<th>$\Delta P_f$ [kPa]</th>
<th>$\Delta P_j/\Delta P_f$ [%]</th>
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**4 Conclusion**

The results of the experiment confirm significant decrease of $\Delta P$ during HFOV in the distal parts of the ventilatory circuit. It is consistent with measurements from animal experiments [3, 4, 5] and also with the results from simulations conducted on the mathematical models [6, 7]. It suggests that HFOV is protective ventilator strategy because $\Delta P$ is quite smaller on the alveolar space compared with CV.

It can be seen that attenuation of $\Delta P$ is dependent on the flow resistance of the ventilatory circuit. Attenuation of $\Delta P$ increases from 68,68% to 85,12% when the flow resistance was changed from $R_p = 0$ cmH$_2$O/L/s to $R_p = 50$ cmH$_2$O/L/s for $C_L = 0,1$ L/cmH$_2$O. Similar changes can be seen in Table 1 for other values of alveolar compliance. This dependence supports that obstructive pulmonary diseases that are characterized by increased airway resistance are contraindicative to the use of HFOV. It is in concordance with [11]. Increased flow resistance decreases $\Delta P$ and also the effective alveolar ventilation will be smaller. The change of the alveolar compliance has significantly smaller effect on the attenuation of $\Delta P$ compared with $R_p$. 

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Attenuation of $ΔP$ changes from 74.46% to 78.08% when the alveolar compliance was changed from $C_l = 0.1$ L/cmH$_2$O to $C_l = 0.025$ L/cmH$_2$O for $R_p = 5$ cmH$_2$O/L/s. There was almost no effect of alveolar compliance upon the attenuation of $ΔP$ for higher values of $R_p$.

The clinical experiences show that mechanical parameters of the respiratory system differ in ARDS$_p$ and ARDS$_{exp}$ patients [8, 9]. It was also observed that efficiency of HFOV is dependent on the pulmonary mechanics [10]. These experiences are consistent with the results of our measurements that also confirm the effects of the changes of alveolar compliance and flow resistance of the airways upon the efficiency of HFOV. Similar effects have been also observed in mathematical and experimental ovine model of the immature lung [12].

The pulmonary mechanics also affects the tidal volume during HFOV because it is pressure controlled ventilatory technique. Tidal volume is not monitored during HFOV commonly and external device have to be added to the circuit to monitor the tidal volume. The tidal volume directly affects the blood gases and its monitoring allows better control of HFOV.

5 References


Acknowledgement

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