Simulated Ventilation of two Patients with a Single Ventilator in a Pandemic Setting

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Abstract

Background: Simultaneous ventilation of two patients, e.g. due to a shortage of ventilators in a pandemic, may result in hypoventilation in one patient and hyperinflation in the other patient.

Methods: In a simulation of double patient ventilation using artificial lungs with equal compliances (70mL•mbar^{-1}), we tried to voluntarily direct gas flow to one patient by using 3D-printed y-adapters and stenosis adapters during volume-, and pressure-controlled ventilation. Subsequently, we modified the model using a special one-way valve on the limited flow side and measured in pressure-controlled ventilation with the flow sensor adjusted to either side in a second and third setup. In the last setup, we also measured with different lung compliances.

Results: Volume- or pressure-controlled ventilation using standard connection tubes with the same compliance in each lung resulted in comparable minute volumes in both lungs, even if one side was obstructed to 3mm (6.6±0.2vs.6.5±0.1L for volume-controlled ventilation, p=.25 continuous severe alarm and 7.4±0.1vs.6.1±0.1L for pressure-controlled ventilation, p=.02 no alarm). In the second setup, pressure-controlled ventilation resulted at 3mm flow limitation in minute ventilation of 9.4±0.3vs3.5±0.1L•min^{-1}, p=.001. In a third setup using the special one-way valve and the flow sensor on the unobstructed side, pressure-controlled ventilation resulted at 3mm flow limitation in minute ventilation of 7.4±0.2vs3±0L•min^{-1}, at the compliance of 70mL•mbar^{-1} for both lungs, 7.2±0vs4.1±0L• min^{-1}, at the compliances of 50 vs. 70mL•mbar^{-1}, and 7.2±0.2vs5.7±0L• min^{-1}, at the compliance of 30 vs. 70mL•mbar^{-1} (all p=.001).

Conclusions: Overriding a modern intensive care ventilator's safety features are possible, thereby ventilating two lungs with one ventilator simultaneously in a laboratory simulation using 3D-printed y-adapters. Directing tidal volumes in different pulmonary conditions towards one lung using 3D-printed flow limiters with diameters <6mm was also possible. While this ventilation setting was technically feasible in a bench model, it would be volatile, if not dangerous in a clinical situation.

Background

The COVID-19 Pandemic revealed in hospitals dramatic shortage of experienced staff, medical equipment, and especially a lack of ventilators. In some cases, intensive care physicians had to perform triage deciding which patient could receive intubation and mechanical ventilation. Such situations were unimaginable in western industrialised countries before the COVID-19 Pandemic. In desperate situations, it was even attempted to share one ventilator for two patients. However, due to different resistance and lung compliances, this may result in hypoventilation in one patient and hyperinflation in the other patient, which may cause lung damage[1], and adversely affect morbidity and mortality.

A simple solution to distribute tidal volume of one ventilator between two patients may be a 3D printed y-adapter, which had been reported before during the 9/11 terrorist attacks and Hurricane Katrina in New

Orleans[2]. Recent reports from the United States show that comparable devices had been used in the Covid-19 Pandemic as well[3]. Historical studies demonstrated this strategy in a laboratory setting[2].However, as the tidal volume cannot be distributed adequately to each patient mainly due to different resistance and lung compliances, this may result in hypoventilation in one patient, and hyperinflation in the other patient. This may result in hypercarbia or even hypoxemia in a given patient with the “worse” lung, while the better lung may be hyperinflated and thus become severely damaged. To avoid such a misdistribution between patients, we developed a 3D printed stenosis adapter that can be inserted in the respirator tube to one patient. Thus, in theory, the tidal volume can be distributed variably between patients.

We evaluated gas distribution in a ventilation bench model using artificial lungs with equal lung compliance during volume-controlled, and pressure-controlled ventilation by inserting stenosis adapters of 2-9mm, respectively. Subsequently, we modified the model using a special one-way valve and evaluated in pressure-controlled ventilation mode the influence of these stenosis adapters on gas distribution between lungs with different lung compliance settings representing ventilation of standard intensive care patients vs. ARDS patients, e.g., due to Covid-19. Our formal hypothesis was that there would be no difference between groups.

Methods

Ethics approval

This study is a completely technical simulation with no participants. Thus, no ethical approval was required.

Experimental setup

We developed a y-adapter fitting to a modern ventilation device according to ISO 5356-1 (Figure 1 A). This adapter can be used both in the in- and the expiratory branch of the ventilation tubes. Accordingly, we developed flow limitation devices with a diameter of 2-9mm that could be inserted in the ventilator tubes (Figure 1B).

Both pieces consisted of carbon fibre (Multec Carbon, Multec, Illmensee, Germany), and were printed on a 3D printer (Multec Multirap M800, Multec, Illmensee, Germany). The average printing time was 4 hrs; the adapter could be disinfected and sterilised. For lung and airway simulation, we used a double test lung with different lung compliances and airway resistance that are adjustable in each test lung separately (dual Adult TTL â - Model 5600i, Michigan Instruments, Grand Rapids, MI). Each side was set to represent airway characteristics of a patient; in result the dual lung represented airway characteristics of two patients. For determining minute ventilation, we included a respirometer Ferraris, Hertford, UK) in the tubes connecting the ventilator with each of the separated test lungs simulating one patient.

We adjusted the y-adapter to a ventilator (Hamilton C6, Bonaduz, Switzerland), allowing parallel ventilation of both test lungs (Figure 2).

In a pilot setup, we used standard double-lumen ventilation tubes distally of the y-adapter enabling gas to flow back to the ventilator in both designs. The flow limiters were adjusted on side B directly behind the inspiratory y-adapter. (Figure 2A). In a second setup, we integrated the flow limiter in circuit B again via a standard double-lumen connection tube, allowing expiratory gas to flow back to the ventilator from lung B; to the double-lumen connection tube to test lung A, we adjusted distally the double-lumen tube a valve of an AMBU Mark V bag-valve device (Ambu Glostrup, Copenhagen, Denmark) thus directing expiratory gas flow directly into ambient air through an HME-filter, and not back to the ventilator via the double-lumen tube. As a result, only expiratory gas of test lung B flowed back to the ventilator. The flow sensor was adjusted to the double-lumen side on circuit B (Figure 2B). In a third setup, we connected test lung A with a standard double-lumen connection tube, allowing expiratory gas to flow from lung A back to the ventilator. The double-lumen connection tube was connected to test lung B; we adjusted distally the flow diameter and the expiratory valve of an AMBU Mark V bag-valve device (Ambu Glostrup, Copenhagen, Denmark) thus directing expiratory gas flow from lung B directly into ambient air. The flow sensor was adjusted to the standard double-lumen side on circuit A (Figure 2C).

Experimental procedure:

In total, six experiments were performed in the aforementioned setups; each experiment was performed for 1 min and repeated six times. In the first experiment (Figure 2A), both lungs were ventilated first in volume-controlled mode with a compliance of 30mL•mbar^{-1} simulating moderate ARDS in both lungs (respiratory rate 12•min^{-1}, tidal volume 1200mL, p_{max} 20mbar, PEEP 5mbar, FiO_2 0.21, I:E 1:2). The first measurement was without a flow limiter. Subsequently, flow limiters were integrated into the inspiratory tube connected to test lung B. The flow limiter diameter was decreased with each attempt, starting with 9mm and ending with 2mm. The second experiment was done in pressure-controlled mode in the same way (respiratory rate 12•min^{-1}, P_{max} 20mbar, PEEP 5mbar, FiO_2 0.21, I:E 1:2). In all pressure-controlled experiments, peak pressure was adjusted to achieve a tidal volume of 600mL in the unobstructed lung A. Since measurements were done with different compliances, this was adjusted for each experiment.

In the third experiment, the pressure-controlled experiment two in the second setup (Figure 2B) was repeated. The peak inspiratory pressure to achieve tidal volumes in test lung A of 600 mL was adjusted. Compliance of test lung A was 70mL•mbar^{-1} simulating a healthy lung. The fourth experiment was pressure-controlled ventilation in analogy to experiment 2 and 3 in the third setup (Figure 2C) with a compliance of 70mL•mbar^{-1} in both lungs. In the last two experiments, the last setup (Figure 2C) was also used and compared test lung A with 50 and 30 mL•mbar^{-1} to 70mL•mbar^{-1} in test lung B, respectively. Pressure for ventilation was always adjusted to achieve tidal volumes of 600mL.

Statistical analysis:

Statistical evaluation was performed with Sigmaplot 14 (Systat, San Jose, CA) using Student's *t*-test after Shapiro Wilk Analysis for normality and Equal Variance Test (Brown-Forsythe). If a normality test failed, the Mann-Whitney rank-sum test was applied. Data are given as mean ± standard deviation. P values <0.05 were considered to be significant.

Results

In experiment one and two, we measured comparable minute volumes in both lungs using either volume or flow-controlled ventilation (Tables 1 and 2). Connecting the flow sensor in circuit A in front of the lung resulted in a severe continuous disconnection alarm in volume-controlled ventilation mode, although ventilation was adequate (Table 1). Using flow limiters in experiments, one and two a larger diameter of the flow limiter (5–9 mm) resulted in larger tidal volumes on the obstructed side. Using flow limiters of 2–4 mm resulted in more considerable minute ventilation on the unobstructed side (Tables 1 and 2). In the third experiment, using flow limitation distributed tidal volumes to one lung. However, this was moderately hyperventilated (Table 3). In experiment four including the ventilator`s flow sensor on the unobstructed side and using a special one-way valve behind the flow limiter, volumes on the obstructed side decreased minute ventilation – in extremis to 0.8 L at a diameter of 2 mm (Table 4). In experiments five and six, simultaneous ventilation of two lungs with different compliances was also possible by including flow limiters of < 6 mm flow diameter to the lung with better compliance (Tables 5 and 6).

Table 1

Volume-controlled mode with the same compliance in both lungs of 30ml/mbar. Flow limitation on circuit B flow sensor on circuit A, double hose circuit for lungs A and B.

Y-adapter inside diameter	Test lung A M.V. (L/min)	Test lung B M.V. (L/min)	F.S. C6 (A) M.V. (L/min)	Hamilton C6 pPeak (mbar)	p Test lung A vs. B
A: Ø 19mm, B: Ø 19mm	6.4 ± 0.4	6.8 ± 0.3*	6.2 ± 0.1	28 ± 0	0.25
A: Ø 19mm, B: Ø 9mm	6.3 ± 0.4	6.9 ± 0.3*	6,3 ± 0.1	28 ± 0	0.03
A: Ø 19mm, B: Ø 8mm	6.4 ± 0.3	6.7 ± 0.1*	6.1 ± 0.1	28 ± 0	0.12
A: Ø 19mm, B: Ø 7mm	6.6 ± 0.3	6.7 ± 0.2*	5.8 ± 0.1	28 ± 0	0.09
A: Ø 19mm, B: Ø 6mm	6.6 ± 0.4	6.7 ± 0.2*	6.1 ± 0.1	28 ± 0	0.12
A: Ø 19mm, B: Ø 5mm	6.7 ± 0.2	6.6 ± 0.1*	6.0 ± 0.2	28 ± 0	0.35
A: Ø 19mm, B: Ø 4mm	6.5 ± 0.2	6.5 ± 0.1*	5.8 ± 0.1	27 ± 0	0.35
A: Ø 19mm, B: Ø 3mm	6.5 ± 0.2	6.4 ± 0.1*	5.7 ± 0.2	27 ± 0	0.92
A: Ø 19mm, B: Ø 2mm	6.6 ± 0.2	6.5 ± 0.1*	5.7 ± 0.1	27 ± 0	0.22

A = Inside diameter of the y-adapter according to ISO 5356-1; B = With and without (19mm) limitation of the inside diameter by the flow limiter; M.V.; minute ventilation; *Critical alert: Disconnection on patient-side alarm. If flow sensor is connected prior to test lung B and not prior the y-adapter stop of ventilation; F.S. = flow sensor on circuit A

Table 2

Pressure-controlled ventilation with the same compliance of 30mL/mbar in both lungs. Flow limitation, and flow sensor in front of y-adapter, double hose circuit for lungs A and B.

Y-adapter inside diameter	Test lung A M.V. (L/min)	Test lung B M.V. (L/min)	F.S. on A M.V. (L/min)	Hamilton C6 pPeak (mbar)	P Test lung A vs. B
A: Ø 19mm, B: Ø 19mm	7.6 ± 0.2	7.8 ± 0.3	7.1 ± 0.1*	28 ± 0*	0.34
A: Ø 19mm, B: Ø 9mm	7.4 ± 0.2	7.9 ± 0.2	6.8 ± 0.1*	28 ± 0*	0.03
A: Ø 19mm, B: Ø 8mm	7.5 ± 0.3	7.9 ± 0.2	7.3 ± 0.1*	28 ± 0*	0.08
A: Ø 19mm, B: Ø 7mm	7.5 ± 0.1	8.0 ± 0.1	7.2 ± 0.1*	28 ± 0*	0.03
A: Ø 19mm, B: Ø 6mm	7.5 ± 0.1	7.9 ± 0.1	7.1 ± 0.1*	28 ± 0*	0.03
A: Ø 19mm, B: Ø 5mm	7.4 ± 01	7.5 ± 0.1	7.1 ± 0.1*	28 ± 0*	0.04
A: Ø 19mm, B: Ø 4mm	7.4 ± 0.1	7.4 ± 0.1	7.1 ± 0.1*	26 ± 0*	0.35
A: Ø 19mm, B: Ø 3mm	7.4 ± 0.1	7.2 ± 0.1	7.0 ± 0.1*	26 ± 0*	0.03
A: Ø 19mm, B: Ø 2mm	7.4 ± 0.1	7.1 ± 0.1	7.1 ± 0.1*	26 ± 0*	0.03

A = Inside diameter of the y-adapter according to ISO 5356-1; B = With and without (19mm) limitation of the inside diameter by the flow limiter; M.V.; minute ventilation; *No machine-side error messages if flow sensor is attached to test lung B; F.S. = flow sensor on circuit A

Table 3

Pressure-controlled ventilation with the same compliance in both lungs of 70mL/mbar. Flow limitation on circuit B and flow sensor on circuit B. Single hose in circuit A and a double hose in circuit B.

Y-adapter inside diameter	**Test lung A M.V. (L/min)**	**Test lung B M.V. (L/min)**	**Hamilton C6 M.V. (L/min)**	**Hamilton C6 pPeak (mbar)#**	**P for M.V. Test lung A vs. B**
A: Ø 19mm, B: Ø 19mm	8.1 ± 0.4	7.8 ± 0.4	6.8 ± 0.2	14 ± 0	0.3
A: Ø 19mm, B: Ø 9mm	8.9 ± 0.2	7.9 ± 0.5	6.9 ± 0.11	14 ± 0	0.04
A: Ø 19mm, B: Ø 8mm	9.3 ± 0.3	8.0 ± 0.4	6.6 ± 0.2	14 ± 0	0.001
A: Ø 19mm, B: Ø 7mm	9.4 ± 0.2	7.8 ± 0.1	6.2 ± 0.2	14 ± 0	0.002
A: Ø 19mm, B: Ø 6mm	9.6 ± 0.1	7.9 ± 0.3	6.7 ± 0	14 ± 0	0.002
A: Ø 19mm, B: Ø 5mm	10.5 ± 0	7.9 ± 0.1	6.9 ± 0.1	14 ± 0	0.002
A: Ø 19mm, B: Ø 4mm	10.2 ± 0.1	6 ± 0.1	5.4 ± 0.1	14 ± 0	0.001
A: Ø 19mm, B: Ø 3mm	9.4 ± 0.3	3.5 ± 0.1	3.0 ± 0.2	13 ± 0	0.001
A: Ø 19mm, B: Ø 2mm	**Failed**				

A = Inside diameter of the y-adapter according to ISO 5356-1; B = With and without (19mm) limitation of the inside diameter by the flow limiter; * M.V.; minute ventilation; No machine-side error messages if flow sensor is attached to test lung B. # Adjustment of the upper ventilation pressure to 14 mbar due to a change in lung compliance test lung A

Table 4

Pressure-controlled ventilation with the same compliance in both lungs of 70ml/mbar. Flow limitation on circuit B and flow sensor on circuit A. Single hose in circuit A and a double hose in circuit B.

Y-adapter inside diameter	**Test lung A M.V. (l/min)**	**Test lung B M.V. (l/min)**	**Hamilton C6 M.V. (l/min)**	**Hamilton C6 pPeak (mbar)#**	**P Test lung A vs. B**
A: Ø 19mm, B: Ø 19mm	8.0 ± 0.4	7.8 ± 0.4	7.2 ± 0,1	14 ± 0	0.3
A: Ø 19mm, B: Ø 9mm	6.7 ± 0.1	6.6 ± 0.1	6.7 ± 0,1	14 ± 0	0.25
A: Ø 19mm, B: Ø 8mm	7.5 ± 0.3	6.5 ± 0.2	6.7 ± 0	14 ± 0	0.001
A: Ø 19mm, B: Ø 7mm	7.9 ± 0	7.0 ± 0	6.7 ± 0	14 ± 0	0.001
A: Ø 19mm, B: Ø 6mm	7.5 ± 0.1	6.5 ± 0.1	6.7 ± 0	14 ± 0	0.001
A: Ø 19mm, B: Ø 5mm	7.8 ± 0.1	6.5 ± 0.1	6.7 ± 0	14 ± 0	0.001
A: Ø 19mm, B: Ø 4mm	7.5 ± 0	5 ± 0	6.7 ± 0	14 ± 0	0.001
A: Ø 19mm, B: Ø 3mm	7.7 ± 0.1	3 ± 0	6.7 ± 0	14 ± 0	0.001
A: Ø 19mm, B: Ø 2mm	7.4 ± 0.2	0.8 ± 0	6.7 ± 0	14 ± 0	0.001

A = Inside diameter of the y-adapter according to ISO 5356-1; B = With and without (19mm) limitation of the inside diameter by the flow limiter; * M.V.; minute ventilation; No machine-side error messages if flow sensor is attached to test lung A. # Adjustment of the upper ventilation pressure to 14 mbar due to a change in lung compliance test lung A

Table 5

Pressure-controlled ventilation with compliance of 50mL/mbar in Lung A and 70ml/mbar in Lung B. Flow limitation on circuit B and flow sensor on circuit A. Single hose in circuit B and a double hose in circuit A.

Y-adapter inside diameter	Test lung A M.V. (l/min)	Test lung B M.V. (l/min)	Hamilton C6 M.V. (l/min)	Hamilton C6 pPeak (mbar)#	P Test lung A vs. B
A: Ø 19mm, B: Ø 19mm	7.4 ± 0.1	9.6 ± 0.2	7.2 ± 0.1	18 ± 0	0.001
A: Ø 19mm, B: Ø 6mm	7.6 ± 0.2	9.4 ± 0.4	6.9 ± 0	18 ± 0	0.001
A: Ø 19mm, B: Ø 5mm	7.6± 0	8.5 ± 0.1	6.9 ± 0	18 ± 0	0.001
A: Ø 19mm, B: Ø 4mm	7.4 ± 0	6.6 ± 0	6.9 ± 0	18 ± 0	0.001
A: Ø 19mm, B: Ø 3mm	7.2 ± 0	4.1 ± 0	6.9 ± 0	18 ± 0	0.001
A: Ø 19mm, B: Ø 2mm	7.2 ± 0	1.2 ± 0.1	6.9 ± 0	18 ± 0	0.001
A = Inside diameter of the y-adapter according to ISO 5356-1; B = With and without (19mm) limitation of the inside diameter by the flow limiter; M.V.; minute ventilation; * No machine-side error messages if flow sensor is attached to test lung A. # Adjustment of the upper ventilation pressure to 26 mbar due to a change in lung compliance test lung A.					

Table 6
Pressure-controlled ventilation with compliance of 30mL/mbar in Lung A and 70ml/mbar in Lung B. Flow limitation on circuit B and flow sensor on circuit A. Single hose in circuit A and a double hose in circuit B.

Y-adapter inside diameter	Test lung A M.V. (L/min)	Test lung B M.V. (L/min)	Hamilton C6 M.V. (L/min)	Hamilton C6 pPeak (mbar)#	P Test lung A vs. B
A: Ø 19mm, B: Ø 19mm	7.0 ± 0.1	14.5 ± 0.2	7.2 ± 0.1	26 ± 0	0.001
A: Ø 19mm, B: Ø 6mm	7.5 ± 0.1	14.5 ± 0.4	6.7 ± 0	26 ± 0	0.001
A: Ø 19mm, B: Ø 5mm	7.4± 0	12 ± 0.1	6.7 ± 0	26 ± 0	0.001
A: Ø 19mm, B: Ø 4mm	7.3 ± 0	9 ± 0	6.7 ± 0	26 ± 0	0.001
A: Ø 19mm, B: Ø 3mm	7.2 ± 0.2	5.7 ± 0	6.7 ± 0	26 ± 0	0.001
A: Ø 19mm, B: Ø 2mm	7.2 ± 0.2	2.0 ± 0.1	6.7 ± 0	26 ± 0	0.001
A = Inside diameter of the y-adapter according to ISO 5356-1; B = With and without (19mm) limitation of the inside diameter by the flow limiter; M.V.; minute ventilation; * No machine-side error messages if flow sensor is attached to test lung A. # Adjustment of the upper ventilation pressure to 26 mbar due to a change in lung compliance test lung A					

Discussion

In this bench model, simulated ventilation of two patients with a single ventilator was technically possible employing both volume- and pressure-controlled ventilation when patient parameters were comparable. The given lung´s compliance did not impact results when it was identical in both lungs. The ventilator was powerful enough to ventilate both lungs with compliances as low as 30mL/mba[1], nor did it oversteer when both lungs were set to 70mL/mbar. In a recently published study, the authors described that changes in airway mechanics in one patient might have less dangerous effects on the second patient when pressure-controlled ventilation was used for double patient ventilation instead of volume-controlled ventilation[4]. We could not validate this effect since volume-controlled ventilation in our model was hampered by continuous disconnection alarms by the ventilator[4]. Thus, we omitted the volume-controlled ventilation design and concentrated on pressure-controlled ventilation.

The limitation of gas flows by flow limiters was not possible in this model in setup one if standard double hose circuits were used on both lungs. We speculate that due to using two y-adapters in the in- and

expiratory circuit, we elevated the pressure first in the unobstructed inspiratory circuit, and then in the expiratory circuit on this side. This may have resulted due to the second y-adapter on the expiratory side of the ventilator in faster pressure elevation on the expiratory side than on the inspiratory side of the obstructed circuit. As a result, the obstructed lung may have been ventilated in a retrograde manner via the expiratory hose on the obstructed side. This is in accordance with the supplemental video, where delayed ventilation of the obstructed side was visible (Supplementary Video File 1 and 2). Further, we were not able to direct gas to one lung using flow limiters.

We changed the setup by inducing a one-way valve with an additional PEEP adapter to lung A. As a result, only gas from lung B streamed back to the ventilator avoiding this error in setup two. If the ventilator's flow sensor was connected to the obstructed side, ventilation was still possible up to a limitation diameter of 3mm, and it only failed at a diameter of 2mm. We suggest that modern, powerful ventilators can ventilate even through a stenosis of 3mm, being comparable to small neonate ventilation tubes, by increasing their ventilation efforts if the flow sensor gave feedback of insufficient ventilation on this side. Thus, the ventilator may have simply overcome our volume limitation attempts on one side by the turbine force, which may explain the turbine noises in this experiment. However, it resulted in a hyperventilation of the non-obstructed side, which could be detrimental in a real clinical setting. Thus, we omitted this setting as well. We then further modified to setup three which worked reliably: Lung A was ventilated in a conventional setting using a standard double-lumen hose and the flow sensor was fixed there, too; the inspiratory lumen of a bypass to lung B was obstructed by a flow limiter and a one-way valve expired ventilation gas into ambient air on side B. With lung compliances of 70mL•mbar^{-1} on both sides, it was possible to variably decrease minute ventilation with our flow limiters on the obstructed side to even 0.8L•min^{-1}. Further, ventilation was possible even in very different lung compliance settings if the worse lung A was ventilated in a conventional setting, whereas a bypass to the healthy lung B was obstructed (Setup C).

To significantly direct minute ventilation to one patient, in general, flow diameters of < 6mm had to be used. This contrasts with previous suggestions where far larger diameters (11-16mm) were suggested for this purpose[5]. In other studies adjustable flow limiters or a Hoffmann clamp adjusted to ventilation had been used to increase inspiratory resistance; however, exact diameters are not known in theses setting and cannot be standardised [[6, 7]]. Use of pressure control valves for distribution of gas between two patients as a completely different solution had been reported before, too [8]. Actually, it is not known which system may be the most promising; our solution is standardised and completely reproducible in regard of used diameters. To further standardise the model, we deliberately decided to combine the flow sensor always with the standard double-lumen hose as an original manufacturer's product; the other side was always connected with the one-way valve. The one-way valve side can be ventilated with the PEEP level preset on the ventilator plus an additional PEEP by adjusting a simple PEEP valve on the one-way valve. In the only functional setup three of this study, the one-way valve was adjusted to lung B, and thus it would only allow higher PEEP levels on the clinically better side, which does clinically make not much

sense. Whether it is possible to adjust both the flow sensor and the one-way valve on one side is not clear and may be subject of further studies.

We must admit that it requires some "criminal energy" to override a modern ventilator's safety features. This would create a setting requiring very skilled monitoring by a highly experienced healthcare worker and would still be dangerous. However, the COVID-19 Pandemic showed us that even health care systems in the highest developed countries might collapse such as in Bergamo/Italy, Strasbourg/France, or New York City, where mechanical ventilation demand exceeded capacity by far. As a result, death tolls rose dramatically in these regions.[9] Case reports from the U.S. show that despite concerns of the SCCM, comparable systems had been used at the height of the Pandemic in spring 2020[4, 6, 9]. Governor Cuomo of New York explicitly authorised the use of such systems on March 26th, 2020[3]; we do not know how countries with less sophisticated intensive care medicine systems handled this challenge.

In a joint declaration, American and European societies for critical care medicine postulate not to ventilate two patients with a single ventilator[10]. Based on our measurements, we definitely support this recommendation - if characteristics of respiratory system parameters change, this will result in hyperinflation of the better lungs, and hypoventilation of the worse. Also, patients cannot be allowed to ventilate spontaneously when flow limiters are applied, as the patient on the limited side would be in a situation of acute airway obstruction thus impairing any weaning attempts. Further, no proper alarm management is possible if the ventilator's safety features have to be overridden before this strategy could work. Also, patients would have to be physically positioned very next to each other thus facilitating exchange of aerosols and cross infections. Accordingly, such a setting is only imaginable in a desperate situation for a limited time after emergency intubation in deeply sedated patients for bridging until a second ventilator becomes available[11, 12]. Also, a self-printed adapter is not in accordance with the E.U. or any other guidelines for medical products. However, in this model our hypothesis could be examined without any risk to patients. In contrast, this laboratory model is even more precise than any human study or animal experiment. Thus, this model always poses a risk to detect statistically significant differences due to its exactness that is clinically merely not relevant. Last, all results cannot be extrapolated to other ventilators.

Conclusions

Overriding a modern intensive care ventilator's safety features is possible, thereby ventilating two lungs simultaneously with one ventilator in a laboratory simulation using 3D-printed y-adapters. Directing tidal volumes in different pulmonary conditions towards one lung using 3D-printed flow limiters with diameters <6mm was also possible. While this ventilation setting was technically feasible in a bench model, it would be volatile, if not dangerous in a clinical situation.

Abbreviations

ARDS - adult respiratory distress syndrome

PEEP - positive end-expiratory pressure

Declarations

Ethics approval

Since this was a completely technical simulation with no participants, no ethical approval was required.

Consent to participate

Not applicable.

Consent for publication

Not applicable.

Availability of data and materials

All data generated used or analyzed during this study are included in the published article.

Competing interests

The authors declare that they do not have any conflict of interest.

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Author contributions

All authors contributed to the study conception. The first protocol was designed by Pascal Schepat, Volker Wenzel and Holger Herff. Material preparation, data collection and analysis were performed by Pascal Schepat, Benjamin Kober, Martin Eble and Holger Herff. The first draft of the manuscript was written by Pascal Schepat and Holger Herff and all authors commented on previous versions of the manuscript. All authors read and approved the final manuscript. This manuscript contains significant parts of the doctoral thesis of P. Schepat, MD.

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References

1. Ranney ML, Griffeth V, Jha AK: **Critical Supply Shortages - The Need for Ventilators and Personal Protective Equipment during the Covid-19 Pandemic**. *N Engl J Med* 2020, **382**(18):e41.
2. Neyman G, Irvin CB: **A single ventilator for multiple simulated patients to meet disaster surge**. *Acad Emerg Med* 2006, **13**(11):1246-1249.
3. **One ventilator, two patients: New York hospitals shift to crisis mode** [https://www.reuters.com/article/us-health-coronavirus-usa-ventilators/one-ventilator-two-patients-new-york-hospitals-shift-to-crisis-mode-idUSKBN21D3M1]
4. Rodriguez-Villar S: **Sharing a single ventilator ("In vitro")**. *Med Intensiva* 2020.
5. Lai BK, Erian JL, Pew SH, Eckmann MS: **Emergency Open-source Three-dimensional Printable Ventilator Circuit Splitter and Flow Regulator during the COVID-19 Pandemic**. *Anesthesiology* 2020, **133**(1):246-248.
6. Levin MA, Shah A, Shah R, Kane E, Zhou G, Eisenkraft JB, Chen MD, Mount Sinai HIG: **Differential Ventilation Using Flow Control Valves as a Potential Bridge to Full Ventilatory Support during the COVID-19 Crisis**. *Anesthesiology* 2020, **133**(4):892-904.
7. Clarke AL, Stephens AF, Liao S, Byrne TJ, Gregory SD: **Coping with COVID-19: ventilator splitting with differential driving pressures using standard hospital equipment**. *Anaesthesia* 2020, **75**(7):872-880.
8. Herrmann J, Fonseca da Cruz A, Hawley ML, Branson RD, Kaczka DW: **Shared Ventilation in the Era of COVID-19: A Theoretical Consideration of the Dangers and Potential Solutions**. *Respir Care* 2020, **65**(7):932-945.
9. Briegel J: **Virusinfektion und fehlende Beatmungsgeräte – ein Déjà-vu?***Anesthesist* 2020, **69**:314-315.
10. **Consensus Statement on Multiple Patients Per Ventilator** [https://www.sccm.org/Disaster/Joint-Statement-on-Multiple-Patients-Per-Ventilato]
11. Notz Q, Herrmann J, Stumpner J, Schmid B, Schlesinger T, Kredel M, Kranke P, Meybohm P, Lotz C: **[Anesthesia and intensive care ventilators: differences and usability in COVID-19 patients]**. *Anaesthesist* 2020, **69**(5):316-322.
12. Bohm SH, Kremeier P, Tusman G, Reuter DA, Pulletz S: **[Volumetric capnography for analysis and optimization of ventilation and gas exchange]**. *Anaesthesist* 2020, **69**(5):361-370.

Figures

Figure 1A.

Figure 1B.

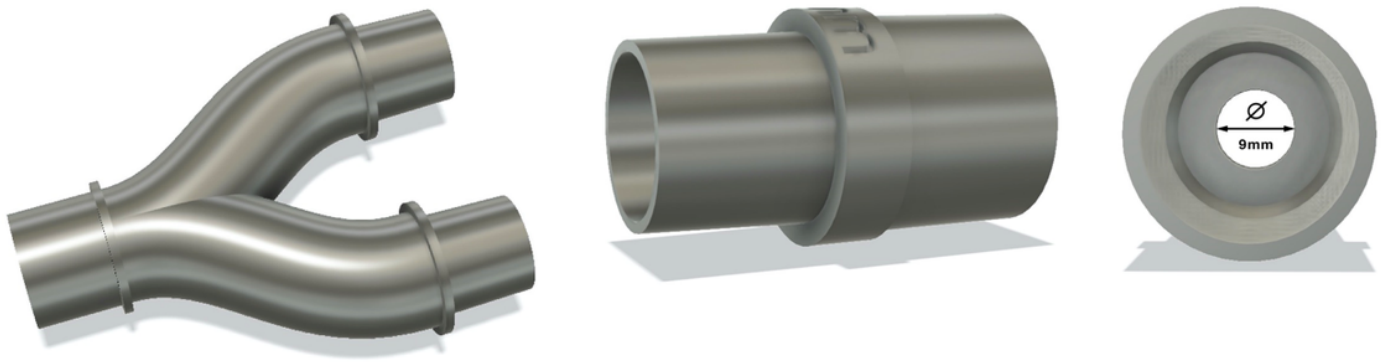

Figure 1

A Draft of a 3D printable y-adapter can be used to divert gas flow to two patients according to ISO 5356-1. B Draft of a 3D printable "Flow limiter "created in Autodesk Fusion 360 according to ISO 5356-1: variable inside diameter from 9 – 2mm.

Figure 2.

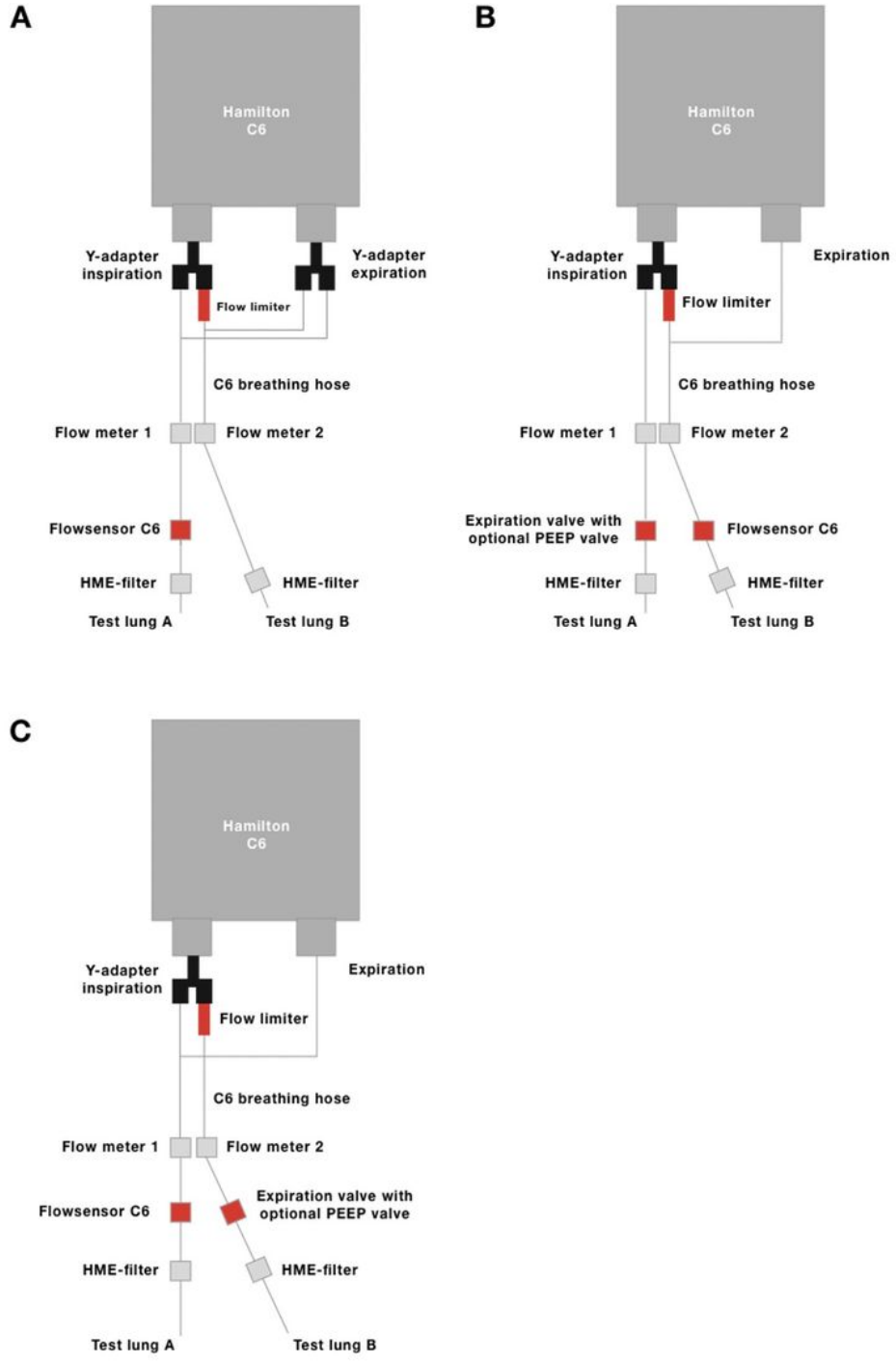

Figure 2

Illustration of a Hamilton C6 ventilator connected to two test lungs; Figure 2A ventilation of two test lungs with a y-adapter in the in- and expiratory ventilator thigh over two double ventilation hoses. The flow limiter is placed in circuit B, the flow sensor in circuit B; Figure 2B ventilation of two test lungs with only one y-adapter in the inspiration thigh. The flow limiter is placed in circuit B. Expiration in circuit A over a one-way expiration valve with an optional PEEP valve. The flow sensor is placed in circuit B (double

hose); Figure 2C ventilation of two test lungs with y-adapter in the inspiration thigh. Expiration in circuit A over a one-way expiration valve with an optional PEEP valve. The flow sensor is placed in circuit A. Circuit B double hose.

Supplementary Files

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