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Anesthesia Workstation Performance in Pediatric Ventilation: Part 1. Precision of Ventilation

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EDITOR'S PERSPECTIVE

What We Already Know about This Topic

- Different ventilator technologies are available in different anesthesia workstations, each representing unique pneumatic characteristics
- The ventilation of children may challenge the performance of these ventilators

What This Article Tells Us That Is New

- A bench study of six modern anesthesia workstations using pressure-controlled and volume-controlled ventilation in models for premature infants, neonates, infants, and toddlers found that during volume-controlled ventilation, the set and delivered tidal volume differed considerably, with the deviations decreasing with increasing tidal volume
- During pressure-controlled ventilation, the delivered and set peak inspiratory pressure were in good agreement
- The compensation for leakage was sufficient during pressure-controlled ventilation but not during volume-controlled ventilation
- The inspiratory to expiratory ratios deviated considerably from the set values in some of the devices

ABSTRACT

Background: Mechanical ventilation of children sets the highest demands on the performance of anesthesia workstations. Different ventilator technologies are available, each representing unique pneumatic characteristics. The hypothesis of this study was that precision of delivered ventilation parameters depends on the ventilator type.

Methods: Six modern anesthesia workstations (GE HealthCare Carestation 650; Mindray Bio-Medical Electronics Co. Ltd. Wato EX-65 Pro and A9; Dräger Medical Atlan A350 and Perseus A500; and Getinge Flow-c) were tested in physical models simulating the respiratory systems of a preterm baby, a neonate, an infant, and a toddler. Volume-controlled and pressure-controlled ventilation were investigated with and without airway leakage. Delivered tidal volume, peak inspiratory pressure, positive end-expiratory pressure, and inspiratory to expiratory ratio were compared to the set values. The data are presented descriptively; a difference greater than or equal to 10% between the delivered and set values was considered clinically relevant.

Results: Deviations of the delivered volume from the set tidal volume during volume-controlled ventilation ranged between $-19.5 \pm 0.6\%$ and $+35.1 \pm 3.0\%$ in the preterm baby model but decreased with increasing tidal volume. Positive end-expiratory pressure and peak inspiratory pressure during pressure-controlled ventilation did not differ relevantly from the set values in any device or lung model. Expiratory ratios varied widely in most devices, leading to relevant differences in tidal volume during pressure-controlled ventilation. With leakage, tidal volume was, on average, 7% lower during volume-controlled ventilation but 2% lower during pressure-controlled ventilation.

Conclusions: Precision of ventilation with modern anesthesia workstations varies depending on the ventilator type, ventilation mode, and set parameters. Pressure-controlled ventilation is advantageous in the six workstations tested considering the good agreement between the applied and the set parameters and leakage compensation. The results imply precision superiority of pressure-controlled ventilation over volume-controlled ventilation in preterm babies and neonates, independent from the ventilator technology in use.

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Abbreviations: **AWS**, anesthesia workstation; **C_{RS}**, compliance of the respiratory system; **ETT**, endotracheal tube; **HME**, heat and moisture exchanger; **ICU**, intensive care unit; **I:E**, inspiratory to expiratory ratio; **PCV**, pressure-controlled ventilation; **PEEP**, positive end-expiratory pressure; **P_{insp}**, peak inspiratory pressure; **P_{pul}**, pressure inside the lung model; **VCV**, volume-controlled ventilation; **V_T**, tidal volume

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Perioperative ventilation of small children is among the most demanding situations in anesthesia.[1] Ventilator technologies that are available for this purpose differ considerably between modern anesthesia workstations (AWSs)[2] and may determine the precision of the provided ventilation measures. Traditional technologies include pneumatically driven bellows and spindle-driven pistons. Approaches that are more modern utilize turbines to accelerate the breathing gas. The latest technology includes volume reflectors for conditioning and rebreathing of expired gas.[3] In conjunction with the technique in use, each system has a unique volume inside the respiratory circuit, specific pressure, and flow sensors and purpose-built mechanisms for supplying fresh gas.[2] It can be assumed that each technology has strengths and weaknesses, depending on the function in question. These differences may be most relevant during the ventilation of small children.[4–6]

With regard to the growing awareness of ventilator-induced lung injury, setting ventilation parameters with respect to the vulnerability of the pediatric lung needs careful individual reflection.[7,8] Considering that ventilation of adult patients with an AWS can generate a relevant volume error,[9] individual performance characteristics may have a significant impact when ventilating children. Already small deviations between intended and delivered tidal volume can lead to proportionally large errors.[10] The consequences may range from derecruitment of the lungs (atelectasis)[11] to overinflation of the lungs (volutrauma)[12] and development of bronchopulmonary dysplasia in the most vulnerable patients.[13]

However, knowledge about the precision of ventilation in neonates and infants during anesthesia is scarce or outdated.[4] From a technical perspective, compressibility of the breathing circuit volume, resistance, and inertia of the breathing gas and timely control of the inspiratory and expiratory valve influence precision of mechanical ventilation. Moreover, airway leakage occurs often in the pediatric patient and is therefore a relevant factor to consider.[1,5,14] Generally, leakage results in a discrepancy between the expired tidal volume measured by the AWS and the tidal volume that is meant to be insufflated to the patient's lungs.

We hypothesized that modern AWSs with different ventilator technologies differ in the accuracy of applied ventilation parameters in the pediatric setting. Independent of the ventilation mode, the central variables of mechanical ventilation concern tidal volume, peak pressure, positive end-expiratory pressure (PEEP), and inspiratory and expiratory timing. Therefore, we investigated six AWSs with respect to their ventilation precision during controlled mechanical ventilation in this regard. Taking into account the dynamic development of the respiratory system in young children, the study was performed in four physical models representing specific compliance, resistance, and functional residual capacity of a preterm baby, a neonate, an infant, and a toddler with and without simulated leakage around the airway device, simulating an experimental environment under body temperature pressure saturated conditions. Using the same setup and models, performance of the six AWSs regarding pressure support ventilation and the ability to change the gas concentration in the breathing system was investigated in a subsequent study.[15]

Materials and Methods

Six modern AWSs were investigated: Carestation 650 (software version 2.00.01; GE HealthCare, USA), Wato EX-65 Pro (software version 05.05.00; Mindray Bio-Medical Electronics Co. Ltd., China), A9 (software version 01.17.00; Mindray), Atlan A350 (software version 2.00.01; Dräger Medical, Germany), Perseus A500 (software version 2.03; Dräger Medical), and Flow-c (software version 04.09.00; Getinge, Sweden). The included AWSs contain different types of ventilator technology (fig. 1): tidal ventilation is driven by a bellow with the Carestation and the Wato, by a piston with the Atlan, by a turbine with the Perseus, and by a volume reflector with the Flow-c and the A9. The Atlan and the Perseus were available as institutional standard. The other devices were made available free of charge for the purpose of the study and clinical testing by the respective manufacturers. All devices underwent safety inspection and calibration by the manufacturers before being handed out for the experiments. The setups of the machines were made with assistance from the companies and professional technicians from our department. The devices were connected to central supply for electricity, medical air, and oxygen (500-kPa supply pressure for both). System testing, including breathing circuit leakage test, compliance test, and sensor calibrations were performed at the beginning of each experimental day and before measurements with a new breathing system. Accordingly, leakage and volume differences due to circuit compliance were compensated automatically by the AWS software, if technically available.

Breathing Circuit and Airway

Single-use breathing systems were used, with the size depending on the patient model in question. For simulated ventilation of preterm babies and neonates, 10-mm-inner-diameter breathing tubes were used (150 cm; reference No. 305/8173; Medtronic, Ireland) with a 0.5-l breathing bag attached. For simulated ventilation of infants and toddlers, 15-mm-inner-diameter breathing tubes were used (150 cm; reference No. 305/8926 [distribution discontinued for

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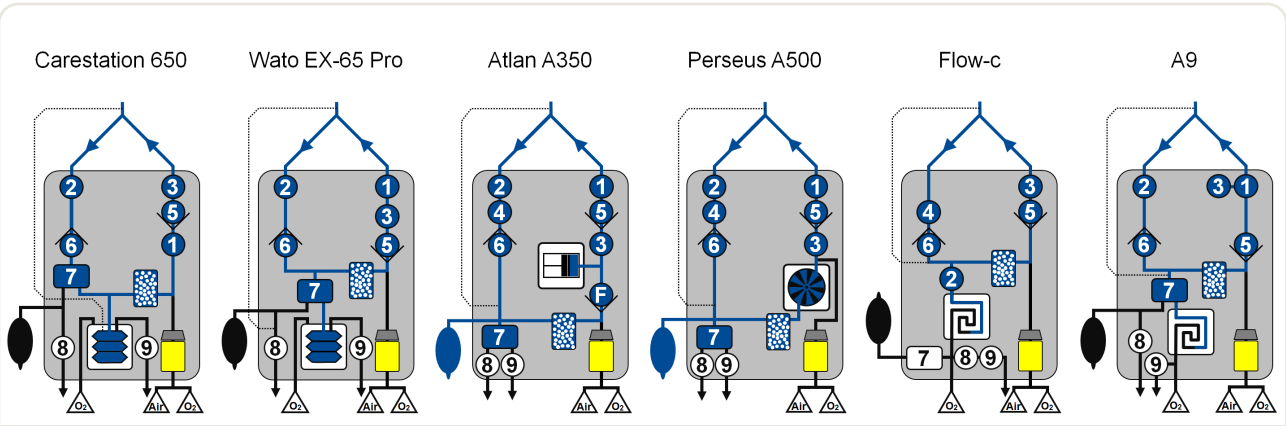

Fig. 1. Simplified pneumatic diagrams of the included anesthesia workstations with different ventilator technologies (from *left* to *right*): Carestation 650 (GE HealthCare, USA) and Wato EX-65 Pro (Mindray Bio-Medical Electronics Co. Ltd., China; both bellow), Atlan A350 (Dräger Medical, Germany; piston), Perseus A500 (Dräger Medical; turbine), and Flow-c (Getinge, Sweden) and A9 (Mindray; both volume reflector). The *numbers* indicate positions and types of sensors and valves: *1*, inspiratory flow sensor; *2*, expiratory flow sensor; *3*, inspiratory pressure sensor; *4*, expiratory pressure sensor; *5*, inspiratory valve; *6*, expiratory valve; *7*, changeover between manual and automatic ventilation; *8*, adjustable pressure-limiting (active during manual ventilation); *9*, electronically controlled expiratory valve (active during mechanical ventilation). Please note that machines with the same ventilator technologies differ in the configuration of the breathing circuit. This includes position of the ventilator with respect to the carbon dioxide absorber, point of fresh gas supply form the gas mixer and sidestream gas measurement (*dashed lines*). In pneumatically driven ventilators, oxygen (O_2) is used as driving gas. Moreover, the Atlan is the only device providing a fresh gas decoupling valve (*F*). Blue-colored elements are streamed by breathing gas in the controlled ventilation mode. Please note that only in the Atlan and the Perseus, the breathing bag is always part of the breathing circuit, independent of the ventilation mode. Air, medical air.

this type]; Medtronic) with a 1.0-l breathing bag attached. Endotracheal tubes (ETTs; Microcuff, Avanos Medical GmbH, Germany) of 3.0-, 3.5-, 4.0-, and 4.5-mm inner diameters were used for simulated ventilation of preterm babies, neonates, infants, and toddlers, respectively. Cuffed endotracheal tubes were used to ensure airtight connection between the tubing system and the patient model. No additional filters or heat and moisture exchangers (HMEs) were used to prevent the additional resistance from obscuring potential differences in the performance of the devices.

Model Setup and Patient Categories

Four models of different patient maturity were developed. Model characteristics were adjusted to typical body weights, as well as compliances (C_{RS}) and resistances of the respiratory system (R_{RS}; resistive pressure gradient including the ETT at 0.1 l · s^{-1}): preterm baby (about 2 kg, static C_{RS}: 2.5 ml · cm H_2O^{-1}, R_{RS}: 16.7 cm H_2O); neonate (about 3.5 kg, static C_{RS}: 5.0 ml · cm H_2O^{-1}, R_{RS}: 7.5 cm H_2O); infant (about 10 kg, static C_{RS}: 12.0 ml · cm H_2O^{-1}, R_{RS}: 2.0 cm H_2O); and toddler (about 14 kg, static C_{RS}: 20.0 ml · cm H_2O^{-1}, R_{RS}: 1.3 cm H_2O).[16–18] Lungs were simulated by glass bottles of a capacity corresponding to the targeted C_{RS} (fig. 2A).[19,20] R_{RS} was approximated through narrowing the artificial tracheal lumen. Respective C_{RS} and R_{RS} were validated *via* multiple linear regression analyses of preliminary pressure and volume recordings. The openings of the glass bottles were sealed with a plug (same in all models), providing ports for required measurements. To counteract nonlinear thermal effects of changing pressures, all bottles were partially filled with copper wool. To generate an inner compartment corresponding to a simulated functional residual capacity of the respective model, a thin-walled plastic bag was air-tightly attached to the bottle plug from inside.

Body temperature pressure saturated conditions were achieved by the following measures. The lung models were heated inside, keeping a temperature between 36° and 38°C. The circulating air was moistened before the experiments by implementing a heated humidifier (MR850; Fisher & Paykell Healthcare, New Zealand) into the breathing system for about 5 min and removed thereafter. Preliminary experiments showed that moisture could only be prevailed over the period of measurements if carbon dioxide is present (due to the chalk generating heat and water). Therefore, carbon dioxide was insufflated into the inner compartment using a roller pump (Stöckert; Sorin Group Deutschland GmbH, Germany), enabling exchange of gas while maintaining volume constancy by in- and outflow of equal volumes (fig. 2B). Insufflation of carbon dioxide and suctioning of breathing gas were realized at different points in the inner compartment of the lung model to ensure uniform gas mixing. Carbon dioxide was supplied from a cylinder, with the pressure regulated to mean airway pressure in the model and moistened using an MR850. Regulated *via* the speed of the roller pump, carbon dioxide partial pressure in the expired gas was maintained between 35 and 45 mmHg and measured *via* sidestream technique with the AWS in question. To avoid condensation of the breathing

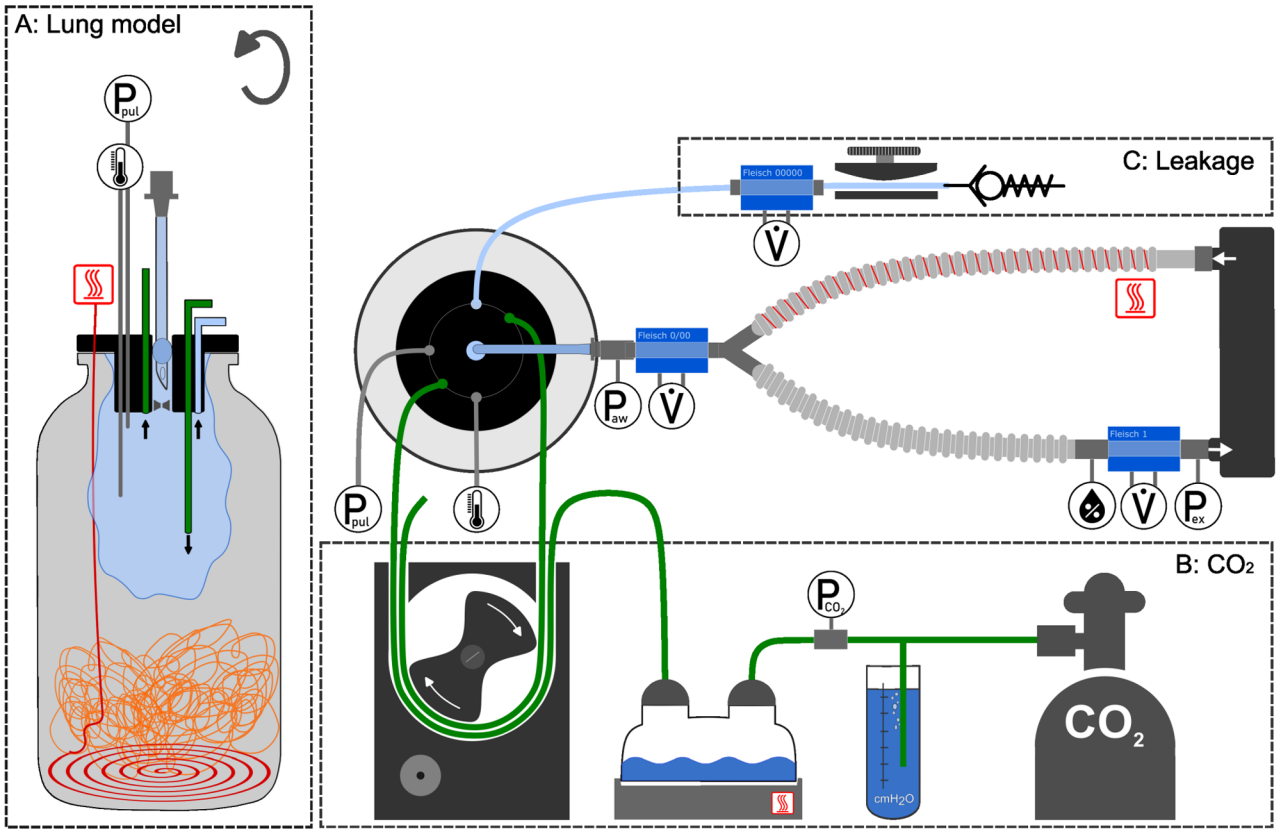

Fig. 2. Experimental setup. (*Left side*, *A*) Schematic drawing of the lung model simulated by a glass bottle with the volume corresponding to the respiratory system compliance of the simulated patient. The bottle includes a heating wire (*red*), copper wool (*orange*) and central opening for the endotracheal tube. Sealing of the bottle includes boreholes for lung pressure and temperature measurement, leakage outflow (*blue tube*) and carbon dioxide (CO_2) inlet and outlet (*green tubes*). Please note the thin-walled plastic foil within the bottle (*bright blue*) separating the ventilated compartment from the bottle's volume without contributing to the compliance of the model. Moreover, tubes for inflow and outflow of CO_2 within the inner compartment end at different heights to ensure uniform gas mixing. (*Right side*) Lung model (bird's eye few) with the breathing system and anesthesia workstation attached to the endotracheal tube. Pressure (P), flow ($\dot{V}$), temperature (thermometer icon), and humidity (drop icon) were measured at several points in the breathing circuit. Delivered inspiratory and expiratory tidal volumes were measured using the pneumotachograph for flow between the endotracheal tube and the Y-connector. Please note the heating wire (*red line*) around the inspiratory limb of the breathing system. (*B*) Technique of insufflation of CO_2 including cylinder, regulation of supply pressure (P_{CO_2}) to mean airway pressure with an underwater tube, moistening CO_2 with an active humidifier and insufflation of CO_2 into the lung model using a roller pump. Please note that the tubes intended for inflow and outflow of CO_2 are inserted in opposite directions to ensure volume exchange in equal parts. (*C*) Throttle valve and mechanical positive end-expiratory pressure valve to regulate leakage flow according to the measured flow ($\dot{V}$).

gas, the inspiratory limb of the breathing system was heated using an externally applied heating wire, and the expiratory limb was isolated to the ambient.

Flow and pressure measurements were performed utilizing a sensor system well established in our group including small tidal volume ranges and rapid pressure changes.[21] Flow was measured using pneumotachographs (Fleisch type; Dr. Fenyves und Gut, Germany) between the ETT and the Y-connector to determine delivered inspiratory and expiratory tidal volumes (preterm and neonatal model: Fleisch 00, suggested range ± 100 ml · s^{-1}, dead space 2.0 ml; infant and toddler model: Fleisch 0, suggested range: ±250 ml · s^{-1}, dead space: 4.0 ml), at the leakage tube to determine leakage volume (Fleisch 00000; suggested range: ±9 ml · s^{-1}, dead space: 0.1 ml), and at the expiratory inlet of the AWS to exclude leakage other than intended in the circuit (Fleisch 1; suggested range: ±1,000 ml · s^{-1}, dead space: 14.0 ml). Pressure was measured between the ETT and Y-connector, inside the inner compartment, in the CO_2 supplying tube, and at the expiratory inlet of the AWS. The measurement range of the pressure sensors was ±50 cm H_2O with an error of ±0.25% (type 2; Special Instruments, Germany). Temperature was measured (Testo 720; Testo SE & Co. KGaA, Germany) inside the inner compartment and moisture (OHT20-A; Omni Elektronik GmbH, Germany) at the expiratory inlet of the AWS. Pressure sensors were

calibrated using a pressure calibrator (OM DM 921; Onneken, Germany), and flow sensors were calibrated using calibration syringes of appropriate volume (Hans Rudolph Inc., USA) at the beginning of each day and before experiments, if required. Additionally, recordings started with the breathing system disconnected to determine and correct potential offset in the subsequent offline analyses.

To simulate the effects of leakage around an airway device (*e.g.*, a noncuffed endotracheal tube),[14] leakage was realized using a rubber tube to drain breathing gas from the inner compartment to the ambient (fig. 2C). Leakage flow was regulated by a throttle valve and a mechanical PEEP valve (Ambu, Denmark), set to 5 cm H_2O, at the tube's ambient end. In preliminary experiments, the position of the throttle valve was determined to achieve a leakage volume of 20% of minute ventilation in each respective model.

Protocol for Precision of Ventilation Measurements

PEEP was set to 5 cm H_2O, fresh gas flow to 0.5 $l \cdot min^{-1}$, and fraction of inspired oxygen to 0.4 in all experiments. Inspiratory to expiratory ratio was set to 1:2 in all devices, except the Atlan and the Perseus in combination with the neonate and infant lung models. In these cases, the closest value to be set was 1:1.9. Further ventilation variables were set dependent on the model in question: ventilation frequencies were set to 40/min, 35/min, 30/min, and 25/min in the preterm baby, neonate, infant, and toddler models, respectively.

For volume-controlled ventilation (VCV), tidal volume (V_T) was set to 10 ml in the preterm baby model, 20 ml in the neonate model, 70 ml in the infant model, and 110 ml in the toddler model. The end-inspiratory pause was set to 20%. For pressure-controlled ventilation (PCV), peak inspiratory pressure (P_{insp}) was set to 6 cm H_2O above PEEP in all models.

Data recordings were started, and thereafter ventilation parameters were set, and the breathing system was connected. Recordings were performed for a minimum of 2 min after breathing system connection. Three repetitive measurements were performed for each condition after complete dismantling and reassembly of the system.

Delivered V_T was defined as measured expiratory volume and compared to the set V_T without or with leakage. Precision of PCV was determined by the differences between measured and the set P_{insp} without or with leakage. For all conditions, measured PEEP was compared to the set PEEP. End-inspiratory pressure inside the lung model (P_{pul}) was compared to P_{insp}. The inspiratory to expiratory ratio was compared by means of the set and measured ratio's denominator from measured flow. Moreover, the range of expiratory V_T displayed at the ventilator was documented during the measurements. Breath-by-breath variations of V_T, P_{insp}, and PEEP were depicted by calculation of their 5th and 95th percentiles. Since we aimed to analyze the quality of ventilation under steady-state conditions, all variables were calculated for the last third of each measurement interval. To compare characteristics of the inspiratory pressure increase between devices, peak inspiratory flow ($\dot{V}$) and maximal inspiratory flow difference ($\Delta\dot{V}$) were calculated.

Data Analyses and Statistics

Pressure and flow signals were recorded at a sampling rate of 200 Hz using a custom-made software written in LabVIEW (Laboratory Virtual Instrumentation Engineering Workbench; LabVIEW version 17.0, National Instruments Inc., USA) and analyzed using self-written scripts in MATLAB (R2019a; MathWorks Inc., USA), GraphPad Prism (version 10.1.2; GraphPad Software, USA), and jamovi (version 2.3, jamovi project [2023], retrieved from https://www.jamovi.org). The results are given descriptively as means ± SD from the three repetitions of the experiments. Absolute differences of a measure greater than 10% to the set V_T, inspiratory to expiratory ratio (I:E) denominator and PEEP during VCV and the set P_{insp}, I:E ratio denominator, and PEEP during PCV were considered relevant.[18]

Results

Precision of Applied Set Variables: VCV

VCV was not available for the preterm baby model in the Carestation, Perseus, and Flow-c since V_T less than or equal to 10 ml could not be set. Deviations from delivered to the set V_T were relevant (outside the ± 10% range) in the preterm baby model for the A9 (−19.5 ± 0.6%) and the Wato (+35.1 ± 3.0%) and in the neonatal model for the Flow-c (−12.9 ± 0.3%), the Wato (+25.6 ± 2.5%), and the Carestation (+14.5 ± 0.5%; fig. 3). Deviations from the set V_T in the infant and toddler model were all less than 10 %, but delivered V_T tended to be lower than the set V_T in all devices but the Wato and the A9. Leakage reduced delivered V_T by 7.0 ± 0.8%, except in the Wato and the A9, in which the reduction was less than 1%, considering all lung models (supplemental table S1, https://links.lww.com/ALN/E557).

Measured ratios of expiratory time were relevantly lower than set in the Carestation (all differences greater than or equal to |−14.4 ± 0.0%|) and the Wato (all differences greater than or equal to |−11.6 ± 0.5%|) in all tested models, with the Atlan in the infant model (−14.5 ± 0.3%) and relevantly higher than set in the Perseus in the neonatal model (+10.5 ± 0.5%). Applied PEEP did not deviate relevantly from the set PEEP in any device or lung model. P_{pul} was always lower than P_{insp} (range: 1.3 to 2.8 cm H_2O).

The breath-by-breath variability of P_{insp} and PEEP (supplemental table S2, https://links.lww.com/ALN/E558), as well as inspiratory flow (supplemental table S3, https://links.lww.com/ALN/E559) presented different between

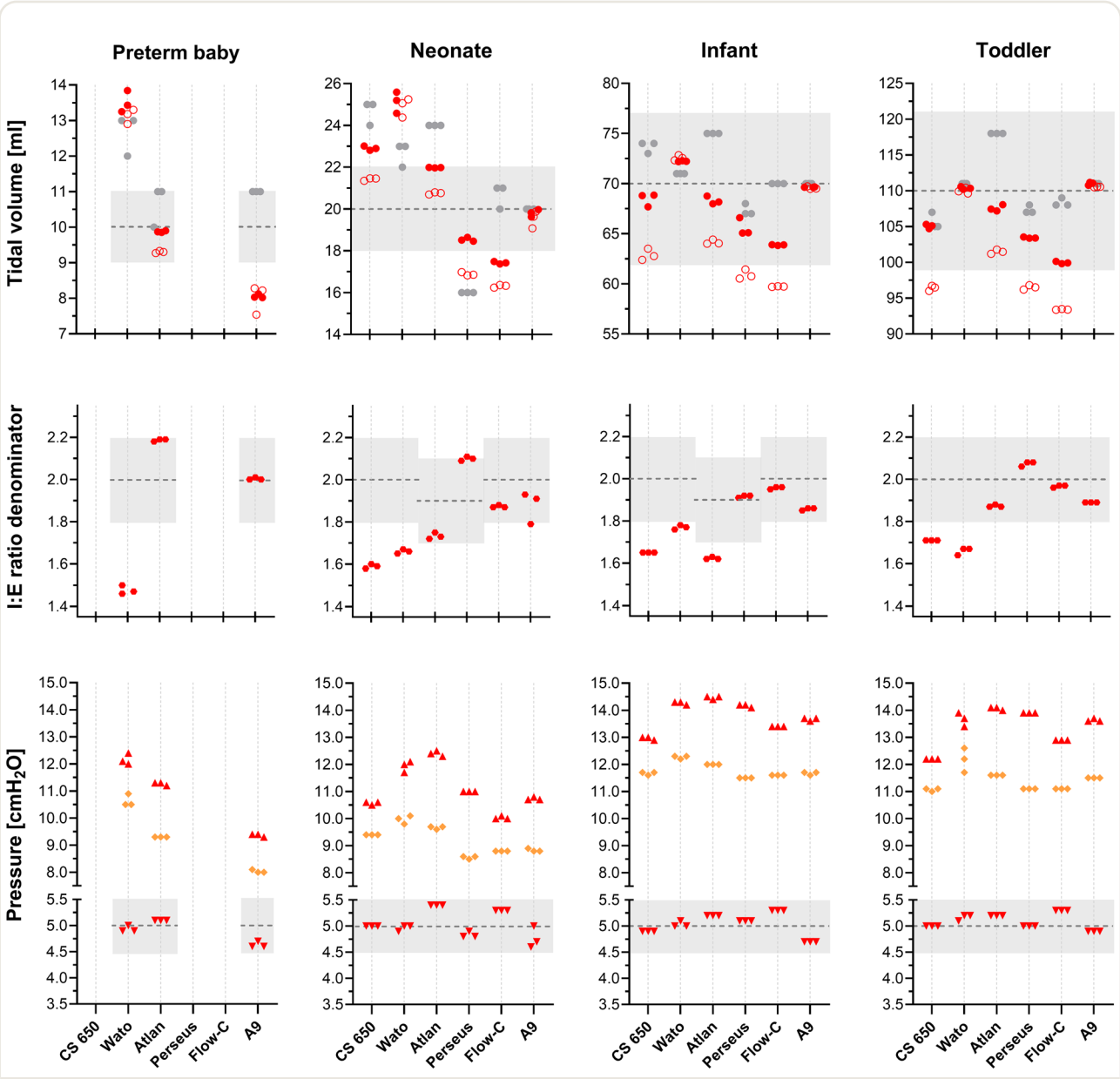

Fig. 3. Ventilation variables during volume-controlled ventilation measured in lung models simulating a preterm baby, neonate, infant, and toddler for each tested anesthesia workstation, respectively. (*Top row*) Delivered tidal volume (V_T) measured at the Y-piece without leakage (*filled orange dots*) and with leakage (*open orange dots*), as well as V_T displayed at the anesthesia workstation (*filled gray dots*). (*Middle row*) Denominator of the inspiratory to expiratory ratio (I:E). (*Bottom row*) Peak airway pressure (*upward-pointing triangles*), peak pulmonary pressure (*diamonds*), and positive end-expiratory pressure (PEEP; *downward-pointing triangles*). Each symbol indicates a single measurement. Please note that setting of 10-ml tidal volume is not possible in the Carestation 650 (GE HealthCare, USA), Perseus A500 (Dräger Medical, Germany), and Flow-c (Getinge, Sweden) during volume-controlled ventilation, and therefore no results are displayed for the preterm baby model. The *horizontal gray dotted lines* indicate set variables (for the Atlan [Mindray Bio-Medical Electronics Co. Ltd., China] and the Perseus I:E ratio could only be set to 1:1.9 in the neonate and infant model). *Gray shaded areas* indicate the tolerable deviation range, defined as ± 10% of the respectively set value. (Wato, Mindray.)

the included devices. Particularly in the initial phase of VCV, the Wato and the A9 needed up to about 80 s to reach constant ventilation conditions in the preterm baby model, as well as the Perseus in the neonatal model (fig. 4). While PEEP was always within the ±10% deviation range, noticeable breath-by-breath variability (determined by 5th and 95th percentile) was present in the Perseus for PEEP and peak inspiratory pressure in the neonatal lung model.

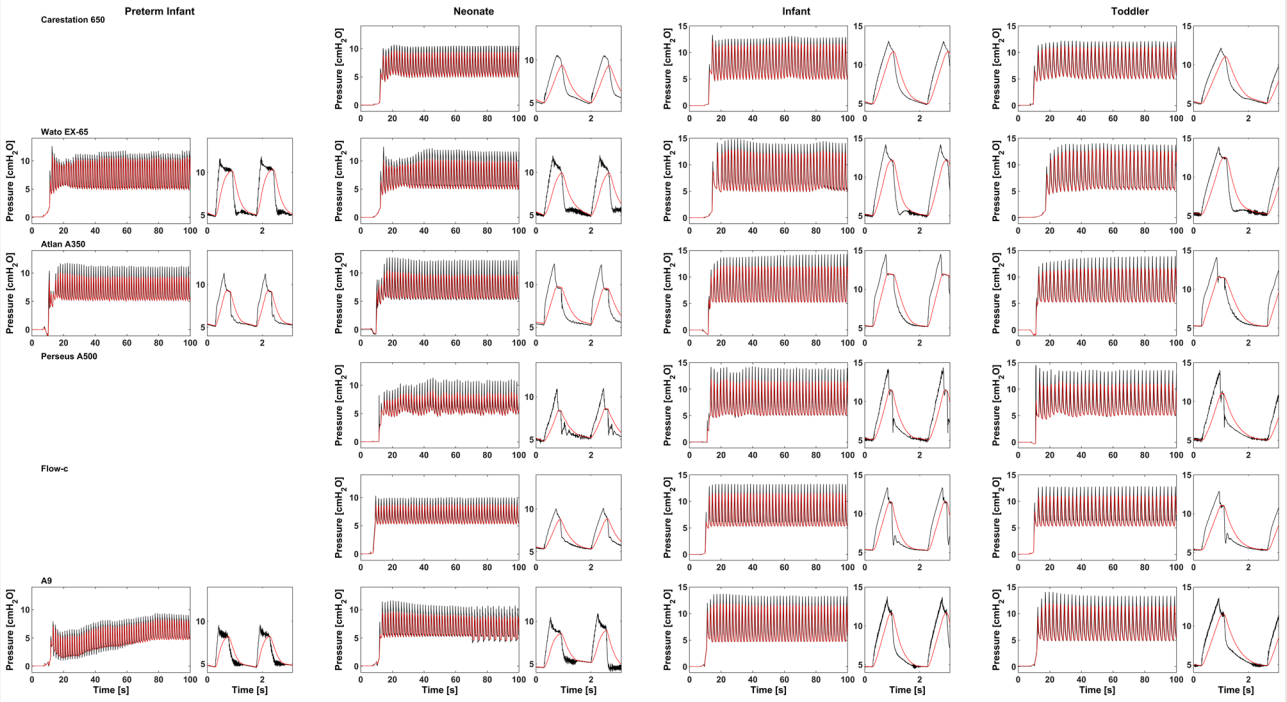

Fig. 4. Exemplary graphs for measured airway pressure (*black*) and pulmonary pressure (*red*) during volume-controlled ventilation. *Rows* indicate the different types of anesthesia workstations. *Columns* indicate different lung models simulating a preterm baby, neonate, infant, and toddler. Of the graphs arranged in pairs in a column, one shows a continuous pressure signal, and the other shows the pressure curves for two exemplary breaths. Please note the differences in scaling of the axes. Moreover, setting of 10-ml tidal volume is not possible in the Carestation 650 (GE HealthCare, USA), Perseus A500 (Dräger Medical, Germany), and Flow-c (Getinge, Sweden) during volume-controlled ventilation, and therefore no results are displayed for the preterm baby model. (Wato EX-65, Mindray Bio-Medical Electronics Co. Ltd., China; Atlan A350, Dräger Medical, Germany; A9, Mindray Bio-Medical Electronics Co. Ltd., China.)

Precision of Applied Set Variables: PCV

P_{insp} measured at the ETT was always within the ±10% range of the set P_{insp} (fig. 5). Leakage had no relevant effect on P_{insp} (supplemental table S4, https://links.lww.com/ALN/E560). Delivered V_T presented different for the AWS and was 2.2 ± 0.3% lower with leakage, considering all comparisons (supplemental table S5, https://links.lww.com/ALN/E561). Inspiratory flows depended on the AWS, too (supplemental table S3, https://links.lww.com/ALN/E559).

Measured ratios of expiratory time were outside the accepted range of ±10% in all devices but the Atlan and the Flow-c considering all patient models. Applied PEEP was comparable to the set PEEP in all devices with each lung model. P_{pul} was always lower than P_{insp}, but the differences presented lower in the infant and toddler model (0.1 to 0.6 cm H_2O) than in the preterm baby and neonatal model (0.3 to 1.6 cm H_2O). Breath-by-breath variability was low during PCV (fig. 6), except for PEEP in the Perseus in the preterm baby and neonatal model (supplemental table S4, https://links.lww.com/ALN/E560).

Discussion

The main findings of the current study can be summarized as follows: (1) during VCV, the delivered and set V_T differed considerably, with the deviations decreasing with increasing V_T; (2) during PCV, the delivered and set P_{insp} were in good agreement; (3) compensation for leakage was sufficient during PCV but not during VCV; and (4) inspiratory to expiratory ratios deviated considerably from the set values in certain devices.

By investigating performance of VCV in a simulated preterm baby, we aimed at evaluating the AWS at the limits of their technical possibilities. From the machines providing VCV, V_T was precisely delivered from the Atlan but exceeded by more than 30% by the Wato and undercut by about 20% by the A9. Delivered V_T also varied considerably in the other lung models, with a trend for undercutting the set values in the infant and toddler lung model.

Technical reasons may account for the observed discrepancies. To achieve the targeted volume, all devices regulate the inspiratory flow based on the set V_T and inspiratory time. Thus, the control variable during VCV is flow.[22] All systems but the Flow-c regulate inspiratory

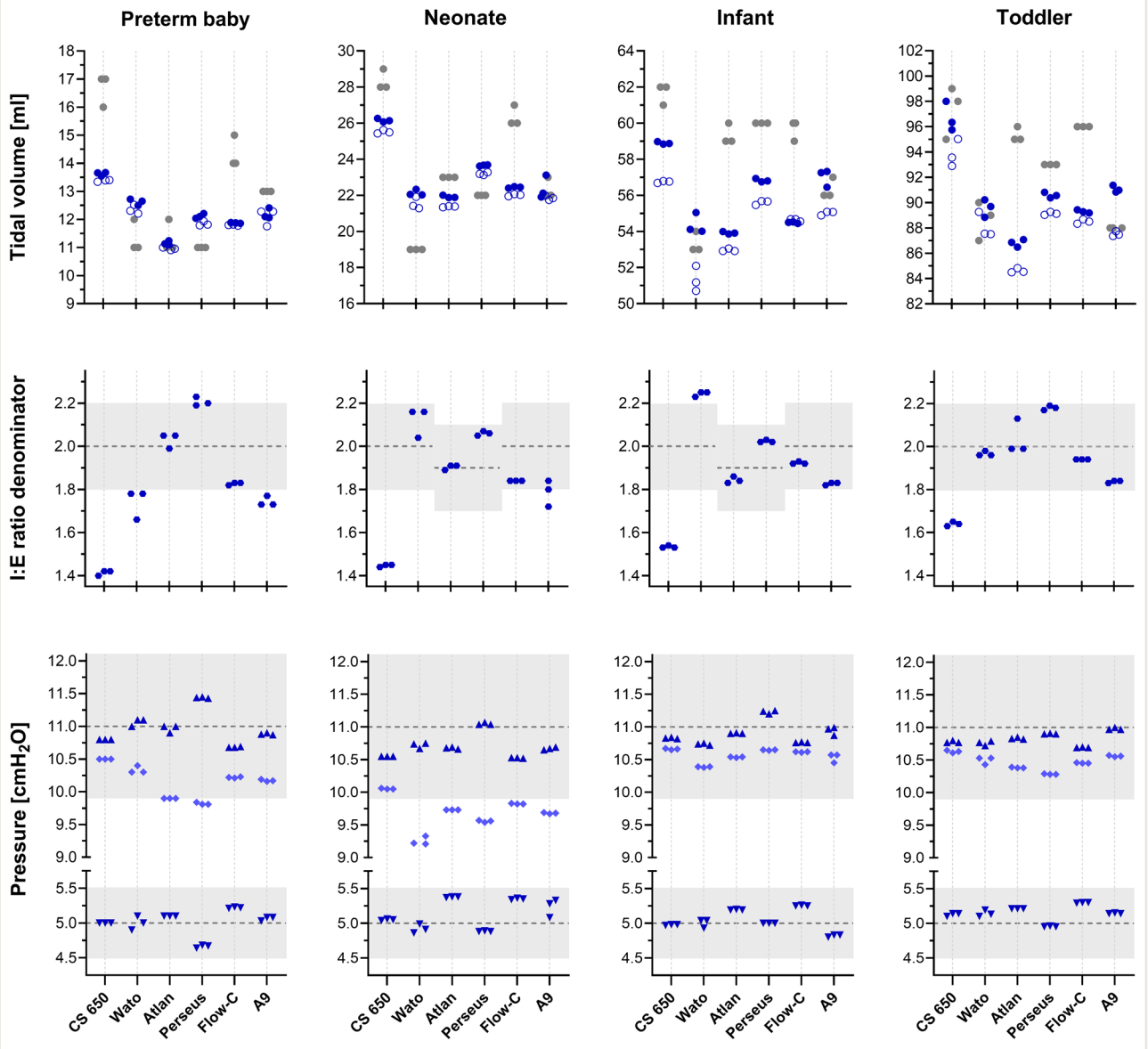

Fig. 5. Ventilation variables during pressure-controlled ventilation measured in lung models simulating a preterm baby, neonate, infant, and toddler for each tested anesthesia workstation, respectively. (*Top row*) Delivered tidal volume (V_T) measured at the Y-piece without leakage (*filled orange dots*) and with leakage (*open orange dots*), as well as V_T displayed at the anesthesia workstation (*filled gray dots*). (*Middle row*) Denominator of the inspiratory to expiratory ratio (I:E). (*Bottom row*) Peak airway pressure (*upward-pointing triangles*), peak pulmonary pressure (*diamonds*) and positive end-expiratory pressure (PEEP; *downward-pointing triangles*). Each symbol indicates a single measurement. The *horizontal gray dotted lines* indicate set variables (for the Atlan [Dräger Medical, Germany] and the Perseus [Dräger Medical] I:E ratio could only be set to 1:1.9 in the neonate and infant model). The *gray shaded areas* indicate the tolerable deviation range, defined as ± 10% of the respectively set value. (CS [Carestation] 650, GE HealthCare, USA; Wato EX-65, Mindray Bio-Medical Electronics Co. Ltd., China; Flow-C, Getinge, Sweden; A9, Mindray.)

V_T during VCV depending on measured inspiratory flow. Whereas after a short settling phase, the Carestation and the Atlan delivered precisely, strong oscillations in P_{insp} occurred in the Wato, the Perseus, and the A9. These oscillations, however, had no effect on P_{pul}. The Flow-c calculates inspiratory V_T from the volume delivered by the ventilator and fresh gas during inspiration. Therefore, it is less affected by adaptive flow regulation, which can be seen in the continuously precise ventilation curves. Underestimation of the set V_T by this device may result from a limitation of our model. In the Flow-c, the delivered inspiratory V_T is regulated to an expected expansion

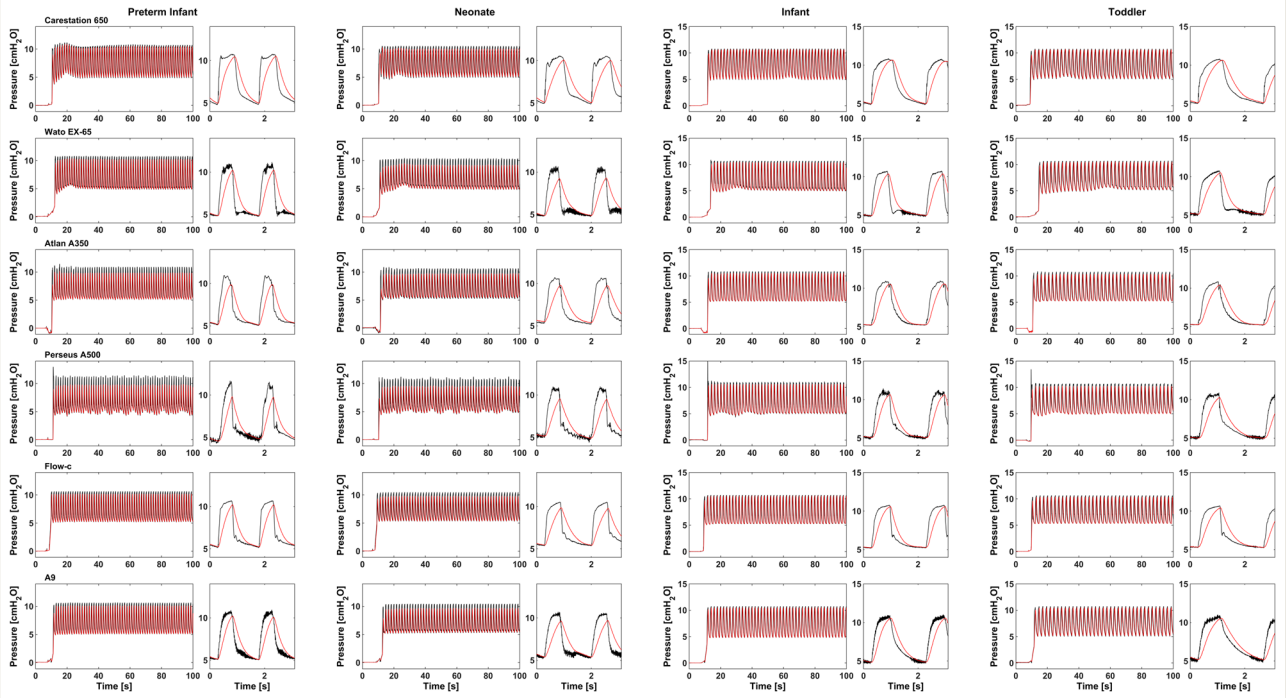

Fig. 6. Exemplary graphs for measured airway pressure (*black*) and pulmonary pressure (*red*) during pressure-controlled ventilation. *Rows* indicate the different types of anesthesia workstations. *Columns* indicate different lung models simulating a preterm baby, neonate, infant, and toddler. Of the graphs arranged in pairs in a column, one shows a continuous pressure signal, and the other shows the pressure curves for two exemplary breaths. Please note the differences in scaling of the axes. (Carestation 650, GE HealthCare, USA; Wato EX-65, Mindray Bio-Medical Electronics Co. Ltd., China; Atlan A350, Dräger Medical, Germany; Perseus A500, Dräger Medical; Flow-c, Getinge, Sweden; A9, Mindray.)

of the breathing gas in the lungs, as a result of an increase in temperature.[23] Homogeneously tempered conditions in our model may have led to delivered V_T being lower by the difference between temperature pressure saturated in the body and in the ambient, which accounts for up to about 10%. Except in the preterm baby model, V_T were delivered most precisely with the A9. By providing leakage compensation during VCV, the Wato and the A9 were the only machines in which leakage did not impair delivered V_T in the vast majority.

We observed a considerable breath-by-breath variability of PEEP and peak inspiratory pressure during VCV in certain devices. In the Perseus, this may result from difficulties in controlling of the turbine at small V_T. If the set V_T is low but a relevant acceleration with the centrifugal pump is required, the centrifugal momentum of the turbine has a significant impact. This may explain that the variability of airway pressures continued during steady-state ventilation in the preterm baby and neonatal model, which became less prominent with higher V_T.

In the Atlan, additionally to flow measurement, information about the piston's position supports determination of the V_T. This may explain the high precision of this ventilator in the preterm neonate model. Moreover, the piston technology allows for immediate termination of volume acceleration. In contrast to the other systems, the piston's end-inspiratory position avoids resonation of the air column of inspiratory V_T. This becomes visible with the immediate drop to end-inspiratory plateau pressure also at high breathing frequencies. Furthermore, a recent software update for the Atlan (2.00.01) seemed to have increased accuracy compared to the preceding software version we used for our preliminary tests.

During PCV, P_{insp} did not deviate from the set value to a relevant extent in any of the devices. The higher precision compared to flow-regulated ventilation may result from the technically less error-prone pressure measurement. Regulation of P_{insp} during PCV is subject to ongoing comparison of the set and actual pressure in all ventilator types.[22] Once the target P_{insp} is achieved, flow is adapted to a level compensating for the pressure loss due to volume redistribution in the lungs, thermodynamic effects, and leakage.

Moreover, comparison of the set and the delivered variables during PCV and VCV must consider that pressure measurement is more precise than flow measurement. In the clinical setting, the range of airway pressures is comparable in patients of all ages, whereas the range of flow depends on the individual minute volume. In our study, peak flow

delivered in the toddler was about six times that delivered in the preterm baby. All included AWSs use a single sensor type to cover the total flow range that may potentially occur. Accordingly, flows typically occurring in small children are at the lower limits of their measurement accuracy, and relatively large errors may occur. Independence from flow measurement may account for the better precision of delivered variables during PCV.

We observed considerable differences in delivered V_T during PCV, resulting from inconsistent performance of control of inspiratory time and pressurization. The inspiratory to expiratory ratio varied from the set values in a wide range. The inspiratory time, however, has a crucial influence on delivered V_T during PCV.[5,8] P_{pul} approximates the set P_{insp} toward the end of inspiration. The longer the inspiratory time, the greater is the chance for pressure equalization, which will result in maximum lung insufflation. The Carestation showed the largest deviations in set and applied inspiratory to expiratory ratio. The comparably long inspiratory times resulted in the largest V_T consequently. However, in most of the other devices, we found a relevant difference between P_{insp} and P_{pul} persisting in the last 10% of inspiration particularly at high breathing frequencies. This resulted in large variations in delivered V_T.

We found that the ventilators compensated for leakage during PCV but not to the full extent during VCV. Continuous flow regulation to targeted inspiratory pressure during PCV accounts for this advantage.[3] The observed greater effect of leakage in the infant and toddler models may be caused by the greater absolute leakage volume in relation to the fresh gas flow, which was the same in all models. Moreover, continuous flow regulation and peak flow during PCV may also account for the observed fast stabilization of tidal ventilation after connection of the AWS to the lung model (fig. 6). These observations may reflect the devices' performances after abruptly opening and reconnection of the breathing circuit.

Individualized ventilation strategies require V_T to be delivered as precisely as possible. Although there is only weak evidence on best V_T for perioperative ventilation of lung healthy children, current guidelines recommend V_T less than 10 ml · kg^{-1} ideal body weight.[24] The range of 10% acceptance of variation in the current study would translate into a difference of ±1 ml · kg^{-1}. Consequently, our results suggest applying caution when using VCV with the set V_T equal or less than 20 ml, as excessively larger volumes may result in devices of certain manufacturers. For larger V_T, effectively delivered volumes were mostly lower than set. The risk of volume trauma resulting from a deficit in technical precision will be of lower relevance in older children accordingly but should always be guided by careful clinical examination of the inspiratory thoracic expansion in the very small children.

The V_T displayed at the ventilator reflected the delivered individual volumes in a few cases only. From a clinical perspective, this discrepancy was of particular concern, as tidal volume is a key target in lung-protective ventilation and misleading display of V_T may misguide clinical decisions. Too-low V_T may lead to lung derecruitment and atelectasis,[11] whereas too-high V_T can result in volutrauma or hypocapnia with severe consequences on cerebral vasoreactivity.[13] In the view of these complications, our findings highlight the need for improved V_T measurement accuracy by the AWS manufacturers. Beyond the differences in quality of flow measurement, the positions of the flow sensors may explain the observed differences. Flow is typically measured in the expiratory and the inspiratory limb. Between the flow sensors and the patient's lungs, the connection of tubing, filters, and adapters may provide leaks, leading to the volume calculated from flow to differ significantly from that actually reaching the patient's lungs. Moreover, flow measurements are affected by changes of temperature, humidity, and composition of the breathing gas.[25] As a consequence, measured raw flow data have to be corrected using complex algorithms. In this regard, our results support the need for precise flow measurement close to the patient. We performed flow measurements with a mainstream flow sensor at the tracheal tube to exclude any leakage other than the intended one. We considered effectively delivered V_T best reflected in the expiration of tempered and humidified breathing gas at undiluted concentrations of carbon dioxide. The discrepancies found between the set, the delivered and the displayed V_T support the suggestion of other authors to make flow measurements, independent from the ventilator, a prerequisite for clinical studies on respiratory mechanics in young children.[26] Moreover, AWS manufacturers should feel encouraged to provide mainstream flow measurement for neonates, infants, and children,[27] as it is available in intensive care ventilators to improve quality of ventilation in this particular group.

Of note, differences between the set and the delivered ventilation parameters have also been observed with ventilators intended for use in the neonatal and pediatric intensive care medicine.[28,29] These data suggest that inaccuracies may not only occur between different ventilator types but also between ventilators of the same type with different frequencies and durations of use. Our study included only a single device per type, professionally calibrated in advance of the experiments. We can therefore not exclude intertype differences. Furthermore, one has to consider that severe differences in effective alveolar ventilation may result from using different ventilators in the same individual, although the same ventilation parameters are set. Such differences were reported for the switches from surgery to transport and then neonatal ICU ventilation.[10] Therefore, our findings may have an impact on the benefit–risk assessment for critically ill newborns being operated on in theaters or in the neonatal intensive care.[30]

Limitations of the Study

With our study, we aimed to present an initial comparison of AWS performance in the pediatric setting. The physical

patient models were designed to generate reproducible conditions for all tested devices. We used a glass bottle filled with copper wool as a model of the respiratory system. This model does not include nonlinear respiratory system mechanics. Furthermore, changes in patient respiratory system mechanics during the ventilation or clinical interventions like changes of the fresh gas flow were not investigated within this framework.

The I:E ratio depends on the compliance and resistance of the respiratory system, and longer inspiratory times may be advantageous in terms of lung recruitment. For the purpose of the study, we decided to choose the most frequently preset value of the AWSs, which covers the range for all age-dependent models investigated. Further, with respect to the small ETTs used in this study, we chose a ratio of 1:2 to eliminate the risk of intrinsic PEEP, which may have potentially influenced the measurements.

Regarding the substitution of the inner compartment of the lung model with carbon dioxide, volume constancy and homogenous mixing of the breathing gas underwent extensive preliminary experiments. Respective control measurements, however, were not possible during the final experiments due to the necessity of an interruption of the ventilation. Therefore, we cannot exclude minimal changes in volume and gas mixing within the inner compartment, but we would not expect such changes to have influenced our measurements.

Further, we did not use an HME. Any additional airway item would contribute to the performance of the total system, typically by adding unfavorable properties, such as increased resistance and deadspace. As a consequence, the relative differences in performance of the systems would have been less apparent when using an HME. Therefore, we omitted airway items other than the essential ones to emphasize the differences between the devices.

Conclusions

Precision of ventilation of young children with modern anesthesia workstations varies depending on the ventilator type, ventilation mode, and set parameters. Our results imply precision superiority of pressure-controlled ventilation over volume-controlled ventilation in preterm babies and neonates, independent from the ventilator technology in use. Setting ventilation with respect to lung protection should account for device-specific characteristics.

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Competing Interests

Dr. Spaeth received a fee as a speaker at a workshop on pediatric ventilation financed by the Dräger Medical Company (Luebeck, Germany) during the 15th European Congress on Pediatric Anesthesiology in Berlin, Germany, in 2025. Dr. Schumann received funding from the Deutsche Forschungsgemeinschaft (Bonn, Germany). The other authors declare no competing interests.

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Supplemental Digital Content

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