Pressure control using sliding modes for the expiratory cycle in mechanical ventilation systems

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***Abstract*—Mechanical ventilation (MV) is crucial in the recovery of patients with severe respiratory failure. The expiratory phase, in particular, presents specific dynamic complexities. In this context, control systems represent fundamental components for the adequate performance of MV systems. However, traditional control approaches employed in the literature have limitations in addressing these complexities and ensuring accurate performance. This article addresses the challenges associated with controlling the expiratory stage in MV systems through a robust and effective solution based on sliding mode control. In particular, the proposed control approach in this article is based on a non-linear model recently validated in the literature. Thus, the objective is to overcome the limitations of traditional approaches and provide a robust and efficient control solution that overcomes the solutions available in the literature. The results show the potential of this control strategy in real clinical settings. The sliding mode control approach improves the performance and convergence of MV systems, contributing to more effective ventilation therapy and better patient recovery.**

***Index Terms*—Mechanical Ventilation, Expiratory Circuit, Sliding Mode Control, Pressure Control.**

I. INTRODUCTION

Mechanical ventilation (MV) plays a crucial role in providing life support to patients with severe respiratory failure. For decades, MV systems have been extensively used in intensive care units to assist or control the respiratory cycle of patients [1]. Particularly, as an illustrative case, during the recent outbreak of SARS-CoV-2 (coronavirus pandemic), special attention has been given to MV technologies due to their fundamental role in the treatment and recovery of COVID-19 affected patients [2].

Respiration, and particularly MV, can be divided into two phases [1]: inspiration and expiration (exhalation). During inspiration, a gas flow is delivered to the patient, and pressures in the circuit and lung are increased by regulating the corresponding valves. During expiration, pressures decrease due to the action of the exhalation valve, reaching an appropriate final pressure [3], [4]. Specifically, the final pressure during the expiratory cycle is referred to as positive end-expiratory pressure (PEEP). The involved dynamic systems and approaches for modelling and control play a crucial role in the correct execution of MV, thus contributing to successful patient recovery.

Achieving an adequate performance in MV systems presents challenges from the perspective of dynamic systems. The parametric dynamic range of the system can span *over four orders of magnitude* [5], which represents a serious challenge [3]. Furthermore, the nonlinear interaction between the patient and the circuit during the expiration stage, commanded through valves based on a force balance scheme [6], [7], introduces an additional level of complexity. Thus, precise control is required to fulfil performance requirements in the time domain, such as achieving a rapid pressure drop without exceeding the PEEP level (avoiding *overshoots*) and maintaining expiratory flow within established limits with no oscillations. Controlling the expiration stage is the most complex challenge in MV systems [5]. To address these difficulties, the use of control technologies is essential.

Regarding existing control approaches, references can be found in the literature, for example, in [4], [8], or in [9]. However, despite being typical cases in the literature, they present serious restrictions, as discussed in [10]. In the study presented in [11], the design of an MV system based on a flow supply system through a turbomachine [5] is proposed, where a control methodology for the inspiration and expiration stages is also presented. However, it is important to note that the model used in [11] for control design is derived under linearity assumptions, while considering only a reduced set of system parameters, only covering one patient for subsequent results. A more recent control proposal for the expiratory system can be found in [10], where a dynamic model of the entire expiratory cycle (including patient, circuit, and valve based on force balance) is derived. However, it is worth noting that the control system proposed in it is not developed with analytical rigour, which prevents providing performance and, fundamentally, stability guarantees.

In virtue of the aforementioned challenges, sliding mode control stands out as a promising strategy to address the control issues in the expiration stage of MV systems [12], [13]. Sliding mode control offers an effective solution to deal with highly nonlinear systems, being robust, and capable of dealing with the wide parametric variation in the expiratory system.

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Thus, the ability of sliding mode controllers to achieve precise control tracking as well as their immunity to disturbances and parametric variations make them particularly suitable for addressing the dynamic complexity of MV systems. It is worth mentioning that recent studies have applied sliding mode-based methodologies to MV systems [14], [15]. In these studies, a sliding mode approach is combined in different ways with fuzzy logic control, such as using adaptive methods [14] or fractional order sliding mode [15], resulting in somewhat more complex-to-implement strategies and with high computational resources consumption [15]. Additionally, the presented results of those studies are simulation-based and, both simulations and control designs, rely on theoretical models of the MV system, which are linear and have not been experimentally validated. In one of the cases, the proposal was simulated under specific conditions without considering parametric variation ranges.

In this context, this paper addresses the design and implementation of a sliding mode controller for pressure regulation during the expiration stage of MV systems. The objective of this work is to overcome the limitations of traditional approaches and provide a robust and effective control solution for expiration cycle control in MV. Specifically, a dynamic model of a real MV expiration subsystem is considered, which has been experimentally validated in [10]. Such model includes the dynamics of the expiration control valve used by many of the MV devices and, with it, it is nonlinear. This paper tackles the challenge of precise and effective control during the expiration stage of MV systems using nonlinear control techniques with a solid theoretical foundation. The sliding mode control approach aims to provide a robust and efficient solution to enhance the performance and stability of MV systems, by means of a relatively simple control law. The obtained results demonstrate the potential of this control strategy and its applicability in real clinical settings.

The remainder of this article is organised as follows. Section II presents the dynamic models used for the expiration phase in MV systems. Section III introduces the theoretical framework for sliding mode control approaches. Section IV presents a case study applying the proposed sliding mode control strategy, considering a wide range of realistic scenarios. Finally, Section V concludes the paper, highlighting the main findings and describing future research directions.

II. Modelling framework

The theoretical framework of the mathematical model that describes the expiratory circuit is introduced here. It is important to note that this section is essentially taken from [10]. Interested readers are referred to it for a detailed discussion on the derivation of the model considered for this study.

Figure 1 illustrates the pneumatic model of a general MV system. The scheme presented in Fig. 1 is developed based on flow resistances (R) and compliances (C). On the one hand, a standard relationship of volumetric flow through an effective area A is established, assuming isotropic flow and a relatively low-pressure difference between upstream and downstream

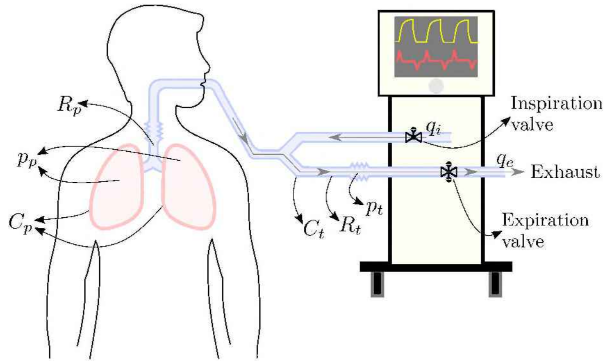

Fig. 1. Scheme of a general mechanical ventilation system. The main components used in the model presented in Section II, such as compliances and resistances (C and R), are indicated with subscripts t and p as appropriate for the circuit and tubing or the patient, respectively. Additionally, the inspiratory and expiratory flows (q_i and q_e) are shown.

locations. As shown in [16], this relation can be expressed as:

$$q = \frac{1}{R}\sqrt{|\Delta p|}\ \mathrm{sign}(\Delta p), \tag{1}$$

where $\mathrm{sign}(\cdot)$ represents the sign operator used to determine the flow direction. The flow is expressed in L/s (litres per second), Δp is the pressure difference measured in cmH_2O, and R corresponds to the effective flow resistance [5]. It is essential to note that the effective area is not actually given by the geometric area, due to higher-order and even nonlinear flow dynamics. On the other hand, the relationship between volume and pressure in a pneumatic system can be described using compliance. According to the analysis in [5], the change in volume V concerning the change in pressure p is approximated as $\frac{\partial V}{\partial p} = C$, where C represents the compliance of the system. Although the compliances of the lung (patient) and circuit are nonlinear in reality, this assumption is widely used for modelling and control design in MV systems [5], [7]. Using compliance, the resulting pressure in a compartment with fixed volume V, affected by a flow $q(t)$, can be calculated as follows:

$$p(t) = \frac{1}{C}\int_0^t q(\tau)d\tau. \tag{2}$$

The exhalation valve in the general MV system is based on a force balance scheme, as shown in Fig. 2, where the main components and variables used for modelling are indicated. The references for the components are given in Table I. In this type of valves, the position is determined by the balance between the command force f_v, controlled by the input control current i_v, and the force of the circuit f_c, generated by the circuit pressure at the valve seat (A_s). It is worth noting that these valves do not have a restoring mechanism (spring). To avoid the restoring effect due to the compression of air, a relief chamber is used, as shown in Fig. 2. During operation, the gas flow from the patient pushes the diaphragm with a force f_c towards the outlet, while the head counteracts with a controlled opposing force f_v through the action of the coil assembly. Thus, a force balance valve for exhalation, as illustrated in

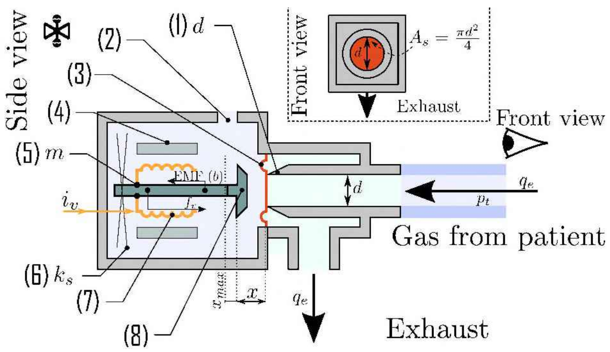

Fig. 2. Force balance type valve for the expiratory circuit.

TABLE I
REFERENCES FOR FIG. 2. THE VARIABLES ASSOCIATED WITH EACH PHYSICAL COMPONENT CONSIDERED IN THE MODEL ARE INDICATED.

Reference	Description	Variable
1	Seat	d
2	Relief chamber	
3	Diaphragm	
4	Magnet	
5	Coil assembly	m
6	Spring	k_s
7	Coil	
8	Poppet	

Fig. 2, can be modelled using a standard approach based on Newton's second law as follows:

$$m\ddot{x} = b\dot{x} + k_s x + f_c - f_v, \tag{3}$$

where x, $\dot{x}$, and $\ddot{x}$ denote the position, velocity, and acceleration of the valve poppet, respectively, b is the energy dissipation coefficient, defined by the reverse electromotive force (EMF), $k_s \approx 0$ is the restitution coefficient, and m is the mass of the coil assembly (see Fig. 2). Additionally, $f_c = A_s p_t$ represents the force due to the pressure of the circuit, while $f_v = k_v i_v$ denotes the control force, with k_v being the force constant provided by the valve.

Using the fundamental principle of mass conservation, along with the expressions in Eqs. (1), (2), and (3), the complete exhalation model can be defined as the nonlinear mapping:

$$\begin{cases} \dot{p}_t = \dfrac{\mathcal{Q}(p_p, p_t)}{C_t(R_p + R_t)} - \dfrac{x_1 A_v \mathcal{Q}(p_t, 0)}{C_t} + \dfrac{R_p}{C_t(R_p + R_t)} q_i, \\ \dot{p}_p = -\dfrac{\mathcal{Q}(p_p, p_t)}{C_p(R_p + R_t)} + \dfrac{R_t}{C_p(R_p + R_t)} q_i, \\ \dot{x}_1 = x_2, \\ \dot{x}_2 = \dfrac{1}{m}\left(k_s x_1 + b x_2 + p_t A_s - k_v i_v\right), \end{cases} \tag{4}$$

where $\mathcal{Q}(p_1, p_2) = \sqrt{|p_1 - p_2|}\,\mathrm{sign}(p_1 - p_2)$, with p_1 and p_2 pressure values. The variables x and $\dot{x}$ in (3), have been replaced by x_1 and x_2, respectively. C_t and C_p denote the compliances related to the circuit and the patient, respectively, and R_t and R_p indicate the flow resistances related to the circuit and the upper airways of the patient, respectively. Additionally, the constant quantity q_i in (4) represents the passing flow, which is commanded by the MV system and used for multiple purposes, such as respiratory reflex detection. In (4), the circuit force and valve control forces have been replaced by their definitions, $f_c = p_t A_s$ and $f_v = k_v i_v$, respectively. Similarly, in (4), $A_v = \pi d c_d$, where d is the diameter of the valve diaphragm (see Fig. 2), and c_d is the coefficient needed to compensate for units and for the geometric and effective areas of the valve [10]. Finally, the expiratory flow (system output) can be expressed as $q_e = x_1 A_v \mathcal{Q}(p_t, 0)$, which is evident from the definition of $\dot{p}_t$ in (4).

III. CONTROL SYSTEM

A. *Reference Trajectory*

As indicated by general medical therapies [10], during the expiratory cycle, p_t must be brought from the *plateau* pressure to the PEEP value in a rapid trajectory. The control specifications can be listed as follows:

- The circuit pressure must decrease to the PEEP value as quickly as possible to preserve the clinical condition. This allows for respecting the natural rate of the patient of free exhalation, avoiding muscle fatigue or carbon dioxide accumulation.
- Overshoot or oscillations in the flow or pressure must be avoided. That is, preventing p_t and q_e from falling below the PEEP level or flow q_i, respectively.
- The system must operate effectively over the entire range of PEEP values and potential patients.

In the steady state of expiration, the following relationships hold:

$$\begin{aligned} p_t^* &= P_{PEEP}, \\ q_e^* &= q_i, \\ i_v^* &= \frac{A_s}{k_v}\left[P_{PEEP} + (R_p q_i)^2\right] \approx \frac{A_s}{k_v} P_{PEEP} \end{aligned} \tag{5}$$

Considering the aforementioned specifications, a pressure reference trajectory (P_{ref}) is designed, consisting of three segments [10]. The first segment is linear, designed with a rapid expiration rate, greater than that of free exhalation, so that the control reaches saturation and, thus, expiration can initially progress without interference with the valve open. The second segment is defined in terms of a quadratic function to provide a smooth transition between the first and third segments. The third segment is defined as the desired constant PEEP value. Thus, the linear, quadratic, and constant segments, denoted as $l_1(t)$, $l_2(t)$, and $l_3(t)$, respectively, are expressed as follows:

$$\begin{aligned} l_1(t) &= m_r t + P_{lt}, \quad 0 \le t < t_{63} \\ l_2(t) &= k_r(t - t_{PEEP})^2 + P_{PEEP}, \quad t_{63} \le t < t_{PEEP} \\ l_3(t) &= P_{PEEP}, \quad t \ge t_{PEEP} \end{aligned} \tag{6}$$

with

$$\begin{aligned} P_{63} &= P_{lt} - 0.63(P_{lt} - P_{PEEP}); \quad & m_r &= \tfrac{P_{63} - P_{lt}}{t_{63}}; \\ t_{PEEP} &= t_{63} - \tfrac{2(P_{63} - P_{PEEP})}{m_r}; \quad & k_r &= \tfrac{m_r}{2(P_{63} - P_{PEEP})} \end{aligned} \tag{7}$$

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where P_{lt} represents the plateau pressure, P_{63} is 63% of the difference between P_{lt} and P_{PEEP}, and $t = 0$ indicates the start of the expiration cycle. Therefore, the reference profile is defined as:

$$P_{ref}(t) = \begin{cases} l_1(t) & \text{if } 0 \leq t < t_{63}, \\ l_2(t) & \text{if } t_{63} \leq t < t_{PEEP}, \\ l_3(t) & \text{if } t \geq t_{PEEP} \end{cases} \tag{8}$$

Using the reference profile in (8), the error signal is gradually regulated to avoid undesirable transient errors that could negatively impact the performance of the system.

B. Sliding mode control

The classical sliding mode theory, for the control of systems, was established in the 1980s [12]. Generally speaking, classical sliding control theory involves a discontinuous control action that ensures the convergence of the system states to a sliding surface (or sliding manifold), which is defined in the state space. Furthermore, it robustly maintains the system on the sliding surface, making sliding mode control systems theoretically insensitive to certain perturbations and modelling uncertainties. Thus, the sliding surface is determined through a sliding variable σ, which depends on the states and exhibits a relative degree of 1 with respect to the control. In turn, the sliding surface is designed in such a way that the control objective is achieved when $\sigma = 0$.

The main drawbacks and limitations of this initial approach, such as chattering (high-frequency oscillations due to discontinuous control) and the requirement of relative degree 1, have been addressed. As a result, several alternative solutions have emerged, one of which is high order sliding mode control [17], [18]. An r-order sliding mode algorithm guarantees theoretical convergence to 0, usually in finite time, for the first $r-1$ derivatives of the designed sliding variable. In other words, it drives the system states to the surface determined by $\sigma = \dot{\sigma} = ... = \sigma^{(r-1)} = 0$. This achievement is possible for sliding variables of relative degree r, by applying a control action that can be either continuous or discontinuous depending on the particular algorithm, but it acts discontinuously on the r-th temporal derivative of σ.

1) Sliding mode control in MV systems: For the system under study, in accordance with the approach adopted in [10], the following sliding variable is adopted:

$$\sigma = p_{\text{ref}}(t) - p_t(t) - k_q \Delta q(t), \tag{9}$$

where k_q is a design parameter and $\Delta q = q_i - q_e(t)$ is a flow compensation term. It was experimentally established in [10] that the inclusion of such a term in the control setup of the actual mechanical ventilator contributed to improving the performance, avoiding oscillations. Once in sliding mode, pressure p_t will track the designed pressure reference, which ultimately, at l_3, reaches the specified PEEP value. Furthermore, the expiratory flow matches the inspiratory flow.

As it can be verified, σ results of relative degree 2 with respect to the valve control action, given by current i_v (or the control force, $f_v = k_v i_v$). In this regard, this work proposes the use of the generalised third-order super-twisting algorithm (3-ST) [19], which applies a continuous control action with a finite-time convergence to the sliding surface under design conditions. The control law of the 3-ST algorithm is:

$$u_{3ST} = -k_1 |\phi|^{1/2} \operatorname{sign}(\phi) - k_3 \int_0^t \operatorname{sign}(\phi(\tau))\, d\tau, \tag{10}$$

$$\phi = \dot{\sigma} + k_2 |\sigma|^{2/3} \operatorname{sign}(\sigma), \tag{11}$$

where k_1, k_2, and k_3 are suitable positive gains [19], [20].

For this application, the applied control structure is proposed as two additive components. Firstly, the sliding mode control algorithm 3-ST and, secondly, for linear translation, the expected steady-state control current value, computed as in (5). Thus, the resulting control action is as follows:

$$i_v = i_{v,3ST} + i_v^* \tag{12}$$

The gains of the 3-ST component were set following [20], based on the nominal model (4) and extensive simulation tests. They considered wide ranges for the parameters and signals, to account for a variety of circuits, patients, and feasible PEEP pressures, relative to potential medical therapies.

IV. Simulations

A. Simulation scenarios

To analyse the feasibility of the proposed sliding mode control, simulation tests are performed by varying parameters over a wide range to cover different realistic scenarios. Additionally, discretisation is included to account for the A/D converter with a sampling frequency of 250Hz.

For the considered cases, a typical inspiratory (passing) flow and PEEP pressure values are considered [10]:

$$q_i = \frac{1}{15} \frac{L}{s} \qquad P_{PEEP} = 5\,\text{cmH}_2O. \tag{13}$$

A standard tubing circuit is assumed as well [10][1]:

$$C_t = 0.00025 \frac{L}{\text{cmH}_2O}, \qquad R_t = 1.04 \frac{\sqrt{\text{cmH}_2O}}{L/s}. \tag{14}$$

Regarding the characteristics of the patient, a wide variety of cases are considered, covering any potential adult case. The flow resistance for a general adult patient usually falls within the range of values R_p^5 to R_p^{20}:

$$R_p \in [R_p^5, R_p^{20}] = \left[\sqrt{2.7}, \sqrt{17.6}\right] \frac{\sqrt{\text{cmH}_2O}}{L/s}, \tag{15}$$

showing results using the extreme values. Additionally, three compliance values are considered, covering the cases of healthy adults and those with some pathology [21]:

$$C_p \in [0.02, 0.03, 0.06] \frac{L}{\text{cmH}_2O}. \tag{16}$$

Finally, the gains for the 3-ST algorithm were set as:

$$k_1 = k_3 = 1.97 \times 10^{-4}, \quad k_2 = 700. \tag{17}$$

[1]Even though the same framework considered in [10] is adopted for this study, a typographical error in the definition of resistance units exists in [10]. The square root term erroneously placed in the denominator is correct here.

B. Sliding mode control results

On the one hand, the outcomes obtained when simulating a patient with $C_p = 0.02\, L/cmH_2O$ and R_p^5 are presented. In particular, Fig. 3 shows the results for the pressure and expiratory flow signals. The passing flow provided by the ventilator, q_i, which begins 150ms after expiration initiates, is also shown with the latter. It must be noted that, even though

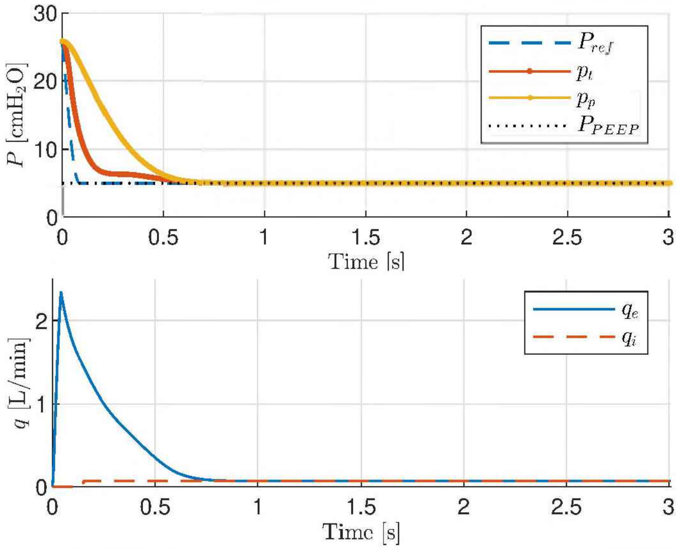

Fig. 3. Pressures (top) and flow (bottom) signals, for $C_p = 0.02$; R_p^5.

in realistic situations only the circuit pressure p_t is accessible, in simulation both circuit and lung pressures, as part of the state-vector, can be studied. Thus, both p_p and p_t, along with p_{ref}, are shown in Fig. 3. In a realistic scheme, lung pressure could be measured using sophisticated and highly invasive techniques.

As it can be appreciated, both the flow error and the pressure error $\Delta P = p_{\text{ref}} - p_t$ are zeroed in finite time, i.e. p_t reaches the designed pressure reference p_{ref}, and the expiratory flow equals the passing flow. The rest of the system variables converge to their steady-state values according to the reduced dynamics determined by the third-order sliding mode algorithm ($\sigma = \dot{\sigma} = \ddot{\sigma} = 0$), and to the stable patient dynamics. In this sense, the sliding variable $\sigma = \Delta P - k_q \Delta q$ is presented in Fig. 4, where its convergence to zero can be observed.

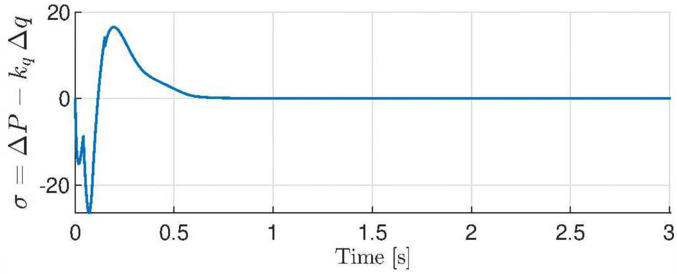

Fig. 4. Sliding variable, $\sigma = \Delta P - k_q, \Delta q$, for $C_p = 0.02$; R_p^5.

In Fig. 5, the control current of the valve i_v can be observed in its temporal evolution according to the proposed algorithm. As can be seen, it is continuous and relatively smooth, ultimately reaching the expected steady-state value.

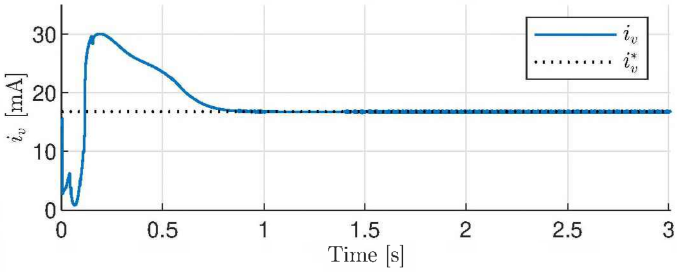

Fig. 5. Valve control current i_v. Patient: $C_p = 0.02$; R_p^5.

On the other hand, to exemplify the ability of the proposed controller to tackle the aimed objective, four results are presented in the following, corresponding to patients with compliance and resistance values at the extremes of the considered ranges. Figure 6 (top) shows the relative pressure error, while the expiratory flow signals obtained in the four cases can be seen at the bottom, jointly with the passing flow.

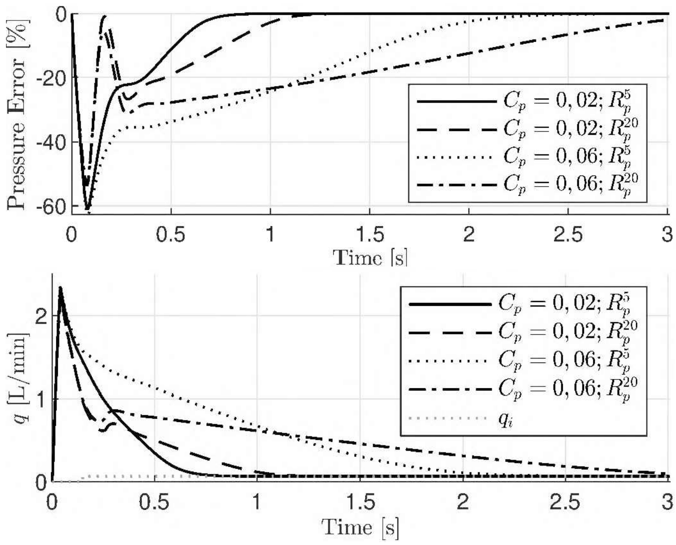

Fig. 6. Pressure errors (top) and flows (bottom) for 4 patients with $C_p = 0.02$ and $C_p = 0.06$, and with R_p^5 and R_p^{20}.

It can be observed that in all cases, the two variables of interest, p_t and q_e converge to their expected references (P_{ref} and q_i, respectively) presenting neither overshoots nor oscillations ('*overdamped*' response). These satisfactory responses were achieved by means of the adequate control currents i_v, displayed in Fig. 7. It can be seen that they are all time-continuous and relatively smooth. Note that the reaching time respects the natural response of free exhalation of each patient, as desired (see III-A).

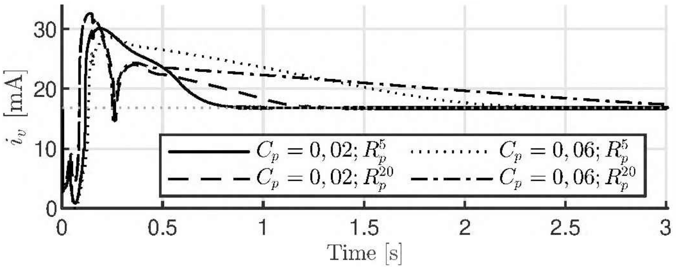

Fig. 7. Control currents i_v (bottom) for the 4 simulated patients, i.e., with $C_p = 0.02$ and $C_p = 0.06$, and with R_p^5 and R_p^{20}.

Additional conditions, including a wide range of patients, circuits, and PEEP values, have also been tested in simulation for this study. However, although these results strengthen those shown in this section, for the sake of brevity additional results are not included here and will be included in an extended version of this article, as detailed in Section V.

In general, the obtained results show an acceptable level of performance, in accordance with the pre-established design specifications (see the introductory part of Section III). The adequate convergence times and dynamics can be highlighted, relative to control specifications and medical requirements, as well as the 'smooth' variations of the actuation signal. It is important to observe that a 'smooth' manipulation of the actuator ensures an extended lifespan of the device (valve), in addition to contributing positively to the proper dynamic performance of the control system.

It must be pointed out that, as shown by the wide range of studied cases, the convergence time is achieved in an almost optimal manner, given the natural discharge time of the patient. Additionally, the system cannot be accelerated beyond its natural limits. Remarkably, the designed controller consistently ensures an *overdamped* response, with no oscillations and nearly optimal performance, with no overshoots. This reliable and precise control strategy guarantees adherence to all medical prescriptions, providing a safe and effective ventilation process for the patients. In addition, the action of the passing flow, a fundamental multipurpose additional input, does not affect the general response of the system.

V. Conclusion and Future Research

This work presents a control system for the expiratory circuit of MV systems, based on sliding mode techniques. The control strategy is designed using a nonlinear analytical approach, offering a robust reference framework for rigorous analysis, marking a significant difference relative to recent references in the literature. Notably, the strategy performs reliably across a wide range of parameter variations, encompassing resistances, compliances, and an extended pressure operating range. The dynamic performance overcomes many existing control strategies in the literature. Another remarkable feature is the smoothness of the control action, vital for the proper functioning of MV systems and the integrity and lifespan of the actuation system.

Future work proposes several research directions. Firstly, it aims to provide analytical proof of convergence and stability, as well as robustness. Additionally, the strategy will be compared with classical control approaches, such as PI and/or PID strategies, to highlight advantages and areas for improvement. In particular, the analysis will be extended to assess the applicability and adaptability of the control system in different clinical scenarios.

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