# THE 30TH EUROPEAN MODELING & SIMULATION SYMPOSIUM

*SEPTEMBER 17-19, 2018*
BUDAPEST, HUNGARY


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# A LUMPED PARAMETER MODEL OF AIRWAY/LUNG MECHANICS

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**ABSTRACT**
In this work we present a nonlinear lumped parameter model of the airway/lung mechanics. The model is able to reproduce the time evolution of the fundamental variables (pressure, flow and volume) within each compartment of the respiratory system during normal breathing. A particular attention is given to the pleural pressure that is the driving force of the system. Our purpose is to build a tool for a qualitative description of the lung/airway dynamics that is able to reproduce the physiological behavior of the main system variables. The aim of this work is to develop a model of the respiratory system and successively insert it into our numerical lumped parameter model of the cardiovascular system. In this way it will be possible to study the interactions between the cardiovascular, the respiratory system and the most used mechanical circulatory and ventilatory assist devices in terms of hemodynamic and energetic variables.

Keywords: airway/lung mechanics, nonlinear lumped parameter model, numerical simulation, normal breathing

## 1. INTRODUCTION

Human cardiovascular and respiratory systems are two complex systems interacting with each other. Zero-dimensional (0D) or lumped parameters models assume a uniform distribution of the fundamental variables (pressure, flow and volume) within each compartment of the respiratory and circulatory systems at any instant of time, thus reducing the complexity in their representation. Typically, a hydraulic and pneumatic electrical analogy is applied, so that the cardiovascular and respiratory systems can be represented by electrical circuits. Voltage and current correspond to pressure and flow (of blood or air); resistances and inductances correspond to friction and inertia (of blood in the vessels or air in the airways); capacitances correspond to elasticity properties or compliances (of cardiovascular or respiratory components such as vessels and lungs); Kirchhoff's laws correspond to mass and energy conservation laws. The governing equations of the model consist on a set of linear or nonlinear ordinary differential equations which has to be solved with proper initial conditions.

Many models can be found describing the respiratory system (Golden, Clark and Stevens 1973; Olender, Clark and Stevens 1976; Liu, Niranjan, Clark, San, Zwischenberger and Bidani 1998; Athanasiades, Ghorbel, Clark, Niranjan, Olansen, Zwischenberger and Bidani 2000; Masana 2015; Bersani, Bersani and De Lazzari 2017) and the cardiovascular system (Formaggia, Quarteroni and Veneziani 2009; Korakianitis and Shi 2006; Rideout and Snyder 1974; Shi 2007) but relatively few models can be found describing the interactions between them (Lu, Clark, Ghorbel, Ware and Bidani 2001; Hemalatha, Suganthi and Manivannan 2010).

A lumped parameter model for the cardiovascular system has been developed at Cardiovascular Numerical/Hybrid Modelling Lab (Institute of Clinical Physiology of the Italian National Research Council of Rome). This 0D model, named CARDIOSIM© (De Lazzari, Neglia, Ferrari, Bernini, Micalizzi, L'Abbate and Trivella, 2009; De Lazzari and Stalteri 2011; Capoccia, Marconi, Singh, Pisanelli and De Lazzari 2018), is made up of several modules describing the sections of the cardiovascular system. The software is a suitable and versatile environment to reproduce heart and circulation main features in both physiological and pathological conditions.

The final goal of this study is to develop a model of the respiratory system and insert it as a new modulus in CARDIOSIM©. In this way it will be possible to study the interactions between the cardiovascular, the respiratory system and the most used mechanical circulatory and ventilatory assist devices in terms of hemodynamic and energetic variables (De Lazzari, D'Ambrosi, Tufano, Fresiello, Garante, Sergiacomi, Stagnitti, Caldarera and Alessandri 2010; De Lazzari, Darowski, Ferrari, Pisanelli and Tosti 2006; De Lazzari, Darowski, Ferrari, Clemente and Guaragno, 2001; De Lazzari, Darowski, Ferrari and Clemente 1998; Colquitt, Colquhoun and Thiele 2011).

In this paper, we present the first results concerning only the airway/lung mechanics. Our purpose is to build a tool for a qualitative description of the lung/airway dynamics that is able to reproduce the physiological behavior of the

main system variables. In the future, the gas exchange will be added to complete the model. Once the model has been completed, the parameters will be properly set by fitting measured quantities and the model will be validated with clinical data.
A particular attention has been given to the pleural (or intra-thoracic) pressure which is the driving force of the model, as it is one of the principal factors that influence the cardio-pulmonary circulation. The model has been tested using some pleural pressure waveforms reproducing different kinds of normal breathing. Some preliminary results are presented.
The model has been implemented in MATLAB and solved in Simulink environment.

## 2. MATERIALS AND METHODS

### 2.1. Respiratory system

The model of the lung/airway mechanics considers the section of the respiratory system below the larynx, where the structures in which air flows divides into airways and alveoli. The lungs are considered as a unique alveolar region connected to the external environment through the airways which divide into the upper airway, the collapsible airway and the small, or peripheral, airways. The upper airway is considered as a rigid tubular structure; the collapsible airway is considered a tubular structure whose volume can change in time; the small airways are the ramified structures which lie in the lungs. Both alveolar region and most of the airways are surrounded by the pleural space. This space (containing the pleural fluid) separates the lungs visceral pleura from the internal parietal pleura of the chest wall (Bates 2009).
The variables of the model are flow, pressures and volumes. The flow ($F$) of the air in the upper airway is the same as the airflow in the mouth. It is considered positive during expiration and negative during inspiration. The alveolar pressure ($P_A$), the collapsible airway pressure ($P_C$) and the pleural pressure ($P_{pl}$) are given with respect to environmental pressure ($P_{ref}$), which is set to zero. In our model the pressure at mouth ($P_e$) equals environmental pressure. Instead, the dynamic lung elastic recoil ($P_{el}$), or transpulmonary pressure, and the transmural pressure ($P_{tm}$) are referred to the pleural pressure. Thus we have:

$$P_A = P_{el} + P_{pl}, \qquad P_C = P_{tm} + P_{pl}. \tag{1}$$

Finally, we consider the alveolar volume ($V_A$) and the collapsible segment volume ($V_C$).
All these quantities vary in time. The spatially averaged pleural pressure is the driving pressure of the entire system. Its variation in time is caused by the contraction and relaxation of the respiratory muscles.

### 2.2. Model description

The electrical analogue of airway/lung mechanics is composed by four resistances, two capacitors and a generator (Fig. 1).

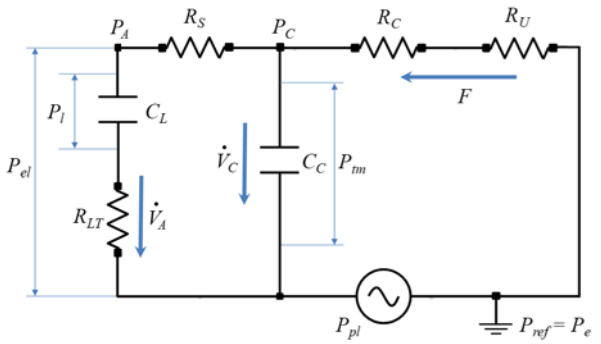

Figure 1: Electrical equivalent circuit of the airway/lung dynamics.

Each element characterizing each compartment of the respiratory system is defined as a function of the main variables (the flow $F$ or the volumes $V_A$ and $V_C$).
The resistive elements characterize the resistance of the corresponding compartment to the flow of air. The resistance of the upper airway ($R_U$) is given by:

$$R_U = A_u + K_u \,|\, F \,| \tag{2}$$

where $A_u$ and $K_u$ are positive constants.
The resistance of the collapsible airway ($R_C$) is given by:

$$R_C = K_c \left( \frac{V_{C\max}}{V_C} \right)^2 \tag{3}$$

where $K_c$ is the value of the resistance when the volume reaches the maximum possible volume of the collapsible airway ($V_{Cmax}$).
The resistance of the small airways is given by:

$$R_S = A_s e^{-K_s \frac{V_A - V_R}{V^* - V_R}} + B_s \tag{4}$$

where $A_s$, $K_s$, $B_s$ and $V^*$ are positive constants and $V_R$ is the residual volume that is the volume of air left in the lungs after a maximum expiratory effort.
Finally, the lung tissue viscous property is modelled by a constant resistance ($R_{LT}$).
The two capacitors simulate the compliance of the alveolar region and of the collapsible airway which expresses their elasticity (i.e. their ability to expand when a pressure change occurs). The currents passing through these elements are at any time the derivatives of the respective volumes and the pressure drop across them are the static lung elastic recoil ($P_l$) and the transmural pressure ($P_{tm}$), respectively. Their expressions are in Bersani, Bersani and De Lazzari (2017).
The generator simulates the pleural pressure ($P_{pl}$) which drives the respiratory process.
The electrical circuit is solved through Ohm's law and Kirchhoff's current and voltage laws:


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$$\begin{cases} I = \dot{V}_A + \dot{V}_C \\ R_U I + R_C I + P_{tm} + P_{pl} = 0 \\ R_S \dot{V}_A + P_l + R_{LT} \dot{V}_A - P_{tm} = 0 \end{cases} \quad (5)$$

Substituting the first equation in the second, we obtain two nonlinear first order ordinary differential equations. $V_A$ and $V_C$ are the unknowns and $P_{pl}$ is the input of the model.
The system in Eq. (5) has to be solved with proper initial conditions:

$$\begin{cases} V_A(0) = V_A^0 \\ V_C(0) = V_C^0 \end{cases} \quad (6)$$

The model has been implemented in MATLAB and solved in the Simulink environment through a variable time step Dormand-Prince method with a relative tolerance of $1\times10^{-6}$.
The parameters of the model are set as in Athanasiades, Ghorbel, Clark, Niranjan, Olansen, Zwischenberger and Bidani 2000 and Liu, Niranjan, Clark, San, Zwischenberger and Bidani 1998.

### 2.3. Pleural pressure

We tested the model response to three input pleural pressures simulating the regular breathing and two kinds of quiet breathing, respectively.
The respiratory frequency for the healthy adult at rest is about 12-18 breaths per minute and the pleural pressure keeps negative in a range of about -8 to -4 $cmH_2O$. In this work we consider a respiratory rate of 12 breaths per minute. Each respiratory act lasts 5 seconds. The pleural pressure oscillates in an overall range of -3.67 to -7.48 $cmH_2O$.
The pleural pressure of the regular breathing ($P^r_{pl}$) is taken from Bai, Zhang, Zhao, and Zhou (1998) (Fig. 2).
The pleural pressure waveform of the quiet breathing is inspired to those in Hemalatha and Manivannan (2010) and Lu, Clark, Ghorbel, Ware and Bidani (2001). These kinds of pleural pressure have been obtained applying a low-pass filter to a pulse generator. Setting the pulse amplitude, the duration of the pulse in each period and the values of the resistance and capacity, it is possible to change the waveform of the response, in order to obtain the desired final pleural pressure (Fig. 3).

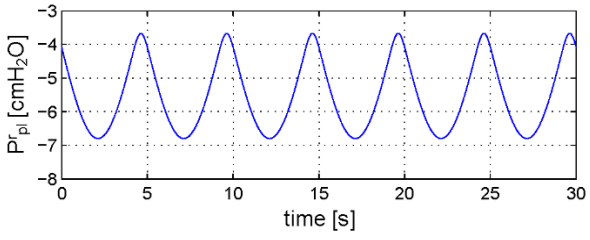


Figure 2: Pleural pressure of the regular breathing.

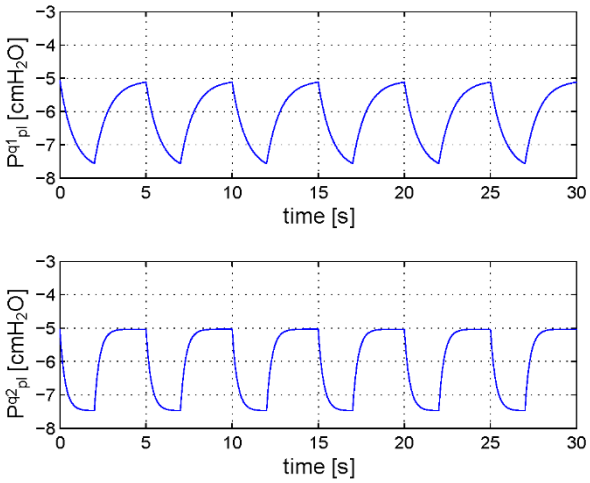


Figure 3: Two different pleural pressures ($P^{q1}_{pl}$ and $P^{q2}_{pl}$) for the quiet breathing obtained from the RC circuit.

## 3. RESULTS

Figures 4 and 5 show the results when $P^r_{pl}$ and $P^{q2}_{pl}$ are applied as input of the model, respectively. In each figure, panel (A) shows the air flow at the mouth. Panel (B) shows the alveolar volume. Panel (C) shows the alveolar and the pleural pressures. Their difference at any instant of time is the dynamic elastic recoil. Panel (D) shows the flow-volume loop, where the volume is the total lung volume ($V_L$) referred to the residual volume ($V_R$). $V_L$ is the sum of the alveolar volume, the collapsible volume and the death space (upper airway) volume ($V_D$):

$$V_L = V_A + V_C + V_D. \quad (7)$$

In normal breathing the difference $V_L$-$V_R$ keeps positive, since the lungs never collapse up to the residual volume. The loop is counterclockwise.
In regular breathing the pleural pressure keeps in a range of -6.80 to -3.67 [$cmH_2O$] while in quiet breathing it varies in a range of -7.48 to -5.03 [$cmH_2O$]. This causes the flow of air at mouth to be lower in regular breathing than in quiet breathing. In the first case the flow varies between -0.36 and 0.34 [$l/s$] (Fig. 4, panels A and D) while in the second it varies between -0.56 and 0.51 [$l/s$] (Fig. 5, panels A and D). Moreover, the flow waveforms show how fast or slow the air enters or exits the mouth in each respiratory cycle. This effect is highlighted in the differences in the flow-volume loops shape (Fig. 4 and 5, panel D).
Alveolar volume varies in a range of 3.10 to 3.59 [$l$] in the regular breathing and in a range of 3.32 to 3.69 [$l$] in the quiet breathing (Fig 4 and 5, panel B). The total lung volume varies accordingly, resulting in a horizontal shift of the flow-volume loops (Fig. 4 and 5, panel D).
The different waveforms of the pleural pressures are reflected also in the waveforms of alveolar pressure and elastic recoil (Fig. 4 and 5, panel C). The alveolar pressure varies between -0.55 and 0.54 [$cmH_2O$] in regular breathing and between -0.80 and 0.62 [$cmH_2O$] in quiet breathing.


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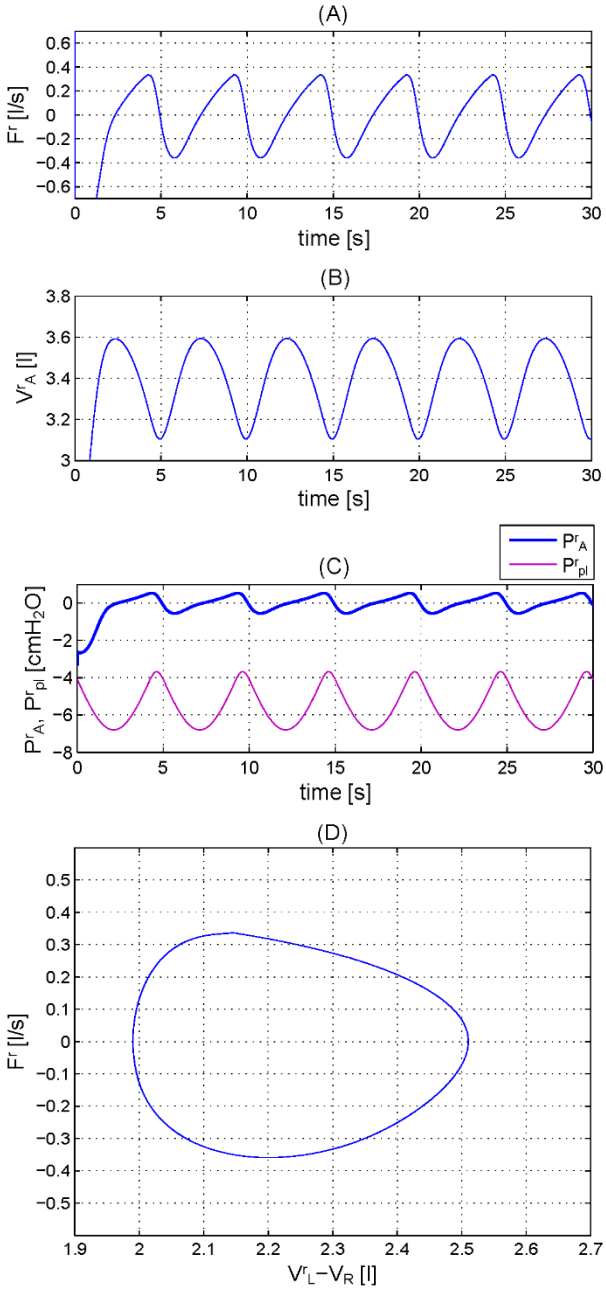


Figure 4: Simulation results: the pleural pressure of the regular breathing is applied as input in the model. (A) flow at the mouth; (B) alveolar volume; (C) alveolar pressure and pleural pressure; (D) flow-volume loop.

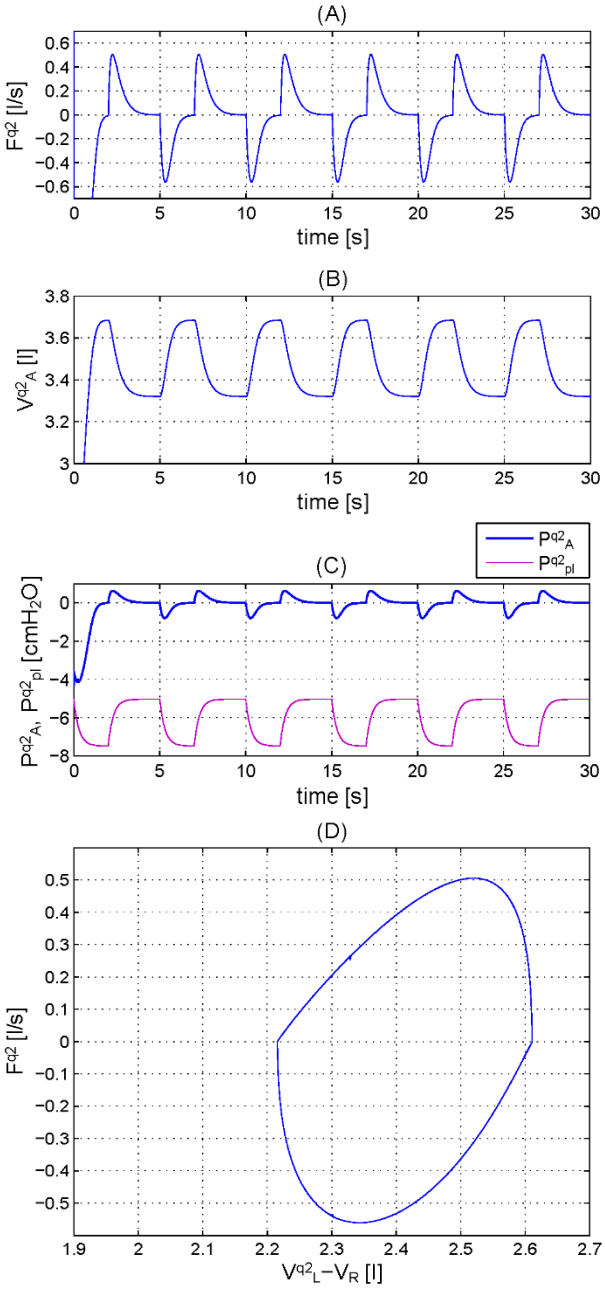


Figure 5: Simulation results: $P^{q2}_{pl}$ is applied as input in the model. (A) flow at the mouth; (B) alveolar volume; (C) alveolar pressure and pleural pressure; (D) flow-volume loop.

These results are in agreement with data and simulations in the literature (Liu, Niranjan, Clark, San, Zwischenberger and Bidani 1998; Hemalatha, Suganthi and Manivannan 2010; Lu, Clark, Ghorbel, Ware and Bidani 2001; Masana 2015).

## 4. CONCLUSIONS

The nonlinear lumped parameter model of the airway/lung mechanics presented in this work is able to describe the time variation and the relations between the main variables of the respiratory system during normal breathing.

Our preliminary results regarding the flow-volume analysis are in accordance with data presented in literature.

We plan to test the model with other kinds of respiration and respiratory maneuvers. Also, increasing (emphysema) or decreasing (fibrosis) lung compliance and other lung diseases can be simulated using the presented model. A more in-depth study of the influence of the parameters on the model's response will come in the future as well as the proper setting of the parameters and the validation of the model by fitting measured data. Furthermore, gas exchange will be added to the model.

From the numerical point of view, suitable numerical schemes for the solution of the model will be considered to find out the most convenient and feasible way to make it dialogue with CARDIOSIM©, which is implemented using Visual Basic program language.

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