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ESTIMATION OF PRESSURE DROP IN PEDIATRIC ENDOTRACHEAL TUBES DURING HFPV

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ABSTRACT

High-frequency percussive ventilation (HFPV) is a ventilation modality which has been proved useful as an alternative to conventional mechanical ventilation. In clinical practice the ventilator measures the pressure that represents the sum of the pressure drop due to the endotracheal tube (ΔP_{EET}) and the pressure dissipated to inflate lung. From the clinical point of view, it is of paramount importance to estimate the real amount of ΔP_{EET}.

This study aimed at identifying in vitro the most adequate model and parameters for estimating ΔP_{EET} of pediatric endotracheal tube during HFPV, under different working pressures, percussive frequencies and resistive and elastic lung loads.

The results show that it is possible to estimate ΔP_{EET} in pediatric endotracheal tubes by using a simple Blasius' model, considering the presence of inertance. The Blasius' model presents the same estimation error as Rohrer's model and its coefficients result largely independent from ventilator settings and lung loads.

Index Terms**—** Tube modeling, Endotracheal tube, Parametric estimation, HFPV, Pediatrics

1. INTRODUCTION

Endotracheal tubes (EET) are regularly used in clinical practice to connect the artificial ventilator to the airway of a patient subject to a mechanical ventilation. However, its presence involves an extra mechanical load to the total respiratory system impedance, causing different pressure values at the proximal and the distal end of the EET [1] [2]. The airway pressure supplied by ventilator represents the sum of the endotracheal tube pressure drop (ΔP_{EET}) and the pressure dissipated to inflate lung. From the clinical point of view, it is of paramount importance to take into account the real amount of pressure dissipated by endotracheal tube to avoid baro and volutrauma [3].

The pressure drop in adult and pediatric endotracheal tubes during conventional mechanical ventilation (CMV) has been widely studied [4-8] and several approaches have been developed to estimate the ΔP_{EET}. Under certain conditions, e.g. laminar flow, the pressure-flow relationship characterizing tracheal tubes may be considered linear [4-5]:

$$\Delta P_{EET}(t) = R_{tube}\dot{V}(t) \tag{1}$$

where, at any time (t), $\dot{V}(t)$ represents flow and R_{tube} is the flow resistance coefficient.

However, in most cases a non-linear tube pressure-flow relationship has been found. In such cases this relationship has been frequently approximated by:

$$\Delta P_{EET}(t) = K_1\dot{V}(t) + K_2\dot{V}(t)|\dot{V}(t)| \tag{2}$$

where, at any time (t), $\dot{V}(t)$ represents the flow, and K_1 and K_2 are the Rohrer's constants [6]. Another approach to estimate in vitro adult and neonatal ΔP_{EET} was proposed by Blasius that suggested a formula for circular tubes [7-8]:

$$\Delta P_{EET}(t) = K_B\dot{V}(t)|\dot{V}(t)|^{0.75} \tag{3}$$

where, at any time (t), $\dot{V}(t)$ represents the flow and K_B is the constant that depends on the inner geometry of the tube and the physical properties of the gas.

During high frequency ventilation (HFV) the aforementioned approaches do not properly describe the pressure-flow relationship in tracheal tubes, possibly because of the turbulence and the presence of mechanical inertance (I). In fact, the latter was taken into consideration under high-frequency oscillatory ventilation (HFOV) [9].

High-frequency percussive ventilation (HFPV) is a non conventional ventilatory strategy that associates the beneficial aspects of CMV with those of high-frequency ventilation. HFPV acts as a rhythmic (sinusoidal) cyclic ventilation with physically servoed flow regulation, which produces a controlled staking tidal volume by pulsatile flow. This particular type of ventilation is a form of HFV that delivers a series of high-frequency sub-tidal volumes in combination with low-frequency breathing cycles [10-12].

Although HFPV has been increasingly used, with encouraging results, also in the pediatric clinical practice, to our knowledge the pressure drop ΔP_{EET} of pediatric endotracheal tubes under this type of ventilation has not been described yet. Thus, we aimed at characterizing in vitro the pressure drop ΔP_{EET} of a pediatric endotracheal tube, by identifying the model that best fits its pressure-flow relationship during HFPV under different working pressures, percussive frequency and imposed resistive and elastic lung loads.

2. MATERIAL AND METHODS

2.1. Experimental setup

Figure 1 shows the experimental setup used in this study.

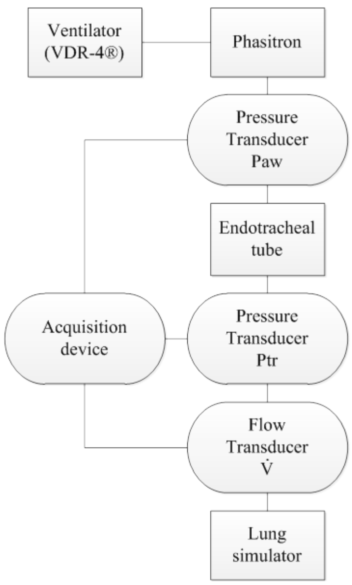

Figure 1. Box diagram of the experimental setup.

A physical model of respiratory system was provided by a single-compartment lung simulator (ACCU LUNG, Fluke Biomedical, USA). In our measurements the lung simulator was set according to the combinations of resistive loads (R) 5 and 20 $cmH_2O/L/s$ and elastic loads (E) 20, 50, and 100 cmH_2O /L.

High Frequency Percussive Ventilation was provided by a volumetric diffusive respirator (VDR-4®; Percussionaire Corporation, USA) that delivers mini-bursts of respiratory gas mixtures in the proximal airways by the Phasitron®, that is the heart of this kind of ventilation [10-11]. In this experimental setup, pulsatile flow was delivered during inspiratory phase (In) while expiratory phase (Ex) was completely passive. The VDR-4® ventilator was set to deliver a pulse inspiratory/expiratory (i/e) duration ratio of 1:1, and inspiratory and expiratory duration (TIn and TEx) ratio of 1:1 [10-11]. Work pressure (P_{work}) was varied from 20 to 45 cmH_2O with increasing steps of 5 cmH_2O. The percussive frequency (f_p) was set to 300, 500 and 700 cycles/min. A ventilator circuit was connected via a dedicated EET connector (ID 15 mm) to the examined EET (size 5.5, Rusch, Germany, ID = 5.5 mm, length = 29.5 cm), the distal part of which is linked to the described physical model of respiratory system (Figure 1).

The respiratory signals (flow and pressures) were acquired by a dedicated acquisition system [13]. The measurement of flow signal $\dot{V}(n)$ was performed using Fleisch pneumotachograph (Type 2, Switzerland) connected to a differential pressure transducer (0.25 INCH-D-4V, All Sensors, USA). The pressure signals were measured with pressure transducers placed before EET connector ($P_{aw}(n)$ - airway pressure) and at the end of EET ($P_{tr}(n)$ - tracheal pressure). Both transducers (ASCX01DN, Honeywell, USA) were connected to the respiratory circuit by identical connectors (ID 20 mm).

Measurements of respiratory signals were performed for the 108 possible combinations of resistive R and elastic E loads, frequencies, and work pressures during three successive respiratory cycles. Data were acquired at a sampling frequency of 2000Hz with 12 bit resolution (PCI-6023E, National Instruments, USA). $\Delta P_{EET}(n)$ was calculated by subtracting, at every sampling instant n, $P_{tr}(n)$ from $P_{aw}(n)$.

2.2. ΔP_{EET} models

In this study we tried to characterize pressure drop during the inspiratory phase of respiratory cycle by parametric identification of coefficients of three models defined in the discrete form by the following equations:

Model 1: $$\Delta P_{EET}(n) = R_{tube}\dot{V}(n) + I\ddot{V}(n) \quad (4)$$

Model 2: $$\Delta P_{EET}(n) = K_1\dot{V}(n) + K_2\dot{V}(n)|\dot{V}(n)| + I\ddot{V}(n) \quad (5)$$

Model 3: $$\Delta P_{EET}(n) = K_B\dot{V}(n)|\dot{V}(n)|^{0.75} + I\ddot{V}(n) \quad (6)$$

Model 1, described by the equation (4), is characterized by a linear resistance term and an inertance term. Model 2, defined by the equation (5), takes into consideration a non linear Rohrer's resistance and an inertance term. In the Model 3, defined by the equation (6), the first term represents the pressure drop caused by air friction (in analogy with the non linear Blasius' expression) and the second term corresponds to the pressure drop due to inertial effects. In this study we assumed that estimated coefficients include both distributed and concentrated pressure drops.

2.3. Estimation of model parameters

In order to estimate the parameters of the proposed models, we used the least squares method. The estimation of the

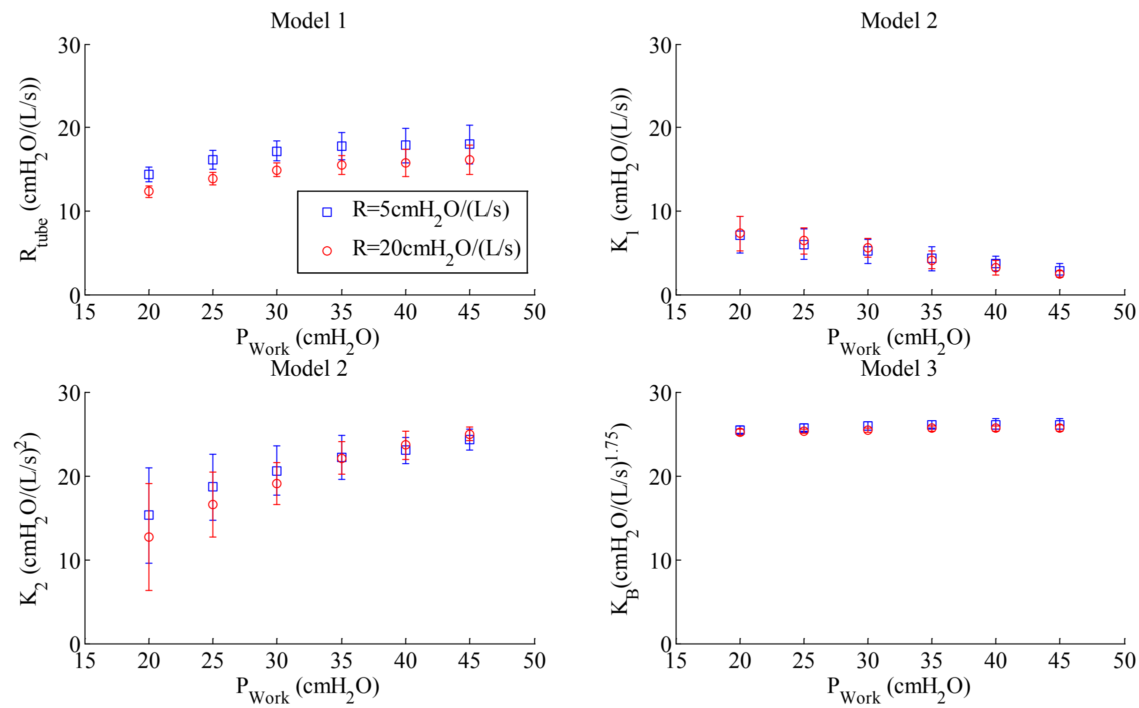

Figure 2. Mean±1SD of the parameters of the three proposed models versus P_{work}.

parameter vectors $\hat{\theta}_1$, $\hat{\theta}_2$ and $\hat{\theta}_3$ for the three Models were defined as:

$$\hat{\theta}_i = (A_i^T \times A_i)^{-1} \times A_i^T \times B \qquad i=1,2,3 \qquad (7)$$

with:

$$\hat{\theta}_1 = [R_{tube}\ I]^T$$

$$\theta_2 = [K_1\ \ K_2\ \ I]^T$$

$$\hat{\theta}_3 = [K_B\ \ I]^T$$

$$A_1 = \begin{bmatrix} \dot{V}(0) & \ddot{V}(0) \\ \dot{V}(1) & \ddot{V}(1) \\ \vdots & \vdots \\ \dot{V}(N-1) & \ddot{V}(N-1) \end{bmatrix} \qquad (8)$$

$$A_2 = \begin{bmatrix} \dot{V}(0) & \dot{V}(0)\cdot|\dot{V}(0)| & \ddot{V}(0) \\ \dot{V}(1) & \dot{V}(1)\cdot|\dot{V}(1)| & \ddot{V}(1) \\ \vdots & \vdots & \vdots \\ \dot{V}(N-1) & \dot{V}(N-1)\cdot|\dot{V}(N-1)| & \ddot{V}(N-1) \end{bmatrix}$$

$$A_3 = \begin{bmatrix} \dot{V}(0)\cdot|\dot{V}(0)|^{0.75} & \ddot{V}(0) \\ \dot{V}(1)\cdot|\dot{V}(1)|^{0.75} & \ddot{V}(1) \\ \vdots & \vdots \\ \dot{V}(N-1)\cdot|\dot{V}(N-1)|^{0.75} & \ddot{V}(N-1) \end{bmatrix}$$

$$B = [\Delta P_{EET}(0) \quad \Delta P_{EET}(1) \quad \dots \quad \Delta P_{EET}(N-1)]^T$$

The residual root mean square error was calculated by means of the equation:

$$\mathrm{RMSE} = \sqrt{\frac{\sum_{n=0}^{N-1}[\Delta \mathrm{P_{EETm}}(n) - \Delta \mathrm{P_{EETe}}(n)]^2}{N}} \qquad (9)$$

where ΔP_{EETm} is the ΔP_{EET} measured, ΔP_{EETe} is the ΔP_{EET} estimated and N is the total number of samples.

3. RESULTS

The estimated inertance (I) was identical in all three models 0.126±0.009 $cmH_2O/L/s^2$ and did not present particular dependence on working conditions.

Figure 2 shows the mean and standard deviations of the parameters of the three proposed models versus P_{work}, obtained for different combinations of lung elastances and percussive frequencies. The values R_{tube} were higher at lower lung resistance R; K_2 parameter also showed a similar behavior up to 30 cmH_2O of P_{work}. On the other hand, K_1 and K_B parameters did not show a significant dependency on lung resistance. However, in the Model 1, R_{tube} slightly increased as the P_{work} increased from 20 to 35 cmH_2O. In the Model 2 K_1 decreased and K_2 increased as P_{work} increased. Finally, in the Model 3, K_B was independent from P_{work} at all the considered experimental setup conditions. The parameters of Model 1 and Model 2 (especially K_2) presented a variability (SD shown in Figure

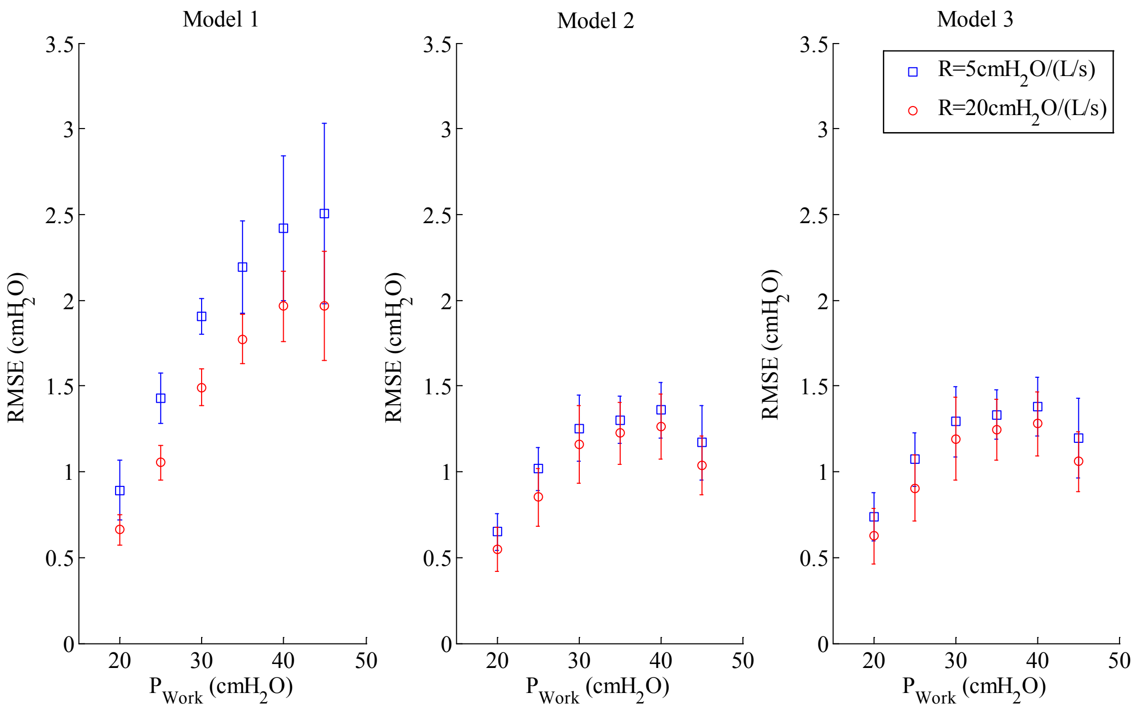

Figure 3. Mean ±1SD RMSE errors of the ΔP_{EET} estimation in the three Models in function of P_{work}.

2) due to the different lung elastances and percussive frequencies, even if without a clear relationship.

Figure 3 shows the mean values (±1SD) of the RMSE errors of the ΔP_{EET} estimation in the three Models in function of P_{work}. Model 1 presented higher values of RMSE for lower value of lung resistance, while in Model 2 and Model 3 these differences were reduced. RMSE errors of Model 1 were directly proportional to P_{work}, while in Model 2 and Model 3 the errors increased from P_{work} of 20 to 35 cmH_2O and decreased at a P_{work} of 45 cmH_2O. In all three models RMSE errors varied with percussive frequency and lung elastance but without a definite relationship with these variables. The linear Model 1 presented RMSE values significantly higher (from 18% at 20 cmH_2O up to 98% at 45 cmH_2O) than those obtained from Rohrer's Model 2 and Blasius' Model 3.

In order to evaluate the slight differences between Models 2 and 3, we plotted the RMSE errors of Model 2 versus the RMSE errors of Model 3 (Figure 4) using all the 108 setup combinations. The Model 2 proved to be slightly better that Model 3 at every setup combination.

4. DISCUSSION

In this study we aimed at identifying the most adequate model and parameters for estimating the pressure drop of pediatric endotracheal tube ΔP_{EET} during HFPV under different working pressures, percussive frequencies and resistive and elastic lung loads.

We studied the linear Model 1 and the non linear Models 2 and 3, all of them taking into account inertance and both distributed and concentrated pressure losses. The estimated inertance values were identical in all the models (0.126±0.009 $cmH_2O/L/s^2$) and slightly smaller than theoretical value (0.152 $cmH_2O/L/s^2$) for the examined EET.

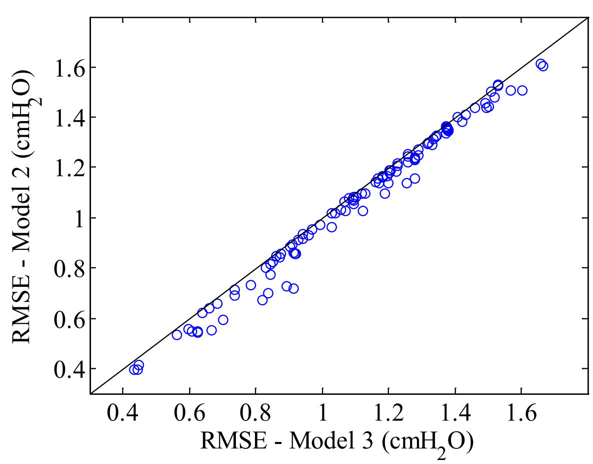

Figure 4. RMSE errors of Model 2 versus the RMSE errors of Model 3.

The linear Model 1 presented RMSE values significantly higher than those obtained from the two non-linear Models, especially at the highest working pressures. This is probably due to the fact that Model 1 does not consider the presence of turbulent flow, that introduces non linearity in pressure-flow relationship. This phenomenon particularly occurs in presence of high values of pulsatile flow; thus we considered the linear model unable to describe properly ΔP_{EET} during HFPV.

In Model 2 coefficients K_1 and K_2 showed strong dependency on P_{work} and in addition they presented a meaningful variability related to different lung elastances and percussive frequencies. The large variability of Rohrer's coefficients was also present in the case of HFOV, as reported in [9].

On the other hand, results showed that the Blasius' Model 3 coefficient K_B was widely independent from the setup conditions as well as from the P_{work}, differently from the parameters of the other models, even if it presented a very slightly higher estimation error than Model 2. Thus this model seems to be the best approximation of the non-linear behavior of the considered EET.

5. CONCLUSIONS

This study offers the possibility to estimate pediatric EET pressure drop by using a simple Blasius' model also evaluating the presence of inertance. Blasius' model coefficient resulted strongly independent from ventilator settings and lung loads and presented the same estimation error of the Rohrer's model, whose parameters were instead affected by a large variability. This model may be implemented as a clinical tool for the estimation of ΔP_{EET} and consequently for the evaluation of tracheal pressure signal during HFPV. The possibility of non invasive continuous monitoring of EET pressure drop will increase the use of this ventilation modality.

6. ACKNOWLEDGEMENTS

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7. REFERENCES

[1] A.D. Bersten, A.J. Rutten, A.E. Veding, and G.A. Skowronski, "Additional work of breathing imposed by endotracheal tubes, breathing circuits, and intensive care ventilators," *Crit. Care Med.,* vol. 17 pp. 671-677, 1987.

[2] P.M. Bolder, T.E.J. Healy, A.R. Bolder, P.C. W. Beatty, and B. Kay, "The extra work of breathing throughout adult endotracheal tubes," *Anaesth. Analg.* Vol. 65, pp. 853–859, 1986.

[3] P.R.M. Rocco, and W.A. Zin, "Modelling the mechanical effects of tracheal tubes in normal subjects," *Eur. J. Physiol.* vol. 8, pp. 121-126, 1995.

[4] W.A. Zin, L.D. Pengelly, and J.Milic-Emili, "Active impedance of respiratory system in anesthetized cats," *J. Appl. Physiol.* vol. 53, pp. 149-157, 1982.

[5] M.G. Conti, R.A. De Blasi, A. Lappa, A. Ferretti, M. Antonelli, M. Bufi, and A. Gasparetto, "Evaluation of respiratory system resistance in mechanically ventilated patients: the role of the endotracheal tube," *Intens. Care Med.* vol. 20, pp. 421-424, 1994.

[6] F. Rohrer, "Der Stroemungswiderstand in den menschlichen Atemwegen und der Einfluss der unregelmaessigen Verzweigung des Bronchialsystems auf den Atmungsverlauf in verschiedenen Lungenbezirken," *Pfluegers Arch. Gesamte Physiol. Menschen Tiere* vol. 162, pp. 225-259, 1915.

[7] F. Lofaso, B. Louis, L. Brochard, A. Harf, and D. Isabey, " Use of the Blasius resistance formula to estimate the effective diameter of endotracheal tubes," *Am. Rev. Respir. Dis.* vol. 146, pp. 974-979, 1992.

[8] P.H. Jarreau, B. Luis, G. Dassieu, L. Desfrere, P. W. Blanchard, G. Moriette, D. Isabey, and A. Harf, " Estimation of inspiratory pressure drop in neonatal and pediatric endotracheal tubes," *J. Appl. Physiol*. vol. 87, pp. 36-46, 1999.

[9] S. Shumann, M. Krappitz, K. Moeller, R. Hentschel, G. Braun, J. Guttmann, " Pressure loss caused by pediatric endotracheal tubes during high frequency oscillation ventilation," *Respir. Physiol. Neurobiol*, vol. 162, pp. 132-137, 2008.

[10] U. Lucangelo, V. Antonaglia, W.A. Zin, L. Fontanesi, A. Peratoner, F.M. Bird, and A. Gullo, " Effects of mechanical load on flow, volume and pressure delvered by high-frequency percussive ventilation," *Respir. Physiol. Neurobiol,* vol. 142, pp. 81-91, 2004.

[11] U. Lucangelo, V. Antonaglia, W.A. Zin, G. Berolt, L. Fontanesi, A. Peratoner, F. Bernabè, and A. Gullo, " Mechanical load modulate tidal volume and lung washout during high-frequency percussive ventilation," *Respir. Physiol. Neurobiol,* vol. 150, pp. 44-51, 2006.

[12] U. Lucangelo, A. Accardo, A. Bernardi, M. Ferluga, M. Borelli, V. Antonaglia, F. Riscica and W.A. Zin, " Gas distribution in a two-compartment model ventilated in high-frequency percussive and pressure-controlled modes," *Intens. Care. Med.* vol. 36, pp. 2125-2131, 2010.

[13] F. Riscica, U. Lucangelo, M. Ferluga, and A. Accardo, "In vitro meas-urements of respiratory mechanics during HFPV using a mechanical lung model," *Physiol. Meas.*32, pp. 637-648, 2011.