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Evaluation of heated humidifiers for use on intubated patients: a comparative study of humidifying efficiency, flow resistance, and alarm functions using a lung model

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Abstract *Objective:* Evaluation of humidification efficiency, flow resistance, and alarm functions of heated humidifiers (HH;(Kendall-Aerodyne-delta, Fisher&Paykel-MR 730; Dräger-Aquapor; Puritan-Bennett-Cascade II) in accordance with ISO/EN-8185:1997 and on a ventilated lung model in accordance with ISO/EN-9360:2000. *Methods:* Humidification efficiency was evaluated by (a) measuring the water content of the inspiratory air on perfusion with different gas flows, (b) measuring the water loss of a lung model, and (c) simultaneous measurement of the in- and expiratory water content with a capacitive hybrid sensor. The resistance characteristics were measured, the data were compared with a mathematical approximation. The alarm functions were determined. *Results:* The humidification efficiency of HHs is a function of gas flow and design characteristics. In HHs with tube heating it is possible to make settings at which the inspiratory humidity falls below the minimal value of 33 mgH_2O/l stipulated by ISO/EN-8185:1997. The inspiratory resistances extend from 0.5 to 4.4 cmH_2O l^{-1} s^{-1}; the expiratory flow resistances of the devices are low. The alarm functions of HHs with tube heating are inadequate for cases involving both "dry start" and "running dry." *Conclusions:* Efficiency data that allow a direct comparison with heat and moisture exchangers data according to ISO/EN-9360:2000 can also be determined for HH. HH do not prevent pulmonary water losses in intubated patients. These losses can exceed the physiological range. The airway resistance of the Cascade II prohibits its use in spontaneously breathing patients. The warning and shut-off features of HH are unacceptable and hazardous.

Keywords Heated humidifier · Airway resistance · Humidification efficiency · Alarm functions · Pulmonary water loss · Capacitive hybrid sensor

Introduction

Intubation and ventilation with dry inspiratory gases result in a distal shift of the isothermic saturation border of the respiratory gas humidity combined with higher unphysiological water losses from the lower airways [1, 2, 3, 4, 5, 6, 7]. An increased incidence of pulmonary complications must be expected even after a brief ventilation period, and to an even greater degree with increasing duration of ventilation [8, 9, 10, 11, 12]. Therefore in patients who are ventilated for a long period of time, humidification systems are used to avoid damage to the respiratory epithelium by ensuring adequate water vapor and heat contents of the inspiratory air. In many cases the conception exists that the highest possible water vapor and heat content of the inspiratory air is synonymous with optimal functioning of the respiratory epithelium. Accordingly, active humidifiers (heated humidifiers,

HH) which add water vapor and heat from external sources to the inspiratory air are preferred – particularly in patients with pulmonary impairment. According to the International Standard Organization (ISO) EN-8185:1997, HH should ensure a water content of at least 33 mgH_2O/l in the inspiratory air (=75% of the saturation humidity of 44 mgH_2O/l at a body temperature of 37°C). To avoid heat damage to the trachea the humidifier's heating unit should shut itself off at temperatures above 41°C.

The present study investigated the efficiency of HHs under various conditions of use. In this context the methods used were oriented toward the currently valid testing procedures for active and passive humidifiers. Since comparative measurements directly on the ventilated patient are difficult, the medically relevant characteristics were investigated on a lung model under standardized conditions. The measurements of the water content of the respiratory air were additionally performed with the aid of a novel hybrid sensor, which also allows high-resolution measurements of the water content in gases saturated with water vapor [13]. The in- and expiratory flow resistances were determined on a lung model in the range of ±1 l/s and described with a mathematical approximation with two coefficients. Possible malfunctions of the investigated HH resulting from technical deficiencies or operator error were analyzed.

Material and methods

Four HH with different design characteristics were investigated: Aerodyne delta (Kendall, Neustadt/Danube, Germany), design identical to SAB (Technomed, Kelheim, Germany; MR 730 (Fisher&Paykel, Auckland, New Zealand); Aquapor (Dräger, Lübeck, Germany); Cascade II (Puritan-Bennett, Carlsbad Calif., USA). The Cascade II is in principle a "bubbling" device, i.e., the vaporization surface is enlarged by introducing the inspiratory gases under the water's surface. The Aerodyne delta and MR 730 devices enlarge the vaporization surface by means of a wick made of blotting paper. In the Aquapor device a special float provides an improved distribution and humidification of the inspiratory gases in the humidification (moisturising) chamber. In the Cascade II, Aerodyne delta, and MR 730 devices the inspiratory gas temperatures can be preselected. In the Aquapor device the inspiratory gas temperature and the resulting humidity can only be set on a dimensionless scale from 1 to 10; the last two fields of which are highlighted in red. The temperature display and monitoring takes place, if at all, in the employed respirator itself.

The Aerodyne delta and MR 730 devices have integrated tube heating. The inspiratory gas temperature desired at the tube is set on these devices and held constant by controlling the tube heating system and the water reservoir temperature. In addition to this parameter, the water bath temperature can be set above or below this value with a so-called "humidity regulator." This results either in a decrease in the inspiratory water content or in condensation of water in the tube system. In the MR 730 device the water bath temperature can be reduced by a maximum of 5°C or raised by a maximum of 2°C with reference to the inspiratory gas temperature. In the Aerodyne delta these limits are each 5°C.

Measurement of the humidification efficiency

Perfusion at a constant flow rate

The HH with adjustable inspiratory gas temperatures were operated at 34 and 37°C; in the Aquapor device the highest water bath temperature below the "red" region of the scale was selected. In the MR 730 and Aerodyne delta devices the humidity regulators were set to their maximum values to achieve the highest possible water vapor saturation. Dry air (<1 mgH_2O/l) from the hospital central gas pipeline system was perfused through the humidifier at a constant flow rate of 10–60 l/min via a flowmeter. The measurements of the water content were performed using a dew-point mirror hygrometer (APS 1200, General Eastern, Watertown, N.Y., USA), which was installed in the bypass of the inspiratory gas flow. The measuring equipment was located in an incubator heated to 38°C to avoid measured value falsification due to bedewing of the dew-point mirror hygrometer.

Interruption of the gas flow

To test the technical safety with regard to the output of hot inspiratory gases subsequent to interruption of ventilation ("hot shot"), the systems were perfused with a continuous gas flow of 30 l/min for 30 min; then the gas flow was interrupted for 30, 60, 120, and 300 s. Subsequently the water bath temperatures, the gas temperatures near the tube, and the humidity at the tube were measured immediately after the interruption. Temperatures were obtained with thermistor probes (Yellow Springs Instruments, Series 400), immersed below the water level of the HH or in the inspiratory gas flow at the Y-piece, respectively.

Ventilation in the lung model

To make comparative, systematic measurements of the humidification behavior of the system under clinical conditions, a lung model whose mode of functioning is oriented to physiological conditions was used [14]. The model consists of a wick vaporizer with a heated and temperature-regulated water bath and a lung simulator (LS 1500, Dräger, Lübeck, Germany). The water bath's temperature ("lung temperature") measured by a thermistor probe (Yellow Spring Instruments, Series 400) was set to 37±0.5°C.

Water content measurements in the inspiratory air were performed using volume-controlled ventilation of the lung model with a Siemens Servo 900 C Respirator (tidal volume 1000 ml; ventilation frequency 10/min and inspiratory flow 30 l/min, no end-tidal plateau). The water content amplitude of the respiratory gases (difference between inspiratory and expiratory water content) at the tube were determined with a capacitive hybrid sensor [13]. After 4 h of ventilation the "pulmonary" water loss of the lung model was determined by weighing and then standardized to liters of respiratory gas. The humidifiers were operated at preselected inspiratory gas temperatures of 34 and 37°C. The effect of the manual humidity correction – caused by the setting of the humidity regulator – on the water content of the inspiratory air was determined at the minimum, maximum and normal positions (only MR 730 and Aerodyne delta).

Flow resistances

Inspiratory resistance characteristics were measured to describe the resistance behavior of the humidifiers. The devices were exposed to constant flows up to 1 l/s, starting at 50 ml/s. Using a differential pressure transducer (Micro Switch, Freeport, Ill., USA) the pressure difference across the system was measured against atmospheric pressure. The flow increment was set to 50 ml/s. Thus

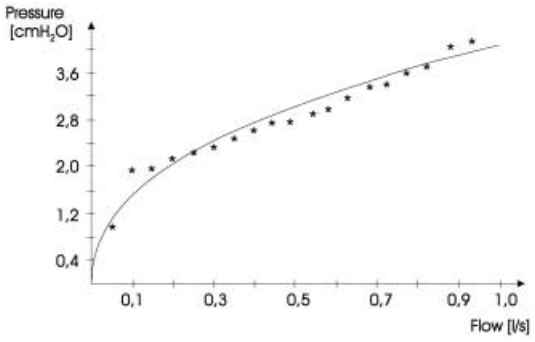

Fig. 1 Inspiratory characteristics of flow resistance for a HH, for example. The measured values and the calculated coefficients *K*1 and *K*2 are given at a flow of 1 l/s. *X-axis* Inspiratory gas flow of 0–1 l/s; *y-axis* resulting pressure (cmH_2O). *Asterisks* Measured values; *solid line* calculated curve

the characteristics could be described by a total of 21 pairs of values including the 0 value (see Fig. 1). The coefficients of the formula $\Delta p = K1 \times \dot{V}^{K2}$, with which the resistances for the investigated flow range can be described, can be determined from these pairs of values by an approximation method in accordance with the Gauss-Newton method. In this context, *K*1 is the value of the resistance at 1 l/s, and the exponent *K*2 is a measure for the "nonlinearity" of the characteristic, for which, for example, the value 1 means that the characteristic of the curve is linear and the value 2 as exponent describes a quadratic course (see "Appendix").

The constant flows were generated by a lung simulator, which was actuated by special software that we developed. It is based on the ASYST 4.0 programming language (Asyst Software Technology, Rochester, N.Y., USA). Actuation of the lung simulator and recording the measured values for flow and differential pressure were performed with the aid of a D/A-A/D card (DT 2801, Data Translation, Marlborough, Mass., USA) and an IBM-compatible computer.

The resistances of the Aquapor and Cascade II devices were measured with a 127-cm-long piece of original silicone tube without tube heating but including water traps. For the Aerodyne delta and MR 730 devices 103-cm-long original hoses with tube heating were used. In the Cascade II device the measurements were additionally performed for various degrees of filling. During measurements of the inspiratory resistances, the expiration limb was closed so that the flow was completely directed through the inspiratory tube.

For comparison the resistance profiles of endotracheal tubes (Portex blue-line, Portex, Hythe, Kent, UK) with various internal diameters (6–9 mm) were determined.

Alarm functions

In the lung model (ventilation parameters: tidal volume 1000 ml, ventilation frequency 10/min, flow 30 l/min, inspiratory to expiratory ratio 1:2, no expiratory plateau) were determined, whether and after what period of time alarm or fault messages were displayed in the following situations: installation and initiation of operation without water; delayed refilling and system running dry; overheating after interruption of operation.

Results

Humidification efficiency

Humidification efficiency for perfusion with constant flow rate

The humidification efficiency of the MR 730 exhibits only a very low flow dependence at both temperature settings. In contrast, higher gas flow rates resulted to a distinct drop in the water content in the Cascade II, Aerodyne delta, and Aquapor devices. In the process, the Aerodyne delta and Aquapor devices dropped below the water content of 33 mg/l of inspiratory air stipulated by the ISO/EN-8185:1997 Standard at 34°C and higher flow rates. In addition, the Aquapor device exhibited substantial humidity variations up to 8 mg/l, which could be traced to periodic oscillations of the water bath's heating unit at 4- to 5-min intervals (Table 1).

Table 1 Humidification efficiency (mgH_2O/l) as a function of flow (l/min) at 34°C and maximum setting of the humidity regulator (the target content 37.6 mgH_2O/l) and at 37°C (t target content 44.0 mgH_2O/l) (*parentheses* proportion of target value)

	10 l/min	20 l/min	30 l/min	40 l/min	50 l/min	60 l/min
34°C						
Aerodyne delta	33.4 (89%)	31.0 (82%)	29.8 (79%)	27.3 (73%)	26.3 (70%)	24.9 (63%)
MR 730	34.2 (91%)	34.1 (91%)	34.3 (91%)	34.4 (91%)	33.9 (90%)	33.4 (89%)
Cascade II	38.0 (101%)	37.7 (100%)	37.8 (101%)	38.5 (102%)	35.8 (95%)	34.2 (91%)
Aquapor[a]						
Minimum	37.7 (100%)	34.2 (91%)	32.4 (86%)	29.3 (78%)	28.0 (74%)	26.2 (70%)
Maximum	34.5 (92%)	30.2 (80%)	24.6 (65%)	21.6 (57%)	20.2 (54%)	18.2 (48%)
37°C						
Aerodyne delta	37.7 (86%)	34.6 (79%)	33.1 (75%)	32.4 (74%)	31.0 (70%)	28.1 (64%)
MR 730	36.1 (82%)	36.2 (82%)	36.2 (82%)	36.2 (82%)	36.2 (82%)	36.2 (82%)
Cascade II	44.1 (100%)	43.9 (100%)	44.1 (100%)	38.8 (88%)	36.1 (82%)	34.6 (79%)
Aquapor[a]						
Minimum	37.6 (85%)	34.1 (78%)	32.4 (74%)	29.0 (66%)	28.1 (64%)	27.1 (62%)
Maximum	34.6 (79%)	30.2 (69%)	24.9 (57%)	22.1 (50%)	20.0 (45%)	18.1 (41%)

[a] Setting the temperature was impossible on the Aquapor device; in addition, large variations in the humidification efficiency were observed for the same device setting. Therefore their minimum and maximum values are given

Table 2 Inspiratory water content changes as a result of alteration in humidity setting. Continuous flow through the HH with 20 or 60 l air/min (*parentheses* proportion of the maximum value)

	Minimum	Normal	Maximum
Aerodyne delta			
34°C			
20 l/min	24.1 (64%)	29.7 (79%)	31.0 (82%)
60 l/min	20.3 (54%)	24.0 (64%)	24.9 (66%)
37°C			
20 l/min	26.3 (60%)	31.5 (72%)	34.6 (79%)
60 l/min	24.8 (56%)	27.2 (62%)	28.9 (66%)
MR 730			
34°C			
20 l/min	26.2 (70%)	32.2 (86%)	33.9 (90%)
60 l/min	22.5 (60%)	29.5 (78%)	33.5 (89%)
37°C			
20 l/min	28.4 (65%)	36.0 (82%)	36.0 (82%)
60 l/min	25.6 (58%)	36.2 (82%)	36.0 (82%)

Changing the settings of the humidification regulator resulted in measurable alterations in the water content. Aerodyne delta and MR 730 provided water content values well below the 33 mg/l when set to their minimum humidity setting (Table 2). The water content of the inspiratory air was practically independent of the level of the water reservoirs. Between the minimum and maximum filling levels the difference was less than 3 mg water/l inspiratory air.

Ventilation of the lung model

In Table 3 the water loss values (ΔFw), which were determined by weighing after 4 h of ventilation and standardized to 1 l respiratory gas, and the humidity amplitudes water loss values (ΔFs) measured close to the tube by means of the heated sensor. The values measured close to the tube were closely correlated with the water losses calculated from weighing data. Additionally, the manual correction of the humidity with the aid of the humidity regulator (MR 730 and Aerodyne delta) also proved to have a substantial influence on the absolute water content of the inspiratory air (Table 3).

Reproducible inspiratory humidity values that were above the theoretically possible values at 34° and 37°C were measured with the Cascade II. This discrepancy was also found in a similar manner in an other Cascade device that was tested for control purposes. This fault was traced back to inadequate functioning of the temperature regulation system. The elevated inspiratory gas temperature also explained the low water losses of the model. Indeed, on setting the inspiratory gas temperature to 37°C, instillation of water into the lung model occurred (Table 3).

Table 3 Water loss of the lung model after 4 h of volume controlled ventilation (V_T 1000 ml, flow 30 l/min, RF 10/min, no inspiratory pause) and respiratory gas: ΔFw was calculated from the weight difference of the model, ΔFs from the difference between the end expiratory and end inspiratory humidity at the tube

	Water loss (mg/l)	
	Δ Fw	Δ Fs
Aerodyne delta		
34°C		
Minimum	7.9	8.0
Normal	7.0	7.3
Maximum	7.9	7.4
37°C		
Minimum	3.5	3.7
Maximum	4.7	4.1
MR 730		
34°C		
Minimum	14.3	16.0
Normal	7.7	7.5
Maximum	6.2	5.9
37°C		
Normal	4.9	4.3
Maximum	3.8	3.0
Aquapor[a]		
1	12.9	13.4
6	6.6	6.5
10	3.4	4.0
Cascade II		
34°C	2.4	3.0
37°C	+1.1	+4.4

[a] Setting of the water bath temperature using the dimensionless scale from 1 to 10.

Flow resistances

The inspiratory flow resistances of the MR 730 and Aquapor devices revealed similar values, 1.3 and 1.5 $cmH_2O\ l^{-1}\ s^{-1}$, respectively. The Aerodyne delta exhibited the lowest resistances with 0.5 $cmH_2O\ l^{-1}\ s^{-1}$. The MR 730, Aerodyne delta, and Aquapor devices showed no dependence on their degree of filling. The approximation procedure was not used for the Cascade II device because the characteristic had an inconstant course in the region of the origin due to the bubble pipe which is immersed below the water's surface. At a flow rate of 1 l/s the maximally filled Cascade II device exhibited the highest measured inspiratory flow resistance with 4.45 cmH_2O. For a minimum filling level of the water reservoir, the measured resistances were approximately 20% lower (Table 4). The utilized tubing systems themselves exhibited a flow resistance of 0.2 cmH_2O at a flow of 1 l/s.

Table 4 Calculated coefficients of the resistance characteristic $\Delta p = K1 \times \dot{V}^{K2}$ for the flow region up to 1 l/s determined for 21 measuring points and measured real values: *K*2 characterizes the form of the characteristic, *K*1 its slope and the pressure drop for a flow of 1 l/s

	*K*1 ($cmH_2O\ s^{-1}\ l^{-1}$)	Real measured value at 1 l s^{-1}	*K*2
Aerodyne delta	0.50	0.46	2.11
MR 730	1.43	1.32	2.12
Aquapor	1.60	1.53	1.81
Cascade 2 min. filling level	–	3.38	–
Cascade 2 max. filling level	–	4.36	–
Tube system, 127 cm, without heating	0.21	0.20	2.15
Endotracheal tube			
∅ 6.0 mm, 31 cm	17.11	16.12	1.98
∅ 6.5 mm, 32 cm	12.58	11.93	1.98
∅ 7.0 mm, 32 cm	8.58	8.26	1.93
∅ 7.5 mm, 33 cm	5.64	5.51	1.86
∅ 8.0 mm, 34 cm	3.98	3.88	1.79
∅ 8.5 mm, 34 cm	2.74	2.65	1.82
∅ 9.9 mm, 34 cm	2.02	1.94	1.79

Table 5 Water content close to tube, inspiratory gas temperature and water bath temperature after ventilation interruption of varying time. The humidifiers were operated at 37°C and maximal humidity setting (inspiratory gas flow 0.5 l/s)

Ventilation interruption (s)	Water content (mg/l)	Inspiratory gas temperature at tube (°C)	Water bath temperature (°C)
Cascade 2			
30	48.5	38.5	47.0
60	49.0	38.5	47.0
120	52.5	38.5	49.0
300	58.0	41.6	53.0
Aquapor			
30	40.0	34.0	51.5
60	34.5	32.1	51.0
120	38.0	32.5	53.0
300	40.5	35.0	54.0
MR 730			
30	42.0	36.2	54.0
60	45.0	37.1	53.0
120	48.5	38.8	55.5
300	47.5	38.0	54.5
Aerodyne delta			
30	42.0	35.1	53.0
60	45.0	36.2	54.0
120	45.0	36.0	54.0
300	36.5	33.3	53.0

In Table 4 the flow resistances are given for the HH investigated and the isolated tubing systems. For comparison resistance measurements were performed with endotracheal tubes of different lengths and diameters. The coefficients of the mathematical approximation are also given (see "Appendix").

Alarm functions

Subsequent to interruption of ventilation for 300 s the Cascade II device with 41.6°C exhibited the highest inspiratory gas temperature (Table 5). No alarm response occurred; the humidifier did not shut itself down. The inspiratory gas temperature always remained under 39°C in the other models.

Safety deficiencies were observed in cases of faulty operation. For example, no alarm occurred on initiating operation of the MR 730 or the Aquapor device without water or with too little water during operation. Even after 10 h of operation without water neither device either showed an alarm or displayed an error message. In contrast, the Aerodyne delta device displayed an error message, which could be suppressed twice, after initiation of operation without water. Thereafter the device shut itself down. In the case of low water level during operation the device shut itself down after 85 min. The Cascade II sounded alarm in cases involving low water level after

only 60 s. The alarm could not be suppressed and resulted in shut-down of the device.

Discussion

The necessity of conditioning the inspiratory gases of intubated patients is indisputable. HH are frequently favored in intensive care medicine over heat and moisture exchangers (HME). The reason for this is often given as the restricted humidification efficiency of HME. HME can only add as much heat and moisture to the inspiratory air as they have reversibly stored in the previous expiration phase. Although numerous investigations have shown that effective HME guarantee a physiological conditioning of inspiratory air [14, 15, 16, 17, 18, 19, 20], many users do not consider this to be adequate. Consequently HH are used above all for long-term ventilated patients with impaired pulmonary gas exchange with the objective of adding additional moisture and heat to the inspiratory air. However, previous investigations have already shown that this goal is also not easily achieved by the use of HH per se. Thus it could be shown in a comparative investigation of active humidifiers and HME in long-term ventilated patients that the inspiratory humidification efficiency of HME and HH are comparable if a inspiratory gas temperature of 34°C has been set on the HH [21].

In the present study we observed that the minimum moisture release stipulated by ISO is not always achieved at continuous flow of the HH despite having set the inspiratory gas temperature to 34°C. Certain device settings, for example, through manual humidity correction with the humidity regulator can lead to a further drop in the inspiratory humidification efficiency. However, the clinical relevance of this effect or possible undesirable effects on the respiratory epithelium cannot be determined because medical science does not yet know which water content in the inspired air is to be considered to be physiological and thus can be considered as being adequate. Accordingly, controversial statements as to the required water content of inspiratory aid are found in the literature. The interpretation of the data is considerably easier if the investigation of the efficiency is conducted using the test stipulations for HME according to ISO/EN-9360:2000 [22]. They are based on the determination of the water losses from the lower airways, which can be equated with pulmonary water loss in the intubated patient. According to the investigations of Ingelstedt [3, 4] and Déry's [1, 2], it amounts to approximately 7 mgH_2O/l for smooth nasal respiration.

The pulmonary water loss determined by weighing is a parameter that is easily interpreted in the ventilation of a lung model: the higher the water loss that has been standardized to 1 l of respiratory gas, the poorer the inspiratory humidification efficiency of the humidification system applied. When using this test specification, which is oriented to physiological conditions, on active humidifiers, one observes that the water losses of the model are determined to a considerable degree by the setting of the HH. Accordingly, the temperature indicator on the device alone does not allow any conclusions about the effective water content of the inspiratory air and thus about the patient's pulmonary water loss. For example, the water loss from the model lung determined by weighing was, for example, between 6.2 and 14.3 mgH_2O/l of respiratory gas for an identical temperature setting on the device.

The use of a hybrid sensor also allows a statement to be made about the efficiency of the active humidifier in the clinical ventilation situation directly on the ventilated patient [13, 14, 20, 21]. In addition to testing the humidification efficiency, an exact setting of the humidifier on the patient in correlation to physiological conditions is concomitantly possible. However, the hybrid sensor used in this investigation is currently only available as a laboratory measuring device (ZSK-Systemtechnik, Katlenburg, Germany) which is not yet suitable for clinical practice. It still seems appropriate, however, to recommend humidity monitoring for active humidification systems in the future. This also appears useful for a second reason, i.e., because the alarm functions of the investigated humidifiers are deficient. The alarms for faulty operation, for example, the operation without water, are absolutely inadequate. Alarms are apparently made via measurement of temperatures. However, this is particularly problematical and erroneous in systems with tube heating units. The fact that hazards due to overheated inspiratory gases are obviously not to be expected should be positively noted.

The investigations on the lung model show that the water losses at inspiratory gas temperatures higher than 34°C definitely decrease and can be found in the physiological range. Nevertheless, under certain conditions water can be instilled into the lungs (Cascade II). However, whether advantages for the respiratory epithelium of the mucous membranes in the airways result is doubtful, because it is obvious that for an optimal mucociliary clearance function a finely coordinated interaction of evaporation and condensation is required. Just as inspiratory gases that are too dry cause disturbances in the mucous viscosity in ventilated patients, those that are too moist also exhibit this effect and thus result in disturbances of the mucociliary clearance function. Degeneration and adhesion of the cilia as well as frustrane cilia movements have been described. Alterations in the surface of mucous drops and increased secretion volumes with reduced viscosity additionally facilitate the undesirable elutriation of contaminated secretions from the upper tracheal region into the peripheral lung sections. Microatelectases, elevated shunt volumes, decrease in compliance, and increase in resistance are the

consequences. The decrease in the surfactant activity should indeed be more frequent and more distinct than in ventilation with dry inspiratory gases [9, 23, 24, 25, 26, 27].

The respiratory gas flow resistances measured showed good agreement with the data that was determined by mathematical approximation in accordance with the Gauss-Newton method (see "Appendix"). By far the highest inspiratory gas resistances were exhibited by the Cascade II. In ventilated patients the importance of elevated airway resistances due to the tube, Y-piece, ventilation tubes, or humidifier are indeed difficult to assess in individual cases. However, they should be kept as low as possible in any case, since they may result in an elevation in the in- and expiratory work of breathing in spontaneously breathing patients. It should be taken into account that the effect of a humidifier on the work of breathing (WOB) is directly related to the position of the ventilator sensors and the type of humidifier. In the case of airway sensors on the expiratory side (Siemens 300, Puritan Bennett 7200/840, Draeger Evita 4, Bear 1000 etc) the HME but not the active humidifier is in the path of gas flow to effect WOB. However, in the case of the airway sensor on the inspiratory side (Hamilton Veolar, Siemens 900, Puritan Bennett MA1) both the active humidifier and the HME increase WOB and effect triggering. In these cases the resistance of the humidifier (HME and active humidifier) may not exceed 2 $cmH_2O\ l^{-1}\ s^{-1}$ of gas flow, according to the ISO/EN-8185:1997. In light of this, the use of the Cascade II in spontaneously breathing patients is not to be recommended if the flow or pressure sensors for ventilator triggering and monitoring is on the inspiratory side, especially since the flow resistances add to the resistances of the tube systems and the tube.

A general criticism with regard to the setting of parameters of active humidification systems appears appropriate. Unfortunately, physicians and nurses of intensive care units often do not know the function of the humidity correction control knob (MR 730 and Aerodyne delta). Correspondingly high is the risk of an improper setting, from which an insufficient condition of the inspiratory gases can result. Dispensing with this additional and indeed unnecessary adjustment function would thus be appropriate and desirable. Inadequate knowledge also exists with regard to the optimal inspiratory gas temperature, which is adjustable in a wide range. A permanent default temperature setting of, for example, 37°C would contribute to simplification and simultaneously increase patient safety. In any case a clinical necessity for the reduction or even elevation in the temperature to a level above body temperature does not exist. This has indeed been repeatedly propagated for the avoidance of heat losses and even for rewarming hypothermal patients, but is ineffective and increases the risks involved [21, 28, 29, 30, 31].

Conclusion

Active humidification systems result in pulmonary water losses in intubated patients, just as HME do. The water losses determined by valid international test specifications can, depending on the device setting, be found considerably outside the physiological range. In future, possibilities of optimization exist by monitoring the water content in the inspiratory air with novel capacitive hybrid sensors. The high flow resistances of the Cascade II prohibit its uncritical use in spontaneously breathing patients. The warning and shut-off behavior of all of the devices is unacceptable and hazardous to patients.

Appendix

Thermodynamic basis

At equilibrium at a given temperature the same vapor pressure (saturation vapor pressure) always exists above a water reservoir, in this case, above the mucous membranes of the airways. The number of water molecules liberated from the water due to molecular movement (vaporization or evaporation) is exactly the same as the number of molecules absorbed by the water (condensation). The maximum possible water content of the gas phase or the water vapor saturation pressure is thus determined by the temperature in an unequivocal manner. The relationship is characterized by thewater vapor pressure curve and the water content curve derived from it.

If the temperature is increased, the saturation vapor pressure and thus the water content in the gas phase increases. If the temperature is decreased, condensation occurs, because the temperature has fallen below the dew point. The condensed water is visible in the air as aerosol or fog, whereas water vapor is invisible.

The ratio of actual, absolute atmospheric humidity, F_{abs}, to the maximum possible humidity in a gas or gas mixture is termed the relative humidity, F_{rel}. This is defined as the proportion of the actual partial pressure, p_D, of the water vapor in the gas volume and the maximum possible saturation vapor pressure, p_S:

$$F_{rel} = p_D / p_S \times 100\ (\%)$$

Without concomitant specification of the temperature the specification of the relative humidity thus does not allow any statement to be made about the water content of the air.

Determination of the flow resistance by mathematical approximation

The measurement principle is based on the conception that the flow resistance of the system is characterized by

the pressure drop across this system when air passes through it. In the most simple case there is a linear relationship between the applied flow and the resulting pressure change. The resistance characteristic of this system can thus be described by the following formula if one value pair is known:

$$\Delta p = \dot{\mathrm{V}} \times \mathrm{R}$$

where Δp corresponds to the pressure drop across the system; R, to the flow resistance and $\dot{\mathrm{V}}$ to the flow. Accordingly, the flow resistance R is:

$$\mathrm{R}\,(\mathrm{cmH_2O})\,/\mathrm{l} \times \mathrm{s}^{-1} = \Delta p\,(\mathrm{cmH_2O})\,/\dot{\mathrm{V}}(\mathrm{l/s})$$

For nonlinear pressure-flow relationships, such as those that exist in respiratory gas filters and HME as well as for endotracheal tubes this formula algorithm is not valid for the entire flow range, but merely for a defined value pair, for example, for a flow of 1 l/s. Since, however, under clinical conditions the perfusion occurs with changing flow velocities and directions, the resistance profile of such systems is not adequately described by a single value pair. Consequently, mathematical approximations, with whose aid model-specific resistance characteristics can be prepared, are required for the assessment of the perfusion behavior in the clinically relevant range. Nonlinear pressure-flow relationships can be characterized by the following calculation algorithm [32, 33]:

$$\Delta p = K1 \times \dot{\mathrm{V}}_{K2} + K3 \times \dot{\mathrm{V}}_{K4} + \ldots + K_n \times \dot{\mathrm{V}}_{Kn+1}$$

The coefficient *K*2 characterizes the form of the characteristic and *K*1 its slope as well as the pressure drop for a flow of 1 l/s (Fig. 1). For deviations less than 0.1 $\mathrm{cmH_2O}$ it is generally sufficient to calculate the formula only for the first term, i.e., up to *K*2 [33, 34].

The Venturi effects, which are determined by the measurement technique and are unavoidable, at the pressure recording site are quantified in a blank experiment without the humidification system. The resulting coefficients were calculated and served as a correction factor in measurements with the gas-conditioning system.

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