METROLOGY AND MEASUREMENT SYSTEMS
Index 330930, ISSN 0860-8229
www.metrology.pg.gda.pl

A FORWARD MODEL OF THE RESPIRATORY SYSTEM DURING AIRFLOW INTERRUPTION

Ireneusz Jabłoński, Janusz Mroczka

Wroclaw University of Technology, Chair of Electronic and Photonic Metrology, Prusa 53/55, 50-317 Wrocław, Poland
(✉ ireneusz.jablonski@pwr.wroc.pl, +48 71 320 6329, janusz.mroczka@pwr.wroc.pl)

Abstract

The paper presents a methodology of complex electrical model formulation for the respiratory system during airflow interruption. Adequacy of both structural and parametric description to the real physiological system has been taken care of. Properties of the valve-transducer unit, upper airways, bronchial tree, lung tissue chest wall and abdomen have been noted in an equivalent description of the electrical circuit. The resulting analog, combining more than 180 parameters, gives the possibility to imitate conditions of normal breathing and airflow interruption. A qualitative verification of the model has been conducted in the time and frequency domain, based on reported numerous experimental findings. The proposed linear description of the respiratory system can be the source of synthetic data for a verification of the interrupter method and for the procedure of model reduction to its identifiable form.

Keywords: respiratory mechanics, interrupter technique, complex model.

1. Introduction

The ability to perform a reliable evaluation of the respiratory system state is an actual issue due to the observed tendency in spreading of pathological cases as well as the level of advance of diagnostic tools. From the range of the techniques dedicated to respiratory mechanics estimation [1-6], the airflow interruption technique offers attractive utilitarian features, the exploitation of which can result in the construction of a portable device realizing medical tests with essentially high diagnostic significance.

The studies reported so far in this area have been based on a simple assumption that it is possible to measure the airway resistance by the estimation of an index called the interrupter resistance R_{int} [7-9]. The following analyses have pointed at limitations of that approach and, at the same time, the possibility of their reduction/elimination[10-14].

A metrological analysis of the interrupter experiment, and also the legitimacy of the proposed consecutive modification, can be conducted in the light of forward-inverse modelling and with the computer simulation methodology [15]. One of the links in such protocol is the complex model – the tool for the method's verification and for interpretation of observations typical for the occlusion manoeuvre.

The article presents the methodology of construction of a complex electrical equivalent model for the respiratory system during airflow interruption. Structural characterizations, selection of the properties and adequate quantities which define its behaviour, have been made. Simulations of the model structure proved the conformity of the behavior of the complex analog with the physiological reports on the respiratory mechanics during airflow interruption and quiet breathing. To broaden the potential of the results for future work and interpretation, both the time and frequency domain features have been investigated qualitatively.

2. Methods

The basic quantities that define the respiratory system are: resistance R, inertance L and compliance C (inversely proportional to elastance), often interpreted in the context of the electrical quantities R, L, C and the same circuit RLC. Modelling of the respiratory system concerns typically three essential structural areas: upper airways, bronchial tree and tissue segment.

2.1. Analog of the respiratory system during airflow interruption

2.1.1. Model of the upper airways

During investigations, the upper airways analog has been assumed as shown in Fig. 1. The resistance R_m = 109.4 Pa·s·dm^{-3} of airflow in the upper airways was described accordingly to the experimental equation [16]

$$\Delta P_m = R_m^{'} \cdot Q_m^r, \tag{1}$$

where $R_m^{'} = 1.2 \cdot 10^4$ Pa·s·dm^{-3} is a factor which scales the airflow resistance and the coefficient r=1.5÷2 results from the nonlinear character of pressure changes ΔP_m in the upper airways. The value of joint gas (C_{gm}) and upper airways wall (C_m) compliance was determined experimentally as $C_m||C_{gm}$ = 1.1·10^{-6} dm^3/Pa and its inertance L_m = 1.374 Pa·s^2·dm^{-3} was set as double value of this quantity in the description of trachea.

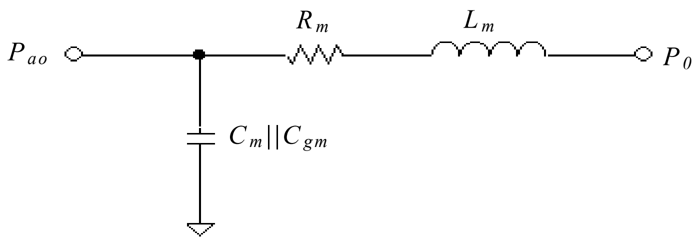


Fig. 1. Model of the upper airways.

2.1.2. Model of the bronchial tree

The starting point for the work in this section is the assumption of symmetry of the structural representation of the bronchial tree according to Weibel [17, 18]. In this frame, the model consists of 24 generations of the airways, where generation 0 corresponds to the trachea and the last 23rd generation is the model of the alveolar space. The total number N of the airways in a given i-th generation equals to 2^i. The assumption on symmetrical dychotomy of the respiratory system enables to aggregate the quantities which describe the airways tree in the following generations according to the formulas [18]

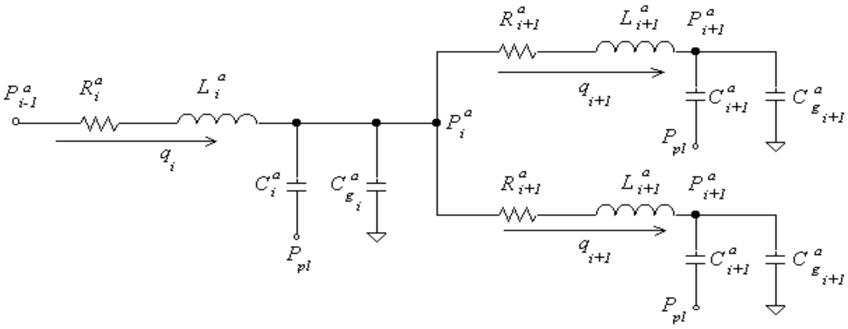
$$R_i = R_i^a / N, \tag{2}$$

$$L_i = L_i^a / N \,, \tag{3}$$

$$C_i = C_i^a \cdot N \,, \tag{4}$$

where R_i^a is an analog of resistance connected with friction during movement of a medium through the individual airway, L_i^a depicts dynamical properties resulting from inertia of the gas in the airway and C_i^a represents compliance of the airway wall. Additionally, the quantity C_{gi}^a was added to model the compression of a gas contained in a single airway of *i*-th segment (Fig. 2 and Fig. 3).

a)



b)

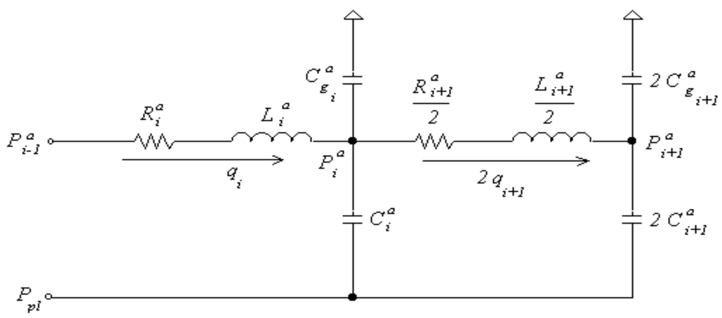


Fig. 2. a) Electrical model of an elementary airway branching. b) Airway bifurcation model according to assumption on homogeneity. q_i – airflow through the single airway of *i*-th generation; other quantities described in the text.

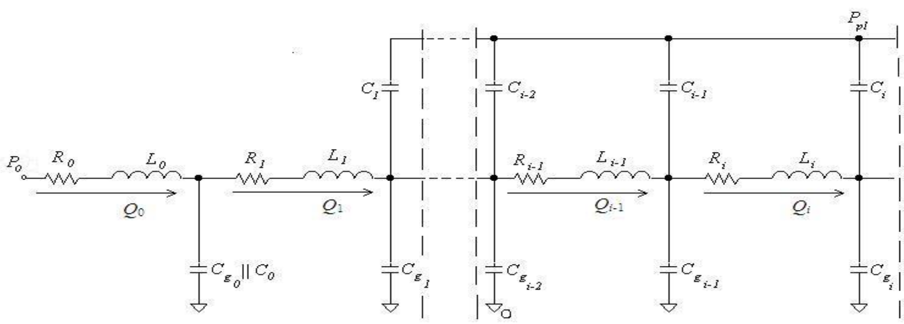


Fig. 3. Model of the bronchial tree; Q_i – airflow in the airways of *i*-th generation, other quantities described in the text.

Each of the quantities was calculated individually for the airways in the following generations, appropriately to (5) – (8); the assumptions on laminarity of the airflow in the cylindrical pipe and physiological data of single airway were used during calculations.

$$R_i^a = \frac{8\pi\mu l_i^a}{\left(A_i^a\right)^2}. \tag{5}$$

$$L_i^a = \frac{\rho}{A_i^a} \cdot l_i^a. \tag{6}$$

$$C_i^a = \frac{\Delta V_i^a}{\Delta P_i^a}. \tag{7}$$

$$C_{gi}^a = \frac{V_i^a}{\left(P_0 - P_{H_2O}^{100\%}\right) \cdot \gamma}, \tag{8}$$

where: l_i^a, A_i^a and V_i^a are length, cross-sectional area and volume of the airway in the *i*-th generation, respectively, ΔP_i^a quantifies the change of pressure in a single airway of *i*-th generation, gas viscosity μ = 18.88·10^{-6} kg·m^{-1}·s^{-1} and density ρ = 1.14 kg·m^{-3} (for the temperature of 37 °C), atmospheric pressure P_0 = 101325 Pa, $P_{H_2O}^{100\%}$ = 6276 Pa – partial pressure of saturated vapour in the air and the coefficient γ equals 1 or 1.4 for an isothermal and adiabatic process (γ = 1 was used in the model), respectively. Distributions of R_i, L_i, C_i and C_{gi} in the bronchial tree were presented in Fig. 4 – Fig. 7; irregularity in the range of 3-rd and 4-th generation, to some extent, is a consequence of approximation of the respiratory duct morphometry [18, 19, 20].

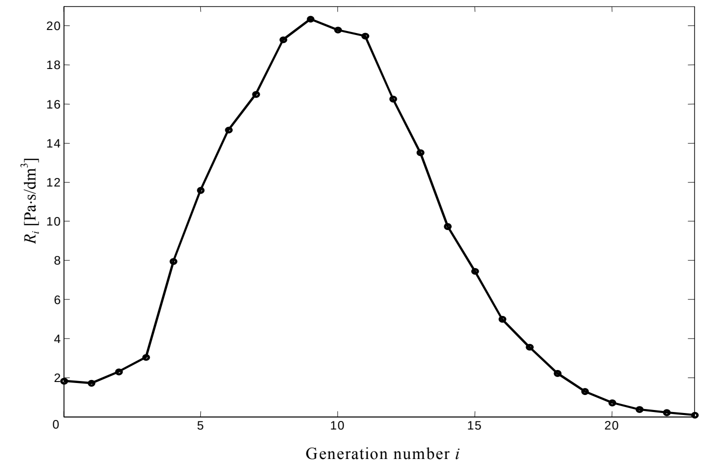


Fig. 4. Resistance R_i in the following generations of the airways tree.

The terminal structures in the bronchial tree are the alveoli. Compressibility of the gas contained in them can be calculated as

$$C_A = \frac{TGV - DS}{\left(P_0 - P_{H_2O}^{100\%}\right) \cdot \gamma}, \tag{9}$$

where TGV means total gas volume (according to [21] it is ~3.5 dm^3) and $DS = \sum_{i=0}^{23} V_i = 0.180\,\mathrm{dm}^3$ is a dead space.

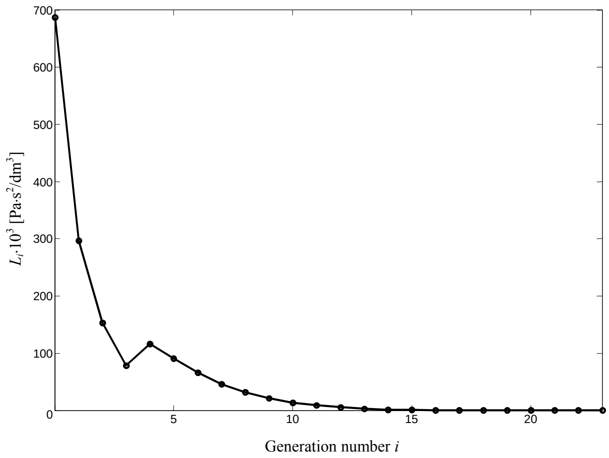


Fig. 5. Inertance L_i along the bronchial tree.

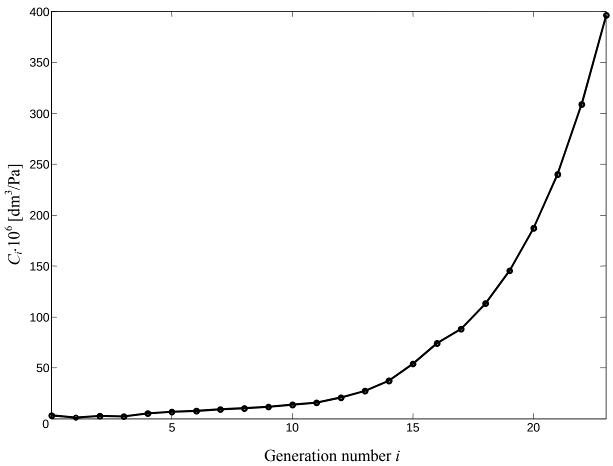


Fig. 6. Airway wall compliance in the i-th generations of the bronchial tree.

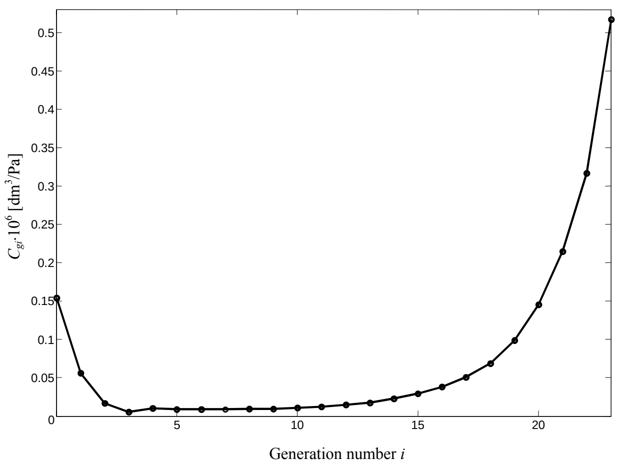


Fig. 7. Compliance of the gas in the i-th generation of the bronchial tree.

2.1.3. Model of lung tissue

The properties of lung tissue are described by three parameters: resistance, inertance and compliance, with their electrical analogs R_l, L_l and C_l, respectively (Fig. 8). The elements are placed between the two potentials different in value: P_{pl} – intrapleural pressure and P_A – alveolar pressure, adequately to the physiological structure configuration. In the model, resistance R_l represents both friction forces (called non-elastic resistance) during tissue movements and elastic resistance connected with its stretching. Inertance L_l is proportional to the acceleration of mass of tissue. The exact parametric description was deduced by studying [21-24]: $R_l = 21.3$ Pa·s·dm^{-3}, $L_l = 0.0638$ Pa·s^2·dm^{-3} and $C_l = 2.9\cdot10^{-3}$ dm^3/Pa.

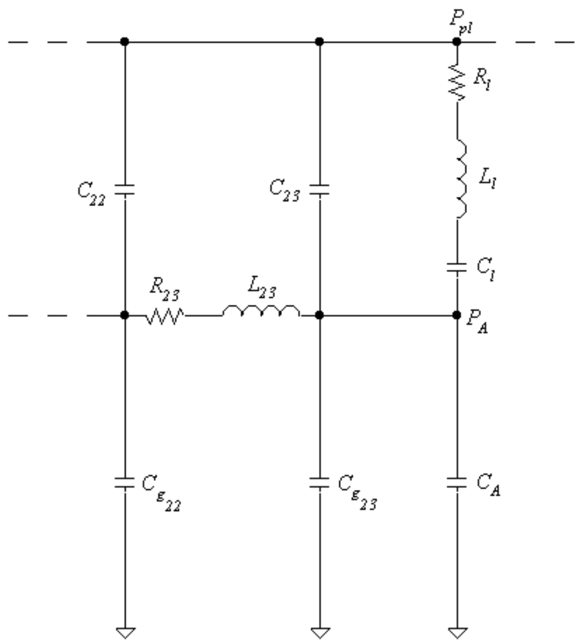


Fig. 8. Electrical analog of lung tissue – inside the dashed contour.

2.1.4. Model of chest wall and abdomen

The electrical representation of chest wall and abdomen characterizes nonhomogeneity, similarly as was stated in [23] (Fig. 9). The physical interpretation of the mechanical description with the values R, L and C is analogous to the lung tissue segment. Additionally, the source P_t = -550 Pa has been introduced to the description which points at occurrence of a pressure component, connected with the tendency of the chest wall to the elastic recoil between inspiration and expiration – the plastoelastic forces of retraction [21]. Respiratory muscle activity was depicted by a voltage source whose characteristics can be formally written as

$$P_m = P_1 + (P_2 - P_1)\cdot\left[e1(t - t_{d1}, t_{c1}) - e1(t - t_{d2}, t_{c2})\right]$$
$$e1(x,\tau) = \left[1 - \exp\left(-\frac{x}{\tau}\right)\right]\cdot\mathbf{1}(x), \tag{10}$$

where:
- t – time,
- P_1 = 0 Pa – initial value of pressure,
- P_2 = -325 Pa – value o pressure after “step”,
- t_{d1} = 0 s – delay time to the falling slope,

– t_{c1} = 0.3 s – time constant for the falling slope,
– t_{d2} = 1.3 s – delay time to the rising slope,
– t_{c2} = 0.07 s – time constant for the rising slope.

Values of the parameters were fixed during computer simulations and according to the suggestions of Lutchen *et. al*. [25]. An example of the time trend for P_m is presented in Fig. 10. Frequency of the signal is ~0.25 Hz and the proportion between duration time of the inspiration and expiration phase $t_I/t_E \approx ½$.

A parametric description of the chest wall-abdomen segment was collected in Table 1. Proportions between the parameters were set according to [21, 23].

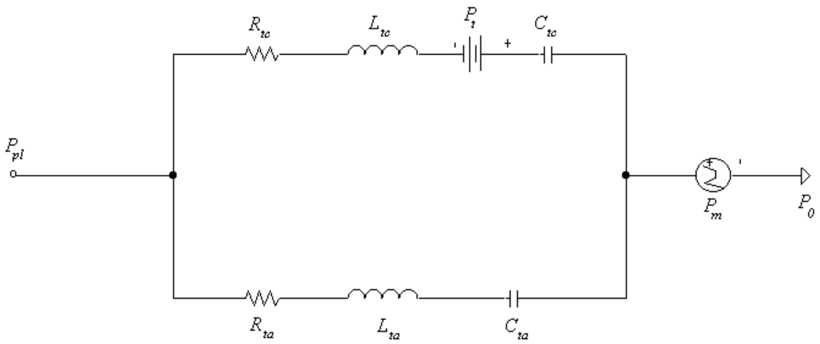


Fig. 9. Nonhomogeneous model of chest wall and abdomen: R_{ta}, L_{ta}, C_{ta} – resistance, inertance and compliance of abdomen tissue, R_{tc}, L_{tc}, C_{tc} – resistance, inertance and compliance of chest wall, P_t – source-analogon to the retractive forces, P_m – source which model the respiratory muscle work.

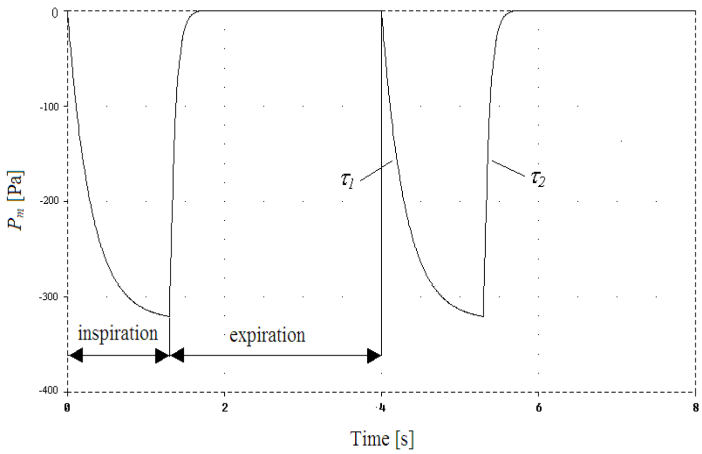


Fig. 10. Time characteristics of the respiratory muscle work; τ_1 and τ_2 – time constant of the inspiratory and expiratory exponential trend, respectively [25].

2.2. Simulational complex model of the respiratory system for the interrupter technique

Simulations of the electrical structure of the complex model for the respiratory system during occlusional experiment were conducted in PSpice. To take into account the conditions in which the model verification has been made, analogs of the interrupter valve (switch *Sbreak* (Fig. 11)) – R_{Sw} resistance (see Eq. (11)) – and an equivalent of the transducers unit (constant resistance R_p) have been additionally introduced in the simulational complex model.

$$R_{Sw} = \begin{cases} R_{open} + R_p \approx R_{open} & Sw \text{ in position } 1 \\ R_p & Sw \text{ in position } 0 \end{cases}, \tag{11}$$

where:

- R_{Sw} – total resistance of the valve-transducer unit,
- $R_{open} \approx 10^6$ Pa·s·dm^{-3} – air-flow resistance through the open electrical switch (valve is closed),
- R_p – resistance of the flow transducer and the open valve.

The task of the interrupter valve (electrical switch *Sbreak* in PSpice convention) is to occlude for about 100 ms the airflow at the mouth at the moment when the expiratory flow (Q_{ao}) is near its maximum (Fig. 12). This idea can be realized in PSpice by proper formation of the profile of the controlling source $P_{control}$.

The simulational complex electrical model of the respiratory system during airflow interruption is an assemblage of analogs of the following parts representing distinguished subsystems of the 'measurement set' of patient-equipment: valve-transducer unit, upper airways, bronchial tree, lung tissue and chest wall with abdomen. The proposed resulting structure for the interrupter technique was presented in Fig. 13. Resistance R_{ZW} introduced in the segment of the chest wall analog results from the method of circuit analysis applied in PSpice (modified method of nodal potential); there a minor influence of this element on the observed outputs was verified by computer simulations.

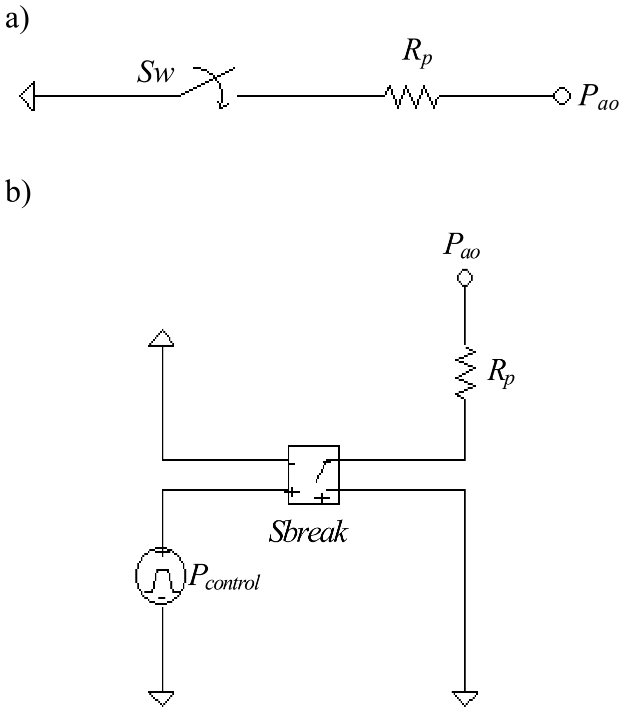
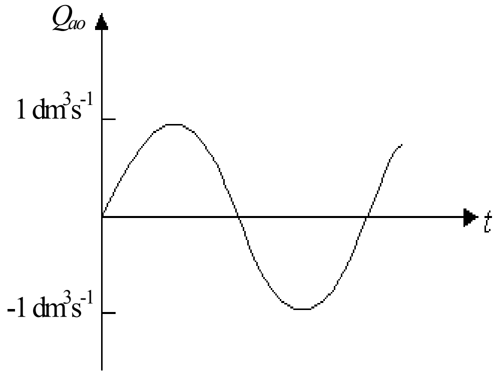


Fig. 11. a) Idea of the model of the valve-transducer unit for implementation in PSpice. b) Analog of the valve-transducer unit realised in PSpice.



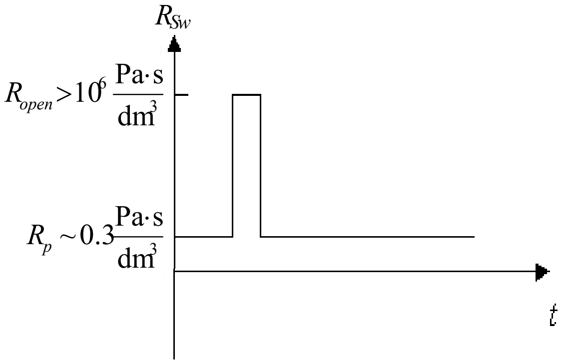


Fig. 12. Idea of synchronisation of time-course of the airflow (Q_{ao}) at the mouth and work of the occlusion valve.

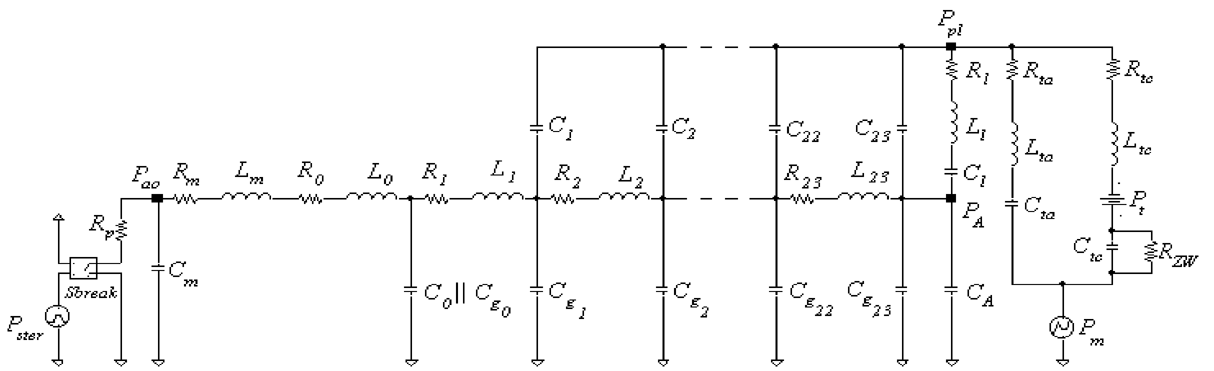


Fig. 13. Simulational complex electrical model of the respiratory system during airflow interruption.

3. Results and discussion

Verification of the complex electrical model of the respiratory system for the interrupter technique has been conducted by computer simulations of the two basic conditions of work of the real system: quiet breathing and occlusion manoeuvre.

In the case of spontaneous breathing, attention was paid to the shapes of resulting imitations of the physiological trends and whether the levels of pressures and flows in the following points of the complex structure are in agreement with the physiological ones, as well. Example plots are shown in Fig. 14 – Fig. 15. Observations have proved the adequacy of the time proportion (t_I/t_E) of the respiratory phases to the real tracings for every investigated signal. The quiet breathing regime, demanded as a measurement condition for the interrupter technique, is expressed in the physiological levels of the model outputs (*e.g.* Q_{ao}) as well as controlled quantities, difficult to measure directly in a non-invasive experiment (*e.g.* P_{pl} or P_A) [21]. Shift of P_{pl} with a constant value (Fig. 15a) is analogous to the intrapleural pressure existing in a real object [21]. Also the shape of trends recorded in the model, with characteristic quick filling of the lungs with the gas and its slower exponential breathing out, is reconstructed.

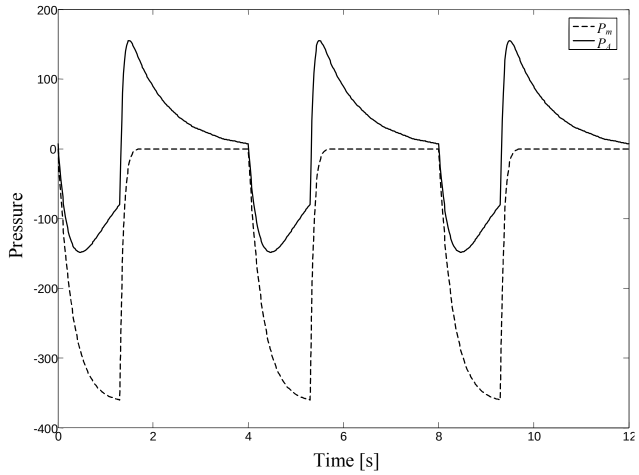


Fig. 14. Characteristics of the respiratory muscle driving pressure (P_m) and alveolar pressure (P_A) during simulations of normal breathing.

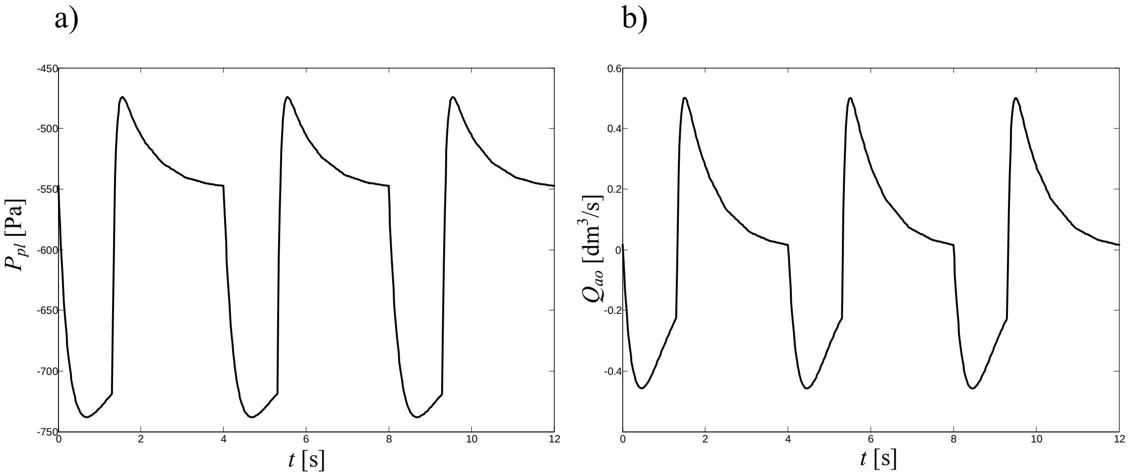


Fig. 15. a) Intrapleaural pressure (P_{pl}) and b) airway opening flow (Q_{ao}) during spontaneous ventilation.

In the model, interruption of airway opening was simulated by changing the resistance of the electrical switch (R_{Sw}) at a moment of 200 ms after the start of the expiratory phase, for the 10-th respiratory cycle. The time of occlusion was fixed at 100 ms and signals were sampled with a frequency of 1000 Hz. From the point of view of future analysis, both time and frequency characteristics are very important [10, 12, 26, 27].During the presented work they were evaluated qualitatively, but dedicated procedures of the post-interrupter data processing should be designed in the next steps.

In the time domain, interest was focused on reconstruction of the three phases of P_{ao}, specific for occlusion: rapid oscillation after valve closure, damped oscillatory transient changes (about 20-40 ms) and slow rise of pressure at the mouth in the final part of occlusion

(Fig. 16). It was also of considerable importance to control the levels of pressure (P_{ao}) and flow (Q_{ao}) signals during interruption (Fig. 17).

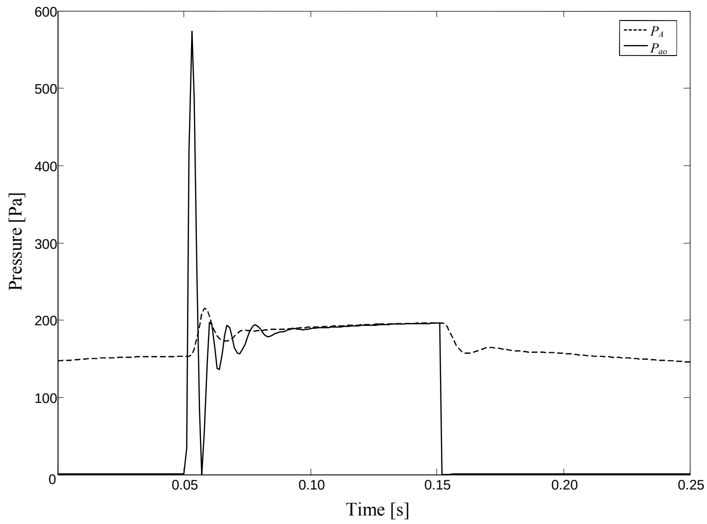


Fig. 16. Airway-opening pressure (P_{ao}) and alveolar pressure (P_A) during simulation of interruption.

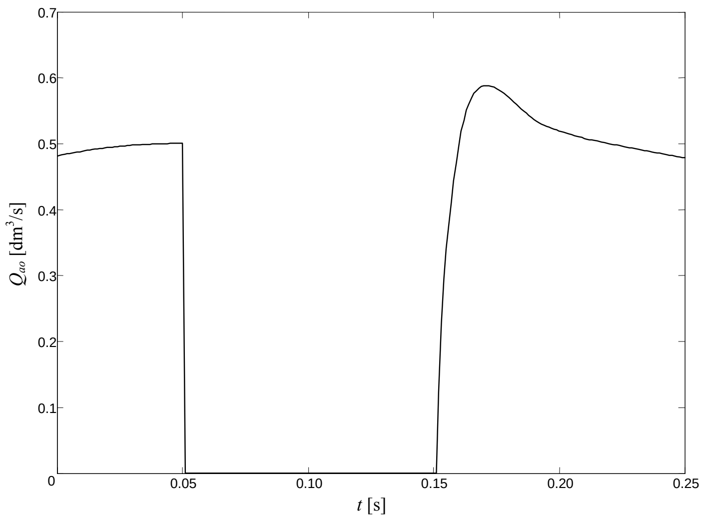


Fig. 17. Simulation of interrupter airflow at the mouth (Q_{ao}).

The proposed complex electrical model show a well-documented difference between post-interrupter pressures: at the mouth and alveolar (Fig. 16), the main component of R_{aw} overestimation by diagnostic factor R_{int} [8, 10, 28]. It is clearly exposed by the initial discrepancy between P_{ao} and P_A at the moment just after occlusion, mainly interpreted in the context of the differences between airways and alveolar resistance, and next inhomogenous pendeluft of the structural volume segments or viscoelasticity of the lung tissue.

Suggestive for investigation of interdependence between oscillatory pressure and flow transients and mechanical properties of the respiratory system can be also an analysis in the frequency domain. To date, there have been single experimental evidences of such potential relations [13,14,26], but there have not been any exact verified algorithmic solutions for post-interrupter data analysis and any advanced analysis of processes and properties of the respiratory system during the occlusion manoeuvre [12, 14]. Typically, the power spectrum density characteristics of the post-occlusional pressure (P_{ao}) signal have been reconstructed, and interesting in this area can be also the calculation of input respiratory impedance [12].

Qualitative verification of the complex electrical model during interruption was conducted on the basis of Frey's *et al.* [13, 14] procedural hints. Time changes of P_{ao} were first normalized with reference to its value at the end of occlusion and then differentiated. Written in Matlab program used such prepared samples to the power spectrum density calculation − it is possible with the *psd*(·) function (the Hanning window was used during calculations). Fig. 18 is a graphical representation of the complex model behavior in the frequency domain. There are two characteristic peaks in estimated power spectrum density of the interrupter P_{ao} signal – PSD_{int}. The first is located at about 80 Hz and second, dominant, falls near 130 Hz. These results are in agreement with experiments conducted with real structures of the respiratory systems in humans and dogs [13, 26]. There is also the evidence of existence of the two resonances in the respiratory system during interruption: the first (50-100 Hz), which reflects tissue resonance and the second (120-180 Hz), connected with the quarter-wave resonance of the gas in the respiratory tract [13]. Moreover, it is worth to note that for a finite time of valve closure (*e.g.* 20 ms) the first (tissue) peak is dominant in respiratory power spectrum density characteristics [13, 26], whereas for an infinitely short time of valve closure (as in simulations) more important is the peak at higher frequencies.

These arguments allow to accept the proposed structure of the complex model of the respiratory system during airflow interruption as a useful tool for analysis of metrological aspects of an occlusional experiment.

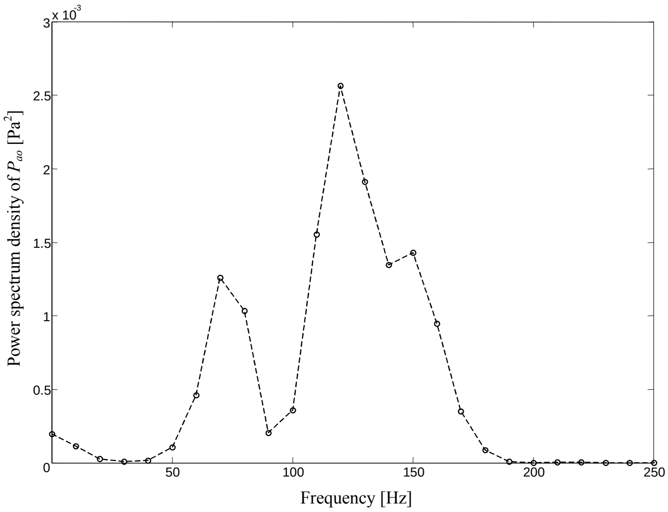


Fig. 18. Power spectrum density of occlusional pressure P_{ao} calculated in the complex model of the respiratory system.

4. Summary

Theoretical and utilitarian foundations of the airflow interruption technique make it an attractive algorithm to evaluate the respiratory mechanics [9, 10, 12, 13]. The principles of measurement have been poorly explored so far, limiting the accuracy and amount of information obtained about the object. To expand the understanding the phenomena and processes proper for pre- and postocclusional states of the measurement settings, numerous computer and real experiments should be made [14, 29].

In the paper, a complex electrical model of the respiratory system for the airflow interruption technique is proposed. It combines both the analog of the respiratory system and the equivalent of the valve-transducer unit. During formulation of the model analogies between pneumatic and electrical systems have been utilized and the knowledge about structural-morphological properties of the real object as well [18, 19, 30]. The resulting circuit consists of more than 180 parameters, being the most advanced reported representation of respiration during an interrupter experiment. Its usefulness to time- and frequency-domain investigations was proved by the computer simulations of the properties and conditions reported in literature [10, 13, 14, 18, 27]. Among other things, the described verified complexity enables future investigations of the airways-tissue interrelations in the forward-inverse experiment [15], reduced model formulation and preparation of identification procedures.

A limitation of investigations is the assumption on laminarity of the airflow in the airways. Including in the description the complex nature of different characters of flow through elastic tubes can be the next step of a modeling study, important for interrupter analysis especially in the context of reconstruction of the phenomenon located in the range of the respiratory duct. Nevertheless, the results of investigations for the proposed complex linear representations of the respiratory system make the next steps – inhomogeneities and nonlinearities included – well-founded.

Acknowledgements

The article published with the support of the Foundation for Polish Science, research grant PB N N505 434 036, and statutory grant no 34 192 3.

References

[1] I. Bruderman, S. Abboud: "Telespirometry: novel system for home monitoring of asthmatic patients". *Telemed. J.,* vol. 3, no. 2, 1997, pp. 127-133.

[2] O. Diaz, C. Villafranca, H. Ghezzo, G. Borzone, A. Leiva, J. Milic-Emili, C. Lisboa: "Role of inspiratory capacity on exercise tolerance in COPD patients with and without tidal expiratory flow limitation at rest". *Eur. Respir. J.*, vol. 16, no. 2, 2000, pp. 269-275.

[3] C. Tantucci, A. Duguet, A. Ferreti, S. Mehiri, I. Arnulf, M. Zelter, T. Similowski, J.P. Derenne, J. Milic-Emili: "Effect of negative expiratory pressure on respiratory system flow resistance in awake snorers and nonsnorers". *J. Appl. Physiol.,* vol. 87, no 3, 1999, pp. 969-976.

[4] K.R. Lutchen, C.A. Giurdanella, A.C. Jackson: "Inability to separate airway from tissue properties by use of human respiratory input impedance". *J. Appl. Physiol.*, vol. 68, no. 6, 1990, pp. 2403-2412.

[5] K.R. Lutchen, A.C. Jackson: "Confidence bounds on respiratory mechanical properties estimated from transfer versus input impedance in humans versus dogs". *IEEE Trans. Biomed. Eng.*, vol. 39, no. 6, 1992, pp. 644-651.

[6] J. Vogel, U. Schmidt: *Impulse oscillometry: analysis of lung mechanics in general practice and the clinic, epidemiological and experimental research*. Frankfurt am Main, pmi-Verl.-Gruppe, 1994.

[7] J. Mead, J.L. Whittenberger: "Evaluation of airway interruption technique as a method for measuring pulmonary air-flow resistance". *J. Appl. Physiol.,* vol. 6, no. 7, 1954, pp. 408-416.

[8] G. Liistro, D. Stănescu, D. Rodenstein, C. Veriter: "Reassessment of the interruption technique for measuring flow resistance in humans". *J. Appl. Physiol.*, vol. 67, no. 3, 1989, pp. 933-937.

[9] J.H.T. Bates, T. Abe, P.V. Romero, J. Sato: "Measurement of alveolar pressure in closed-chest dogs during flow interruption". *J. Appl. Physiol.,* vol. 67, no. 1, 1989, pp. 488-492.

[10] I. Jabłoński, J. Mroczka: "Computer-aided evaluation of a new interrupter algorithm in respiratory mechanics measurement". *Biocyb. & Biomed. Eng.,* vol. 26, no. 3, 2006, pp. 33-47.

[11] I. Jabłoński, J. Mroczka: "Interrupter valve kinematics in the issues of parameter estimation of the respiratory system model". *Metrol Meas Syst,* vol. 14, no. 3, 2007, pp. 339-350.

[12] I. Jabłoński, J. Mroczka: *Frequency-domain identification of the respiratory system model during the interrupter experiment. Measurement*, vol. 42, no. 3, 2009, pp. 390-398.

[13] U. Frey, A. Schibler, R. Kraemer: "Pressure oscillations after flow interruption in relation to lung mechanics". *Respir. Physiol.,* vol. 102, no. 2-3, 1995, pp. 225-237.

[14] U. Frey, R. Kraemer: "Oscillatory pressure transients after flow interruption during bronchial challenge test in children". *Eur. Respir. J.,* vol. 10, no. 1, 1997, pp. 75-81.

[15] K.R. Lutchen, K.D. Costa: "Physiological interpretations based on lumped element models fit to respiratory impedance data: use of forward-inverse modelling". *IEEE Trans. Biomed. Eng.,* vol. 37, no. 11, 1990, pp. 1076-1085.

[16] A.G. Polak: "A forward model for maximum expiration". *Comput. Biol. Med.,* vol. 28, no. 6, 1998, pp. 613-625.

[17] E.R. Weibel : *Morphometry of the human lung*. Academic Press, New York, 1963.

[18] I. Ginzburg, D. Elad: "Dynamic model of the bronchial tree". *J. Biomed. Eng.*, vol. 15, no. X, 1993, pp. 283-288.

[19] B.R. Wiggs, R. Moreno, J.C. Hogg, C. Hilliam, P.D. Paré: "A model of the mechanics of airway narrowing". *J. Appl. Physiol.,* vol. 69, no. 3, 1990, pp. 849-860.

[20] R.K. Lambert, T.A. Wilson, R.E. Hyatt, J.R. Rodarte: "A computational model for expiratory flow". *J. Appl. Physiol.: Respirat. Environ. Exercise Physiol.*, vol. 52, no. 1, 1982, pp. 44-56.

[21] J. Milic-Emili J. (*ed.*): *Respiratory mechanics. European Respiratory Society*. Sheffield, 1999.

[22] K.R. Lutchen, A.C. Jackson: "Confidence bounds on respiratory mechanical properties estimated from transfer versus input impedance in humans versus dogs". *IEEE Trans. Biomed. Eng.*, vol. 39, no. 6, 1992, pp. 644-651.

[23] W. Tomalak: "Models of the respiratory system for interpretation of the transfer respiratory impedance". *VIII Symposium "Modelling and Simulations of Measurement Systems",* Krynica Górska, 21-25 September 1998, pp. 183-191. (in Polish)

[24] J. Mroczka, A.G. Polak: "Non-invasive method for measurement of respiratory system parameters". *Proceedings of the XIII IMEKO World Congress*, Torino, Italy, 5-9 September 1994, vol. 2, pp. 1561-1565.

[25] K.R. Lutchen, F.P. Primiano Jr, G.M. Saidel: "A nonlinear model combining pulmonary mechanics and gas concentration dynamics". *IEEE Trans. Biomed. Eng.,* vol. 29, no. 9, 1982, pp. 629-641.

[26] P.V. Romero, J. Sato, F. Shardonofsky, J.H.T. Bates: "High frequency characteristics of respiratory mechanics determined by flow interruption". *J. Appl. Physiol.,* vol. 69, no. 5, 1990, pp. 1682-1688.

[27] U. Frey, B. Suki, R. Kraemer, A.C. Jackson: Human respiratory input impedance between 32 and 800 Hz, measured by interrupter technique and forced oscillations. *J. Appl. Physiol.*, vol. 82, no. 3, 1997, pp. 1018-1023.

[28] A.C. Jackson, H.T. Milhorn Jr, J.R. Norman: "A reevaluation of the interrupter technique for airway resistance measurement". *J. Appl. Physiol.,* vol. 36, no. 2, 1974, pp. 264-268.

[29] M. Oswald-Mammosser, A. Charloux, I. Enache, E. Lonsdorfer-Wolf, B. Geny: "A comparison of four algorithms for the measurement of interrupter respiratory resistance in adults". *Respir. Med.*, vol. 103, no. 1, 2009, pp. 729-735.

[30] J.A. Bloom: *Monitoring of respiration and circulation*. CRC Press LLC, Boca Raton, Florida 2004.