# Robust Flow Control in a Mechatronic Test Lung with Spontaneous Breathing

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*Abstract*—To systematically evaluate mechanical ventilators and their automation concepts, test lungs can be used in hardware-in-the-loop (HiL) simulations. However, most test lungs do not simulate lungs with spontaneous breathing. This study presents a mechatronic test lung that consists of a bellow and a voice coil actuator. After modeling and parameter estimation, the model was used for the synthesis of a robust controller. The whole system was validated with a reference lung model and recorded measurement data of volume flow. The model identification resulted in a root-mean-square error (RMSE) of 0.96 mm. The controller achieved a maximum RMSE of 0.062 liters per second and a relative compliance error of 5.88 %. The test lung has the potential to enable HiL simulations of mechanical ventilation with spontaneous breathing.

*Index Terms*—mechanical ventilation, hardware-in-the-loop, robust control, multivariable control, voice coil actuator

## I. INTRODUCTION

Mechanical ventilation is used in intensive care units to maintain gas exchange in patients with respiratory insufficiency. Approximately 33% of patients in intensive care units receive mechanical ventilation [1]. During the COVID-19 pandemic, the importance of mechanical ventilation was reported by the media worldwide. Mechanically ventilated patients are usually in the intensive care unit, where the workload is increasing due to the aging population, as predicted by Angus et al. [2]. To reduce the workload of clinical staff, automated mechanical ventilation systems are promising. Automated mechanical ventilation systems provide optimal gas exchange while preventing ventilator-induced lung injuries [3].

Mechanical ventilators and their automation systems need to be sufficiently validated prior to clinical use, often through animal experiments. Although animal testing has enabled advances in medicine, the use of animals is ethically controversial [4]. Based on the 3 Rs principle (*Replacement*,

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Test lungs simulate the mechanical characteristics of a lung. In contrast to test lungs that are passive, there are few test lungs that can simulate the patient's spontaneous breathing. Commercially available test lungs with this capability include the *ASL 5000*® *Breathing Simulator* (Ingmar Medical, Pittsburgh, USA) and the *Spontaneous Breathing Lung Simulator* (*SBL*<sup>TM</sup>) (Michigan Instruments, Grand Rapids, USA). While the commercial test lungs use bellows for volume storage, David et al. [6] proposed a blower-based test lung with spontaneous breathing. Mecklenburgh et al. [7] developed a test lung with a pneumatic piston.

However, none of the above test lungs have an interface for phrenic nerve (diaphragm's nerve) stimulation. Phrenic nerve stimulation is a support or an alternative to mechanical ventilation in which the phrenic nerve is stimulated by sending electrical pulses. These pulses cause the patient's diaphragm to contract, thus keeping it active. By keeping the diaphragm active, phrenic nerve stimulation can prevent diaphragm atrophy (muscle weakness) [8], which can occur during mechanical ventilation within hours of the initial start [9].

Current test lungs are not suitable for HiL simulation of mechanical ventilation and phrenic nerve stimulation. As a first step towards such a test lung, we propose a mechatronic test lung that can follow variable flow patterns, which builds upon our previous work [10]. The test lung is based on a bellow piston concept and simulates the passive characteristics of the lung and the diaphragm. The automation concept uses multivariable  $H_{\infty}$  control.



Fig. 1. Mechatronic test lung with VCA.



Fig. 2. Concept of the proposed test lung (adapted from [10]).

#### II. METHODS

### A. Hardware

The proposed HiL lung is shown in Fig. 1. A bellow (F-1218-NBR, Thodacon Werkzeugmaschinenschutz GmbH, Kolbermoor, Germany) simulates the lung volume capacity. Passive and active lung components are modeled via a voice coil actuator (VCA) (AVM 60-25-0.5, TDS Precision Products GmbH, Dielsdorf, Switzerland). A position sensor (T-0100, Novotechnik Messwertaufnehmer OHG, Ostfildern, Germany) was added to track the VCA position. Two pressure sensors (CODAN Xtrans pressure measurement system), one at the test lung entrance  $(p_{aw})$  and one at the metal plate  $(p_{bel})$ , and a flow sensor (SFM3300-D, Sensirion, Stäfa, Switzerland) were installed. An airway filter reduces the effect of the VCA motion on  $p_{aw}$ . Another airway filters is used as a standard connection to a mechanical ventilator (EVE IN, Fritz Stephan GmbH, Gackenbach, Germany).

The conceptual design of the test lung with automation is shown in Fig. 2. A real-time capable embedded computer



Fig. 3. Mechanical model of the test lung (adapted from [10]).

(Microlabbox, dSPACE, Paderborn, Germany) receives the measurements of the pressures  $p_{aw}$  and  $p_{bel}$ , the flow  $\dot{V}_{aw}$ , and the plate height h. The sensor data is processed, and then the pulse width modulation (PWM) signal of the VCA is set. The software used for programming was *MATLAB Simulink* (The MathWorks Inc., Natick, USA) and *ControlDesk* (dSPACE).

## B. Modeling

The assumed mechanical model is shown in Fig. 3. The mass m includes the weight of the plate, the VCA, and the bellow. k denotes the spring constant, and d the damping constant of the system. The VCA generates the force  $F_{vc}$ , the weight the gravitational force  $F_g$ , and the pressure induced by the ventilator the force  $F_p$ . The force equation of the system is:

$$F_p + F_{vc} = F_i + F_d + F_s + F_q \tag{1}$$

$$pA + F_{vc} = m\ddot{h} + d\dot{h} + kh + mg \tag{2}$$

 $A = 1.24 \times 10^{-2} \text{ m}^2$  is the cross-sectional area of the bellow.  $F_{vc}$  is obtained from the force constant  $K_{vc} = 17 \text{ N/A}$  and the electrical resistance  $R_{vc} = 5.37 \Omega$  from the manufacturer's specifications of the VCA and the selected operating voltage of  $V_{vc} = 24 \text{ V}$ . With the PWM signal  $u \in [0, 100 \%]$ ,  $F_{vc}$  is:

$$F_{vc} = K_{vc} \frac{V_{vc}}{R_{vc}} u \tag{3}$$

The resulting flow  $\dot{V}$  was derived by the first differential of the height h:

$$\dot{V} = \dot{h}A \tag{4}$$

#### C. Parameter estimation

Based on the model from (1), the parameters m, d and k needed to be estimated for system identification. In preliminary experiments, the system behavior of the test lung was not reproducible, presumably due to friction. To minimize the effect of friction, a virtual downward force  $F_{vt} = 5.3 \text{ N}$  and a virtual spring  $k_{vt} = 490 \text{ N/m}$  were introduced, whose forces were realized by the VCA. These forces are dominant compared to the friction force when  $\dot{h}$  is close to zero. The virtual spring has a rest length of 16.6 mm, which is the center of the VCA stroke length. The subsequent experiments showed



Fig. 4. Measured pressure  $p_{bel}$  during identification.

reproducible behavior. The resulting model during parameter estimation is:

$$\ddot{mh} + d\dot{h} + kh + mg + F_{vt} = F_{vc} - k_{vt}(h - 16.6 \,\mathrm{mm}) + F_{n}.$$
 (5)

During identification, the mechanical ventilator was operated in pressure-controlled mode so that the output pressure alternated between the inspiratory pressure  $p_{insp}$  during inspiration and the PEEP (*positive end-expiratory pressure*) during expiration. The PEEP was constant at 6 hPa,  $p_{insp}$ increased from 10 hPa to 22 hPa in 2 hPa steps per breath, and then decreased to 10 hPa in equal steps. The identification experiment was automated on the real-time computer. The measured curve of  $p_{bel}$  is shown in Fig. 4. The height h was chosen as the output to avoid fluidic effects during parameter estimation of test lung.

Parameter estimation was carried out using the *parameter* estimation toolbox of MATLAB2022b. The sum of squared errors between the simulated height  $\hat{h}$  and the measured height h was minimized using the nonlinear least squares method. To validate the estimated parameters, the data of 12 breaths were used, which were not used during the identification but were measured in the same experiment and look similar to the pressures from Fig. 4.

### D. Control design

For the control design, the model identified in section II-C was used. Based on (1) to (4), the model was defined in state-space form with u as an input and with h and  $\dot{V}$  as outputs:

$$\mathbf{x} = \begin{pmatrix} h \\ \dot{h} \end{pmatrix} \qquad , \mathbf{y} = \begin{pmatrix} h \\ \dot{V} \end{pmatrix} \tag{6}$$

$$\dot{\mathbf{x}} = \mathbf{A}\mathbf{x} + \mathbf{B}u \qquad , \mathbf{y} = \mathbf{C}\mathbf{x} + \mathbf{D}u \qquad (7)$$

$$\mathbf{A} = \begin{pmatrix} 0 & 1\\ -\frac{k}{m} & -\frac{d}{m} \end{pmatrix} \qquad , \mathbf{B} = \begin{pmatrix} 0\\ -\frac{K_{vc}V_{vc}}{mR_{vc}} \end{pmatrix} \tag{8}$$

$$\mathbf{C} = \begin{pmatrix} 1 & 0 \\ 0 & A \end{pmatrix} \qquad , \mathbf{D} = \begin{pmatrix} 0 \\ 0 \end{pmatrix} \qquad (9)$$

The controller must mimic the volume flow of the simulated lung while keeping the metal plate in operating range. Therefore, the controller objective was to track a reference height  $h^*$ 



Fig. 5. Loop shaping procedure by McFarlane and Glover [11]. The controller  $\mathbf{K}(s)$  comprises the compensators  $W_1(s)$  and  $\mathbf{W}_2(s)$ , and the controller  $\mathbf{K}_{\infty}(s)$  gained by  $H_{\infty}$  synthesis.

and a reference flow  $\dot{V}^*$ , summarized in the reference vector **r**:

$$\mathbf{r} = \begin{pmatrix} h^* \\ \dot{V}^* \end{pmatrix} \tag{10}$$

The controller was synthesized using the  $H_{\infty}$  loop-shaping approach proposed by McFarlane and Glover [11] with the structure given in Fig. 5. First, the plant  $\mathbf{G}(s)$  is shaped with the compensators  $W_1(s)$ , in our case a scalar, and  $\mathbf{W}_2(s)$ , leading to the shaped open-loop transfer function  $\mathbf{G}_s(s)$ :

$$\mathbf{G}_s(s) = \mathbf{W}_2(s)\mathbf{G}(s)W_1(s) \tag{11}$$

Afterwards, a stabilizing  $H_{\infty}$  controller was synthesized for the system.

 $W_1(s)$  was set to 1 and the loop was shaped with  $W_2(s)$ . To normalize both reference tracking goals of **r**, we introduced a scaling matrix  $\mathbf{D}_e$  that contains the maximum possible height  $h_{max} = 0.035 \,\mathrm{m}$  and maximum observed flow  $\dot{V}_{max} = 1.8 \,\mathrm{L/s}$ :

$$\mathbf{D}_e = \begin{pmatrix} h_{max} & 0\\ 0 & \dot{V}_{max} \end{pmatrix} \tag{12}$$

The height open-loop transfer function must be shaped to allow the metal plate to move up and down without exceeding the VCA's height operating space. As long as the operating space is maintained, the closed-loop response to the reference height can be slow. The reference tracking of the height was prioritized in frequency regions below  $0.07 \,\mathrm{rad/s}$  by adding a gain of  $5 \,\mathrm{dB} = 1.77$  and introducing a low-pass filter with a time constant of  $80 \,\mathrm{s}$ .

For the flow open-loop transfer function, the controller must respond quickly. First, a gain of 25 dB = 17.8 was introduced to prioritize this goal for frequencies above 0.07 rad/s. To limit the controller bandwidth, a low-pass filter with a time constant of 2.0 s was introduced. In the last step, a zero with the time constant of 0.02 s was introduced to increase the phase margin. The final loop-shaping compensator  $\mathbf{W}_2(s)$  is given by:

$$\mathbf{W}_{2} = \mathbf{D}_{e}^{-1} \begin{bmatrix} \frac{1.77}{1+80s} & 0\\ 0 & 17.8 \frac{1+0.02s}{1+2s} \end{bmatrix}$$
(13)

The command notsyn from *MATLAB 2019b* was used to synthesize the loop-shaping  $H_{\infty}$  controller  $\mathbf{K}(s)$  of order five. After inspecting the Hankel singular values, the controller was reduced to the model order of three via the command balred. The Bode plots of the desired and the actual open-loop transfer functions are given in Fig. 6.



Fig. 6. Bode plots of the open-loop transfer function for the height (blue) and flow (red). The dashed lines denote the desired loop shape  $\mathbf{G}_s(s)$  of (11), the solid lines the actual open-loop transfer function  $\mathbf{G}(s)\mathbf{K}(s)$  of the  $H_{\infty}$  synthesis.



Fig. 7. Equivalent circuit model of the lung with spontaneous breathing [12, p. 149].

#### E. Reference lung model

A model of the respiratory system was used to obtain a reference flow for the test lung. The respiratory system is modeled as a one-compartment system as depicted in the equivalent circuit in Fig. 7. The model has a volume flow resistance R and a lung compliance C. The spontaneous respiratory muscles generate the pressure  $p_{mus}$ , the ventilator generates the airway pressure  $p_{aw}$ . Using the volume flow  $V_{aw}$ , the differential equation is obtained:

$$p_{aw} = R\dot{V}_{aw}(t) + \frac{1}{C}\int \dot{V}_{aw}(t)dt + p_{mus}(t)$$
 (14)

A description of  $p_{mus}$  was used from the supplementary material of Leonhardt and Walter [12]. The amplitude  $\hat{p}_{mus}$ , the spontaneous breathing frequency  $f_{mus}$ , the inspiratory time  $T_i$ , and the expiratory time  $T_e$  are setting parameters. Repeated with  $f_{mus}$ ,  $p_{mus}$  is given by:

$$\left( \left( 1 - \exp\left(-\frac{5t}{T_i}\right) \right), 0 < t \le T_i \quad (15)$$

$$p_{mus} = -\hat{p}_{mus} \left\{ \exp(-\frac{4(t-T_i)}{T_e}), T_i < t \le T_i + T_e \quad (16) \right\}$$

$$D, T_i + T_e < t \le \frac{1}{f_{mus}} (17)$$

During evaluation, the parameters were set to the values R = 1.0 hPa s/L,  $C = 1.5 \times 10^{-2} \text{ L/hPa}$ ,  $T_i = 1.3 \text{ s}$ ,  $T_e = 1.5 \text{ s}$ ,  $f_{mus} = 15 \text{ min}^{-1}$ ,  $\hat{p}_{mus} = 10 \text{ hPa}$ , and  $p_{aw} = 5 \text{ hPa}$ .

## F. Reference phrenic nerve stimulation data

For reference data of phrenic nerve stimulation, measurements from previous animal studies similar to previously published results [13] were used. The animal study was approved by the appropriate governmental institution (Landesamt für Natur, Umwelt und Verbraucherschutz Nordrhein-Westfalen, LANUV NRW, Germany, reference number: 81-02.04.2020.A080, date of approval: 07.07.2020) and performed in accordance with German legislation governing animal studies following the "Guide for the care and use of Laboratory Animals" (NIH publication, 8th edition, 2011), the principles for care and use of animals based on the Helsinki declaration and the Directive 2010/63/EU on the protection of animals used for scientific purposes (Official Journal of the European Union, 2010).

From the data, breaths induced by stimulation were extracted and in total, 16 minutes of flow measurements were used. The data include breaths with varying inhaled volumes and therefore different flow rates. The reference flow was scaled by  $\frac{1}{3}$  so that the maximum volume induced by the reference flow did not exceed the maximum volume of the test lung.

#### **III. RESULTS**

#### A. Parameter estimation

The system identification resulted in the parameters m = 0.95 kg, d = 81 N m/s, k = 157 N/m. The validation plot is given in Fig. 8. While the simulated values were similar to the measured values when the VCA was rising or falling, the differences became greater when the VCA moved more slowly.

The root-mean-square-error (RMSE) and the maximum error  $\bar{e}$  are given in Table I. With a maximum stroke of 33 mm of the VCA, the RMSE corresponds to 2.91 % and  $\bar{e}$  corresponds to 11.36 % of the maximum stroke during validation.

 
 TABLE I

 RMSE and maximum error  $\bar{e}$  of the parameter estimation during training and validation.

Error	Training	Validation
RMSE	$0.89\mathrm{mm}$	$0.96\mathrm{mm}$
$\bar{e}$	$3.61\mathrm{mm}$	$3.75\mathrm{mm}$



Fig. 8. Comparison between the measured (blue line) and the simulated (red line) height during validation.

#### B. Controller performance

The total recording time of the test lung performance against the reference flow from the lung model of section II-E was 300 s. An example time window of the test lung performance is given in Fig. 9. While the flow curve has a periodic shape, the height changed according to different reference height  $h^*$ settings. The RMSE between reference and measured flow was  $6.2 \times 10^{-2}$  L/s. An illustrative single breath is shown in Fig. 10 (a). During inspiration, the peak measured flow is 0.01 s later than the peak reference flow while during expiration, there was an overshoot of approximately 13 %  $(4.5 \times 10^{-2}$  L/s). Integrating the positive flow over each breath and dividing it by 10 hPa yielded a mean compliance of  $1.41 \times 10^{-2}$  L/hPa. This corresponds to a relative error of 5.88 % compared to the reference compliance.

For the test lung performance with respect to stimulation measurements, a 60 s time window is shown in Fig. 11. The controller achieved an RMSE of  $2.9 \times 10^{-2}$  L/s. The single breath in Fig. 10 (b) showed no delay or overshoot, in contrast to Fig. 10 (a).

#### **IV. DISCUSSION**

The proposed mechatronic lung was able to follow the reference flow curves of the lung model as shown in Fig. 9, and the reference flow curves of experiment measurements with phrenic nerve stimulation as depicted in Fig. 11. Therefore, the novel approach of a test lung with a VCA can simulate lung behavior.

Both Fig. 8 and Table I show that the presented model behaves similarly to the physical system. However, it is noticeable that the difference between the measured and simulated deflection increases as soon as the VCA slows down. Since the plate velocity decreased during this time, unmodeled friction effects may have occurred. Nevertheless, the model accuracy was sufficient for the proposed robust control design, but it remains unclear whether the model accuracy is sufficient for other control designs such as model predictive control.

During control design, a proportional-integral flow controller was evaluated, but the height h had a drift that caused the VCA to leave its operating range after a few breaths. To solve this problem, the proposed controller concept was



Fig. 9. Flow (top) and plate height (bottom) during control validation via the reference model of the lung. The blue lines are measurements, the red dashed lines are the references.

extended to a multiple-input-single-output controller. In Fig. 9, the controller kept the alternating height levels constant between breaths, changing them according to adjustments of the reference height. Simultaneously, the controller continued to follow the reference flow. This allowed the VCA height to be kept within operating range. A low-order controller was chosen for its computationally efficient structure, so that it can be implemented on low-cost hardware in the future. However, if computationally feasible, model predictive control, which defines the height as a constraint rather than a control goal, is a promising alternative.

The relative compliance error of 5.88% is similar to the error of 5.00% given by the specifications of the *ASL 5000*® *Breathing Simulator* and the *Spontaneous Breathing Lung Simulator* (*SBL*<sup>TM</sup>). In the future, the proposed test lung may become more flexible as physiological based models of patient response to the mechanical ventilator and the phrenic nerve stimulation interface are implemented.

In terms of limitations, a maximum deflection of the VCA results in a volume range of 0.40 L. Adding a safety margin of 2 mm to the limits so that the metal plate does not hit them reduces the maximum possible volume to 0.36 L. Since adults inhale and exhale an average of 0.50 L per breath, the test lung would have to be duplicated and connected in parallel to simulate adults.

The close-up performance of a single breath from Fig. 10



Fig. 10. Top: Close-up comparison from (a) Fig. 9, and (b) Fig. 11 between the measured airway flow from the test lung (blue line) and the scaled reference flow (red dashed line). Bottom: Corresponding PWM values. Negative values indicate that the VCA applied force downward.



Fig. 11. Measured flow of the test lung (blue line) compared to the scaled reference flow (red dashed line) from animal studies.

shows that the measured flow of the test lung differed from the reference flow of the lung model. These were probably caused by the high frequencies in the reference flow. While the high dynamics of the VCA should allow the simulation of high frequency reference changes, the controller concept may need to be modified to simulate high frequency reference changes during e.g. coughing. One possible solution is to introduce additional feed-forward control.

While the test lung was able to replicate flows from stimulation measurements, no phrenic nerve stimulation models were used. To extend the test lung to HiL stimulation, a stimulation interface and reference models for stimulation need to be developed and implemented. Overall, the test lung has the potential to be used for HiL simulations to test either mechanical ventilators or their automation concepts.

#### REFERENCES

- A. Esteban, "Characteristics and Outcomes in Adult Patients Receiving Mechanical VentilationA 28-Day International Study," *JAMA*, vol. 287, no. 3, p. 345, Jan. 2002.
- [2] D. C. Angus, "Current and Projected Workforce Requirements for Care of the Critically III and Patients With Pulmonary DiseaseCan We Meet the Requirements of an Aging Population?" *JAMA*, vol. 284, no. 21, p. 2762, Dec. 2000.
- [3] P. von Platen, A. Pomprapa, B. Lachmann, and S. Leonhardt, "The dawn of physiological closed-loop ventilation—a review," *Critical Care*, vol. 24, no. 1, p. 121, Dec. 2020.
- [4] S. Festing and R. Wilkinson, "The ethics of animal research: Talking Point on the use of animals in scientific research," *EMBO reports*, vol. 8, no. 6, pp. 526–530, Jun. 2007.
- [5] W. M. S. Russell and R. L. Burch, *The Principles of Humane Experimental Technique*. Potters Bar, Herts: Universities Federation for Animal Welfare, 1992.
- [6] P. David, H. Sandra, M. Georg, and R. Philipp, "Control of a blowerbased lung simulator for testing of mechanical ventilators," *Current Directions in Biomedical Engineering*, vol. 8, no. 2, pp. 719–722, Sep. 2022.
- [7] J. Mecklenburgh, T. Al-Obaidi, and W. Mapleson, "A Model Lung With Direct Representation of Respiratory Muscle Activity," *British Journal* of Anaesthesia, vol. 68, no. 6, pp. 603–612, Jun. 1992.
- [8] M. Soták, K. Roubík, T. Henlín, and T. Tyll, "Phrenic nerve stimulation prevents diaphragm atrophy in patients with respiratory failure on mechanical ventilation," *BMC Pulmonary Medicine*, vol. 21, no. 1, p. 314, Dec. 2021.
- [9] S. N. A. Hussain, M. Mofarrahi, I. Sigala, H. C. Kim, T. Vassilakopoulos, F. Maltais, I. Bellenis, R. Chaturvedi, S. B. Gottfried, P. Metrakos, G. Danialou, S. Matecki, S. Jaber, B. J. Petrof, and P. Goldberg, "Mechanical ventilation-induced diaphragm disuse in humans triggers autophagy," *American Journal of Respiratory and Critical Care Medicine*, vol. 182, no. 11, pp. 1377–1386, 2010.
- [10] Lohse, Arnhold, Hamidov, Farid, Bing, Junxiao, von Platen, Philip, Leonhardt, Steffen, and Walter, Marian, "Entwicklung einer Mechatronischen Testlunge mit Zwerchfellfunktion," in VDI Mechatroniktagung 2024. Dresden, Germany: In press.
- [11] D. McFarlane and K. Glover, "A loop-shaping design procedure using H infinity synthesis," *IEEE Transactions on Automatic Control*, vol. 37, no. 6, pp. 759–769, Jun. 1992.
- [12] S. Leonhardt and M. Walter, Eds., *Medizintechnische Systeme*. Berlin, Heidelberg: Springer Berlin Heidelberg, 2016.
- [13] A. Lohse, P. Von Platen, C.-F. Benner, S. Leonhardt, M. Walter, M. M. Deininger, D. Ziles, T. Seemann, and T. Breuer, "Identification of the Tidal Volume Response to Pulse Amplitudes of Phrenic Nerve Stimulation Using Gaussian Process Regression," in 2022 44th Annual International Conference of the IEEE Engineering in Medicine & Biology Society (EMBC). Glasgow, Scotland, United Kingdom: IEEE, Jul. 2022, pp. 135–138.