ESTIMATION OF THE DIFFERENCE BETWEEN PEAK AIRWAY AND TRACHEAL PRESSURES DURING HFPV

Miloš Ajčević[1], Umberto Lucangelo[1,2] and Agostino Accardo[1]

[1]Dept. of Engineering and Architecture, University of Trieste, Via Valerio 10, 34127 Trieste, Italy
[2]Dept. of Perioperative Medicine, Intensive Care and Emergency, Cattinara Hospital, University of Trieste, Strada di Fiume 447, 34149 Trieste, Italy

ABSTRACT

High frequency percussive ventilation (HFPV) is an advanced ventilatory strategy which has proven very effective in patients with acute respiratory failure. The airway pressure measured by HFPV ventilator represents the sum of the endotracheal tube pressure drop and the tracheal pressure dissipated to inflate a lung. The estimation of the difference between the peak airway and tracheal pressure ΔP_p may be very useful to the clinician to avoid lung injury. The aim of this study is to provide an *in vitro* estimation of ΔP_p based only on the ventilator set parameters (i.e. peak pressures, pulsatile frequencies) and the patient's respiratory system resistance and compliance. The model for the estimation of ΔP_p was determined by using the Least Absolute Shrinkage and Selection Operator (LASSO) regularized least-squares regression technique. The identified model was successively assessed on test data set.

Index Terms— HFPV, respiratory signal processing, model identification, LASSO, endotracheal tubes

1. INTRODUCTION

High frequency percussive ventilation (HFPV) is a non-conventional ventilatory strategy which associates the beneficial aspects of conventional mechanical ventilation (CMV) with those of high-frequency ventilation (HFV) [1]. HFPV acts as a rhythmic cyclic ventilation with physically servoed flow regulation, which produces a controlled staking tidal volume by pulsatile flow [2]. This particular high frequency ventilation modality delivers a series of high-frequency sub-tidal volumes, by pulsatile flow (200-900 cycles/min), in combination with low-frequency breathing cycles (10-15 cycles/min).

HFPV was developed and introduced by F. M. Bird in the early 1980's and initially used for the treatment of burn patients with acute respiratory failure caused by burns and smoke inhalation [3]. Over the years, HFPV has proven highly useful in the treatment of several different pathological conditions: closed head injury [4], newborns with hyaline membrane disease and/or acute respiratory distress syndrome (ARDS) [5], patients with severe gas exchange impairment [6]. The efficacy of HFPV has been demonstrated also in removing bronchial secretions under diverse conditions [7].

Endotracheal tubes (ETT) are regularly used in clinical practice to connect an artificial ventilator to the airway of a patient undergoing 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 ETT [8]. The airway pressure measured (Paw) by ventilator (Figure 1) represents the sum of the endotracheal tube pressure drop (ΔP) and the tracheal pressure (Ptr) dissipated to inflate lung. Hence, peak pressure (Paw_{peak}) measured by the ventilator does not correspond to the peak pressure actually present in the patient's trachea (Ptr_{peak}). Measurement of Ptr during HFPV cannot be done invasively in every day clinical practice. The estimation of ΔP may be very useful to the clinician, especially to avoid the development of barotrauma and an erroneous estimate of the volume delivered to the patient [9].

The ETT pressure-flow relationship was widely studied in conventional [9,10] and high frequency ventilation [11-12]. Several approaches for estimation of ΔP based on flow dependent model and flow measurement have been developed [9-12]. Recently, the Blasius flow dependent model was identified as the most adequate for the pressure drop $\Delta P(t)$ estimation over time in adult [11] and pediatric [12] endotracheal tubes during HFPV.

Currently, the HFPV ventilator measures only airway pressure and corresponding Paw_{peak}, while flow measurement is not provided. For a correct $\Delta P(t)$ estimation over time, the flow measurement is mandatory, but in some clinical scenarios this measurement may be difficult to perform. In these cases, estimation of difference between airway and tracheal peak pressure ΔP_p may represent an alternative approximate estimation of $\Delta P(t)$ over time. ΔP_p flow independent model should be based on ventilatory parameters, which values are known at the beginning of the HFPV treatment.

HFPV is characterized by a pulsatile flow delivery that can be varied by changing the device's peak pressure (Paw_{peak}) and percussive frequency (f). Furthermore, the performance of the HFPV varies according to the physiological/physical feedback, i.e., resistive (R) and compliance (C) parameters of the respiratory system of the ventilated subject [2].

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The aim of this study is to provide an *in vitro* estimation of ΔP_p based only on the ventilator set parameters (i.e. peak pressures, pulsatile frequencies) and the patient's respiratory system resistance and compliance.

2. MATERIAL AND METHODS

The experimental setup used in this study is shown in figure 1.

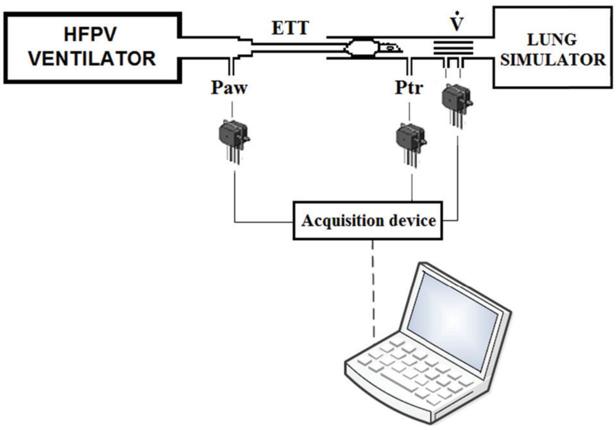

Fig. 1. Schematic diagram of respiratory circuit and acquisition system.

HFPV ventilator (VDR-4®, Percussionaire Corporation, USA) was connected via endotracheal tube (size 8, Rusch, Italy, ID = 8 mm, length = 33cm) and flow transducer to the physical model of the respiratory system provided by a single-compartment lung simulator (ACCU LUNG, Fluke Biomedical, USA). The airway Paw(t) and tracheal Ptr(t) pressures were measured by pressure transducers (ASCX01DN, Honeywell, USA) placed at the proximal and distal part of ETT, respectively (Figure 1). Flow signal $\dot{V}(t)$ was measured using Fleisch pneumotachograph (Type 2, Lausanne, Switzerland) linked to a differential pressure transducer (0.25 INCH-D-4V, All Sensors, Morgan Hill, CA, USA).

The VDR-4® HFPV ventilator delivered a pulse flow with pulse inspiratory/expiratory (i/e) duration ratio of 1:1, and overall inspiratory to expiratory (I/E) time ratio of 1:1. The peak inspiratory pressure measured by the VDR-4® ventilator, was progressively increased in 5 cmH_2O steps from 20 to 45 cmH_2O, resulting in 6 Paw_{peak} values. The percussive frequency (*f*) was set to 300, 500 and 700 cycles per minute, corresponding to 5, 8.33 and 11.67 Hz, respectively. The lung simulator was set according to all the combinations of lung resistance (R): 5, 20 and 50 $cmH_2O\ L^{-1}$ s and compliance (C): 10, 20, and 50 mL cmH_2O^{-1}. Lung simulator loads were chosen to represent normal subjects, obstructive and restrictive patients, and ventilator settings were chosen to represent usual clinical application range.

Measurements of respiratory signals were performed for all possible 162 combinations of resistances, compliances, frequencies, and peak pressures during a respiratory cycle. For each measurement setting two respiratory cycles were recorded in order to create two data sets. The first data set was used for the parameter estimation and cross-validation (training set), while the second data set was used to test model on new unseen data (test set) Data were acquired at a sampling frequency of 2kHz with 12 bit resolution (PCI- 6023E, National Instruments, USA). Respiratory signals were filtered with low-pass second order Butterworth filter at a cut-off frequency of 35 Hz. ΔP_p was calculated by subtracting for each respiratory cycle Ptr_{peak} from Paw_{peak} (Figure 2).

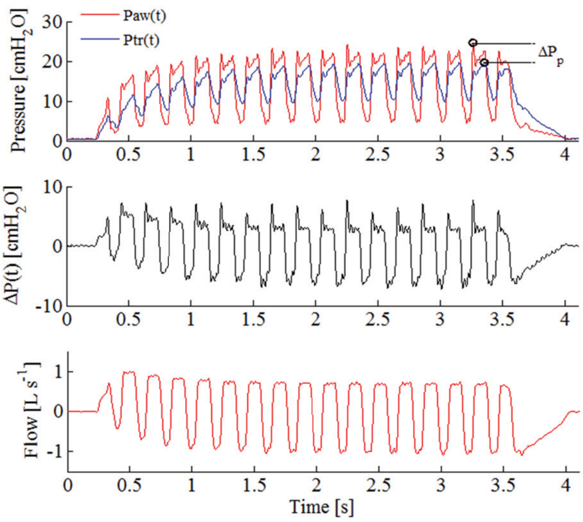

Fig. 2. From top to bottom: airway and tracheal pressure (peak values are marked with circles), measured endotracheal tube pressure drop and flow tracing during a single respiratory cycle.

In order to estimate ΔP_p we proposed a quadratic model including all terms (linear, interactions, squared and intercept) of the Paw_{peak}, *f*, R and C variables (Table 1). Model parameters were estimated by regularized least-squares regression using Least Absolute Shrinkage and Selection Operator (LASSO). LASSO minimizes the cost function consisting of residual sum of squares (RSS) and of a regularization term over the model parameter vector [13]:

$$\hat{\vartheta} = argmin_{\hat{\vartheta}} \left(RSS(\hat{\vartheta}) + \lambda \sum_{j=1}^{p} |\hat{\vartheta}_j| \right). \qquad (1)$$

where $\hat{\vartheta}$ is a parameter vector, p is number of coefficients and λ is a parameter which controls the model complexity and prevents the coefficients of the linear model from having large absolute values, in order to prevent overfitting. In order to minimize cross validation mean square error (MSE), λ was chosen using 10-fold cross validation [14].

The identified model was tested on a previously unseen test dataset. The predicted ΔP_p values were compared to measured ΔP_p by means of Root Mean Square Error (RMSE) and the coefficient of determination (R^2).

The RMSE error values were also compared to those obtained by using the Blasius model that derives ΔP_p values from the peak values of tracheal pressure estimate Ptr(t)=Paw(t)-ΔP(t). ΔP(t) was calculated using the following Blasius equation [11]:

$$\Delta P(t) = 5.57 \cdot \Delta\dot{V}(t) \cdot \left|\Delta\dot{V}(t)\right|^{0.75} + 0.081 \cdot \ddot{V}(t) \quad (2)$$

where $\dot{V}$ is the measured flow, $\ddot{V}(t)$ is the volume acceleration calculated by numerical differentiation of flow. The model coefficients, for a tube of size 8, were estimated in a previous study [11].

3. RESULTS

The identified model and the corresponding coefficients (Table 1) were determined by regularized least squares regression using LASSO.

Model terms	Estimated coefficients
Paw_{peak}	0.25
f	1.35
R	-0.36
C	0.25
Paw_{peak}^2	-0.0043
f^2	-0.042
R^2	0.003
C^2	-0.0023
$Paw_{peak} \cdot f$	-0.01
$Paw_{peak} \cdot R$	0.0039
$Paw_{peak} \cdot C$	-0.0029
$f \cdot R$	-0.0053
$f \cdot C$	0.0007
$R \cdot C$	0.012
Intercept	-3.52

Table 1. Model terms and corresponding estimated coefficients.

	Quadratic Model		Blasius model
	Training set	Test set	Test set
RMSE [cmH_2O]	1.16	1.17	1.27
R^2	0.81	0.81	0.76

Table 2. Calculated RMSE and R^2 values.

In Figure 3 are plotted measured ΔP_p values versus those predicted by the identified quadratic model on training set, for all experimental setup combinations. The results showed dispersion around identity line between measured and predicted values. The estimation error was lower than 1 cmH_2O in 82% of cases (Figure 3). For higher values of ΔP_p a slight underestimation trend was detected.

In order to assess generalization performance we plotted measured ΔP_p values versus those predicted by the identified model on the test data set (Figure 4). The estimation errors on test set were very similar to the errors obtained on training set, thus indicating good generalization performance of the proposed model to predict new data. The error was lower than 1 cmH_2O in 79% of data samples on test set (Figure 4).

In the figure 5 the measured ΔP_p were compared to ΔP_p estimated using Blasius model on test set. The Blasius model presented more accurate estimate of ΔP_p in the range 0-5 cmH_2O then that of the quadratic model; for higher values, greater estimation errors with significant underestimation trend were found.

In Table 2 are summarized the RMSE and R^2 values obtained using quadratic model on training and test set, and using Blasius model on test data set. The values were almost equal between testing and training data set for the quadratic model, while Blasius model showed higher RMSE and lower R^2 values than the quadratic one.

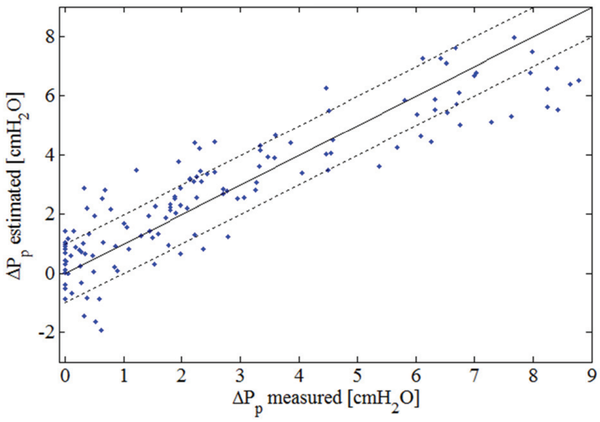

Fig. 3. Measured ΔP_p plotted against ΔP_p predicted with quadratic model (training set). Dashed lines represent ± 1 cmH_2O errors.

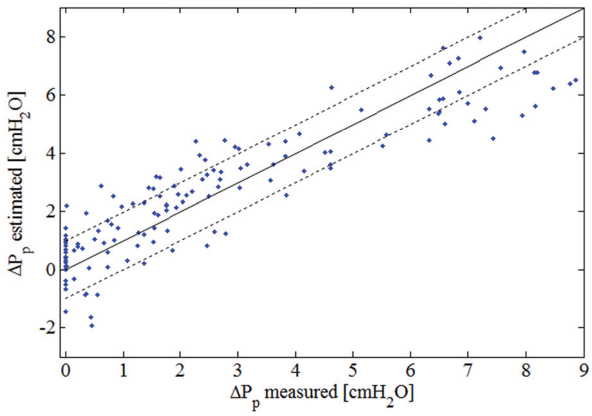

Fig. 4. Measured ΔP_p plotted against ΔP_p predicted with quadratic model (test set). Dashed lines represent ± 1 cmH_2O errors.

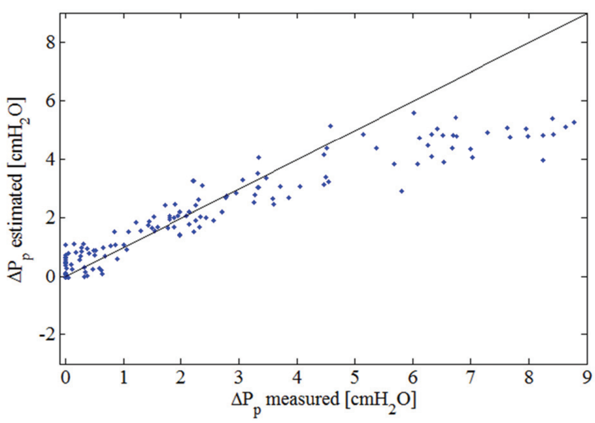

Fig. 5. Measured ΔP_p plotted against ΔP_p predicted using the Blasius model (test set).

4. DISCUSSION

The estimation of endotracheal tube contribution to the total amount of the airway peak pressure is of paramount importance to avoid lung damage during artificial ventilation [9]. Several models has been proposed to estimate pressure drop ΔP(t) over time across the tracheal tube during mechanical ventilation [9-12]. Nevertheless, all of them depend on flow measurement, which currently is not present in high frequency percussive ventilators. Thus, a model that does not require flow measurement is clinically wanted. However, the ventilation and respiratory mechanics parameters (Paw_{peak}, f, R and C) have influence on flow $\dot{V}(t)$ generated by ventilator, which creates the pressure drop [2].

Hence, a regression model was proposed and identified by using the LASSO technique. Such model is able to predict ΔP_p considering only the variables set on the HFPV ventilator and the patient's respiratory system resistance and compliance that can be easily measured at the bed side, before the application of HFPV [15].

The proposed quadratic model presented an estimation error lower than 1 cmH_2O in 79% of data samples, both on training and test set (Figure 3, Figure 4). This value is comparable with the HFPV ventilator resolution of Paw_{peak} measurement. The overall RMSE and R^2 calculated on both data set were very similar (Table 2) underlining a good performance in predicting unseen data. The LASSO technique avoided over-fitting and improved prediction accuracy by forcing parameters to low absolute values (Table 1).

The model presented slightly lower RMSE error than the Blasius model which was already used for ΔP(t) estimation in patients undergoing HFPV [16]. ΔP_p prediction using Blasius model resulted more accurate for lower ΔP_p values while, for higher ΔP_p values, this model presented a progressive underestimation trend (Figure 5). On the other hand, the Blasius model better estimates ΔP(t) over time (Figure 6) and does not accurately estimate the peak value of Paw (and the derived ΔP_p).

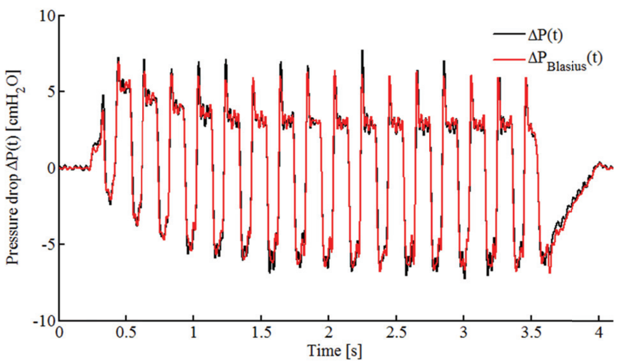

Fig. 6. Comparison between measured and estimated (Blasius model) pressure drop over time.

5. CONCLUSIONS

This study proposes an innovative approach to ΔP_p estimation based only on Paw_{peak}, f, R and C ventilatory parameters. The results showed that the proposed quadratic ΔP_p model identified by using LASSO method may present a valid alternative to the ΔP(t) estimation in case of lack of flow rate measurement during HFPV treatment. The results will be clinically confirmed in a future study.

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