Evaluation of Resistance in 8 Different Heat-and-Moisture Exchangers: Effects of Saturation and Flow Rate/Profile

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INTRODUCTION: When endotracheal intubation is required during ventilatory support, the physiologic mechanisms of heating and humidifying the inspired air related to the upper airways are bypassed. The task of conditioning the air can be partially accomplished by heat-and-moisture exchangers (HMEs). OBJECTIVES: To evaluate and compare with respect to imposed resistance, different types/models of HME: (1) dry versus saturated, (2) changing inspiratory flow rates. MATERIALS AND METHODS: Eight different HMEs were studied using a lung model system. The study was conducted initially by simulating spontaneous breathing, followed by connecting the system directly to a mechanical ventilator to provide pressure-support ventilation. RESULTS: None of the encountered values of resistance (0.5–3.6 cm H_2O/L/s) exceeded the limits stipulated by the previously described international standard for HMEs (International Standards Organization Draft International Standard 9360-2) (not to exceed 5.0 cm H_2O with a flow of 1.0 L/s, even when saturated). The hygroscopic HME had less resistance than other types, independent of the precondition status (dry or saturated) or the respiratory mode. The hygroscopic HME also had a lesser increase in resistance when saturated. The resistance of the HME was little affected by increases in flow, but saturation did increase resistance in the hydrophobic and hygroscopic/hydrophobic HME to levels that could be important at some clinical conditions. CONCLUSIONS: Resistance was little affected by saturation in hygroscopic models, when compared to the hydrophobic or hygroscopic/hydrophobic HME. Changes in inspiratory flow did not cause relevant alterations in resistance. *Key words: heat-and-moisture exchangers, respiratory mechanics, resistance, mechanical ventilation*. [Respir Care 2005;50(5):636–643. © 2005 Daedalus Enterprises]

Introduction

During spontaneous breathing, inspired air is warmed and humidified during passage through the nasal and oral cavities. The upper airways are responsible for the delivery of gas at approximately 32°C and a relative humidity of more than 90% to the lower respiratory tract at the tracheal carina.[1] Upon reaching the alveolar level, inspired air is warmed to body temperature (about 37°C) and has achieved 100% saturation with water vapor.[2,3] The point at which gases reach body temperature and full saturation is known as the isothermic saturation boundary. The isothermic saturation boundary can and does move up and down the respiratory tract as ambient conditions change or when there is a change in the patient's disease state.[4] When positive-pressure ventilation is delivered through oro/naso endotracheal intubation or via tracheostomy tube, the physiologic mechanisms of heating and humidifying the air related to the upper airways are bypassed.[3,5] Dry gas delivery has been associated with damage to the tracheobronchial mucosa and undesirable clinical manifestations.[5–8] So it is vital to precondition the inspired air in order to provide adequate heating and humidification.[9–13]

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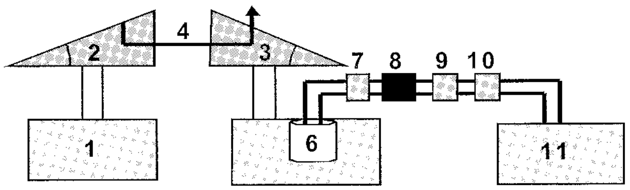

Fig. 1. Mechanical model. Components include: Bear 1000 mechanical ventilator (1) connected to a respiratory system analog constructed using a training test lung 1600 and a Bear lung model. The training test lung had 2 compartments, with the first compartment (2) connected to a Bear 1000 mechanical ventilator (used to simulate 3 levels of inspiratory effort), and the second compartment (3) displaced by the first compartment with the aid of a lift bar (4). The second compartment was connected to the Bear lung model that was composed of a bellows (6) within a rigid box. The bellows represented the lung and the space between the rigid box and the bellows represented the pleural cavity (5). A section of tubings/connectors represented the airways, within which different heat-and-moisture exchangers (HMEs) (8) were tested. Flow (10) and pressure (7 and 9) sensors were connected to the model. The study was conducted initially simulating spontaneous ventilation, then the analog was connected to a Puritan Bennett 840 mechanical ventilator (11) to simulate pressure-support ventilation.

The task of preconditioning the air can be accomplished by heated humidifiers (active-action humidifiers) or by heat-and-moisture exchangers (HMEs) (passive-action humidifiers).[9,14,15] The use of heated humidifiers has some advantages, like temperature control independent of the patient's temperature, and disadvantages, such as cost,[11,14,16–19] circuit water condensate,[14,20] requirement of an energy source,[10,16] and continuous water supply.[16,21,22] However, HMEs can also cause clinical problems that may preclude their use. Complications associated with HME use include increased resistance,[15,23–26] increased work of breathing,[15,22,25,27–30] and hypercapnia due to increased dead space.[15,31,32]

Gas-flow resistance through the HME increases after several hours of use,[23–25,33] as the HME material density increases[18,34] and as flow is increased.[23,35] Resistance can be markedly increased when the HME is occluded with secretions, blood, or water.[23,28,34,36–38]

Investigations of HME performance have reported differences between models, as evaluated by several variables. Results vary, even within the same type of HME.[39,40] The present study had the following objectives: to evaluate and compare, in a lung model system, resistance measured across currently available types and models of HME, with particular attention to the effects of saturating the unit and by variations in inspiratory flow.

Methods

A respiratory system analog (Fig. 1) was constructed, using a training test lung (model 1600, Michigan Instruments, Grand Rapids, Michigan) and a Bear lung model (Bear Medical Systems, Palm Springs, California). The training test lung has 2 compartments. We connected the first compartment to a Bear 1000 mechanical ventilator (Bear Medical Systems, Palm Springs, California), and the second compartment was displaced by the first compartment with the aid of a lift bar. The second compartment was connected to the Bear lung model, which is a bellows within a rigid box. In our analog, the bellows represents the lung, and the space between the rigid box and the bellows represents the pleural cavity.

There are no clear standards in mechanical models about the effort simulation that should be used. In a pilot study we tried to design efforts in order to obtain 3 different levels of airway occlusion pressure 0.1 s after the onset of inspiratory effort ($P_{0.1}$), to simulate normal and high demand situations. The arbitrary efforts that we used were based on the findings of this pilot study. The Bear ventilator was used to simulate 3 levels of inspiratory effort, by adjusting pressure levels and slopes in pressure control mode: effort 1 (E1): change in pressure (ΔP) = 18 cm H_2O and slope = −3; effort 2 (E2): ΔP = 22 cm H_2O and slope = 0; effort 3 (E3): ΔP = 26 cm H_2O and slope = +4. The effort levels were set to deliver normal $P_{0.1}$ (3.3 cm H_2O), an effort above normal $P_{0.1}$ (6.1 cm H_2O), and a high $P_{0.1}$ (8.0 cm H_2O), respectively. To achieve these targets, the compliance of the first compartment of the training test lung was set at 54 mL/cm H_2O. The compliance of the second compartment of the training test lung, representing the thoracic cage, was set at 200 mL/cm H_2O, resulting in a respiratory system compliance of 60 mL/cm H_2O. Constant settings for each effort level were respiratory frequency 12 breaths/min, inspiratory time 1 s, and positive end-expiratory pressure 0 cm H_2O.

The testing region was a section of tubings/connectors that represented the airways. We added a resistive element (4.0 cm H_2O/L/s) to simulate resistance similar to that of a normal patient.[41] Flow and pressure sensors were connected to the model immediately before and after the HME; the data were transmitted to a personal computer with an analogic-digital data-acquisition interface (PCI-MIO-16XE-50, National Instruments, Austin, Texas). In this same simulated airway segment, 8 different HMEs were sequentially tested. Our choice of HME was based on the most commonly used units in hospitals of our city (Table 1) (Fig. 2). The study was initially conducted while simulating spontaneous ventilation, then the analog was connected to a mechanical ventilator (model 840, Nellcor Puritan Bennett, Carlsbad, California) to simulate pressure-support ventilation (PSV) set at 10 cm H_2O.

Software developed in LabView (National Instruments, Austin, Texas) was used for the analysis. Using customized subprograms, we analyzed 10 cycles for each simulated condition in order to generate a "mean" cycle and to

Table 1. Heat-and-Moisture Exchangers

Hygroscopic	
	Humid-Vent 2S (G2S), Gibeck, Upplands-Väsby, Sweden
	Thermovent 1200 (Portex), Sims Portex, Hythe, Kent, United Kingdom
	Hygroscopic Condenser Humidifier (HCH), HCH Newmed, Bromma, Sweden
Hydrophobic	
	BB100MFS (Pall), Newquay, Cornwall, United Kingdom
Hygroscopic-Hydrophobic	
	Humid-Vent Filter Light (G light) and Humid-Vent Filter Compact (G compact), Gibeck, Upplands-Väsby, Sweden
	Hygrobac S DAR (Hygrobac S) and Hygroster DAR (Hygroster), Mallinckrodt Medical SpA, Mironda, Italy

Fig. 2. Eight heat-and-moisture exchanger (HME) models under investigation (details in Appendix). Top row (left to right): HCH, Portex, G compact, Hydrobac S. Bottom row (left to right): G2S, Pall, G light, Hygroster.

calculate the imposed resistance due to the HME. Resistance was measured using the isovolume method, based on an analysis of transpulmonary pressure, flow, and volume information; points were chosen in both the inspiratory and expiratory phases of a respiratory cycle when lung volumes were identical and flow rates were about maximal. In the tidal range of volume-change, the elastic component of transpulmonary pressure is the same at these instants. Accordingly, the observed change in the pressure between these points relates solely to flow resistance. The ratio of this pressure change to the corresponding change in flow between these points has been calculated. This value represents an "average" flow resistance for inspiration and expiration.[42]

After basal measurements were obtained from each HME still dry, the HMEs were connected to a nebulizer (micro mist nebulizer #1884, Hudson RCI, Temecula, California) that aerosolized 6.0 mL of 0.9% saline into the HME at a flow of 5.0 L/min O_2. The HMEs were then tested in this saturated state (our definition of saturated for this study). The saline aerosolization amount was empirically determined to obtain a wet weight for the HME similar to the weight attained after 24 hours of use in patients. This weight information was derived from samples of each of the studied HMEs used in different patients after 24 hours of use. These HMEs were then sealed and weighed in a high-precision analytic balance.

Results

Initially we performed repeated measurements under different respiratory mechanics situations with different HMEs, and our model showed excellent reproducibility, with a standard deviation for resistance at 0.019. The measured resistances would certainly be statistically different ($p < 0.05$) when they differed from each other in more than 2 times the standard error of the mean, based on a 95% confidence interval.[43] Many comparisons were found to be statistically different, but their clinical importance will be considered in the discussion.

The resistance measures during spontaneous ventilation and PSV are shown in Tables 2 and 3. There are clear differences between control and HME, HME models, and between dry versus saturated conditions.

Although increases in efforts resulted in, at times, statistically significant increases ($p < 0.05$) in resistance, the magnitude of this increase reached a maximum value of 0.28 cm H_2O/L/s. Therefore, we chose to show only the medium effort (E2) in the next set of figures.

Figure 3 displays resistance during spontaneous ventilation in dry and saturated conditions. Figure 4 displays these values during PSV. We observed from these figures that the Portex, G2S, and HCH had lesser resistance levels, independent of their condition (dry or saturated), and there were smaller resistance differences between dry and saturated states. The Pall, Hygroster, Hygrobac S, G compact, and G light models had greater resistance when dry, and resistance increased after saturation. Among these HMEs, G light (both in spontaneous ventilation and PSV) and G compact (PSV) did not present a significant increase in resistance after saturation.

Figure 5 compares resistance of the dry HME when using spontaneous ventilation or PSV. Figure 6 makes the same comparison for the saturated state. We observed that the resistance of the HME had almost no variation with an increase in flow (during PSV), from a clinical point of view.

Discussion

HMEs have had growing acceptance in recent years because of their low cost, simplicity of use, reduction in

Table 2. Resistance Without HME (Control) and With 8 Different HMEs During Spontaneous Ventilation

	Inspiratory Effort	Resistance (cm H_2O/L/s)								
		Control	Pall	Hygroster	Hygrobac S	Portex	G compact	G light	G2S	HCH
Dry	E1	0.283	2.157	2.314	2.005	0.964	2.325	1.710	0.793	1.102
	E2	0.316	2.251	2.368	2.074	1.067	2.392	1.813	0.857	1.216
	E3	0.345	2.336	2.417	2.141	1.161	2.453	1.908	0.915	1.320
Saturated	E1	NA	2.842	3.311	3.788	0.994	3.088	1.93	1.154	1.424
	E2	NA	2.936	3.357	3.842	1.102	3.082	2.021	1.229	1.582
	E3	NA	3.009	3.388	3.896	1.190	3.024	2.085	1.286	1.699

HME = heat-and-moisture exchanger
E1 = low effort
E2 = medium effort
E3 = high effort
NA = not applicable

Table 3. Resistance Without HME (Control) and With 8 Different HMEs During Pressure-Support Ventilation

	Inspiratory Effort	Resistance (cm H_2O/L/s)								
		Control	Pall	Hygroster	Hygrobac S	Portex	G Compact	G Light	G2S	HCH
Dry	E1	0.365	2.392	2.475	2.196	1.216	2.499	2.004	0.996	1.393
	E2	0.387	2.454	2.509	2.246	1.315	2.552	2.101	1.050	1.490
	E3	0.405	2.498	2.537	2.293	1.387	2.598	2.173	1.092	1.573
Saturated	E1	NA	2.929	3.323	3.765	1.262	2.851	2.075	1.287	1.723
	E2	NA	2.947	3.339	3.701	1.355	2.765	2.165	1.342	1.819
	E3	NA	2.942	3.347	3.655	1.425	2.791	2.224	1.368	1.880

HME = heat-and-moisture exchanger
E1 = low effort
E2 = medium effort
E3 = high effort
NA = not applicable

circuit condensate, avoidance of an energy source, and in some HMEs, microbiological filtration. Several studies have been published that tested HME characteristics, employing the models of HME available at that time, under a range of conditions/techniques with the use of respiratory-system mechanical simulators.[22–24,28,35,44]

Adding an HME to the patient's respiratory system does increase airway resistance[15,23–26] and work of breathing.[15,22,25,27–30] This increase in resistance can be ignored in patients with normal lungs having low airway resistance that is clinically unimportant; therefore, an increase in resistance caused by an HME may not be a factor for patients with normal pulmonary parenchyma. But, theoretically, a resistance of 1.78 cm H_2O/L/s (mean resistance found in our study) or of 3.60 cm H_2O/L/s (maximum value of resistance found), could in some clinical settings, such as patients with chronic obstructive pulmonary disease, together with the intrinsically high resistance in the airways of these patients and the endotracheal-tube resistance, lead to increased WOB and to the development of dynamic hyperinflation.[41]

Previous studies stated that the international standard for the HME (International Standards Organization Draft International Standard 9360-2) has set a maximum increase of resistive pressure not to exceed 5.0 cm H_2O with a flow of 1.0 L/s, even when saturated.[24,33] The imposed resistance by an HME ranged, in our study, between 0.50 cm H_2O/L/s (G2S dry, spontaneous ventilation, low effort) and 3.60 cm H_2O/L/s (Hygrobac S saturated, spontaneous ventilation, high effort). None of our encountered values of resistance exceeded the limits stipulated by the international standard for the HME (International Standards Organization Draft International Standard 9360-2).

HMEs are disposable devices that are inserted between the endotracheal tube and the Y-piece of the mechanical ventilator.[12,45,46] HMEs recover heat and moisture during exhalation and return a portion of the heat and humidity

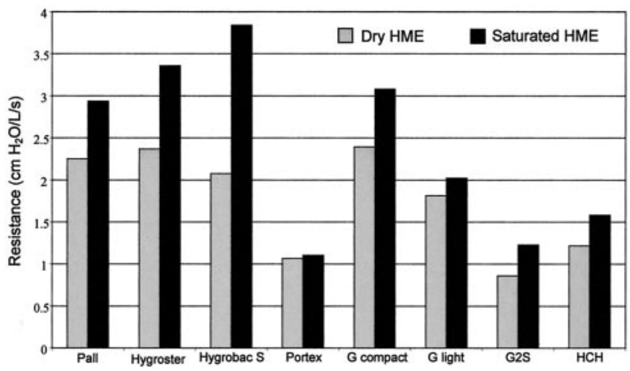

Fig. 3. Graph of resistance with dry and saturated heat-and-moisture exchangers (HMEs) during spontaneous ventilation for effort level E2 (change in pressure [ΔP] = 22 cm H_2O and slope = 0) at a flow of 0.70 L/s. The hygroscopic HMEs (Portex, G2S, and HCH) had lesser resistance levels, independent of their state (dry or saturated), and there were smaller resistance differences between dry and saturated states. G light did not present a significant increase in resistance after saturation. The Hygrobac S had a higher resistance level than other HMEs when saturated and a greater resistance difference between dry and saturated states.

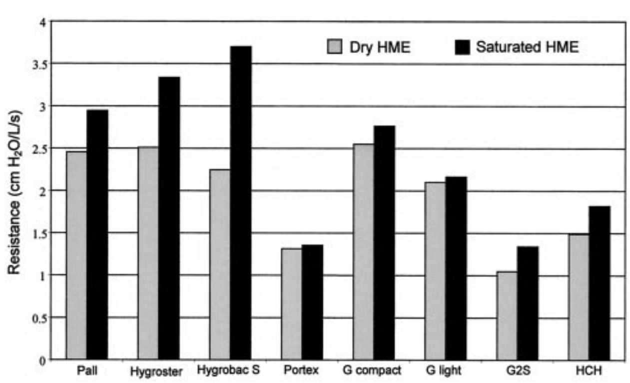

Fig. 4. Graph of resistance with dry and saturated heat-and-moisture exchangers (HMEs) during pressure-support ventilation for effort level E2 (change in pressure [ΔP] = 22 cm H_2O and slope = 0) at a flow of 1.13 L/s. The hygroscopic HMEs (Portex, G2S, and HCH) had lesser resistance levels, independent of the condition (dry or saturated), and there were smaller resistance differences between dry and saturated states. G light and G compact did not present a significant increase in resistance after saturation. The Hygrobac S had a higher resistance level when saturated and a greater resistance difference between dry and saturated states.

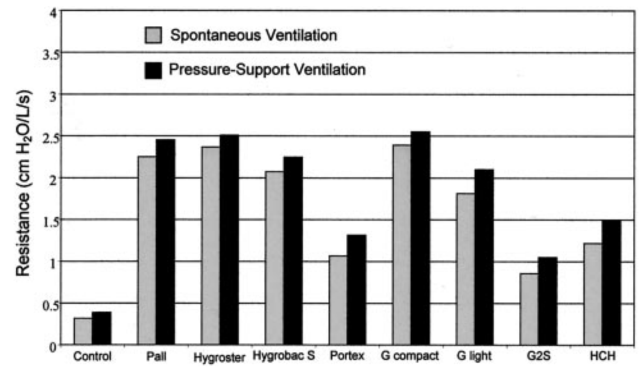

Fig. 5. Graph that compares resistance during spontaneous ventilation and pressure-support ventilation without a heat-and-moisture exchanger (HME) (control) and with a dry HME, for effort level E2 (change in pressure [ΔP] = 22 cm H_2O and slope = 0) at a flow of 0.70 L/s during spontaneous ventilation and at a flow of 1.13 L/s during pressure-support ventilation. The resistance of the HME was not affected by an increase in flow (during pressure-support ventilation), from a clinical point of view.

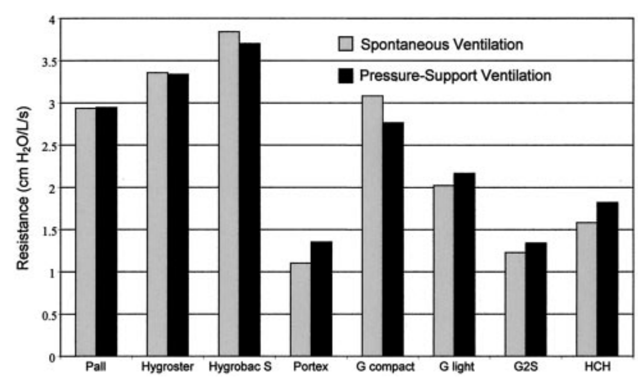

Fig. 6. Graph of resistance during spontaneous ventilation and pressure-support ventilation in a saturated heat-and-moisture exchanger (HME) for effort level E2 (change in pressure [ΔP] = 22 cm H_2O and slope = 0) at a flow of 0.70 L/s during spontaneous ventilation and at a flow of 1.13 L/s during pressure-support ventilation. The resistance of the HME was not affected by an increase in flow (during pressure-support ventilation), from a clinical point of view.

during the next inspiration.[16,18,47–50] Their recovering ability is dependent on materials within the HME, and there are 3 basic types of HME: hygroscopic, hydrophobic, and combined (hygroscopic-hydrophobic).[12] The hygroscopic HMEs (HHMEs: Portex, G2S, and HCH) contain materials of low thermal conductivity, impregnated with a hygroscopic chemical. The hydrophobic HME (HMEF: Pall) has a larger surface area, because of pleating of the material.[18] The HMEF has a substance covering the filter that prevents the water's exodus during exhalation, and the HMEF serves as an efficient microbiologic filter as well.[51] The combined HMEs (hygroscopic-hydrophobic) (HHMEF: Hygroster, Hygrobac S, G compact, and G light) have humidification properties and the bacterial-retention properties of the filter membrane.[12,48,52] In the HHMEF, the hygroscopically-treated material is located between the patient's airway and the filter.

HMEs with hydrophobic membranes have different performance characteristics, which can be explained by the properties and positioning of the different materials within the HME, which constitute an intrinsic property that even varies among HMEs of the same group. This finding is of particular importance because it shows that it is not possible to make any general rule of behavior based on the 3 classical groups of HMEs, in agreement with Thiéry et al.[40]

When we compared dry with saturated HMEs for each effort level and ventilation mode, we observed that the types/models of HME influenced the resistance measurements. The main factor is the hydrophobic membrane, because the HHME (without hydrophobic membrane) showed a distinct behavior compared to the others (HMEF, HHMEF), presenting less resistance in all situations and less effect on resistance when saturated.

The hydrophobic components seem to have an important role in causing higher resistance values. The HMEs with a hydrophobic membrane (HMEF and HHMEF) show an increase in resistance after saturation, although this phenomenon differed among models. Except for the G light, this increase in resistance ranged between 0.69 cm H_2O/L/s and 1.80 cm H_2O/L/s for the low effort during simulated spontaneous ventilation. In high-demand situations, this additional increase in resistance could cause iatrogenic increases in airway pressure. G light, although belonging to the HHMEFs, was the model that had the least resistance in this group when dry, and also exhibited a smaller increase when saturated in each ventilatory mode. When PSV was used, G compact also had little change in resistance when saturated.

As seen in the tables/figures, only by saturating an HME do we produce a significant increase in resistance. If secretions become entrapped inside the HME, this could lead to a significant further increase in resistance. Therefore, when an HME is in use and secretions become abundant, a heated humidifier should be used to replace the HME. Other studies have reported increases of resistance with the use of HME when they were filled with secretions,[34] blood,[36] or water.[23]

One of the limitations of our study was the use of saline to saturate the HME. Although we tried to simulate the weight-gain observed in patients, the saturation with saline may be different than the humidity exchange that happens in patients during the respiratory cycle. Thus, the magnitude of our results may not reflect exactly the clinical situation, although this consideration does not change the interpretation of our findings.

HMEF and HHMEF models displayed significant increases in resistance after saturation, but they had little increase in resistance due to increased inspiratory flow. Overall, we observed that HMEs had little variation in resistance in response to an increase in flow (when pressure support and/or higher efforts were used). Some studies[23,35] have reported significant increases in resistance when flow was increased. In our study, although there was an increase in resistance as a function of flow, the magnitude of this increase was smaller than that observed in other studies and certainly of little clinical importance. The fact that different HME models were used in our study can partially explain these findings.

In our study, the only objective was to analyze respiratory mechanics, and, based on our findings, HHMEs are slightly better suited for patients with disorders of respiratory mechanics, because they show low resistance when dry and little variation in resistance with saturation. It is important to remember, though, that other factors can influence the choice of an HME, such as humidification capacity and microbiological filter ability.

We must not forget that factors that increase resistance of the ventilator-patient circuit, like the endotracheal tube and/or the presence of humidification devices, are often neglected when respiratory system mechanics are determined. This lack of consideration can result in errors in therapy, such as a delay in weaning from mechanical ventilation.

Conclusions

1. None of the encountered values of resistance (dry or saturated HME) exceeded the limits stipulated by the international standard for the HME (International Standards Organization Draft International Standard 9360–2).

2. Among the HMEs we studied, the HHME had less effect on resistance, when compared to the HMEF and HHMEF models.

3. Changes in inspiratory flow did not cause relevant alterations in resistance. Saturation did, however, increase resistance, mainly in the HMEF and HHMEF models.

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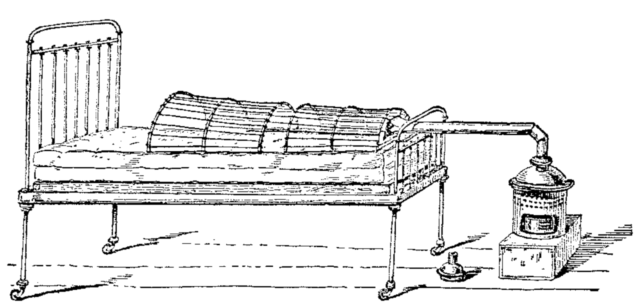

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