Within the last decade, lung protective mechanical ventilation during anesthesia has become increasingly important. Nowadays, it is widely accepted that mechanical ventilation settings should be adjusted to a patient’s individual condition in order to avoid ventilator-induced lung injury (VILI). In the absence of outcome-related data, such a physiologically guided approach is of particular interest in pediatric patients. Respiratory monitoring in the anesthesia workstation (AWS) is the basis for respective analyses. The analysis of a single breath already allows the creation of a sophisticated characterization of the patient’s respiratory system’s conditions. In the synopsis of emerging and rapidly evolving technologies, this information opens a potential for lung protective mechanical ventilation.

In order to support the understanding of a rationale-based perioperative ventilation therapy, this educational review aims to explain the clinical impact, technological background, and validity of monitoring variables routinely provided by modern AWS. In a second part, translation of this information into a rationally based perioperative ventilation therapy is discussed.

Core components of modern AWS have changed only marginally over the past decades. Today, an AWS basically contains gas and electricity supply, a ventilator circuit operated either automatically...
(with a piston or a turbine) or manually, and a monitoring unit displaying set and measured ventilation variables (Figure 1).

With regard to the curves displayed on an AWS monitor, it is of note that directly measured signals contain artifacts resulting from external electrical fields, switching of the ventilator’s valves, and limitations of the measurement techniques, for example. For smooth visualization, signals are therefore highly processed. Typical artifacts of high frequency are eliminated using low pass filters and baseline drifts are artificially set to zero before each breath. These examples illustrate how complex the requirements of information processing in the AWS need to be, to allow for user-friendly and safe operation. Such automated processing, however, entails the risk of error propagation in the system.

3 | MONITORING OF RESPIRATORY MECHANICS—VOLUME

Tidal volume is probably the most essential measure directly provided by the AWS. The calculation of a number of secondary parameters is based on this variable.

3.1 | TECHNIQUES OF VOLUME MEASUREMENT

Volume is calculated as the integral of the directly measured flow rate over time. To date, the two most reliable methods of flow measurement in an AWS utilize pneumotachographs or thermal anemometers.

In the first case, the pressure gradient across a defined resistor element in a pipe (the pneumotachograph) is measured via highly sensitive differential pressure sensors. The two pitot tubes of the sensor piece are pneumatically connected to the pressure sensor. Secretion and condensation may change the characteristics of the resistor. Heating and regular calibration of the device are supposed to prevent this. Following calibration (including the compensation of nonlinearities of the real pressure-flow relationship), the pressure gradient is proportional to the flow rate of the gas passing through the resistor. The advantages of this type of sensors lie in a long operation life time as well as in its robustness to vibration and other sources of noise such as electromagnetic interference.

Thermal anemometry works via a wire heated by a defined electrical current inside an airway tube. The wire’s temperature and resulting resistance, change depending on the amount of passing gas. The measurement of flow rate is based on the change in voltage to maintain the defined current or on controlling the wire’s temperature via complex controlling systems. In the range of physiological measurements, the performance of this sensor type is not affected by temperature or gas composition.

3.2 | LOCATION OF VOLUME MEASUREMENT

The pneumatic conditions in the breathing system can be considered quasi-stationary and continuity can be assumed. It results that flow is sufficiently location independent if no leakage is present. A medical ventilator circuit typically contains two flow sensors, one in the inspiratory branch and the other in the expiratory branch of the breathing system (Figure 1). With two sensors in place, leaks can be easily detected by comparison of inspiratory and expiratory volume. Expiratory volume can be slightly lower than inspiratory volume due to the resorption of oxygen whereas the changes in gas temperature and humidity are automatically taken into account.

Of note, some ventilators suited for pediatric patients reduce volume measurement error by utilizing an external flow sensor, positioned close to the patient’s airways, that is, distal to the y-piece. In doing so, the compliance of the breathing circuit (which is about 0.4 ml/cmH2O in neonatal breathing circuits with an inner diameter of 10 mm) does not influence the measurements of the flow rates toward and from the patient. Thus, accuracy of exhaled tidal volume measurement can be increased, particularly in presence of a leaking airway device. Moreover, proximal flow measurements have
a better signal-to-noise ratio and ventilator response to patient’s breathing effort can be better synchronized during assisted spontaneous breathing. However, additional dead space and risk of disconnection may limit the benefits of this method.

3.3 | PRECISION OF TIDAL VOLUME MEASUREMENT

Modern AWS can provide tidal volumes (VT) as low as 5 ml for pressure-controlled ventilation (PCV), and as low as 10 ml for volume-controlled ventilation (VCV). According to manufacturer’s information, the accuracy ranges between ±10% and ±10 ml below a VT of 150 ml, the, respectively, greater value applying. In AWS using a piston pump for volume generation, the delivered volume is directly related to the linear movement of the piston, which provides additional information ensuring precision regarding VT via internal processing. Alternatively, a turbine is used to drive the ventilation, similar to some ICU ventilators. This approach reveals its benefits whenever highly accelerating inspiratory flow is needed, for example, during PCV and when high respiratory rates are required. Another technical achievement to enable accurate titration of very small VT is the consideration of fresh gas supply during application of the set VT. Historically, fresh gas flow continued during the volume delivery, which altered VT depending on the respective flow. Nowadays, fresh gas flow is either taken into account computationally or discontinued during the inspiratory period. Lastly, sensor calibration and compliance of the breathing circuit can impact the precision of volume measurement. These points will be discussed further down.

From a physical point of view, the composition of the breathing gas influences flow measurement and thus precision of VT application. Given a certain temperature and pressure, gases come with a certain viscosity and density. Viscosities of modern volatile anesthetics are lower than those of the typical carrier gases (i.e., mixtures of air and oxygen) whereas their densities are considerably higher (e.g., the density of Sevoflurane is sixfold of that of air). At larger volume fractions, these characteristics can promote turbulent flow and thus influence flow and pressure measurement in particular. Although the net effect can be considered relatively small, AWS with integrated gas measurement units automatically account for correction of gas mixture. However, in AWS without gas measurement, compensation has to be activated manually, if volatile anesthetics are in use.

3.4 | DISPLAY OF FLOW AND VOLUME

The flow rate itself is typically displayed as a time-dependent curve (Figures 2 and 3). Among the most relevant information which can be visually estimated from the flow curve are airway obstruction, detection of incomplete expiration and inspiration, and detection of spontaneous breathing efforts. Moreover, leakage can be easily estimated by comparing the area enclosed by the inspiratory curve (representing inspiratory volume) with that enclosed by the expiratory curve (representing expiratory volume).

4 | MONITORING OF RESPIRATORY MECHANICS—PRESSURE

The second variable directly measured in the breathing circuit is pressure. Airway pressure is usually displayed continuously against time (Figures 2 and 3), together with several related metrics (e.g., peak airway pressure, plateau pressure, and positive end-expiratory pressure (PEEP)).

4.1 | TECHNIQUES OF PRESSURE MEASUREMENT

The most widespread method of pressure measurement, in the context of ventilation therapy, is piezoelectricity. Hereby, the applied pressure deforms a piezo-element which is part of a Wheatstone bridge circuit. Changes in the materials strain result in changes in output voltage. In modern AWS, pressure can be measured from −20 to +120 cmH2O, with a precision of ±4% or ±2 cmH2O, depending on the respective device.

4.2 | LOCATION OF PRESSURE MEASUREMENT

The location of the pressure sensor within the circuit is of crucial importance as pressure can change within the airways due to their resistance. Although the pressure sensor would be best placed close to the patient’s airways, they are usually mounted remotely from the patient, for reasons of sensor patency. As a minimum, one pressure sensor is placed in the expiratory limb of the breathing circuit (Figure 1). Pressure measurement in the expiratory part facilitates detection by low-pressure alarms as disconnection of the breathing system may not register immediately (or be delayed) at the inspiratory site, due to backward pressure that may result from high resistance of the artificial airways (as it applies typically for artificial airway components for pediatric ventilation). Turbine driven AWS typically have pressure sensors in the inspiratory limb and the expiratory limb of the breathing circuit. This enables alternating airway pressure measurement depending on the phase of the breathing cycle. This way, there is no flow present in the limb of measurement and its resistance does not add to the measured pressure. For security reasons, most AWS include an additional mechanical pressure gauge with resolution of ±5 cmH2O, which serves as a backup for electricity failure.

4.3 | INTERPRETATION OF PRESSURE MEASUREMENT

Although pressure measurement technology can be considered highly precise, it is less obvious which information is reflected by the airway pressure displayed in the AWS. Applying Newton’s equation of motion, it appears that the pressure signal consists of different...
components (Equation 1). Generally spoken, the airway pressure \( P_{aw} \) is composed of (a) the expansion of the respiratory system’s elastic forces (i.e., compliance; \( C_{RS} \)), depending on volume \( V \), (b) the pressure generated by the respiratory system’s resistance \( R_{RS} \), depending on flow \( V' \), and (c) a constant pressure portion \( P_0 \), reflecting PEEP.

\[
P_{aw} = \frac{1}{C_{RS}} \cdot V + R_{RS} \cdot V' + P_0
\]  

FIGURE 2  Computer-generated time curves of airway pressure \( P_{aw} \) (black solid line), tracheal pressure \( P_{trach} \) (gray dotted line) and flow (blue line), and pressure-volume (p-V) loops and flow-volume loops during pressure-controlled ventilation, for various combinations of compliance (C) and resistance (R) of the respiratory system. In the p-V loops, the lower left and upper right dot mark the beginning and the ending, respectively. Left column: typical conditions of the respiratory system. The difference between \( P_{aw} \) and \( P_{trach} \) corresponds to the resistance of the artificial airways. Middle column: curves in the presence of high and fixed resistance of the upper airways (e.g., kinking of the endotracheal tube, secretion, or laryngeal spasm); please note that \( P_{trach} \) does not reach \( P_{aw} \) at the end of inspiration and PEEP and the end of expiration and that maximum flow is symmetrically reduced during inspiration and expiration (by contrast, obstruction of the lower airways, e.g., bronchospasm, would mainly reduce flow in expiration). As a result, tidal volume is reduced, which appears from p-V loop and flow-volume loop and both loops do not close due to incomplete expiration. Therefore, the steepness of the diagonal between start-inspiratory pressure (left lower dot) and end-inspiratory pressure (right upper dot) in the p-V loop does not correctly reflect compliance. Right column: curves in the presence of low compliance (e.g., capnoperitoneum or external compression of the thorax, e.g., the surgeon bracing on the thorax). That tidal volume is considerably diminished which appears from p-V loop and flow-volume loop. Moreover, in the p-V loop, the steepness of the diagonal between start-inspiratory pressure (left lower dot) and end-inspiratory pressure (right upper dot) is reduced. Please note that changes in R or C are not recognizable in the course of \( P_{aw} \) compared to volume-controlled ventilation (see Figure 3) [Colour figure can be viewed at wileyonlinelibrary.com]

Another pressure portion also includes the respiratory system’s inerterance. This term relates to volume acceleration and is not displayed within the AWS’ monitoring and mostly neglected.

Equation 1 illustrates that whenever flow is present, the resistive component adds to the displayed pressure signal. Since \( R_{RS} \) is mainly caused by artificial airways, this pressure component does in fact not contribute to the mechanical forces expanding the biological tissue. Accordingly, during dynamic conditions, \( P_{aw} \) does not necessarily reflect the physiologically effective pressure in the alveoli.
This is particularly relevant at high respiratory rates, where periods of no-flow are short, if present at all. However, only within such no-flow periods, measured Paw reflects the pressure in the alveoli. The duration of the equilibration during a no-flow period depends on viscoelasticity and homogeneity of the lungs, which can be significantly impaired in children undergoing anesthesia. If during VCV, a ventilation pause is present at the end of inspiration, the equilibration can be observed as an exponential pressure decrease approaching plateau pressure (Figure 4).

It is of note that Paw measurement in AWS generally includes the resistance of the artificial airways distal to the Y-piece and particularly that of the endotracheal tube. Thus, Paw generally overestimates alveolar pressure during inspiration and underestimates pulmonary pressure during expiration (Figures 2 and 3). Various methods have been evaluated to approximate the pressure in the airways. Since direct measurement (e.g., with specific endotracheal tubes) could not prevail, specific algorithms were developed to calculate tracheal pressure based on knowledge about the size of the endotracheal tube. Such monitoring is already available in some ICU ventilators. Secretions, however, limit validity of these methods.

**Figure 3** Computer-generated time curves of airway pressure (Paw; black solid line), tracheal pressure (Ptrach; gray dotted line) and flow (V; blue line) and pressure-volume (p-V) loops and flow-volume loops during volume-controlled ventilation, for various combinations of compliance (C) and resistance (R) of the respiratory system. In the p-V loops, the lower left and upper right dot mark beginning and ending of inspiration, respectively. Left column: typical conditions of the respiratory system. The difference between Paw and Ptrach corresponds to the resistance of the artificial airways. This difference is only present when flow is present. Thus, Paw equals Ptrach during the end-inspiratory plateau but not peaks of both pressures differ. Middle column: curves as typically displayed in the presence of increased resistance of the upper airways (e.g., kinking of the endotracheal tube, secretion, or laryngeal spasm); please note that the p-V loop becomes wider at same maximum tidal volume and that maximum flow is lowered during expiration. Moreover, both loops do not close due to incomplete expiration. Right column: curves as typically displayed in the presence of low compliance (e.g., capnoperitoneum or external compression of the thorax as when the surgeon bracing on the thorax). Please note that the slope of the p-V loop is less steep at same maximum tidal volume and that peak expiratory flow is increased. Please note that changes in C and R are reflected in the steepness of the inspiratory limb and breadth of the p-V loop, in contrast to pressure-controlled ventilation (Figure 2) [Colour figure can be viewed at wileyonelibrary.com]
5 | MONITORING OF RESPIRATORY SYSTEM MECHANICS—SECONDARILY CALCULATED INDICES

The monitoring of modern AWS provides various respiratory indices, secondarily calculated from the above mentioned directly measured variables. In terms of pediatric ventilation therapy, respiratory system compliance and resistance appear the most relevant.

5.1 | CALCULATION OF RESPIRATORY SYSTEM COMPLIANCE (CRS)

CRS is considered an important indicator of the conditions of the patient’s respiratory system. A multitude of methods exists to calculate CRS. Each approach, however, provides different information. Historically, CRS was measured via step-by-step inflation of the lungs with defined volume portions, during a super-syringe maneuver, later with a continuous slow inflation maneuver. These maneuvers were required to generate conditions of zero or very low flow to generate (almost) static conditions and to eliminate the resistance-related pressure share in the measured signal. Later algorithms allowed for estimating CRS without interruption of the breathing cycle, using a so-called semi-static approach (Figure 5, left panel). Therefore, the volume applied within a single breath is divided by the difference between plateau pressure and end-expiratory pressure of the same breath. The value of semi-static CRS turns out lower than the static CRS, due to the limited time for pressure equilibration in the respiratory system. From a clinical perspective, semi-static CRS reflects better physiological conditions during the ongoing ventilation than static CRS.

Particularly during ventilation of infants and neonates, it can be difficult to achieve reliable values for semi-static CRS. As mentioned earlier, resistive pressure portions influence pressure measurement and consequently challenge accuracy of the calculated CRS. Taking the high respiratory rates into account, the end-inspiratory airway pressure is likely to include resistive pressure components. Thus, driving pressure will be estimated to high and CRS to low, accordingly.

In the light of these limitations, modern AWS determine the dynamic CRS. Thereby, resistance and compliance are determined by multilinear regression analysis fitting the pressure-, flow-, and volume curves of a breath to the equation of motion (Equation 1). Current research has enhanced this approach in a fashion to show the intratidal change of compliance within a breath. One approach, proved reliable in children from the age of 9 months, separates tidal volume mathematically into isovolumetric steps (Figure 5, right panel). CRS is then calculated for each step, respectively, from data surrounding this step within a defined volume slice. The resulting compliance-volume curve provides information on the recruitment state of lungs since a compliance increasing with volume indicates intratidal recruitment/derecruitment and a compliance decreasing with volume indicates overdistension. Taken this information into account, PEEP and VT could be set based on actual individual conditions. Such methods are of particular interest in the pediatric population, where ventilation pressure settings remain to be guided on individual parameters due to rapidly changing respiratory system characteristics with adolescence.

To date, most AWS allow for displaying the pressure-volume (P-V) loop, which provides a wealth of information. Figure 2 simulates the most relevant changes of the P-V loop during PCV and Figure 3 during VCV, respectively. Moreover, the P-V loop may help to discriminate spontaneous breathing from mandatory ventilation. During spontaneous breathing, the loop shows negative pressure values and clockwise cycling.

5.2 | CALCULATION OF RESPIRATORY SYSTEM RESISTANCE (RRS)

Resistance of the respiratory system (RRS) plays a key role in pediatric ventilation therapy, although clinically often disregarded. To determine RRS in a closed system, pressure is measured at the end of expiration (zero flow) in the beginning of inspiration (maximal flow). Under the assumption that when the flow is maximal, the filling of
the lung is negligible (initial beginning of the inspiration), therefore compliance does not yet contribute to pressure. $R_{RS}$ results from the division of the determined difference in pressure. To increase accuracy, this value is corrected by the internal resistance of the breathing system (including breathing hose and Y-piece), as determined during initial calibration of the device. Consequently, the displayed $R_{RS}$ includes both artificial and biological airways. In order to determine resistance of the biological airways only, calculating tracheal pressure as mentioned above, would be a promising tool. This calculation could be further improved by regarding the resistance of other artificial airway components.16

Interpretation of the flow-volume loop (V-V loop) may be of diagnostic value regarding the ethology of changes in $R_{RS}$. Figure 2 shows the most relevant changes of the V-V loop during PCV and Figure 3 during VCV, respectively. Above that, a saw-toothed shape of the loop indicates secretion in the airways under clinical conditions.

6 | GAS MONITORING—PRINCIPLES OF MEASUREMENT

In the AWS, gases are measured using a specific tool, which includes sensors for all substances of interest. Depending on the device, this measuring bench can either be integrated in the AWS or kept separate, sending only the result of the analysis to the AWS. In machines without an integrated measuring bench (e.g., Fabius, Dräger Medical, Germany), an isolated oxygen measurement device is mandatory. For such purpose, a galvanic fuel cell is employed, which has a limited lifespan due to the need of a chemical reaction.

The measuring bank is fed by a gas sample withdrawn at the Y-piece of the breathing system. The length of the sample line (usually 2 meters) determines the delay in the analysis. The sample gas is drafted with about 200 ml per minute with the sampled volume being returned into the circuit. In order to deliver precise tidal volume, the loss in inspiratory volume will be automatically compensated by additional volume depending on the sample flow. Some devices offer lowering of the sample gas flow during ventilation of pediatric patients. This, however, can increase the delay of gas analysis and the risk of measurement inaccuracy in case of high respiratory rates. Moreover, particularly during low- and minimal-flow anesthesia, the high humidity in the system can damage the measuring bank due to condensation. Such problems can be encountered by including water traps into the sampling line and the breathing system.

6.1 | LOCATION OF GAS SAMPLING

Basically, sample gas should be aspirated as close to the patient airways as possible, to ensure collection of reliable mixture of the gas in question. In the need of small tidal volumes, however, the measurement can be affected by the volume of dead space. Heat and moisture exchangers (HME), with or without filter function, are a source of relevant dead space volume. During pediatric ventilation, they are therefore preferably positioned at the AWS’s inlet and outlet. If used
between Y-piece and airway device, the following consequences have to be considered (Figure 6).

Without leakage in the breathing system, breathing gas sampled between Y-piece and HME filter will most likely mirror a reliable gas composition (Figure 6, left panel). With very small tidal volumes, however, a sample may contain dead space volume in the beginning of exhalation (i.e., less CO₂), but fail to catch the least portion of the exhaled volume, as it is captured within the dead space volume of the filter. Since the volume in the very end of exhalation reflects alveolar gas composition at best, various techniques are employed to ensure accurate measurement in this phase. For example, the ATLAN (Dräger Medical, Germany) deploys a slightly decelerating PEEP during expiration, thus generating a discrete pressure gradient to maintain expiratory flow.¹⁷

With a leakage in the breathing system (e.g., caused by uncuffed endotracheal tube), the permanent flow in the breathing circuit of the AWS can displace exhaled gas from the point of gas sampling and thus lower the quality of the gas sample (Figure 6, middle panel).

With the use of special filters or connectors for neonatal ventilation, breathing gas can also be sampled between the filter and the airway device. With respect to the sample flow, it has to be considered that filter obstruction may result in sub-atmospheric pressure in the patient’s lungs (Figure 6, right panel). Sudden changes in filter resistance may be detected by a timely corresponding decrease of peak expiratory flow (compare Figures 2 and 3, middle column). To prevent such an event, HME without filter function should be used in this place.

### 6.2 | GAS MONITORING—OXYGEN

Standards for basic anesthetic monitoring are defined by the American Society of Anesthesiologists as well as other national institutions, constituting that adequate oxygen concentration in the inspired gas has to be ensured in all patients undergoing anesthesia.

Today all AWS comply with these requirements, most commonly using a paramagnetic analyser to measure oxygen concentration of the gas mixture. In contrast to other medical gases, oxygen molecules are attracted inside a magnetic field. Different techniques are used to translate the influence of oxygen concentration on the paramagnetic field into a proportional change of an electrical signal.

 Particularly when anesthetizing pediatric patients, fast increase of the inspired oxygen concentration may be of importance. The speed of this increase, that is, the time constant of the respiratory system, depends on various factors. First of all, fresh gas flow has to be increased to a maximum, which is up to about 18 L per minute. As mentioned previously, the fresh gas inlet is isolated from the breathing circuit in order to ensure precise delivery of small tidal volume. Thus, in AWS using a piston pump, the increased oxygen concentration enters the breathing circuit in a breath by breath fashion. Consequently, the speed of oxygen increase depends on the patient’s minute ventilation and the volume enclosed by the inspiratory limb of the breathing circuit (e.g., breathing circuit, piston, and breathing hose). Assuming a breathing volume of 1.6 L per minute (infant 9 kg, VT 54 ml, RR 30 per minute) and a volume of 2.4 L in the inspiratory limb of a Primus (Dräger Medical, Germany) equipped with neonatal breathing hoses, it would take up to 90 s to achieve an inspired oxygen concentration of 100%. Therefore, newer generations of piston driven AWS adjust volume of the piston automatically in order to decrease the total system volume and thus accelerate change in gas mixture.

By contrast, in AWS using a turbine to drive ventilation, this process is accelerated using the continuously applied circulatory flow.

### 6.3 | GAS MONITORING—CARBON DIOXIDE

The concentration of carbon dioxide (CO₂) impacts lung perfusion, cerebral autoregulation, or postoperative mental status, for

![Figure 6](https://wileyonlinelibrary.com)
example. Ventilation during anesthesia of small infants, newborns, and neonates should therefore target a physiological range. Today, a bundle of different technologies are available to measure expired CO₂ concentration in the anesthetized patient. Each method has its strength and weaknesses which to compare is beyond the scope of this review.

In the AWS, the expired CO₂ concentration is continuously measured and displayed referred to as capnography. The maximum value of the capnography trace within a single breath refers to end-tidal CO₂. In comparison with the currently most reliable method of CO₂ measurement in the body, that is, partial pressure of arterial carbon dioxide (PaCO₂), the discrepancy between PaCO₂ and end-tidal CO₂ should not exceed ±5.2 mmHg.⁰⁹

Capnography follows infrared absorption spectroscopy at specific wavelength, just as the measurement of volatile anesthetics (see below). The energy of the infrared radiation leads to a molecular vibration which can then be measured (e.g., by temperature change).

### 6.4 GAS MONITORING—VOLATILE ANESTHETICS

As with CO₂, infrared absorption spectroscopy is employed to measure the concentration of volatile anesthetics in the sidestream method. Whereas the absorbance peak of CO₂ is between a wavelength of 4–5 μm that of typical anesthetic agents lie between 8 and 13 μm. As a source of error, it is to consider that the agents’ spectrums can overlap, requiring complex algorithms to discriminate between the respective components.

The concentration of the anesthetic agents is measured in a continuous fashion but is usually displayed as maximum values (percent of volume) for the inspiratory and expiratory phase of a tidal breath. Certain AWS automatically calculate the ratio of current expiratory concentration of a volatile anesthetic and age-corrected minimal alveolar gas concentration (MAC). This facilitates age-specific anesthetic agent delivery. To ensure correct calculation, it is important to enter the patient’s age into the system. For example, Dräger refers for their devices to an equation published by Mapleson, applying to patients older than 1 year.

### 7 REFLECTIVE QUESTIONS

1. Which pressure amplitude should be used in order to calculate semi-static compliance of the respiratory system and why would false measurement underestimate the true compliance of the respiratory system?

2. Why is it important to calibrate the AWS with the breathing hose in question?

3. What is the mean sample rate for gas measurement in the AWS and where should the sample gas be aspirated in order to avoid sub-atmospheric pressures in the patients’ airways?

4. Which factors impact the time required to increase oxygen concentration in the breathing system?

### DATA AVAILABILITY STATEMENT

Data sharing is not applicable to this article as no new data were created or analyzed in this study.

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