Eur Respir J, 1993, 6, 767–769
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European Respiratory Journal
ISSN 0903 - 1936

EDITORIAL

The forced oscillation technique in intubated, mechanically-ventilated patients

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In a recent discussion concerning the use of the forced oscillation technique (FOT), PRIDE [1] stated that, although anaesthesia and intensive care are among the obvious areas of application of the method, there is very little information on these applications. The paper by PESLIN *et al.* [2] in the present issue of the Journal describes one of the first attempts to apply the FOT to mechanically-ventilated patients.

FOT, first described by DUBOIS *et al.* [3] in 1956, is a technique in which externally produced pressure oscillations are applied to the respiratory system. The resulting oscillations in airflow, related to the corresponding pressure oscillations, allow a direct measurement of the impedance, *i.e.* the pressure-flow relationships of the system. Generally, the oscillations are produced by a loudspeaker, and are applied at the mouth of the subject. The impedance is calculated from the signals of a pressure transducer and a pneumotachograph placed close to the subject's mouth. Because of the complex mechanical properties of the respiratory system, consisting of various structures with capacitive (compliant), resistive and inertive properties, pressure and flow signals are not in phase (except at given frequencies, called resonant frequencies): recorded in X-Y coordinates, the pressure-flow relationships describe a loop. Therefore, the impedance is characterized by a modulus, $|Z|$, the ratio of pressure *versus* airflow, and a phase, φ, expressing the shift in time of these two variables. Another, more usual, way of dealing with this time shift is to divide the impedance into a real part or resistance (R), and an imaginary part or reactance (X), in which R is $|Z| \cos \varphi$, and X is $|Z| \sin \varphi$. In a simple circuit, consisting of a capacity, a resistance and an inertance, connected in series, R corresponds to the actual resistance of the circuit and X to a combination of capacity and inertance. Generally, in a complex system, R and X vary with oscillatory frequency. Therefore, to define the mechanical properties of the respiratory system, R and X are measured at various frequencies.

In the original device of DUBOIS *et al.* [3], the response of the respiratory system was investigated by applying a succession of oscillations varying in frequency, during voluntary apnoea. This took time, and made the technique very impractical. This is why the FOT did not gain widespread acceptance, until it was modified so that several oscillatory frequencies could be applied simultaneously during spontaneous breathing. This became possible due to advances in microcomputer technology, permitting the application of a composite signal containing several oscillatory frequencies. The signals of pressure and flow are then analysed frequency per frequency, by a mathematical technique called Fourier analysis, in a version adapted to computers: the fast Fourier transform (FFT). This technique allows the frequencies which are not of interest to be filtered out immediately, and thus also the low frequencies of the breathing signal [4, 5]. The remaining problems were the accurate recording of pressure and flow signals in the investigated frequency range (generally 2–32 Hz), and the development of an optimal oscillatory signal, yielding reliable estimates of R and X at each frequency. The latter problems were investigated systematically by an international working group, supported by the Commission of the European Communities (COMAC-BME). The results of this concerted work were published in an issue of the European Respiratory Review [6].

The fundamental requirement for the use of the FFT is that the investigated system is linear. Obviously, this is not the case for the respiratory system: pulmonary and chest wall compliance vary with volume, airway resistance varies with airflow and lung volume. Under these circumstances, however, FFT can still be used if the forced oscillations are sufficiently small with respect to the breathing signal, so that they can be applied over small portions of the pressure-volume and the pressure-flow curves of the respiratory system, portions which are nearly linear, *i.e.* over which compliance and resistance are constant. It turned out that this was possible: oscillations were generated, which were sufficiently large to allow for reliable measurements (satisfactory signal to noise ratio), but not too large to prevent the application of FFT, because of the alinearity of the respiratory system. As a check on the validity of the measurement, a so-called coherence function is generally used. The latter corresponds to a correlation coefficient, and gives an estimate of the amount of extraneous noise and alinearities in the signals. In the generally used set-up, a coherence function of 0.95 (a value of 1 corresponding to total absence of noise and alinearities) is often used as a lower limit for a satisfactory measurement.

In this way, a technique was developed, requiring a minimal cooperation from the subject and yielding values of total respiratory resistance (Rrs), and reactance

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(Xrs), at various frequencies, together with an estimate of the reliability of these values after a relatively short recording time (*e.g.* 16 s). Is this technique also suited for use in intubated, artificially-ventilated subjects, without interfering with the ventilation itself?

The first problem met in the application of FFT is that endotracheal tubes (ETT), even wide ones, have a high impedance, which is markedly less linear than the respiratory system itself. For instance, at a frequency of 2 Hz, the resistance of an ETT (internal diameter 8 mm) doubles when the peak-to-peak amplitude of the signal increases from 10 to 150 $ml \cdot s^{-1}$ [7]. Under these circumstances, an input pressure signal, consisting of several frequencies which are multiples of a fundamental frequency (*e.g.* for a fundamental frequency of 2 Hz, the harmonics 4, 6, 8... Hz), will produce a flow signal which may vary alinearly with the corresponding changes in pressure at each of the investigated frequencies. The FFT will treat the flow changes as if they were linear but containing, next to the flow at the investigated frequency, additional flow signals at higher order harmonic frequencies. The latter "spurious" harmonics will be added to the flows generated by the input pressures with the same higher periodicity. In the presence of such a "cross-talk" between harmonics, the FFT will yield biased estimates of impedance [8]. To avoid this error, single frequency oscillations may be used, thus allowing a correction for the alinearity of ETT [9]. This is the solution to which PESLIN *et al.* [2] resorted. They discovered that the resistance of ETT does not vary with the amplitude of oscillatory flow superimposed on the breathing flow, unless the latter (approximated by a steady flow) is zero. Accordingly, it is sufficient to correct for the influence of breathing flow. The drawback of this approach is double: 1) to measure the frequency characteristics of the respiratory system, the frequencies should be investigated one by one. This is time-consuming, and acceptable only if the mechanical properties of the system do not vary too rapidly with time; and 2) the impedance of the ETT is large: therefore, the correction should be very accurate. In the course of an intubation, mucus deposition will occur, increasing the impedance of the ETT [10]. The latter influence cannot be estimated unless the ETT is removed. These problems are avoided by recording pressure at the distal end of the ETT [7, 11, 12]. This technique has been applied by NAVAJAS *et al.* [13] to anaesthetized paralysed patients during short periods of apnoea. They used an ETT developed for high frequency ventilation, which is provided with a lateral catheter, thus allowing pressure recording at the distal end of the ETT, the pressure transducer being outside the patient. An alternative technique is to use a tip manometer introduced *via* the ETT inside the trachea [14].

Another technical problem in a ventilated patient is that the forced oscillations should be generated and superimposed on the large pressure variations produced by the respirator. This has been solved by PESLIN *et al.* [2] by placing the loudspeaker in parallel with the respirator: the loudspeaker is enclosed in a box and its back and front side are connected to the respirator circuit *via* a piece of tubing. In this way, both sides of the loudspeaker are exposed similarly to the pressure variations produced by the respirator, and the high frequency signals generated by the loudspeaker are transmitted undisturbed to the patient. Another technique, which does not require a modification of the ventilatory circuit, was used by NAVAJAS *et al.* [15]: they connected the loudspeaker to the expiratory side of the circuit, distally from the expiratory one-way valve. In this way, the loudspeaker was not exposed to the large pressure variations inside the respirator circuit, and the oscillations of the loudspeaker travelled easily to the entrance of the respiratory system during the passive expiration of the patient, yielding values of Rrs and Xrs during expiration only.

Although satisfactory solutions are now available for these various technical problems, the relevance of this application of FOT for clinical use is not yet clear. In this respect, the paper of PESLIN *et al.* [2] is important, because it contains the first systematic description of data obtained with this technique in patients ventilated for acute respiratory failure. A first observation made by PESLIN *et al.* [2] is the large variation of Rrs and Xrs during the respiratory cycle. Rrs is markedly flow-dependent: it increases in the course of inspiration and at the beginning of expiration. It is lowest during the pause between inspiration and expiration. Xrs varies less with flow. Similar, but less pronounced, variations of Rrs and Xrs have been observed in patients with chronic obstructive lung disease and upper airway obstruction in the course of spontaneous breathing [16]. Most patients in the study of PESLIN *et al.* [2] demonstrated a negative frequency dependence of Rrs but, in contrast with the findings in spontaneously breathing subjects, in whom Xrs increases with frequency, Xrs regularly did not vary, and even decreased with frequency (between 5–20 Hz) during the inspiratory phase. This unexpected pattern was met more often in subjects with more severe airway obstruction, and could be mimicked by a model containing a shunt in parallel with the airway (or part of the airway), lung and chest wall. This shunt approximated the compliance of the airway walls. The results of this simulation suggest that in some patients peripheral airway resistance is extremely large during expiration, implying dynamic airway compression (flow limitation). The latter is accompanied characteristically by a marked decrease of Xrs. The simultaneous increase of Rrs may be misleading when considered separately, since variations of Rrs may result from (expiratory) flow limitation as well as from flow dependence of Rrs (inspiratory and expiratory). It is interesting to note that the variations of Xrs and Rrs, suggestive of dynamic airway compression, were less marked when a positive end-expiratory pressure (PEEP) was added to the artificial ventilation.

The study of PESLIN *et al.* [2] suggests that the measurements of Rrs and Xrs by means of a FOT lend themselves to an interpretation of the changes of lung mechanics occurring during the ventilatory cycle. To this end, the measurements should be performed at several frequencies, and yield data on Rrs and Xrs at each of these frequencies. This can be done easily with the FOT, in contrast to the interruption methods, which yield a single value of resistance (precisely a value computed from the sudden initial pressure drop and a value computed following pressure equilibration), the meaning of which is difficult to interpret in terms of mechanical properties of the respiratory system.

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