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The New Hybrid Pneumo-electrical Piston Model of Lungs Mechanics – Preliminary Tests

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A design principle, construction and results of preliminary tests of a new hybrid physical--electrical model of lungs mechanics has been presented. The methods leading to development of lungs models of different complexity have been also included. The basic component of the model is a voltage controlled flow source build up with a piston – cylinder system driven by a servomotor. This is used to develop a functional module playing a role of an impedance converter transforming an input electrical impedance Z_0 of any electrical network connected to its electrical terminals into a pneumatic impedance Z_{in}. Static and dynamic characteristics of the model connected to different pneumatic signal sources have been presented i.e. for the model connected with the respirator (expiration by the respiratory valve) and for the model with free unobstructed expiration. The very good dynamic features (time constant of the piston flow source less than 1ms) and a small resultant error of impedance conversion (less than 1%) enable the model to be applied in many application especially when new methods of lung ventilation are developed.

K e y w o r d s: lungs mechanics, hybrid modeling, lungs models, respiratory system tests.

1. Introduction

Physical models reproducing mechanical properties of human lungs are still an irreplaceable tool in many important applications e.g.:

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– developing of new methods of artificial or assist ventilation,
– comparative testing of medical instruments (ventilators, spirometers etc.),
– training of academic education and medical staff,
– as a bridge between *in vitro* and *in vivo* tests replacing many experiments with animals.

The classic pure physical lung *RC* models designed as a connection of pneumatic resistor *R* and capacitor *C* reproducing basic mechanical properties of human lungs i.e. its ability to accumulate potential energy (by capacitance *C*) and to dissipate energy (by resistance *R*) do not take into account complexity of physical phenomena existing in the human lungs. The lungs constitute a very complex, multi-lobe tissue structure [1, 2] enclosed in the not less complicated ribbon-muscle cage. Additionally we have to deal with the very specific lung behavior, e.g. while alveoli recruitment, and with nonlinear [3, 2] static and dynamic interdependence of the lung parameters describing their mechanical features.

It is clearly seen that only mathematical models [1, 2, 4] applied as computer programs [5] may be used to describe such a complicated physical structure. Nevertheless the computer model of lungs will suffer from the one inborn disadvantage – it can not be directly interconnected to physical medical devices and instruments working in a different signal environments: electrical – of the computer and pneumatic – of the lungs.

The hybrid pneumo-electrical lungs' model developed in the IBBE [6] resolves that interfacing problem. It is made as a pure analog system however the applied interfacing idea can be extended to pneumo-numerical (computer) hybrid models. It has been realized by means of functional subcircuits playing a role of the proportional impedance converters. They have two electrical terminals connected to electrical RLC network and two pneumatic terminals (one actual and one symbolic "mass" terminal short-circuited to the atmosphere) where medical devices can be connected to. In the IBBE two different design approaches are being developed: with the voltage controlled piston flow source and with the valve flow source.

Growing needs of lungs' model performance stimulate replacement of existing pure mechanical lungs' models by new hybrid (pneumo-numerical) models lately offered as market products [7] enabling users to include some important lungs' nonlinearities by simple computer programs manipulations, however in the frame of a very limited RC structure of the lungs' model.

2. Materials and Methods

A typical laboratory setup is shown in Fig. 1 where the respirator is connected to the physical model representing mechanical properties of the patient's lungs. The model may be pure pneumatic or hybrid pneumo-electrical. The second one is presented in the paper.

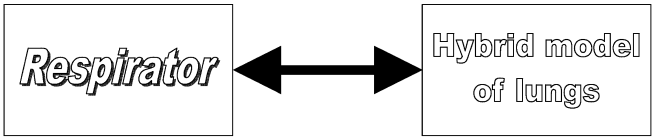


Fig. 1. A typical laboratory setup

A method of designing of hybrid models developed in the IBBE [3,8] is presented in Fig. 2b. It is assumed that one has at his disposal a special four terminal module i.e. impedance converter *TR* able to convert proportionally an electrical impedance (of any origin) connected to pair of the electrical terminals with voltage u and current i variables into a pneumatic impedance observed in pair of pneumatic terminals with pressure p and flow q variables (one the terminal plays a role of a symbolic mass). The input pneumatic impedance Z_{in} is expected to be proportional to electrical impedance Z_0 placed between the electrical terminals (Fig. 2a). This converting module can be built up using voltage controlled electrical and pneumatic sources. In the exemplary design presented in Fig. 2a these sources are the voltage controlled flow source *VCFS* and the voltage controlled voltage source *VCVS*. If the following relations are fulfilled

$$q = k_i \cdot i; \quad u = -k_p \cdot p \tag{1}$$

we obtain the needed final result i.e.

$$Z_{in} = \frac{p}{q} = \frac{1}{k_i \cdot k_p} \cdot \frac{u}{i} = \frac{1}{k_i \cdot k_p} \cdot Z_0 \tag{2}$$

where:

k_i, k_p – proportionality factors,

u, i – voltage and current at the electrical terminals of *TR*,

p, q – pressure and flow at the pneumatic terminals of *TR*.

Z_0 represents an input impedance of any electrical network (containing linear and nonlinear elements, voltage and current sources, etc.) reproducing static and dynamic properties of lungs transformed to the pneumatic terminals of the impedance converter (transformer).

The impedance converter, shown in Fig. 2a contains two important components: pressure transducer *p/u* measuring pressure drop p between pneumatic terminals and current to voltage transducer *i/u* measuring current i in electrical terminals. Electrical output signal u_p proportional to pressure p controls output voltage u of voltage source *VCVS*. Similarly current i controls flow q delivered by flow source *VCFS*. So, the required relations given by equations (1) are fulfilled.

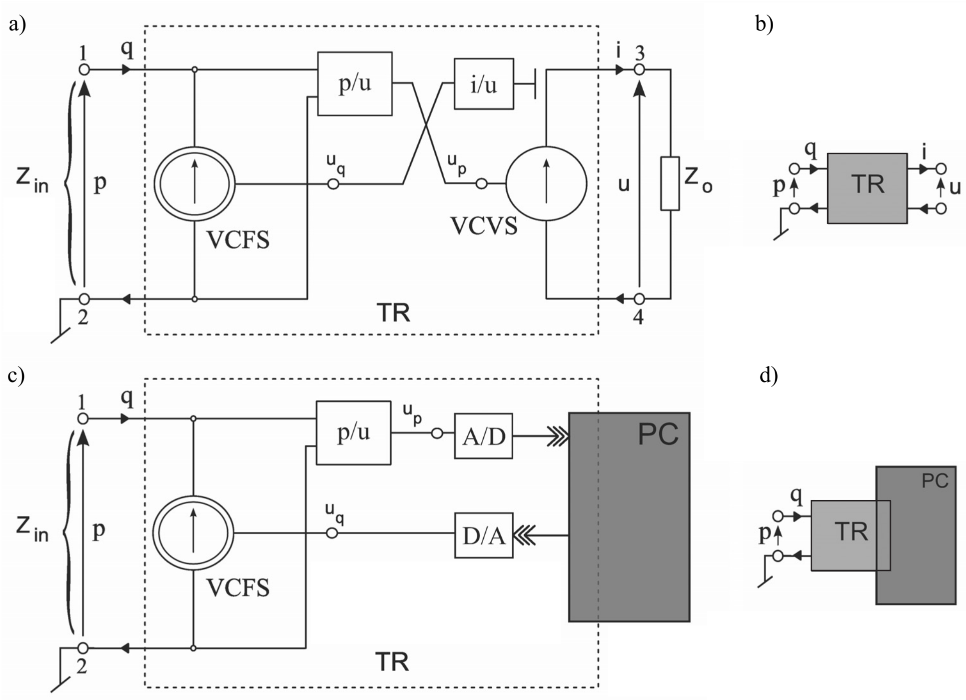


Fig. 2. Impedance transformer (description in the text)

Three examples of lungs' mechanics models presented as its electrical analogs are shown in Fig. 3. The *RC* model in Fig. 3a is the most popular in biomedical applications constituting a basis for the routine respirators assessment. Dubois model [4] in Fig. 3b takes into account inertial properties of the lungs and the chest (inertances L_a, L_t) pointing out two compartment structure of the lungs divided into air and tissue parts. These two models are sufficiently simple to be realized as a pure analog *RLC* network. The model presented in Fig. 3c is considerably complex but "working". It was developed by T. Gólczewski (the IBBE) and includes among others a lobe structure of lungs [1, 3]. It should by pointed out that there exists more complicated models of lungs e.g. developed by Barbini et. al [9].

Such complex models may exist only as a computer program and can be a part of the hybrid models of lungs when using the hybrid pneumo-numerical impedance converter presented in Fig. 2c and 2d, where analog-digital *A/D* and digital-analog *D/A* conversion is applied to interface computer *PC* with the analog pneumo-electrical part of the impedance converter. The digital output of *A/D* proportional to u_p, being an input for mathematical equations (solved by *PC*) plays the same role as the output voltage *u* of *VCVS* from Fig. 2a. A solution of the set of equations describing the *RLC* network gives the current which is converted by the *D/A* converter into voltage u_q similarly as in the pure analog case from Fig. 2a.

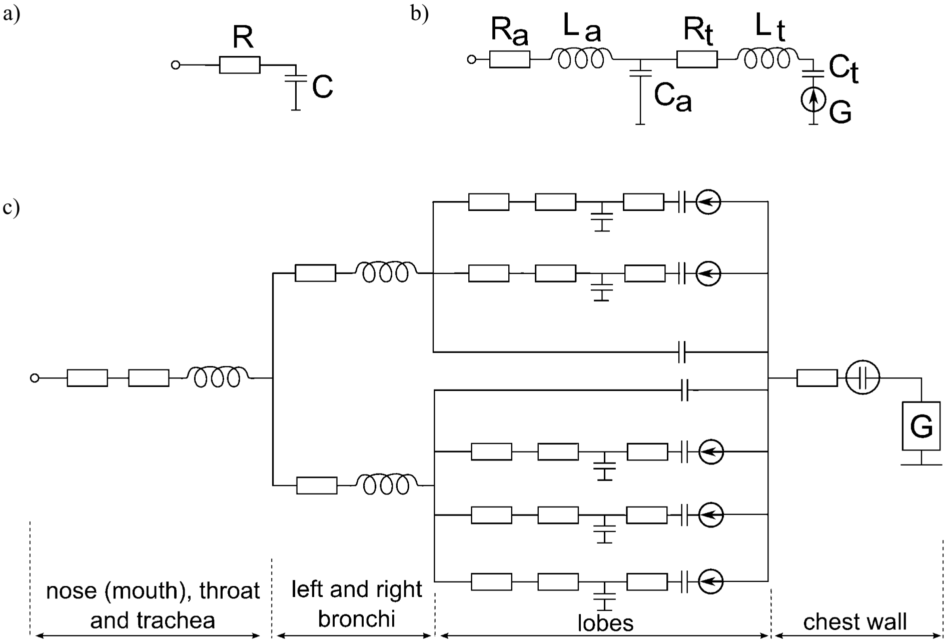


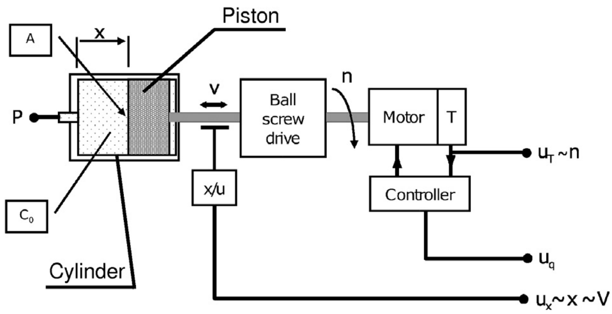
Fig. 3. Examples of lung models presented as its electrical analogs. a) Basic *RC* model: *R* – total lungs' resistance, *C* – total lungs' capacitance; b) DuBois model: R_a – airway resistance, L_a – airway inertance, C_a – airway capacitance (originally as alveolar gas compressibility), R_t – tissue resistance, L_t – tissue inertance, C_t – tissue capacitance, *G* – spontaneous breath generator; c) model by Gólczewski – a general structure (detailed description in [1, 3]) where the particular lumped components respresent resistances, inertances and capacitances (mostly dependent on the current flow and the lung volume); components marked with arrows represent gravitational hydrostatic forces effecting the lung lobes

This numerical procedure must be executed "in real time" what means that time needed for numerical calculation, data transfer and conversion must be negligible as compared to the shortest time constant of the *RLC* network.

These models of lung mechanics described by electrical analogy when connected to electrical terminals of the impedance converter are seen in its pneumatic terminal as a connection of equivalent pneumatic elements. Thanks to direct proportionality and linearity of the conversion the whole structure of the electrical network is preserved at the pneumatic side of the hybrid model. If the model of lungs contains nonlinear elements, they are also "transformed" proportionally to the pneumatic terminal with no change of its nonlinear character.

In our experiments we applied pure linear model of lungs because in this case an analysis of the system behavior is simplest and most accurate. It is worth stressing that all time constants of the network are invariants of the conversion and in this way preserve its values "observed" in the pneumatic terminal of the impedance converter, and have to be equal to these found in the real human lungs.

a)



b)

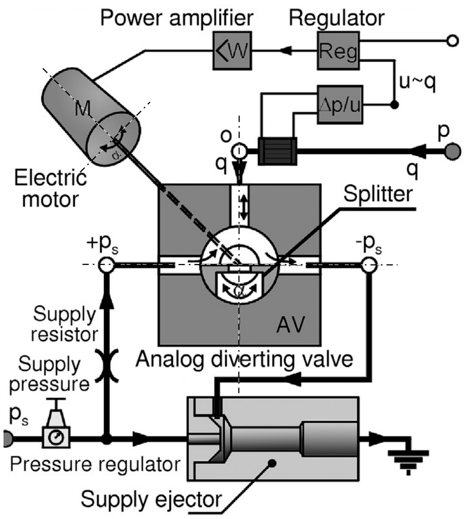


Fig. 4. Voltage controlled flow source *VCFS* realizations. a) with the piston: A – piston area; x – piston displacement; v – piston velocity; q – flow produced by piston movements ($q = A \cdot v$); n – angular motor shaft velocity; C_0 – pneumatic compliance; p – pressure in cylinder; u_T – voltage output of tacho (proportional to n); u_q – preset (input) value of flow q; u_x – voltage output of displacement transducer (proportional to V); V – volume, b) with the analog electro-pneumatic valve (description in the text)

As voltage controlled voltage source *VCVS* it is simply a voltage amplifier, we only concentrated on voltage controlled flow source *VCFS*. In Fig. 4 two different solutions are presented just to stress that there are different design possibilities. This one in Fig. 4a applied in our final design is based on the piston placed inside the cylinder and driven by the ball screw (SKF SH 12.7x12.7R) connected with *AC* motor (Maxon RE75 – tacho combination) and servo-controller (Maxon ADS50/10). Velocity v of the piston is proportional to angular velocity n of the motor shaft and proportional to voltage u_q delivered to the controller set input. Basic characteristics of *VCFS* developed by us are presented in Fig. 5. The static characteristic in Fig. 5a is fairly linear and stable while the characteristic in Fig. 5b exhibits very good

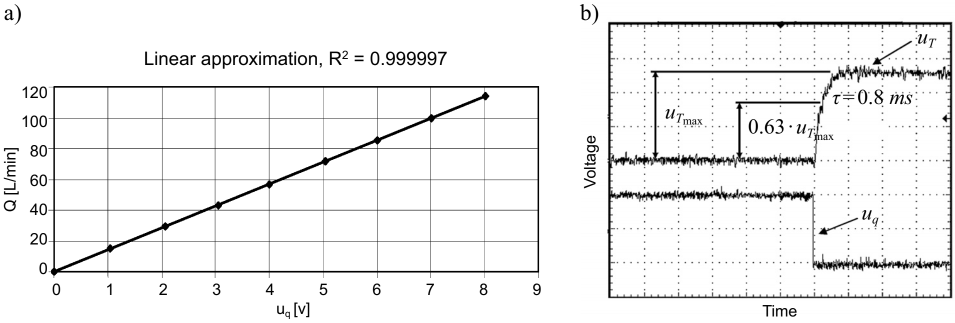


Fig. 5. Basics characteristics of the voltage controlled flow source. a) Static characteristic: Q – static flow, u_q – input voltage. b) Dynamic characteristic: u_T – tacho output voltage [0.2V/div], τ – time constant of the piston response [4ms/div] to input voltage u_q [0.5V/div]

dynamics features of the flow source. Flow Q has been calculated as a product of angular velocity n, cylinder-cross section A and the screw lead equal to 12.7 mm.

An excellent linearity of the static characteristic is achieved thanks to a very good linearity of the tacho-generator exhibiting also a very small voltage ripple (p.p < 0.7% u_T). Time constant τ of the step response in respect to input voltage u_q is less than 1ms which is much better than is needed to reproduce dynamic flow and pressure courses of the respiratory system even for a high frequency mode of lung ventilation.

Another possible solution of *VCFS* is shown in Fig. 4b. The total supply pressure p_s is converted to positive $+p_s$ and negative $-p_s$ pressure supply of the analog valve. The splitter connects or disconnects these supply sources with output channel O. So output pressure can be positive or negative dependently on angular position α of the splitter. To realize the voltage controlled flow source it is necessary to apply a control system where measured signal $u\sim q$ from the Fleish-type flow-meter (laminar head) is fed by the controller back to the motor M which rotates the valve splitter. Thus the flow q produced by the described system is proportional to input voltage u_q. A weak point of this design is connected with long term stability of the flow-meter and its limited dynamic properties.

Unlike the valve flow source, the piston flow source from Fig. 4a exhibits perfect flow accuracy. It was assessed by measurements of cylinder diameter (0.01 mm accuracy) resulting in the final piston area error less than 0.02%. This is negligible as compared to the tacho error. Similarly, an error introduced by the ball screw lead uncertainty (less than 0.005mm) can be neglected.

As the static characteristic was measured with a free unobstructed flow to the atmosphere a contribution of the shunting flow through the volume capacitor was negligible too. The only considerable source of errors is the tachometer. In our case it is less than 0.4% in the whole measuring flow range (Hübner GTL-5 tacho) with only 0.1% nonlinearity.

Now the final step might be done. According to the general concept in Fig. 2a it was sufficient to connect the piston *VCFS* from Fig. 4a with the power operational amplifier (OPA134) playing a role of the voltage controlled voltage source (*VCVS*) as shown in Fig. 6, to obtain the working hybrid model of lungs. Pressure transducer *p/u* (MPX2010) produces voltage u_p proportional to pressure p in the cylinder chamber. In fact this is an input signal for voltage controlled voltage source *VCVS* (a voltage follower with OPA134 able to measure its output current by measuring a voltage drop on resistor R). The instrumentation amplifier (INA128) produces the voltage u_q proportional to the output current of *VCVS* and delivers it to the controller of VCFS. So a needed cross-connection inside the impedance transformer is completed. The last design step was connecting of the *RLC* network (reproducing statical and dynamical properties of lungs) to the output of *VCVS*.

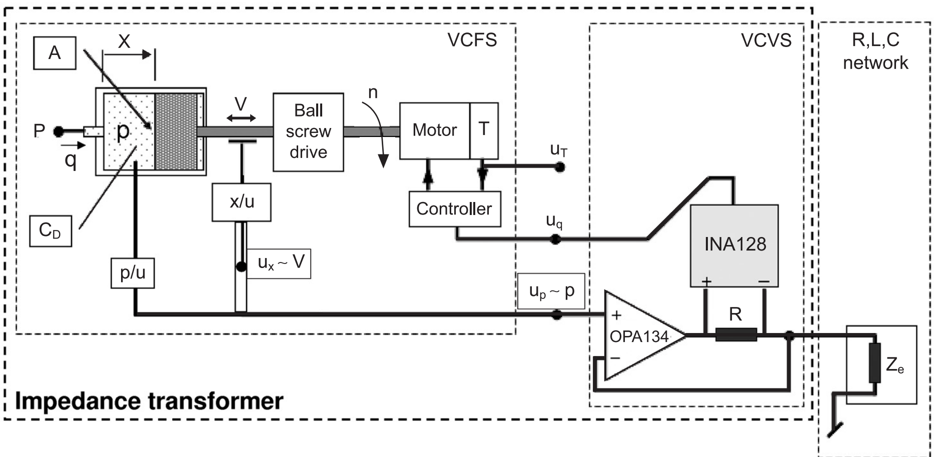


Fig. 6. The piston type pure analog proportional impedance converter-transformer: u_q – control voltage of *VCFS*, u_p – input voltage of the *RLC* network

An actual form of the chosen *RLC-like* network depends on the application. It may be reduced to the model as simple as *RC* connection presented in Fig. 3a or more complicated as in Fig. 3b where for low frequency of the conventional ventilation all inertances can be neglected what leads to the $R_aC_aR_tC_t$ model of lungs.

A photo of the circuitry of the analog hybrid model lunges (made in the IBBE) is shown in Fig. 7. In our model the piston position x has been measured directly by linear variable differential transformer *LVDT* (PELTRON PJ z 150).

One important matter must be here commented. The piston type lungs' model (Fig. 6) needs to have an initial cylinder volume V_0 to make possible a bidirectional move of the piston. It creates an initial lung capacitance C_0 that varies in time with piston movements and depends on thermo-dynamical parameters:

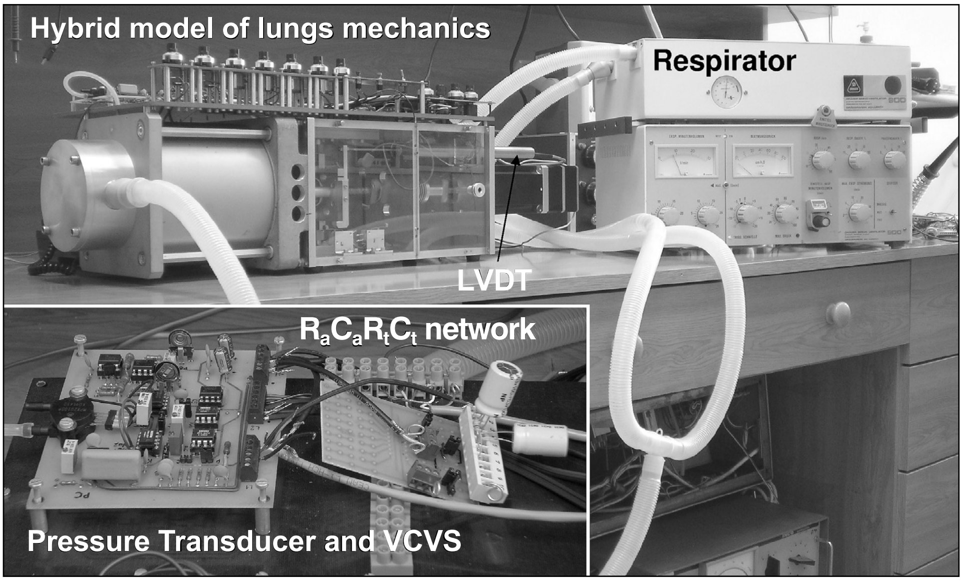


Fig. 7. A general view of the tested hybrid model

$$C_0 = \frac{V_0}{np_a} = \frac{AX_0}{np_a} \tag{3}$$

where:

n – polytrophic exponent ($1.3 < n < 1.4$),

p_a – absolute pressure in the cylinder,

A – cylinder cross-section area (2.01 dm^2),

X_0 – initial piston position (less than 0.5 dm).

A presence of the initial volume V_0 does not create any design problems as in reality we have also such an initial volume in human lungs which varies in time similarly to volume V_0 and additionally this initial volume component constitutes only few percent of the total lungs volume and can be easily taken into account by simple calculations (3).

3. Results

The main purpose of the laboratory experiments was preliminary testing of the presented piston hybrid model of lungs to examine its system performance and prove its proper functioning. Two types of networks representing lungs' structure have been chosen i.e. basic RC and $R_aC_aR_tC_t$ reduced DuBois model with inertances L_a and L_t

equal to zero what is acceptable in the case of normal low frequency mandatory or supported spontaneous ventilation (frequency < 1Hz).

Experiments were performed for two different situations:

− with the model connected to the respirator (as in Fig. 1),
− with the model exposed to forced inspiration (inflation "by mouth manoeuvre") and free unobstructed expiration to the atmosphere.

Findings on dynamic properties and stability of the model for a wide range of varying lungs' parameters and lungs' loads starting from zero load impedance (the model pneumatic terminal opened to the atmosphere) to infinite load impedance (model terminal closed) were of the first importance.

Exemplary results of the model tests are presented in two groups: one regarding resultant dynamic properties of the lungs' model (Fig. 8) and another referring to different lungs' model types and different working condition of the respirator (Fig. 10, 11, 12).

3.1. Dynamic Properties of the Hybrid Lung Model

The investigated system is presented in Fig. 8a. It contains three way electro-pneumatic valve *EV* supplied with pressure p_s and driven by the electric rectangular signal generator *G* (Tektronix AFG320). Pressure *p* and flow *q* traces have been registered in the input channel (terminal) of the hybrid model (*TR*, *RC*) and are shown in Fig. 8b. It evidences very good dynamic features of the impedance converter *TR* found as an instant flow response to a pressure excitation with no observed time delay.

a) b)

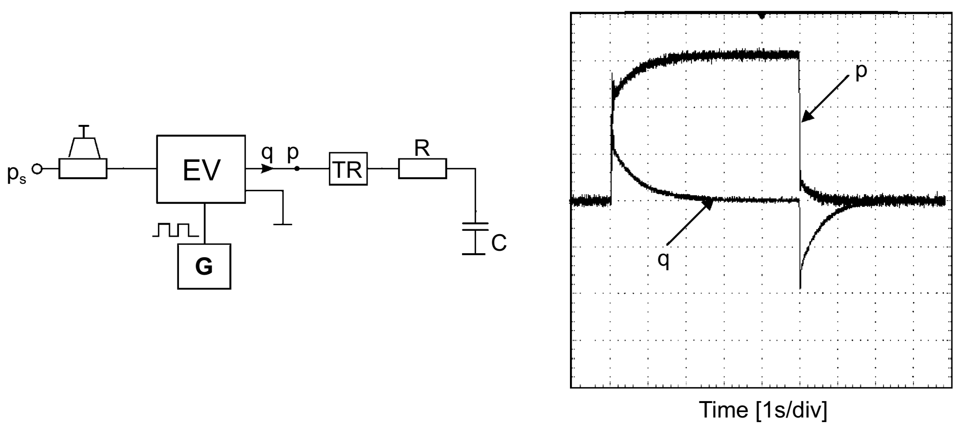


Fig. 8. Model dynamic test. a) laboratory setup, b) pressure and flow traces. p_s – supply pressure, *EV* – electro-pneumatic valve, *G* – square wave generator, *TR* – impedance converter-transformer, *R* – lungs' resistance, *C* – lungs' capacitance, *p* – pressure, *q* – flow

3.2. Respirator – lungs' Model Cooperation

A setup of the system is shown in Fig. 9. The Respirator Dräger Servo 900 working in a constant flow and constant volume mode of ventilation with the activated end-inspiratory pause was used. Two types of the electrical network have been applied in the lungs' model i.e. RC and $R_aC_aR_tC_t$ where the total resistance and the total capacitance fulfilled the following conditions to make results easier comparable:

$$R \approx R_a + R_t,\ \ C \approx C_a + C_t \tag{4}$$

Volume V (measured by the piston displacement transformer), flow q (measured by the motor tachometer) and pressure p (measured by the pressure piezoresistive transducer) have been observed and registered by a digital oscilloscope Tektronix TDS3014.

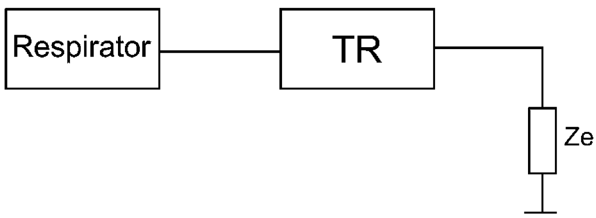


Fig. 9. Test with the respirator – hybrid model system: *TR* – analog impedance converter, Z_e – input impedance of the electrical analog lungs' model (*RLC-type* model of lungs), Respirator – Dräger Servo 900

3.3. Pressure and Flow Time Traces

Some exemplary pressure volume and flow traces (registered by the digital oscilloscope) evidencing the proper model operation are presented in Fig. 10 and Fig. 11 for the various model resistances and capacitances. The end-inspiratory pause was set at 10% and inspiration time was preset at 25% of the respiratory cycle. The characteristic end-inspiratory pressure drop is clearly visible in the pressure p traces. As the applied network is linear, a linear in time increase of volume and consequently lung pressure can be observed. Small flow oscillations, superimposed on the main flow signal, are induced by the respiratory servo-valve driven by a stepper motor. This is an inborn feature of the applied type of a servo-control feedback loop and, as a matter of fact, is of the minor importance regarding therapeutic respiratory applications.

The applied unit conversions are shown in the upper left corner of the each picture. Some characteristic features of the traces are to be pointed out. In Fig. 10 an influence of resistance R and capacitance C on input pressure is shown. Flow q traces are invariable as well as volume V traces, what should be expected in constant flow and constant volume mode of the respiratory operation. Double increase of capacitance C (Fig. 10ab) results in double decrease of the amplitude of pressure p (excluding an initial pressure jump associated with a pressure drop on R). The pressure drop accompanying the end respiratory pause activation remains the same in both pictures, as it depends on R and q maintained constant by the respirator (q) and by model (R). The situation presented in Fig. 10cd is very similar but because of double increase of the model resistance R also the pause pressure drop is two times higher as compared to the previous case.

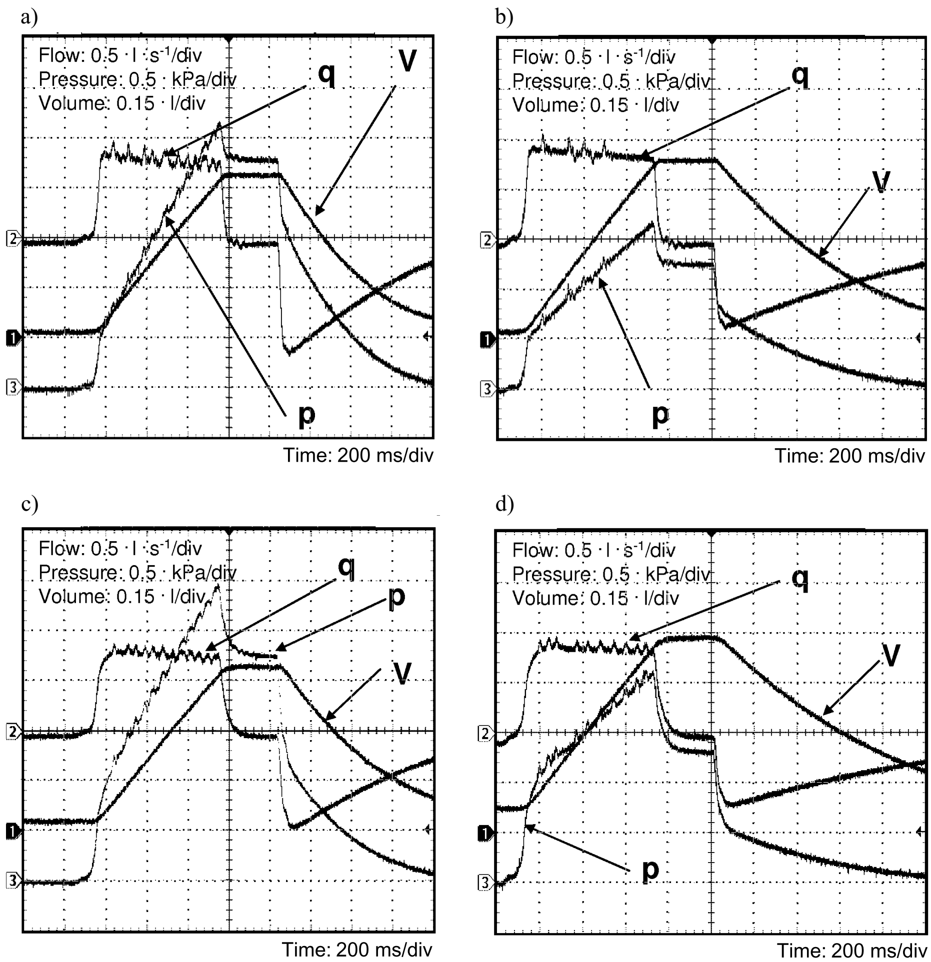


Fig. 10. Pressure, flow and volume traces registered by the digital oscilloscope in the respirator – hybrid RC model. a) $R = 1$ kΩ, $C = 97.4$ μF; b) $R = 1$ kΩ, $C = 195.4$ μF; c) $R = 2$ kΩ, $C = 97.4$ μF; d) $R = 2$ kΩ, $C = 195.4$ μF. 1, 2, 3 – zero levels

In Fig. 11 an interesting case is evidenced. The traces are obtained for the lung $R_aC_aR_tC_t$ model where the total resistance (R_a+R_t) and the total capacitance (C_a+C_t) are approximately equal to R and C values from Fig. 10 respectively. It can be easily find that in this case no substantial differences in equivalent traces especially as pressure drops are taken into account. It shown that for low frequency respiratory applications the model of lungs may in many cases be reduced to *RC* model.

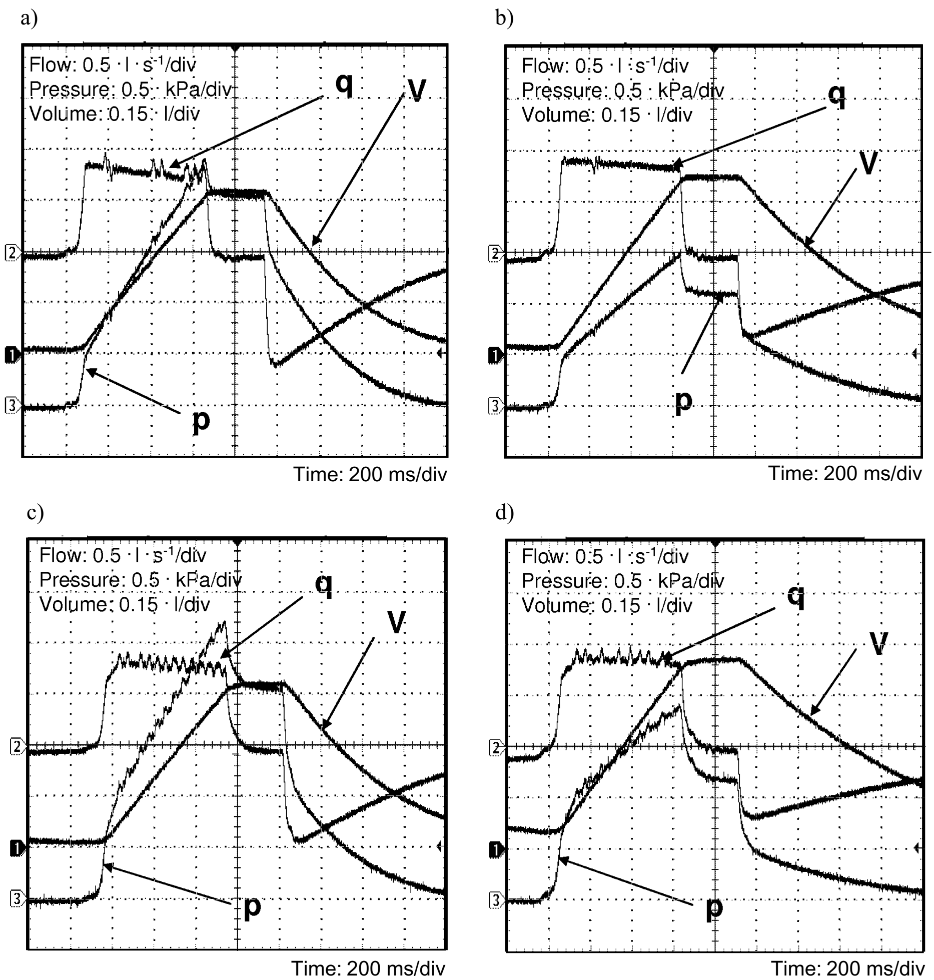


Fig. 11. Pressure, flow and volume traces registered by the digital oscilloscope in the respirator – hybrid $R_aC_aR_tC_t$ model. a) $R_a = 500\ \Omega$, $R_t = 500\ \Omega$, $C_a = 10\ \mu F$, $C_t = 98\ \mu F$; b) $R_a = 500\ \Omega$, $R_t = 500\ \Omega$, $C_a = 21.2\ \mu F$, $C_t = 195.4\ \mu F$; c) $R_a = 1\ k\Omega$, $R_t = 1\ k\Omega$, $C_a = 10\ \mu F$, $C_t = 98\ \mu F$; d) $R_a = 1\ k\Omega$, $R_t = 1\ k\Omega$, $C_a = 21.2\ \mu F$, $C_t = 195.4\ \mu F$. 1, 2, 3 – zero levels

3.4. Volume-flow Loops

Volume-flow loops are widely used in diagnostic procedures as a source of important information on mechanical lungs' parameters. They can be also used to assess mechanical parameters of the respirator. This is evidenced in Fig. 12 where a response of the linear *RC* lungs' model is compared in two different situations:

- the lungs' model is connected to the respirator (expiration by the respiratory valve) – Fig. 12a,
- the lungs' model is exposed to mouth inflation followed by an unobstructed expiration to the atmosphere (zero expiratory resistance) – Fig. 12b.

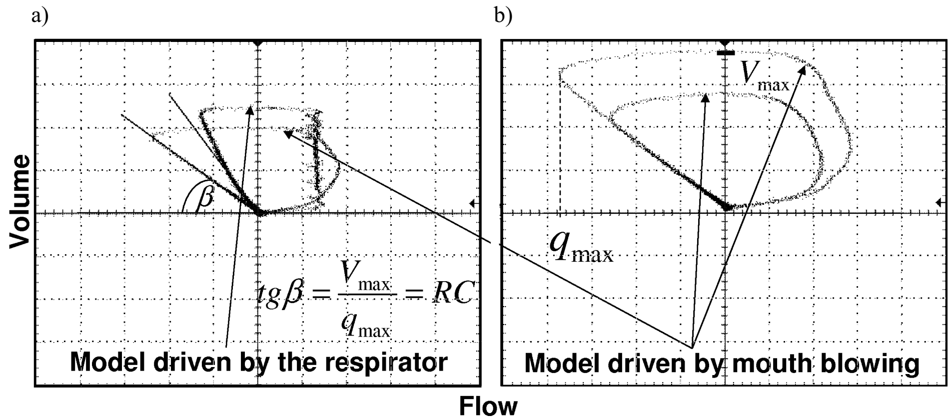


Fig. 12. Volume – flow loops as a quick qualitative assessment of the model performance

In case of the RC-type linear lungs' model for the unobstructed expiration we have

$$q = q_{\max} \cdot e^{-\frac{t}{R\cdot C}} \quad \text{and} \quad V_{\max} = q_{\max} \cdot \int_0^\infty e^{-\frac{t}{R\cdot C}} dt = q_{\max} \cdot R \cdot C \tag{5}$$

so finally

$$R \cdot C = tg\beta = \frac{V_{\max}}{q_{\max}} \tag{6}$$

where:

q_{max} – maximum expiratory flow,
V_{max} – maximum expiratory volume.

So angle β (for constant *C*) may be used as a parameter characterizing the total resistance of the system.

In Fig. 12a an influence of a nonlinear resistance of the expiratory valve can be clearly observed. The volume-flow loop obtained in the case of the model driven by the respirator is superimposed on the loop obtained for the model driven by a mouth inflation then opened to the atmosphere. We can observe the constant flow performance of the respirator (vertical part of the loop) and discharging of lungs by the nonlinear (turbulent) resistance of the respiratory valve. The valve resistance is high at the beginning and considerably lower at the end of the expiratory phase. Angle β of the line tangential to the volume-flow trace clearly shows a change of the resultant resistance of the model-respirator system.

The loops presented in Fig. 12b give good evidence of the model performance. For two different intensities of mouth inflation the expiratory trace goes to zero flow exactly along the same straight line characterized by constant resistance R of the model, as it should be expected.

4. Discussion and Conclusions

The results of model examinations presented in the previous part of the paper prove its good performance, e.g.:

– Dynamic and static properties of the hybrid model under tests are very good (Fig. 8b). Its time response to any excitation is sufficiently short to be negligible for any possible application in wide range of a respiratory frequency starting from low frequency ventilation (0.1 Hz) up to high frequency ventilation (more than 20 Hz).

– The model reproduces lung properties represented by the *RLC-type* network with practically no conversion error. The main sources of error are flow and pressure transducers but in the discussed case they are really small: 0.4% for flow (tacho error) and 0.2% for pressure measurements. So the resultant error of the impedance conversion is considerably less than 1%.

– The model is absolutely stable if loaded by an infinite resistance (closed pneumatic terminal) or zero resistance (open pneumatic terminal).

– The respirator – hybrid model cooperation does not create any problem. It may be seen analysing the pressure and flow traces presented in figures from 10 to 12. They look like a posters prepared for educational purposes illustrating a performance of the servo ventilator working under different loading conditions. Especially instructive are *V-q* loops giving information both on the lungs and the ventilator.

The hybrid model developed in the IBBE opens interesting perspectives in the field of physical modelling of mechanical parameters of lungs. Limitations on complexity of the model are connected rather with clinical identification of the model parameters [2, 9–13] than with impedance conversion from the electrical to pneumatic side of the model. Even relatively simple (e.g. 6 parameter) models of lungs' mechanics, especially if parameter nonlinearities are taken into account, have key significance as far as creation and testing of the new ventilatory algorithms is concerned. The hybrid model merges all positive features of pure mechanical models (ability to be directly connected with real medical instruments) and electrical models (an easy change of model parameters and high accuracy and stability of the impedance transformation).

The presented results of the model tests proved correctness of the developed analog impedance conversion method of synthesis of the hybrid lung mechanics models. The next step should be a hybrid (physical – numerical) model of lungs – now under development in the IBBE. It is based on the use of the hybrid impedance converter presented in Fig. 2cd. The most important nonlinearities of lung parameters will be introduced in to the hybrid physical – numerical model of lungs. There are also opportunities, nonexistent in physical models, i.e. ability to take into account cardiopulmonary interactions connected with intra-thoracic pressure and mechanisms of gas exchange and gas transport.

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