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Dynamic Characteristics of a Mechanical Ventilation System With Spontaneous Breathing

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ABSTRACT Mechanical ventilation is an important and effective method for the treatment of pulmonary diseases patients with spontaneous breathing. Spontaneous breathing refers to the physiological breathing activity caused by the respiratory muscle. These patients retain some ability to breathe spontaneously, but do not reach the level of normal breathing. Mathematical simulation and modeling of the mechanical ventilation system are crucial for research on mechanical ventilation. In this paper, a novel pneumatic model of a mechanical ventilation system considering patients' spontaneous breathing is presented. Mathematical equations are accurately derived to explain the principles of the respiratory system and mechanical ventilation system. An experimental prototype is designed to confirm the correctness and validity of the pneumatic model. The goodness of fit shows that the mathematical simulation curve fits well with the experimental curve, thus confirming the accuracy of the pneumatic model. For patients with a certain degree of spontaneous breathing, the mechanical ventilation mode is set to the pressure support ventilation (*PSV*) mode, and variations in the flow, pressure and tidal volume curves are observed by changing specific respiratory mechanics parameters such as the compliance (C), the effective area of the throttle in the pneumatic model (A), and the muscle pressure difference (ΔP_{mus}). From the results, it can be concluded that the resistance of the mechanical ventilation system can be equivalent to A. The dynamic characteristics (mainly flow characteristics, tidal volume characteristics and pressure characteristics) of the mechanical ventilation system are directly influenced by variations in C, A and ΔP_{mus}. This study is an important reference for setting ventilation levels and ventilator control parameters. The results of this research are valuable for the diagnosis and treatment of respiratory diseases.

INDEX TERMS Mechanical ventilation, spontaneous breathing, pneumatic model, lung simulator, dynamic characteristics.

I. INTRODUCTION

Respiratory diseases have become an important factor affecting the health of human beings and quality of life. According to research by the World Health Organization (WHO), respiratory diseases have become some of the most deadly diseases in the world [1], [2]. For example, chronic obstructive pulmonary disease (COPD), a chronic lung disease, has become the 5th most common cause of death and the 10th most burdensome disease [3]–[5]. In the United States, COPD has become the 4th leading cause of morbidity and mortality [3], [6], [7]. In China, respiratory diseases account for 1 million deaths and over 5 million disabilities each year [8]. With the development of medical technology, mechanical ventilation has become an important technical means for the treatment of respiratory diseases [9].

In the treatment of chronic lung diseases, the patient's spontaneous breathing has an important influence on the control of the ventilator [10], [11]. Spontaneous breathing refers to the physiological breathing activity caused by the respiratory muscle. Hence, mathematic simulation is vital to study the patient-specific breathing effort in mechanical ventilation systems. In addition, studies on the airflow dynamic characteristics of a mechanical ventilation system are useful for the treatment of respiratory diseases.

Several active lung models have been proposed in past studies for the simulation of mechanical ventilation considering the patient's spontaneous breathing [10], [12]–[18]. Jodat, Ronald W. et al. conducted a comprehensive analysis of the respiratory system and obtained the mutual mechanical

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relationship among the various parts of the respiratory system [13]. Lutchen et al proposed a viscoelastic model, considering airway resistance, static compliance, viscous resistance and compliance, but the estimation error was large [14]. The resistance-inductance-capacitance (RIC) model was proposed to explain the mathematical relationship among airway resistance, lung inertia, and lung compliance, but lacked a comprehensive description of the respiratory system [15]. The extended RIC model was proposed as an improvement to the RIC model and added peripheral impedance [16], [17].

The model proposed by Jodat et al. considered the respiratory system as a spring-damping system [13]. This model can describe the mechanical properties of the respiratory system well, but cannot accurately describe the physiological characteristics. Most lumped parameter models consider the characteristics of the respiratory system as electrical characteristics to explain the relationship among the airflow resistance, compliance and lung inertia [19]–[24]. However, these models cannot alter the intensity of lung lesions and physiological parameters of the respiratory system, which reduces their applicability in medical research. In addition, too many parameters increase the complexity and reduce the efficiency when analyzing the dynamic characteristics of the respiratory system.

In this paper, a novel method is proposed to study the dynamic characteristics of a mechanical ventilation system considering patients' spontaneous breathing. The mechanical ventilation system is modeled as a pure pneumatic system, which facilitates the research of the influences of parameter changes. Furthermore, the mathematical relationship of parameters in this pneumatic model is proposed, considering the effect of the patient-specific breathing effort. This method reduces the number of parameters in the mechanical ventilation system and increases the efficiency and feasibility of the research. In addition, a prototype of a mechanical ventilation system based on ASL5000 is constructed to verify the pneumatic model. The ASL5000 is a digitally controlled real-time breathing simulator, which consists of a piston moving inside a cylinder. The principle of the lung simulator is based on the equation of motion of an active respiratory mechanics system [25], [26]. The ASL5000 can simulate various types of breaths as spontaneous ventilation. Therefore, it is commonly used in clinical research [22], [27]–[29]. The dynamic characteristics of the mechanical ventilation system considering patients' spontaneous breathing can be obtained by analyzing the results of experiments and mathematical simulations. The novel method and the conclusions are of great significance for respiratory diagnostics, treatment, clinical research, and intelligent control of ventilators.

II. MODELING OF A MECHANICAL VENTILATION SYSTEM CONSIDERING PATIENTS' SPONTANEOUS BREATHING

A. MODELING OF THE RESPIRATORY SYSTEM

Before studying the mechanical ventilation system, the respiratory system, which directly reflects the physiology of the human body, must be modeled accurately. During spontaneous breathing, two lung lobes of the respiratory system may be considered to function together as a single pneumatic viscoelastic compartment. As shown in FIGURE 1, the lungs in the respiratory system are equivalent to a single, elastic compartment. The chest wall is represented as an elastic shell that wraps around the lung compartment. There is a thin intrapleural space between the chest wall and the lung compartment, and its volume is negligible. A flow-resistive airway connects the lung compartment to the outside world, providing an airflow path for spontaneous or mechanical ventilation.

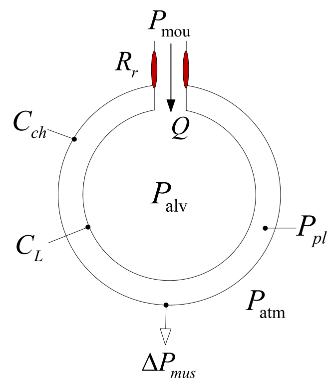


FIGURE 1. A single-compartment pneumatic model of the respiratory system.

In FIGURE 1, P_{mou} is the pressure at the airway opening; P_{pl} is the pressure within the intrapleural space; P_{atm} is the atmospheric pressure; P_{alv} is the alveolar pressure; ΔP_{mus} is the muscle pressure difference; Q is the volumetric flow rate, whose value is positive during inspiration, and negative during expiration; R_{r} is the resistance of the respiratory system; C_{L} is the lung compliance; and C_{ch} is the chest wall compliance.

The working principle of the respiratory system is explained as follows based on the pneumatic model shown in FIGURE 1. Unlike the passive lung breathing pattern, ΔP_{mus} provides the power for spontaneous breathing. When inhaling, the respiratory muscle works, changing the volume of the abdominal cavity and thoracic cavity. This process causes a change in the pleural pressure (P_{pl}). Then, transpulmonary pressure, i.e., the difference between the pleural pressure (P_{pl}) and alveolar pressure (P_{alv}), produces variation in the lung volume. The air enters the lungs through the airway. The volumetric flow rate (Q) is influenced by the resistance of the respiratory system (R_{r}), which is considered equal to the airway resistance. When exhaling, the elastic retractive force of the chest wall reduces the volume of the abdominal cavity and thoracic cavity, and then the pleural pressure (P_{pl}) is varied. The elastic retractive force of the lungs expels the air from the alveoli, changing the lung volume. To summarize the working principle, we can derive the following equations [12], [13], [30].

Pressure drop due to the airway resistance:

$$P_{mou} - P_{alv} = R \cdot Q, \tag{1}$$

Pressure changes caused by the compliance:

$$P_{alv} - P_{pl} = \frac{\Delta V_L}{C_L}, \tag{2}$$

$$(P_{pl} - P_{atm}) + \Delta P_{mus} = \frac{\Delta V_L}{C_{ch}}, \tag{3}$$

In these equations, ΔV_{L} is change in lung volume, and the other terms have been explained above. Taking the airway as the research object, equations (1) and (2) can explain the mechanical principle of the process of airflow entering the lung compartment from the mouth. The airway pressure simultaneously overcomes the elastic retractive force of the lungs and the airway resistance. Taking the chest wall as the research object, equation (3) can explain the mechanical principle of the process in which the muscle pressure difference provides the power for spontaneous breathing. In this process, the respiratory muscles mainly overcome the pressure changes inside the intrapleural space and the elastic retractive force of the chest wall.

By combining equations (1), (2) and (3), the effective mechanical equation for a single-compartment pneumatic model of the respiratory system can be obtained and is expressed as equation (4).

$$(P_{mou} - P_{atm}) + \Delta P_{mus} = \frac{\Delta V_L}{C_{\mathrm{tot}}} + R \cdot Q, \tag{4}$$

In the above equation, C_{tot} is the total compliance of the respiratory system, which is given by equation (5). Considering the compliance as a lumped parameter facilitates the control of the ventilator during mechanical ventilation.

$$C_{\mathrm{tot}} = \frac{C_L \cdot C_{ch}}{C_L + C_{ch}}, \tag{5}$$

For comparison between different parameters, the units are unified in this paper. Time is reported in s (seconds) and ΔV_{L} in mL (milliliter). All the pressure values are reported in $\mathrm{cmH_2O}$, which is often used clinically ($1\ \mathrm{cmH_2O} = 98$ Pa). In plotting and comparisons, all pressure values are actually the difference between a particular pressure and atmospheric pressure. The units of Q are $\mathrm{cmH_2O/L/s}$, and the units of all compliance (C_{L}, C_{ch}, and C_{tot}) are $\mathrm{mL/cmH_2O}$.

B. PNEUMATIC MODEL AND MATHEMATICAL RELATIONSHIP OF MECHANICAL VENTILATION SYSTEM

Based on the working principle of the mechanical ventilation system, a novel pneumatic model is developed, as shown in FIGURE 2. In this system, the ventilator can be regarded as an air supply resource, whose pressure is adjusted by the control valve. A flexible tube connects the human respiratory system to the ventilator.

Compliance (C_{tot}) and resistance (R_{r}) are two important parameters of the mechanical ventilation system. Compliance is a physical term, referring to the change in capacity of a container at unit pressure ($C = \mathrm{d}V/\mathrm{d}P$). During mechanical ventilation, the pressure of the mechanical ventilation system is 2 to 40 $\mathrm{cmH_2O}$, so the compliances of the ventilator tube and respiratory tract can be neglected [31]–[34]. In the pneumatic model, the compliance of the variable-volume container can reflect C_{tot}. During experimental measurements, the value of C_{tot} is adjusted by the initialization settings on the control panel of the active lung simulator ASL5000. The experimental prototype is shown in FIGURE 3 and is described below.

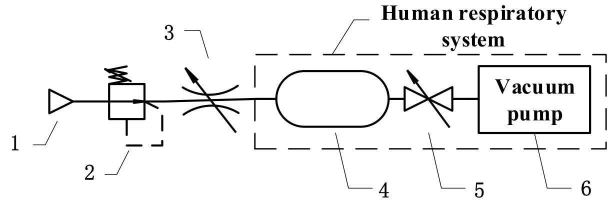
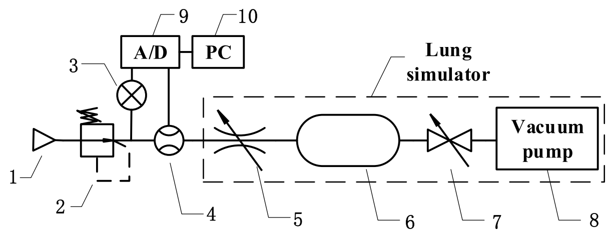
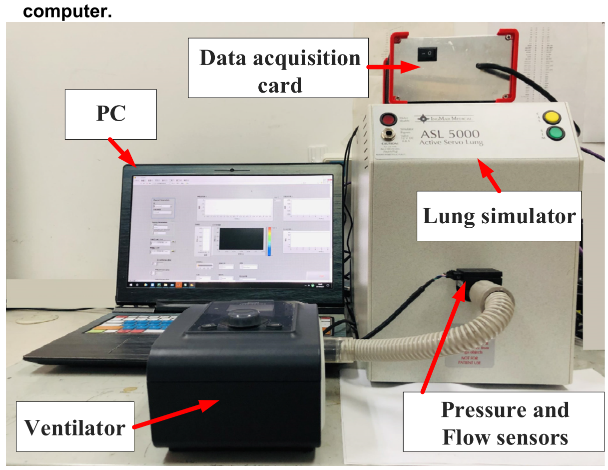


FIGURE 2. The pneumatic model of the mechanical ventilation system considering spontaneous breathing. (1) Air supply resource, (2) Control valve, (3) Throttle, (4) Variable-volume container, (5) Brake valve, (6) Vacuum pump.



(a) Principle diagram of the experimental prototype (1) Air supply resource, (2) Control valve, (3) Pressure sensor, (4) Flow sensor, (5) Throttle, (6) Variable volume container, (7) Brake valve, (8) Vacuum pump, (9) Data acquisition card, (10) Data analysis computer.



(b) Physical image of the experimental prototype

FIGURE 3. Experimental prototype of the mechanical ventilation system considering spontaneous breathing.

In the mechanical ventilation system, the total respiratory resistance is represented by the friction loss of the equivalent throttle, including the effects of the resistance of the tube and human respiratory tract. The lengths of the ventilator tube and the human respiratory tract are fixed, so that the friction loss of the throttle is just affected by its equivalent effective area (A). The values of the effective area during complete spontaneous breathing and mechanical ventilation are different, which represents the difference in respiratory resistance. This is reflected in the analysis of the dynamic characteristics of the airflow.

In the pneumatic model, the human lung is regarded as a variable-volume container. A vacuum pump provides a negative pressure to the container, which simulates the process that the contraction of the respiratory muscles causes a negative pressure in the lungs, generating an inhalation action during spontaneous breathing. The brake valve controls the on and off of the vacuum pump to simulate the conversion between the inhalation and exhalation processes.

The effect of spontaneous breathing on mechanical ventilation is mainly reflected in the pressure changes caused by respiratory muscle contraction, whose waveform can be obtained by the working principle of the active lung simulator. Therefore, the mathematical relationship of the mechanical ventilation system considering human spontaneous breathing is proposed as follows.

1) PRESSURE EQUATION

The temperature and density field of air in the mechanical ventilation system are uniform; hence, the prototype can be assumed to be an isothermal system. According to the state equation of the ideal gas (i.e., equation (6)), both sides of the equation are differentiated, and an expression of the pressure can be proposed as follows:

$$PV = mRT, \tag{6}$$

$$\frac{dP_L}{dt} = \frac{RTq}{V_L} - \frac{mRT}{V_L^2} \cdot \frac{dV_L}{dt}, \tag{7}$$

$$\frac{dP_L}{dt} = \frac{RTqV_L}{V_L^2 + CmRT}, \tag{8}$$

where P_L is the intrapulmonary pressure affected by changes in lung volume (Pa), t is the time (s), V_L is the volume of the lung (m^3), q is the air mass flow (kg/s), m is the mass of air (kg), C is the compliance of the mechanical ventilation system (m^3/Pa), R is the gas constant (equal to 287 J/(kg·K)), and T is the temperature (K).

Here, the total compliance of the mechanical ventilation system is used as the compliance parameter ($C = C_{tot}$) for determining the applicability of the control system.

2) FLOW EQUATION

In this pneumatic model, the effects of the tube and the human respiratory tract are equivalent to the influence of the throttle. Hence, the air mass flow equation through a restriction can be calculated, using the ratio of downstream to upstream pressure (P_d/P_u). As the pressure of the mechanical ventilation system is 2 to 40 cmH_2O, the airflow is subsonic ($P_d/P_u > 0.528$) [35], [36]. Therefore, the volume flow equation of this system can be obtained as follows:

$$Q = \frac{n_f A P_u \sqrt{1-b}}{\rho_a \sqrt{RT}} \sqrt{1 - \left(\frac{\frac{P_d}{P_u} - b}{1-b} \right)^2}, \tag{9}$$

In this equation, n_f is the flow coefficient, which is equal to 1 during the inhalation period and equal to -1 during the exhalation period. Q is the volumetric flow rate (m^3/s), ρ_a is the air density (kg/m^3), A is the effective area of the throttle (mm^2), P_u is the upstream pressure (cmH_2O), P_d is the downstream pressure (cmH_2O), and b is equal to 0.528, which is the critical pressure ratio.

3) VOLUME EQUATION

The compliance of the mechanical ventilation system can be described by equation (10) based on the definition of respiratory compliance.

$$C = C_{tot} = \frac{dV}{dP_L}, \tag{10}$$

$$dV = C \cdot dP_L, \tag{11}$$

4) GOVERNING EQUATION

The values of V and P_L can be calculated by using the discrete state equations as follows:

$$\Delta V = V_t(t) - V_t(t-\tau), \tag{12}$$

$$\Delta P_L = P_L(t) - P_L(t-\tau), \tag{13}$$

where V_t is the real-time volume of the lung (m^3), P_L is regarded as the real-time pressure of the lung (Pa), ΔV is the volume change of the lung caused by the pressure change (m^3), and τ is the sampling time.

Considering human spontaneous breathing, the pressure change produced by the respiratory muscles must be added to the pressure of the lung. Therefore, the effective pressure difference driving the mechanical ventilation system during spontaneous breathing is given by:

$$\Delta p = \Delta P_L + \Delta P_{mus}, \tag{14}$$

The value of ΔP_{mus} is analyzed in the next section.

III. EXPERIMENTAL STUDY AND VERIFICATION OF MATHEMATICAL SIMULATION

A. EXPERIMENTAL APPARATUS

In this study, to demonstrate the novel pneumatic model and the mathematical relationship proposed above, an experimental apparatus is constructed, as shown in FIGURE 3. In this prototype, pressure and flow sensors, a data acquisition card and a data analysis computer have been added to the mechanical ventilation system to collect pressure and flow data. The part in the dashed box can be achieved by using an active lung simulator. The experimental prototype used the active lung simulator ASL5000 to simulate the spontaneous breathing of the human lung, which has been verified to be accurately fitted in clinical medicine.

B. EXPERIMENTS AND MATHEMATICAL SIMULATION OF SPONTANEOUS BREATHING

To simulate the spontaneous breathing of a human being, the lung simulator is separated from the ventilator and works independently. According to the principle of the lung simulator, ΔP_{mus} is defined as a piecewise function (cmH_2O), as shown in equation (15) [22], [20]. This function represents the change in the value of ΔP_{mus} over a breathing cycle. A breathing cycle is set to 4 s, and the curvature of the pressure change produced by respiratory muscles approximately satisfies the sinusoidal curvature. During a respiratory period, the inspiratory time is set to 1 s and the expiratory time is set to 3 s [22], [37]. It is assumed that the pressure change generated by respiratory muscles acts only during inhalation, and the work is mainly performed by the elastic recoil force of the lungs and the chest wall during exhalation.

$$\Delta P_{mus} = \begin{cases} -10 \cdot \sin(\dfrac{\pi t}{1.2}) & 0 \le t < 0.6 \\ -10 & 0.6 \le t < 0.8 \\ -10 \cdot \sin(\dfrac{\pi (t-0.6)}{0.4}) & 0.8 \le t < 1 \\ 0 & 1 \le t < 4, \end{cases} \tag{15}$$

According to the literature, the resistance of the respiratory tract is 1 to 4 cmH_2O/L/s for healthy adults, so it is set to 3 cmH_2O/L/s [38]–[40]. The compliance is approximately 100 mL/cmH_2O for healthy adults [41], [42]. Under the parameter settings above, the flow, volume and pressure curves of the mechanical ventilation system are shown in FIGURE 4.

From the figures, it can be seen that the simulation curves fit the experimental results accurately, verifying the correction of this pneumatic model. The goodness of fit (R^2) is used to evaluate the degree of fit of the simulation curve to the experimental curve. The value of R^2 is calculated by equation (16). In this equation, $\hat{y}_i$ is the value of the simulation data, y_i is the value of the experimental data, $\bar{y}$ is the average value of the experimental data, and n is the number of samples. In FIGURE 4, the R^2 of the flow curve is equal to 0.9358, the R^2 of the volume curve is equal to 0.9641, and the R^2 of the pressure curve is equal to 0.9614.

$$R^2 = \frac{\sum_{i=1}^{n} (\hat{y}_i - \bar{y})^2}{\sum_{i=1}^{n} (y_i - \bar{y})^2}, \tag{16}$$

The pressure curve reflects the effort of spontaneous breathing. The pressure value is the difference from atmospheric pressure. In the case of negative pressure, the intrapulmonary pressure is less than atmospheric pressure, which means that the respiratory muscles are working to generate inhalation. At the end of inspiration, the chest wall retracts, the diaphragm rises, and the exhalation action occurs. At this time, the intrapulmonary pressure is greater than atmospheric pressure, and the value of the pressure curve is positive. The flow curve corresponds to the pressure curve, demonstrating the accuracy of the above analysis. From the volume curve, it can be seen that the tidal volume of spontaneous breathing under these parameter settings is approximately 500 mL, which is the normal level of adult males.

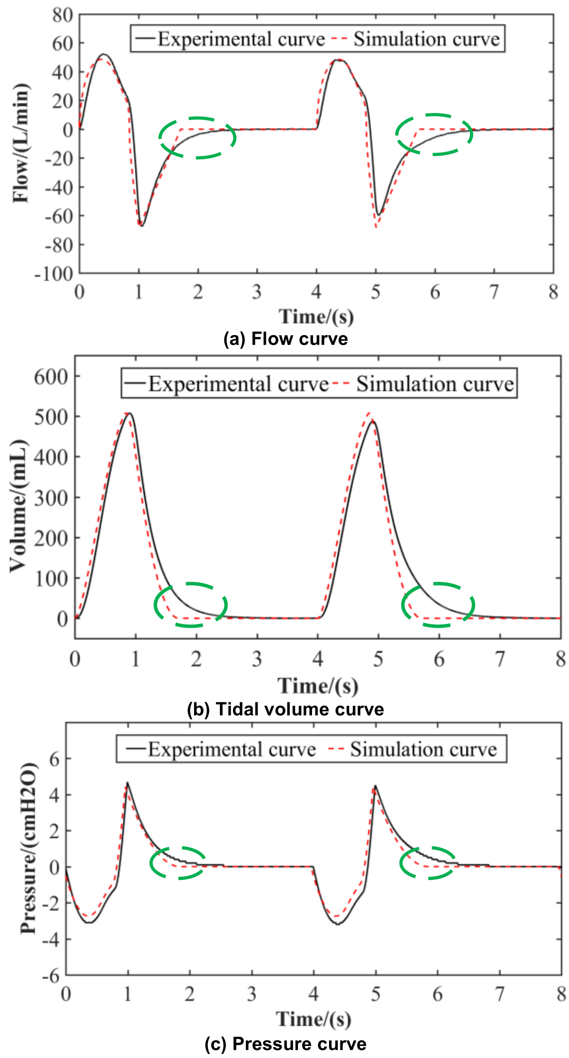


FIGURE 4. Flow, tidal volume and pressure curves of the mechanical ventilation system.

At the turning points in the flow and volume curves, as shown in the green dotted lines in FIGURE 4, the curvature of the experimental curve is smaller than that of the simulation curve, and the experimental values lag behind the simulation values. This is due to the presence of piston resistance in the ASL5000 simulator and some differences from the simulated airflow resistance and compliance variance. The experimental pressure and volume lag behind the simulation pressure and volume. The main reason is that the presence of resistance and compliance blocks the rise of pressure and tidal volume. By taking these extra resistances into account and correcting the pneumatic model, the fit of the curve will improve.

C. EXPERIMENTS AND MATHEMATICAL SIMULATION OF MECHANICAL VENTILATION IN CPAP MODE

In this study, mechanical ventilation for patients with spontaneous breathing is simulated by connecting the ventilator to the lung simulator. For patients with strong spontaneous

TABLE 1. Values of the main parameter settings in *CPAP* mode.

	Value	Unit
CPAP	10	cmH_2O
A	25	mm^2
C	70	mL/cmH_2O
R	5	$cmH_2O/L/s$
$\Delta P_{mus(max)}$	10	cmH_2O

breathing ability, such as patients with obstructive sleep apnea hypopnea syndrome (*OSAH*), the adopted mode of ventilation is continuous positive airway pressure (*CPAP*). The values of the main parameter settings are shown in TABLE 1.

The results are shown in FIGURE 5.

The auxiliary effect of the ventilator on patients in this mode can be well reflected from the curves. The continuous positive pressure from the ventilator makes the patient's breathing easier and smoother. The effective area of the throttle (A) is set to 25 mm^2, corresponding to the respiratory resistance of 5 $cmH_2O/L/s$. The compliance is set to 70 cmH_2O. The maximum ΔP_{mus} is 10 cmH_2O, which is the same as the value of healthy people. This means that the patient's spontaneous breathing ability remains good; however, the resistance of the patient's respiratory system is higher than normal, and the elastic retraction force of the lungs is reduced.

As shown in FIGURE 5, using mechanical ventilation in *CPAP* mode, the patient's tidal volume can reach normal levels (approximately 500 mL). The patient's peak value of volume flow doesn't increase significantly compared to the normal level and is approximately 57 $cmH_2O/L/s$. Due to the continuous positive pressure from the ventilator, the value of the lung pressure is always positive. The pressure value varies on the basis of 10 cmH_2O. The peak-to-peak value of pressure (14.3 cmH_2O) is larger than that of healthy people without mechanical ventilation. The peak inspiratory pressure is approximately -5 cmH_2O, which is larger than the simulation value of the healthy pressure (approximately -3.1 cmH_2O).

Through calculation, the R^2 of the flow curve is 0.9398, the R^2 of the tidal volume curve is 0.9983, and the R^2 of the pressure curve is 0.9418. The goodness of fit is sufficient to verify the correction of the parameter settings and the pneumatic model. The differences at the turning points of the curves are similar to those of the spontaneous breathing curves. The experimental environment is the same. Therefore, the reasons for the differences between the experimental curve and the simulation curve are similar to the reasons for the differences in the spontaneous breathing curves, which have been explained in the previous chapter. These differences do not affect our analysis of this pneumatic model or the clinical application of this model in mechanical ventilation.

D. EXPERIMENTS AND MATHEMATICAL SIMULATION OF MECHANICAL VENTILATION IN PSV MODE

In addition, pressure support is used to facilitate spontaneous breathing during mechanical ventilation. For patients with varying degrees of spontaneous breathing, pressure support ventilation (*PSV*) is a commonly used mode. Therefore, the experiments and mathematical simulation of mechanical ventilation in *PSV* mode are highlighted below. Taking the mechanical ventilation treatment of patients with *COPD* as an example, the respiratory mechanics parameters are set to specific constant values, and the *PSV* ventilation mode is adopted. The values of the main parameter settings are shown in TABLE 1.

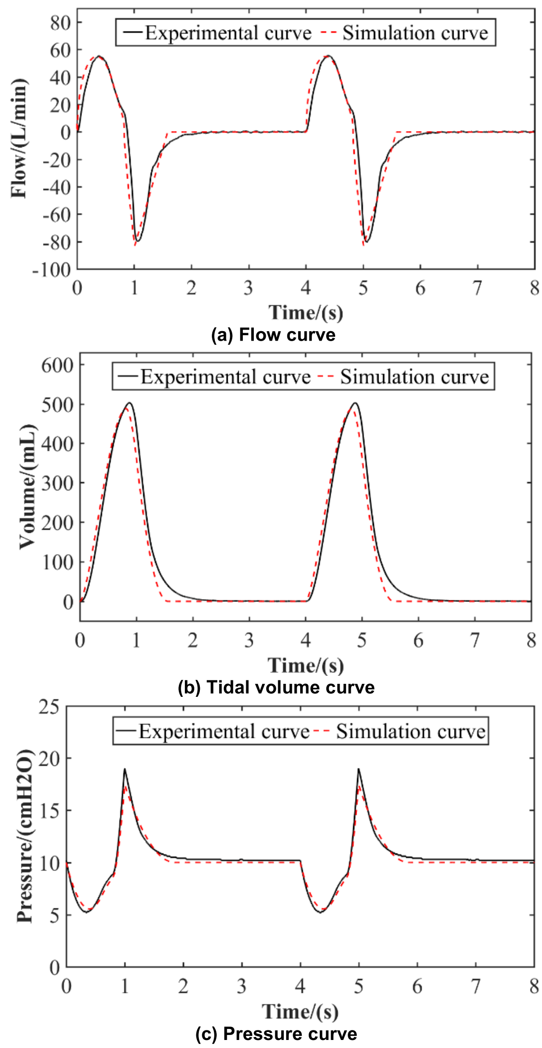


FIGURE 5. Flow, tidal volume and pressure curves of the mechanical ventilation system in *CPAP* mode.

TABLE 2. Values of the main parameter settings in *PSV* mode.

	Value	Unit
PSV	17	cmH_2O
PEEP	5	cmH_2O
A	15	mm^2
C	50	mL/cmH_2O
R	15	$cmH_2O/L/s$
$\Delta P_{mus(max)}$	5	cmH_2O

The function type and curve shape of the muscle pressure difference (ΔP_{mus}) are the same as in the mathematical simulation of healthy people. However, the maximum ΔP_{mus} is adjusted to 5 cmH_2O here, indicating a significant reduction in the patient's ability for spontaneous breathing. The function is expressed as equation (17).

$$\Delta P_{mus} = \begin{cases} -5 \cdot \sin(\frac{\pi t}{1.2}) & 0 \le t < 0.6 \\ -5 & 0.6 \le t < 0.8 \\ -5 \cdot \sin(\frac{\pi (t-0.6)}{0.4}) & 0.8 \le t < 1 \\ 0 & 1 \le t < 4, \end{cases} \tag{17}$$

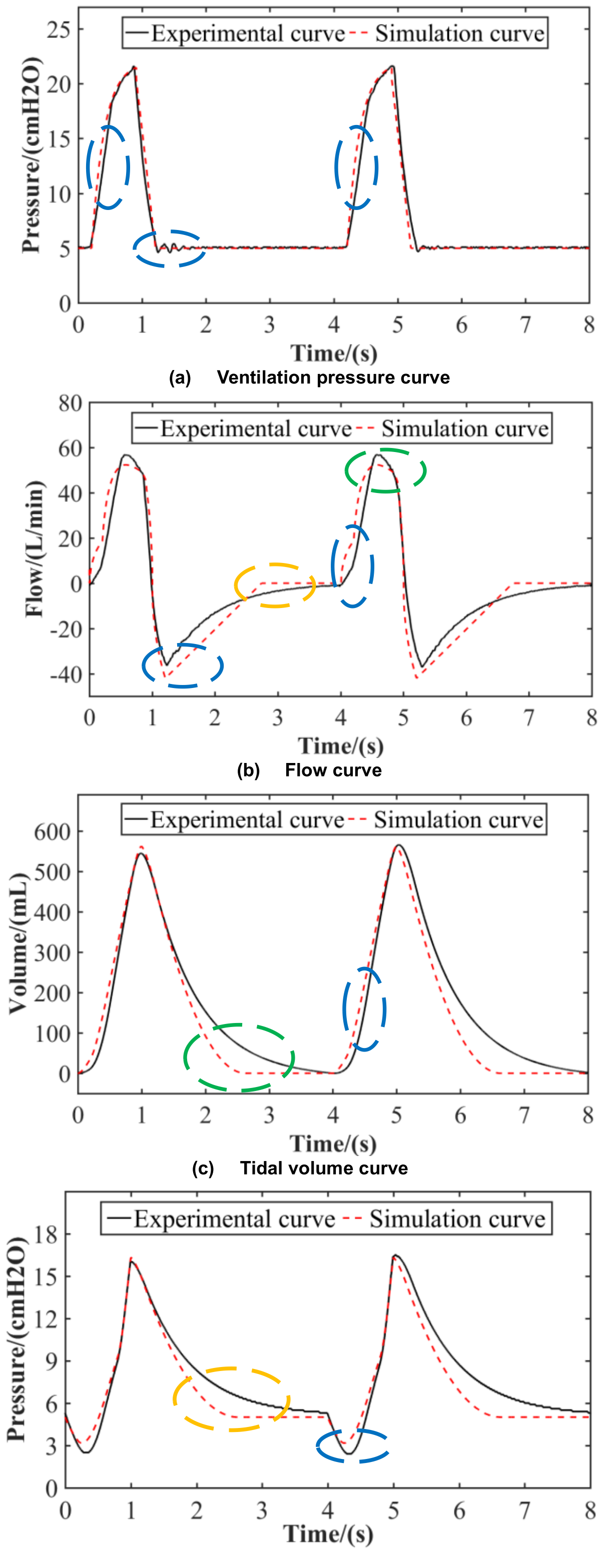
In *PSV* mode, the operating pressure of the ventilator works as a piecewise function, as shown in equation (18). This equation is obtained by interpolation fitting the experimental curve based on the working principle of the active lung simulator ASL5000. It has been modified according to the physiological principle. In the equation, *PEEP* represents the positive end expiratory pressure, and the value is 5 cmH_2O. The nominal *PSV* value is set to 17 cmH_2O, which represents the pressure support level of the ventilator. When an inspiratory flow is detected, the ventilator triggers and the trigger time is expressed as t_{o}. The rise time of the inhalation pressure is expressed as t_{rise}. The ventilation pressure at the end of the rise phase is P_{peak}. Then, the pressure drops, and the drop time is expressed as t_{drop}. When exhaling, the ventilator pressure remains at *PEEP*. In this experimental example, t_{o} is 0.2, t_{rise} is 0.7, and t_{drop} is 0.3.

$$P_m = \begin{cases} PEEP & 0 \le t < t_o \\ PEEP + PSV \\ \cdot \left(1 - e^{\left(\frac{-(t-t_o)}{t_{rise}}\right)}\right) & t_o \le t < (t_o + t_{rise}) \\ P_{peak} - (P_{peak} - PEEP) \\ \cdot \dfrac{(t - t_o - t_{rise})}{t_{drop}} & (t_o + t_{rise}) \le t \\ & \quad < (t_o + t_{rise} + t_{drop}) \\ PEEP & t \ge \left(t_o + t_{rise} + t_{drop}\right) \end{cases} \tag{18}$$

The results are shown in FIGURE 6.

The values of R^2 are shown in TABLE 3. The goodness of fit is sufficient to verify the correction of the parameter settings and the mathematical equations. Hence, the following research can be carried out under these settings.

FIGURE 6 shows that the experimental values lag behind the simulated values, and the experimental absolute values of the expiratory flow and the inspiratory pressure are less than the simulated values. These phenomena are circled by blue dotted lines in the figure. The reasons can be attributed to the resistance of the piston in the active lung simulator, the resistance of the channel in the flow sensor and other additional dissipation. For the same reasons, the pressure value exhibits a short-term oscillation at the corner in FIGURE 6 (a).



(a) Ventilation pressure curve

(b) Flow curve

(c) Tidal volume curve

(d) Lung pressure curve

FIGURE 6. **Ventilation pressure, flow, tidal volume and pressure curves of the mechanical ventilation system in *PSV* mode.**

The green dotted circles in FIGURE 6 (b) (c) mean that the pneumatic model is not completely constructed in accordance with the principle of the active lung simulator. This model

TABLE 3. Values of the goodness of fit.

	Flow curve	Tidal volume curve	Lung pressure curve	Ventilation pressure curve
Healthy	0.9358	0.9641	0.9614	\
CPAP mode	0.9398	0.9983	0.9418	\
PSV mode	0.9240	0.9941	0.9766	0.9203

has been modified according to the physiological principle. Collecting patients' clinical data to verify this pneumatic model will be scheduled for further work in the future.

The orange dotted circles in FIGURE 6 (b) (d) mean that the pneumatic model of the exhalation phase needs further correction. In this paper, it is considered that breathing muscles do not work when exhaling and the expiratory flow is produced by the elastic retractive force of the lungs and chest wall. However, in fact, in the expiratory phase, the breathing muscles provide some force to help exhaust air from the lungs. Moreover, in the case of mechanical ventilation, the presence of *PEEP* also affects the patient's spontaneous breathing. In the next work, the pneumatic model of the expiratory phase can be modified to an exponential function.

IV. DYNAMIC CHARACTERISTICS OF A MECHANICAL VENTILATION SYSTEM WITH SPONTANEOUS BREATHING

Dynamic characteristics are important reflections of the mechanical ventilation system under changing parameters and are crucial for the control of the system and research on the system parameters. According to clinical research, resistance and compliance can directly affect the flow, volume and pressure characteristics of the mechanical ventilation system. Therefore, the effective area of the throttle (A), which produce changes of resistance, and the compliance (C) are discussed. This approach can provide theoretical support for the clinical operations of the ventilator. In this paper, we focus on the dynamic characteristics of the mechanical ventilation system in *PSV* mode, which is commonly used in mechanically ventilated patients with spontaneous breathing. This chapter is divided into three parts according to analysis of the flow, tidal volume and pressure characteristics.

A. ANALYSIS OF FLOW CHARACTERISTICS

As shown in FIGURE 7, flow curves can reflect the influence of respiratory parameters on the flow characteristics. The peak flow during inspiration can reflect the intensity of breathing under mechanical ventilation. The value of A represents the change in resistance of the mechanical ventilation system. The magnitude of A is inversely proportional to the magnitude of the resistance. The value of C reflects the pulmonary elasticity. As the value of C increases, the degree of the pulmonary elasticity improves. From FIGURE 7, it can be seen that:

(1) As the value of A decreases, the value and curvature of the flow curve decrease, and the change is slower. This is because a decrease in the value of A represents an increase in the resistance of the mechanical ventilation system, hinders the change in the ventilation flow rate, and lowers its value.

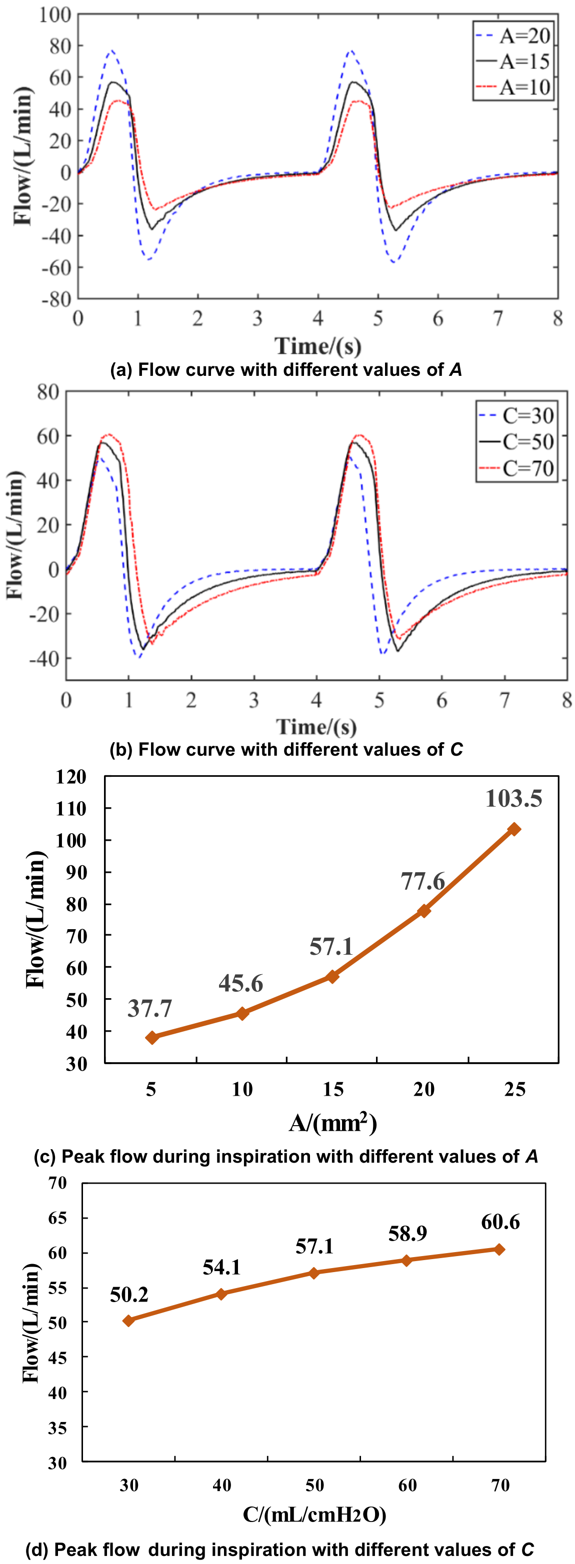


FIGURE 7. Influence of *A* and *C* on the flow characteristics.

(2) As the value of C increases, the value of the flow during inspiration increases. This is because an increase in compliance indicates that the pulmonary elasticity improves. The unit pressure can cause a larger volume change. In addition, an increase in elasticity provides a buffer for changes in lung volume, resulting in a lag in the flow curve. A decrease in the peak flow during exhalation can be observed in FIGURE 7 (b). This is because the increase in the C value leads to a decrease in the intrapulmonary pressure (the specific reason will be described later), and exhalation mainly relies on the intrapulmonary pressure to provide power; hence, the value of the flow decreases.

(3) The value of the peak flow during inspiration is positively correlated with the values of A and C, and the change in the curve is obvious. From FIGURE 7 (c) (d), it can be clearly seen that a change in the value of A can cause a larger change in the value of the peak flow compared with the influence of C. This is because the ventilation flow is mainly affected by the resistance of the mechanical ventilation system. The mathematical relationship can be derived from equation (9).

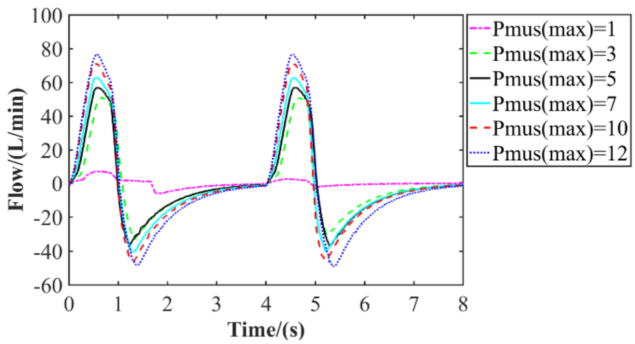


FIGURE 8. Influence of ΔP_{mus} on the flow characteristics.

FIGURE 8 illustrates the effect of the value of ΔP_{mus} on the airflow of the mechanical ventilation system. The value of ΔP_{mus} represents the patient's spontaneous breathing intensity. As the value of ΔP_{mus} increases, the patient's spontaneous breathing intensity becomes stronger, and the value of the flow also increases. However, when $\Delta P_{\text{mus(max)}} = 1$, the airflow of the mechanical ventilation system is very weak, and false triggering can be found in the curve. At this point, the level of PSV has been seriously insufficient. Therefore, different PSV levels should be used for patients with varying degrees of spontaneous breathing. In clinical treatment, the strength of the patient's respiratory muscle can be improved by adjusting the level of PSV.

B. ANALYSIS OF TIDAL VOLUME CHARACTERISTICS

As shown in FIGURE 9, tidal volume curves can reflect the influence of respiratory parameters A and C on the tidal volume characteristics. The peak tidal volume can reflect the intensity of breathing under mechanical ventilation and the level of treatment. By analyzing the curves, it can be found that:

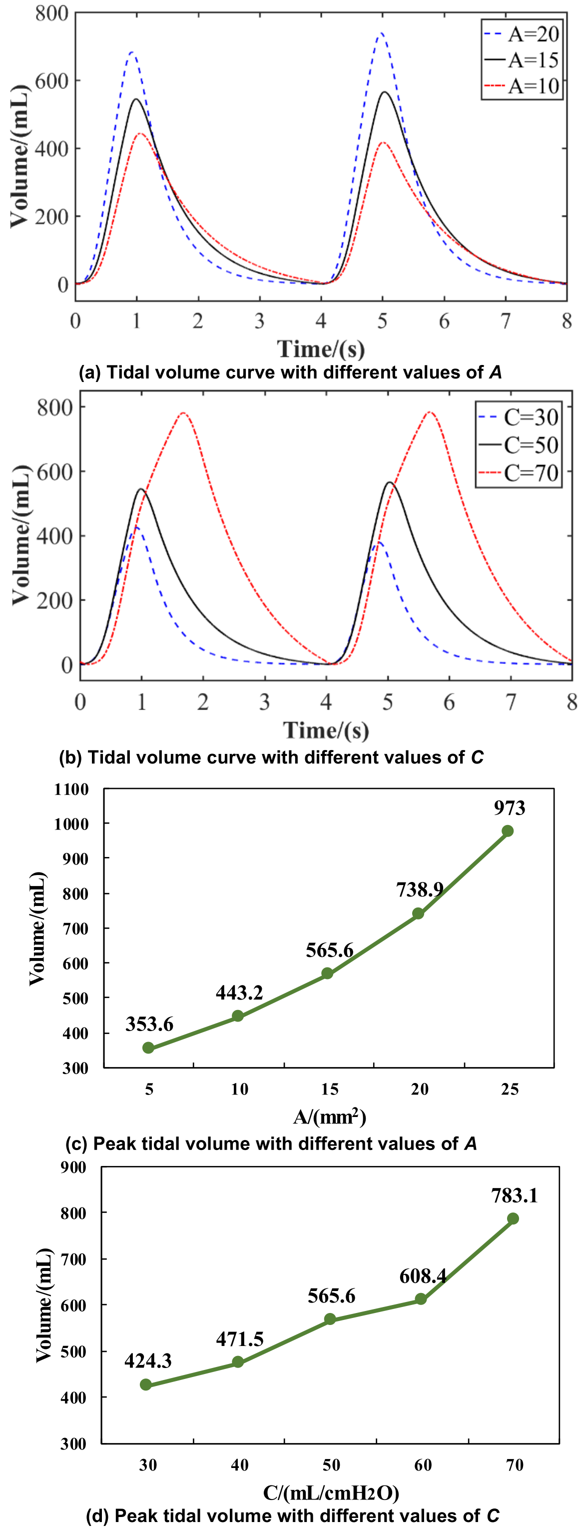


FIGURE 9. Influence of A and C on the tidal volume characteristics.

(1) As the value of A decreases, the value and curvature of the tidal volume curve decrease, and the change is slower. Similar to the flow curve, the reason for this phenomenon is that the increase in resistance hinders the operation of the mechanical ventilation system.

(2) As the value of C increases, the value of the tidal volume increases. Referring to equation (10), it is known that the tidal volume is positively proportional to C. As the value of C increases, the pulmonary compliance increases, and the per unit pressure produces more tidal volume. At the same time, the value of the airflow decreases. Therefore, as the value of C increases, the change in tidal volume lags behind and the change is more moderate.

(3) From FIGURE 9 (c) (d), it can be seen that as the values of A and C increase, the peak tidal volume increases significantly. When $A = 25$ mm^2, the peak tidal volume reaches 973 mL, which means that the mechanical ventilation level is too high. When $A = 5$ mm^2, the peak tidal volume is only 353.6 mL, which means that the level of mechanical ventilation is seriously insufficient. Within the effective working range, a change in the value of C has a greater impact on tidal volume. This is because compliance directly affects the degree of lung expansion, while resistance affects tidal volume by affecting airflow.

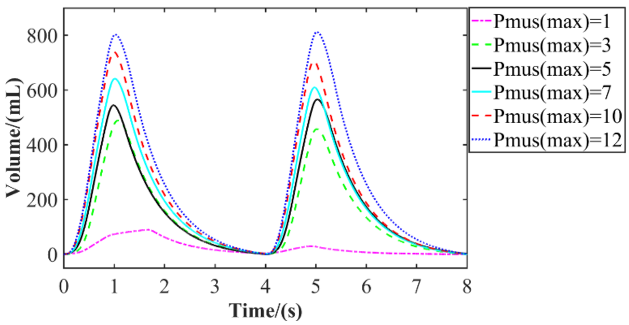


FIGURE 10. Influence of ΔP_{mus} on the tidal volume characteristics.

FIGURE 10 illustrates the effect of the value of ΔP_{mus} on the tidal volume of the mechanical ventilation system. Similar to the flow curve, the value of the tidal volume increases significantly as the value of ΔP_{mus} increases. When $\Delta P_{\text{mus(max)}} = 1$ cmH_2O, mechanical ventilation is ineffective because the *PSV* level is set too low. When ΔP_{mus} changes from 3 cmH_2O to 12 cmH_2O, the peak value of the tidal volume increases by approximately 60% (300 mL).

C. ANALYSIS OF PRESSURE CHARACTERISTICS

As shown in FIGURE 11, pressure curves can reflect the influence of respiratory parameters A and C on the pressure characteristics. The peak pressure during exhalation can reflect the intensity of breathing under mechanical ventilation. By analyzing the curves, it can be found that:

(1) As the value of A decreases, the value of the pressure curve decreases, and the delay of the pressure curve increases. Similar to the flow curve, the reason for this phenomenon is that the increase in resistance hinders the operation of the mechanical ventilation system.

(2) As the value of C increases, the value of the pressure curve decreases. Referring to equation (10), it is known that the pressure change is inversely proportional to C. As the value of C increases, the elasticity of the lungs improves. When breathing, the change in pressure caused by volume changes is smaller. Therefore, the larger the compliance is, the smaller the value of the pressure curve.

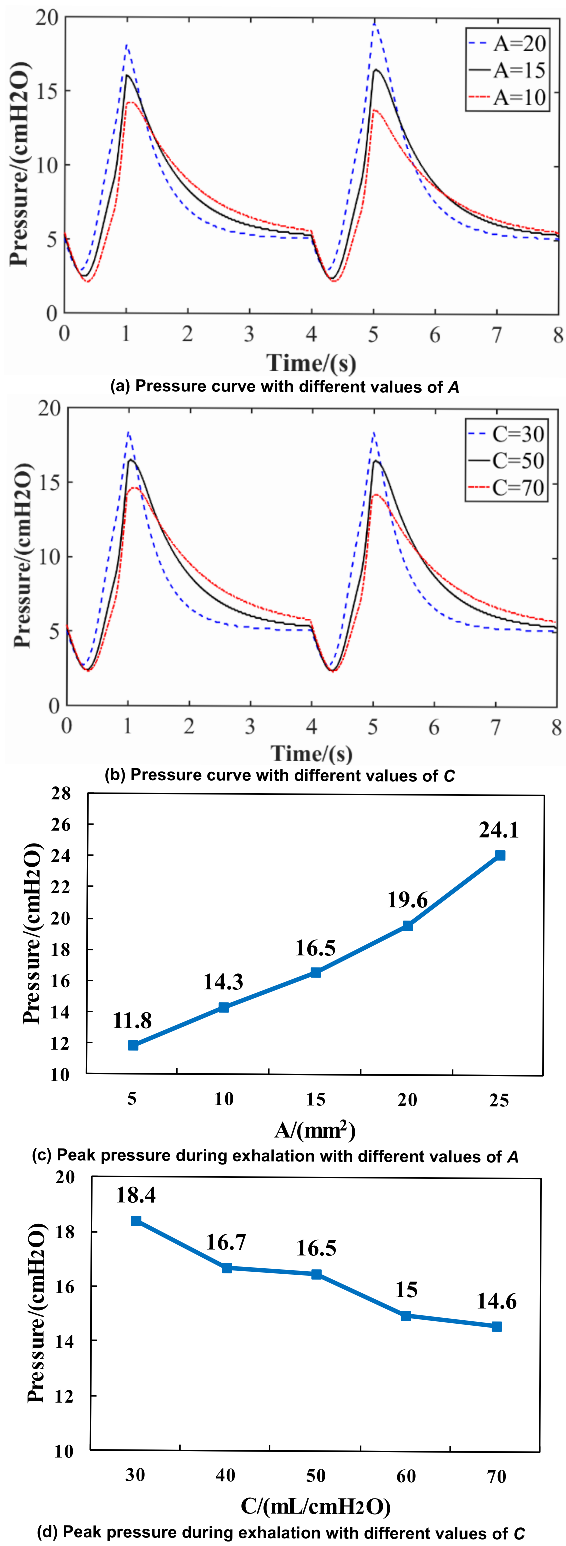


FIGURE 11. Influence of A and C on the pressure characteristics.

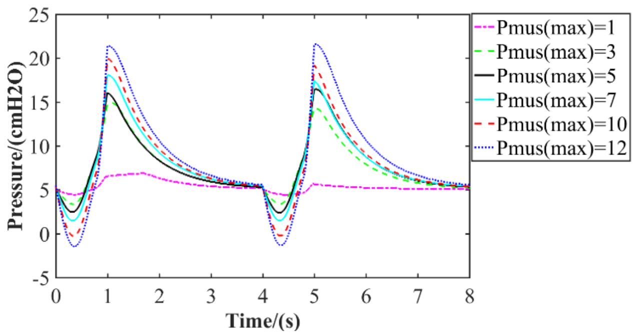


FIGURE 12. Influence of ΔP_{mus} on the pressure characteristics.

(3) From FIGURE 9 (c) (d), it can be seen that as the value of A increases, the peak pressure during exhalation increases significantly, and as the value of C increases, the peak pressure during exhalation decreases. Therefore, patients with poor lung compliance should not rely solely on the pressure curve as a reference for treatment. At this point, clinical monitoring of the tidal volume is very important.

FIGURE 12 illustrates the effect of ΔP_{mus} on the pressure characteristics of the mechanical ventilation system. The pressure increases significantly as the value of ΔP_{mus} increases. This confirms that the degree of effort in breathing muscle is positively correlated with the intensity of breathing. When $\Delta P_{mus(max)} = 1$ cmH_2O, mechanical ventilation is ineffective because the *PSV* level is set too low. When ΔP_{mus} changes from 3 cmH_2O to 12 cmH_2O, the peak value of the pressure increases by approximately 52% (7.5 cmH_2O).

V. CONCLUSION

In this research, mathematical equations that explain the principles of the respiratory system and mechanical ventilation system are accurately derived. A novel pneumatic model is proposed to simulate the mechanical ventilation system with spontaneous breathing. An experimental device is designed to verify the correctness of this model and provide data for analyzing the dynamic characteristics of the mechanical ventilation system. In addition, mathematical simulations and experiments under three different physiological conditions were designed using the experimental prototype, namely, the healthy human state, mechanical ventilation state in *CPAP* mode, and mechanical ventilation state in *PSV* mode. Finally, through the results of experiments and mathematical simulations, the influences of respiratory parameters (A, C and ΔP_{mus}) on the dynamic characteristics of the mechanical ventilation system (flow, tidal volume and pressure characteristics) are obtained. Analysis of the dynamic characteristics is performed for mechanical ventilation using the *PSV* mode, which is commonly used for mechanically ventilated patients with spontaneous breathing. The analysis results are summarized in the following three points.

(1) The mathematical simulation results are consistent with the experimental results, which verifies the novel pneumatic model. The goodness of fit is shown in TABLE 3. The design of the experiments is proven to be effective and reliable.

(2) The negative pressure in the initial phase of inspiration reflects the presence of spontaneous breathing. The variation characteristics of the flow curve, tidal volume curve and pressure curve correspond with each other. For patients with pulmonary diseases, the respiratory parameters are abnormal, and mechanical ventilation can help patients breathe better.

(3) The resistance (R_r), the compliance (C), and the muscle pressure difference (ΔP_{mus}) are three respiratory parameters that affect the dynamic characteristics of the mechanical ventilation system. In the pneumatic model proposed in this paper, the effective area of the throttle (A) replaces the resistance as the respiratory parameter. The dynamic characteristics of the mechanical ventilation system include the flow characteristics, tidal volume characteristics and pressure characteristics. The variation in the respiratory parameters has a significant effect on the dynamic characteristics. The specific impact results have been explained clearly in this paper.

This research provides a novel method to study the dynamics of the mechanical ventilation system with spontaneous breathing. It is helpful absolutely for respiratory diagnostics and the treatment of respiratory diseases, such as *OSAHS* and *COPD*. The pneumatic model of a mechanical ventilation system with spontaneous breathing established in this paper can simulate the treatment plans for patients with respiratory diseases, and then calculate and optimize the pressure support level. This pneumatic model can be used in the design and performance optimization of active lung simulators, ventilators, and some other medical devices. However, asynchrony is a common problem in mechanical ventilation during spontaneous breathing activity. In the next step, we will focus on the identification of asynchrony and novel control solutions for ventilators to solve this problem. In further work, more complex physiological models will be constructed to research the regional ventilation distribution and concomitant effects on ventilator-induced lung injuries.

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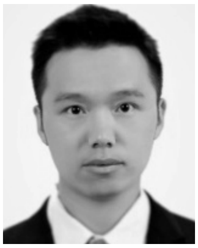

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