Exploiting respiratory mechanics by forced oscillations: principles and pitfalls

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Submitting a physical system to forced oscillations is a very general approach to investigation of its structure and/or properties. Its application to respiratory mechanics was first proposed by DuBois et al. in 1956 [1], but it has also been used in other fields of respiratory physiology, e.g., to investigate the control of breathing and the response to exercise. As used in respiratory mechanics, the method consists of applying sinusoidal pressure variations to the respiratory system (or one of its parts) with an external generator, and in studying the relationship between the pressure applied and the resulting respiratory flow. The pressure-flow relationship at a given frequency is termed “impedance” (Z) and may be expressed by the amplitude ratio of the variables (modulus of impedance (Z)) and by their phase angle (°), or by two related parameters: the effective resistance or real part of the impedance (Re = (Z)·cos °) and the reactance or imaginary part of the impedance (Im = (Z)·sin °).

Although the method may already be helpful when used at a single frequency, it is especially fruitful to make measurements at a number of different frequencies, i.e., to obtain the frequency response of the system. Then, using an appropriate model, several properties of the system may be computed from the impedance data. The extent to which a specific property influences the impedance depends very much upon the frequency. By properly choosing the frequency range, it is therefore possible to study selectively different aspects of respiratory mechanics. The method is increasingly used since digital computers are available for data processing. Computers also made it possible to explore many frequencies simultaneously by using non-sinusoidal inputs with a large frequency content. Then, the analysis in terms of elementary sine-waves is made using Fast Fourier Transforms [2].

When oscillating the respiratory system at frequencies above a few Hz, substantial differences are seen between instantaneous flow at the mouth and at the chest wall in relation to alveolar gas compression. It follows that several types of respiratory impedance may be obtained according to where pressure oscillations are applied and which flow is considered [3]. The most commonly measured is termed "input impedance" (Zin) and is obtained by varying the pressure at the airway opening and measuring flow at the same place. An alternative approach is to measure the so-called respiratory transfer impedance (Ztr), either by applying pressure variations at the chest and measuring flow at the airway opening, or by applying pressure variations at the airway opening and measuring flow at the chest.

We will focus on input impedance. A major problem with this approach is the shunt impedance constituted by upper airway walls, in particular the cheeks, which are mechanically in parallel with the proper respiratory system. It may be responsible for a large underestimation of respiratory impedance, especially at high frequencies. Several methods have been proposed to reduce or suppress this artefact, among them the application of the pressure input around the head, so as to minimize transmural pressures across the upper airways [4]. The input effective resistance (Re(Zin)) varies little with increasing frequency in normal subjects. It is larger and becomes frequency dependent in patients with airway obstruction. Negative frequency dependence of Re(Zin) is commonly taken to reflect non-homogeneous behaviour of the respiratory system, due to differences between lung regions or between airway wall and peripheral lung. Input reactance is often analysed in terms of respiratory compliance and inerance. It is decreased in chronic obstructive pulmonary disease (COPD) patients, a finding also attributed to mechanical non-homogeneity. Both of these features of impedance curves in COPD patients may be exaggerated by the above-mentioned upper airway artefact.

Provided that this artefact is corrected for, what kind of abnormalities may total respiratory input impedance reveal? Computer simulations show that in the frequency range commonly explored (4–30 Hz), Zin should be sensitive to both central and peripheral airway obstruction and, to some extent, should permit separation of these two abnormalities. Indeed, central obstruction should uniformly increase Re(Zin) without changing Im(Zin), while peripheral obstruction should only increase Re(Zin) at low frequencies, so inducing a negative frequency dependence of Re(Zin), and decrease of Im(Zin). Zin should also be sensitive to mechanical non-homogeneity, which it may be difficult in some instances to distinguish from peripheral airway obstruction. On the other hand, Zin is not expected to be very sensitive to lung and chest wall compliance at frequencies above a few Hz. Its sensitivity to the different types of airway...
obstruction and the fact that the measurements do not require active co-operation from the subject and may be frequently repeated make the method particularly suitable for studying airway response to bronchodilator or bronchoconstrictor agents.

References


Clinical applications and modelling of forced oscillation mechanics of the respiratory system

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Over the past few years we have conducted a number of studies to assess the usefulness of the pseudorandom noise forced oscillation technique (FOT) [1], described by LÀNDSÈR et al. [2], and NAGELS et al. [3], for clinical practice. This technique consists of applying the complex signal at the mouth and recording of the resulting flow and pressure signal also at the mouth (input impedance). Changes in total respiratory resistance (Rrs) and reactance (Xrs), measured between 6–26 Hz, were investigated in the following disorders or conditions: a) upper airway obstruction [4]; b) asthma, chronic bronchitis and emphysema; c) bronchial challenge tests [5]; d) diffuse interstitial lung disease [6]; e) chest wall deformities (kyphoscoliosis and ankylosing spondylitis); and f) strapping of the thoracic cage [7], which can be considered as a physiological tool to mimic chest wall disorders.

Irrespective of the underlying disorder, changes in input impedance always presented the same characteristics, which consisted of an increase in Rrs, together with a decrease in Rrs with frequency and a decrease in Xrs with an increase of the resonant frequency. Differences in impedance data among the various obstructive and restrictive disorders were quantitative, not qualitative.

In airflow obstruction, i.e. upper airway obstruction, asthma, chronic bronchitis and emphysema, Rrs (and Xrs) were closely correlated with airway resistance (Raw). The changes in Rrs and Xrs were determined mainly by the degree of increase in Raw, whereas the site of this increase was of minor influence on the impedance curves. A model study indicated that this fact is connected with the important influence of the shunt properties of the upper airway wall on the values of Rrs and Xrs in obstructive patients. Concerning practice, in upper airway obstruction impedance is neither very sensitive nor able to assess the dynamic behaviour of the obstruction; in asthma, chronic bronchitis and emphysema input impedance can be used as an alternative for plethysmographic Raw and gives complementary information with respect to forced inspiratory volume in one second (FEV1); for bronchial challenge tests in patients with normal baseline resistance FOT is more sensitive than the measurement of specific airway conductance (sGaw).

In the various restrictive disorders of the respiratory system the changes in Rrs and Xrs were proportional to the reduction in vital capacity (VC) and total lung capacity (TLC). For a similar reduction in TLC changes in Rrs and Xrs were more pronounced in kyphoscoliosis than in diffuse interstitial lung disease. In the former disease there is generally a substantial difference between Raw and Rrs. In diffuse interstitial lung disease a model study suggested in addition to the measured decrease in lung compliance and increase in lung tissue resistance, an increase in peripheral airway resistance and a decrease in lung compliance. In kyphoscoliosis and in ankylosing spondylitis a model simulation indicated that the changes in Rrs and Xrs were mainly attributable to an increase in chest resistance and to a decrease in chest wall compliance, while in kyphoscleriosis changes in airway and lung mechanics were also produced. With respect to clinical practice we concluded that FOT can detect changes in chest wall mechanics, but the method lacks sensitivity. Finally the study on strapping of the thoracic cage demonstrated that partitioning of Rrs and Xrs into lung and chest wall components may be a more promising tool for the assessment of chest wall disorders.

References