Measuring and evaluating system designed for high frequency oscillatory ventilation monitoring

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Abstract

The study deals with design and testing of a measuring and evaluating system suitable for high frequency oscillatory ventilation (HFOV). The main features of the system are the real time monitoring of HFOV including precise evaluation of ventilatory parameters including tidal volume and minute ventilation, real time modeling of the respiratory system and its mechanical parameters and a model-based computing of alveolar pressure. The study also deals with a calibration and testing of the designed monitoring system.

1 Introduction

High frequency oscillatory ventilation (HFOV) used as an alternative ventilatory strategy for treatment of cute respiratory distress syndrome (ARDS) patients offers potential advantages of lower tidal volumes and lower pressure changes in the alveolar space conducted on such a value of continuous distending pressure on which the oxygenation reaches its maximum value [1]. HFOV uses very high ventilatory frequencies between 2 to 15 Hz, what is up to 100 times higher than the physiological breathing frequencies or ventilatory frequencies used during conventional mechanical ventilation (CV). Therefore, the respiratory monitors designed for CV cannot be used during HFOV. There are not commercially available respiratory monitors suitable for HFOV monitoring.

Commercially available HFOV ventilators for clinical use measure only pressure Paw in the airways. The ventilator calculates other pressure parameters as positive end-expiratory pressure, mean airway pressure (referred to as continuous distending pressure during HFOV), peak inspiratory pressure and pressure amplitude ΔPaw. These parameters are determined mostly to assure maximum safety of HFOV for patients. Measurement of airflow Qaw in the airways, that can be used for calculation of tidal volume (V_T), does not exist in any commercially available HFOV ventilator.

The inability of proper monitoring during HFOV affects the initiation and control of HFOV, which is mostly empirical based on table values, observation of patient's chest wall movement magnitude and a frequent analysis of arterial blood gases. A long-time stability of the HFOV ventilatory mode cannot be achieved easily as any change in the mechanical parameters of the patient's respiratory system causes a change in delivered tidal volume, minute alveolar ventilation and gas exchange.

As the airway flow rate and volume parameters are not measured during HFOV, other important variables as mechanical parameters of the respiratory system (airway resistance and lung compliance) cannot be evaluated. The amplitude of alveolar pressure $\Delta Palv$ is very important as it is the pressure acting directly inside the lungs. Contrary to conventional ventilation, $\Delta Palv$ is very different from the proximal airway pressure amplitude ΔPaw during HFOV, and its magnitude depends strongly on mechanical parameters of the respiratory system. Unfortunately, none of the HFOV ventilators is able to evaluate alveolar pressure.

The aim of the study is to design and test a monitoring system suitable for HFOV that allows measurement of all common ventilatory parameters including parameters of the lung mechanics and alveolar pressure.

2 Methods

The designed monitoring system consists of electronic pressure and airflow transducers, analogue-to-digital converter and a computer (notebook) with original software for evaluation and monitoring of HFOV. The overall scheme of the system is depicted in fig. 1.

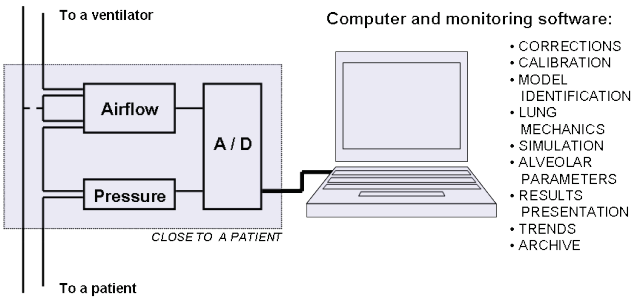

Figure 1 Scheme of the monitoring and evaluating system for HFOV.

2.1 Airflow and airway pressure measurement

Airway airflow Qaw measurement is assured by a differential pressure transducer that measures a pressure drop developed on a fixed orifice as a consequence of an existing airflow. There are two different types of a coupling con-

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taining the orifice designed for use in adult or in neonatal or pediatric patients.
The adult version of the coupling, depicted in Fig. 2, is equipped with standard medical male and female 15 mm cones so that the coupling might be inserted between the Y-piece of a ventilator circuit and an adaptor of the patient's endotracheal tube. The orifice is 0.5 mm long and has three possible diameters of 4, 6 or 8 mm so that the airflow measurement covers a wide range of the patients' weights.

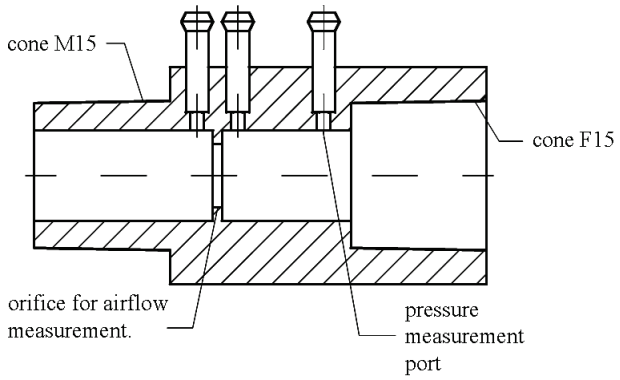

Figure 2 An adult version of the coupling for airflow and pressure measurement during HFOV.

The neonatal/pediatric version of the coupling is depicted in Fig. 3. It ends with a standard medical 15 mm male cone that fits the Y-piece of a ventilator circuit on one side and a cone for a direct attachment of the patient's endotracheal tube (ETC) on the other side. Such a solution allows minimizing the apparatus dead space that is important especially for neonatal use. The orifice is 0.5 mm long and 2 mm (for ETC 2.5 – 3) or 3 mm (for ETC 3.5 – 5) in diameter.

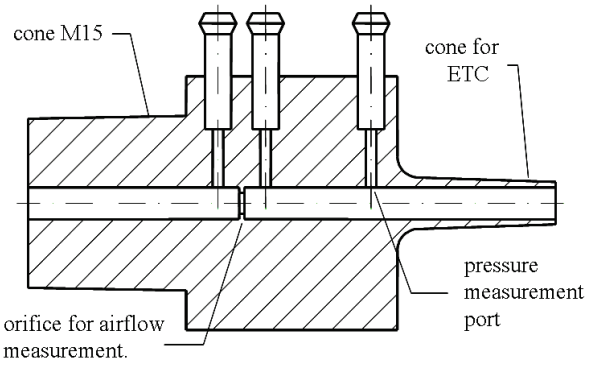

Figure 3 A neonatal/pediatric version of the coupling for airflow and pressure measurement during HFOV.

A significant nonlinearity of the fixed orifice is corrected in the computer using the following equations:

$$Qaw = a \cdot |Umeasured|^{b} \cdot \mathrm{sign}(Umeasured),$$

where *Qaw* is the airflow, *Umeasured* is the output voltage from the pressure transducer followed by an amplifier, *a* is a constant dependent mainly on the diameter of the orifice and a gain of the used voltage amplifiers, and *b* is a constant characterizing the non-linearity. The value of the *b* constant is very close to 0.5 as the structure of the coupling is a parabolic resistor used very often in equipment for respiratory care. The orifice has a symmetric response in inspirium and expirium. A minor difference between its inspiratory and expiratory pressure-volume characteristics are corrected in the computer based on its precise calibration.
The airway pressure is measured 10 mm behind the airflow measurement orifice proximally to the patient. The diameter of the measuring port is 1.2 mm. Because of a very high velocity of the gas along the pressure measurement port in the coupling an error in measured pressure develops according to the Bernoulli effect. The error is corrected in the computer using the following equation:

$$Paw = Pmeasured + c \cdot |Qaw|,$$

where: *Paw* is the corrected airway pressure, *Pmeasured* is the pressure measured by the transducer, *Qaw* is the airflow and *c* is a constant characterizing the Bernoulli effect.
A differential pressure transducer for airflow measurement and a pressure transducer for airway pressure measurement are connected to the coupling by three pieces of PVC hose 120 mm long and 2 mm in diameter. Signals from the transducers are amplified and sampled by a 12-bit analog-to-digital converter with a sampling rate of 2 500 Hz. The obtained data are sent to a computer using a serial line or USB port.
The constants *a*, *b* and *c* in the above stated equations were obtained during calibration of the system by interpolation of a measured static pressure-flow curve using the least square technique. Calibration gas mixture comprised 60% of oxygen and 40% of air at 37 °C and 100% humidity.

2.2 Modeling of the respiratory system, alveolar pressure evaluation

A special algorithm has been developed for the real-time evaluation of the lung mechanics and alveolar pressure. The only way how to determine these parameters and variables noninvasively is their computing from the courses of proximal airway pressure *Paw* and airflow *Qaw* using a model of the respiratory system. A special model (Fig. 4) of the respiratory system has been derived suitable both for conventional ventilation and HFOV [2].

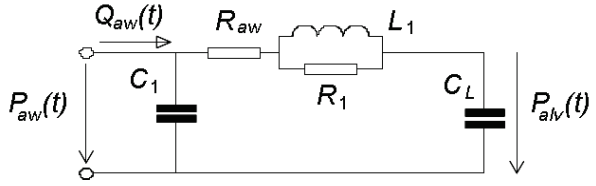

Figure 4 Model of the respiratory system.

The first part of the model represents the endotracheal tube and the bronchial tree. The second part of the model represents the lungs in the chest with a compliance as their main component.
The parameters of the model are evaluated from the courses of *Paw* and *Qaw*. The iterative method uses the Fast Fou-

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rier Transform (FFT). An auxiliary proximal pressure is computed from the proximal airflow and the model input impedance. Parameters of the model are periodically changed until the measured and the computed auxiliary proximal pressures are the same or very similar. Some of the model parameters represent important characteristics such as compliance of the lungs (C_L), airway resistance (R_{aw}) and inertance of the airways (L_1). They describe a state of the respiratory system. The input impedance of the model in frequency domain can be expressed by equation:

$$Z(\omega)=\frac{P_{aw}(\omega)}{q_{aw}(\omega)}=\frac{\left(R_{aw}+\frac{j\omega L_1\cdot R_1}{j\omega L_1+R_1}+\frac{1}{j\omega C_L}\right)\cdot\frac{1}{j\omega C_1}}{R_{aw}+\frac{j\omega L_1\cdot R_1}{j\omega L_1+R_1}+\frac{1}{j\omega C_L}+\frac{1}{j\omega C_1}}$$

where C_1 and R_1 are additional parameters representing connection of the respiratory system to the ventilato circuit, $P_{aw}(\omega)$ and $Q_{aw}(\omega)$ are the Fourier spectra of the proximal pressure and airflow. Constant j represents the imaginary unit.

The error of approximation of measured proximal airway pressure and the computed auxiliary pressure is evaluated in time domain as a sum of absolute deviations by equation:

$$\xi=\sum_{i=1}^{N}\left|P_{aw}(i)-P_{aux}(i)\right|$$

where N is number of samples of the signals and P_{aux} is the computed auxiliary pressure. During the iterative optimisation process the model parameters are being changed with the aim to reduce this error of approximation. When the error is minimal the mathematical model represents the respiratory system best and the parameters of the model are equal to the mechanical parameters of the respiratory system.

After identification of the model, the course of the alveolar pressure P_{alv} is computed using the transfer function $T_{Pq}(\omega)$ of the model:

$$T_{Pq}(\omega)=\frac{P_{alv}(\omega)}{q_{aw}(\omega)}=\frac{-\frac{1}{\omega^2 C_L C_1}}{\frac{1}{j\omega C_1}+R_{aw}+\frac{j\omega L_1\cdot R_1}{j\omega L_1+R_1}+\frac{1}{j\omega C_L}}$$

The transfer function describes relation between airflow $Q_{aw}(\omega)$ and alveolar pressure $P_{alv}(\omega)$ in the frequency domain.

2.3 Testing of the monitoring system

For verification of the accuracy of airflow and volume measurement, a special apparatus generated precisely known volumes has been constructed. The apparatus consists of a calibrated 10 or 20 mL glass syringe and a crank mechanism providing cyclical linear shifts of the syringe piston. The piston mechanism is driven by a DC motor. Scheme of the apparatus is presented in Fig. 5.

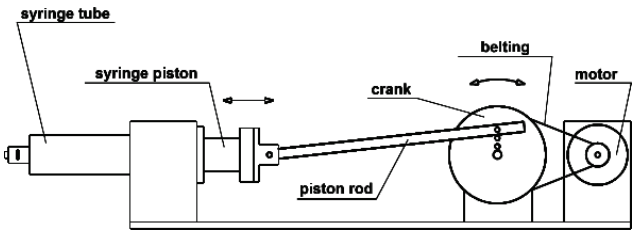

Figure 5 Scheme of the airflow and volume calibrator.

The calibrating volumes are adjustable to 3, 6 and 9 mL with the 20 mL syringe and to 1.7, 3.4 and 5.1 mL with the 10 mL syringe. Frequency of the generated waves is adjustable by the CD motor speed. Airflow signals generated by this apparatus are very close to the sinusoidal waves. The output port of the syringe is connected to the orifice coupling being calibrated.

The connected orifice represents a pneumatic resistance for the moving gas (approx. 30 kPa.s/L measured at 0.2 L/s) and the actual inner space in the syringe represents a compressible compliance. These parameters cause that the volume generated by the apparatus is slightly lower then the volume corresponding to the volume displacement of the piston. This phenomenon may be described by a differential equation which can be easily solved numerically using an iterative solving of real-gas transport process and its consequent pressure and volume changes.

3 Results

The designed monitoring and evaluating system has been constructed for use during HFOV in neonates, pediatric and adult patients. The system has been thoroughgoing tested by HFOV ventilation of physical models and also animal objects.

With the aim to increase clinical applicability of the system, an increased attention was paid to the presentation of the measured variables. The software can display all the measured and computed curves and computed values. A copy of the basic monitoring screen is presented in Fig. 6, captured during HFOV in a neonate at 10 Hz on a ventilator SensorMedics 3100A (CareFusion, Yorba Linda, CA, USA).

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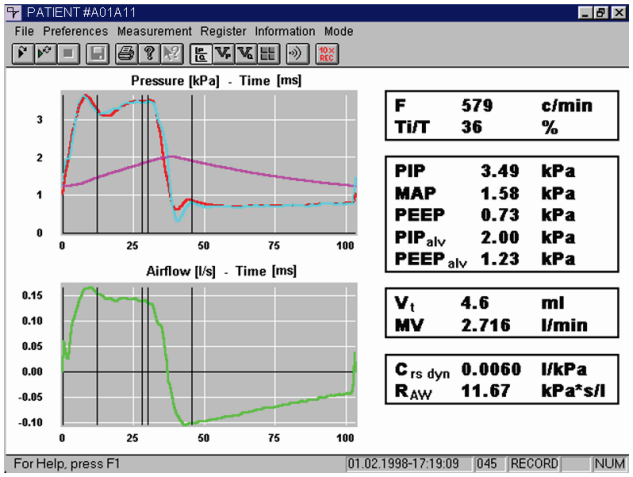

Figure 6: Basic screen of the designed respiratory monitor. Curves displayed: measured airway pressure (cyan), computed auxiliary pressure (red), alveolar pressure (purple) and airflow (green).

In the first graph field the measured proximal pressure *Paw* is displayed. A curve of the computed auxiliary pressure *Paux* is displayed behind the measured proximal pressure *Paw* so that a functioning and precision of the respiratory system modeling may be visually verified. The course of the computed alveolar pressure *Palv* is displayed in the same field. The curve of the measured airflow *Qaw* is displayed in the second graph field. Computed values characterizing ventilatory parameters are displayed in the tables on the right hand side.

The system can be used not only for HFOV, but for conventional ventilation as well. For this case additional display modes are available, for example pressure-volume loop, airflow-volume loop etc. Other capabilities of the program are possibility of visualization of history of ventilation and data archiving and export.

Accuracy of the airflow and volume measurement has been tested using the apparatus described in section 2.3. Results of the test are presented in Table 1 for tidal volumes 1.7, 3, 6, and 9 mL (this values cover all range of tidal volumes used in neonatal patients) and ventilatory frequencies 5 and 10 Hz. These adjusted tidal volumes corresponding to the working space of the piston and oscillatory frequencies are displayed in the first and the second columns. The third column shows a relative deviation of the delivered volumes computed by the numerical simulations described above caused by the resistance of the orifice and the internal compliance of the syringe. These values show that the volume generator error is very low and its maximum error was 1.26%. The next column of the table shows the volumes measured by the tested monitoring system. The values are calculated as mean values from 10 consecutive measurements. Standard deviations are less then 0.06 mL in all the cases. The last column displays a relative deviation of the monitoring system for each setting, calculated as a difference between the measured and the generated volumes divided by the generated volume and expressed in percents. The results of the test prove the precision of the volume measurement as the maximum error of volume determination is 3.75% over the whole range of possible neonatal tidal volumes.

Table 1 Results of the volume measurement accuracy tests.

Adjusted V_T (ml)	Breathing frequency (Hz)	Relative error simulated (%)	Measured V_T (ml)	Relative error of measured V_T (%)
1.7	5	0.061	1.72	1.24
1.7	10	0.24	1.74	2.60
3	5	0.26	2.88	-3.75
3	10	1.01	3.00	1.03
6	5	0.28	6.04	0.95
6	10	1.11	6.02	1.45
9	5	0.33	9.10	0.33
9	10	1.26	9.00	1.27

4 Discussion

The monitoring system, particularly the software, is not able to record and evaluate every ventilation cycle because the mathematical algorithm is quite complicated and evaluation takes some time. Consequently, the monitor records only some ventilation cycles. A typical refresh rate is between 2 and 5 s. The response time of the pressure sensors is 1 ms (f_m = 1 000 Hz) and represents the biggest time constant in the system. This constant is enough low in comparison with the maximum spectral components of the measured signals (about 400 Hz) to reproduce the signals without any frequency distortion.

5 Conclusion

The designed monitoring system is suitable for HFOV monitoring in neonates, pediatric and adult patients wit a very good accuracy of the delivered tidal volume measurement. Continuous monitoring of HFOV parameters, evaluation of lung mechanics and computation of the alveolar pressure may help optimization and control of the HFOV ventilatory strategy.

6 Acknowledgment

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7 References

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