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Application of equal loudness contour in assessing the impact of random noise: Case study in continuous positive airway pressure systems with helmet

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ABSTRACT

Empirical evidence is presented on the application of equal-loudness contour for assessing the exposure to noise to which patient are subjected when receiving continuous positive airway pressure respiratory support via a helmet. Sound pressure levels are compared in one third-octave bands with respect to the isophon lines and the influence of the heat and moisture exchanger filters coupled to the helmets on the sound pressure levels during the supply of air–oxygen is analyzed. This study has demonstrated that an appropriate change of scale enables a more objective assessment of the whole auditory spectrum and the levels of noise produced by the continuous positive airway pressure respiratory support via a helmet exceed the 40-phon line and levels of 45 dB(A) recommended by various organizations.

1. Introduction

Several studies have revealed that, in general, high noise levels contribute to hearing loss and other physiological factors [1,2]. The medium and long term effects of stimulation by repeated high-intensity noises have been detected in infants; these could also have serious repercussions on their behavior [3].

In the survival of infants, one of the modalities of respiratory support is continuous positive airway pressure (CPAP) by means of a helmet. Recent studies demonstrate that these systems generate a noise level that can be harmful to the developing hearing system of premature babies [4–6].

One of the tools that enable discomfort or harm to be correlated with the levels of sound pressure to which humans are subjected is the determination of curves of equal loudness. These normalized curves are determined for normal individuals aged between 18 and 30 years (ISO 226:1987), but despite this, they serve as a useful reference for quantifying the relative discomfort suffered in individuals of other ages.

Curves of equal loudness were presented for the first time by Munson and Fletcher in 1930, and were later modified by Robinson and Dadson [7,8]. These curves calculate the relationship existing between the frequencies and intensities (in dB) of two pure tones so that these sounds are perceived equally strongly by the human ear. Thus all the points configured in this way represent an isophone curve that has the same loudness.

It is established that the value in phons corresponds to a loudness in dB (dB-SPL) at the frequency of 1 kHz, that is, 0 phon corresponds to a loudness of 0 dB at 1 kHz, based on the free-field measurements.

One of the characteristics presented by these curves is that, as the level of sound pressure increases, the curves become flatter; in other words, their dependence on the frequency diminishes as the level of sound pressure increases.

These curves gave rise to the filters employed today in the sound level meters, commonly termed A, B and C filters; in these meters the A, B and C weighting filters have profiles similar to the isophone curves of 30, 70 and 100 phons, respectively.

The weighting filter A is the type most frequently employed in practice; however, the indiscriminate use of any type of filter for weighting without taking into account the level of sound measured makes the value of measurement rather remote from the reality of perception.

The present study is focused on a comparison of two measurement scales, dB and phons, to assess the impact of random noise. These measurement scales will be used to study the noise exposure of people treated with CPAP helmets. Similarly, it will be analyzed

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the influence that HME (Heat and Moisture Exchangers) filter (Fig. 1) placed on the helmet, has on sound pressure levels.

2. Materials and methods

Two different systems of CPAP using neonatal helmets have been compared: one with the StarMed Castar Helmet (Mirandola, Italy) and the other with the DimAir helmet (Mirandola, Italy). The measurements of the levels of sound pressure were performed placing these two CPAP system helmets over the head of a mannequin (Type 4100, of Brüel&Kjær). For the measurements of the sound pressure levels a calibrated pistonphone (Model 4231 of Brüel&Kjær) was employed to check the chain of measurement at the start and end of each measure; a Pulse analyzer (model 3560C, of Brüel&Kjær) was also used; the microphones were placed in the internal ears of the mannequin; another microphone measured the ambient noise level. The Velocical Plus instrument of TSI was employed to measure the environmental conditions [8].

All the measurements were performed in the Neonatal Intensive Care Unit (NICU) of the "Puerta del Mar" University Hospital (Cadiz, Spain). The equipment was configured, after the pretest, in a dynamic range between 0 and 140 dB, a bandwidth of 20 Hz and 20 kHz, and for linear measurements in A weighting. The human ear does not respond equally to all frequencies: it is very much more sensitive to sounds of 1–4 kHz [9]. Moreover, in the postnatal environment, the noises are unpredictable and predominantly of high frequency [10], basically the tones at which the alarms of the monitoring instruments are configured. Given the above, the measurements of sound pressure were made in frequency bands of one third of an octave, and with Fast Fourier Transform (FFT) spectral analysis, with the object of evaluating the real behavior of the noise emitted by the air-oxygen flow rate in the different frequency bands. In addition, it was decided to compare the spectra with the curve of 40 phons, to evaluate in the most favorable conditions the adverse effects that a healthy person would experience [11].

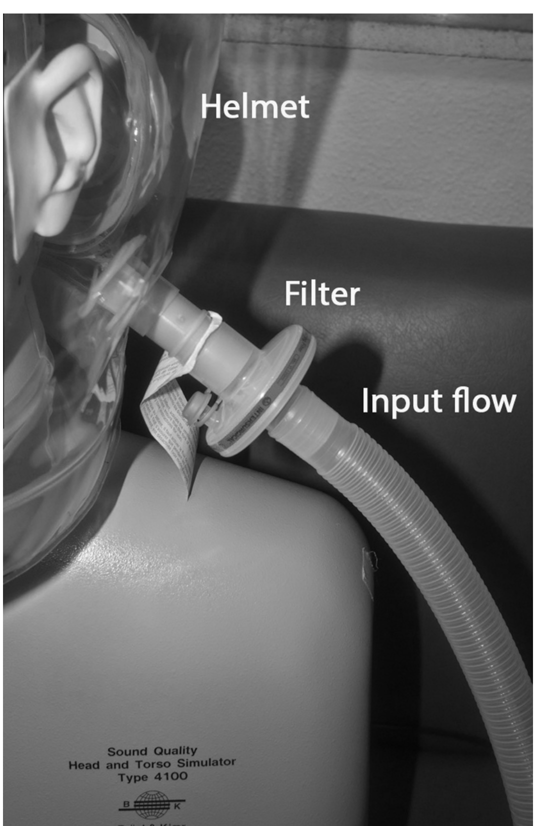

Fig. 1. System CPAP-helmet.

A total of 4968 measurements of noise of 5 s each were made; 216 of the measurements were of background noise. Due to the wide variations in the noise levels detected in the NICU [10], all the measurements were made in thirds of an octave, without filters and with A weighting. The various different parameters of the sound pressure levels, L10, L90, Lleq, LeqA, Lmax and Lmin, were determined in periods of 3 min, and the central trend statistics were used for the corresponding analyses. The sensitivity values of the equipment were also checked with a calibrated pistonphone at the start and end of the measurements.

To be able to correlate the noise levels recorded in the interior and exterior of the helmet, three microphones were employed, one each for the left and right ears (inside the helmet), and another for measurements external to the helmet. In this way it was possible to discount measurement samples where the external noises were higher than those in the helmet interior; these would have been generated by activations of alarms and other typical occurrences within a room of the NICU.

Flow values represent a controversial aspect. If low flows were used rebreathing [12] of CO_2 could occur which can lead to the pCO_2 increase in the patient's blood. It has been found in healthy adult volunteers that the minimum flow for this to occur is 30 lpm. The CO_2 production in young infants is lower so the minimum flow is probably less; in this sense, some authors recommended in neonates 10 lpm flows up, but without ensuring their safety so far given the impossibility, for ethical reasons, of conducting such experiments with healthy infants. Therefore, many authors recommend flows above 20–25 lpm [13,14].

The noise measurements were made for different air–oxygen flow rates: 20, 30 and 40 lpm without filter and with a Heat and Moisture Exchange (HME) filter (Clear Therm-Micro model 1441, Tyco, Mirandola, Italy) and a respirator CGM filter (Clear-Guard Midi model 1644, Tyco, Mirandola, Italy) placed in the inspiratory branch of the CPAP circuit. During the study, the pressure in the interior of the helmet was kept constant, in all the conditions, at 5 cm H_2O. The environmental conditions recorded were as follows: ambient temperature of 22.1 ± 1 °C; relative humidity within the range of 52.9 ± 5%.

In order to determine the temporal and spatial conditions most favorable for the performance of the study, a pre-test was carried out over the course of a working day of 16 h, with the object of discriminating the time band when background noise was highest. The purpose was to avoid those times when very high levels of background noise that typically occur in the NICU would, according to various previous research studies [15,16], influence the noise generated by the gas introduced into the interior of the helmet.

For the comparison with the curve of equal loudness it is considered that the Munson and Fletcher curves and those of Robinson and Dadson are only valid for direct sound fields, since they do not take into account that sounds are not perceived equally if they originate from different directions [17]. For this reason curves of equal loudness have been employed, with the added corrections proposed for diffuse sound fields by the standard ISO 226:1987 (UNE 74003:1992 in Spain is still in effect), as expressed by Eq. (1). The curves represented in Fig. 2 have been constructed under these directives; they illustrate a comparison of the advantages and disadvantages of applying the A weighting filter to the curves of equal loudness.

$$L_N(\text{fon}) = 4,2 + [a_f(L_f - T'_f)]/[1 + b_f(L_f - T'_f)] \quad (1)$$

$$L_f(\text{fon}) = T'_f + [L_N - 4,2]/[a_f - b_f L_N] \quad (2)$$