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TECHNIQUE

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An experimental set-up to test heat-moisture exchangers

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Abstract *Objectives:* The purpose of this study was to build an experimental set-up to assess continuously the humidification, heating and resistance properties of heat-moisture exchangers (HMEs) under clinical conditions.
Design: The experimental set-up consists of a patient model, measurement systems and a ventilator.
Setting: Surgical ICU, University Hospital of Rotterdam.
Materials: A clinically used HME.
Measurements and results: The air flow, pressure in the ventilation circuit, pressure difference over the HME, and partial water vapour pressure and temperature at each side of the HME were measured. The resistance, absolute humidity, humidification efficiency and temperature difference at the patient side of the HME were calculated. Measurements were performed during 24 h. The temperature output, humidity output and lung mechanics of the patient model were similar to values found in mechanically ventilated patients. The measurement system was in agreement with the ISO draft standard and was capable of measuring dynamic variation of water and heat exchange over the range of a clinically used ventilator setting.
Conclusion: The experimental set-up described is reliable for evaluating HMEs and can also be used for future clinical evaluation of HMEs. The main advantages of this set-up over those described previously are: (i) measurements of dynamic variations of water and heat exchange; (ii) on-line measurements of expiratory, as well as inspiratory resistance.

Key words Humidity · Heat and Moisture exchangers · Mechanical ventilation · Mass spectrometry · Temperature · Resistance

Introduction

The heat and humidity exchange functions of the nose and upper airways are bypassed during endotracheal intubation and tracheostomy. Use of dry medical gases for mechanical ventilation increase the water and heat loss from the lower airways, which can produce serious airway damage and worsen pulmonary function [1–5]. Heated humidifiers (HHs) or heat-moisture exchangers (HMEs) are therefore used to both heat and humidify the air before being delivered to the patient. HHs, however, have some disadvantages such as the potential to deliver excessive heat with the consequent problem of thermal injury and of acting as a reservoir for bacterial growth resulting in nosocomial infections [6].

The HME filter is a simple solution to the problems of humidification of inspired gases and contamination of ventilatory circuits. HMEs are passive devices which absorb the expiratory moisture and heat, and return it partially to the patient at the next inspiration [7, 8]. However, previous publications have reported some drawbacks of

the use HMEs such as inadequate humidification and heating efficiency, high resistance to airflow, clogging by sputum and endotracheal tube occlusion [8–12]. Nonetheless, continued developments of the HMEs have improved their heating and humidification efficiency, decreased their resistance to airflow and improved their qualities as a bacterial-viral filter [8, 13–15]. A variety of HMEs are available with different physical properties and various experimental set-up models have been developed to evaluate these properties [7–10, 13, 16–22]. The International Organization for Standardization (ISO) has released a draft standard for testing HMEs [18]. This draft standard specifies the minimum requirements of a patient model and measurement system to test HMEs. An ideal experimental set-up should not only imitate the respiratory properties of mechanically ventilated patients but also be able to record dynamic variations of humidity, temperature and resistance to calculate more accurately the performance characteristics of HMEs in time.

The present study concerns the design and validation of an experimental set-up to test HMEs in accordance with the technical standards of the ISO and with the ability to measure dynamic variations of humidity, temperature and resistance.

The aims of the study are: (i) to construct a patient model which imitates a mechanically ventilated patient in terms of compliance, resistance, expiratory temperature and humidity output; (ii) to measure dynamic variations of flow, pressure, pressure difference, temperature and humidity continuously, thereby allowing intra- and inter-breath interpretation of the results; (iii) to calculate inspiratory and expiratory flow-weighted mean values of humidity and temperature to assess the performance characteristics of the HME (efficiency of humidification and heating); (iv) to determine the intrinsic properties of the experimental test system with a commonly used ventilator setting and to validate the set-up during long-term measurements with a commonly used HME.

Materials and methods

The experimental set-up

The experimental set-up includes a patient model, measurement systems and a ventilator.

The patient model consisted of a 1 l training thorax (Übungsthorax, M 13333, Dräger, Germany), a HH (Conchatherm 3, Kendall Company Limited, London, UK), standard ventilatory tubing, two one-way valves, connectors and an incubator (Intensivpflege Incubator 6500, Drägerwerk AG, Lübeck, Germany) as shown in Fig. 1. A calibration bag with a capacity of 650 ml was used during high tidal volume settings to prevent changes in compliance and pressure of the patient model. The output of the patient model was adjusted to produce 100% relative humidity at 34.5±1.0° C. The incubator was kept at 36.0±0.5° C to prevent condensation.

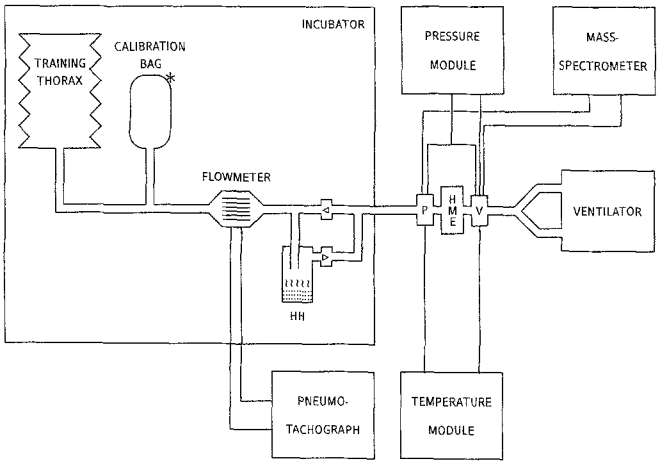

Fig. 1 The measurement set-up. The calibration bag (*) was only used during high tidal volume settings. ◄ denotes the direction of the flow on the one-way valve

A heated flowmeter (Fleisch No. 2, Sensormedics, Bilthoven, The Netherlands), located between the training thorax and the HH, connected to a pneumotachograph (Type 17212, Godart-Statham, Bilthoven, The Netherlands) was used for flow measurements. The flow measurement system was linear between 0–100 l/min, and used to measure the inspiratory and expiratory flow ($\dot{V}_I$ and $\dot{V}_E$).

Two sampling ports were used to introduce the temperature probes, humidity sampling capillary and pressure lines; one sampling port was located between the patient model and HME ("P" site), and the other was located between the HME and Y-piece of the ventilatory tubing ("V" site). Two differential pressure transducers (Hewlett-Packard model 270, HP International, CA) and two signal conditioners (Hewlett-Packard model 8805 B carrier amplifier, HP International, CA) were used to measure the pressure in the ventilation circuit at the "V" site (P_V) and the pressure difference between the "P" and "V" site (ΔP_{HME}). The response time and ranges of the pressure transducers were 5 ms and −40 to 40 cmH_2O, respectively.

Two precalibrated, small bead NTC thermistors (Fenwal Electronics, American Power Devices, MA) and a two channel temperature module (Temperature module 78204 B, Hewlett-Packard, HP International, CA) were used for temperature measurements at the "P" and "V" sites. The accuracy of the temperature module was ±0.2° C. The 0–90% step response times of the thermistors in flowing air were 200 and 300 ms respectively. The temperature probes were calibrated between 20–40° C with a mercury thermometer.

A quadrupole mass spectrometer (MGA 3000, Case, Biggin Hill, UK) was used to measure partial water vapour pressures at the "P" and "V" sites. A transparent, unheated, constricted tip capillary was used for humidity measurements (sampling flow: 40 ml/min). The delay time of the mass spectrometer was 460 ms with a 10–90% step response time of 600 ms.

The barometric pressure was measured by a barometer (Fues barometer, Berlin, Germany) before each measurement and used for calibration of the mass spectrometer.

Bottled helium with a dew point of −30° C (0.038 kPa or 0.34 mg/l humidity) and a vapour generator (Type WG-600 water vapour generator, The Analytical Development Co. Ltd., UK) were used to calibrate the mass spectrometer.

All signals (flow, pressure, pressure difference, temperature and humidity) were amplified (Medium gain differential input DC am-

plifier, Model 56-1340-00, Gould Electronics, OH) and plotted continuously (Gould TA 2000 thermal array recorder, Gould Electronics, OH). All signals were recorded concomitantly on magnetic tapes (Store-14, Racal Instrumentation Recorder, Racal, Southampton, UK). During replay of the tapes, the analog signals were digitized (50 Hz), saved on a PC and corrected for the delay of the mass spectrometer.

The incubator and room temperatures (T_{inc}, T_{room}) were measured by two mercury thermometers with an accuracy of $\pm 0.2°$ C and $\pm 0.1°$ C respectively.

An electronic scale (ED-60T, Berkel, The Netherlands) with an accuracy of ± 0.5 g was used to weigh the patient model and the HME.

A volume controlled ventilator (Dräger-Evita, Drägerwerk AG, Lübeck, Germany) was used to ventilate the patient model with the following settings: an expiratory tidal volume of 1.0 l, a frequency of 10 breaths/min, an I:E ratio of 1:2, and an inspiratory flow of 0.5 l/s. Central medical air with a dew point of $-20°$ C (equal to 0.1 kPa or 0.85 mg/l humidity) was used to ventilate the patient model.

The HME used in this study was the Dar Hygroster (Dar SpA Mirandola, Italy).

Measurement protocol

For stabilization, the patient model was ventilated for 1.5 h without a HME in the system. After attachment of the HME, measurements were made every 10 min in the first hour and continued hourly for 24 h. Between the measurements the signals were plotted continuously. The humidity sampling capillary was placed from one sampling site to the other to measure the humidity at each side of the HME. At regular intervals the calibration, step-response time and delay time of the mass spectrometer were checked and updated if necessary in combination with frequent inspection of the transparent capillary to detect condensation. The weight of the HME and patient model without the incubator were measured before and after the measurement period.

Calculated parameters included (see appendix for formulae): (i) inspiratory-maximum, inspiratory-plateau and end-expiratory pressures at the "V" site ($P(V)_{I,max}$, $P(V)_{I,plat}$, $P(V)_{E,end}$); (ii) Flow-weighted mean inspiratory and mean expiratory partial water vapour pressures at the "P" and "V" sites of five successive breaths ($P_{H2O}(P)_{I,mean}$, $P_{H2O}(P)_{E,mean}$, $P_{H2O}(V)_{I,mean}$, $P_{H2O}(V)_{E,mean}$); (iii) Mean inspiratory and mean expiratory absolute humidity values (equal to the water content of the inspiratory and expiratory air) of five successive breaths in mg/l ($AH(P)_{I,mean}$, $AH(P)_{E,mean}$, $AH(V)_{I,mean}$, $AH(V)_{E,mean}$); (iv) Flow-weighted mean inspiratory and mean expiratory temperature values at the "P" and "V" sites of 5 successive breaths ($T(P)_{I,mean}$, $T(P)_{E,mean}$, $T(V)_{I,mean}$, $T(V)_{E,mean}$; (v) Resistance values: the total resistance of the patient model with the HME (R_{TOT}), mean inspiratory and maximum expiratory resistance of the HME ($R_{I,mean}$, $R_{E,max}$) and the resistance of the patient model (R_{PM}); (vi) Compliance of the patient model (C_{PM}); (vii) Weight differences: Weight gain of the HMEs and condensation in the patient model; (viii) Water losses at the "P" and "V" sites; (ix) Humidification efficiency of the HME (H_{EFF}); (x) Temperature difference at the "P" site of the HME ($\Delta T(P)$) [17].

Results are expressed as mean$\pm$SD over 24 h ($n = 30$). The Kruskal-Wallis one-way analysis of variance test was used for the comparison of averages over time, using a statistical computer program (SPSS/PC+). $p < 0.05$ was considered as being significant.

Results

The compliance and resistance values of the patient model over 24 h were 30.0 ± 1.0 ml/cmH_2O and 10.9 ± 1.5 cmH_2O/l/s respectively. $T(P)_{E,mean}$ and $P_{H_2O}(P)_{E,mean}$ were $34.1 \pm 0.8°$ C and 5.2 ± 0.3 kPa respectively during the whole study (equal to an absolute humidity of 36.5 ± 1.7 mg/l or a relative humidity of $97.3 \pm 3.2\%$). The incubator temperature over 24 hours was $35.8 \pm 0.2°$ C. The condensation in the patient model was only 3 g over 24 h.

Figure 2 shows an example of the continuous recording of the different signals. Figure 3 shows the mean inspiratory and expiratory absolute humidity at each site of the HME. As can be seen in Table 1, there were no significant changes in absolute humidity over 24 h.

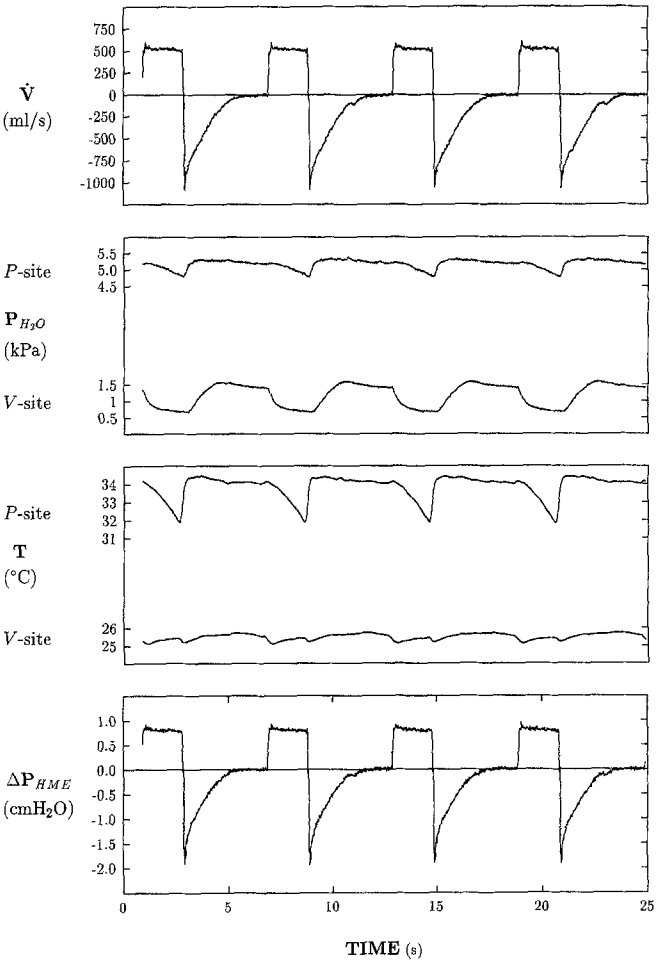

Fig. 2 A continuous registration of the flow ($\dot{V}$), humidity ($P_{H2O}(P)$, $P_{H2O}(V)$: partial water vapour pressure recordings from "P" and "V" sites), temperature ($T(P)$, $T(V)$: temperature recordings from "P" and "V" sites) and pressure difference (ΔP_{HME}: between "P" and "V" sites) signals after correction of the delay times

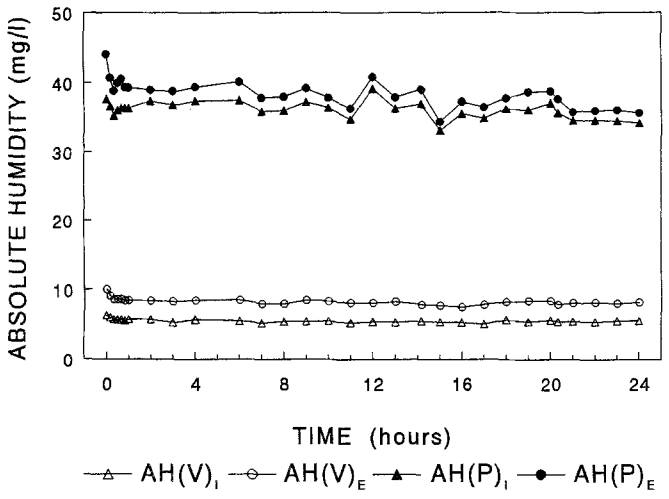

Fig. 3 Inspiratory and expiratory absolute humidity values over 24 h measured at the "P" site and "V" site

Table 1 The measured and calculated parameters during the 24 h study (mean ± SD) with the results of the statistical analysis ($n = 30$)

T_{room}	(°C)	24.0	±0.7	
T_{inc}	(°C)	35.8	±0.2	
$T(V)_{I,mean}$	(°C)	26.3	±0.5	
$T(P)_{I,mean}$	(°C)	32.3	±1.3	
$T(P)_{E,mean}$	(°C)	34.1	±0.8	
$T(V)_{E,mean}$	(°C)	26.2	±0.4	
$P_{H_2O}(V)_{I,mean}$	(kPa)	0.76	±0.0	
$P_{H_2O}(P)_{I,mean}$	(kPa)	4.85	±0.2	
$P_{H_2O}(P)_{E,mean}$	(kPa)	5.17	±0.3	
$P_{H_2O}(V)_{E,mean}$	(kPa)	1.14	±0.1	
$AH(V)_{I,mean}$	(mg/l)	5.5	±0.2	$p = 0.3$
$AH(P)_{I,mean}$	(mg/l)	34.4	±1.3	$p = 0.4$
$AH(P)_{E,mean}$	(mg/l)	36.5	±1.7	$p = 0.2$
$AH(V)_{E,mean}$	(mg/l)	8.2	±0.4	$p = 0.2$
WL_P	(mg/l)	2.1	±0.8	$p = 0.2$
WL_V	(mg/l)	2.8	±0.2	$p = 0.2$
$R_{I,mean}$	($cmH_2O/l/s$)	1.7	±0.1	$p = 0.2$
$R_{E,max}$	($cmH_2O/l/s$)	1.8	±0.1	$p = 0.2$
H_{eff}	(%)	93.44	±1.0	$p = 0.2$
$\Delta T(P)$	(°C)	1.7	±0.7	$p = 0.5$

Figure 4 shows the mean inspiratory and maximal expiratory resistances of the HME during the 24 h study period. There was no significant increase of resistance over 24 h (Table 1, $p = 0.2$ and $p = 0.2$ respectively).

Table 1 shows the flow-weighted mean temperature and humidity values at each side of the HME, the mean room temperature and the mean water losses at the each side of the HME over 24 h. As it should be, WL_P equals WL_V within the error range. The mean inspiratory fresh air temperature at the "V" site was always higher than the room temperature, as the ventilator heats the expiratory air to prevent condensation. The produced heat is conducted to the inspiratory limb by the metal parts of the ventilator and heats the inspiratory air to a temperature above the room temperature. Because of the heat and moisture exchange capacity of the "V" site sampling port and Y-piece of the tubing set, the inspiratory air humidity content at the "V" site was higher than the content of the central medical air.

Figure 5 shows the humidification efficiency (H_{EFF}) of the HME and the temperature difference at the "P" site ($\Delta T(P)$) during 24 h. There were no significant changes in humidification efficiency ($p = 0.2$) or $\Delta T(P)$ ($p = 0.5$) over 24 h.

The weight gain of the HME was 4 g during the study. The sensitivity and dynamic response of the mass spectrometer remained constant during the whole study. During the study the barometric pressure ranged between 100.7 and 101.9 kPa.

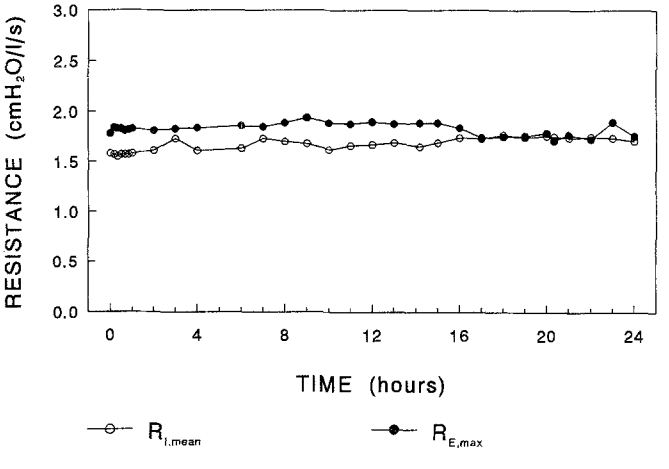

Fig. 4 Mean inspiratory and maximal expiratory resistance values of the HME over 24 h

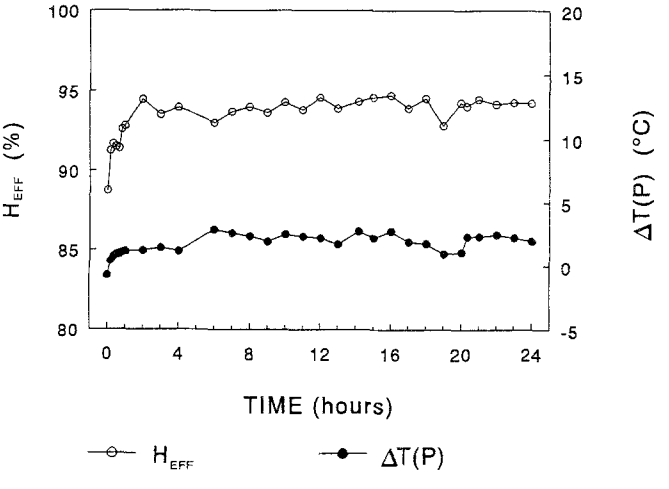

Fig. 5 Humidification efficiency of the HME and temperature difference at the "P" site over 24 h

Discussion

This study describes an experimental set-up and method to test HMEs, which is in close agreement with clinical conditions during mechanical ventilation. As can be seen from Fig. 2 and Table 1, the patient model characteristics of flow, expiratory temperature ($T(P)_{E,mean}$), and expiratory humidity ($AH(P)_{E,mean}$), were indeed comparable to values found in humans [23–27] and were in agreement with the ISO draft standard [18].

During pilot studies we observed a decrease in the expiratory saturation level with an incubator temperature of more than 38° C and the occurrence of condensation in the patient model with a temperature lower than 34° C which interfered with the measurements. The ISO draft standard allows the incubator temperature to be in a wide range [18]. A mean incubator temperature of 35.8±0.2° C over 24 h was sufficient to prevent condensation in this study. Vickers et al. stated that heating the patient model to prevent condensation, can result in incomplete saturation of expiratory air [19]. As shown above, complete saturation of the expiratory air leaving the patient model was reached with the set incubator temperature.

The ISO draft standard suggests compliance and resistance values for the patient model at various tidal volumes [18]. Our patient model showed a lower compliance and higher resistance compared to the recommendations of the ISO draft standard. The compliance value was in the same range as observed clinically by Eckerbom et al. [17]. Furthermore the efficiency of the HME is related to the flow rate, tidal volume, and transmission time of air through the HME [8, 9, 16, 20–22, 28]. However, the draft standard does not specify any inspiratory and expiratory flow rate. The expiratory flow profile of the patient model is more important for the function of the HMEs than the individual flow, compliance and resistance values. As can be seen in Fig. 2, the expiratory flow profile of the patient model closely resembles the expiratory profile during mechanical ventilation in normal patients.

In order to measure the dynamic variations in water exchange of HMEs during mechanical ventilation, we chose to use a mass spectrometer. Although the ISO draft standard specifies detailed technical requirements of the measurement equipment, the technique to measure humidity, an essential parameter for evaluation of HMEs, is only defined for the gravimetric method [18]. Gravimetric humidity measurements, however, are only reliable during long term studies and give an overall, average value without information of transient changes [29]. Thermocouple psychrometers and hygrometers have been used to measure humidity in various studies. However, thermocouple psychrometers have long time constants and require high flow rates for accurate measurements [29]. Hygrometers are particularly suitable for measurements at low saturation levels but generally have long response times [13, 29].

The use of a mass spectrometer avoids the above mentioned problems and allows measurement of dynamic variations of the water content of the inspiratory and expiratory air [26, 29, 30]. However, condensation and evaporation on the internal surfaces of the sampling capillary and the mass spectrometer can prolong the response time to the order of seconds. As others, we used a specially devised capillary (unheated, constricted tip capillary) to minimize these effects [31]. A heated sampling capillary was not used because this would possibly interfere with the local heat and water transport at the measurement site [31]. The dynamic response of the mass spectrometer was assessed at other ventilator settings (frequency 10 and 15 breaths/min, T_V of 1000 and 500 ml, $\dot{V}$: 30 and 60 l/min) and proved to be adequate to measure dynamic variations. To minimize the response time we also subjected the water vapour signals to an off-line deconvolution procedure based on a double-exponential approximation to the average of increasing step-responses [33]. Comparison of absolute humidity values calculated from compensated water vapour measurements were not significantly different from those calculated with uncompensated measurements. For a better signal to noise ratio we used, therefore, the uncompensated water vapour measurements for subsequent absolute humidity calculations. One practical problem with the use of a constricted tip capillary is the increased risk of obstruction. In general, when an obstruction is present, this can be deduced from the quality of the water vapour signals and an increase of step-response time. However, frequent recalibration of the mass spectrometer and assessment of the step-response time at regular intervals did not reveal any indication of obstruction of the sampling capillary during our measurements.

The main advantages of our system for measuring humidity is the application of an online, relative fast and accurate method for measurements of the water content of the inspiratory and expiratory air, which allows measurement of dynamic variations in water exchange of a HME.

As expected, the inspiratory and expiratory resistances of the HME were different, caused by the difference in inspiratory and expiratory flow rate and the amount of water captured in the HME. Expiratory resistance of the HME is related to the expiratory flow rate which is mainly dependent on the mechanical time constant of the patient model. In this study, we tried to imitate the expiratory flow profile of a mechanically ventilated patient in order to assess the influence of the HME on lung mechanics. The expiratory resistance of the HME we used did not influence the mechanics of our patient model as found by Conti et al. [32]. However, in diseases such as asthma, expiratory resistance has to be assessed for different HMEs.

A continuous and obvious increase in the inspiratory resistance as reported by Ploysongsang et al. [10] could not be observed in our study.

The ISO draft standard suggests only 2 resistance measurements during 24 h measurement period by using dry air and a measurement system separated from the patient model [18]. The use of a dry gas flow to measure resistance outside the system, as suggested by the draft standard, can cause an underestimation of the resistance values. Moreover, disconnection can cause possible water loss from the HME which also can lead to an underestimation of the resistance. Therefore, resistance measurements will be more accurate when measured in the patient model circuit without disconnection of the HME.

The thermic gradient across the HME determines its humidity output [7, 8, 12, 21]. This thermal gradient changes with the ambient temperature or the patient model output. The change in these variables makes it difficult to compare the different measurements performed at different times and conditions. The influence of the temperature and humidity difference of the inspiratory fresh gas and the patient model output on the results can be eliminated by calculating the humidification efficiency and the temperature difference at the "P" site as suggested in this and other papers [17]. Calculation of these parameters is easy and essential for comparison of different HMEs and protocols to test HMEs.

It is concluded that the experimental set-up described in this paper is in accordance with the ISO draft standard and closely imitates the use of HMEs in a clinical setting. This set-up and method is therefore a reliable means to evaluate HMEs. The principle advantages of this set-up over those described previously include: (i) continuous measurements of dynamic variations of water and temperature exchange; (ii) calculation of expiratory as well as inspiratory resistance of the HME; (iii) calculation of humidification efficiency and temperature difference at the patient site, which can be used for comparison of the performance of different HMEs.

Future studies will be directed to compare different HMEs during different clinical an experimental conditions.

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Appendix

$$AH = \frac{P_{H_2O} \times M_{H_2O}}{R \times T} \times 1000 \quad (mg/l) \tag{1}$$

$$C_{PM} = \frac{V_{T_I}}{P(V)_{I,plat} - P(V)_{E,end}} \times 1000 \quad (ml/cmH_2O) \tag{2}$$

$$R_{TOT} = \frac{P(V)_{I,max} - P(V)_{I,plat}}{\dot{V}_I} \quad (cmH_2O/l/s) \tag{3}$$

$$R(HME)_{I,mean} = \frac{\Delta P_{I,mean}}{\dot{V}_{I,mean}} \quad (cmH_2O/l/s) \tag{4}$$

$$R(HME)_{E,max} = \frac{\Delta P_{E,max}}{\dot{V}_{E,max}} \quad (cmH_2O/l/s) \tag{5}$$

$$R_{PM} = R_{TOT} - R(HME)_{I,mean} \quad (cmH_2O/l/s) \tag{6}$$

$$WL_P = AH(P)_{E,mean} - AH(P)_{I,mean} \quad (mg/l) \tag{7}$$

$$WL_V = AH(V)_{E,mean} - AH(V)_{I,mean} \quad (mg/l) \tag{8}$$

$$H_{EFF} = \left(1 - \frac{\text{mean water loss}}{\text{water output patient model}}\right) = \left(1 - \frac{(WL_P + WL_V)/2}{AH(P)_{E,mean}}\right) \times 100 \quad (\%) \tag{9}$$

$$\Delta T(P) = T(P)_{E,mean} - T(P)_{I,mean} \quad (°C) \tag{10}$$