CHAPTER 10

Mechanics of proportional-assist ventilation

A.C. Lua & K.C. Shi
School of Mechanical and Aerospace Engineering, Nanyang Technological University, 50, Nanyang Avenue, Singapore 639798.

Abstract

Proportional-assist ventilation (PAV) is an effective and improved mode of ventilation over the conventional pressure-support ventilation due to its attractive features of better patient harmony and comfort. The conventional ventilation systems in hospitals will either deliver a specified tidal volume to the patient or raise the airway pressure to a preset level during the inspiratory phase of each breath. These systems do not take into consideration the patient’s breathing patterns and varying requirements during each inspiration. To assist patients with breathing difficulties, PAV has been used as a mode of synchronized partial ventilatory support, by applying airway pressure in proportion to volume and flow rate to overcome the patient’s higher respiratory elastance and resistance, respectively, as a result of pulmonary disease. A new PAV system using a proportional solenoid valve to control the air supply to patients suffering from respiratory disabilities, was studied. The outlet flow and pressure from the valve at various air-supply pressures were tested and proven to be suitable for pressure and flow control in a PAV system. In vitro tests using a breathing simulator, which has been proven to possess the general characteristics of the human respiratory system in spontaneous breathing tests, were conducted and the results demonstrated the viability of this PAV system in normalizing the breathing patterns of patients with abnormally high resistances and elastances as well as neuromuscular weaknesses. In vivo tests using this PAV system were also conducted on healthy volunteer subjects with artificial resistance and elastance incorporated. The test subjects felt comfortable to breathe, with reduced breathing effort when PAV was active, demonstrating the positive assist effect of PAV.
Human Respiration

1 Introduction

Respiration consists of two processes – external respiration that involves the intake of oxygen and removal of carbon dioxide from the body as a whole, and internal respiration that is the utilization of oxygen and the production of carbon dioxide by cells and the gaseous exchanges between the cells and their fluid medium. The respiratory system is made up of a gas-exchanging organ (the lung) and a pump that ventilates the lungs. The pump consists of the chest wall and the respiratory muscles that increase and decrease the size of the thoracic cavity. At rest, a normal human breathes 12–15 times a minute. Five hundred milliliters of air per breath or 6–8 l/min are inspired and expired.

Inspiration is an active process. The contraction of the inspiratory muscles increases the intrathoracic volume. During quiet breathing, the intrapleural pressure at the base of the lungs is about –2.5 mmHg gauge pressure at the start of inspiration and it finally decreases to about –6 mmHg. The lungs are pulled into a more expanded position. The pressure in the airway and therefore the intrapulmonary pressure become slightly negative and air flows into the lungs (Fig. 1). At the end of inspiration, the lung recoil begins to pull the chest back to the expiratory position where the recoil pressure of the lungs and chest wall balance. The pressure in the airway becomes slightly positive and air flows out of the lungs. Expiration during

![Figure 1: Changes in intrapleural and intrapulmonary pressures relative to atmospheric pressure during respiration [1].](image-url)
quiet breathing is passive in the sense that the muscles that decrease intrathoracic volume do not contract.

Mechanical ventilators are used to artificially ventilate the lungs of patients who are unable to breathe normally. Mechanical ventilation is generally classified into positive-pressure, negative-pressure and high-frequency ventilation. Positive pressure means that the air that is at above atmospheric pressure, is applied to the respiratory tract with the result of pushing the air into the patient’s lungs in order to inflate them. Negative-pressure ventilation is performed with the use of a body ventilator that surrounds a patient with adjustable negative-pressure and creates a vacuum around the chest wall. The negative pressure moves the thoracic walls outward expanding the intrathoracic volume and decreasing the pressure inside the lungs. The pressure difference between the atmosphere and the lungs allows the air to flow into the lungs. High-frequency ventilation delivers the air at a much higher frequency of 50–150 cycles per minute and at a low air volume, providing adequate gas exchange while minimizing the degree that the lungs are actually inflated. Positive-pressure ventilation is the most widely used and this principle is used in the assist-mode ventilation to be described below.

In the assist-mode ventilation, the delivery of ventilatory support can be coordinated by the patient’s effort. This mode allows the patient to participate in the breathing process and the initialization of the machine-cycled breath is triggered by the patient’s own effort. This is in contrast to the other mode, i.e., the controlled-mode ventilation that is totally predetermined by the ventilator settings and therefore the patient’s spontaneous breathing is totally depressed. Currently, research in ventilation is mainly directed towards the assist mode because it can (i) synchronize patients’ effort and ventilatory response, (ii) reduce the need for sedation, (iii) prevent disused atrophy of the respiratory muscles, (iv) improve haemodynamic tolerance of ventilatory support, and (v) facilitate the weaning process.

The assist mode can be subdivided into five modes, namely, assist-control ventilation, synchronized intermittent mandatory ventilation (SIMV), pressure-support ventilation (PSV), continuous positive airway pressure (CPAP) and positive end expiratory pressure (PEEP). Assist-control ventilation provides preset volume or pressure support every time the patient initiates a spontaneous breath. In addition, the ventilator is preset to provide a minimum number of breaths if the patient fails to trigger inspiration. SIMV coincides the mandatory machine breaths to occur with the patient’s inspiratory efforts, but not every breath is supported. SIMV is usually used as a weaning mode because it requires the patient to take on some of the work of breathing. The intermittent frequency can be gradually reduced according to the patient’s recovery condition. PSV is a mode of ventilation that supplements the patient’s spontaneous inspiration with a preset amount of positive airway pressure. During PSV, the patient determines the rate and duration of the breath. It is useful in compensating for high airway resistance and it is generally regarded as comfortable for spontaneous inspiration for patients. CPAP provides a constant flow of air (positive airway pressure) during both inspiratory and expiratory phases. The ventilation supply is mostly controlled by a solenoid valve or a proportional solenoid valve that can be triggered by either pressure or flow signal. PEEP functions to maintain
positive pressure at the end of expiration, serving to keep the lungs and the alveoli inflated at the end of expiration and reducing the potential for alveolar collapse, thereby improving oxygenation levels. Therefore, PEEP is useful for patients who have difficulties in maintaining adequate oxygenation and this will eliminate the need for high levels of applied oxygen.

However, none of the above modes normalizes the continuous relationship between the patient’s effort and ventilatory consequences. The patient’s spontaneous effort triggers the ventilator and it then runs its own course according to its predetermined settings. All these systems do not take into consideration the patient’s breathing patterns and varying requirements during each inspiration. Therefore, PAV is proposed as a mode of ventilatory support that will normalize the patient–ventilator interaction throughout the respiration process. PAV provides synchronized partial ventilatory support by applying airway pressure in proportion to volume and flow rate to overcome respiratory-system elastance and resistance, respectively. The PAV system makes use of the patient’s spontaneous breathing patterns and amplifies proportionally the instantaneous effort throughout each inspiration cycle. It uses positive-feedback control theory to determine the anticipated requirements from a knowledge of its existing values and make instantaneous adjustments accordingly to achieve the necessary flow requirements for the patient. As the patient pulls his/her chest muscles more to inspire, the machine will generate more airflow to supplement this. The ventilator simply amplifies the patient’s effort without imposing any ventilatory or pressure targets. The role of PAV is to allow the patient to comfortabaly attain whatever ventilation and breathing pattern his or her control system sees fit. The responsibility for determining the level and pattern of breathing is shifted entirely from the caregiver to the patient. In this respect, PAV differs fundamentally from all other methods of ventilatory support where ventilatory variables are, by and large, determined by machine settings.

2 Development of proportional-assist ventilation

The early PAV system was described by Younes et al. [2] for an initial clinical trial on some ventilator-dependent patients. Essentially, this prototype consisted of a rolling-seal piston coupled to a motor to generate a force in proportion to the airway pressure. This PAV system is schematically shown in Fig. 2. The current supplied to the motor is proportional to the instantaneous flow rate \( \dot{V} \) and its volume of inspiration \( V \). The control gain \( G \) determines the proportionality. At the early stage of inspiration, as the patient pulls, air moves from the cylinder to the patient. The flowmeter measures the flow rate and the volume can be calculated by integrating the flow rate with respect to time. As flow rate and volume signals are generated, the motor applies pressure according to the flow rate and volume as well as the control gain \( G \). If the patient pulls harder, more flow rate and volume are generated; the assisted ventilation will increase accordingly.

The relationship between the instantaneous effort and ventilatory consequences in healthy persons and in patients with diseases is shown in Fig. 3. The effort is defined as a fraction of maximal inspiratory muscle activation. The application
of PAV compensates for the deficiencies in the diseased case to attain the actual instantaneous requirements of a normal person for a given effort.

Another PAV system that used a linear actuator to push a bellow to provide ventilation to the patients according to the calculated airway pressure was reported by Chua et al [3] and Li et al [4]. This system is shown in Fig. 4 in which a mechanical-driven unit is used to generate a pressurized-air supply.

A summary of the research progress on PAV is as follows:

(i) Clinical trials of PAV on patients with respiratory diseases and evaluation of the short- and long-term effects on the respiratory system, breathing patterns as well as the whole physiological system [2, 5–7].
(ii) Comparison of PAV with other conventional ventilation methods [8].
(iii) Evaluation of the effect of varying degree of assistance on patients and the appropriate determination of proportionalities [3, 4, 9].
(iv) Evaluation of the effect of PAV in some special circumstances such as exercise tolerance and infant ventilatory support [10, 11].
3 Theory of proportional-assist ventilation

During tidal breathing, the elastic and resistive properties of the total respiratory system determine the pressure required to inflate the lungs. This pressure ($\Delta P$) can be divided into its elastic component (pressure required to change volume, $V$) and resistive component (pressure required to generate flow, $\dot{V}$) by the simplified equation of motion of the lungs [12]:

$$\Delta P = E \times V + R \times \dot{V},$$

where $E$ is the elastance (the reciprocal of the commonly used compliance, $C$) and $R$ is the resistance. The relevant elastic properties can be divided into two components, those of the lungs and those of the chest wall, which are arranged in series. Similarly, the resistive pressure can be defined as the total pressure drop in phase with the flow through the separate segments of the tracheobronchial tree and across the lung tissue and chest wall, which are arranged in series. The elastance and resistance of the respiratory system are defined as:

$$E = \frac{\Delta P}{\Delta V},$$

$$R = \frac{\Delta P}{\Delta \dot{V}},$$

where $\Delta V$, $\Delta P$ and $\Delta \dot{V}$ are the changes in lung volume, airway pressure and flow rate, respectively. The typical normal adult values for elastance and resistance are 10 cmH$_2$O/l and 2.5 cmH$_2$O/l/s, respectively.

In the operation of a mechanical ventilator, parameter settings are based on a prior knowledge of the values of elastance and resistance of the respiratory system. Therefore, the assessment of elastance and resistance is important in lung-function tests.
The static pressure–volume curve of the lungs, constructed by measuring the difference between the alveolar and pleural pressures (transpulmonary pressure) at several lung volumes, can be used for the estimation of the static compliance of the respiratory system. Normally, it is obtained during breath holding with a balloon-catheter system or a small pressure transducer in the lower esophagus to approximate pleural pressure. This pressure is compared to the pressure at the airway opening that during breath holding, reflects the alveolar pressure. Measurements of transpulmonary pressure and volume can also be recorded continuously during tidal breathing; the so-called ‘dynamic compliance’ can then be obtained. In normal subjects, dynamic compliance is only slightly less than the static compliance [13, 14].

Generally, spontaneous breathing is entirely accomplished by the respiratory muscles via the generating muscle pressure (\(P_{\text{mus}}\)) while the ventilator provides ventilatory support by changing the airway pressure (\(P_{\text{aw}}\)). Therefore, for patients receiving synchronized ventilatory support, breathing is accomplished through the combined actions of respiratory muscles and ventilator. Thus, at any instant the total applied pressure (\(P_{\text{app}}\)) is given by:

\[
P_{\text{app}} = P_{\text{mus}} + P_{\text{aw}}.
\]  

This total applied pressure is dissipated primarily against two forces, the elastic recoil of the respiratory system (\(P_{\text{el}}\)) and the resistance to gas flow offered by the airways (\(P_{\text{res}}\)), i.e.,

\[
P_{\text{mus}} + P_{\text{aw}} = P_{\text{el}} + P_{\text{res}}.
\]  

During the inflation phase, the pressure difference between the airway pressure (\(P_{\text{aw}}\)) and the alveolar pressure (\(P_{\text{alv}}\)) is the pressure dissipated against the resistance of the airways (\(P_{\text{res}}\)). This difference is related to the flow rate and the resistive properties of the airways (\(R\)). Thus,

\[
P_{\text{res}} = \dot{V} \times R.
\]  

The elastic recoil of the respiratory system (\(P_{\text{el}}\)) is related to the extent the lung volume is above the passive functional residual capacity (\(FRC\)) and the elastance (\(E\)) of the respiratory system. If a linear relationship between the pressure and volume in the tidal volume range is assumed, then

\[
P_{\text{el}} = V \times E,
\]  

where \(V\) is the volume above passive \(FRC\). Substituting eqns. (6) and (7) into eqn. (5) gives

\[
P_{\text{mus}} + P_{\text{aw}} = V \times E + \dot{V} \times R.
\]  

In the PAV system, \(P_{\text{aw}}\) is a function of \(V\) and \(\dot{V}\) that are generated by the patient’s effort. Thus

\[
P_{\text{aw}} = f_1(V) + f_2(\dot{V}).
\]
In the case of linear functions:

\[ P_{aw} = K_1 \times V + K_2 \times \dot{V}, \]  

(10)

where \( K_1 \) is the proportionality between the airway pressure and inspired volume (in elastance units, \( \text{cmH}_2\text{O/l} \)) and \( K_2 \) is the proportionality between the airway pressure and inspired flow rate (in resistance units, \( \text{cmH}_2\text{O/l/s} \)). \( K_1 \) and \( K_2 \) values are determined according to both abnormal neuromuscular and abnormal mechanical disabilities of the patient. Substituting eqn. (10) into eqn. (8) yields

\[ P_{\text{mus}} + K_1 \times V + K_2 \times \dot{V} = V \times E + \dot{V} \times R. \]  

(11)

In this method, if the patient increases or decreases his effort, \( V \) and \( \dot{V} \) will be altered accordingly. Also, as \( P_{aw} \) changes in relation to \( V \) and \( \dot{V} \), machine assist will increase or decrease in concert with \( P_{\text{mus}} \). The net effect is to amplify the patient’s effort while, at the same time, allow the patient to have total control over all aspects of breathing.

In the PA V system, the appropriate ventilatory support, to a great extent, is dependent on the determination of \( K_1 \) and \( K_2 \). Younes [15] proposed the following 4 ways to determine \( K_1 \) and \( K_2 \).

(i) Support in proportion to estimated \( P_{\text{mus}} \).

\( P_{\text{mus}} \) is amplified by the same factor regardless of how it is partitioned between elastic and resistive components and regardless of the time during the inspiration. This approach would be ideal in cases where the problem is primarily in the inability to generate pressure while the mechanical properties are reasonably normal.

(ii) Adjustment according to the nature of mechanical abnormality.

In this mode, the machine-generated pressure will compensate for the extra work resulting from disease, leaving the patient’s muscles to cope with the normal load.

(iii) Combination of (i) and (ii) above.

When the disorder involves both neuromuscular weakness and abnormal mechanics, \( K_1 \) and \( K_2 \) can each be the sum of two components, one specifically for the mechanical abnormality and the other for neuromuscular weakness.

(iv) Adjustment of gain factors according to the patient’s comfort.

Implementation of the above three approaches requires a knowledge of the patient’s mechanical properties (i.e., \( E \) and \( R \)). This is not always possible and the data may not be reliable because of incomplete relaxation. It would seem reasonable to utilize the patient’s own sensory mechanisms to adjust the type and magnitude of assist. The \( K_1 \) and \( K_2 \) settings may be adjusted by trial and error to the values that result in the greatest comfort for the patient.

4 Proportional solenoid valve PA V system

The PA V system to be described here uses a proportional solenoid valve (PSV) to regulate airflow from a compressed air source. The schematic diagram and
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Photograph of this system are shown in Figs. 5 and 6, respectively. The proportional solenoid valve (MYPE-5-M5-010B, FESTO) controls the flow continuously and proportionally to the analogue input. When the input analogue voltage increases from 5 to 10 VDC, the flow will increase from 0 to a maximum value.

Figure 5: Schematic diagram of the proportional solenoid valve PAV system.

Figure 6: Photograph of the proportional solenoid valve PAV system.
The outlet flow and pressure of the PSV are related to its inlet air pressure. Connecting the PSV to a compressed-air source via a pressure regulator and adjusting its analogue input by varying the output signal from a computer, the outlet flow from the PSV can be varied. The flows at different analogue input values are measured by a pneumotachograph (4700A, Hans Rudolph). The pressure drop across the pneumotachograph is measured using a pressure transducer (DP103-14, Validyne) and carrier demodulator (CD15, Validyne). The outlet pressure of the PSV is measured by another pressure transducer (DP15-30, Validyne). Calibrating the PSV at a fixed inlet supply air pressure, the outlet pressure and flow from the PSV for a range of analogue inputs from 5.75 to 9 VDC for various inlet supply pressures are shown in Figs. 7 and 8, respectively. In Fig. 7, the relationship between the outlet pressure and the analogue input can be approximated by a second-order polynomial except for an inlet supply pressure of 0.5 bar. The outlet flow is almost linearly proportional to the analogue input for any fixed inlet supply pressure as shown in Fig. 8. Hence, the use of this solenoid valve to control the desired flow during inspiration can be easily attained. The PSV has a very short response time of 20 ms to an analogue input and this is crucial for short breathing cycles, typically about 3 seconds. If the assisted airway pressure based on the inspired flow and integrated volume, as expressed in eqn. (10), is amplified and taken as the analogue input signal to the proportional solenoid valve, then a proportional-assisted air flow can be delivered to the patient.

A commercial breathing simulator (Series 1101, Hans Rudolph Inc.) was used to mimic human spontaneous breathing for the in vitro tests. The spontaneous breathing pattern of the breathing simulator is determined by the lung-model.

Figure 7: Pressure output from the proportional solenoid valve at various inlet air source pressures.
parameters and the patient effort. The two main lung-model parameters are airway resistance \((R)\) and total elastance \((E)\) of the patient. Normal airway-resistance values for adults range from 5 to 20 cmH\(_2\)O/l/s, whilst normal values of elastance for adults are from 20 to 50 cmH\(_2\)O/l. The patient effort embodies the breathing rate, effort amplitude, effort slope and inhale percentage (% inhale). The effort-amplitude parameter controls the maximum patient effort or peak negative plural pressure generated. By adjusting this parameter, the tidal volume and flow waveform can be changed. The normal amplitude value ranges from 5 to 30 cmH\(_2\)O. The effort slope parameter is used to adjust the shape of the patient-effort waveform by controlling its rise and fall rate.

The PSV-controlled PAV support system incorporating the breathing simulator (Fig. 5) and the control program (Fig. 9) function as follows. As the breathing simulator generates spontaneous effort, air is inhaled through the free airway via a one-way valve and its flow is measured by a pneumotachograph (4700A, Hans Rudolph). This flow signal is sent to a computer via a data acquisition card (PCI-20428W, Intelligent Instrumentation Inc.). The flow signal is then integrated to obtain the volume. The airway pressure is measured by a pressure transducer (DP15-30, Validyne). A closed-loop control program (Visual Designer, Intelligent Instrumentation Inc.) is used in the PAV system. The theoretical airway pressure calculated according to eqn. (8) is used as the setpoint for the PID (proportional, integral and derivative) process value whilst the actual airway pressure will be treated as the current input data. The pressure difference between the calculated and the actual airway pressure is the error input of PID. The output of the PID is multiplied by an amplification factor \((Kc)\) to transfer the pressure signal to an analogue voltage signal. As the operating input signal of the proportional solenoid valve ranges from 5 to 10 VDC, a 5-VDC offset value is added to the analogue voltage signal. This total voltage is

![Figure 8: Flow output from the proportional solenoid valve at various inlet air source pressures.](image-url)
then used as the analogue input to the proportional solenoid valve to control the flow. The PAV-assisted air joins the atmospheric air intake through the free airway via a ‘Y’ type connector (Fig. 5). The exhalation valve (ADK12-15A, CKD Corporation) shown in Fig. 5 is a normally open-mode solenoid valve. It operates as a switch that will open during exhalation and close during inhalation. To ensure that PAV support is cut off at the end of each inspiration, the parameter $K_c$ is set to zero when the flow declines and falls below a certain preset value that is 0.05 l/s in the tests reported here. Immediately, the exhalation valve is triggered to open so that the exhaled air can exit into the atmosphere. The exhalation valve will be triggered to close at the beginning of inspiration when the inspired flow rises above zero. Flow rate, volume and airway pressure are continuously monitored and displayed on a strip-chart recorder. These signals are also stored as data files for subsequent analyses.

In a PSV-controlled PAV system, pressure ‘run-away’ is a common phenomenon when the proportionality settings are inappropriately used and an excessive-pressure limit mechanism is not incorporated. Pressure ‘run-away’ occurs when the pressure provided by the ventilator at the end of inspiration exceeds the opposing elastic and resistive pressures. Flow delivery continues during patient’s neural expiration and airway pressure continues to increase during expiration [16]. A peak inspiratory pressure greater than 30–40 cmH$_2$O increases the risk of barotrauma. Therefore, a pressure-limit mechanism is designed to prevent the occurrence of pressure ‘run-away’ by switching off the proportional solenoid valve when it occurs. Basically, the control program has the following safety features. A timer is used to monitor the opening duration of the proportional solenoid valve. The PSV will be triggered to switch off when the opening duration reaches a preset value. Simultaneously, the inspired flow, volume and airway pressure are monitored and compared with clinically determined maximum thresholds. If any of these values exceeds the...
corresponding threshold value, the proportional solenoid valve will be switched off and the exhalation valve triggers to open. In this way, pressure ‘run-away’ can be effectively curbed.

For the in vivo tests, the human subject replaces the breathing simulator in the PAV system in Fig. 5 and the modified system is shown in Fig. 10. The subject breathes spontaneously from a two-way nonrebreathing facemask (Hans Rudolph). The proportional solenoid valve used in this system is an air/O₂ servoid metering valve (South Bend Controls) that is medically safe for patients. As the activating signal of this valve is based on current instead of voltage, a 1-A output transmitter is used to convert a 0–10-VDC to 50–350 mADC. An additional pneumotachograph is installed at the expiration line to monitor the expired flow. Also, the pressure in the facemask is measured by a pressure transducer (Type 7261, Kistler) and charge amplifier (Type 5011, Kistler) to monitor the subject’s effort.

5 Experimental results and discussion

Experimental results on the PAV system are presented for the in vitro tests based on the breathing simulator and in vivo tests for the healthy subjects.

5.1 Spontaneous breathing tests on breathing simulator

To verify whether the breathing simulator does mimic the human respiratory system, a group of spontaneous tests was conducted to determine the response of tidal
To determine the influence of the airway resistance to spontaneous breathing, the response of the tidal volume to different resistances was tested under a given elastance and patient-effort combination. In these tests, various elastance and patient-effort combinations were used. The resistance was varied from normal (low) to abnormal (high); these values were 10, 20, 30, 50 and 80 cmH2O/l/s. Figure 11 shows the tidal-volume response to different resistances under various elastance and effort combinations. Increasing the resistance from 10 to 80 cmH2O/l/s results in reducing the tidal-volume progressively for every elastance and effort combination. Thus this breathing simulator can be used to mimic patients with high airway resistance.

Similarly, the resistance and the patient effort were fixed whilst the lung elastance was varied to determine its effect on spontaneous breathing. The elastance was varied from 20 to 100 cmH2O/l for various resistance and effort combinations; the tidal volume response can be observed in Fig. 12. Increasing the elastance progressively decreases the tidal volume, which is to be expected in the human respiratory system.

Normally, a patient with neuromuscular weakness cannot generate enough negative pleural pressure to pull the lungs to a more expanded position. Therefore, the normal required tidal-volume for the patient cannot be met. This neuromuscular weakness can be simulated by having a small effort amplitude in the breathing simulator. In the following tests, the effort amplitude was set at 5, 10, 15 or 20 cmH2O in each test. These simulated values would mimic from a very weak effort or

Figure 11: Tidal-volume response to different resistances under various elastance and effort combinations.
breathing at rest to heavy breathing or a person carrying out strenuous exercise. Figure 13 shows the tidal-volume response to different effort amplitudes under various resistance and elastance combinations. At each constant resistance and elastance setting, the tidal-volume increases almost linearly with increasing patient-effort amplitude. The combination with lower resistance and elastance values corresponds to a larger gradient; and consequently a larger lung expansion with a greater
tidal-volume. In all the combinations, lung expansion is directly proportional to the patient effort. Therefore, this breathing simulator can simulate both normal subjects and patients with neuromuscular weaknesses.

In summary, the breathing simulator can mimic a wide range of patient conditions by setting different lung-mechanics parameters and patient-effort shapes. Therefore, it is suitable to be used in the evaluation of the PAV system.

5.2 Performance tests on PAV system using breathing simulator

Tests were conducted on the breathing simulator to mimic patients with either high resistance, high elastance or weak effort. In this study, the normal resistance, elastance and patient effort values were assumed to be 20 cmH2O/l/s, 20 cmH2O/l and 10 cmH2O, respectively, and the corresponding normal tidal volume given by the breathing simulator was about 0.269 l as shown in Fig. 11. The abnormal resistance, elastance and patient-effort values were set at 30 cmH2O/l/s, 50 cmH2O/l and 5 cmH2O, respectively. In all the tests, the spontaneous breathing effort was assumed to be constant for simplicity. The breath rate, effort slope and % inhale were set at 20 breaths per minute, 10 and 40%, respectively.

5.2.1 PAV tests on breathing simulator with high resistance

In this study, the elastance and patient effort of the breathing simulator were set at normal values of 20 cmH2O/l and 10 cmH2O, respectively, but the resistance was set at a relatively higher value of 30 cmH2O/l/s. Hence, the PAV should mainly overcome the extra resistance and the proportionalties $K_1$ and $K_2$ were set accordingly. The breathing patterns in terms of flow, volume and airway pressure with and without the implementation of PAV are compared in Fig. 14. During the first six breaths, the ‘patient’ (breathing simulator) breathed without assistance, and from the seventh breath onwards, the PAV commenced to supply air to assist the ‘patient’. It can be observed that, with the implementation of PAV, the flow, volume and airway pressure increase significantly. During spontaneous breathing without assistance (Fig. 14(c)), the airway pressure was observed to be slightly negative during inspiration. However, when the PAV was initiated, the airway pressure increased sharply. Subsequently, different $K_1$ and $K_2$ combinations were tested. The breathing patterns in a typical breathing cycle before and after the implementation of PAV are illustrated in Fig. 15. The solid line represents the breathing pattern without the implementation of PAV, whilst the rest represent the breathing patterns with the implementation of PAV under different $K_1$ and $K_2$ combinations.

It can be observed that with the implementation of PAV, the peak inspiration and expiration flow, tidal-volume, airway pressure and inspiratory time increase for each $K_1$ and $K_2$ combination as compared to those without PAV. However, the flow only increases marginally at the beginning of the inspiration period, i.e. during the first 0.75 second (Fig. 15(a)). This may be due to two reasons. First, at the beginning of the inspiration, both flow and inspired volume are relatively small and therefore the calculated theoretical airway pressure is also small.
Figure 14: Tracings of breathing patterns before and after the implementation of PAV ($R = 30 \text{cmH}_2\text{O}/l/s$; $E = 20 \text{cmH}_2\text{O}/l$; Effort = 10 cmH$_2$O; $K_1 = 0$; $K_2 = 30 \text{cmH}_2\text{O}/l/s$).
Hence, the computed signal for the proportional solenoid valve, which is proportional to the difference between the theoretical and the actual airway pressures, is also small, resulting in a small opening of the valve. Secondly, there are time delays in (i) the opening of the proportional solenoid valve (this can be possibly a very
small time delay as the valve response time is 20 ms) and (ii) gas transmission from PSV to the patient-connection port. One possible way to reduce the delay time is to decrease the tube length between the PSV and the patient-connection port. The proportional solenoid valve was not triggered to close at the end of spontaneous inspiration but only after the ‘patient’ had generated sufficient positive pressure to balance the airway pressure. The inspired flow will persist a little longer after the commencement of the exhalation phase of spontaneous breathing. This accounts for the PAV inspiration time to be longer than that of spontaneous breathing. The sharp ‘spike’ in the flow (Fig. 15(a)) during exhalation is due to the sudden opening of the exhalation valve at the end of inspiration, resulting in the venting of the airway pressure at the patient connection port to the atmosphere.

In Fig. 15(b), it can be observed that one of the benefits of PAV is that the tidal-volume increases. However, the minimum volume at the end of each breathing cycle increases slightly when PAV is initiated. This is due to the initial higher resistance set and shorter exhalation time such that the lung cannot be emptied sufficiently, resulting in an increase in the residual volume.

A comparison between two groups of proportionality settings $K_1 = 0$, $K_2 = 30 \text{cmH}_2\text{O/l/s}$ and $K_1 = 10 \text{cmH}_2\text{O/l}$, $K_2 = 20 \text{cmH}_2\text{O/l/s}$ shows that the tidal-volume in both cases has exceeded the original normal value (0.269 l in this case), demonstrating a positive effect of PAV in overcoming high resistance. However, the tidal volume and airway pressure increase more for the former settings than those of the latter. Since the abnormality is resistance in this case, the effect of increasing flow-assist proportionality ($K_2$) is better than that of increasing volume-assist proportionality ($K_1$). This result clearly shows that proper proportionality settings should be based on a good knowledge of the patient’s resistance and elastance.

### 5.2.2 PAV tests on breathing simulator with high elastance

In these tests, the resistance and patient effort of the breathing simulator were set at normal values of $20 \text{cmH}_2\text{O/l/s}$ and $10 \text{cmH}_2\text{O}$, respectively, but the elastance was set at a relatively higher value of $50 \text{cmH}_2\text{O/l}$. Hence, the function of the PAV is to overcome the higher elastance and thus the proportionalities $K_1$ and $K_2$ were set accordingly. The breathing patterns in a typical breathing cycle before and after the implementation of PAV for different $K_1$ and $K_2$ combinations are illustrated in Fig. 16. The solid line represents the breathing pattern without the assistance of PAV, whilst the rest represent the breathing patterns with the implementation of PAV under different $K_1$ and $K_2$ combinations. From Fig. 16, it can be observed that the flow, tidal-volume and airway pressure increase significantly with the implementation of PAV as compared to those without PAV. In Fig. 16(a), with the implementation of PAV, the reasons for the marginal increase in flow at the beginning of the inspiration period, the longer inspiration time and the ‘spike’ in flow at the beginning of exhalation are similar to those for Fig. 15(a). The tidal volumes attained for these different $K_1$ and $K_2$ combinations have exceeded the normal tidal volume, demonstrating a positive effect of PAV in overcoming high elastance.
Comparing the proportionality settings of $K_1 = 40\text{ cmH}_2\text{O/l}$, $K_2 = 0$ and $K_1 = 30\text{ cmH}_2\text{O/l}$, $K_2 = 10\text{ cmH}_2\text{O/l/s}$ in Fig. 16(b) and (c) show that increasing volume-assist proportionality $K_1$ is more effective than flow-assist proportionality $K_2$ to overcome the patient’s high elastance abnormality.
5.2.3 PAV tests on breathing simulator with weak effort

Tests were carried out for a weak patient-effort amplitude of 5 cmH$_2$O whilst the resistance and elastance were set at normal values of 20 cmH$_2$O/l/s and 20 cmH$_2$O/l, respectively. The $K_1$ and $K_2$ values were increased gradually at interval of 10, with the starting values at 10 cmH$_2$O/l and 10 cmH$_2$O/l/s for $K_1$ and $K_2$, respectively. The PAV system performed well for the first two values of $K_1$ and $K_2$. However, when $K_1$ and $K_2$ were increased to 30 cmH$_2$O/l and 30 cmH$_2$O/l/s, respectively, pressure ‘run-away’ occurred. The stable breathing patterns in one breath before and after the implementation of PAV using the first two $K_1$ and $K_2$ combinations are shown in Fig. 17, whilst Fig. 18 shows the breathing patterns when pressure ‘run-away’ occurred after PAV was initiated.

It can be seen in Fig. 17 that although there are some increases in the flow, tidal-volume and airway pressure, the small PAV enhanced effect is not capable of providing sufficient tidal-volume to ‘pull’ the patient to his/her normal breathing patterns. This implies that the PAV system is not quite suitable for patients with very weak muscle effort and therefore some kind of forced mechanical ventilation should be provided for them.

To avoid damage to the breathing simulator, the proportional solenoid valve is automatically switched off when pressure ‘run-away’ occurs. Otherwise, the flow will continue to increase with increases in the valve opening until it is completely open. The flow will then remain at that maximum value and therefore the volume will increase unboundedly. Then pressure ‘run-away’ ensues. The upper limit of the opening duration for the proportional solenoid valve was set at a large value of 5 s. If the inspiration period is 5 s or more, the solenoid valve will automatically cut off the supply flow to the breathing simulator. The 5-s limit is longer than the normal tolerant inspiration time. The cutoff of supply flow is only momentary; in the next inspiration cycle, the proportional solenoid valve will open and continue to control the supply flow until the next pressure ‘run-away’ occurs.

In Fig. 18(a), the flow patterns and breathing rates before and after the implementation of PAV differ significantly although the spontaneous breathing effort remains unchanged. In the pressure ‘run-away’ situation, the breathing patterns are quite irregular after PAV is initiated. During the spontaneous expiration when PAV is on, the proportional solenoid valve is still supplying inflow to the breathing simulator as reflected by the increasing inspired volume (Fig. 18(b)) and the increasing airway pressure (Fig. 18(c)). However, when the opening duration of the solenoid valve has reached the pre-set 5 s, it suddenly shuts off and stops any further inflow, as shown in Fig. 18(a). This demonstrates that the pressure ‘run-away’ limit mechanism functions properly. As the proportional solenoid valve suddenly cuts off the inflow completely, the air continues to flow into the breathing simulator and momentarily the peak alveolar pressure is even greater than the airway pressure. Therefore, the exhalation valve is triggered to open and the airway pressure drops sharply (Fig. 18(c)), accompanied by a high peak expiratory flow as shown in Fig. 18(a).

In practice, the maximum opening duration of the solenoid valve will not be set for more than 2 s in the program for safety consideration. The reason for setting
Figure 17: Illustrations of the effects of PAV in overcoming weak effort under different $K_1$ and $K_2$ combinations.
Assisted airflow was terminated by switching off the solenoid valve.

Figure 18: Illustrations of the breathing patterns when pressure 'run-away' occurs and the triggering of the pressure-limit mechanism.
5 s in the above instance is to demonstrate clearly that without a pressure-limit mechanism, the airway and alveolar pressures will continue to increase unabatedly and lead to a serious pressure ‘run-away’ situation.

5.3 Simulated tests on healthy human subjects

Breathing tests were carried out on six normal subjects with artificial resistance and elastance incorporated. Four males and two females, with ages ranging from 23 to 32, volunteered for this study. In order to mimic patients with high resistance, a series of orifice resistors were incorporated in the airway upstream of the pneumotachograph (PNT 1) in Fig. 10. The resistance of the resistor was varied by changing the diameter of the center hole from 5 to 10 mm at a step interval of 1 mm. The resistances of the various orifice resistors are given in Table 1. For simplicity, the orifice resistors are referred to here as resistor 5 to 10 where the number specifies the diameter of the orifice. Before the in vivo PAV tests, it is necessary to determine the influence of the orifice resistors on the normal subjects’ breathing conditions. Each person was instructed to breathe through a facemask that was connected to an orifice resistor and a pneumotachograph. Figure 19 shows the typical flow patterns when a subject breathed through resistors 10 to 5. Increasing resistance (or decreasing orifice diameter) decreases the flow rate and hence the tidal volume also decreases. The subjects were able to maintain a stable breathing effort when breathing through resistors 10 to 8 and they could sustain this for a longer time without assist. Their flow patterns were relatively stable, unlike those for resistors 7 to 5 as shown in Fig. 19. For resistors 7 to 5, the subjects

<table>
<thead>
<tr>
<th>Orifice Diameter (mm)</th>
<th>5</th>
<th>6</th>
<th>7</th>
<th>8</th>
<th>9</th>
<th>10</th>
</tr>
</thead>
<tbody>
<tr>
<td>Resistance (cmH₂O/l/s)</td>
<td>100.88</td>
<td>54.80</td>
<td>37.08</td>
<td>15.81</td>
<td>10.71</td>
<td>8.68</td>
</tr>
</tbody>
</table>

Figure 19: Typical flow patterns of a normal subject breathing through different orifice resistors.
Resistor 9

Resistor 8

Resistor 7

Resistor 6

Figure 19: Continued.
opined that it was very difficult to breathe with a normal breathing effort. Therefore, incorporating orifice resistors could, to some extent, mimic patients with high resistance.

To mimic patients with high elastance, a slim belt was strapped to the chest of the six normal subjects at varying degrees of tightness. The muscular movement of the subject was restricted in some extent by the slim belt and the subject would exert much more breathing effort to obtain the desired ventilation. This is similar to a patient with a higher elastance whose breathing effort is mainly dissipated towards expanding the stiff lungs. The typical flow-rate patterns of a healthy subject breathing with different belt tightness exerted on the chest are shown in Fig. 20. Increasing belt tightness that corresponds to increasing elastance of the respiratory system, decreases the inspiratory flow rate. Therefore, to some extent, it is possible to mimic a patient with higher elastance by a slim belt strapped on the subject’s chest.

5.3.1 In vivo PAV tests to overcome high resistance
In vivo PAV tests were carried out on the six normal subjects to overcome the high resistance induced by the orifice resistors. The subjects were requested to remain as relaxed as possible and to maintain a relatively steady breathing effort throughout the entire test cycles. Figure 21 shows the breathing patterns before and after the implementation of PAV for a subject when breathing through resistor 7 with proportionalities $K_1$ and $K_2$ set at 5 cmH$_2$O/l and 10 cmH$_2$O/l/s, respectively. Before PAV was initiated, the subject breathed unassisted for several cycles. Figure 21 shows that the flow rate, inspired volume and airway pressure of the subject increase after the activation of PAV. The most significant change with the implementation of PAV is the airway pressure that shows a sharp increase when PAV is activated. It also shows that the assist in each cycle is not consistent due to varying spontaneous efforts. Increasing effort increases the assist, which is one of the advantages of PAV.

In some of the tests, the flow rate and inspired tidal-volume decreased after the activation of PAV as illustrated in Fig. 22. In these cases, the subjects felt...
comfortable and therefore they relaxed their efforts when breathing with the help of PAV. This is reflected in the positive airway pressures at the commencement of each breathing cycle after the activation of PAV, indicating that the subject had not maintained the same breathing efforts as compared to those before the activation of PAV. If the subject pulls less, PAV will also reduce its assist. Therefore, it is reasonable to observe in Fig. 22 that the tidal-volumes after the activation of PAV are less than those before its activation. This phenomenon demonstrates the positive effect of PAV as it relieves the subject’s effort to obtain a comfortable tidal-volume.

Figure 20: Typical flow-rate patterns of a normal subject with different belt tightness exerted on the chest.
**Figure 21:** Representative breathing patterns before and after the implementation of PAV for a healthy subject when breathing through resistor 7 with proportionalities $K_1$ and $K_2$ set at 5 cmH$_2$O/l and 10 cmH$_2$O/l/s, respectively.

### 5.3.2 In vivo PAV tests to overcome high elastance

*In vivo* PAV tests were carried out on the six normal subjects to overcome the high elastance induced by the slim belt. To overcome high elastance, PAV is mainly employed as a volume-assist mode. Therefore, $K_1$ was varied according to different belt tightness. Flow rate, volume and airway pressure patterns in a typical breathing cycle before and after the activation of PAV at different $K_1$ levels are shown in Fig. 23. Increasing $K_1$ increases the ventilatory consequences, demonstrating the positive effect in assisting patients with higher elastance than normally should be.
Figure 22: Breathing consequences of PAV in relieving a healthy subject’s breathing effort.

The subjects opined that they felt much more comfortable to breathe when PAV was activated.

6 Conclusions

A proportional-assist ventilation system based on a proportional solenoid valve to control air delivery from a pressurized source to a patient, was evaluated. Respiratory tests using this proposed system were conducted on a breathing simulator that is capable of mimicking the spontaneous breathing responses of patients suffering from pulmonary diseases and respiratory disabilities. The results demonstrated the viability of the PAV system in normalizing the breathing patterns of patients with abnormal resistance, elastance and weak effort after appropriate selection of $K_1$ and $K_2$ proportionalities. Pressure ‘run-away’ can be prevented.
Figure 23: Representative breathing consequences with and without PAV for a healthy subject strapped with a slim belt.

by incorporating a back-up safety mechanism in the control program to ensure the safe operation of the PAV system. In vivo tests using the proposed PAV system were carried out on volunteer healthy persons with incorporated artificial resistance and elastance. Significant increases in inspired flow, tidal-volume
and airway pressure after the implementation of PAV were obtained. The test subjects felt comfortable to breathe with reduced breathing effort when PAV-assist ventilation was initiated, thereby demonstrating the positive assist effect of PAV.

References


