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Danger of low pressure alarm failure in preterm infants on continuous positive airway pressure

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Abstract In certain settings of conventional continuous positive airway pressure (CPAP) application, the ventilator may not be able to detect dislodgement of the prongs. This occurs especially in settings with high flow and small prongs. We investigated the relation between ventilator flows, size of the nasal prongs, and pressure generated within the ventilator circuit due to the flow resistance of the prongs. We studied a Baby-flow® CPAP connected to a Babylog 8000plus® ventilator. Five prongs of increasing size (x-small, small, medium, large, x-large) and one nose mask were connected to the CPAP in turn. Starting at 30 lpm, the flow was reduced in 2 lpm steps. The dynamic pressure caused by the flow resistance of the prongs within the ventilator circuit was recorded. For all devices, we observed a correlation between the reduction of the flow and the reduction in pressure within the ventilator circuit. However, the flow resistance of the x-small prongs generated the highest dynamic pressure (30 mbar at 22 lpm) within the ventilator circuit while the mask gave rise to the lowest pressure (9 mbar at 30 lpm). The pressure value generated with x-small prongs at low flow rate was observed at high flow rate with x-large prongs or with a mask. We conclude that in settings with high flow rates, low CPAP levels, and small prongs, the resistance of the prongs will create enough dynamic pressure within the ventilator circuit to permit the ventilator to compensate a large leakage flow by closing the expiratory valve. Thus, in case of dislodgement of the prongs, the pressure within the ventilator circuit will not decrease below the alarm level, and the machine will not be able to generate an alarm.

Keywords CPAP · Monitoring · Low pressure alarm failure · Preterm infants

Introduction

Since the implementation of nasal continuous positive airway pressure (CPAP) for the treatment of respiratory distress syndrome in preterm infants, the rate of invasive ventilation has decreased considerably [1, 2]. Implementation of binasal CPAP devices was a second milestone in the history of neonatal ventilation [3, 4]. CPAP can be applied either by means of a conventional constant flow ventilator or by jet ventilation. In conventional constant-flow ventilators, the pressure necessary for using CPAP is created by closing an expiratory valve, while in jet-CPAP devices, the pressure is generated by injecting one or two high-speed flow jets [5].

In our ward, we observed that when CPAP was applied with a conventional constant-flow ventilator, the prongs were sometimes completely dislodged from the nose of the patient and still the machine triggered no low-pressure alarm. This

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happened especially when high gas flow was applied to very small premature babies using very small prongs.

So far, no studies exist about the optimum gas flow in CPAP applications with conventional constant-flow ventilators. However, it is generally supposed that "too much flow might be better than too little flow" [5]. In this study, we aimed at determining the optimum flow rate for any CPAP applied by conventional constant-flow ventilators to preterm infants by investigating the correlation between the ventilator flow, the size of the nasal prongs, and the dynamic pressure generated within the ventilator circuit by the flow resistance of the prongs.

Methods

In this experimental study, the binasal CPAP Baby flow® device (Draeger Medical AG & Co. KG, Lübeck, Germany) produced for application of CPAP with constant-flow ventilators was connected to a Babylog 8000plus ventilator (Draeger Medical AG & Co. KG, Lübeck, Germany). The ventilator was set on CPAP mode with the maximally achievable positive end-expiratory pressure (PEEP) of 25 mbar. The PEEP was set to such a high level in order to ensure that the expiratory valve remained closed all the time. Otherwise, the dynamic pressure within the ventilator circuit could have become higher than the adjusted PEEP, and in this case, the machine would have opened the expiratory valve and so it would have been impossible to measure the changes in the dynamic pressure. In order to investigate the full spectrum of possibly adjust-able flow rates, the ventilator flow was initially set to the maximum of 30 lpm. Nasal prongs of increasing size (x-small (XS): inner diameter 2 mm, length 6 mm; small (S): inner diameter 2.5 mm, length 8 mm; medium (M): inner diameter 3 mm, length 9 mm; large (L): inner diameter 3.5 mm, length 9 mm; x-large (XL): inner diameter 4 mm, length 9 mm) and one nasal mask were connected in turn to the CPAP device. The gas streaming out of the nasal prongs could enter the room air without any additional resistance. The different pressure levels generated within the ventilator circuit using the different prongs and the mask were documented in a pressure/flow graph. Then the flow was gradually reduced in steps of 2 lpm, and the measurements were repeated with all prongs and the mask in turn until the pressure within the ventilator circuit had decreased to a level of 2 mbar. All measurements were repeated three times.

Results

Figure 1 shows the flow/pressure graph for all nasal prongs and for the mask. For all devices, we observed a correlation between the reduction of the flow and the reduction in pressure within the ventilator circuit.

With XS-prongs, the pressure measured within the ventilator circuit was 25 mbar when the flow was set to more than 22 lpm. The pressure fell to below 2 mbar only when the flow was reduced to 4 lpm. The decrease in pressure when reducing the flow from 22 lpm down to

Fig. 1 Flow-pressure graph of all nasal devices examined. (*XS-Prongs* prong size X-small, *S-Prongs* prong size small, *M-Prongs* prong size medium, *L-Prongs* prong size large, *XL-Prongs* prong size X-large, *Nose Mask* nasal mask). The unit of measurement is the millibar (*mbar*), which is also the unit that appears on the display of the Babylog 8000plus ventilator. One millibar corresponds to approximately 1 cm H2O
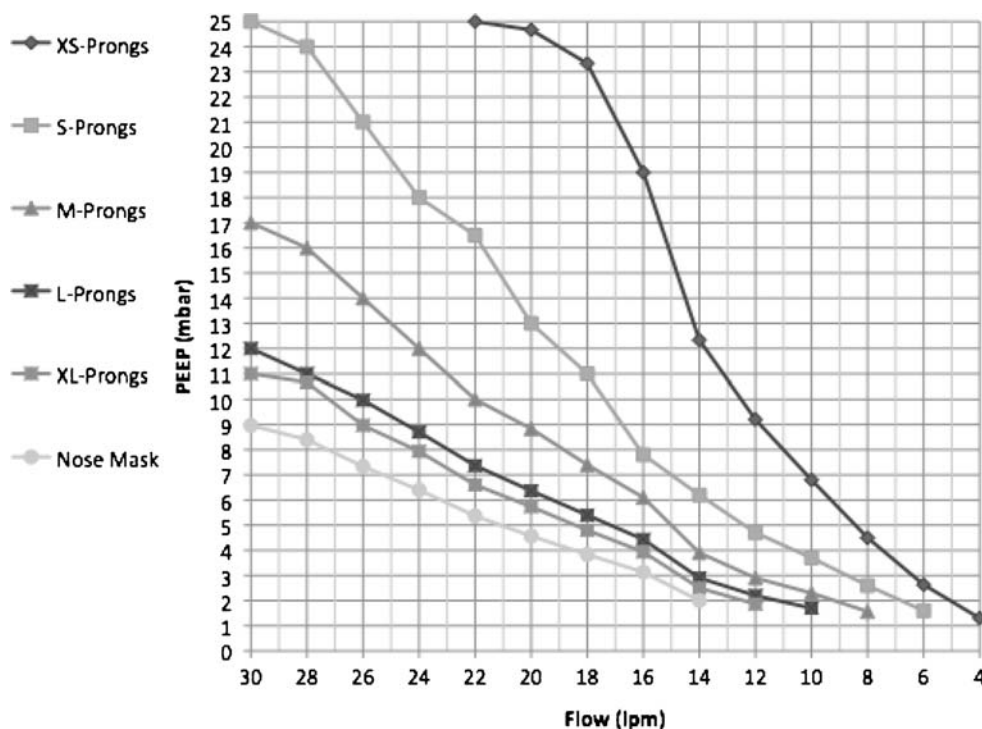
4 lpm was not linear; the steepest decrease occurred between 16 and 14 lpm. For example, a pressure of 6 mbar within the ventilator circuit was generated at a flow rate of 9.5 lpm.

With S-prongs, the pressure measured within the ventilator circuit was 25 mbar when the flow was set to 30 lpm. The pressure fell to below 2 mbar only when the flow was reduced to 6 lpm. The decrease in pressure when reducing the flow from 30 lpm down to 16 lpm was nearly linear. At lower flow rates the decrease was slower. For example, a pressure of 6 mbar within the ventilator circuit was generated at a flow rate of 14 lpm.

With M-prongs, the pressure measured within the ventilator circuit was 17 mbar when the flow was set to 30 lpm. The pressure fell to below 2 mbar only when the flow was reduced to 8 lpm. The decrease in pressure when reducing the flow was nearly linear. For example, a pressure of 6 mbar within the ventilator circuit was generated at a flow rate of 16 lpm.

With L-prongs, the pressure measured within the ventilator circuit was 12 mbar when the flow was set to 30 lpm. The pressure fell to below 2 mbar only when the flow was reduced to 10 lpm. The decrease in pressure when reducing the flow was nearly linear. For example, a pressure of 6 mbar within the ventilator circuit was generated at a flow rate of 19 lpm.

With XL-prongs, the pressure measured within the ventilator circuit was 11 mbar when the flow was set to 30 lpm. The pressure fell to below 2 mbar only when the flow was reduced to 12 lpm. The decrease in pressure when reducing the flow was nearly linear. For example, a pressure of 6 mbar within the ventilator circuit was generated at a flow rate of 21 lpm.

With the nose mask, the pressure measured within the ventilator circuit was 9 mbar when the flow was set to 30 lpm. The pressure fell to below 2 mbar only when the flow was reduced to 14 lpm. The decrease in pressure when reducing the flow was nearly linear. For example, a pressure of 6 mbar within the ventilator circuit was generated at a flow rate of 23 lpm.

In Fig. 2, the x- and the y-axis of the pressure/flow graph for all nasal prongs and for the mask are switched. In this graph, the areas under the curves define the flow range where in case of high leakage flow, the pressure measured within the ventilator circuit is similar to the pressure that is applied inside the patient's lung.

Discussion

The results of our study show that the flow resistance of the nasal prongs and of the mask creates a dynamic

Fig. 2 Pressure/flow graph for all nasal prongs and for the nose mask of the Baby flow® CPAP ventilation device with x- and y-axis switched. The area under the curves indicates the flow range where in case of high leakage flow, the pressure measured within the ventilator circuit is equal to the pressure created within the airways of the patient. Adjustments of flow within this area are "safe." The *area above* the *curves* indicates the flow range where in cases of high leakage flow the pressure measured within the ventilator circuit is higher than the pressure created within the airways of the patient. At such high flow ranges, failure of the ventilator alarm may occur. Adjustments of flow within this area are therefore "not safe." (*XS-Prongs* prong size X-small, *S-Prongs* prong size small, *M-Prongs* prong size medium, *L-Prongs* prong size large, *XL-Prongs* prong size X-large, *Nose Mask* nasal mask)
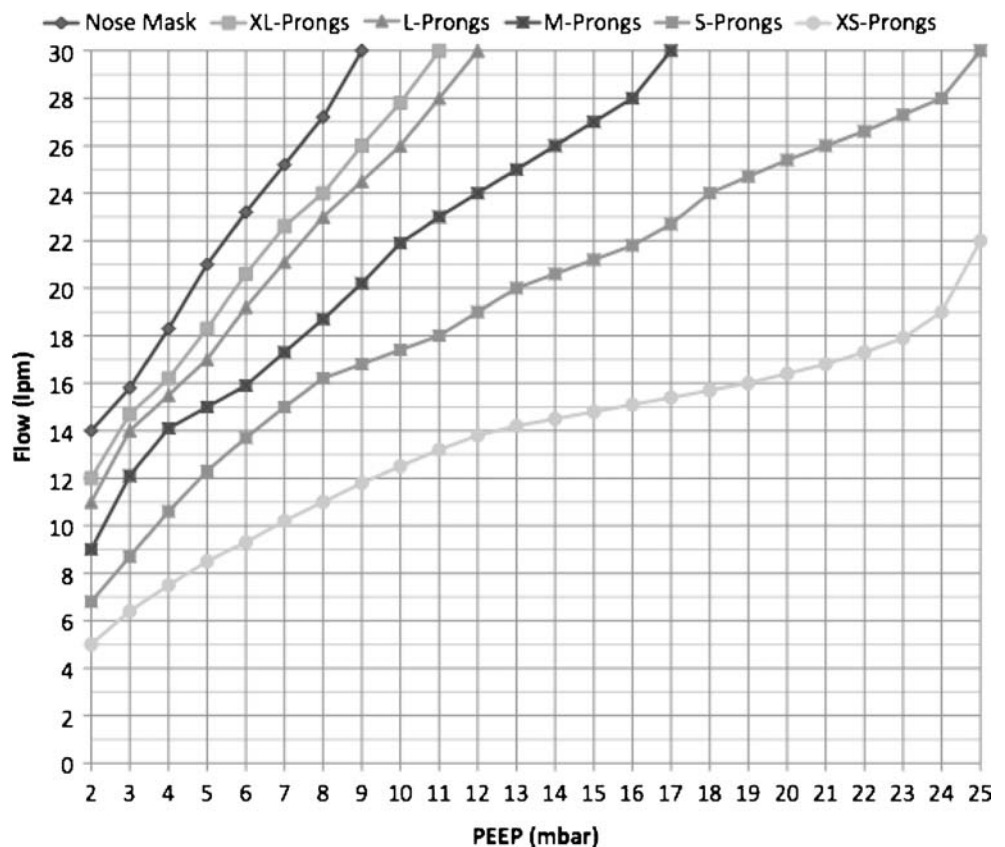
pressure inside the ventilator circuit. This dynamic pressure depends on the flow rate and on the size of the prongs. It increases for higher flows and smaller inner diameters of the prongs [6].

In everyday use, normally this does not matter as long as the prongs remain in the correct position. Without leakage flow, all the flow created by the machine must pass the expiratory valve at the end of the ventilator circuit. This flow varies only little due to the patient's inspiration and expiration. During inspiration, the flow that reaches the expiratory valve is reduced by the tidal volume of the breath while during expiration, the breath volume is added to the constant flow of the machine. The pressure created within the patient's airways is only regulated by the opening and closing of the expiratory valve and is thus independent of the ventilator flow.

In case the prongs are partially dislodged and part of the flow is lost, the machine attempts to compensate for this leakage flow by closing the expiratory valve a little more in order to create a constant CPAP. However, when the leakage flow becomes higher—this happens when the prongs are completely dislodged—the pressure within the ventilator circuit will decrease even when the expiratory valve is completely closed. The machine will detect the problem and alert the staff by activating an alarm [7] so that the nasal prongs can be repositioned only with a short interruption of the respiratory support. This occurs as long as the applied flow is within the flow range at the areas under the curves in Fig. 2 where at all times, the pressure within the ventilator circuit is similar to the pressure applied to the patient's lungs.

However, not in every situation of dislodged nasal prongs is the machine able to generate an alarm signal. One reason for failure of the low pressure alarm is an occlusion of the dislodged prongs, for example by the cheek of the patient. Manufacturers often mention this during training courses as a possible complication. In this case, there will not be any leakage flow so that the pressure does not decrease and the machine cannot identify the problem. A second reason for an alarm failure caused by dislodgment of the prongs can be demonstrated in our experimental setup. It concerns in particular very small preterm infants on CPAP who require XS- or S-prongs. Due to the small inner diameter of these prongs and the resultant maximum resistance, any possible leakage flow will be very low, and especially when applying high flows, a high dynamic pressure will be generated within the ventilator circuit. In such a situation, the leakage flow through these small prongs can be lower than the ventilation flow from the machine even when the prongs are completely dislodged. In this case, the ventilator will compensate for this small leakage flow by closing the expiratory valve as much as necessary. Thus,

the pressure inside the ventilator circuit will not decrease below the alarm level, and the machine will not be able to detect the problem and to generate an alarm. This situation may occur if the applied flow is within the flow range at the areas above the curves in Fig. 2.

In contrast to the leakage flow described above which is caused by dislodgement of the prongs, a "normal" leakage flow caused by an open mouth of the patient during binasal CPAP application [8–10] can be compensated by closing the expiratory valve and/or by increasing the ventilation flow [5].

Our results contradict the common opinion that in conventional CPAP ventilation "too much flow might be better than too little flow" [5]. We conclude that the machine ventilation flow should be chosen low enough to detect critical alarm situations but at the same time high enough to compensate the normal leakage flow due to an open mouth. The curves generated by our measurements allow finding the maximum flow that can be used with different prongs without risking the alarm system failure described above.

In our study, the lowest flow resistance was created by the nasal mask. Thus, nasal masks seem to be most suitable for binasal CPAP ventilation for extremely small preterm infants. With nasal masks, the highest ventilation flows can be chosen without danger of low-pressure alarm failure. The only disadvantage of nasal masks is the theoretical danger of causing midface bone deformities. However, so far this complication has never been described in the literature.

The flow resistance generated by the inner diameter of the nasal prongs is a well-known physical problem [6]. Therefore, the main findings of this study are applicable to all conventional CPAP ventilation devices used in neonatal intensive care medicine. However, this does not mean that all existing nasal prongs with the same labeling create the same dynamic pressure at the same flow rate because prongs from different manufacturers vary both in their inner diameter and in their length. Therefore, all prongs from different manufacturers have their own pressure/flow graphs as well as their own maximum flow rates that should not be exceeded in order to avoid failure of the low-pressure alarm. In none of the commercially available devices has this problem of potential alarm failure been resolved in a satisfactory way.

Our results show that ventilators with automatic adjustment of the machine ventilation flow hold important potential risk factors. These ventilators compensate for leakage flow by closing the expiratory valve and/or increasing the ventilation flow. Theoretically, these machines are able to create a constant pressure with prongs of all sizes. However, as we have shown, these machines need very complex alarm systems.

Jet-CPAP devices work according to a completely different physical principle. They do not need an expiratory tube connected with an expiratory valve [11]. Therefore, jet-CPAP devices have no problem to detect the pressure loss induced by dislodgement of the prongs. Each of them is able to create an appropriate CPAP, and no differences in clinical efficiency have been described in the literature [12, 13]. However, jet-CPAP devices have other disadvantages, one of the biggest problems being the extreme noise created by these machines. Noise levels of more than 100 dB were measured in the pharynx of infants on jet-CPAP devices [14, 15].

Conclusion

Based on our results, we conclude that the pressure created inside the ventilator circuit is equal to the pressure delivered to the patient's air ways only up to a certain maximum flow rate. The smaller size of the prongs and the lower the adjusted CPAP, the lower is the maximum flow that may be used without danger of low pressure alarm failure. More information about alarm settings during noninvasive CPAP ventilation of preterm infants is needed.

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