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Dynamic modeling of patient breathing effort in response to change in pressure support in mechanical ventilation

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***Abstract* — A patient-ventilator system is introduced where the input is pressure support level (PSV) from the ventilator and the output is the patient's mechanical work of breathing per minute, or power of breathing (POB). With higher level of pressure support from the ventilator, less effort is needed by the patient to attain the needed the level of ventilation. However, due to several physiological control loops regulating the level of respiration, the dynamic response of the patient due to a change in PSV level can be very complicated. In this study, this dynamic behavior is characterized using a first order system. The dynamic model of the patient-ventilator system has been obtained numerically based on data collected from five patients receiving mechanical ventilation with different levels of pressure support. The results show a wide range of response times among the patients but very limited range in steady state behavior of the system, i.e. DC Gain.**

Keywords—Mechanical Ventilator; Work of breathing; respiratory control; Power of breathing

I. INTRODUCTION

Mechanical ventilators are life-saving machines that help the patients with impaired, or lack of, respiratory drive either temporary or long term. A proper level of respiratory support is very critical to help the lung improve and avoid additional damage to the respiratory system.

Studies show that 24 percent of all patients mechanically ventilated will develop ventilator-associated lung injury VALI for reasons other than ALI or ARDS [1]. Another problem is ventilator dependency due to respiratory atrophy where the patient may depend on the machine, hence causing longer recovery time [2].

Numerous mechanical ventilation control modes from different manufacturers have been developed to address this issue [3]. Some ventilation modes can be complicated for the health care provider to in terms of selection and knob settings. A solution would be to have an adaptive mode that can adjust to the physiological need of the patients. In other words, a ventilation mode that can get feedback from the response of the patient to the ventilator and adjust its gas delivery method according to the need of the patient.

The first step toward this direction is to characterize the dynamic behavior of the patient response to changes in ventilator setting. For example, if we want to regulate the patient's breathing effort by adjusting the level of PSV, we need to answer two basic questions: 1) how much will the patient effort change for a PSV level (i.e. the DC gain of the system)? 2) How long does it take for the patient to adjust to the new level of PSV (i.e. the bandwidth or response time of the system)?

Answering these two questions is not straightforward as there are many factors that are in play. These factors can be summarized into three different categories: Demographic, pathophysiologic and external settings.

Demographics of the patient are the main determinants of the dynamic characteristics of the patient-ventilator system. Factors such has age, gender, weight and race can affect the speed of response of the patient to a new setting in a significant way. This is due to varying levels of strength of the respiratory muscles based on the patient demographic class as well as different sensitivities of the chemoreceptors.

While these dynamic characteristics may change from patient to patient, they may also change within a patient depending on the consciousness level of the patient, the type of the disease and the type and level of the medicine the patient is on. These all fall into the pathophysiological factors.

The third factor affecting the dynamics of the patient-ventilator is the ventilator setting. Any setting on the ventilator affects this dynamics. For example, changing the triggering sensitivity will change the response time of the patient.

In this study, we are interested in investigating the variation in patient-ventilator dynamic model from a pool of five random patients. These patients have different medical conditions, and they all receive PSV ventilation. A first order model has been fitted to the data collected from these subjects and the parameters of the first order model which represents the time constant and gain of the system have been obtained and compared to each other.

II. DEFINITION OF PATIENT-VENTILATOR SYSTEM

With the patients receiving mechanical ventilation, we can define a system of patient plus ventilator which has many inputs and outputs (Fig. 1). All ventilator settings such as PSV (pressure support ventilation) levels, triggering and cycling limits and maximum flow and volume as well as FiO_2 and humidity level are considered as inputs to the system. Patient vitals and ventilation parameters such as tidal volume and respiratory rate are considered as system outputs.

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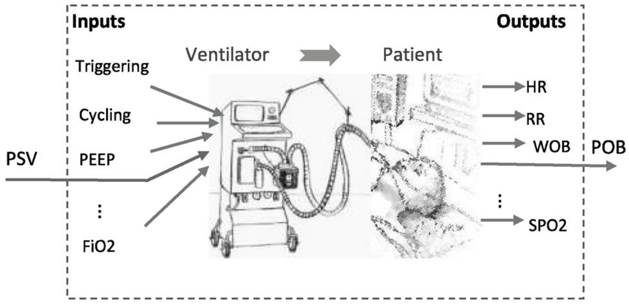

Fig. 1 General patient-ventilator system

Since the focus of this study is the patient's effort in response to changes in the PSV level, the only input and output considered for our model are PSV and POB. POB is defined as the amount of mechanical work done by the patients for breathing in one minute. It is calculated as:

$$POB = WOB * RR, \tag{1}$$

where RR is the respiratory rate and WOB is the mechanical work done by patient's respiratory muscles to complete a breath. It is calculated by integrating muscle effort P_{mus} over tidal volume V [5].

$$WOB = \int P_{mus}\, dV = \int P_{mus}\, Q dt \tag{2}$$

In this equation, Q and V are the inhaled airflow and volume, respectively. Fig. 2 demonstrates a simplified block diagram explaining the relationship between the internal and external respiratory drives. In the absence of mechanical ventilation, the respiratory muscle effort is regulated by several feedback mechanisms. The fastest of these mechanisms is through mechanoreceptors. When the volume in the lung reaches a certain threshold, the mechanoreceptors shut down the respiratory muscles and exhalation begins. The threshold for the lung volume is dependent on the blood gases sensed by the chemoreceptors [6].

When the level of pH in the blood decreases due to exercise or insufficient ventilation, the central chemoreceptor is then activated and it increases the volume threshold. As a result, the tidal volume and minute ventilation increase and the pH returns to normal. Similarly, the peripheral chemoreceptors act upon the levels of oxygen and carbon dioxide in the arterial blood and regulate blood gases by controlling the depth of ventilation.

In the presence of external ventilation, i.e. with a ventilator, these feedback mechanisms are affected. When the level of pressure support increases due to external increase in the flow, the long volume reaches the threshold sooner than normal so the maximum level of muscle pressure decreases in the next immediate breath. However, the resulting tidal volume will be higher than normal because of external pressure supply. Due to the increase in total ventilation, the oxygen in the blood increases gradually and the CO_2 level reduces. As a result, the chemoreceptors gradually reduce the volume threshold until the point where the total minute ventilation gradually returns to its initial value. Therefore, the response of the patient to the change in the ventilator setting is indeed the combination of the three feedback mechanisms. The dynamic behavior of this system is being investigated by modeling it with a linear first order model.

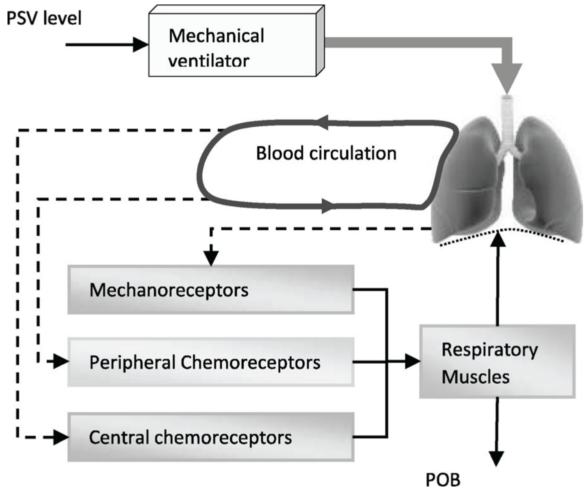

Fig. 2 Mechanism of control of breathing in the presence of mechanical ventilation

III. PATIENT DATA DESCRIPTION

The patient data used in this study are collected from five patients who had the ability to breathe spontaneously. Patient demographics are listed in Table 1. These patients have received invasive ventilation under PSV mode. The patients remained under constant level of pressure support for a period of 10 to 20 minutes before the PSV level changed to a new value. During the entire period, the pressure and flow waveforms where collected from the patient's month. The esophageal pressure was also collected, invasively via an esophageal balloon, synchronously to the flow and pressure measurement, as to use it for calculation of the gold standard muscle pressure P_{mus}. Muscle pressure is calculated from the esophageal pressure P_{esoph} by the following formula:

$$P_{mus} = P_{esoph} - \frac{1}{C_{cw}} V - R_{cw} Q, \tag{3}$$

where C_{cw} and R_{cw} are compliance and resistance of the chest wall respectively. The value of resistance in this case is assumed to be 0.5 and the chest-wall compliance has been calculated by analyzing the esophageal pressure during the exhalation.

Table 1 List of patients used in this study

Pt	*Age*	*Gender*	*Height (in)*	*Weight (kg)*	*Disease, Issue*
1	79	Male	68	70	Ischemic gut
2	70	Female	60	60	Sepsis, COPD
3	55	Male	67	71	Hemopneumothorax
4	54	Male	67	76	Pneumonia
5	77	Male	70	71	Bilateral Pneumonia

Fig. 3 shows the input and output data used for system identification for patient #4 (Table 1). The top graph shows the gold standard POB calculated from Eqs. 1&2. The lower graph shows the corresponding PSV level.

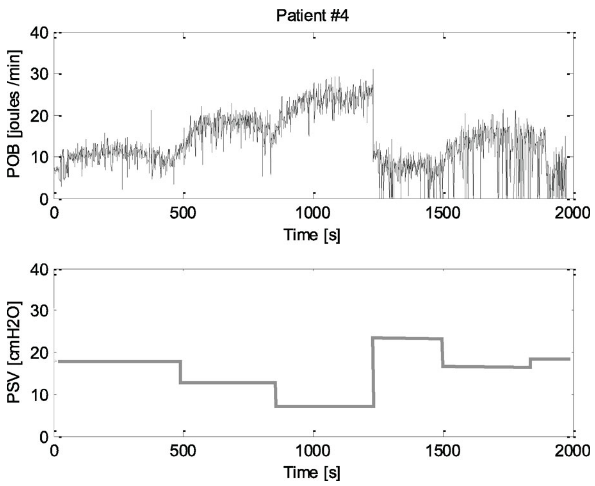

Fig 3 – Output signal, POB (top) and input signal, PSV level (bottom) for the patient-ventilator model

IV. SYSTEM IDENTIFICATION

In order to do system identification, the input and output data were first re-sampled at a fixed sampling time. The system identification was done using a standard transfer model system with one pole. So the transfer function model has the following generic form:

$$G(s) = \frac{K}{\tau s + 1}$$

The parameters of this system were estimated by fitting the model to the input and output data using the least square method and the output of the model were then compared with the real system output. Fig 4 shows the real system output vs the model output for one case.

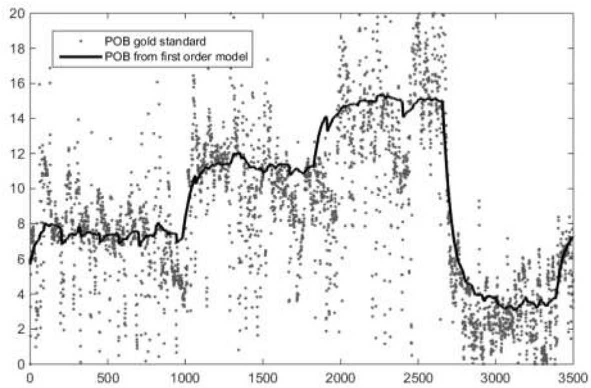

Figure 4 Comparison of the system and model outputs

Obviously, the first order model will not reflect the high frequency variation seen in Fig 4. However, these variations are not important for the purpose of our study. We are only interested in the average of work done over a few minutes. Table 2 shows the summary of results for the identified systems for these five patients. Notice that the system gains are negative since an increase in PSV leads to a decrease in POB.

Table 2 Parameters of the identified systems

Pt	*K (DC gain)*	*τ (Time constant)*
1	-0.90	33.3
2	-0.90	27.7
3	-0.87	491
4	-1.02	50.6
5	-0.81	203.5

V. ANALYSIS OF THE RESULTS

The comparison of the models gives us a good understanding of the dynamic response of the patients to changes in pressure support level. The first and most important observation from the results is the large variation in response time of the patients to ventilator PSV level changes. The time constant of the system 27.7 second for patient #2 to maximum of 491 second in patient #3. It was somehow expected considering the different conditions of the patients but these variations have been quantified for the first time in the literature. The information about the range of time constant for the patient-ventilator model is also very valuable for automatic control of the ventilator to regulate the work load on the patient. Without such information, the bandwidth of the system is unknown, therefore we don't have any reference point for the design of filter for the estimated POB. It will not only helps to design the controller for a closed loop control of the ventilator, but also provide a hint about the desired controller performance. For instance, expecting a controller to change the level of work of breathing on the patient in one or two minutes is not practical since the natural response time of the patient to the change in the PSV can be much slower.

Another interesting observation is that unlike the large variation in the time constants, there is very limited variation in DC gain among the five patients. This mean that regardless of patient's respiratory mechanics, the amount of change in the patient's respiratory workload is more or less the same for the same change in the level of PSV from ventilators. This can also make the design of the closed loop controller for POB regulation much easier since with a single controller, overall gain of the system will not change a lot from patient to patient and as a result, the margin of stability does not change a lot.

The frequency responses of these five models have been compared in Fig. 5 which shows a large variation in the system bandwidth. The bandwidth of the system changes from 0.002 rad/s in slowest patient to 0.02 rad/s second for the patient with the fastest response.

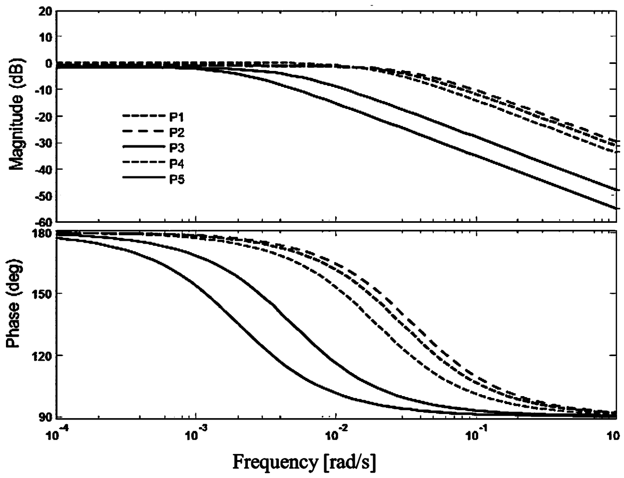

Figure 5 Variations of frequency responses among models P# represents the patient number from Table 1

VI. CONCLUTION

A dynamic system consisting of the patient and the ventilator was considered. A first order single input singe output (SISO) system with PSV (pressure support ventilation) level as input and output of POB (power of breathing) as output was identified for five patients with different diseases and demographics. The results show a large variation in response times of the patients but very limited variation in their DC grains. This information is very useful for the design of a controller to regulate the work load of the patients. At the first place, it give us a good perspective about such system to define a realistic set of expectations for the design performance and criteria of the controller. It can also be used as a reference point for designing such controller using classical control design theories.

VII. REFERENCES

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