G. Smit’s Ventilator Project “Inspiration”
Delft, The Netherlands, compiled April 24, 2020

Experimental Evaluation of the “Project Inspiration” Rapidly Manufactured Ventilator

(Draft Report, Version 1.5)

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***Abstract*—This document details experimental evaluation of the second prototype of ”Project Inspiration”, in terms of pressure and volume curves, and in terms of reliability of information that can be read from the electronic sensors and also in a manual fashion, without electronics.**

I. Introduction

In response to the Corona crisis, G. Smit from the TU Delft launched a project to (re-)build a ventilator that consists of simple mechanical components. He collected the East-Radcliffe ventilator (Fig.1) from the Boerhaven museum in Leiden and disassembled it.

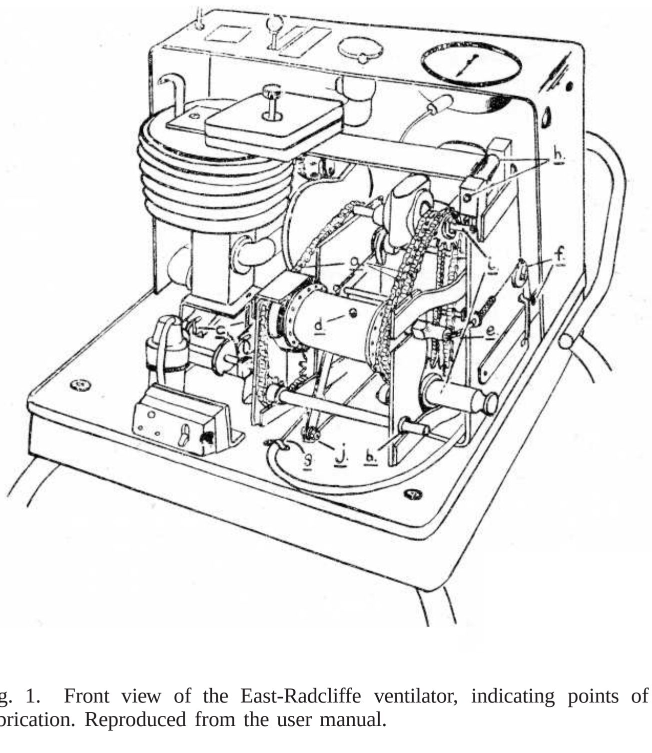

Fig. 1. Front view of the East-Radcliffe ventilator, indicating points of lubrication. Reproduced from the user manual.

Pressure is controlled by modifying the number and location of weights pushing down on a bellow.

The device was first replicated close to the original design, with a few improvements. Then, a series of the second prototype was realized, with further improvements and a sturdy frame.

This document deatils the experimental evaluation of the second prototype. First, the machine was submitted to two systematic evaluation experiments, covering a grid of operating conditions. Both experiments were designed to characterize pressure, flow, and tidal volume, as functions of breathing rate, set peep, and number of weights.

Several more experiments were conducted to assess sensing of plateau pressure, the information content of the motor current signal, and the effectiveness of a hardware design choice to avoid pressure overshoot.

This document first describes the experimental protocols, and then the results.

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II. Working Principle

An arm with an adjustable number and location of weights on it drives a bellow and thereby generates pressure and volume flow to the patient’s lung.

A single DC motor (PM11S, Parvalux, UK), which runs at user-controlled speed, drives a) a main cam that moves the arm up during the expiration phase, and b) two smaller cams that control valve opening and closing. All cams are mounted to the same shaft and thereby synchronized.

Peep is controlled manually, via a disposable peep valve at the outlet.

Due to the mechanical working principle, several variables can be read without electronics, in particular pressure by the number of weights and their location on the arm, and supplied volume by the angle of the arm. Manual scales for pressure and volume are integrated in the system and calibrated agains electronic sensors. In principle, also breathing rate can be assessed by the movement of the arm in combination with a stopwatch.

An additional humidifier system controls temperature and humidity of the air.

A separate monitoring system supervises pressure, flow, temperature, humidity and oxygen content.

Details on device design are documented in the reports on the individual components.

III. Experimental Protocol

First, two systematic experiments were conducted, covering a grid. Compared to the first set, the second evaluates a different type of peep valve, as well as a new simple sensor for volume. Further experiments assessed plateau pressure sensing, motor current information content, and effectiveness of a hardware measure to reduce pressure overshoot.

A. First set of grid experiments

In the first experiment, an adjustable peep valve from HUDSON RCI was used (see details in A).

The test lung used was from B&B Medical Technologies, type 1 l, P/N 20110.

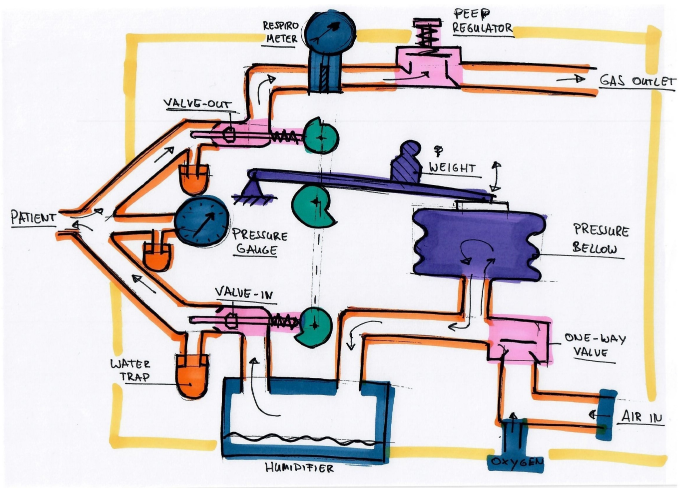

Fig. 2. Working principle of the ventilator.

The first test grid was designed in the following way: Seven different peep settings were investigated, from 5 to 19 mbar.

For each peep setting, five different weight settings were evaluated, namely 2 to 6 pieces of weight (whereby 1 piece describes the situation where only the carrier is on the arm, 2 pieces describes the situation where one weight element is placed *on* the carrier). The weights were always placed in the configuration with maximum lever arm, which means that the carrier touched the screw head.

For each weight setting, three different breathing rates were set by manually adjusting the dial. This adjustment was not done with high precision, allowing the breathing rate to vary by about 1 rpm.

Each trial was run for at least four cycles of the machine.

That led to a total of 7 times 5 times 3, so 105 data files.

All experiments were conducted with the machine that has number 1.

B. Second set of grid experiments

In the second set of experiments, a different peep valve was used, namely from Intersurgical (see details in A).

The same test lung was used as in experiment 1.

The second grid was designed in the following way: Four different peep settings were investigated, from 5 to 20 mbar.

For each peep setting, again five different weight settings were evaluated, namely 2 to 6 pieces of weight.

In contrast to the setting in experiment 1, where all weights were placed at the maximum lever position $L_{\max}$=19.7 cm, the weights' centers were now located about 4 cm closer to the arm's hinge

For each weight setting, again three different breathing rates were set.

Each trial was run for at least four cycles of the machine.

That led to a total of 4 times 5 times 3, so 60 data files.

All experiments were conducted with the machine that has number 1.

C. Experiment on plateau pressure

Modern ventilation machines briefly halt the flow at the end of the inspiration phase, in order to measure the pressure at zero flow. Since the mechanical machine considered here cannot hold the flow, the pressure in the alveoli needs to be measured in a different way.

Therefore, we performed an experiment to determine whether the driving pressure can safely be assumed to be equal to the pressure in the patient's alveoli at the end of the inhale phase, which means that pressure equilibrium is reached.

For this experiment, four weights were placed on the arm at maximal distance. The flow sensor was now not put in the exhale stream, but at the lung side, such that it measured both inhaled and exhaled flow. Breathing rate was set to 10, 20 and 30 breaths per minute.

This way, it was possible to investigate whether the flow would reach zero before the end of the inhale phase.

Particularly the high breathing rate is of interest, because this can be considered a worst case in terms of reaching equilibrium.

This experiments was conducted with the machine that has number 1. The test lung was the same as above. The peep valve used was the one from Hudson.

D. Experiment on motor current

A third set of experiments was conducted to assess the information content of motor current, as measured by the drive (Maxon, Escon 50/5). The goal was to detect mainly the impact of the follower on the cam, because the timing of this event with respect to other events can be used to assess the angle of the arm.

This is possible because for a given breathing rate, the motor angle is a function of time, and the radius of the cam is a monotonically increasing function of angle within the considered range. The underlying assumptions are a) breathing rate is constant over the course of the cycle, and b) motor speed tracking is highly precise. In order to be robust against violation of the two assumptions, a) averaging over multiple cycles should be performed, and b) volume should be measured for a range of breathing rates and pressures. This has not yet been performed.

In the experiments, 2 to 6 weights were placed at the maximal arm length, at breathing rates of 10 and 40 breaths per minute, respectively. Subsequently, four weights were placed and tested at an intermediate breathing rate of 21.3 breaths per minute and different peep settings.

From the resulting graphs, the motor current was inspected, and the visibility of certain events (opening and closing of valves, impact of follower and cam) verified.

This experiments was conducted with the machine that has number 3, which also has already undergone more than a week of endurance test. The test lung was the same as above. The peep valve used was the one from Hudson.

E. Experiment on the influence of an orifice plate on overshoot

In pilot experiments, we had noticed that pressure tended to overshoot at the beginning of inspiration, due to passive dynamics resulting from inertia of the arm and weights in combination with the compliance of the air in the system.

Therefore, we added a small printed plate with orifices between the bellow and the inlet valve. The orifice has 10 holes of 3 mm diameter each. The purpose is to generate damping for high flow rates, and thereby reduce the overshoot.

The orifice plate is needed in the path between bellow and inlet valve. The first hardware realization consists of a small printed component that is to be inserted into the tube connector below the bellow.

One risk of this component is that it could be forgotten at assembly or after cleaning the machine. To avoid this, the component is now placed at the bottom of the bellow, integrated into the laser-cut bottom plate of the bellow. For easier manufacturing, there are now only 9 orifices of 3 mm diameter each.

If the orifice plate is placed directly at the bottom of the bellow, however, it might also affect flow at the inlet of the bellow. This flow is slow (given by slow upward movement the cam), this effect is expected to be small. Remark: In order to quantify the effect, in pilot experiments we also compared the situation where an additional plate with orifices is placed in a tube at the bellow inlet, in order to assess any potential difference in volume. We did not observe any difference.

In pilot experiments, we further saw that the overshoot is highest at high breathing rates and high pressures. Therefore, we recorded at 30 breaths per minute, with four weights.

Note that all experiments described in previous sections included an orifice plate already between bellow and inlet valve.

IV. Data acquisition and processing

A. Data acquisition

Data was acquired using custom electronics, and sent via serial communication (USB) to a Windows laptop, where they were recorded. The sampling rate was approximately 1 kHz without motor current measurements, and approximately 500 Hz when including motor current. Time stamps were also sent and recorded, to control for jitter.

The electronics included a pressure sensor (Differential Pressure Sensor: NXP MPXV7007DP) and a flow sensor (SFM3000-200C, Sensirion). The pressure was measured at the machine side of the bacteria filter, against atmosphere. The flow sensor was placed at the outlet valve, before the peep.

For each trial, the pressure and the flow were recorded. Hereby, the pressure sensor was mounted at the filter element, so close to the patient, but on the machine side. In all experiments except the one on plateau pressure, the flow sensor was mounted at the outlet valve, so measuring exhaled flow.

Only in the second grid experiment, a new “sensor” was available, namely a disk with an angle grid on it, to enable reading the angle of the weight-carrying arm with a precision of around $2.5°$. This angle was read and noted down manually for each setting. We expected a relationship between this angle and the tidal volume.

In the experiment on motor current, the custom electronics and code were modified to also receive an analog signal from the motor drive (Maxon Escon 50/5), which provided the information on current. This signal was already filtered by the Escon 50/5.

B. Processing of each file

The following features were extracted from sensor signals for each cycle:

- Volume was calculated via integration of flow over time, making use of the time stamps of the recorded data.
- Plateau pressure for each cycle was extracted from the pressure curves by filtering the signal back and forth, each time with a first-order filter with a cutoff frequency of 5 Hz. This procedure removes overshoot and allows extracting the plateau as the maximum value of the filtered signal.
- Peep pressure for each cycle was calculated by 1. filtering the raw pressure signal only weakly, using a second-order Butterworth filter with cutoff frequency of 50 Hz, again applied backward and forward, 2. multiplying the resulting signal with -1, and 3. finding peaks in the resulting signal, which correspond to valleys in the original signal.

For each file in the two grids, the data was further condensed for further analysis: Over the number of cycles recorded, the median was stored for volume, plateau pressure, and peep pressure.

The maximal value of the raw pressure signal was also stored, to assess overshoot.

C. Relationships between signals in the grids

To investigate relationships between signals across the different settings, the grid data was further processed in the following way:

First, for each file, it was checked that 1. the measured peep was above 5 mbar, 2. the plateau pressure was below 35 mbar, and 3. the overshoot, which was calculated as the difference between maximum and plateau pressure, was below 2 mbar.

A main purpose of this characterization is to establish a table for operators of the machine, such that they can choose the correct number of weights for a certain desired combination of breathing rate, volume, and peep setting.

Then, relationships were sought between volume and arm angle, between plateau pressure and number of weights, and between measured and set peep. Lines were fit for all signals.

V. Results

A. Device characteristics

Figure 3 shows some exemplary data of the second experiment.

Figure 4 shows, for each breathing rate, an interpolated 3D surface and a contour plot for the first experiment. The volume is shown as a function of different peep and weight settings. The contour also contains a convex hull of the black points in the 3D plots, which are the ones that fulfill the UK specs (accessed version April 4) in terms of plateau pressure, overshoot, and peep pressure. The contour can be seen as a graphical representation of a table that can be

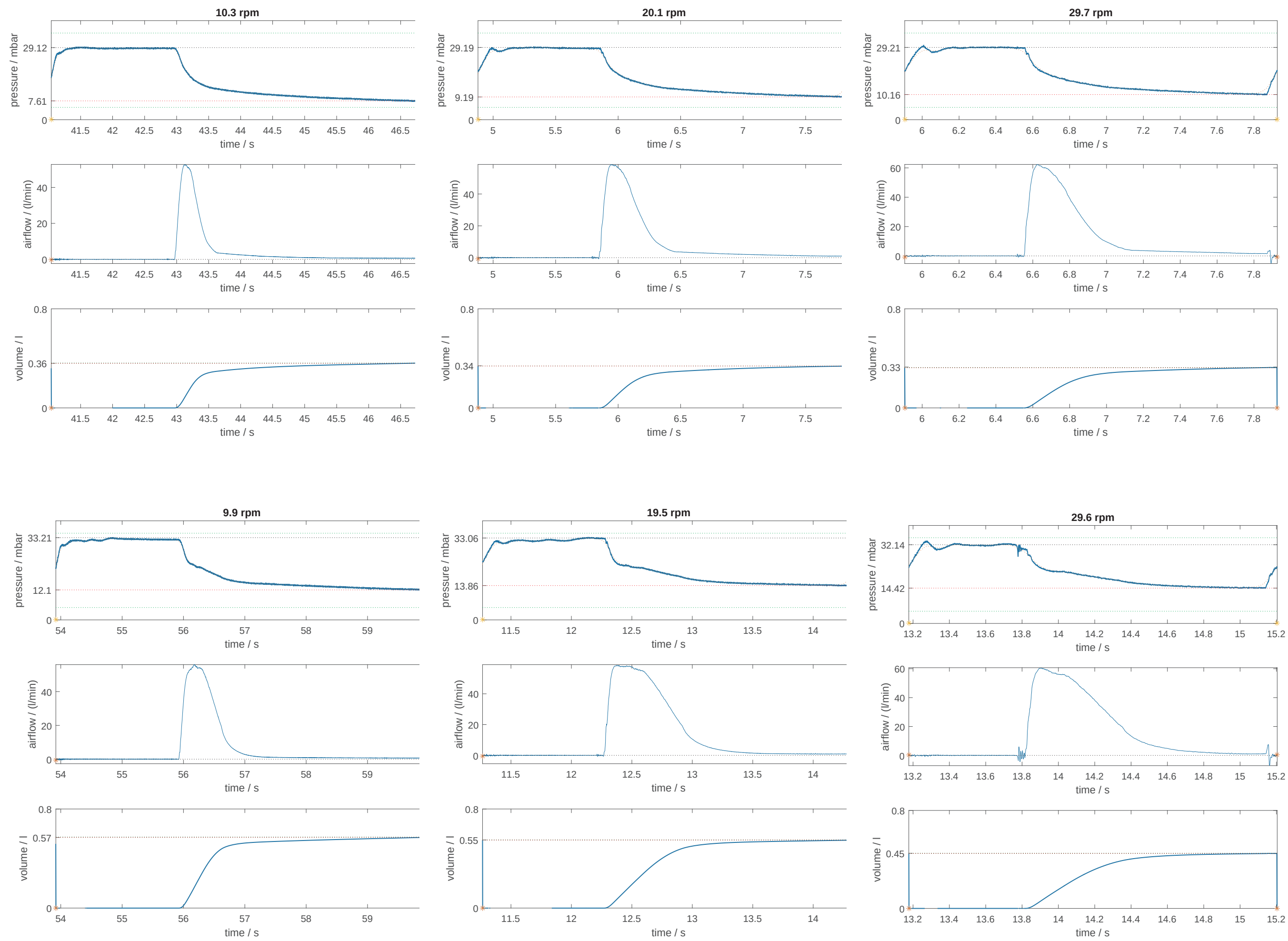

Fig. 3. Exemplary results for pressure, flow and volume, for different settings of peep and weight. Signals are not filtered.

extracted for users. Such a table has already been extracted as well.

Figure 5 shows the equivalent data of the second experiment.

It must be noted that the peep shown on the axes is the one that was set, not the one that was measured. In the case of the second experiment, the measured peep was always lower than the set peep.

B. *Relationships between signals*

Figure 6 shows the relationships between set peep and actually measured peep (left), between number of weights and plateau pressure (center), and volume and angle of the arm (right, only available for experiment 2). Note that the weights were placed closer to the arm's hinge in experiment 2, leading to lower pressure per weight.

Note that number of weights is counted starting already at the weight of the carrier. So, a setting of two weights means that the user has placed one loose weight on the carrier. See B for further calculations regarding the user interface.

From the second grid, the fitted relationship between volume V and arm angle φ was:

$$V = m\varphi + c, \tag{1}$$

with $m = 0.0485 \cdot \mathrm{l/deg}$ and $c = -0.1062 \cdot \mathrm{l}$.

For the first experiment, the measurement errors are: mean error between set and measured peep, and std of this error: 0.51 mbar $\pm$ 1 mbar, std of manual plateau pressure reading in the fit: $1.1 \cdot \mathrm{l}$.

For the second experiment, these values are: mean error and std of manual peep setting:4.82 mbar$\pm$ 2.16 mbar, std of the fit error on manual volume reading:$\pm$ 0.02 l, std of the fit error on manual plateau pressure reading:$\pm$ 0.28 mbar.

An observation that existed only in the first experiment was high-frequent noise (around 144 Hz) in the pressure and flow signal, though these only occurred for very high peep values and a high number of weights and plateau pressures. The sound was also audible.

In the second experiment, with the different peep valve, no such oscillations or sound were observed.

C. *Plateau pressure*

Figure 7 shows that for the high breathing rate, the flow does not reach zero before the inlet valve is closed (shortly before 11.5 s). Note that definition was inverted here with respect to the other experimental conditions, such that a positive flow was measured during the inhale phase.

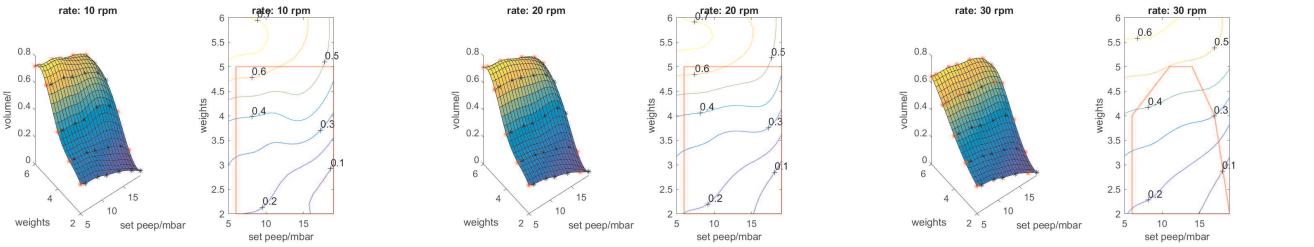

Fig. 4. Volume as a function of set peep and number of weights, for the first experiment and 10,20,30 breaths per minute. Left: Surface plot, red points do not fulfill the specs. Right: Contour and convex hull of all allowed points.

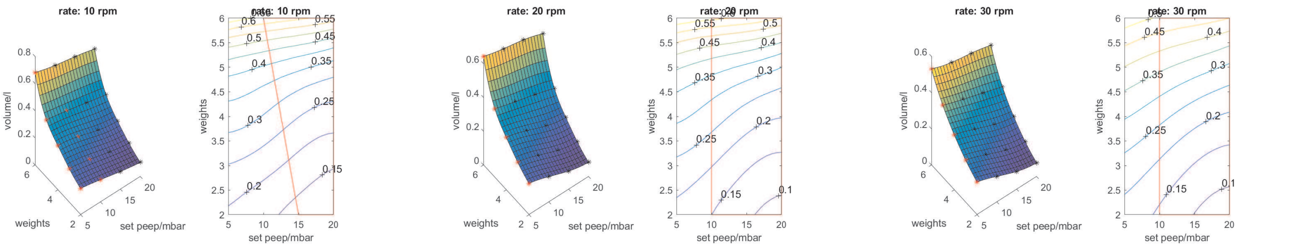

Fig. 5. Volume as a function of set peep and number of weights, for the second experiment and 10,20,30 breaths per minute. Left: Surface plot, red points do not fulfill the specs. Right: Contour and convex hull of all allowed points.

Nonzero flow means that the pressure before the closing of the valve is not yet an equilibrium pressure, which means it is not equal to the pressure in the alveoli. When the inlet valve is closed (and the outlet valve is not yet open), also a temporary drop in pressure can be observed.

D. *Information content of motor current*

The motor current signal, as shown in Figure 8, contains the information sought for: The increase in motor current indicates an impact between follower and cam. Further events are also detectable and indicated in the same figure.

E. *Influence of orifice plate on overshoot*

Figure 9 shows that the added orifice plate eliminates the overshoot in pressure that is present otherwise.

Note that in the experiment without orifice plate, the pressure sensor was located closer to the inlet valve than in the experiment with orifice plate, where the sensor was located directly at the filter. In earlier experiments, it was verified that this did not have an observable effect on the height of the first overshoot.

VI. Discussion

A. *Device characteristics*

The device achieves an almost ideal block curve for pressure, which is necessary for maximal volume without exceeding peak pressure also for high peep values. Overshoot is minimal. Note that the machine's mechanical feature that enables this is a small piece with orifices after the bellow outlet, to add damping at high air speeds only, without restricting low-frequent flow and thereby volume.

Specifically, the following specific UK requirements were checked and compared to the device characteristics in the above-described experiments:

- Plateau pressure should adapt to achieve volume and be limited to 35 cmH2O: *This was achieved*.
- Peak pressure should be no more than 2 cmH2O greater than plateau pressure: *This was achieved within the range of allowed plateau pressure values*.
- Peep range 5 to 25 cm H2O adjustable in 5 cmH2O increment: *This is determined by the peep valve used. Both peep valves have a range of 20 cm H2O, and the second peep valve seems to set a lower peep than indicated. This component may need to be exchanged to set higher peep values.*
- Patient breathing system must remain pressurised to at least the PEEP level setting at all times: *This was achieved within the allowed peep settings*.
- Inspiratory:Expiratory ratio (I:E) 2.0 (i.e. expiration lasts twice as long as inspiration): *This was achieved, it is mechanically fixed. Note that the optional requirement of different ratios cannot be achieved without hardware changes and/or variation of motor speed within one cycle, which is currently not possible.*
- Respiratory Rate. The number of breathing cycles every minute: Range 10 to 30 breaths per minute in increments of 2 (only in mandatory mode) can be set by the user: *This was achieved, and adjustment is continuous, not only in discrete increments*.
- Tidal Volume (Vt). The volume of gas flowing into the lungs during one inspiratory cycle must have at least one setting of 400ml +/- 10 ml: *This tidal volume was achieved. To set volume, the machine requires setting pressure*.
- Tidal Volume ideally has 350ml and 450 ml options: *This was achieved as well, and adjustment is continuous by means of varying number or location of weights, not only in discrete increments*.

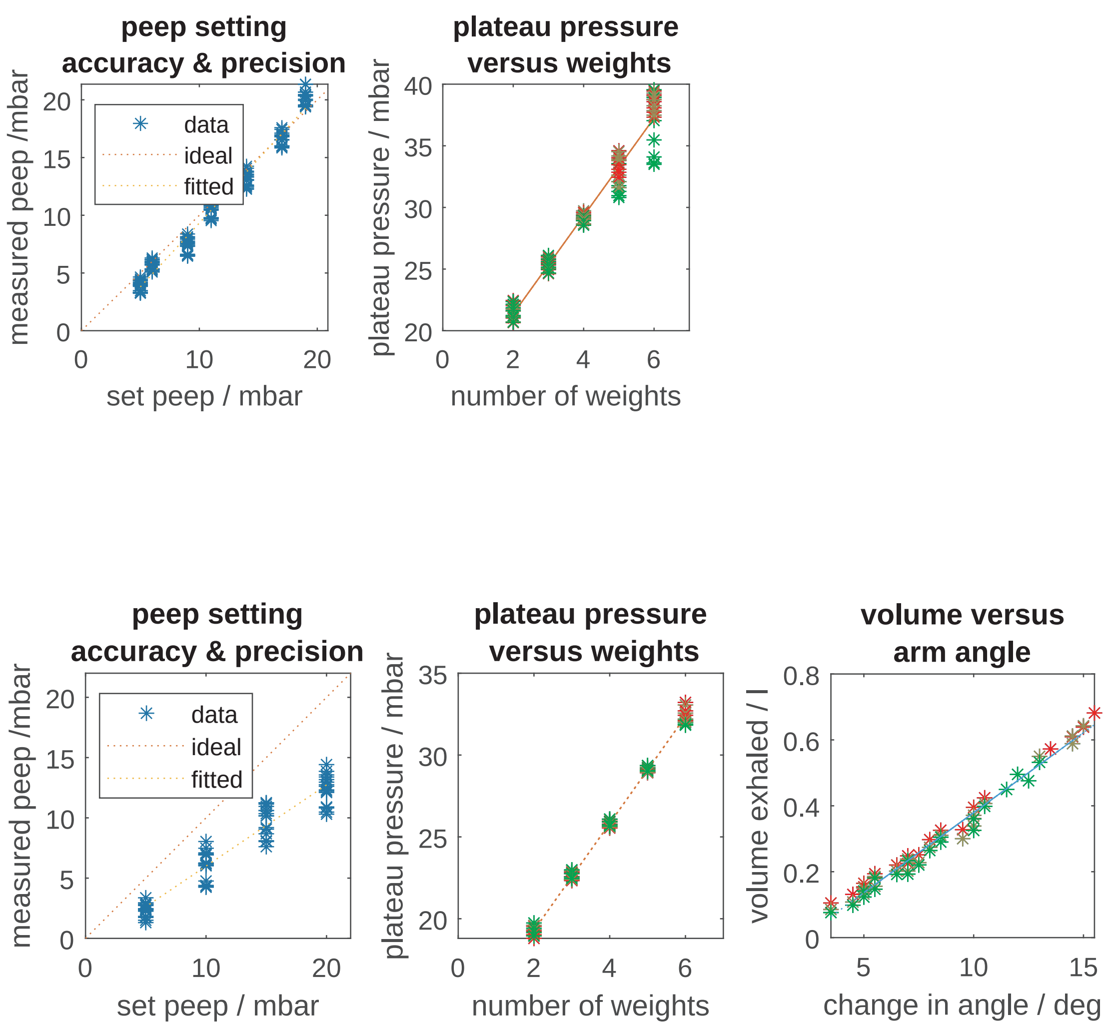

Fig. 6. Relationships between mechanically available information and sensor signals. Top: First experiment. Bottom: Second experiment. Red: low breathing rates, green: high breathing rates. Note that the weights were placed closer to the hinge in the second experiment.

- Tidal Volume optionally has range 250 to 600 ml in steps of 50ml: *The machine did easily achieve lower volumes such as 250 ml. 600 ml were achieved as well, but only for low-to-medium breathing rates. Variation is continuous, not discrete..*

B. Peep valve

The second peep valve (green, from Intersurgical) does not cause noise like the first (blue, HUDSON RCI). However, it does have a lower accuracy and precision. Both differences may be explained to some extent by higher dry friction in the Intersurgical valve. The difference in scaling could also be explained by a difference in peep definition between the two manufacturers. If the Intersurgical valve is to be used, readings may need to be re-calibrated.

The oscillations might be explicable by resonance involving the HUDSON peep valve and the specific test lung.

The difference in scaling needs to be kept in mind when interpreting Figure 5: The peep set at the valve is used for the axes, while the actual peep measured by the machine is much lower in the second experiment. So, the axis ticks should in fact be shifted.

C. Mechanical volume sensor

The angle sensor at the arm provides a surprisingly high precision in estimating volume, assuming proper calibration. This has two advantages: 1. Angle sensing could provide redundancy of volume calculation. 2. Angle sensing could replace volume calculation in settings where there is shortage of electrical components, in particular flow sensors.

The sensing of the angle could continue to be done manually, but it could also be done with an encoder (optical or magnetic, to avoid wear) at the joint, enabling automated processing and comparison with flow-calculated volume. It could even be done by only measuring motor current. Preliminary

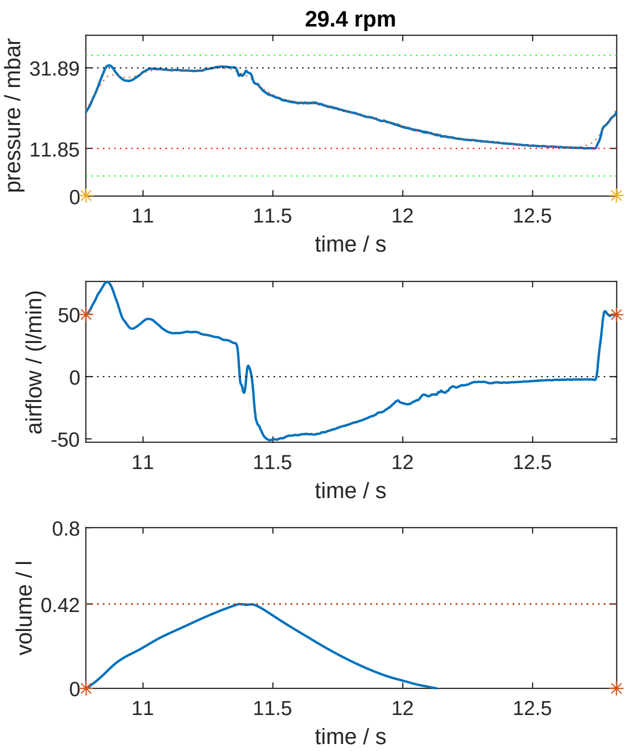

Fig. 7. Curves resulting from the experiment on plateau pressure. For this relatively high pressure and breathing rate, the flow in the inspiratory phase does not reach zero before the inlet valve is closed.

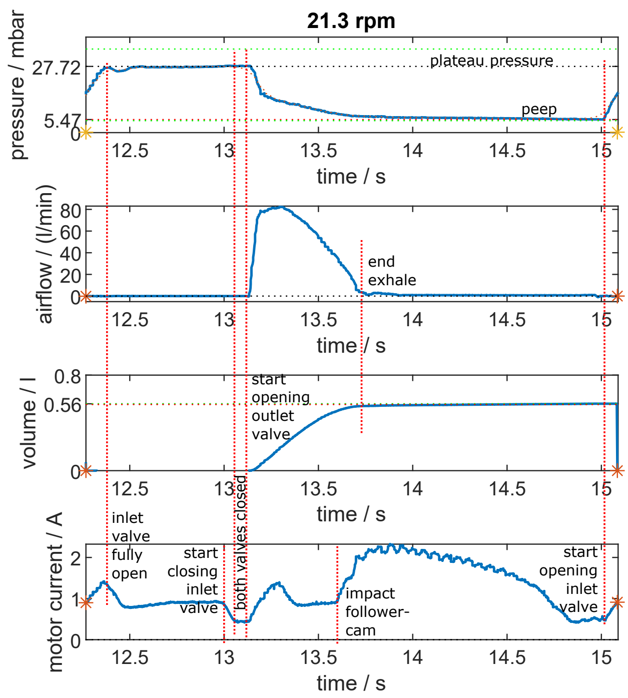

Fig. 8. Curves resulting from the experiment addressing motor current. The different events are clearly visible, including follower-cam impact.

testing shows that the instant of cam-follower contact is measurable in motor current, as are other instances such as valve opening and closing. Because the cam has a monotonically increasing radius as a function of angle, and the motor speed is constant, this timing information allows determining arm angle and thereby volume.

D. *Mechanical pressure sensor*

Plateau pressure can be predicted from the number of weights and moment arm. Whether the resolution and accuracy is clinically acceptable remains to be investigated.

E. *Plateau pressure*

For lower breathing rates and driving pressures, the inhale flow goes to zero before the inlet valve closes. Therefore, we assume that the pressure shortly before that closing is equal to the pressure in the alveoli.

However, for higher pressures and breathing rates, this assumption does not hold. Possibly, the pressure measured during the short interval where both valves are closed can be used as an estimate for pressure in the alveoli.

This remains to be investigated.

F. *Information content of motor current*

The motor current seems a valuable source of information. Not only could it help identify increased wear and friction, it also seems suitable to estimate inhaled volume, based on the arm angle at impact between follower and cam.

The precise relationship between relative timing of events and inhaled volume remains to be quantified in calibration experiments.

The signal could also be used to identify the short time window where both valves are closed, where the pressure is closer to the pressure in the alveoli, the to-be-measured plateau pressure.

In principle, motor current could also be used to control the motor in such a way that the expiration phase is extended. The expiration phase can be well-detected as a period of prolonged increase in motor current, because the arm is moved up. If motor speed is selectively reduced in this phase, the inspiration-expiration ratio could be set to values such as 1:3 or 1:4, possibly even without requiring an encoder, and especially without connection to the monitoring system. Therefore, this could be done autonomously in the drive.

G. *Limitations of these experiments*

A limitation of this analysis is that between all experiments, the device was transported from one room to another, and also all tubes were decoupled and re-coupled, and some tubes were exchanged with others.

A further limitation is that all measurements were done against a single test lung. Behavior may change in interaction with a real, damaged lung.

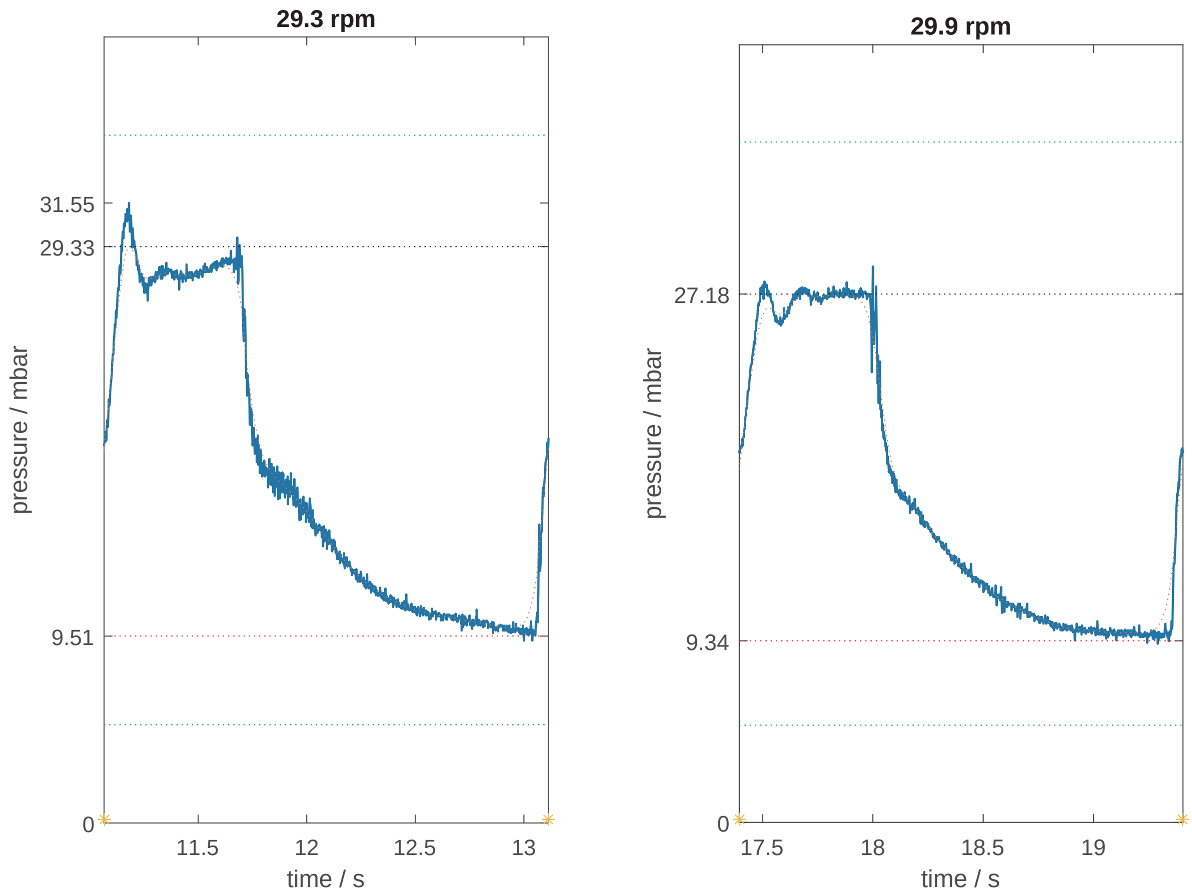

Fig. 9. Curves resulting from the experiment on the influence of an orifice plate. Left: No orifice plate. Right: With orifice plate, at bottom of bellow. It can be seen that the added orifice plate eliminates the otherwise observed overshoot in pressure. Note that breathing rates do not match exactly.

In reality, the pressure sensor might be placed less close to the lung. It remains to be investigated what the effect is of moving it. Preliminary (not systematic) tests indicate that that influence is very small. However, the norm (NEN-EN-ISO 80601-2-, Medical electrical equipment - Part 2-12) indicates that the pressure sensor is either placed close to the patient, or that two pressure sensors are placed, one in the inhale side and one in the exhale side.

Finally, a few machine components were omitted when performing these experiments, for simplicity of setup, namely the humidifier and the water traps.

Note that the open-source release of the system is a newer version, which has an updated setup for the weights (fewer weights and shorter travel along the arm). The main functional parts are the same, such that behavior is not expected to be different. However, the characterization experiment is still to be repeated with the final setup.

VII. Conclusion

The device is within the UK specs listed above, but performance remains sensitive to changes in components, such as a different peep valve. The experiments need to be repeated with the final configuration.

It is also recommended to install an angle sensor at the arm's joint, ideally both in form of an encoder and in form of a manual gauge, as in experiment 2 described above.

It is further recommended to include the motor current in the monitoring system, as it contains rich information that can be used to improve sensing and for fault detection.

Appendix A
Peep valves used

Both peep valves used in this experiment are adjustable.

The first (blue) PEEP valve is of the brand HUDSON RCI, distributed by Teleflex.

The second (green) PEEP valve is from Intersurgical (https://www.intersurgical.com/products/anaesthesia/mapleson-a-and-c-breathing-systems). It is not delivered separately, but taken from a Mapleson C ventilation system.

Figure 10 shows the two peep valves.

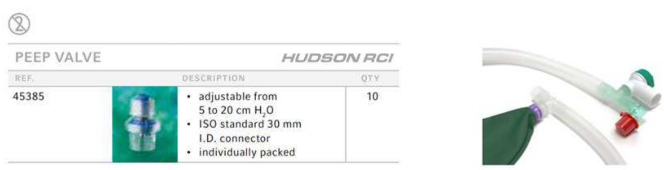

Fig. 10. Peep valves used. Left: Blue valve, used in experiment 1. Right: Green valve, used in experiment 2.

Appendix B
User interface for the weights

The weights can be adjusted to vary pressure in a linear fashion, as can be seen in Figure 6.

The precision of the measurement suggests to insert at least one or two additional steps (by shifting the weights along the lever) between the discrete steps of adding weights.

For a lever length of L and N weights of mass m, the moment is given by

$$\tau = Lg(mN + c), \tag{2}$$

whereby c is additional weight of the carrier structure, so three pins and the necessary screws, bolts, and washers. The main element of the carrier is counted as the first weight.

In order for a higher number of weights $N + 1$ at $L_{\min}$ to be equivalent to one weight less, but at a lever of $L_{\max}$, the torques must be equal:

$$L_{\max} g(mN + c) = L_{\min} g(m(N + 1) + c) \tag{3}$$

That means that the relationship of lever arms must be:

$$L_{\min} = \frac{Nm + c}{(N + 1)m + c} L_{\max}. \tag{4}$$

so the distance Δ to be shifted can be calculated from:

$$\Delta = L_{\max} - L_{\min} = \frac{m}{(N + 1)m + c} L_{\max} \tag{5}$$

So, this distance varies with the number of weights placed.

A reasonable number of weights on the machine at maximal lever length, given the experimens, is larger or equal to 3 and lower or equal to 6 weights, mostly likely it will mostly be around 4 weights, to achieve a volume around 400 ml. We therefore only consider N in the range from 3 to 5.

For these settings, neglecting c, the shift range Δ would be approximately $\frac{1}{4}$, $\frac{1}{5}$, and $\frac{1}{6}$ of the maximal length $L_{\max}$, respectively for the carrier carrying $L+1 = 4$, 5, or 6 weights.

The exact relationships, considering c as well, are:

- $\Delta = \frac{m}{4m+c} L_{\max}$ for $N + 1 = 4$ weights,
- $\Delta = \frac{m}{5m+c} L_{\max}$ for $N + 1 = 5$ weights, and
- $\Delta = \frac{m}{6m+c} L_{\max}$ for $N + 1 = 6$ weights

In the investigated prototype, the lever length that is maximally set is approximately $L_{\max} = 19.7$ cm.

One weight of the type used in the experiments has a mass of approximately $m = 0.74$ kg. The additional carrier mass is $c \approx 0.152$ kg.

Substituting these numbers above, the range of lever lengths needs to be:

- $\Delta = 4.7$ cm for 4 weights,
- $\Delta = 3.8$ cm for 5 weights, and
- $\Delta = 3.2$ cm for 6 weights

It may be noted that the influence of a variation of c is small, as completely neglecting c only changes Δ by less than two millimeters.

It is recommended to have at least one in-between tick on the arm for each weight setting (equivalent to the ability of adding "half" weights), so that leads to 6 ticks minimally. It remains to be investigated how users can best interact with the system.

In all of the above, the weight functioning as the carrier is already counted as the first weight.

When shifting the weight, one needs to pay attention that the carrier does not slide too fast. This must therefore be appropriately handled by either a modification of the adjustment mechanism, procedure, or user training.

A similar concern is that placing additional weight in a dynamic way (dropping the weight in the worst case) could lead to pressure peaks for the user. This could be prevented by only placing weight during the upward movement of the arm. However, also here procedure and proper training are important.

Acknowledgments

This work was only part of a larger initiative headed by G. Smit, with multiple members who realized the setup and helped conduct tests. The custom electronics and embedded software used for these experiments were designed by M. Fritschi.