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A Hybrid Model of the Respiratory System

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The aim of this work is building a hybrid model of the human respiratory system which enables connecting the real clinical devices (respirators) with the computerized virtual lungs. A simulation of the artificial ventilation of lungs, with the use of the hybrid model and the Siemens Servo 900 respirator, was made. Waveforms of pressure inside the lungs, flow in the respiratory tract, and the lung volume during the simulated artificial ventilation were recorded. The compliance and resistance of the hybrid model of the respiratory system were calculated on the basis of the inspiratory pause algorithms and compared to the values set in the model. The initial tests have shown that the calculated values of the parameters differ by 20% (worst result) from the values set in the model. The model will enable the investigation of the different modes of lung ventilation, as well as educational presentation of the respirator-patient interaction.

K e y w o r d s: model of lungs, respiratory system

1. Introduction

The values of pressures, flows and volumes in a real respiratory system are usually very difficult to measure. In practice, it is possible to measure pressure and flow on the level of the patient's mouth, the oesophagal pressure, the measurements pressure on the tip of the intubation tube, or measurement of the volume of the chest [1, 2].

Some of this measurements may be invasive, and thus the possibility of their use, as well as the quality of the obtained information, is limited. To explain the physical phenomena inside the respiratory system a lot of models of human respiratory system have been built [3, 4]. They can be classified into two main groups: the physical models and the mathematical models. The physical models consist mostly

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of bellows (representing compliances) and pneumatic resistances. Such models enable measurements of flows, pressures and volumes, but their ability of changing parameters such as compliances and resistances is limited). However, a physical model can be easily connected with respirator. Main advantage of the mathematical models is their flexibility, i.e. facility of modification of the parameters and the model itself. For example, in a simple RC model the resistance and the compliance can be set to the values typical either for newborns or for adults (see Fig. 1). In the more complex model, many parameters of the respiratory system, like resistances of the trachea, main bronchi and bronchioles, compliances of the lobes, influence of gravity and others, are represented separately [5]. However, the models described previously by the authors in [4] and [5], are mathematical models. The mathematical models cannot be, unlike physical models, easily connected to a real device (respirator). The hybrid model of the respiratory system makes it possible to connect a real respirator or spirometer to a computerized virtual lungs. Our hybrid model of the respiratory system is based on a different concept (a controlled source of flow) than some commercially available breathing simulators, e.g. Hans Rudolph 1101 Series or IngMar Medical ASL 500 (which consist of mechanically driven bellows).

Hybrid models create the ability of investigation of the phenomena in patient's lungs during artificial ventilation, simulation of different pathologies, and even situations potentially dangerous for real patient, like, for example, high pressures in lungs or auto PEEP (Positive End Expiratory Pressure) impairing ventilation of the lungs. Also testing of respirators working with different respiratory modes is possible.

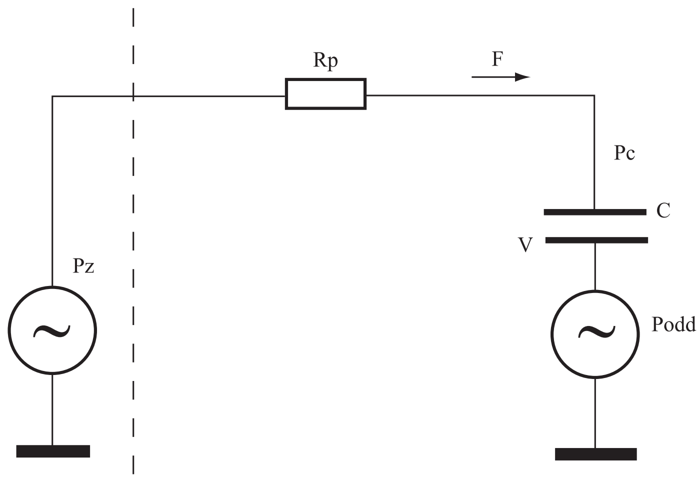

Fig. 1. The RC (resistance and compliance) model of lungs. The virtual respiratory system is represented by the compliance C and resistance Rp. Podd is the pressure generated during spontaneous breathing, V – volume of air inside the lungs, Pc – pressure inside the lungs, F – flow of air to the lungs, Pz – is the external pressure, which may be generated by the respirator

2. The Concept of the Hybrid Model of the Respiratory System

The hybrid model of the respiratory system consists of two parts: the physical part of which main part is the flow generator, and the computer part which contains mathematical, computerized model of the lungs (virtual lungs). The computer part includes also time control and input-output block. The device has the ability to collaborate with any kind of mathematical model which meets the following conditions: 1. The model works in real time. 2. The model can output the current values of the flow at the flow generator tip. The scheme of the model is presented in Fig. 2.

3. The Physical Part of the Model

The tasks of the physical part of the model are: measuring the pressure and generating the flow at the flow generator tip. The flow generator (Fig. 3) transforms the electrical, digital input signal (called the input flow) to the real flow. The pneumatic valve that is

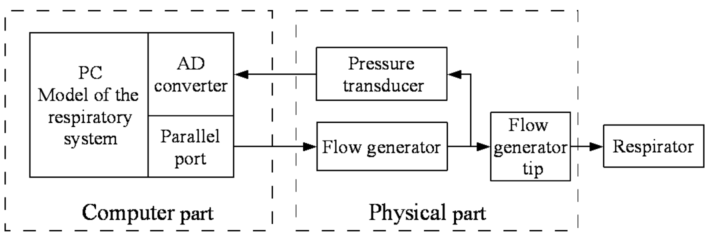

Fig. 2. Scheme of the hybrid model of the respiratory system

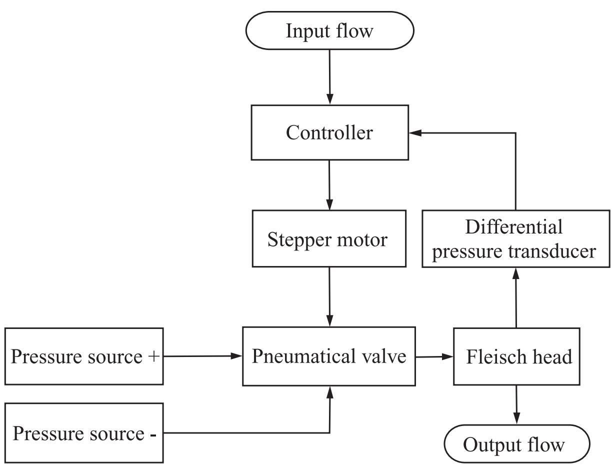

Fig. 3. The flow generator

driven by the stepper motor opens the flow supply sources (positive or negative flow). The currently generated flow depends on the position of the valve. The output flow is measured with the use of a Fleish head (a linear resistor) and differential pressure transducer (DCXL 01DN Honeywell). The flow signal is then filtered, amplified, and fed to the feedback controller built with the use of a single chip microprocessor. The on board analog to digital converter of the microprocessor is used to convert the signal. The input flow is compared with the current one. The difference between the two signals drives the controller of the stepper motor that regulates the flow in the simulator. An algorithm of a PID regulator was applied. The amplification coefficients were adjusted arbitrary after several tests. The signal of the input flow is transmitted to the controller from the PC computer (Fig. 2) via the parallel port, and parallel data bus. To reduce noise the signals on the data bus are separated by transoptors.

4. The Computer Part of the Model

The computer part of the model was based upon a PC computer. The program was written in Delphi Pascal working under the Windows operating system. The computer program for driving of the simulator consists of three main functional parts:

1. The signal input-output and time control block.
2. The computerized model of the respiratory system.
3. The user interface.

The flow diagram of the control program is shown in Fig. 4. The time the program needs to get the input signals, calculate the output signal from the virtual lungs, and draw the output signals on the screen is denoted as *T*calc. The time of waiting loop adjusted by the program is denoted as *Tw*. The algorithm forces a constant time (*Tw* +*T*calc) by adjusting *Tw* time depending on the actual Tcalc. Thus the cycle time of the main program loop is constant. The *T*calc time is variable because of the variable time of calculations and screen output operations. To achieve the maximal possible frequency of the main loop, the *Tw* time should be as small as possible. On the PC with Pentium 4 with 2.4 GHz, 512 MB RAM and Windows XP SP2 the typical *T*calc time was about 3 milliseconds. The (*Tw* +*T*calc) time finally chosen for tests was 5 milliseconds.

The triggering signal is generated by clock in constant time intervals. The computer performs all the calculations and input-output operations between the successive triggering signals. The *Tw* time is continuously checked and should be greater than zero. If this time would fall to the critical zero value, it indicates, that the calculations are too slow. In such case, an alarm is generated.

For the measurement of time intervals an 8253 type timer/counter was used. This chip (or compatible 8254) is present in all PC computers and contains three independent timer counters. Although all these timers are used by PC hardware, timer 2 can be used by software for different time measurement or triggering purposes [6]. It is clocked with 1.19318 MHz, the accuracy of time interval measurement is about 1 microsecond.

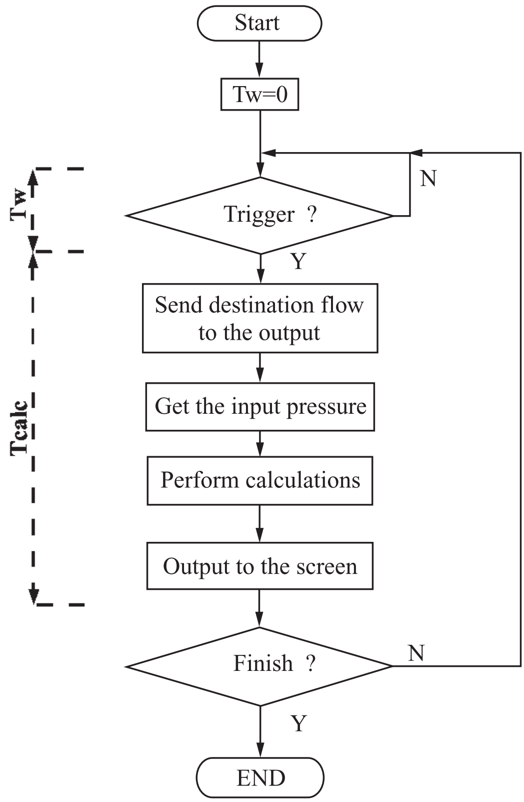

Fig. 4. The flow diagram of the control program

In the testing measurements the simplest RC model of the respiratory system was used.

The simple RC model of the respiratory system can be described with the following equations:

$$F = \frac{Pz - Pc}{R} \tag{1}$$

and

$$Pc = P_0 + \frac{1}{C}\int_0^t F(t)dt \tag{2}$$

where:

Pz – the external (input) pressure,

Pc – pressure inside the lungs,

P_0 – pressure inside the lungs at the beginning of integration,

F – flow of the air to the lungs,
R – resistance of the respiratory system,
C – compliance of the respiratory system,
t – time.

When the program starts, the values of the variables are set as follows: $F = 0$, $Pc = 0$, $P_0 = 0$. In one step of the loop the program uses the equation (1) to calculate the current flow. The flow is then numerically integrated and a new value of the Pc pressure is calculated based upon the equation (2). The programmed loop continues with calculating the F using the equation (1).

The user interface was built for controlling of the work of the program. The main screen of the user interface displays the current values of the volume of air in the lungs (V), pressure inside the lungs (Pc), external pressure (Pz), and flow of the air to (from) the lungs (F). The values are displayed on the screen as numbers and as color bars. They can be written to a disk file (together with current time). There are two sliders for setting the parameters of the RC model: the resistance R [cmH_2O*s*l^{-1}] and the compliance C [$cm\ H_2O*\ l^{-1}$] of the respiratory system. Both values are presented as numbers on the screen, above the sliders, and can be adjusted on-line (while the program is running). In the lower part of the screen there are indicators and buttons that activate calibrations, tests, and special control functions. There are two kind of calibrations: 1. Calibration of the input pressure, 2. Calibration of the generated flow. The testing functions test the generated respiratory volumes and the measured input pressure. Special control functions enable setting of the frequency of the main program loop, as well as switching off of the graphic presentation of parameters. The current value of the time of the waiting loop (Tw) is presented in a special window.

5. Test of the Model

5.1. Characteristics of the Flow Generator

The characteristics of the generator are presented in Fig. 5 and 6. The set-up for measurement of the generator characteristics were following: the input of the generator (the input flow) was connected to the D/A converter and PC computer, while the output flow of the generator was measured by the A/D converter in the same PC. A special program were created to collect the two characteristics of the generator. For the static characteristic (Fig. 5) the desired flow signal was generated in the range –100 +100 [l/ min] in steps of 5 [l/min]. For each generated input flow the output flow was measured. To measure the characteristic (Fig. 6) the sine wave signal of the input flow was generated and the measured amplitude of the output flow was divided by the amplitude of the input flow. With two different amplitudes of the input flow its frequency was changed from 0 to 50 Hz in steps of about 2 Hz. For each frequency the amplitude was measured. According to the assumptions the output flow depends linearly on the input flow (Fig. 5). The correlation coefficient between the output flow and the input flow is very high (0.99). The response times to the Heaviside (unit) step function (time between switching the step and the moment when output signal reaches 99% of the input value) is about 4 ms both with the signal of the input flow rising from 0 [l/min] to 10 [l/min] as well as with the signal falling from 0 [l/min] to 10 [l/min]. The critical frequency of the generator is about 12 Hz (Fig. 6).

5.2. Test with Respirator

The model was tested by connecting it with constant flow and volume controlled respirator Siemens Elema Servo 900. Several tests have been done with different parameters of the RC model. The settings of the respirator were following: constant

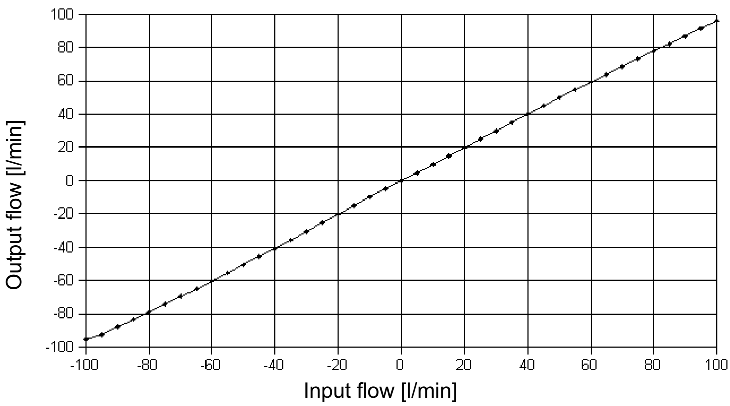

Fig. 5. The output flow of the generator as a response to the constant target flow signal in range (from –100 to 100 l/min). The coefficients of the linear regression $Fo = a*Fd + b$ are: $a = 0.981$ $b = 0.34$ $\Delta a = 0.003$ $\Delta b = 0.158$ where: Fo – output flow (dependent variable), Fd – destination flow (independent variable), Δa and Δb errors of the a and b coefficients

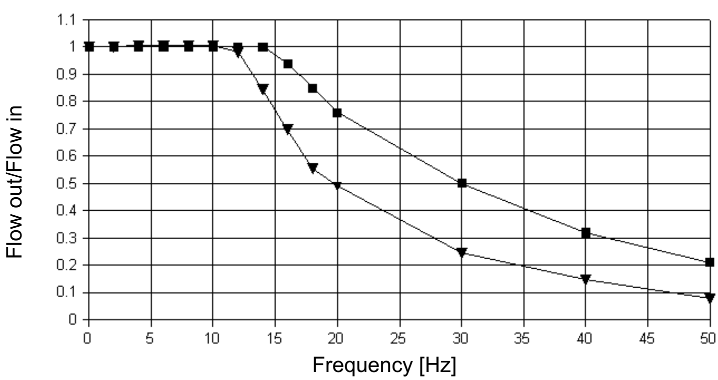

Fig 6. The output flow/target flow ratio of the generator for two different values of the amplitudes of the target flow: 30 l/min (squares) and 60 l/min (triangles). The numbers are peak to peak values of the sinusoidal signal

inspiratory flow, the respiratory frequency = 12 breaths/min, the minute ventilation 7.5 l/min, the end inspiratory pause time 0.5 s.

The values of the human respiratory system can vary in a wide range [7]: the compliance 5 ml/cmH_2O (newborn) – 200 ml/cmH_2O (adult) and the resistance 30 cmH_2O*s*l^{-1} (newborn) – 1.5 cmH_2O*s*l^{-1} (adult). The value of the airway resistance and compliance of the individual patient depends on the body weight and on the pathology of the respiratory system. For the initial tests we chose resistances R = 10 [cmH_2O*s*l^{-1}] which correspond to a healthy child or a child (or also adult) with mild lung pathology and R = 20 [cmH_2O*s*l^{-1}] which correspond to a child or adult with severe lung pathology. The compliance of the C = 0.15 [l*cmH_2O^{-1}] corresponds to an adult and the C = 0.04 [l*cmH_2O^{-1}] to a child. The above values, as well as the settings of the respirator are arbitrary (they are not a real patient's data), and were used for initial tests of the model only.

For initial tests there were chosen three sets of the model parameters shown in the columns R and C of the Table 1. For each set of the parameters of the model three time courses: the airway pressure, the respiratory flow, and the tidal volume were recorded. The time courses observed (Fig. 7) are typical for artificial ventilation of

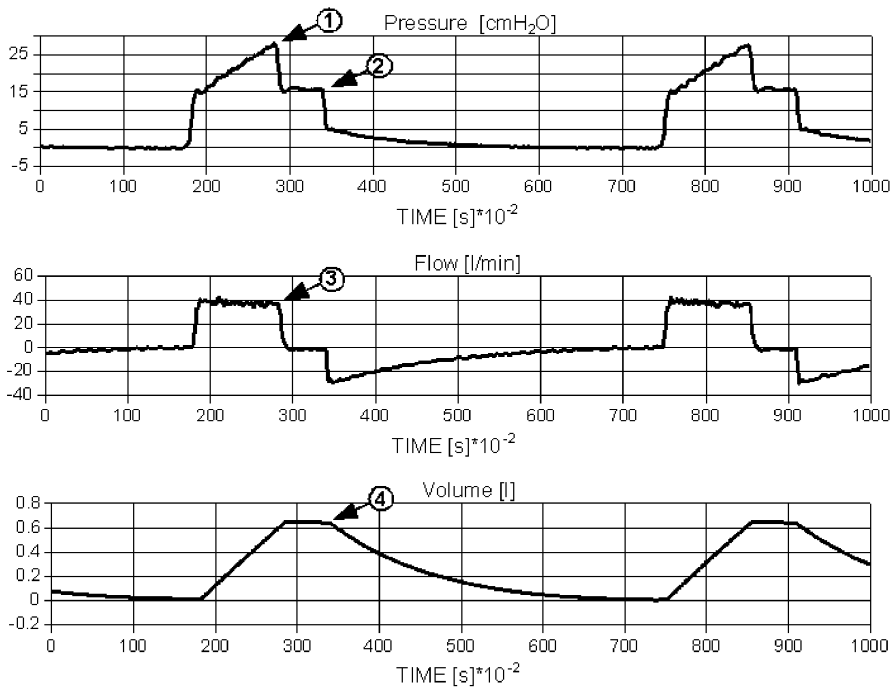

Fig. 7. Pressure, flow, and volume time courses during artificial ventilation of the RC model with the Servo 900 respirator. Parameters of the model R = 20 [cmH2O*s*l^{-1}] C = 0.04 [l*cmH_2O^{-1}]. The numbers in the circles denote: 1 – the peak inspiratory pressure (PIP), 2 – the end inspiratory pressure – (EIP), 3 – the inspiratory flow (F_i) , 4 – and the end inspiratory volume – (V_i)

the lungs or mechanical models of the lungs. An almost linear pressure increase during inspiration time is due to the constant flow pattern. Also the pressure drop during the inspiratory pause is characteristic. The characteristic points (Fig. 7) (values) can be observed: the peak inspiratory pressure – PIP (1), the end inspiratory pressure – EIP (2), the inspiratory flow – F_i (3), and the end inspiratory volume – V_i (4). The characteristic points of the pressure and flow in the time courses were chosen manually. The commonly used end inspiratory pause algorithms [8] for calculating of the resistance and compliance were applied. The results are shown in Table 1.

Table 1. Resistances and compliances measured for different settings of the RC model. *R* – set resistance, *Rm* – measured resistance, *C* – set compliance, *Cm* – measured compliance, *Rerr* – difference between the set and the measured resistances, *Cerr* – difference between the set and the measured compliances. Units for *R* and *Rm*: [cmH_2O*s*l^{-1}]; for *C* and *Cm* [$l*cmH_2O^{-1}$] *Rerr*, *Cerr* [%]

Nr	*R*	*Rm*	*C*	*Cm*	*Rerr*	*Cerr*
1	20	18.8	0.04	0.04	–1	0
2	20	22.5	0.15	0.12	12.5	–20
3	10	10.5	0.1	0.09	5	–10

6. Conclusions

1. The hybrid model of the respiratory system enables connecting of real clinical devices, like respirators or spirometers to the virtual, computerized model of lungs. Practically it is possible to use any kind of numerical model of lungs (also simulating the active patient's breathing) fulfilling the criteria of the real time work and producing appropriate data output.

2. The initial tests with the use of the real respirator (Siemens Servo 900) have shown the ability of the model to simulate the properties of the respiratory system with different parameters of the airway resistances and compliances. The flow generator is fast enough to simulate the artificial ventilation of lungs, and the output flow depends linearly on the desired flow. The resistances and compliances calculated on the basis of the inspiratory pause algorithms differ by 20% or less from the real settings of the model.

3. The model can be used for educational purposes, and investigation of different modes of lung ventilation. Using of the model will allow to avoid the *in vivo* tests, and tests that are not possible to be performed in clinics on technical or ethical reasons.

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