Respiratory System Model Based on PSPICE

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***Abstract*—The aim of the present work is the building of a lumped nonlinear dynamic model of lung/airway mechanics using generic instead of specific software, in an attempt to offer an open simulation environment.**

Based on the analogy between pneumatic and electric magnitudes, an electrical equivalent circuit of the lung/airway mechanics is derived. Then, the nonlinear circuit elements are constructed by means of the powerful Analog Behavioral Modeling (ABM) building blocks and the system is solved using PSPICE.

Following the approach in [3] and [4], five lumps are defined: two capacitors (elastances) corresponding to lung and collapsible airway segment and three resistors, corresponding to lung, collapsible airway segment and upper airway. The element definition involves as much as five parameters for the lung, four parameters for the collapsible segment and two parameters for the upper airway.

The model does not attempt to mimic any particular system adjusting a given set of parameters but instead to provide a tool to explore the relationship between a given parameter or set of parameters and the system response. In particular, Forced Vital Capacity (FVC) maneuver and tidal breathing will be explored.

***Index Terms*—Respiratory System Model, Analog Behavioral Modeling, Nonlinear model, Open simulation environment, PSPICE, FVC maneuver, Tidal breathing.**

I. Introduction

THE human respiratory system, as many other complex systems, requires its complexity to be reduced if we want to analyze it and be able to draw sufficiently general conclusions. That is precisely the objective of any model.

Two different groups of models can be identified: fundamental models that use some numerical method, usually finite differences or finite elements, to solve the system fundamental differential equations, and functional models, that simplify the system before trying to solve it, usually by some spatial averaging or lumping of the original system that reduces its complexity while still keeping some or all of its functional description.

Fundamental models achieve a large amount of detail that sometimes may hide the conclusions on system functionality and are complex to describe due to the large number of input parameters involved. On the other side, they offer a very detailed description of any particular system.

Functional models, although having in common the spatial averaging or lumping of the system, can present a varying degree of complexity by using more or less lumps and defining them through linear or nonlinear expressions, according to the degree of precision required. In the study of the human respiratory mechanics, numerous models have appeared in the related literature, both fundamental [1], [2] and functional [3]-[9].

In this work, we present a lumped nonlinear dynamic model following the approach in [3], [4], [7] and [8] by defining five lumps, two capacitors (elastances) corresponding to lung and collapsible airway segment and three resistors, corresponding to lung, collapsible airway and upper airway, all of them defined by nonlinear analytical expressions. A significant difference with the preceding models is that it will be defined inside an electrical circuit solving environment (PSPICE) by means of the so called Analog Behavioral Modeling (ABM) building blocks that will allow an easy definition and parameterization of the nonlinear functions describing the model lumps and will in addition provide a very powerful graphic interface for the output of results.

All this will set on an open simulation environment allowing us to easily explore the relationship between any model parameter or set of parameters and the system response that may eventually lead to a deeper physiological insight into the respiratory mechanics.

Apart from the model itself and its components, we will focus on the role of the system excitation. Although pleural pressure will be used as usual, it cannot be considered an independent driving force in the sense that, as stated in [10], the system can only be completely described if all three variables, i.e., pressure, flow and volume are simultaneously defined. But the pneumatical to electrical analogy that allows us to use an electrical circuit solver as a general solving environment will use the standard voltage-current representation, whose pneumatical equivalence is pressure-flow. Then, volume has to be calculated as the integral of flow and care has to be taken in order to preserve coherence of the results.

Previous works do not attempt to build a model first and then calculate the response to a given excitation, but instead they measure a set of system variables, say pleural pressure, flow at the mouth and lung volume, and then calculate the system parameters to get the best fit for that particular set of measurements.

This is why we will use their findings to set the expressions for system elements and values of model parameters, with some modifications to preserve convergence. Once this is done and the model is built, we will work the other way by calculating the response to various excitations and parameter values, in a kind of "what if" approach.

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II. Model Equations

We have already mentioned in the introduction that our model will contain five lumps: two capacitors (elastances) corresponding to lung and collapsible airway segment and three resistors, corresponding to lung, collapsible airway and upper airway. Fig. 1 shows the model elements and its relationship with anatomy.

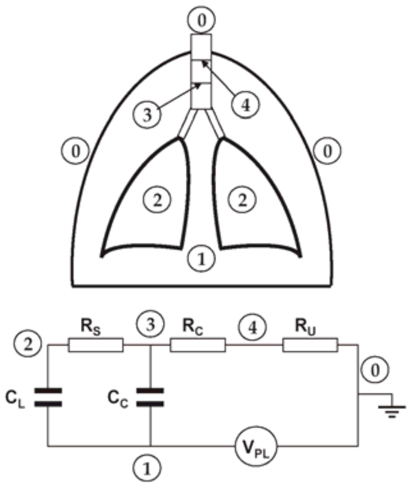

Fig. 1. Lung/Airway model

In our lumped model, node 1 will be the pleural cavity, node 2 the alveolar space, node 3 the output side of the trachea, node 4 the output of collapsible segment and node 0 the outside world at atmospheric pressure that will be the reference for pressures and will be given the value of zero.

All the model elements are nonlinear and their expressions are developed in the following paragraphs.

A. Lung capacitance C_L

The expression for lung capacitance C_L is derived from the lung pressure-volume relationship in [4] and [8]:

$$P_L = A_L \cdot \exp(k_L \cdot V_L) + B_L \tag{1}$$

where P_L is the transpulmonary pressure, V_L the lung volume and A_L, B_L and k_L are model parameters. To calculate the lung capacitance is more convenient to put the volume as a function of pressure so, from (1):

$$V_L = \frac{1}{k_L} \cdot \ln \frac{P_L - B_L}{A_L} \tag{2}$$

Then we have:

$$C_L = \frac{dV_L}{dP_L} = \left[k_L \cdot (P_L - B_L)\right]^{-1} \tag{3}$$

This expression yields negative C_L values for $P_L<B_L$, which has in fact no physical sense. Because the solving method starts its first guess at $P_L = 0$ then we have to make sure that B_L is negative, which is not necessarily true if its value is the result of an experimental value fitting as in [8]. To solve this problem we will use the following expression instead of (1):

$$P_L = A_L \cdot \left[\exp(k_L \cdot V_L) + 1\right] \tag{4}$$

which yields the following result for C_L:

$$C_L = \frac{dV_L}{dP_L} = \left[k_L \cdot (P_L + A_L)\right]^{-1} \tag{5}$$

In this way we solve the convergence problem and make $V_L = 0$ for $P_L = 0$, which is coherent with physics. Moreover, we spare one model parameter.

B. Small airway resistance R_S

The expression for small airway resistance R_S follows from [7] and [8]:

$$R_S = A_S \cdot \exp\left[k_S \cdot (V_L - V_R)/(V^* - V_R)\right] + B_S \tag{6}$$

where once again V_L is the lung volume, V_R is the residual volume and A_S, B_S, k_S and V^* are model parameters.

Although in [7] the term B_S is made dependent on pleural pressure P_{PL}, for the sake of simplicity we will use a constant value instead. We consider the difference of second order to the results.

To reduce the number of parameters and introduce P_L as the independent variable instead of V_L, we will further manipulate (6) as follows:

$$R_S = A_S \cdot \exp \frac{k_S V_L}{V^* - V_R} \cdot \exp \frac{-k_S V_R}{V^* - V_R} + B_S = A \cdot \exp kV_L + B_S \tag{7}$$

where:

$$A = A_S \cdot \exp \frac{-k_S V_R}{V^* - V_R} \tag{8}$$

$$k = \frac{k_S}{V^* - V_R} \tag{9}$$

Now, if we substitute V_L for its value as a function of P_L calculated from (4) into the second term of (7) we get:

$$R_S = A \cdot \left[\frac{P_L}{A_L} + 1\right]^{\frac{k}{k_L}} + B_S \tag{10}$$

C. Collapsible airway capacitance C_C

When considering the collapsible airway capacitance we have to keep in mind the structural properties of this particular airway segment. At high positive transmural pressure, the airway can hardly stretch past its maximum volume so volume is almost constant and capacitance is low. At high negative transmural pressure, the airway is completely collapsed and its volume can hardly decrease below its minimum (fully collapsed) value so capacitance is again low. This leads to a sigmoid like volume to pressure curve. Accordingly, we will use in our model a generic sigmoid curve of the form:

$$\frac{V_C}{V_{C\max}} = \frac{1}{1 + \exp\left[-(P_C - P_0)/P_1\right]} \tag{11}$$

where P_C is the transmural pressure, V_C is the collapsible segment volume, V_{Cmax} is a parameter indicating the maximum achievable volume of that segment and P_0 and P_1 are the

TABLE I
MODEL EQUATIONS

Equation	Model Element	Parameters	Reference
$P_L = A_L \cdot [\exp(k_L \cdot V_L) + 1]$	$C_L = [k_L \cdot (P_L + A_L)]^{-1}$	k_L, A_L	[4], [8], t.w.
$R_S = A_S \cdot \exp[k_S \cdot (V_L - V_R)/(V^* - V_R)] + B_S$	$R_S = A \cdot \left[\frac{P_L}{A_L} + 1\right]^{\frac{k}{k_L}} + B_S$	A, K, K_L A_L, B_S	[7], [8]
$\frac{V_C}{V_{Cmàx}} = \frac{1}{1 + \exp[-(P_C - P_0)/P_1]}$	$C_C = \frac{V_{Cmàx}}{P_1} \cdot \frac{1}{\{\exp[-(P_C - P_0)/2P_1] + \exp[(P_C - P_0)/2P_1]\}^2}$	V_{cmax}, P_0, P_1	this work
$R_C = k_C \cdot \left(\frac{V_{Cmàx}}{V_C}\right)^2$	$R_C = k_C \cdot \{1 + \exp[-(P_C - P_0)/P_1]\}^2$	k_C, P_0, P_1	[3], [4], [8]
$R_U = A_U + k_U \cdot \lvert I_U \rvert$	$R_U = A_U + k_U \cdot \lvert I_U \rvert$	A_U, K_U	[3], [4], [8]

sigmoid parameters. P_0 gives the pressure value at half volume and P_1 adjusts the slope of the curve. From (11), the expression for the collapsible airway capacitance turns to be:

$$C_C = \frac{dV_C}{dP_C} = \frac{V_{Cmàx}}{P_1} \cdot \frac{1}{\{\exp[-(P_C - P_0)/2P_1] + \exp[(P_C - P_0)/2P_1]\}^2} \tag{12}$$

D. Collapsible airway resistance R_C

We will use for the collapsible segment resistance the well accepted expression from [3], [4] and [8]:

$$R_C = k_C \cdot \left(\frac{V_{Cmàx}}{V_C}\right)^2 \tag{13}$$

where V_{Cmax} has already been defined and k_C is another parameter. If we now substitute (11) into (13) we get:

$$R_C = k_C \cdot \{1 + \exp[-(P_C - P_0)/P_1]\}^2 \tag{14}$$

E. Upper airway resistance R_U

The upper airway resistance will be modeled according to [3], [4] and [8] as:

$$R_U = A_U + k_U \cdot \lvert I_U \rvert \tag{15}$$

where I_U is the flow at the mouth and A_U and k_U are model parameters. We will name the flow using the electric current symbol I throughout.

As usual, upper airway resistance has a linear term A_U plus a term proportional to air flow that takes turbulence into account.

Table I presents a summary of model equations, parameters and references.

III. MODEL ELEMENTS AND STRUCTURE

There are two kinds of elements in our model, apart from excitation: nonlinear capacitors and nonlinear resistors. We will implement in a general form the elements that emulate these capacitors and resistors by means of ABM building blocks.

A. Nonlinear capacitor

To emulate a nonlinear capacitor, we will consider the general relation between voltage and current in a capacitor:

$$i_C = C\frac{dv_C}{dt} \tag{16}$$

where i_C is the current through the capacitor and v_C the voltage across it. According to (16) to emulate a capacitor we have to make the current across our device proportional to the derivative of its voltage. Putting it the other way, if we draw a voltage through a unit capacitor and measure the current through it we obtain the voltage derivative. Then, if we multiply this value by an expression making for the value of our nonlinear capacitance and we force that magnitude to be the current through it we will emulate a nonlinear capacitor.

All this will be better understood by looking at Fig. 2. The elements E are called "Voltage controlled voltage source" and replicate a voltage magnitude at the secondary side without implying any load to the rest of the circuit at the primary side.

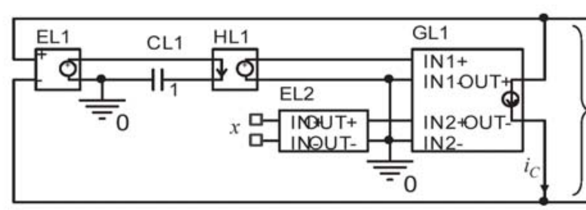

Fig. 2. ABM emulation of a nonlinear capacitor

EL1 has only "gain", meaning that the relation between primary and secondary side is a constant factor (default unity) whereas in *EL2* can be any mathematical function among those available at the ABM library. The element *H*, called "Current controlled voltage source" converts a primary current to a secondary voltage through a constant factor (default unity). Finally, the element *G*, called "Multiplying voltage controlled current source" multiplies both input voltages and converts the product into a current.

If we now analyze the circuit, we realize that the upper input to element *GL1* is the derivative of the voltage at the input of *EL1*, which is also the output voltage of *GL1* whereas the lower *GL1* input is a mathematical function, realized by

EL2 on the input labeled “*x*”. Accordingly, the output current of *GL1* is the current through a capacitor whose value is the mathematical function of *EL2* on “*x*”. We will see that in our final model, input “*x*” is the capacitor voltage itself.

It is worth to say now that the element *EL2* (called E-function) can also be of a type called E-Table, where the relation between input and output is given through a table, meaning that we can eventually synthesize a graphical function instead of an analytical one.

B. Nonlinear resistor

The emulation of a nonlinear resistor follows the same steps as for the capacitor but it is a bit simpler. The voltage current relationship in a resistor is:

$$v_R = R \cdot i_R \tag{17}$$

where v_R and i_R are resistor voltage and current respectively. Following the same path we can draw the circuit of Fig. 3.

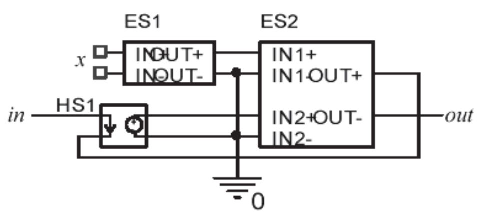

Fig. 3. ABM emulation of a nonlinear resistor

Here, *HS1* samples the current in the resistor and multiply it by the expression in *ES1* by means of *ES2* that now is a “Multiplying voltage controlled voltage source”. Then, the voltage between “*in*” and “*out*” realizes (17).

Our final model will include two nonlinear capacitors and three nonlinear resistors whose functional expressions are realized by E-Function elements implementing the functions listed in the previous paragraph and summarized in Table I. Needless to say that constant capacitors and resistors are standard elements of PSPICE.

IV. System Excitation

It has already been mentioned that we will use pleural pressure as system excitation. However, the flow that results from this pressure cannot take any value or form. In particular, it has to fulfill a condition: in steady state, the net volume increment that results from the integral of flow over one respiratory cycle has to be zero, otherwise an unlimited lung inflation or deflation will result, which is of course impossible. Thus, we have to make sure that the function we use for pleural pressure, apart of reasonably emulating the respiratory cycle is consistent with that fact.

The pleural pressure generator we will use in our model is illustrated in Fig. 4. The generator V_{PL} is a pulse generator of which we can adjust both rise and fall times, pulse duration, period and high and low voltage (pressure) values. The network C_{PL}-R_{PL} is a high pass filter to remove any DC component from signal and guarantee that the positive and negative areas are equal. V_O is a DC generator to finally add a controlled amount of offset and *E1* is a voltage controlled voltage source to isolate the simulated circuit from the high pass network.

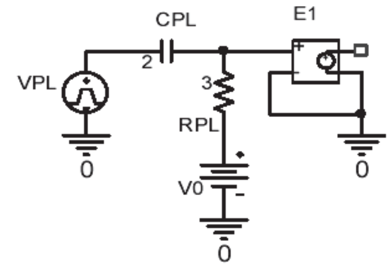

Fig. 4. Pleural pressure generator

Due to the capacitances in the circuit, some time has to elapse before the steady state can be reached, as depicted in Fig. 5. Lower trace is pleural pressure and upper trace is lung volume. It takes three to four respiratory cycles to reach steady state so analysis is not valid until then.

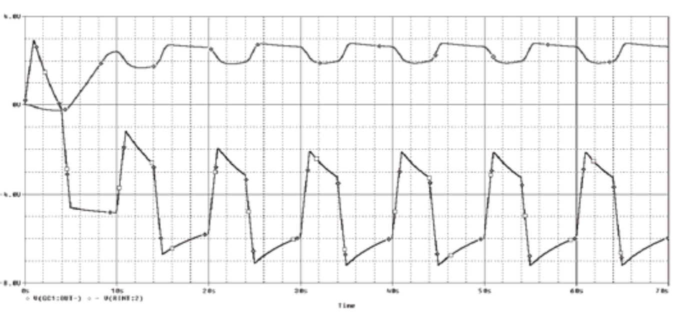

Fig. 5. Transient to the steady state

V. The Complete Model

Now that we have described all the necessary elements, we can build the complete model. Fig. 6 shows the model schematics that include the five mentioned lumps, the pleural pressure generator and two integrators to compute lung and collapsible segment volumes. Numbered nodes correspond with those of Fig. 1. Also, both integrators are sampled by V_{INT} to skip the transient phase. More on that later.

A remark has to be made concerning integrator 2 used to calculate the collapsible segment volume. Because this volume is not zero for zero transmural pressure, an initial condition has to be introduced into C_{INT2}. Its value can be calculated from (11) for $P_C = 0$.

$$V_C\big|_{IC} = \frac{V_{C\max}}{1 + \exp\left(P_0 / P_1\right)} \tag{18}$$

This value is not parameterized into the model so each time V_{Cmax}, P_0 or P_1 are changed, it has to be reintroduced. This is because the solving program does not allow to parameterize initial conditions.

In the section “*PARAMETERS*” all the model parameters are listed: five for the lung, four for the collapsible segment and two for the upper segment.

Three more parameters are defined for the pleural pressure generator: the peak to peak amplitude V_{val}, the offset V_{off} and the period T_C.

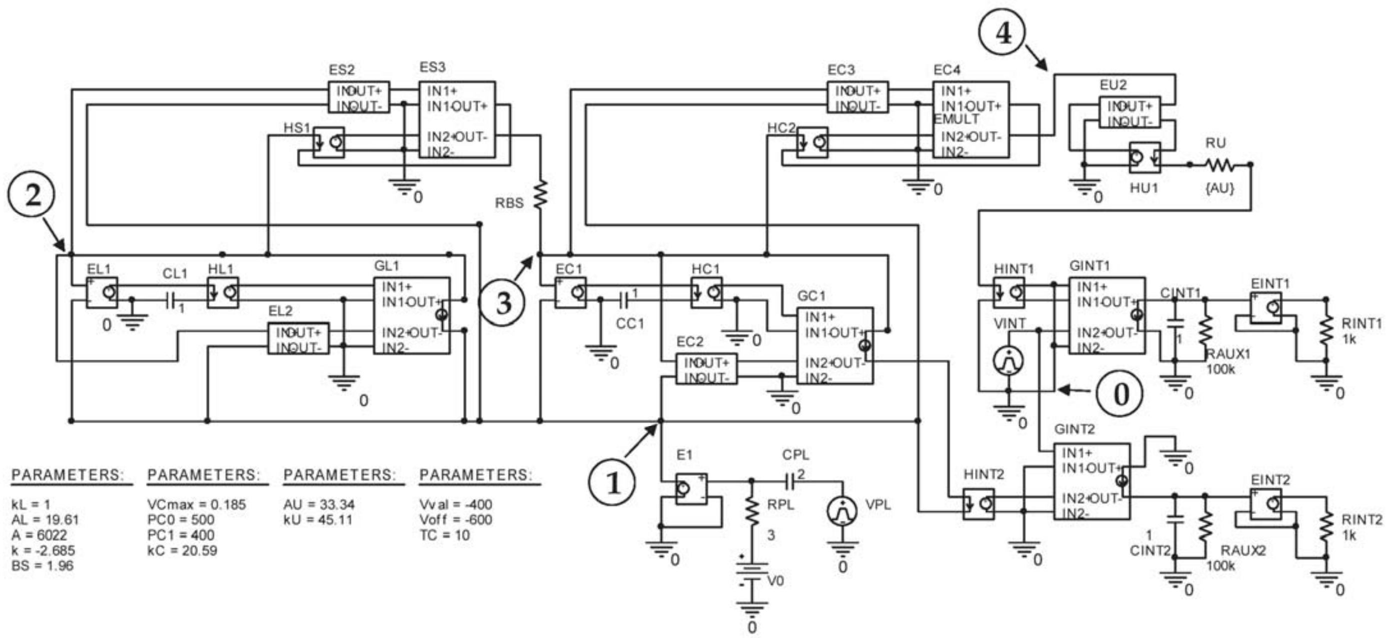

Fig. 6. Complete model schematic

VI. Simulation Results

Once the model is complete and parameterized, we will start with simulations. The parameter values are extracted from [8] and conveniently modified to the proposed model expressions and units. In particular, P_0 and P_1 have been adapted to fit the original (non-analytical) curve. Table 2 shows the values for four different subjects as in [8], along with the units. Pascal is used for pressure and liter for volume.

TABLE II
Parameter values

Param.	Subject nº				Units
	1	2	3	4	
k_L	1.00	1.00	1.00	0.80	l^{-1}
A_L	19.61	3.92	9.81	55.90	Pa
k	-2.685	-1.198	-0.956	-0.940	l^{-1}
A	6022	3166	1129	3228	Pa/l.s
B_S	1.96	1.96	1.96	1.96	Pa/l.s
k_C	20.59	48.05	49.03	23.54	Pa/l.s
P_0	500.00	500.00	500.00	500.00	Pa
P_1	400.00	750.00	1200.00	1350.00	Pa
V_{Cmax}	0.185	0.125	0.165	0.101	l
A_U	33.34	30.40	30.40	30.40	Pa/l.s
k_U	45.11	39.23	31.38	19.61	$Pa/(l.s)^2$
V*	5.30	10.30	8.41	7.37	l
V_R	1.24	2.04	1.61	1.91	l
V_T	5.19	8.27	6.34	7.20	l

For those used to pressure in $cm \cdot H_2O$, the equivalence is approximately $1cm \cdot H2O \approx 100$ Pa (98.0665 to be more exact).

We will run some simulations to show the influence of different parameter values on model elements and then proceed with FVC maneuver and tidal breathing.

We have to note at this point that lung volumes displayed are relative, that is, considering lung residual volume is zero. We could add residual volume as an initial condition to the integrator but it has no much interest. Nevertheless, the small airway resistance calculation is done taking into account residual volume which influence is included in *A* and *k*.

On the other side, collapsible segment volume is the real volume because in that case the initial condition is a must.

A. Model elements

Fig. 7 shows the simulation results for: *a)* Lung compliance, *b)* Small airway resistance, *c)* Collapsible segment compliance and *d)* Collapsible segment resistance. The excitation applied is the same as in the FVC maneuver.

In Fig. 7a we see the influence of A_L on the lung volume at a given pressure. In the curve of Subject #4 it can also be seen a higher slope, i.e., higher lung elastance due to a lower k_L value.

In Fig.7b, the *y* axis is in logarithmic scale for more clarity. The lower the value of *k*, the smoother the knee whereas the value of *A* sets the maximum R_S value.

In Fig. 7c, the value of P_1 sets the slope and thus the elastance. In Fig. 7d, the value of P_1 sets again the slope and k_C the value of resistance when segment is not collapsed, i.e., when transmural pressure is high and positive.

TABLE III
Excitation values for FVC

Param.	Value	Units
t_r	500.00	ms
tf	100.00	ms
pw	500.000	ms
T_C	10	s
V_{val}	-7000.00	Pa
V_{off}	0.00	Pa
C_{PL}	1.00	l/Pa
R_{PL}	0.70	Pa/l·s

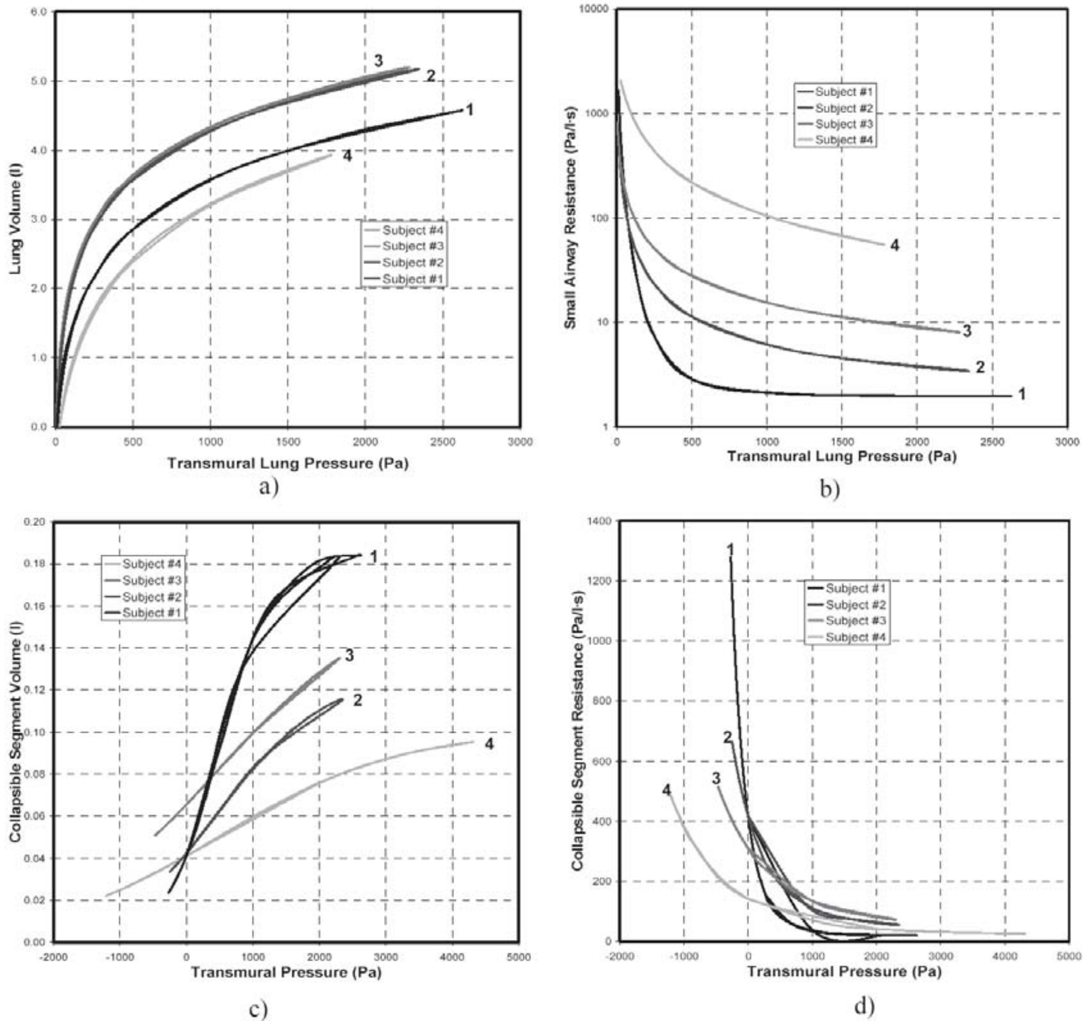

Fig. 7. Simulation results for: *a)* Lung compliance, *b)* Small airway resistance, *c)* Collapsible segment compliance and *d)* Collapsible segment resistance.

B. FVC maneuver

To simulate the FVC maneuver, the excitation parameters are set as listed in table 3 whereas Fig. 8 displays the excitation waveform for pleural pressure.

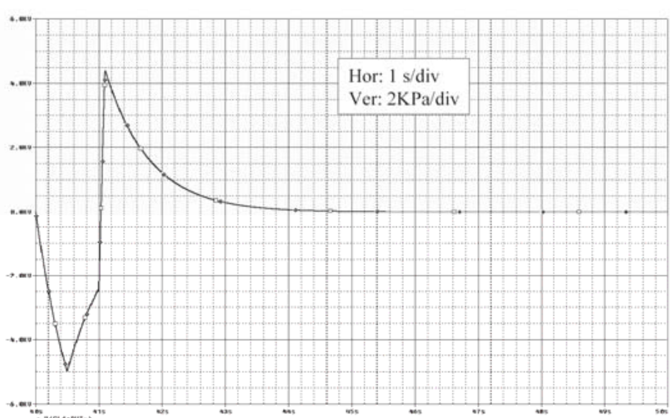

Fig. 8. Pleural pressure waveform for FVC

Fig. 9 shows the resulting spirometric curves relating lung volume and flow at the mouth. Subject #4 has the lowest volume due to the low lung elastance and the highest flow due to the lower upper airway resistance.

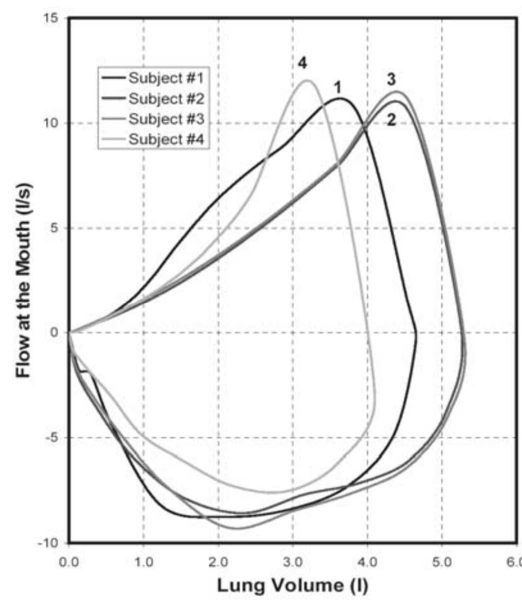

Fig. 9. Simulated spirometric curves for FVC maneuver

Subject #1 has a slightly lower flow-volume curve than subjects #2 and #3 due to lower lung elastance affecting the volume whereas the flow is quite similar.

C. Tidal Breathing

Tidal breathing is simulated using the excitation parameters listed in table 4 whereas Fig. 10 displays the excitation waveform for pleural pressure, once steady state is reached.

TABLE IV
EXCITATION VALUES FOR TIDAL BREATHING

Param.	Value	Units
t_r	1.00	s
tf	1.00	s
pw	3.00	s
T_C	10.00	s
V_{val}	-400.00	Pa
V_{off}	-600.00	Pa
C_{PL}	2.00	l/Pa
R_{PL}	3.00	Pa/l·s

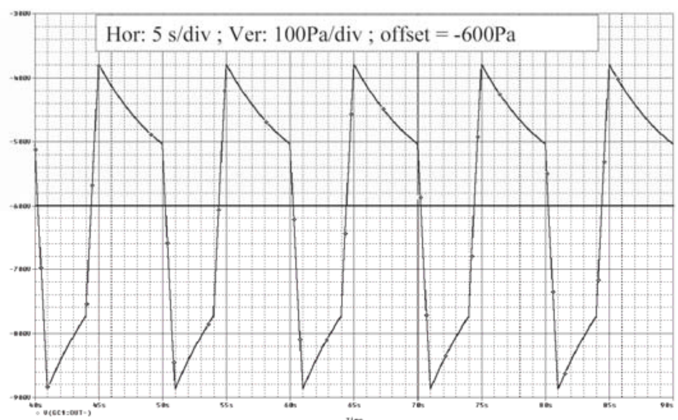

Fig. 10. Pleural pressure waveform for Tidal Breathing

Simulation results are shown in Fig. 11, where flow at the mouth vs. alveolar pressure is the chosen representation for tidal breathing because it shows at a glance the differences in lung compliance and total resistance from one subject to another. Positive flow corresponds to expiration.

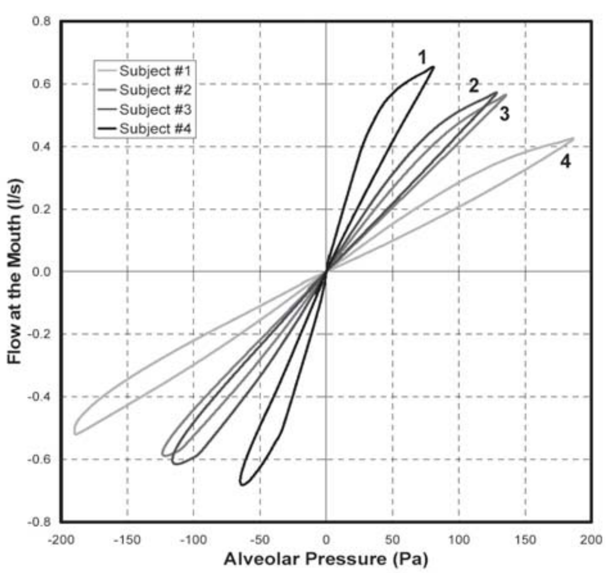

Fig. 11. Flow at the Mouth-Alveolar Pressure relation for Tidal Breathing

Subject #1 has the steepest curve because of its low small airway resistance value, in contrast with Subject #4 that has the highest value. Collapsible segment resistance has probably less influence because transmural pressure is kept in the region where resistance is small as are differences between subjects.

VII. VOLUME CALCULATION

It has been said that the simulation environment works on the basis of pressure-flow variables and consequently volume has to be calculated as the integral of flow.

This is the trickiest part of the simulation. On one side, the value of a definite integral depends on the integration limits but, on the other side, the way the circuit solver works is by applying a time step and calculating the next point as a function of the previous one, starting from zero. So the difficulty arises when we have to define integration limits because, unless something is done, the lower limit will always be zero and this may not correspond to the real situation.

We have implemented a way around this problem with the use of what we call a "gated integrator". The schematic of such a device is shown in Fig. 12.

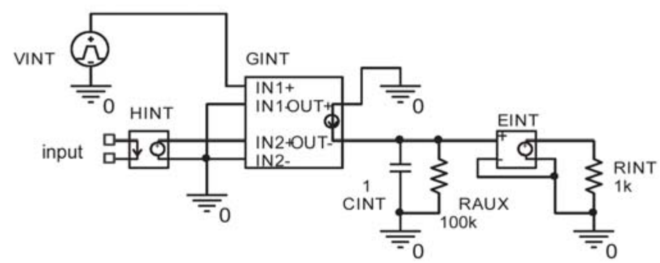

Fig. 12. Gated integrator

The gated integrator works as follows:

Capacitor C_{INT} charges up with the current through H_{INT}, which is the current to be integrated, for a time that in principle is the simulation time. Accordingly, the voltage through C_{INT} is the integral of input current along the simulation time that includes the initial transient to steady state. However, the "Multiplying voltage controlled current source" G_{INT} is in fact acting as a gate. If input *IN1+* is zero, output current will be zero and capacitor will not charge, whereas if it is 1V, capacitor will charge with input current. Then, by making V_{INT} initially zero and introducing a variable delay after which it becomes 1V, we will be able to set the initial integration time. Final time will always be the end of simulation, unless we take V_{INT} back to zero after a given time lapse. In this description we have used the term current but it should be read as flow in the pneumatic world.

In the preceding simulations we had two different situations:

FVC is a transient maneuver that starts from a steady state situation, goes from zero volume (in fact from residual volume) to maximum lung volume and lasts one respiratory cycle. Accordingly, to get the right value for the volume, we have to run some respiratory cycles (say three to four) to reach steady state and then perform the test, in a way that we start integrating at the beginning of the first steady state cycle and perform the integration for just one cycle. To do so we set the delay in V_{INT} at the end of the transient and end the simulation one cycle later.

On the other side, tidal breathing is completely different in the sense that it is a periodic event whose steady state is set by the mean value of the excitation, i.e., V_{off} . Thus, we have to perform the integration from the beginning and then make sure we reach a steady state, as seen in Fig. 5. In that case V_{INT} delay has to be very small (it cannot be zero for convergence reasons).

Apart from that, we can always introduce an initial condition to C_{INT} as has already been done for collapsible segment.

VIII. Conclusions

In the preceding paragraphs we have set up a model for Lung/Airway dynamics that is solved in PSPICE, a powerful electric circuit solver.

After developing analytical expressions for the model elements we have shown how to implement them by means of ABM building blocks and finally constructed a complete model with five lumps: two capacitors (elastances) and three resistors. Also we have worked out an excitation circuit to mimic the best the pleural pressure waveform while keeping consistency. Finally, a gated integrator has been built to calculate the lung and collapsible segment volume in a consistent way.

Results from simulations have not been compared to other author's because this is not the aim of this work. The present model, as any general model, does not try to mimic any particular situation, so the fitting accuracy to a particular experiment is not the question. The question is consistency of results with physical facts. A good fitting could eventually be obtained by adjusting parameter values, as in [8].

Being a general model that works in a standard solving environment as PSPICE, anyone can use it to its own purposes and even modify it at will to fit to a particular situation. Also, the fact that element functions can be defined through a table allows the model to be adapted more precisely to a particular set of experimental data, not forgetting that convergence to a solution may impose some restrictions.

Finally, the "what if" approach can easily be implemented and will be especially useful to investigate the influence of the different model parameters on the result and gain some insight into the way some characteristics of respiratory system can affect its functionality.

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