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REGULAR ARTICLE

A Simple Dynamic Model of Respiratory Pump

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Abstract To study the interaction of forces that produce chest wall motion, we propose a model based on the lever system of Hillman and Finucane (J Appl Physiol 63(3):951–961, 1987) and introduce some dynamic properties of the respiratory system. The passive elements (rib cage and abdomen) are considered as elastic compartments linked to the open air via a resistive tube, an image of airways. The respiratory muscles (active) force is applied to both compartments. Parameters of the model are identified in using experimental data of airflow signal measured by pneumotachography and rib cage and abdomen signals measured by respiratory inductive plethysmography on eleven healthy volunteers in five conditions: at rest and with four level of added loads. A breath by breath analysis showed, whatever the individual and the condition are, that there are several breaths on which the airflow simulated by our model is well fitted to the airflow measured by pneumotachography as estimated by a determination coefficient $R^2 \geq 0.70$. This very simple model may well represent the behaviour of the chest wall and thus may be useful to interpret the relative motion of rib cage and abdomen during quiet breathing.

Keywords Respiratory pump model · Respiratory inductive plethysmography · Pneumotachography rib cage · Abdomen

1 Introduction

The interaction of forces that produce chest wall motion is complex and not completely understood. This interaction has been studied for several years

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(Agostoni and Mognoni 1966; Gilbert et al. 1981; Goldman and Mead 1973; Konno and Mead 1967; Sharp et al. 1975; Wade 1954).

Mathematical models are used to understand these interactions and the mechanics of respiratory system better. Hillman and Finucane (1987) have produced a simple model of the respiratory pump that "appears to be appropriate for most breathing maneuvers and allows predictions to be made of the effect on chest wall of changes in applied forces". An advantage of this model compared with previous models (Macklem et al. 1979; Primiano 1982) and analyses (Loring and Mead 1982; Macklem et al. 1983; Mead et al. 1985) is that it provides a simple conceptual aid to understand how the motion of the chest wall observed at the body surface varies with changes in the active (muscle) and passive forces acting on it. The model is a lever system, represented by an imaginary bar on which forces act at four fixed sites. Although this model did not take into account the dynamic component of the system, it appears valid for different respiratory maneuvers and has the advantage of being simple and easy to use compared with other more complicated models of chest wall. One of these models (Macklem et al. 1983; Ward et al. 1992) divided the rib cage into two compartments, one that apposed to the lung and one that apposed to the diaphragm. Ricci et al. (2002) proposed a two-compartment model of the inspiratory pump, which used a realistic modelisation, distinguishing an active, an elastic and a viscous component. Their model parameters identification derived from actual measurement obtained by magnetic resonance imaging in normal humans.

We developed a simple dynamical model of respiratory pump based on the lever system of Hillman and Finucane (1987). We used experimental data provided by (1) respiratory inductive plethysmography (RIP), a noninvasive method of measurement of the rib cage and abdomen cross sectional area changes, which allows to estimate by a linear combination breathing volume changes (Konno and Mead 1967) and (2) pneumotachography, to calculate model parameters at rest and at four levels of added resistive load.

The fit between the airflow simulated by our model and the airflow measured by pneumotachography was estimated by the determination coefficient.

2 Methods

2.1 Description of the Model

Our model is based on the lever system proposed by Hillman and Finucane (1987) and introduces some dynamic properties of the respiratory system. Roughly, the respiratory system can be divided into the rib cage and abdominal elements, considered as passive, and the active respiratory muscles. The passive elements are considered to be elastic compartments linked to the open air via a resistive tube, an image of airways. The respiratory muscles force is applied to both compartments.

This simplified representation of the respiratory system is modeled in the following way (Fig. 1). The passive elastic rib cage and abdominal compartments are represented by two springs characterized by their elasticity (respectively E_{RC} and E_A). The resistances of respiratory system are represented by a dashpot

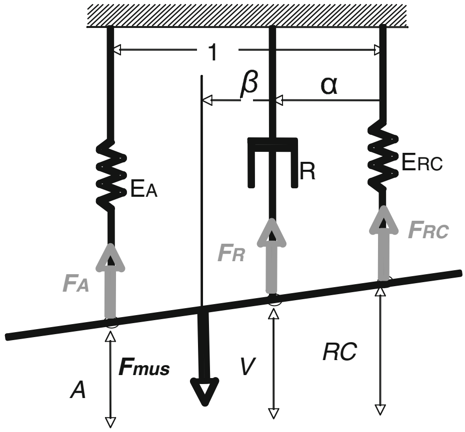

Fig. 1 Mechanistic representation of the respiratory pump

(characterized by its viscous resistance R). All three elements are attached to a rigid roof. The respiratory muscles force (Fmus) is applied to this system through a stiff bar without inertia sliding on the bottom ends of the springs and the piston.

Cross sectional area changes of abdomen (respectively rib cage) are represented by changes of the vertical position of the bottom end of the abdominal (respectively rib cage) spring A (respectively RC) as showed on Fig. 1. The vertical displacement of any point of the bar is a linear combination of A and RC. The dashpot is attached to the bar at the point where this linear combination corresponds to the one used for the estimation of breathing volume changes from rib cage and abdomen cross sectional area changes provided by respiratory inductive plethysmography (RIP) measurement. Then, the vertical displacement of the bottom end of the dashpot represents volume changes (V).

In reaction to the respiratory muscles force (Fmus), three forces are opposed: two elastic tension forces of the springs (F_{RC} and F_{A}) and the viscous resistance force (F_{R}) of the dashpot. Some simplification allows to minimize the number of parameters. The simplifications that have been taken into account are: the displacements are parallel and vertical, and the horizontal distance between abdominal and rib cage elements is normalized (=1). There are two geometric parameters (1) α the horizontal distance between the dashpot and the "rib cage" spring with $0 < \alpha < 1$ and (2) β the horizontal distance between the dashpot and the application point of Fmus, with $\alpha - 1 < \beta < \alpha$, and three mechanical parameters, namely rib cage (E_{RC}) and abdominal (E_{A}) elasticity and respiratory resistance (R).

2.2 Equations of the Model

By construction of the geometric properties of the model, vertical displacement of the bottom end of the dashpot is a linear combination of vertical displacements of the bottom end of abdominal (respectively rib cage) spring A (respectively RC) (Eq. 1).

$$V = \alpha A + (1 - \alpha)\mathrm{RC} \tag{1}$$

The viscous resistance force (F_R) is in the opposite direction to the direction of displacement of the lower end of the dashpot, and is the product of the simulated flow (dv/dt) and the viscous resistance of the dashpot (R) (Eq. 2).

$$F_R = -R\frac{\mathrm{d}V}{\mathrm{d}t} \tag{2}$$

The elastic tension forces of the springs (F_{RC} and F_A) are in the opposite direction of the deformation of the spring, and also the product of the elastances (E_{RC} and E_A) characterizing the springs and the displacement of the springs (Eqs. 3, 4).

$$F_{RC} = -E_{RC}\mathrm{RC} \tag{3}$$

$$F_A = -E_A A \tag{4}$$

In rotational dynamics, if a rigid body is submitted to several forces, the resulting moment of all forces exerted on the rigid body is equal to the product of its angular acceleration and its moment of inertia. In our model, we have stated that the bar has no inertia. It follows that the resulting moment of all forces on the bar is null, and this is true whatever the rotation point considered.

Then, the resulting moment of all forces applied on the bar relative to the application point of Fmus is null (Eq. 5).

$$\beta \mathrm{R}\frac{\mathrm{d}V}{\mathrm{d}t} + (\alpha + \beta - 1)(E_A A) + (\alpha + \beta)(E_{RC}\mathrm{RC}) = 0 \tag{5}$$

The derivative of V is consequently a linear combination of A and RC (Eq. 6) and, if the model is valid, in the true respiratory system the flow should be a linear combination of abdomen and rib cage cross sectional areas.

$$\frac{\mathrm{d}V}{\mathrm{d}t} = \frac{(1 - (\alpha + \beta))E_A}{\beta R}A - \frac{(\alpha + \beta)E_{RC}}{\beta R}\mathrm{RC}$$

which may be written

$$\frac{\mathrm{d}V}{\mathrm{d}t} = K_A A + K_{RC}\mathrm{RC} \tag{6}$$

with

$$K_A = \frac{(1 - (\alpha + \beta))E_A}{\beta R} \tag{7}$$

$$K_{RC} = -\frac{(\alpha + \beta)E_{RC}}{\beta R} \tag{8}$$

2.3 Validation with Experimental Data

Experimental data from recordings on eleven healthy volunteers (8 female, 3 male) were used to calculate the coefficients K_A and K_{RC} of Eq. 6. The healthy volunteers were between 19 and 55 years old (mean ± SD: 28.5 ± 12.6). The weight varied between 46 and 87 kg (mean ± SD: 61.6 ± 10.6) and the height was between 154 and 185 cm (mean ± SD: 168 ± 10).

Airflow was recorded with a pneumotachograph (Fleisch head no.1) placed on a face mask and a differential transducer (163PC01D36, Micro Switch). Mouth pressure was measured with a differential pressure transducer (142PC01D, Micro Switch). Leaks from around the mask were checked for before the recording was initiated using an infrared CO2 analyser (Engström Eliza/Eliza MC). Rib cage and abdominal signals were recorded by inductive plethysmography (Visuresp, RBI®).

Subjects were comfortably seated in a quiet room and were asked to relax and to breathe freely. The recording was performed with eyes open and lasted about 5 min for each of the five conditions: at rest and with four level of added loads. Data acquisition was started approximately 1 min after the addition of the resistance. Resistive loads were added throughout the entire breath and for each recording, the value of the resistance was calculated from the mouth pressure versus flow plot throughout the total recording. The mean value (±SD) for all resistances on the 11 subjects were at rest $R_0 = 0.75$ (±0.03) for the apparatus resistance, $R_1 = 3.26$ (±0.25), $R_2 = 5.27$ (±0.34), $R_3 = 8.23$ (±0.43) and $R_4 = 12.26$ (±0.63) cm H_2O l^{-1} s. The recordings were performed in a random order unknown to the subject and on two subjects (#4 and #10) only four recordings were performed (Calabrese et al. 2000).

K_A and K_{RC} of Eq. 6 were estimated breath-by-breath with a multiple linear regression using the airflow signal measured by pneumotachography, rib cage (RC) and abdominal (A) signals. Then the flow estimated by the model (dv^*/dt) (Eq. 6) and the determination coefficients R^2 were calculated breath-by-breath for the evaluation of the level of fit between experimental flow measured by pneumotachography and flow estimated by the model.

We have arbitrarily fixed at 0.70 the threshold above which we consider that a breath complies with the model ($R^2 \geq 0.70$). Figure 2 is an example of measured

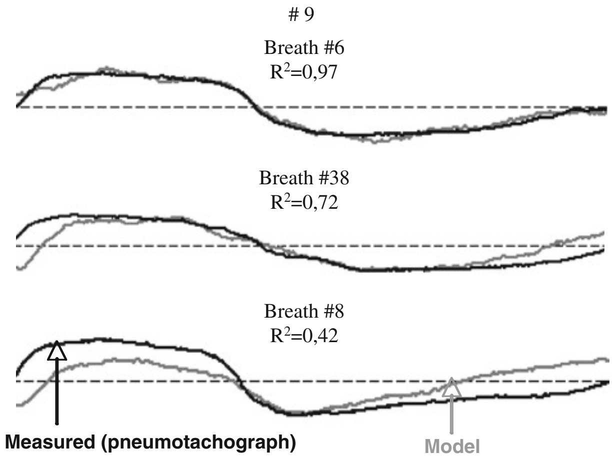

Fig. 2 Airflow measured by pneumotachography (*dark bar*) and airflow calculated with the model (*grey bar*) for subject 9 at rest (R_0) for 3 breaths yielding different values of determination coefficient R^2

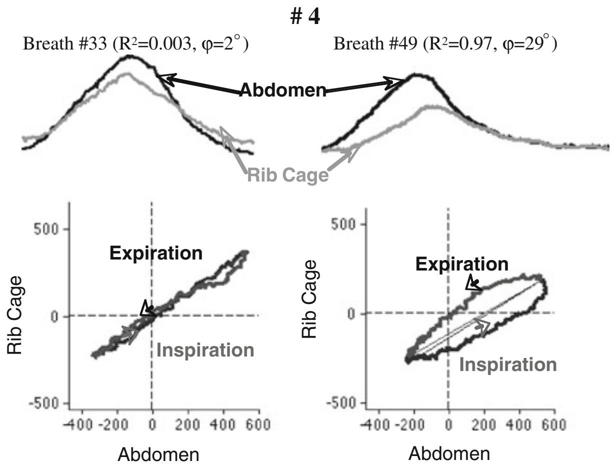

Fig. 3 Rib cage and abdomen signals recorded with RIP, and rib cage versus abdomen plots in arbitrary units. Two breaths are represented for subject 4 with the corresponding determination coefficient R^2 and phase difference φ

and reconstituted flows signals from five breaths with different R^2 of subject 9. Mean values of K_A and K_{RC} are calculated only on breaths with $R^2 \geq 0.70$.

Abdominal and rib cage signals from RIP on one hand and airflow from pneumotachography on the other hand are used to estimate K_A and K_{RC} values. However, if rib cage and abdomen signals are closely in phase, such parallel changes introduce indecision in the in the multiple linear regression calculation (badly conditioned matrix) and may decrease the determination coefficients R^2. Phase differences were therefore calculated from rib cage versus abdomen plots (Lissajous curves) in order to estimate the amount of these parallel changes (Agostoni and Mognoni 1966). The phase difference is expressed as angle (°). Indeed, in Fig. 3 where are presented rib cage versus abdomen plots for two breaths from one recording along with the value of R^2, one can see that a higher value of the phase difference (φ) is associated with a high value of R^2. For breath #33, the phase difference is 2° and the fit as evaluated by R^2 is very poor ($R^2 = 0.003$), whereas for breath #49 a high phase difference (29°) is associated with a good fit ($R^2 = 0.97$).

3 Results

The results of the fit between the recorded flow signal and the model flow signal are gathered in Table 1 for all subjects and for all five conditions. For each recording two groups have been considered according to the value of the determination coefficients R^2. For each resistance, the number of breaths in the group with $R^2 \geq 0.70$ followed by the number of breaths in the group with $R^2 < 0.70$ are given in the upper line. The mean value of phase difference (°) for each group is given in the lower line.

Table 1 Number of breaths with $R^2 \geq 0.70$ (first number) and with $R^2 < 0.70$ (second number)

	Subjects																					
	#1		#2		#3		#4		#5		#6		#7		#8		#9		#10		#11	
R_0																						
n	31	44	13	63	1	67	47	15	27	28	28	27	5	76	16	49	60	3	0	41	1	75
φ	*21*	*16*	*11*	*5*	*11*	*6*	*15*	*8*	*11*	*9*	*16*	*12*	*25*	*7*	*20*	*14*	*29*	*18*		*5*	*8*	*2*
R_1																						
n	23	40	7	67	2	68	49	23	34	17	13	25	1	72	10	41	58	5	[a]		43	7
φ	*17*	*10*	*12*	*8*	*22*	*6*	*15*	*9*	*19*	*13*	*18*	*15*	42	*12*	*18*	*13*	*34*	*25*			*14*	*9*
R_2																						
n	5	49	10	56	1	52	55	12	39	13	20	11	2	68	5	41	60	3	3	36	57	3
φ	*17*	*9*	*12*	*6*	*16*	*8*	*14*	*10*	*21*	*9*	*18*	*10*	*34*	*10*	*19*	*8*	*33*	*23*	*25*	*12*	*13*	*7*
R_3																						
n	4	36	11	51	4	60	53	14	41	7	33	26	0	64	5	34	53	4	9	28	33	4
φ	*19*	*10*	*12*	*6*	*20*	*8*	*20*	*12*	*19*	*10*	*22*	*16*		*8*	*19*	*8*	*38*	*29*	*33*	*21*	*13*	*8*
R_4																						
n	4	36	12	50	4	57	[a]		56	5	13	16	5	68	8	25	33	13	18	14	34	3
φ	*23*	*7*	*18*	*8*	*19*	*7*			*25*	*18*	*23*	*13*	*17*	*10*	*31*	*9*	*33*	*20*	*31*	*22*	*20*	*7*

Above are the corresponding mean values (in italics) of the phase difference (°) for all subjects and all resistances (R_0, R_1, R_2, R_3, R_4)

[a] Missing recording

The number of breath with $R^2 \geq 0.70$ is higher than $R^2 < 0.70$ in subjects #4, #5 and #9.

The number of breath with $R^2 \geq 0.70$ increases with the load in subjects #5, #10 and #11 (up to R^2 only).

On the whole, $R^2 \geq 0.70$ for 1,193 breaths while $R^2 < 0.70$ for 1,778 breaths.

One can see that for each condition and all subjects, the mean value of the phase difference is higher for the group of breath with $R^2 \geq 0.70$ than for the group of breath with $R^2 < 0.70$. Phase difference versus R^2 plots for each subject and all resistances show a significant positive correlation in all subjects.

Figure 4 is the plot of the mean values of K_A and K_{RC} calculated for each condition on those breaths with $R^2 \geq 0.70$. At a given condition K_A and K_{RC} have opposite signs. K_A and K_{RC} increase or decrease (according to their sign) with increasing added load approaching zero at the highest added load.

In our model, K_A/K_{RC} ratio is mathematically independant of R (from Eqs. 7 to 8):

$$-K_A/K_{RC} = \frac{(1-(\alpha+\beta))E_A}{(\alpha+\beta)E_{RC}} \tag{9}$$

We have calculated this ratio for each subject in all conditions, it appears that this ratio is roughly constant for subjects #2, 5, 10 and 11 (SD/mean < 17%). For subject # 1, 4, 6, 7 and 9, the variability is higher (24% < SD/mean < 40%). For

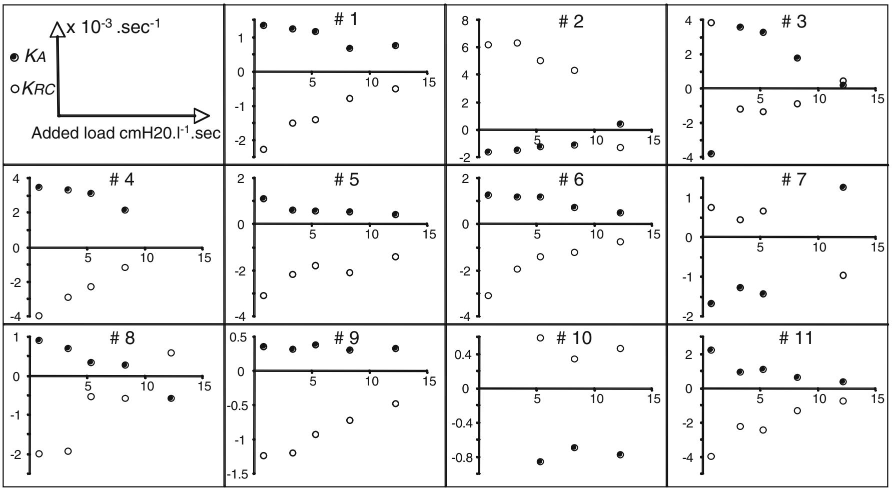

Fig. 4 Representation of K_A and K_{RC} (mean value on those breaths with $R^2 \geq 0.70$) for each added load condition and each subject

two subjects #3 and 8, the variability of K_A/K_{RC} ratio is very high (80 and 48% respectively).

4 Discussion

This study shows that 1,193 breaths on a total of 2,971 breaths (40%), recorded in eleven subjects in five conditions (rest and four level of resistive loading), validate the model. Then, our simple model represents satisfactorily (i.e. with a determination coefficient $R^2 \geq 0.70$) the behavior of the chest wall motion in a significant number of breaths.

Starting from the lever system of Hillman and Finucane (1987), we developed a simple dynamical model of the respiratory pump. This allowed to compare model simulation of airflow (obtained with abdominal and rib cage signals) with airflow measured by pneumotachography.

Many model of the respiratory system have been developed (Loring and Mead 1982; Macklem et al. 1979, 1983; Mead et al. 1985; Primiano 1982; Ricci et al. 2002). Our model is realistic because it takes into account the two compartments considered passive, rib cage and abdomen, and the active respiratory muscles, elastic and viscous forces in reaction to the respiratory muscle. The simplifications performed to minimize the number of parameters (i.e. parallel and vertical displacements, normalized horizontal distance between abdominal and rib cage elements) are minor. The geometric properties of the model (Eq. 1) are coherent with the Konno and Mead hypothesis which asserts that tidal volume is a linear combination of rib cage and abdomen cross sectional areas. The hypothesis of a system without inertia is a common assumption in respiratory mechanics.

The limits on the model are directly related to the hypothesis it includes: the geometric and mechanical properties of respiratory system are constant, at least over the breathing cycle. That mechanical properties remain stable over a respiratory cycle is a common assumption. That α and β remain constant during a respiratory cycle seems to be verified in a number of cycles where the model fits the data, but may be wrong for the others. The stability of α for a given subject is implicitly admitted as it is the condition of application of RIP technique. This leads to question about the stability of β: our results indicate that this parameter is stable for a number of cycles (those with $R^2 \geq 0.70$). The value of β depends upon the relative contributions of diaphragm, rib cage muscles and abdominal muscles to the act of breathing because their net action determines the relative motion of rib cage and abdomen. Because this net action can vary from breath to breath and from one condition to another, β may change from one breath to the other. Our results indicate, however, that β remains stable for some subjects (#2, 5, 10 and 11) presenting a low K_A/K_{RC} ratio variability.

Experimental data were obtained at rest and with the addition of four level loads throughout the entire breath. Adding ventilatory external loads has been used to simulate respiratory system disorders (Milic-Emili and Zin 1986). Although, resistive loading is not entirely analogous to internal respiratory loading induced by airway diseases it has often been used as a tool to characterize the compensatory mechanisms of the ventilatory system (Altose et al. 1979; Axen et al. 1983), the subject's tolerance to added loads (Freedman and Campbell 1970) and to study the sensations induced by these loads (Kelsen et al. 1981; Chonan et al. 1990). For Milic-Emili and Zin (1986) there are several reasons for studying loads: (1) elucidate basic physiology, (2) improve understanding and treatment of disturbances in patients with respiratory diseases and (3) assist in the design and use of breathing equipment for medical or other purposes. Another reason would be: explore respiratory muscles activity by non invasive techniques in conscious subjects (Gozal et al. 1995, 1996). Our study is clearly in line with the latter reason.

The threshold of the determination coefficient R^2 (estimation of the level of fit between experimental flow measured by pneumotachography and flow estimated by the model) has been arbitrarily fixed at 0.70. This value is rather high but it can be seen in Fig. 2 that the fit appears satisfactory at this level of threshold.

The number of breath with $R^2 \geq 0.70$ for each recording is gathered in Table 1. This number varies amongst subjects and conditions. No trend was observed with increasing resistances. The accuracy of K_A and K_{RC} calculation may explain this variability in the number of breath with $R^2 \geq 0.70$. Indeed, the accuracy of coefficient estimation is related to the difference between abdominal and rib cage signals. This difference can be estimated from the Lissajous curves and expressed as the phase difference. Figure 3 and Table 1 show that high phase differences were often associated with high R^2.

The variability in the number of breath with $R^2 \geq 0.70$ on one individual may have various origins. It can be explained in terms of fluctuations of the application point of the respiratory muscles force (Fmus) on the model's bar (Fig. 1). Indeed, when the position of the muscle force application point is at the bottom end of the dashpot, this results in parallel changes in RC and A signals, in other words a very low phase difference, inducing indecision in the multiple linear regression calculation (badly conditioned matrix). On the other hand, elastances are greatly influenced by thoracic muscle tone which may vary in the course of the breath (Josenhans et al. 1975). This may impair the adequacy of the model which relies on the hypothesis that elastances are constant.

Figure 4 show that estimated K_A and K_{RC} have always opposite signs as it is imposed on theoretical values by Eq. 6. Furthermore, the coefficients increase or decrease (according to their sign) when the added load increases (except for subject #7 for both coefficients and for subject #9 for K_A). This is in agreement with our model hypotheses: Eqs. 7 and 8 indicate R is inversely proportional to K_A and K_{RC}, on the hypothesis of E_A and E_{RC} constant.

RIP is often used to obtain either a measure of tidal volume or the rib cage and abdomen signals indicating the contribution of the two compartments to ventilation. This remains valid during different respiratory conditions, as hyperventilation (Calabrese et al. 2007) and especially during loaded breathing (Carry et al. 1997; Eberhard et al. 2001).

RIP measurements and our model are complementary to understand chest wall motion. RIP is largely used to estimate thoracoabdominal asynchrony that is considered to be noninvasive indicators of airway obstruction (Hammer et al. 1992; Sackner et al. 1984; Cantineau et al. 1992). The hypothesis that thoracoabdominal asynchrony is related with airway obstruction, is in accordance with our model: if the resistance of respiratory system (R) increases, the viscous resistance force (F_R) of the dashpot applied on the bar is stronger involving rib cage and abdomen asynchrony (elastances E_A and E_{RC} supposed constant).

In conclusion, our model proved to be able to represent the behavior of the chest wall and thus may be useful to interpret the relative motion of rib cage and abdomen during quiet breathing. The model is simple in its conception and easy to use since three signals airflow (pneumotachography) and abdominal and rib cage signals (respiratory inductive plethysmography) are needed to calculate the model's parameters. The model could be a useful and complementary tool to experimental data to understand respiratory mechanics.

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