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Research article

Modeling and control of an invasive mechanical ventilation system using the active disturbances rejection control structure

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A B S T R A C T

We propose a mandatory invasive mechanical ventilator prototype for severe COVID-19 patients with volume and pressure control operation modes. This system comprises basic pneumatic elements and sensors. Its performance is similar to commercial equipment, and it presents robustness to external disturbances and parametric uncertainties. To develop a control strategy, we propose a mathematical model with a variable structure that incorporates the dead zone phenomenon of the proportional valve, and considers external disturbances and parametric uncertainties. Based on this model, we propose a global control strategy that is based on pressure and flow regulation controllers, which use the active disturbances rejection control structure (ADRC). In this strategy, we propose robust state observers to estimate disturbances and the signals necessary for implementing the controllers. We illustrate the performance of the prototype and the control strategy through numerical simulations and experiments. We also compare its performance with PID controllers. These results corroborate its effectiveness and the possibility of its application in invasive mechanical ventilators with a simple structure, which can significantly help critical care of COVID-19 inpatients.

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1. Introduction

Invasive mechanical ventilation (IMV) is essential in treating patients with acute respiratory distress syndrome (ARDS), which is a feature in severe COVID-19 [1,2]. The respiratory symptoms of COVID-19 can go from mild flu-like symptoms to respiratory failure in minutes to hours, depending on the state of health before infection. Complications caused by viral infection mainly occur in older adults and are the most serious in those with comorbidities. Mechanical ventilation can decrease the work of breathing, increase oxygenation and remove carbon dioxide in patients with pneumonia, ARDS, or respiratory failure associated with COVID-19 [3,4].

During the years 2020 and 2021, the world experienced an overwhelming scenario due to a rapid spreading disease with a globally poor medical surge capacity. No country was prepared to meet such a large number of critically ill patients. In some cases, the shortage of ventilators was significant.

Given the current scene of a delta variant and the predicted behavior of the disease, the urgency of the availability of mechanical ventilators has remained because respiratory support is still the first-line treatment for COVID-19.

Mechanical ventilators that use pneumatic valves to carry out the assisted or controlled breathing cycle are the most commonly used today in intensive care departments. This type of ventilator is generally connected to separate sources of compressed air and oxygen, which allows them to deliver the amount of oxygen concentration required by the patient for long periods. In addition, there are many proportional valves that enable the supply of oxygen needed for the patient with high precision [5,6]. However, this type of device introduces non-linear dynamics, which are more difficult to control. Borrello et al. [7] present a complete analysis of the characteristics, advantages, and disadvantages of the different technologies used in mechanical ventilators and the challenges in developing control algorithms.

Some adaptive control techniques have been applied to solve the pressure tracking control problem. An example can be found in [8], where the authors propose an adaptive controller that is based on an inverse model of a patient’s lung and the ventilator. This controller estimates the parameters of the system

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online using the recursive least square with a forgetting factor. This proposal achieves robust performance over a wide range of patient conditions. This controller was applied on NPB 840 mechanical ventilator and obtained good results; however, the controller produces overshoots at the beginning of the expiration stage.

Active Disturbance Compensation Control (ADRC) is a control structure that presents good robustness properties and has a certain simplicity in its implementation. This control structure has been successfully applied in mechanical systems to solve regulation and tracking control objectives. A detailed description of this control structure and some applications can be found in [9,10]. This control structure was successfully applied in [11] to track trajectories in a mechanical system of one degree of freedom, with linear movement, using pneumatic actuators. This control strategy has also been applied to control a Bag Valve-Based mechanical ventilator prototype, which is a portable mechanical ventilator that is based on bag-valve compression through a flexible belt and a DC motor [12]. This kind of ventilator is beneficial in providing short-term care for patients; however, there is a risk of accumulation of CO2 inside the bag. Here, a double integrator with matched nonvanishing disturbances is used to model the volume and pressure control design. Because the proposed state observer is linear and there are nonvanishing disturbances in the plant, it does not guarantee that the observation error will converge to zero, and therefore it does not guarantee adequate compensation. The experimental results show an acceptable performance compared to a commercial mechanical ventilator, presenting a considerable error between the pressure and the PEEP pressure level in the expiration period.

This work presents a mandatory invasive mechanical ventilator prototype that is based on valves for severe COVID-19 patients with volume and pressure control operation modes. This system comprises a service unit, a proportional valve for flow control, an on/off valve, an artificial lung, pressure and flow sensors, and a controller device. We propose a mathematical model with a variable structure to develop a control strategy that incorporates the dead zone phenomenon in the proportional valve, and which considers external disturbances and parametric uncertainties. Based on this model, we propose pressure and flow regulation controllers that use the ADRC structure. We then propose a global control strategy that, depending on the operation mode selected, volume or pressure, commutes the pressure and flow controllers to generate the desired performance.

To implement the controllers in this strategy, we propose robust state observers to estimate disturbances and the necessary signals. The result is a prototype with minimum error in pressure and flow, in both operation modes, and in robustness.

We illustrate the prototype’s performance and the control strategy through numerical simulations and experiments, both with and without external disturbances. We also compare its performance with PID controllers. These results corroborate its effectiveness and the possibility of its application in invasive mechanical ventilators with a simple structure, which can significantly help COVID-19 critical care inpatients.

The rest of this article is organized as follows. The second section presents the basic definitions of an invasive mechanical ventilator and it defines the problem. The dynamic ventilator model is proposed in the third section, which allows the design of the control strategy. Meanwhile, sections four and five present the design of the volume and flow controllers, respectively. Through the flow control, the volume control design is presented in Section 4. In Section 5, the pressure controller is presented. Section 7 presents the overall control strategy, which switches the flow and pressure controllers, depending on the mode of operation and the state of the breath cycle. Section 8 presents the performance of the proposed control strategy through numerical simulations. Section 9 presents the experimental results, which illustrate the adequate performance of the proposed control strategy. Section 10 presents the mechanical ventilator’s experimental performance for volume and pressure operation modes, using PID controllers, and concludes that the ADRC control structure has a better performance. Finally, the conclusions and final comments are presented in Section 11.

2. Description of the pneumatic system and problem statement

The proposed invasive mechanical ventilation prototype can be classified as a mandatory or assisted continuous ventilation system, with two modes of operation: volume control mode and pressure control mode. The parameters and variables that establish its operation follow.

- Ventilation period \( T \). Period of time, in minutes, that a respiration cycle takes, which is divided into two stages: the inspiration time \( T_I \) and the expiration time \( T_E \), where \( T = T_I + T_E \).
- Inspiration–expiration ratio \( I : E \). This is the relationship that exists between the inspiration time \( T_I \) and the expiration time \( T_E \); it is calculated as \( I : E = T_E / T_I \).
- Ventilation frequency \( F \). This is the inverse of the ventilation period \( T \), whose units are cycles per minute.
- Tidal volume \( V_t \). This the volume, in units of milliliters, of gas entering, or leaving, the lungs in a given amount of time.
- Positive End-Expiration Pressure PEEP. This is the positive pressure that must remain at the end of expiration time to keep alveoli distended and avoid alveolar collapse, it is measured in centimeters of water (cm H2O).
- Peak inspiratory pressure PIP. This is the maximum reference pressure in pressure control mode, its units are cm H2O.
- Limit pressure \( P_{\text{lim}} \). This is the safety pressure level in units of cm H2O. In volume operation mode, the ventilator cannot exceed this pressure level.

In addition to these operating parameters, there are two variables that determine the operation of the mechanical ventilator.

- “On/Off”, if this variable is equal to 1, then the system is in operation; if it is equal to 0, then the system is deactivated.
- “Mode”, this variable indicates the operating mode: when “Mode = 0”, the system operates in volume control mode; while if “Mode = 1”; then it operates in pressure control mode.

A diagram of the pneumatic system within the invasive mechanical ventilator and its instrumentation is shown in Fig. 1. In this figure, the gas inlet path, in the inspiration process, is indicated by the red arrows; while the gas outlet path, in the expiration process, is indicated by the green arrows. The inlet route starts with a constant inlet pressure \( P_0 \), which is obtained through a service unit. Next is a proportional valve for flow control, controlled by the voltage \( V_I(t) \). Pressure and flow sensors are placed at the outlet of this valve, which provide measurements of the inspiration flow \( F_I(t) \) and the pressure \( P(t) \); we assume that lung pressure \( P_{\text{lung}}(t) \) is equal to pressure \( P(t) \). In the gas outlet path, there is a flow sensor to measure the outlet flow \( F_O(t) \). In this way, the net flow \( F(t) \) in the lung is \( F(t) = F_I(t) - F_O(t) \), while the tidal volume \( V(t) \), which is defined as the volume of gas entering the lungs in the respiration period \( T \), is

\[ V(t) = \int_0^T F(t) \, dt. \]
Next to the flow sensor is an on/off valve that is activated by the voltage \( V_e(t) \), which allows the gas to escape. Finally, \( P_{env} \) is the ambient pressure, which is considered the reference of the system. Throughout this paper, the pressure units are centimeters of water (cm H2O), the flow units are standard liters per minute (SLMP), the volume units are milliliters (ml), the \( V_i(t) \) control units are volts (V), while the control \( V_e(t) \) of the outlet valve has the interpretation of on and off.

The problem addressed is to propose a robust control strategy for the pneumatic system, shown in Fig. 1, to operate as a mandatory invasive mechanical ventilator for critically ill COVID-19 patients, with volume control and pressure control modes of operation. The first step is to propose a mathematical model of the system that incorporates its main dynamic characteristics but at the same time must be as simple as possible to allow its parameters and the design of the controllers to be estimated. Next, robust controllers must be proposed to solve volume regulation problems through flow control and pressure regulation, despite the presence of external disturbances, parametric uncertainties, and unmodeled dynamics. Finally, the third step is to propose a global control strategy that allows the system to switch between the pressure and volume control modes of operation that is based on parameters set by the user.

### 3. Pneumatic system modeling

The dynamics of the system shown in Fig. 1 are strongly nonlinear due to the dead zone in the valves, fluid dynamics, and delays caused by the lines connecting the valves to the patient. This section proposes a model, which is as simple as possible, that includes the most representative dynamics of the system and which allows the design of robust controllers that solve the flow and pressure control objectives.

First, the inspiration process is analyzed. The flow \( F(t) \) has a behavior that is very similar to a first-order system. Consequently, the following model is proposed

\[
\dot{V}_i(t) + a_1 F(t) = b_1 z_m(V_{di}, V_i(t)) + \gamma_1(t),
\]

where \( a_1 \) and \( b_1 \) are positive constants, \( z_m(V_{di}, V_i(t)) \) is the dead-zone function defined as

\[
z_m(V_{di}, V_i(t)) = \begin{cases} 
0, & V_i(t) \leq V_{di} \\
V_i(t), & V_i(t) > V_{di} 
\end{cases},
\]

and \( \gamma_1(t) \) is a disturbance term that incorporates non-modeled dynamics and external disturbances, which are considered bounded and with bounded derivatives for all \( t \) and \( F(t) \).

The function \( z_m(\cdot) \), Fig. 2(c), can be represented as the subtraction of a linear function, 2(a), with a saturation function \( \text{sat}(\cdot) \), 2(b), so that the system (2) can be rewritten as

\[
\dot{V}_i(t) + a_1 F(t) = b_1 V_i(t) - b_1 \text{sat}(V_i(t), V_{di}) + \gamma_1(t),
\]

For \( \dot{s}(t) = 1 \) the model of the system is given by the equation

\[
\dot{P}(t) = -\alpha P(t) + \frac{1}{C_{lung}} F(t) + \gamma_{pat}(t),
\]

where \( \gamma_{pat}(t) \) is a term that contains external disturbances and modeled dynamics, with bounded amplitude and derivative.

For \( \dot{s}(t) = 0 \) the model of the system is given by the equations

\[
\dot{P}(t) = -\alpha P(t) + \frac{1}{C_{lung}} F(t) + \gamma_{pat}(t),
\]

where \( \gamma_{pat}(t) \) is a disturbance exerted by the patient’s muscles, which is considered limited in amplitude and its derivative.

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\dot{V}_i(t) + a_1 F(t) = b_1 V_i(t) - b_1 \text{sat}(V_i(t), V_{di}) + \gamma_1(t).
\]
where \( x(t) \in \mathbb{R} \) is the state, \( f(x) \) and \( g(x) \) are known functions, \( u(t) \) is a control input and \( \gamma (x, t) \) is a disturbance term that satisfies the condition \( |\dot{\gamma}(x, t)| < \delta \) where \( \delta \) is a constant. The problem is to estimate the disturbance term \( \gamma (x, t) \).

For this purpose, we propose the observer given by

\[
\dot{x}(t) = f(x) + g(x)u(t) + c_1(x(t) - \hat{x}(t)) + \omega(t),
\]

\[
\dot{\omega}(t) = c_2(x(t) - \hat{x}(t)) + c_3 \text{sign}(x(t) - \hat{x}(t)),
\]

\[
\hat{y} = \hat{x}.
\]

where \( \omega(t) \) is an auxiliary state acting as an estimate of the disturbance \( \gamma (x, t) \) in system (8) and the coefficients \( c_i, i = 1, 2, 3 \) are positive. It is important to mention here that the solutions of system (9) are defined in Filippov’s sense [14].

To demonstrate the estimation of the disturbance term in system (8), the error variable \( \varepsilon (t) = x(t) - \hat{x}(t) \) is defined, whose dynamics are given by

\[
\dot{\varepsilon}(x) = \gamma(x, t) - c_1 \varepsilon(t) - \omega(t),
\]

\[
\dot{\omega}(t) = c_2 \varepsilon(t) + c_3 \text{sign}(\varepsilon(t)).
\]

Now we make the change of variables \( z_1(t) = \varepsilon(t) \) and \( z_2(t) = \gamma(x, t) - c_1 \varepsilon(t) - \omega(t) \), whose dynamics are given by

\[
\dot{z}_1(t) = \dot{z}_2(t),
\]

\[
\dot{z}_2(t) = -c_2z_1(t) - c_3z_2(t) - c_3 \text{sign}(z_1(t)) + \dot{y}(x, t),
\]

then there exists a positive definite matrix \( P \), which is the solution of the Lyapunov equation

\[
A^TP + PA = -I,
\]

where \( I \) is the identity matrix and \( A \) is given by

\[
A = \begin{bmatrix} 0 & 1 \\ -c_2 & 1 \end{bmatrix}.
\]

Here, \( \lambda_{\text{min}}(P) \) and \( \lambda_{\text{max}}(P) \) denote the minimum and maximum eigenvalues of matrix \( P \), respectively. We then have the following theorem.

**Theorem 1.** For system (10), suppose that \( |\dot{\gamma}(x, t)| < \delta \). If

\[
c_3 > 2\lambda_{\text{max}}(P) \left[ \begin{array}{c} \lambda_{\text{max}}(P) \\ \lambda_{\text{min}}(P) \end{array} \right] \left( \frac{c_3 \delta}{\theta} \right)
\]

where \( 0 < \theta < 1 \), then the origin of the state space will be an asymptotically stable equilibrium point in the Lyapunov sense. Consequently,

\[
limit_{t \to \infty} \omega(t) = \gamma(x, t).
\]

**Proof.** The proof of this Theorem can be found in [15].

In practice, the value of \( \delta \) is not known but it exists, so a tuning process is carried out to define the values of the observer gains.

### 5. Volume controller design

Volume control in the lung is done indirectly through flow control entering the lung over a period of time. For simplicity, a constant flow reference \( F_{\text{ref}} \) is considered whose value is such that at the inspiration time \( T_i \), the tidal volume \( V_t \) is achieved; that is, \( F_{\text{ref}} = V_t/T_i \).

Because flow control is only applied in the inspiration process, the plant model is given by

\[
\dot{F}(t) = -a_1F(t) + b_1V_t(t) + G(\cdot),
\]

\[
y_2(t) = F(t).
\]
To design the control, first the disturbance term $\Gamma^{\prime}(\cdot)$ is estimated using the observer (9), which takes the form
\[
\dot{\hat{y}} = -a_F(t) + b_iV_i(t) + c_iF(t) - \dot{\hat{F}}(t) + \omega_F(t),
\]
(12)
\[
\dot{\omega}_F(t) = c_{iF}F(t) - \dot{\hat{F}}(t) + c_i\text{sign}(F(t) - \dot{\hat{F}}(t)),
\]
\[
\hat{y}_2 = \dot{\hat{F}}.
\]

Once the disturbance is estimated, the controller for flow regulation is proposed. Let a constant flow reference $F_{\text{ref}}$, define the error $e_F(t) = F(t) - F_{\text{ref}}$, whose dynamics are given by
\[
\dot{e}_F(t) = -a_Fe_F(t) - a_{iF}F_{\text{ref}} + b_iV_i(t) + \Gamma^{\prime}(\cdot)
\]
(13)
based on the active disturbance rejection control structure we propose the control signal
\[
V_i(t) = \frac{1}{b_i} \left(-k_e e_F(t) + a_{iF}F_{\text{ref}} - \omega_F(t)\right),
\]
(14)
by substituting (14) in system (13) we get
\[
\dot{e}_F(t) = -(a_i + k_e) e_F(t) - \omega_F(t) + \Gamma^{\prime}(\cdot),
\]
where the term $-\omega_F(t) + \Gamma^{\prime}(\cdot)$ vanishes asymptotically, so making a suitable selection of the constant $k_i$ the convergence to the origin of the error $e_F(t)$ is guaranteed.

6. Pressure control design

In this control objective, there is a significant restriction in the dynamics of the system. The way to increase the pressure is through the injection of gas using the proportional flow valve, while the decrease in pressure can only be achieved by releasing gas through the on/off valve and suspending the injection of gas. In this sense, the dynamics of the pressure increase is through a controllable system, system (6), but the dynamics of the pressure decrease is not controllable, system (7); the combination of the operation of both dynamics should result in a system with a stable equilibrium point.

Let $P_{\text{ref}}$ be a constant reference for the pressure level and let $e_p(t) = P(t) - P_{\text{ref}}$ be the error between the actual pressure and the reference. The strategy to be implemented is as follows: if $e_p(t) < 0$, then a control $V_i(t)$ is designed that guarantees the convergence of the error $e_p(t)$ to zero asymptotically. However, if $e_p(t) > 0$, then the output $V_i(t)$ will be set to zero to stop the injection of air and will activate the $V_i(t)$ output to release air and decrease pressure. The combination of both structures must ensure the convergence of the pressure $P(t)$ to the reference $P_{\text{ref}}$.

To increase the pressure, $e_p(t) < 0$, the invasive mechanical ventilator model is given by (6). First, based on (9), a state observer is implemented to estimate the disturbance $\gamma_{\text{pat}}(t)$, which takes the form
\[
\dot{\hat{y}} = -a_F\hat{y} + \frac{1}{C_{\text{lung}}}F(t) + c_{iP}(y_1(t) - \hat{y}(t)) + \omega_F(t),\]
(15)
\[
\dot{\omega}_P(t) = c_{iP}(y_1(t) - \hat{y}(t)) + c_{iP}\text{sign}(y_1(t) - \hat{y}(t)),
\]
\[
\hat{y}_2 = \dot{\hat{F}}.
\]
where $\omega_F(t)$ is the estimation of $\gamma_{\text{pat}}(t)$.

The design of $V_i(t)$ is as follows. The dynamics of the error $e_F(t)$ is given by
\[
\dot{e}_F(t) = -a_F e_F(t) - a_{iF}P_{\text{ref}} + \frac{1}{C_{\text{lung}}}F(t) + \gamma_{\text{pat}}(t)
\]
(16)
\[
\dot{F}(t) = -a_F F(t) + b_iV_i(t) + \Gamma^{\prime}(\cdot),
\]
\[
y_1(t) = P(t),
\]
\[
y_2(t) = F(t),
\]
Due to the structure of this system, the control of $e_F(t)$ can only be done through the flow $F(t)$. To guarantee the convergence of the error $e_F(t)$ to zero, the flow must be
\[
F(t) = C_{\text{lung}}(a_Rp_i - \gamma_{\text{pat}}(t) - k_p e_F(t)),
\]
(17)
by substituting (17) in (16), we get
\[
\dot{e}_F(t) = -\left(a + k_p\right) e_F(t)
\]
(18)
where its stability can easily be guaranteed with a gain $k_p > -\alpha$.

Now, to satisfy (17) a tracking controller is implemented for the flow, where the reference signal $R_{\text{ref}}(t)$ is given by
\[
R_{\text{ref}}(t) = C_{\text{lung}}(a_Rp_i - \omega_F(t) - k_p e_F(t)),
\]
(19)
Now we define the error variable $\epsilon(t) = F(t) - R_{\text{ref}}(t)$, whose dynamics are given by
\[
\dot{\epsilon}(t) = -a_F \epsilon(t) - a_F R_{\text{ref}}(t) + b_iV_i(t) + \Gamma^{\prime}(\cdot) - \dot{\hat{F}}(t),
\]
(20)
a control signal that stabilizes the origin $\epsilon(t) = 0$ is
\[
V_i(t) = \frac{1}{b_i} \left(a_F R_{\text{ref}}(t) - \Gamma^{\prime}(\cdot) + \dot{\hat{F}}(t) - k_F \epsilon(t)\right),
\]
(21)
substituting (21) in (20) gives
\[
\dot{\epsilon}(t) = -\left(a + k_F\right) \epsilon(t),
\]
where
\[
\lim_{t\to\infty} \omega_F(t) = -\dot{\hat{F}}(t)
\]
then the control signal that is implemented is
\[
V(t) = \frac{1}{b_i} \left(a_F R_{\text{ref}}(t) - \omega_F(t) - \omega_F(t) - k_F \epsilon(t)\right),
\]
(23)
by substituting the control signal (23) in the system (20), we have
\[
\dot{\epsilon}(t) = -\left(a_i + k_F\right) \epsilon(t) - \Gamma^{\prime}(\cdot) - \dot{\hat{F}}(t) - \omega_F(t) - \omega_F(t),
\]
(24)
because the term $\Gamma^{\prime}(\cdot) - \dot{\hat{F}}(t) - \omega_F(t) - \omega_F(t)$ vanishes asymptotically, a value of the gain $k_F$ can be chosen such that the origin $\epsilon(t) = 0$, is an asymptotically stable equilibrium point.

Now we must guarantee that despite the presence of communications between pressure increase and decrease, it converges to the reference pressure. For this purpose, it is crucial to force that the flow $F(t)$ remains limited when the pressure increases and that it tends to zero when the pressure decreases. Therefore, only the pressure dynamics are considered, which will be analyzed in terms of the pressure error $e_p(t)$.

The model that represents the dynamics of the error $e_p(t)$ is a first-order system with variable structure given by
\[
\dot{e}_p(t) = -\left(\alpha + k_p\right) e_p(t), \text{ for } e_p(t) < 0
\]
(25)
\[
\dot{e}_p(t) = -\alpha e_p(t) - \alpha P_{\text{ref}} + \frac{1}{C_{\text{lung}}}F(t) + \gamma_{\text{pat}}(t), \text{ for } e_p(t) > 0
\]
(26)
where there is a discontinuity surface at $e_p (t) = 0$. The behavior of the trajectories in a neighborhood of this surface is analyzed. When the trajectories tend to the discontinuity surface by the right, we have

$$
\lim_{e_p \to 0^+} \dot{e}_p (t) = -\alpha P_{ref} + \frac{1}{C_{lung}} F (t) + \gamma_{pat} (t),
$$
in this situation gas is released. Therefore the flow $F (t)$ has a negative value, and then

$$
\lim_{e_p \to 0^+} \dot{e}_p (t) < -\alpha P_{ref} + \frac{1}{C_{lung}} F (t) + |\gamma_{pat} (t)|,
$$

if $|\gamma_{pat} (t)| < \alpha P_{ref} + \frac{1}{C_{lung}} F (t)$,

then

$$
\lim_{e_p \to 0^+} \dot{e}_p (t) < 0,
$$

which implies that the trajectories cross the discontinuity surface.

When the trajectories tend to the discontinuity surface by the left, we have

$$
\lim_{e_p \to 0^-} \dot{e}_p (t) = 0,
$$

which implies that it arrives in asymptotic form.

Based on the limits (27) and (28) it is shown that the error $e_p (t)$ converges to the origin as follows. If $e_p (t) < 0$ then there is an asymptotic convergence to zero, there are no overshoots because it is a first order system. If we have an initial condition $e_p (t_0) > 0$ or if due to some disturbance we have the condition $e_p (t) > 0$, then the trajectories cross the discontinuity surface and subsequently the error converges to zero in asymptotic form.

7. Global invasive mechanical ventilator control strategy

Fig. 5 gives a block diagram that shows the overall strategy of the mechanical ventilator operation control, which is described below.

Based on the operating parameters of the system, established by the user, the reference signal for the flow $F_{ref} (t)$ is generated to control the volume indirectly, and the reference signal $P_{ref} (t)$ is used for pressure control. For this purpose, the respiration period $T$ is calculated as the inverse of the frequency $F$. Based on the $I:E$ ratio, the inspiration $T_i$ and expiration $T_e$ times are calculated. Using a real-time clock from the Microlabbox platform, a time variable modulus $T$ is generated, which is defined as $t_m$. To generate the reference signal for the volume control, the amplitude $A_F$ of the flow is calculated such that the volume $V_i$ is reached during the period $T_i$, Finally, the reference signals are generated simultaneously, which corresponds to block (a) of Fig. 5, which are square signals that are defined as

$$
F_{ref} (t) = \begin{cases} 
A_F, & 0 \leq t_n < T_i, \\
0, & T_i \leq t_n < T_F, \\
\end{cases}
$$

$$
P_{ref} (t) = \begin{cases} 
\text{PIP}, & 0 \leq t_n < T_i, \\
\text{PEEP}, & T_i \leq t_n < T_F. \\
\end{cases}
$$

At the same time, the reading of the output voltages of the pressure and flow sensors, as well as their conditioning, are carried out in block (b). The volume calculation is carried out in block (c), where the flow $F (t)$ is integrated, and the integration is restarted in each cycle of respiration through the “Reset” terminal, which is activated by the positive edge of the flow reference signal.

The state observers (12), (15) and (22), which estimate the disturbances and signals necessary to implement the controllers, correspond to blocks (d), (e) and (f).

The pressure controller, Eq. (23), corresponds to block g). This controller generates two outputs that control the valves in the system as follows. If the pressure $P_{ref} (t) > P (t)$, which implies that the pressure needs to be increased, then $V_{L} = V_i (t)$ and $V_P = \text{Off}$. Otherwise, $V_{L} = 0$ and $V_{P} = \text{On}$; that is, gas is released to lower the pressure.

The flow controller, Eq. (14), corresponds to block (h). Like the pressure controller, this controller generates the control signals for the two valves. If the breathing process is in the period of inspiration, that is to say $F_{ref} (t) > 0$ and the pressure in the lungs is below the limit pressure; $P (t) \leq P_{lim}$, then $V_{L} = V_i (t)$ and $V_{P} = \text{Off}$, otherwise $V_{L} = 0$ and $V_{P} = \text{On}$. This ensures that if the pressure in the lungs exceeds the limit pressure, then gas is released to decrease pressure and thus prevent patient harm.

All of the blocks that have been previously described operate at the same time. However, the application of the signals from the controllers to the valves is governed by the logic established in block (i), called the “Logic stage”. The commutation of the controllers depends on the variables “On/Off” and “Mode” and their operation logic is presented in Fig. 6. If the ventilator is deactivated, $\text{On/Off} = 0$, then $V_i (t) = 0$ and $V_{P} (t) = \text{On}$, which implies that no gas is introduced to the patient and it can freely leave the lungs. Meanwhile, when $\text{On/Off} = 1$, the operation depends on the value of the “Mode” variable. If $\text{Mode} = 0$, then the ventilator operates by volume control; and if $\text{Mode} = 1$, then the ventilator operates in pressure control mode.
In the volume control mode, the state of the variable $s_{sw}$ is checked. If it is in the inspiration period, $s_{sw} = 1$, then $V_{c}(t) = \check{V}_F$ and $V_{e}(t) = \check{V}_E$. It is important to mention that in the flow control block, block $h$), it is previously ensured that the pressure in the lungs does not exceed the limit pressure $P_{lim}$. If it is in the expiration period, $s_{sw} = 0$, then the control outputs are $V_{c}(t) = \check{V}_P$ and $V_{e}(t) = \check{V}_E$. In this way, the PEEP pressure is maintained in this time interval.

Finally, the control signals $V_{c}(t)$ and $V_{e}(t)$ are applied to the respective valves through block (j). It is important to note that the performance of the closed loop system does not depend on the initial conditions in both control operation modes.

### 8. Results of numerical simulations of the closed-loop system

This section presents the numerical results of the simulation of the proposed strategy for the control of the mechanical ventilator. These simulations correspond to a frequency $F = 10$ c/min and with the relation $I : E = 2$. We use the fixed step Euler solver with 0.001 s of step time. The parameters of the observers and controllers are $c_{IF} = 100$, $c_{2F} = 30$, $c_{3F} = 100$, $c_{IF} = 50$, $c_{2F} = 5$, $c_{3F} = 25$, $c_{t_{ini}} = 200$, $c_{2E} = 50$, $c_{3E} = 100$, $k_{ref} = 10$, $k_{ref} = 3$ and $k_{h} = 3$.

Fig. 7 shows the performance of the mechanical ventilator in the volume control mode of operation. The upper graph shows the behavior of the flow, red line, and the flow reference signal, black line. Meanwhile, the lower graph shows the behavior of the pressure $P(t)$, a black line, and the PEEP pressure, green dotted line. It can be seen that in the inspiratory period, the flow $F(t)$ converges to the reference signal $F_{ref}(t)$; while in the expiration period the pressure $P(t)$ converges to the base pressure PEEP. The control signals $V_{c}(t)$ and $V_{e}(t)$ are shown in Fig. 8. Here it can be seen that the control signal $V_{c}(t)$ does not reach the value of $1.5V$, which allows us to predict that the proportional valve will not saturate in the experiments. It is interesting to observe the switching of the signal $V_{e}(t)$ that allows the pressure to be released in the period of expiration because it is important to see that it does not have many switches for pressure regulation.

The behavior of the state observers that estimate the disturbances and the signals necessary to implement the control signals is shown in Fig. 9. Here, the state variables, black lines, and the estimated states, dotted red lines, are shown in the graphs in the left-hand column. Meanwhile, the estimated disturbances and auxiliary signals are shown in the graphs in the right-hand column. It is important to note that in all cases, the error between the real and estimated states is minimal. Consequently, the estimate of disturbances is considered reliable.

The results of numerical simulations of the mechanical ventilator in the pressure control mode are shown below. In Fig. 10, in the upper graph, the behavior of the flow $F(t)$ is presented. In the lower graph, the behavior of the pressure $P(t)$, red line, the reference signal of pressure $P_{ref}(t)$, black line, and the base pressure PEEP, dotted green line, are presented. Here, it can be seen that the pressure adequately converges to the reference signal without overshooting in the inspiration period and with small fading oscillations in the expiration period.

The control signals $V_{c}(t)$ and $V_{e}(t)$, as well as state observers, have the same performance than in volume control mode.
9. Experimental performance of the invasive mechanical ventilator

This section presents the experimental results of the implementation of the proposed strategy for the control of the invasive mechanical ventilator. These experiments correspond to a frequency $F = 10$ c/min and with the relation $I : E = 2$. As in the numerical simulations, in the experiments we use the fixed step Euler solver with 0.001 s of step time. The parameters of the observers and controllers are $c_{1F} = 90, c_{2F} = 30, c_{3F} = 50, c_{1P} = 150, c_{2P} = 50, c_{3P} = 100, c_{1e} = 200, c_{2e} = 50, c_{3e} = 100, k_F = 6, k_{pi} = 2.3$ and $k_I = 10$.

It is important to mention that, as in the simulation, to avoid abrupt changes in flow and pressure, the reference signals are smoothed to avoid damage to the patient. Fig. 11 shows the performance of the mechanical ventilator in the volume control mode of operation. The upper graph shows the behavior of the flow, red line, and the flow reference signal, black line, while the lower graph shows the behavior of the pressure $P(t)$, black line, the base pressure $PEEP$, dotted line in green, and the limit pressure $P_{lim}$, dotted line in blue. During the first two cycles, the level of the reference signal causes the pressure $P(t)$ to reach a pressure close to $P_{lim}$. However, because the flow does not show overshoots, the on/off valve is not activated to release pressure. The subsequent two cycles decrease the amplitude of the flow reference signal and, as can be seen, the convergence of the flow to the reference signal has minimal errors.

In the next two cycles, the value of $P_{lim}$ is decreased in such a way that the safety condition is met and the controller blocks the gas supply and releases the pressure so as not to exceed the $P_{lim}$ level. In the next cycle, it is returned to the previous $P_{lim}$ level and the mechanical ventilator returns to normal operation. In the final two cycles, a change is made in the base $PEEP$ pressure. It can be seen that the pressure level in the expiration period is adjusted correctly.

The behavior of the control signals $V_i(t)$ and $V_e(t)$, as well as the state observers, are qualitatively similar to the obtained in numerical simulations.

Finally, an additional experiment is carried out where external disturbances are applied to the system, which consists of applying, in a random and manual way, pressure disturbances to the lung. The results are shown in Fig. 12, where it can be seen that the performance of the flow control does not present any perceptible change. Although there are considerable changes in the pressure level in the inspiration period, in the expiration period the pressure continues to converge to the $PEEP$ pressure level.

The experimental results in the pressure control mode of operation are described below. In Fig. 13, in the upper graph, the behavior of the flow $F(t)$ is presented. Meanwhile, in the
10. Experimental performance of the invasive mechanical ventilator using PID controllers

This section presents the performance of the invasive mechanical ventilator using PID-type controllers to draw a comparison with those obtained with the ADRC control structure. To make this comparison, the same control switching strategy described in Section 7, shown in Fig. 5, is used, while eliminating the flow and pressure controllers based on the ADRC structure, blocks (h) and (g), as well as state observers correspond to (d), (e) and (f) blocks.

A block diagram of the global control strategy using PID controllers is given in Fig. 15. Here the PID controllers are placed; for pressure control, block (2) and for flow control, block (3), and a logical structure that restarts the integrators at the moment of switching between flow control and pressure control, block (1). The gains of each of the controllers are, for pressure controller $k_p = 0.6$, $k_i = 0$ and $k_d = 0.05$, and for flow controller $k_p = 0.015$, $k_i = 0.14$ and $k_d = 0$.

Detailed comparative performance analysis of ADRC control structure and PID control in the volume control mode, where flow and pressure controllers operate, is in Fig. 16. Here, flow controllers are active in the first period and pressure controllers in the second. For flow control, we can see that the PID controller cannot compensate for the effect of the dead zone in the proportional flow valve. Consequently, there are time-lapses where the flow is zero. The same thing happens in the pressure control operation mode; there is an error between the pressure $P(t)$ and the baseline pressure PEEP. These results show that the ADRC control structure, Fig. 16(a), performs better than the PID control, Fig. 16(b), for flow and pressure control.

11. Conclusions

This paper has demonstrated, in an analytical, numerical, and experimental way, that the control structure with active compensation of disturbances can be used successfully in the implementation of invasive mechanical ventilators, which are strongly non-linear systems, with delays, disturbances, and parametric uncertainties. In addition, a simple pneumatic circuit that is only composed of a service unit, a proportional valve for flow control, an on/off valve, two flow sensors, a pressure sensor, and the controller is proposed as a viable and economical option to build an invasive mechanical ventilator for critically ill COVID-19 patients.

For this option to be massively implemented, it is necessary to replace the dSPACE Microlabbox platform with a compact and economic control platform. However, making this replacement is not an easy task because the operation of the state observers depend on a real-time execution and a sampling time of 1 ms maximum. Consequently, platforms such as Raspberry Pi have to be discarded. However, it has been shown in [16] that the ADRC control structure can be implemented in analog circuits. Therefore, the controllers and observers could be implemented in analog circuits. Meanwhile, the blocks to generate the reference signals, capture the flow and pressure signals, as well as the logic block that switches the pressure and flow controllers, can
be implemented in a compact platform such as the Raspberry Pi or myRIO from National Instruments. These platforms will also allow a user-friendly interface. They can also implement all of the security measures and alarms that the different international standards established for this type of equipment enable.

Declaration of competing interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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