In vitro investigations of jet-pulses for the measurement of respiratory impedance in newborns

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In vitro *investigations of jet-pulses for the measurement of respiratory impedance in newborns. M. Schmidt, B. Foitzik, O. Hochmuth, G. Schmalisch.* ©ERS Journals Ltd 1999.
ABSTRACT: The aim of this *in vitro* study was to investigate the measuring range and accuracy of a miniaturized equipment for respiratory impedance (Z_{rs}) measurements in newborns using jet-pulses.

Brief flow pulses (peak flow=16 L·min^{-1}, width=10 ms) were generated by a jet-generator consisting of a solenoid valve and an injector, situated between pneumotachograph and outflow resistance. Serially arranged resistance-inertance-compliance (R-I-C) lung models (R_M=1.3-6.4 kPa·L^{-1}·s, C_M=7.4-36.9 mL·kPa^{-1}, I_M=1.5 Pa·L^{-1}·s^2) were used to measure the real and imaginary part of Z_{rs} between 4 and 50 Hz and to determine R, C and I by means of the method of least squares.

The median errors for R, C and I were -0.1 kPa·L^{-1}·s (-2%), 2.4 mL·kPa^{-1} (13%) and -0.2 Pa·L^{-1}·s^2 (-13%) for measurements without breathing signals and 0.11 kPa·L^{-1}·s (3%), 3 mL·kPa^{-1} (16%) and 0.28 Pa·L^{-1}·s^2 (19%) in mechanically ventilated models. During spontaneous breathing the influence of the breathing flow on Z_{rs} was negligible. The equipment did not show any nonlinearity when different pulse amplitudes were used ($\dot{V}_{max}$=13–22 L·min^{-1}).

The investigations have shown that jet-pulses allow reliable measurements of respiratory impedance and have the potential to provide valuable information about lung mechanics in spontaneously breathing and mechanically ventilated newborns. The developed measuring head has a low apparatus dead space, is easy to disinfect, has standard connections and can be used as the T-piece in a ventilator circuit.
Eur Respir J 1999; 14: 1156–1162.

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Suitable techniques for measurements of respiratory impedance (Z_{rs}) in infants by using the forced oscillation technique (FOT) are still being developed and commercial devices for this age group are not available [1]. Commercial devices for FOT developed for adults cannot be adapted easily for measurements in infants firstly due to differences in the technical requirements (*e.g.* apparatus dead space, signal resolution), and secondly because of another interesting measuring range (*e.g.* higher respiratory impedance, problems in the lower frequency range due to the high respiratory rate, higher resonant frequency). One of the most important problems for FOT in newborns is the application of the oscillations into the airways *via* a face mask [2]. BISGAARD and KLUG [2] and KLUG and BISGAARD [3] developed a special face mask which allowed measurements to be taken in young children using the IOS-MasterScreen (Jaeger, Würzburg, Germany).

For FOT in infants, monofrequent sinusoidal signals [4, 5] and polyfrequent signals in the form of pseudorandom noise [6–9] or pulses [10] have already been used. Regardless of which test signal is used, so far it has always been generated by loudspeakers. Despite, their advantages of optimizing the spectra of the test signal, they are cumbersome in bedside application and difficult to insert into a ventilator circuit.

Therefore, the authors have investigated the suitability of jet-pulses for FOT in spontaneously breathing and mechanically ventilated newborns. Jet-pulses can be easily generated by an injector which allows for a miniaturization of the measuring head. In contrast to loudspeakers, jet-generators are easier to disinfect and fulfil the clinical demands for measurements in intensive care units better.

So far, pulses have rarely been used for FOT in infants. Therefore, the aim of this study was to investigate the measuring range and accuracy of respiratory impedance measurements by jet-pulses using mechanical lung models with known parameters. *In vitro* measurements were performed because no reference method for FOT in newborns was available to investigate the accuracy of the developed technique.

Methods

Measuring principle

The measuring principle used is shown in figure 1. The flow pulses generated by the jet-generator are fed through the measuring head into the patient's airways while the patient can breathe through the pneumotachograph (PNT) and the outflow resistance (R_{out}) This means that only one part of the flow pulse reaches the airways while the other escapes through R_{out}. An example of the superimposition of the breathing signals with the jet-pulses is shown in figure 2. The breathing signal must be separated from the measured flow and pressure signals and the resulting
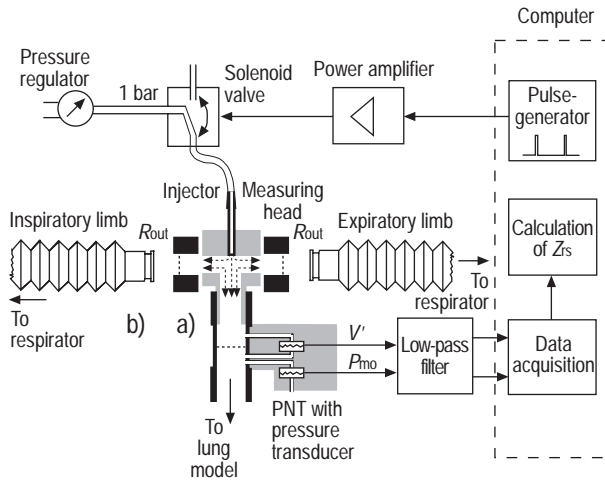
Fig. 1. – Principle of the developed forced oscillation technique (FOT) setup using a jet-generator to produce brief flow pulses (V′: air flow; Pmo: mouth pressure). a) Setup for measurements on spontaneously breathing patients (Rout: outflow resistance). b) Setup for measurements during mechanical ventilation. The measuring head was used as a T-piece in the ventilator circuit. Zrs: respiratory impedance; PNT: pneumotachograph.
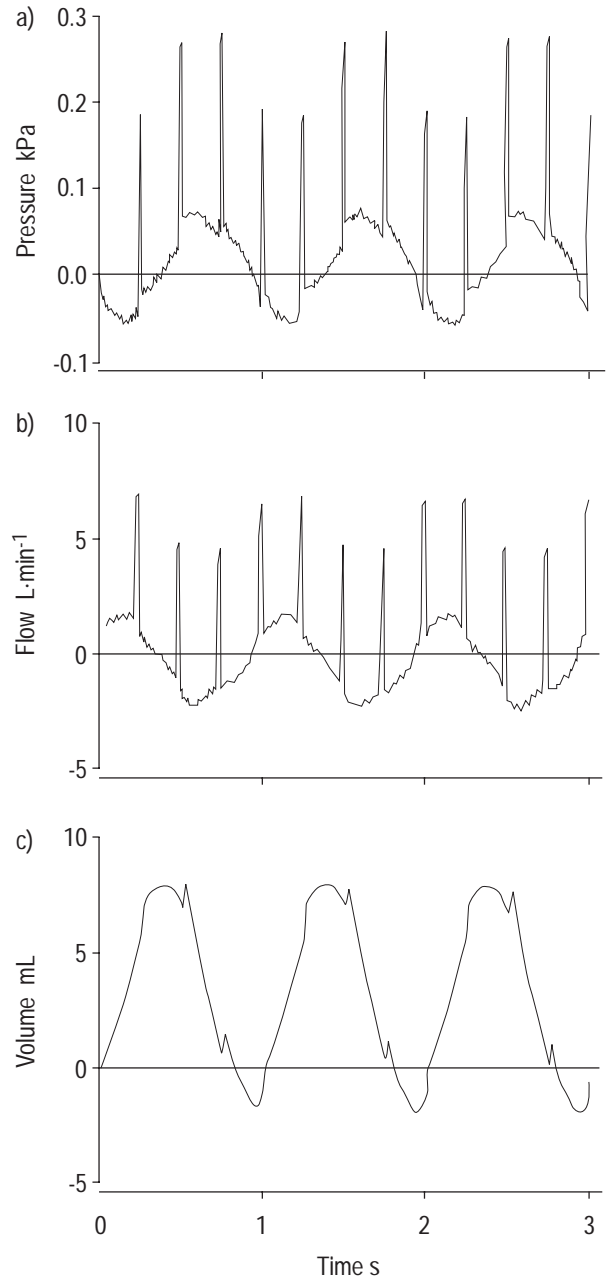

Fig. 2. – Spontaneous breathing signals superimposed by the pulse responses of the respiratory system measured by the developed forced oscillation technique (FOT) for pressure (a), flow (b) and volume (c).

pulse response was used to calculate $Z_{rs}(j\omega)$ according to the method of MICHAELSON *et al.* [11]:

$$Z_{rs}(j\omega) = \frac{\overline{S_{PP}(j\omega)}}{\overline{S_{V'P}(j\omega)}}, \quad (1)$$

where $S_{PP}(j\omega)$ is the mean pressure auto-spectra, $S_{V'P}(j\omega)$ the mean flow-pressure cross-spectra averaged over a given number of pulses, j the unit of imaginary numbers and $\omega = 2\Pi f$ the angular frequency with f being the measuring frequency.

With the mean flow auto-spectra ($S_{V'V'}(j\omega)$) the coherence function ($\gamma^2(\omega)$) was calculated to assess the reliability of the measurements and to describe the causality between the flow and pressure signal and the presence of extraneous noise:

$$\gamma^2(\omega) = \frac{\overline{S_{V'P}(j\omega)}}{\overline{S_{V'V'}(j\omega)} \times \overline{S_{PP}(j\omega)}} \quad (2)$$

Mechanical respiratory models

Linear one-compartment respiratory models were used to simulate the respiratory impedance:

$$Z_{rs}(j\omega) = R_M + 1/(j\omega \times C_M) + (j\omega \times I_M) \quad (3)$$

where compliance (C_M), resistance (R_M) and inertance models (I_M) were connected in a series.

The parameters of the respiratory models used were chosen to be similar to those of newborns [12]. The resistances ($R_{M1}=1.3$ kPa·L^{-1}·s, $R_{M2}=2.5$ kPa·L^{-1}·s, $R_{M3}=3.7$ kPa·L^{-1}·s, $R_{M4}=6.4$ kPa·L^{-1}·s) were made by using different meshes of polyamide in a tube (ID = 1.4 cm, length = 6 cm). The compliance models were built from rigid copper chambers with volumes of 1, 2.5 and 5 L and had almost adiabatic measuring conditions during the pulse response time. This is due to thermal time constants for heat exchange between the gas inside the copper chamber and the inner wall of the chamber being >4 s, *i.e.*

much longer than the evaluated pulse response of 250 ms). The adiabatic compliance values of the used chambers were $C_{M1}=7.4$ mL·kPa^{-1}, $C_{M2}=18.7$ mL·kPa^{-1}, and $C_{M3}=36.9$ mL·kPa^{-1}. The inertance of the lung models (I_M) was mainly determined by the gas inside the connection tubes (ID=1.4 cm, length= 18 cm) and was analytically calculated from the geometrical dimensions by:

$$I_M = \varsigma \times 1/A \quad (4)$$

where ς is the gas density, 1 the length and A the cross-sectional area of the connection tubes.

The same connection tubes were used in all models meaning that $I_M=1.5$ Pa·L^{-1}·s^2 was constant in all measurements.

Spontaneous breathing was simulated by a motor driven pump connected to a lateral port of the copper chambers, whereby the volume of the pump is negligible compared to the volume of the copper chambers (<1%). For the measurements, breathing rates of 0, 30, 60 and 90 min^{-1} with a constant tidal volume of 10 mL were used.

Jet-generator

Brief flow pulses were generated by a jet-generator consisting of an injector (ID=1.8 mm, length=15 mm) and a solenoid valve (D 18220; DYNAMCO, Inc., McKinney, TX, USA) driven by a computer-controlled power-amplifier. A driving pressure ($P0$) of 1 bar was used to prevent any influences of the measured impedance on the shape and amplitude of the generated pulses. The injector and the valve were connected by a flexible tube (ID=3.2 mm, length=1 m) to make the measuring head flexible and moveable. The measurements were performed with pulses of 3.5 mL volume, a pulse width of 10 ms, a peak flow of 16 L·min^{-1} and a pulse rate of 4 pulses·s^{-1} (fig. 3).

Measuring technique

The flow signal was measured by a miniaturized neonatal PNT (Jaeger; RPNT=0.6 kPa·L^{-1}·s at 5 L·min^{-1}) which
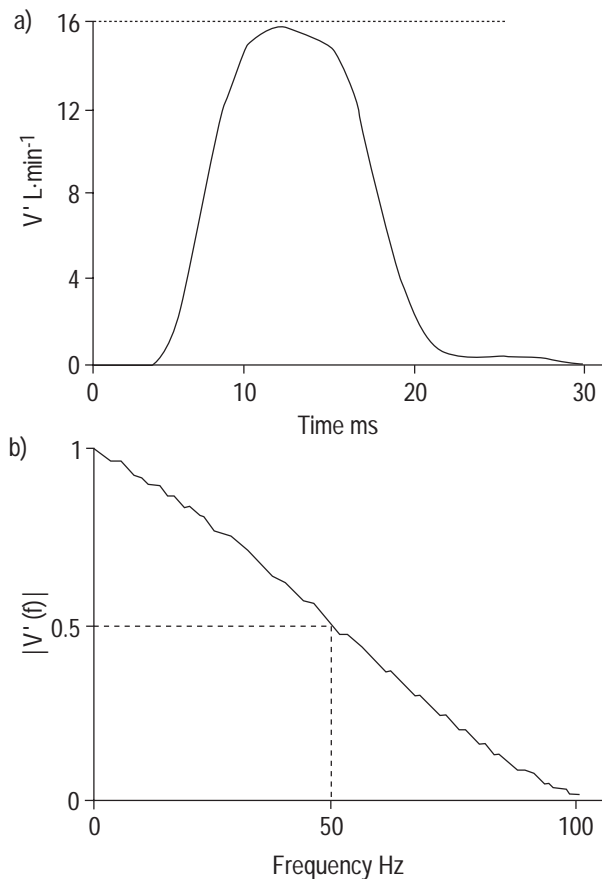

Fig. 3. – Shape of the flow pulse generated by the jet-generator with a maximal flow of 16 L·min^{-1} (- - - -) and a volume of 3.5 mL (a) and the amplitude spectrum of the flow pulse (b). - - - -: lowest amplitude in the frequency range 0–50 Hz. V': airflow; $|V'$ (f)|: standardized amplitude spectrum.

contained the differential pressure transducer for the flow signal and a separate pressure transducer for the mouth pressure. Both signals were low-pass filtered to avoid aliasing (8th-order Bessel-filter, cut-off frequency 100 Hz) and sampled at 1,024 Hz by using a 16 bit analogue digital converter (AT-MIO 16XE10; National Instruments, Austin, TX, USA). The common-mode rejection ratio (CMRR) of the flow channel was >50 dB up to 50 Hz, measured with sinusoidal oscillations.

For measurements during spontaneous breathing, two parallel outflow screen resistances Rout with negligible frequency dependence were used (fig. 1a). The total resistance of both parallel Rout was 1.4 kPa·L^{-1}·s measured at a constant flow of 5 L·min^{-1}. Thus, the resistance of the whole measuring system including RPNT was 2 kPa·L^{-1}·s so that the peak pressure inside the system for an occluded PNT was 0.53 kPa. The apparatus dead space of the measuring head was 3.5 mL and was determined by water displacement.

Measurements during mechanical ventilation were performed by using the authors′ measuring head instead of the T-piece of the ventilation circuit without any additional outflow resistance (fig. 1b). The gas flow into the lungs depends on the impedance of the respiratory system and the impedance of the respiratory circuit. The measuring head and the respirator (BP 2001; BEAR Inc., Palm Springs, CA, USA) were connected with tubes of 1.2 m length and an ID of 9 mm. For the respiratory models used, the volume lost of the generated flow pulses (volume 3.5 mL) through the respirator tubings was between 1.0 and 2.1 mL for inspiration and between 1.3 and 2.2 mL for expiration.

The measurements were performed with dry gas from the central gas supply (22±2°C, FI,O$_2$=0.21) in the intermittent positive-pressure ventilation (IPPV) mode without positive end-expiratory pressure. The respiratory rate was 30 min^{-1} and the inspiration to expiration time ratio was 1:1. The peak pressure was 2.5 kPa for CM1, and 1.5 kPa for CM2 and CM3.

Signal processing

The separation of the spontaneous breathing from the measured flow and pressure signals was performed by using a digital high-pass filter with a 2 Hz cut-off frequency. After filtering, a time window of 250 ms (256 sample points) was used to evaluate the pulse response, and an additional base line correction was carried out in order to ensure that the pulse response was zero at the beginning and end of the time window. The spectra of the resulting flow and pressure signals were calculated by fast Fourier transform on blocks of 1,024 sample points, where the missing samples were zero padded to realise a frequency resolution of 1 Hz for the given time window.

The measured impedance Zrs(jω) was determined from equation 1 and was corrected according to FARRÉ et al. [13] using the impedance of the occluded system Zocc(jω):

$$Z_{rs,corr}(j\omega) = (1/Z_{rs}(j\omega) - 1/Z_{occ}(j\omega))^{-1} \quad (5)$$

to reduce the influence of any asymmetry in the differential inputs of the pressure transducer and its connected PNT on measured impedance.

The real part $R_{rs}(f)$ and imaginary part $X_{rs}(f)$ of $Z_{rs}(f)$ ($Z_{rs}(f) = R_{rs}(f) + j \times X_{rs}(f)$) of the corrected impedance were finally plotted in the frequency range between 4–50 Hz in 1 Hz steps. This frequency range includes the resonant frequency ($f$res) of the models ($X_{rs}(f$res$) = 0$).

Data analysis

After elimination of the breathing signals, the spectra of the resulting flow and pressure signals were averaged over 12 pulses (3 s) to reduce random errors in the calculation of $Z_{rs}(j\omega)$ according to equation 1 and to calculate $\gamma^2(\omega)$. The measured $Z_{rs}(j\omega)$ data were interpreted as an *R-I-C* series model as described by equation 3 where R is the mean real part of $Z_{rs}(j\omega)$. C and I were calculated from $X_{rs}(j\omega)$ by the method of least squares (MLS) in the frequency range of 4–50 Hz.

To investigate the short-term reproducibility of the measured $Z_{rs}(j\omega)$, 5 time intervals of 3 s were used for each model to calculate the mean and SD for R, C and I.

Results

The comparison of the calculated and measured real and imaginary part of $Z_{rs}(j\omega)$ is shown in figure 4. Thereby, either the resistance was changed and the compliance kept constant, or vice versa. The impedance of the models was calculated according to equation 3 using RM, CM and

IM. The agreement with the measured $R_{rs}(f)$ and $X_{rs}(f)$ was good and there were no interdependencies from changes of RM on the imaginary part and from changes of CM on the real part, with the exception of the model with the highest resistance (RM4=6.4 kPa·L^{-1}·s). For this model a slightly increased imaginary part compared to the other models was measured at $f > 10$ Hz. For a constant CM (fig. 4a and b) the calculated resonant frequency was 30 Hz and the measured resonant frequencies were only slightly underestimated (28.5±1.97 Hz). For a constant resistance (fig. 4c and d) the differences between the calculated and the measured resonant frequencies were <1 Hz.

For the impedance measurements shown in figure 4, lung mechanic parameters were estimated by the MLS and were compared to the model parameters (table 1). The median absolute errors (range in brackets) in the resistance were -0.1 (-0.45–0.03) kPa·L^{-1}·s, in the compliance 2.4 (0.7–6.5) mL·kPa^{-1} and in the inertance -0.2 (-0.24–0.28) Pa·L^{-1}·s^2. The relative errors were -2.2% (-7.0%–2.3%), 12.8% (9.5%–17.6%), -13.3% (-16.0%–18.7%) for R, C and I respectively.

The linearity of the measuring system and the models was assessed by applying flow pulses of different amplitudes from 12.6–22.5 L·min^{-1} to an *R-I-C* lung model (RM3 =3.7 kPa·L^{-1}·s, CM2=18.7 mL·kPa^{-1}, IM=1.5 Pa·L^{-1}·s^2). The different flow amplitudes were generated by driving pressures from 0.5 to 2 bar. In this flow range the

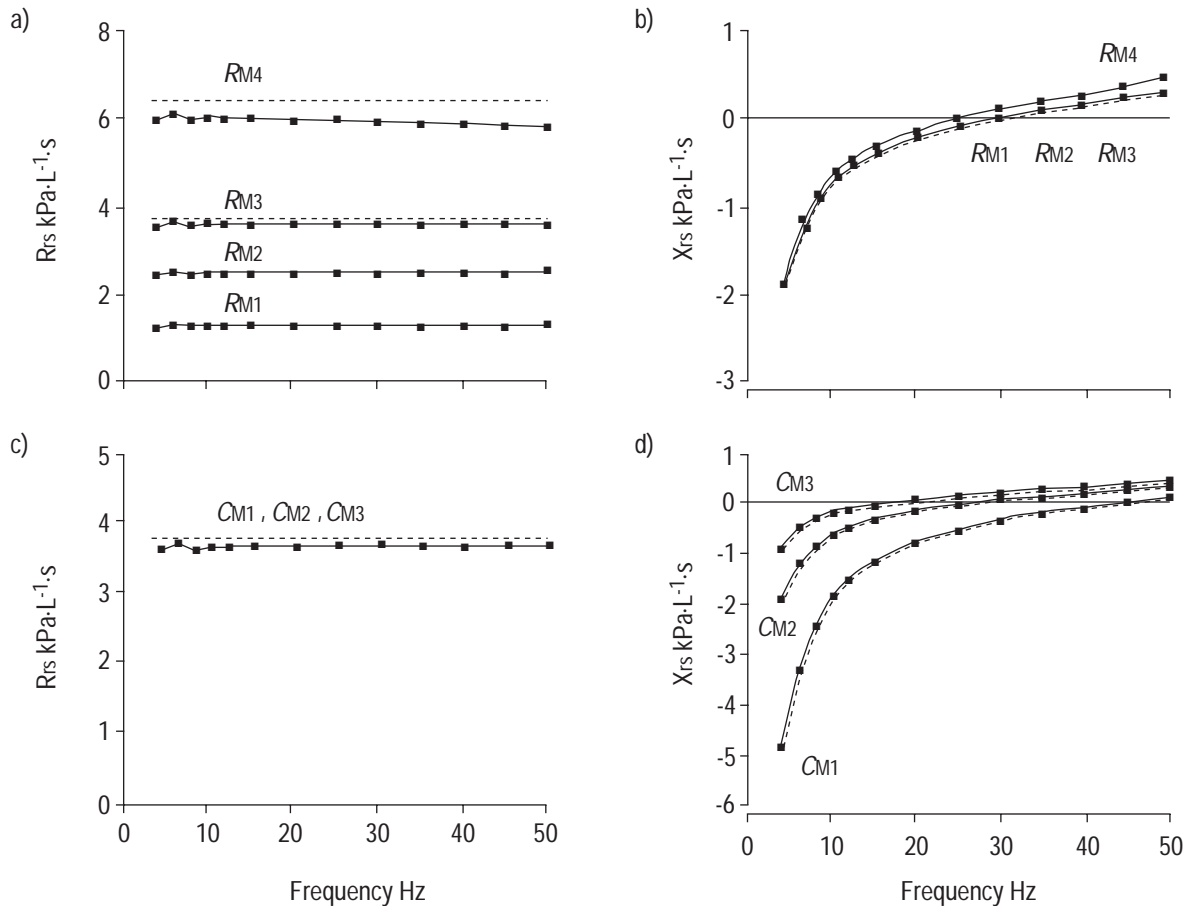Fig. 4. – Comparison of the calculated (- - -) and measured (——) real and imaginary part of passive resistance-inertance-compliance (*R-I-C*) lung models. a and b) Changed resistance at constant compliance of the model CM=18.7 mL·kPa^{-1}; c and d) changed compliance at constant resistance of the model RM=3.7 kPa·L^{-1}·s. The inertance of the model (IM) was kept constant (1.5 Pa·L^{-1}·s^2).

Table 1. – Accuracy of resistance (*R*), compliance (*C*) and inertance (*I*) determined by the method of least squares from measured respiratory impedance on passive lung models

	Model		Measured parameters			Absolute (relative) errors		
No.	R_M	C_M	R kPa·L^{-1}·s	C mL·kPa^{-1}	I Pa·L^{-1}·s^2	ε_R kPa·L^{-1}·s	ε_C mL·kPa^{-1}	ε_I Pa·L^{-1}·s^2
1	3.7	7.4	3.60±0.00	8.1±0.00	1.30±0.00	-0.10 (-2.7%)	0.7 (9.5%)	-0.20 (-13.3%)
2	3.7	18.7	3.62±0.00	21.1±0.19	1.26±0.05	-0.08 (-2.2%)	2.4 (12.8%)	-0.24 (-16.0%)
3	3.7	36.9	3.57±0.14	43.4±0.63	1.30±0.00	-0.13 (-3.5%)	6.5 (17.6%)	-0.20 (-13.3%)
4	6.4	18.7	5.95±0.00	21.5±0.09	1.78±0.04	-0.45 (-7.0%)	2.8 (15.0%)	0.28 (18.7%)
5	3.7	18.7	3.62±0.00	21.1±0.19	1.26±0.05	-0.08 (-2.2%)	2.4 (12.8%)	-0.24 (-16.0%)
6	2.5	18.7	2.52±0.00	20.9±0.19	1.40±0.00	0.02 (0.8%)	2.2 (11.8%)	-0.10 (-6.7%)
7	1.3	18.7	1.33±0.00	20.8±0.08	1.44±0.05	0.03 (2.3%)	2.1 (11.2%)	-0.06 (-4.0%)

Data are mean±SD. ε_R, ε_C and ε_I were the absolute errors between measured and calculated values for *R*, *C* and *I*, respectively. The inertance of the model (I_M) was kept constant (1.5 kPa·L^{-1}·s^2). R_M: resistance of the model; C_M: compliance of the model.

amplitude of the generated flow pulse only had a minimal influence on the estimated *R* (<0.1 kPa·L^{-1}·s), *C* (<0.9 mL·kPa^{-1}) and *I* (<0.2 Pa·L^{-1}·s^2). These deviations are smaller than the measuring errors shown in table 1.

The influence of the breathing rate on $Z_{rs}(j\omega)$ was investigated using the model with R_{M3}=3.7 kPa·L^{-1}·s, C_{M2}= 18.7 mL·kPa^{-1} and I_M=1.5 Pa·L^{-1}·s^2. For breathing rates 30–90 min^{-1} no measurable influences on *R* and *I* were found. By increasing the breathing rate the measured compliance decreased by about 1.6 mL·kPa^{-1}. However, with a higher breathing rate the SD of $R_{rs}(f)$ and $X_{rs}(f)$ increased, especially in the lower frequency range (*f* <10 Hz). For a measuring frequency of 4 Hz and a respiratory rate of 30 min^{-1} the SD of $R_{rs}(f)$ and $X_{rs}(f)$ were 0.22 kPa·L^{-1}·s and 0.27 kPa·L^{-1}·s and for a respiratory rate of 90 min^{-1} these SD increased to 1.05 kPa·L^{-1}·s and 1.28 kPa·L^{-1}·s respectively. For measuring frequencies greater than 10 Hz the SD of $R_{rs}(f)$ and $X_{rs}(f)$ were <0.18 kPa·L^{-1}·s for all investigated breathing rates.
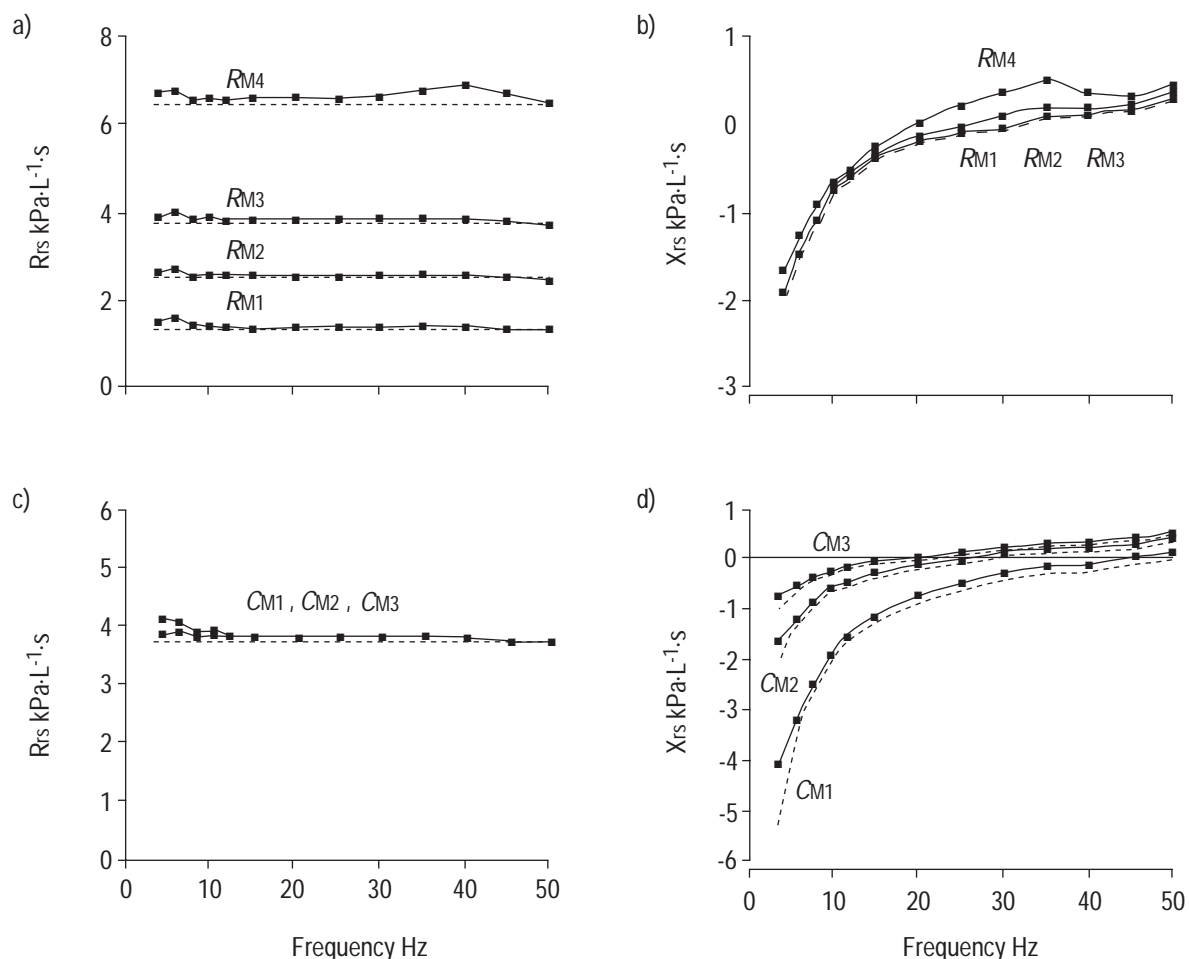

Fig. 5. – Comparison of the calculated (- - -) and measured (——) real and imaginary part of mechanically ventilated resistance-inertance-compliance (*R-I-C*) lung models. a and b) Changed resistance at constant compliance of the model (C_M)=18.7 mL·kPa^{-1}; c and d) changed compliance at constant resistance of the model (R_M)=3.7 kPa·L^{-1}·s. The inertance of the model (I_M) was kept constant (1.5 Pa·L^{-1}·s^2).

In all measurements the γ^2 was greater than 0.95 in the frequency range of 4–50 Hz, with the exception of the breathing rate of 90 min^{-1}. For this breathing rate the γ^2 was slightly decreased at low measuring frequencies (lowest value γ^2=0.86 at 4 Hz).

In figure 5 the calculated and measured real and imaginary parts of the same lung models as shown in figure 4 were compared. In contrast to figure 4 the models were mechanically ventilated to investigate the accuracy of impedance measurements during mechanical ventilation. A similar high agreement was found between the calculated and measured $R_{rs}(f)$ and $X_{rs}(f)$ as shown in figure 4, with the one exception of the model with the highest resistance R_{M4}=6.4 kPa·L^{-1}·s where the impedance curves had a higher roughness. In the models (fig. 5a and b) with constant resonant frequencies (30 Hz) the measured resonant frequencies were underestimated (24.7±3.7 Hz). In the models with different C_M (fig. 5c and d) the mean difference between calculated and measured resonant frequencies was 3.9±1.3 Hz. The accuracy of measurements during mechanical ventilation did not differ significantly from the measurements in the passive models (table 1). The median absolute errors (range in brackets) in R were 0.11 (0.03–0.19) kPa·L^{-1}·s, in C 3.0 (1.0–7.8) mL·kPa^{-1} and in I 0.28 (-0.12–1.1) Pa·L^{-1}·s^2. The relative errors were 3.0% (1.2%–4.6%), 16.2% (5.3%–21.1%), 18.7% (-8.0%–73.3%) for R, C and I respectively.

Discussion

Measurement of $Z_{rs}(j\omega)$ in newborns at bedside requires a miniaturized measuring head with low apparatus dead space and low weight. Furthermore, the measuring head must fulfil clinical requirements, especially easy disinfection and application in infants requiring ventilatory support or mechanical ventilation. In contrast to loudspeaker systems commonly used to generate the test signal, jet-generators fulfil these requirements better since they allow very small and lightweight measuring heads to be developed. Our measuring head has standard connections and was designed in the shape of a T-piece, so that it can also be used in a ventilator circuit.

The generated flow pulses contain a sufficiently wide frequency spectrum to allow for the determination of $Z_{rs}(j\omega)$ between 4 and 50 Hz where $Z_{rs}(j\omega)$ of infants is influenced by several respiratory diseases [14, 15]. The resonant frequency of the respiratory system also lies in this frequency range. SLY et al. [8] have measured resonant frequencies in nonintubated infants between 10–20 Hz and SULLIVAN et al. [10] in intubated infants between 6–16 Hz. MARCHAL et al. [7] found a mean resonant frequency of 10.5 Hz in 24 spontaneously breathing infants.

One significant advantage of the impedance measurement by means of pulses is the short duration of the measurement, thus allowing for a high time resolution so that changes in the lung mechanics during the breathing cycle can be detected, especially in the case of very low breathing rates [8]. In contrast to periodical test signals which require a steady state, the use of aperiodical signals allows the dynamic properties of the respiratory system to be determined from the transient response. In the equipment for this study a time window of 250 ms was used to analyse the pulse response and a pulse rate of 4 pulses·s^{-1}. Because of several signal disturbances (e.g. from the breathing signal, noise) it is necessary to average the signal spectra over a number of pulse responses in order to calculate $Z_{rs}(j\omega)$. However, the pulses can be triggered by the breathing cycle so that the determination of $Z_{rs}(j\omega)$ from pulse responses at different breathing levels is possible.

One disadvantage of the jet-pulses for FOT is that high flow amplitudes are necessary to achieve sufficient signal amplitudes in the frequency range used. Therefore, in contrast to periodical multifrequent signals with discrete frequencies, the FOT by means of pulses needs higher signal amplitudes that increase the influence of nonlinearities on the impedance measurement. The measurements have shown that in the investigated flow range the equipment did not show any nonlinearity.

In contrast to FOT with bidirectional flow pulses generated by loudspeakers [2, 3, 16] the flow pulses in the equipment used in this study are always positive, which can inadvertently lead to a positive end-expiratory pressure inside the respiratory system if the gap between 2 pulses is too short. Calculations have shown that the end-expiratory pressure is <0.01 kPa for a respiratory time constant of 150 ms in the current system and negligible from the clinical point of view.

In newborns the measurement of $Z_{rs}(j\omega)$ is hampered at lower frequencies by their high respiratory rate (up to 120 breaths·min^{-1}). Previous studies have shown that the lower frequency ranges have proven very informative in adults [2, 17–19], children [20] and infants [6, 8, 15]. Furthermore, especially in preterm newborns the breathing pattern is often irregular and disturbed by various physiological occurrences (sighs, swallows, cardiac activities) and technical artefacts (drift, noise, air leaks) [21]. Therefore, in newborns the lower cut-off frequency of the respiratory signals may be additionally increased. DESAGER et al. [6] reported measuring problems of $Z_{rs}(j\omega)$ at low frequencies in infants due to spontaneous breathing. They found a high coefficient of variation and a low coherence at f <16 Hz. These findings match the current results.

Newborns have a much higher $Z_{rs}(j\omega)$ than adults and therefore the impedance of the occluded measuring head ($Z_{occ}(j\omega)$) must be much higher than in equipment developed for adults. This requires a low compressible volume of the measuring head and extremely symmetrical pressure transducers with high CMRR and identical frequency response [13, 22–24]. The CMRR of the Jaeger-pneumotachograph developed for infants which was used in this study was lower than in other set-ups which had a CMRR >60 dB [2, 4, 7]. This might explain some of the results. As shown in figure 4, the measuring error increased with increased resistance. FARRÉ et al. [13] have shown that the lower the CMRR the higher the measuring error for high $Z_{rs}(j\omega)$. Thus, a numerical correction of $Z_{rs}(j\omega)$ according to equation 5 is necessary to reduce this error. Therefore, the correct determination of $Z_{occ}(j\omega)$ has an important influence on the measuring accuracy for measurements in infants with a high $Z_{rs}(j\omega)$.

Although the volume of the current flow pulse is relatively low (<3.5 mL), the accuracy of the impedance measurements was acceptable and the short-term reproducibility of the measurements was high. The good correlation between impedance data obtained with and without

superposition of breathing signals for $f > 4$ Hz showed that the pulses had been successfully separated from the measured pressure and flow signals.

In conclusion, the model investigations have shown that jet-pulses allow reliable measurements of the respiratory impedance and a miniaturization of the measuring technique. The measuring equipment developed has the potential to give valuable information about lung mechanics in newborns breathing spontaneously as well as in newborns requiring mechanical ventilation.

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