The acoustic reflection technique for non-invasive assessment of upper airway area

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ABSTRACT: Non-invasive assessment of upper airway area by acoustic reflections has been developed in the last 12 yrs. The technique is based on the analysis of sound waves reflected from the airways. Measurement of the amplitudes of the reflections and their times of arrival at the sensing microphone permits construction of a plot of airway area vs distance from the microphone. We describe the theoretical foundations of the method and review the underlying assumptions. This is followed by a summary of the results of in vitro and in vivo validation studies, with particular attention to the reproducibility, accuracy and variability of the technique. The description of clinical and physiological applications of this technique includes detection of tracheal stenosis, demonstration of structural and functional abnormalities of the pharynx and glottis in patients with sleep apnoea, dichotomous response of the airway area to exercise, and airway vs parenchymal hysteresis. Finally, we comment on the future directions that might be investigated using this technique.


Determination of mechanical properties of the lung from its response to pressure oscillations began with the pioneering experiments of DuBois et al. in 1956 [1]. They employed sinusoidal oscillations at the airway opening to measure the impedance of the respiratory system up to 15 Hz. At a frequency of 15 Hz in air, the corresponding acoustic wavelength is approximately 24 m. Such waves, being 40 to 1,000 times longer than respiratory structures of interest, effectively preclude our ability to learn anything about the fine structure of the respiratory system. These measurements essentially reduce the lung to a point, with all the properties of central airways, peripheral airways, pulmonary parenchyma and chest wall lumped into effective system compliances, resistances, and inertances.

If we desire to learn about the properties of the respiratory system on a finer scale, such as by looking at 1 cm segments of the airway starting at the mouth and proceeding toward the chest wall, we need to employ pressure waves with wavelength comparable to the linear dimensions of the object we are studying. To do so requires pressure waves with frequencies extending to several kHz. In doing so, the respiratory system can no longer be modelled in terms of "lumped" parameters, but must be considered in terms of "distributed" properties along the length of the system.

One such "distributed" property of the respiratory system is the total cross-sectional area of the airways. The aim of the acoustic reflection technique is to measure this area at any point within the respiratory system. We review the basic principles of the acoustic technique, its underlying assumptions, sensitivity and resolution, validation and applicability.

Theoretical background of the acoustic method

The underlying principle of the method is quite simple, although in practice the technique is based on strict assumptions and complex computational algorithms. Let us first consider the basic principle of the acoustic method. When a sound pulse travels along a tube and encounters an area change from A₁ to A₂, part of the pulse is reflected, and part is transmitted. The reflected pulse travels back along the tube. If the wavespeed (c) is known, from the travel time (t) the length of the tube (d) is easily calculated to be d = ct. In the case of one-dimensional wave propagation, the measurement of travel time of the wave is entirely equivalent to the measurement of distance.

The amplitude of the reflected pulse (Pr), on the other hand, is determined by the amplitude of the incident pulse (Pi), and the physical property of the tube. Consider a tube with a single discrete area change from A₁ to A₂. Assuming constant and uniform gas composition, the amplitude of the reflected pulse is given by:

Pr = Po[(A₁-A₂)/(A₁+A₂)]
By measuring the amplitude of incident and reflected pulses, the cross-sectional area $A$ of the duct is easily calculated, since $A$ is presumed known. Consequently, the problem of determining the length and area of the straight tube reduces to measuring the travel time of the pressure pulses reflected from the area change of the tube, and the amplitudes of the incident and reflected waves. If instead of a single straight tube we have a duct consisting of many segments, each having different area, the incident sound wave (pressure wave) will be reflected in part every time it encounters a new segment. The measurement of the arrival times and amplitudes of the reflections permits determination of the lengths and areas of the individual segments. This gives the area of the duct vs distance from the inlet - the area-distance function, or airway echogram.

Although conceptually simple, the actual calculation of area as a function of travel time is computationally quite intensive due to occurrence of multiple reflections and multiple arrival times. Fortunately, the problem is a general one, with applications in many situations involving propagation of waves through stratified media. This general problem has been studied by WARE and AXI [2], who provided a computational algorithm for recovering the characteristic impedance of a medium as a function of travel time from the response of the medium to an impulse. This algorithm serves as a backbone of the programme for calculating areas as a function of distance.

The theoretical foundations and limitations of this method with particular regard to subglottic airways have been analysed by SIDELL and FREDBERG [3]. They identified and discussed several idealizations and assumptions of the method, which we will briefly review together with the recent experimental evidence regarding the importance of these assumptions in inferring airway areas. A full discussion of these assumptions has been given by FREDBERG [4].

**Assumptions and approximations**

**Branching**

The algorithm for computing area does not recognize branching per se. In the case of symmetric branching, in which all parallel pathways have identical properties, the area obtained by the algorithm will be equal to the sum of the areas across all branches. However, in the case of asymmetric branching, or different gas composition, we would have no fundamental grounds upon which to expect the acoustic area inference to closely approximate the anatomical areas. Nevertheless, JACKSON et al. [5] used this technique to measure cross-sectional area of airway casts and found that the acoustic area compared favourably with the anatomical tracheal area as well as with the summed cross-sectional area of the main stem bronchi for up to 6 cm past the carina.

Another situation where branching influences the acoustic area inference is in the case of open nasopharyngeal velum. In this case we have two parallel pathways: mouth to subglottic airways, and mouth to nasal airways. Since the acoustic energy is propagated along the latter pathway, the algorithm interprets this as increase in the area of subglottic airway, and the area inference becomes meaningless. Fortunately, this situation is easily recognized in practice [6] and measures may be taken to prevent opening of the velum (e.g. removing the noseclips or instructing the subject to breathe through the mouth).

**Uniform gas composition**

To calculate the areas and travel times we need to know the speed of sound waves through the respiratory system. This wavespeed depends, among other things, on gas composition. During measurements, gas composition may vary due to accumulation of carbon dioxide in the expired gas. This problem has been examined by D'Urzo et al. [7] who measured cross-sectional area of a branching system of tubes stimulating pharynx, larynx and trachea) containing gas mixtures with varying concentrations of CO$_2$ (from 0 to 10%). These authors found that at 10% CO$_2$ the areas of the pharynx, larynx, and trachea were overestimated by 5–6%, and the distance was overestimated by 1–2.5 cm, depending on whether it was located in the pharynx or in the trachea. They concluded that under normal measurement conditions, with CO$_2$ concentrations of about 5%, the error introduced in the measurement of area is negligible, but the distance is systematically shifted chestward; however, the magnitude of these shifts is well within the inherent variability of this technique for repeated measurements.

**One dimensional wave propagation**

This assumption is fundamental to the acoustic method of area inference. It ultimately determines the spatial resolution of area measurements. This assumption also dictates many of the technical details of the method, such as frequency range (bandwidth), gas composition, the type of mouthpiece, etc.

Interpretation of the acoustic data using the algorithm developed by WARE and AXI [2] assumes that the pressure waves are one-dimensional plane waves, described by one-dimensional equation of motion. If the phase or amplitude of pressure waves vary across the tube radius, then the mathematical formalism used to compute acoustic impedance as a function of travel time is no longer applicable.

The major limitation in this connection is on the upper limit of frequency (bandwidth) of the pressure waves which are allowed to propagate through the respiratory tract. In long rigid straight tubes one-dimensional wave propagation is assured as long as the frequency of the waves is less than $c/2d$, where $c$ is the wavespeed, and $d$ is the diameter of the duct. In the human respiratory system the greatest diameter occurs between the cheeks (about 5 cm), hence the maximum frequency is $35440/10 = 3.5$ kHz. By using a special mouthpiece which fits
between the cheeks leaving only a lumen of about 2.5 cm diameter, the bandwidth may be increased to 7 kHz. This is why many of the original measurements of airway area [8-13] employed an unusual rubber mouthpiece. Recently, Rubinstein et al. [6] showed that a regular scuba mouthpiece may suffice, provided that the care is taken to instruct the subject to keep the cheeks tightly closed, perhaps by applying light pressure on the cheeks with the hands.

However, the walls of the respiratory system (pharynx, trachea) are not rigid. Consequently, the energy of the incident pressure waves may be reflected not only because of the axial variations in the cross-sectional area of the duct, but also because of axial variations in wall compliance.

Airway wall impedance has been extensively measured at relatively low frequencies (<100 Hz), where the static compliance of the airway walls is the major contributing factor; at higher frequencies (100-1,000 Hz) airway wall resistance is important in reducing the dynamic compliance [4, 8, 14, 15]. At still higher frequencies (>1,000 Hz) airway wall inertia limits the motion of the airway walls, causing the dynamic compliance to fall at a rate inversely proportional to the square of the frequency.

The effect of airway wall non-rigidity on the accuracy of area determinations has been examined by Brooks et al. [16] in excised canine tracheae. They found that when the mass of the airways is increased by surrounding the walls with petroleum jelly, the accuracy of the acoustic measurements was significantly enhanced.

When the energy of the incident wave is reflected by axial variations in wall impedance, the algorithm would falsely interpret this change as a variation in duct area. As a result of such errors it becomes necessary to extend the bandwidth to higher frequencies, where airway walls are dynamically more rigid by virtue of their inertia. But this requirement conflicts with that of bandwidth limitation. The bandwidth cannot be extended to higher frequencies, causing the dynamic compliance to fall at a rate inversely proportional to the square of the frequency.

The equipment

The experimental apparatus for measurement of airway areas by acoustic reflections has been described by Jackson and co-workers [17] and Fredberg et al. [8]. The two systems, although based on the same principle, are somewhat different in terms of the equipment used to generate the incident sound wave and in terms of signal processing. The system described by Jackson and co-workers [17] has been used to measure airway cross-sectional areas in animals (in vivo and in vitro) and in airway casts. The system of Fredberg et al. [8] has mainly been used to study human airways. In this section we describe in general the components used in the system for measuring airway areas and provide the rationale for their use. Details of the actual apparatus may be found in the original publications [8, 17].

Figure 1 shows a simplified diagram of the equipment. Probably the most important component of the acoustic system is the sound source. It generates the incident pressure waves which travel toward the subject, get reflected, and are used to compute airway areas. Two types of sound sources have been used. Jackson and co-workers [17] employed a spark source. In this system a high voltage signal is discharged across a spark gap; the heat produced during such discharge generates a pressure disturbance which is the incident sound wave. Fredberg et al. [8] used a loudspeaker (horn driver) which generates an audible acoustic impulse serving as the incident wave. The latter system provides more reproducible signal, which permits some simplification in the measurement technique and the calibration procedure.

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**Fig. 1.** Diagram of experimental equipment. Pm: mouth pressure; atm: atmosphere.
The incident signal is sensed by a microphone located near the subject. The same microphone also senses the reflected waves originating within the respiratory tract. The microphone must be capable of good signal-to-noise ratio and have rapid response time. However, since the same transducer senses both incident and reflected waves, it need not have flat frequency response.

The only other electronics necessary to operate the system includes a power amplifier which provides the voltage signal necessary to drive the loudspeaker, and an amplifier-filter which amplifies and filters the intensities of the signals sensed by the microphone.

A computer equipped with the digital-to-analogue (D/A) and analogue-to-digital (A/D) converters controls the time sequence of the measurements starting with the activation of the loudspeaker, waiting until the incident wave enters the respiratory system, and starting data acquisition to measure incident and reflected signals. The output signal of the microphone is sampled by the A/D converter and stored in the computer memory as a sequence of numbers representing the pulse response of the respiratory system. This sequence is then processed by the customized software (employing the Ware-Aki algorithm and real time domain deconvolution techniques) to obtain airway area as function of distance from the microphone.

During data acquisition the wave recorded by the microphone represents a superposition of the incident and reflected waves. This occurs because the incident pulse is of duration such that by the time it travels completely past the microphone, the first reflections already begin to arrive. In order to obtain the pulse response of the airways, the reflected and incident waves must first be separately identified and then deconvoluted. To isolate the reflected wave, a special calibration procedure is employed. This procedure is based on measuring the pressure signal present when the wavetube is extended by a long straight tube of constant cross-sectional area. Because there are no reflections arising from a constant area tube, the waveform recorded by the microphone is simply the incident wave, which is subsequently subtracted from the superposed incident and reflected waves obtained during the actual data acquisition to yield the reflected wave alone. Then the incident and reflected waves are deconvoluted (using real time domain deconvolution techniques) to obtain the pulse response of the airway system. The temporal separation between the incident and the reflected waves depends on the distance between the microphone and the mouth. This distance varies between different publications, but as long as the proper calibration procedure is used to obtain the reflected wave, the accuracy of the area determinations remains unaffected.

The loudspeaker and the microphone are mounted on a long (200 cm) tube (the wavetube) which serves to conduct the incident wave toward the subject (fig. 1). The loudspeaker may be mounted either in the middle or at the end of the tube, the length of which is determined by travel time considerations. We found it convenient to mount the loudspeaker in the middle of the wavetube to allow a spirometer to be mounted at its distal end.

For the reasons described above it is important to keep the wavetube filled with 80% He - 20% O2 mixture, which is accomplished by providing a slow bias flow through the wavetube.

One end of the wavetube (the subject end) contains a valve which permits the subject to breathe either through the wavetube (during data acquisition) or from a reservoir bag containing the 80% He - 20% O2 mixture (during equilibration prior to data acquisition). The valve has ports for fitting the microphone and other auxiliary equipment such as pressure transducers and gas analysers. Distal to the valve there is a small (4-5 cm) section of the wavetube which serves for mounting the mouthpiece.

The system is capable of performing 64 measurements in rapid succession (every 200 ms) while the subject is breathing through the wavetube. Any breathing pattern may be employed, although all of the studies to date were performed either during quiet tidal breathing or during slow vital capacity inspiration and expiration. An attractive possibility of measuring upper airway area during forced expiration (or inspiration) has not yet been investigated.

An example of the area-distance plot, or airway echogram, obtained during quiet tidal breathing is shown in figure 2. Some major anatomical landmarks may be identified. The first few centimeters (about 6 cm) correspond to the end of the wavetube and the mouthpiece, the large area flare is the pharynx, the local minimum in area immediately distal to the pharynx is the glottis, and a relatively stable plateau region distal to it is the trachea. We note that the area of the upper airway (pharynx, glottis and trachea) is quite reproducible, as evidenced by small standard deviations. We also note that the reproducibility of the measurements decreases for the distal structures, where the accumulated numerical error becomes larger. The accuracy of the acoustic measurements (i.e. anatomical vs acoustical equivalence) is expected to be lower in the distal structures because the assumptions of the method are satisfied only for the central airways. In particular, the assumptions regarding symmetrical branching, lossless gas, etc. [4] are not valid for the distal airways.

Validation of the acoustic method

In vitro and animal measurements

The earliest measurements of area using the acoustic method were performed by Jackson and co-workers [5, 17, 18]. All of the measurements were carried out using air rather than the helium-oxygen mixture. The measured structures included either the excised dried lungs, or airway casts, or biological preparations in situ. These measurements were generally restricted to structures with rigid walls. Although these measurements cannot be considered validisations of the acoustic technique, they established the general applicability of the method...
to the measurement of airway areas. In their initial study [17], Jackson and co-workers measured the acoustic and radiographic airway areas of dried excised airways in the dog; the airways ranged in diameter from 1–11 mm. Radiographic areas were obtained from the bronchograms assuming circular geometry. These authors found a strong correlation between the acoustic and radiographic areas (correlation coefficient ranging from 0.81–0.96, depending on the measurement conditions); furthermore, the acoustic and radiographic measurements were clustered around the Line of identity, with the absolute acoustic areas somewhat (9–15%) lower than the radiographic ones. In addition, Jackson and co-workers [19] used radiographic bronchograms and the acoustic technique to measure airway areas in dogs in-vivo before and after vagal stimulation and histamine. They were able to determine that even following the above interventions, acoustic areas remained in close agreement with the radiographic ones, with the ratio of the acoustic area to the radiographic one being 0.95. They also made a very important observation regarding the more distal airways, where the acoustic areas are not expected to track the radiographic ones: the changes in anatomical area due to bronchoconstricting agents are generally reproduced by the changes in the acoustic areas.

Jackson and Krevans [18] also validated the technique in vivo in tracheostomized dogs. They found that the ratio of the acoustic area to the radiographic area ranged from 0.91–1.03 depending on the assumed airway geometry for radiographic area measurements.

**In vivo human measurements**

There are several studies comparing the acoustic and radiographic areas in humans in vivo. In all of these measurements the same apparatus and the experimental protocol as described by Fredberg et al. [8] and Brooks et al. [16] was used. All of the measurements were performed using 20% O₂ - 80% He mixture. The incident pulse was generated by a loudspeaker, and a customized mouthpiece was employed. With this system an echogram shown on figure 2 is produced. Different anatomical regions are indicated. We employ a convention whereby the “pharynx” is defined as that portion of the airway contained between the end of the mouthpiece and the glottis: it includes anatomical oropharynx and laryngopharynx. The “glottis” is defined as a 3–5 cm segment of the airway centred on the minimum in airway area immediately distal to the pharynx; anatomically, this segment includes the laryngeal vestibule, false and true vocal cords, and the immediate subglottic region. The “tracheal” segment is defined as beginning immediately distal to the glottic minimum and extending 18–20 cm distally; anatomically, this segment corresponds to the extrathoracic and intrathoracic trachea.

The most extensive validation in human subjects has been done for the trachea. The first study which assessed the value of this technique in measuring tracheal areas was that of Fredberg et al. [8], who compared the acoustic areas with those determined from the anteroposterior and lateral radiographs of the airway in six subjects. Figures 3 and 4 reproduced from that paper show a good correlation between the radiographic and acoustic tracheal areas; for the mid-trachea, the average coefficient of variation of area determinations for five trials in each subject was 16%. These authors also found that if the area measurements were performed breathing room air, rather than He-O₂, gas mixture, acoustic areas exceeded the radiographic areas, presumably due to the violation of the assumption of dynamic rigidity of the airway walls. They concluded that the use of 80% He - 20% O₂ mixture is required for accurate measurements of area in humans in vivo. If the measurements are performed in rigid structures such as airway casts or excised dried airways, lower bandwidth may be used.
and the measurements may be carried out in air, as demonstrated by Jackson and co-workers [5, 17-19]; nevertheless, some of their measurements performed in "wet" lungs using air agreed well with the radiographic data. The issue of whether air or helium-oxygen gas mixture should be used for acoustic area measurements still remains unresolved.

In a subsequent study by Brooks et al. [16], the authors compared tracheal areas at 6-10 cm distal to the glottis measured using the acoustic and the radiographic technique in ten healthy volunteers. Once again, they confirmed close correspondence between the acoustic and the radiographic areas, with the acoustic/radiographic area ratio of 1.06±0.13.

D'Urzo et al. [11] employed computerized tomographic (CT) scans to compare tracheal areas in seven patients with the history of upper airway abnormalities. The comparison was carried out at 1 cm distances starting at the glottis for as long as 13 cm in some patients. Very good correlation was found between the acoustic and the radiographic areas (correlation coefficient r=0.92); the absolute areas were also in good agreement, with the acoustic/radiographic area ratio of 0.96.

Acoustic measurements of glottic areas have not been as extensively validated as for the trachea. Only one study, by D'Urzo et al. [20] addressed this question. Using similar techniques as for the validation of tracheal areas [11], D'Urzo et al. [20] obtained CT scans of the larynx and airway echograms in 11 patients with laryngeal pathology. The results indicate a very close correspondence between the two methods: mean (±SD) glottic areas were 1.8±0.8 cm² for the acoustic method and 1.7±0.9 cm² for the CT method. Furthermore, there was a good correlation between the two methods with the correlation coefficient of 0.95. An important finding of that study was the demonstration that acoustic technique was capable of measuring glottic areas as low as 0.4 cm². These results suggest that the acoustic technique may be useful for measuring glottic areas, possibly not only during quiet tidal breathing, but during various other respiratory manoeuvres to assess the dynamic movements of the vocal cords.

There are no data dealing with the validation of pharyngeal area in humans in vivo. However, there is some indirect evidence indicating that the acoustic technique probably describes the pharynx fairly accurately. Most of the indirect evidence comes from the radiographic data. Pharynx has been extensively studied in patients with obstructive sleep apnoea (OSA) using a variety of techniques, including CT scans and the acoustic reflection technique [9, 21, 22]. The areas obtained using both techniques are of similar magnitude. Furthermore, the differences in pharyngeal areas
in patients with various sleep-related breathing disorders and in normal subjects, measured using CT scans, are reproduced by the acoustic technique. Other indirect evidence attesting to the fact that pharyngeal areas are probably well reproduced by the acoustic technique comes from the considerations involving the accuracy of the acoustic method. The nature of the algorithm for calculating acoustic areas is such that errors accumulate with distance. Any errors in the proximal areas will be only compounded when the areas of the more distal airways are computed; since the glottic and tracheal areas are faithfully reproduced by the acoustic technique, this implies that the pharyngeal areas must also be accurate.

It is clear that more validations are necessary, particularly for the pharynx and glottis. In addition, the use of newer techniques such as ultrafast CT scans and magnetic resonance imaging could improve the accuracy of the comparison procedures and extend it past the carina. Another important issue which needs to be addressed before the acoustic technique can be applied to clinical practice, is the reproducibility and variability of the method. This problem has been extensively studied by Brooks et al. [16]. By performing repeated measurements in the same subjects on different days and using different mouthpieces, they were able to determine that the coefficient of variation for measurements performed on the same day was 10±4%, day-to-day variability was 9±4%, and mouth-piece-to-mouthpiece variability was 7±6%. These values are similar to the variabilities encountered in conventional measurements of pulmonary function.

Table 1 shows the summary of all studies comparing the acoustic and anatomical areas. Although these studies were performed in three different laboratories, there is a remarkable consistency of the data, indicating that the ratio of acoustic to radiographic area in humans in vivo ranges from 0.96–1.06.

**Applications**

Being relatively new, the acoustic reflection technique has so far been applied to only a few physiological and clinical problems. Some, but not all of them had been studied previously using indirect techniques to assess tracheal area. In this section we mention, rather than

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* In these studies although no acoustic/radiographic area ratio is given, the entire acoustic area-distance plot is compared to the radiographic one. PA: postero anterior; CT: computerized tomographic.
discuss in detail, some of the physiological and clinical problems that have been studied using the acoustic reflection technique. This discussion will be organized "anatomically", starting with the pharynx and ending with the trachea. We point out how the acoustic technique contributed to our understanding of the properties of these anatomical structures in health and disease.

Pharynx

Pharyngeal structure and function has been studied very extensively by the acoustic technique, almost exclusively in patients with sleep apnoea. There are several areas where the application of this technique provided new insights into the pathogenesis of this disorder.

Rivlin et al. [9] were the first to apply the acoustic technique to measurements of pharyngeal area in patients with obstructive sleep apnoea (OSA). They showed that patients with OSA had smaller pharyngeal areas than the non-apnoeic controls, even when measured in the awake state. This was not an entirely new finding, but it confirmed similar observations obtained previously using CT scans [21, 22]. It is now considered to be an almost established fact that patients with OSA have abnormal pharyngeal structure.

A novel finding obtained using the acoustic method was the observation that these patients not only have abnormal pharyngeal structure, but function as well. Hofferstein et al. [23] found that patients with OSA exhibit greater changes in pharyngeal area with lung volume and applied pressure [12], than the non-apnoeic controls, indicating that these patients have a more "floppy" pharynx than controls, even in the awake state. Fouke et al. [24] and Brown and co-workers [13] showed that pharyngeal area also depends on posture, being lower in the supine, than in the upright position. These postural changes are greater in patients with OSA [13] than in the non-apnoeic controls, thus predisposing the former group to developing complete pharyngeal occlusion during sleep.

Perhaps the most important contribution of this technique to our understanding of pharyngeal structure and function in patients with sleep disordered breathing was the recognition that not only sleep apnoea, but snoring alone (which is considered to be a precursor to sleep apnoea) may be associated with subtle pharyngeal abnormalities. Bradley et al. [25] found that non-apnoeic snorers have similar pharyngeal area at functional residual capacity (FRC) and higher pharyngeal area at residual volume (RV) than the apnoeic snorers. A related observation was that obesity (which is frequently present in apnoeic and non-apnoeic snorers) also influences pharyngeal properties. Rubinstein et al. [26] examined pharyngeal mechanics in apnoeic and non-apnoeic snorers before and after weight loss. He found that although pharyngeal areas were not significantly different before and after 26 kg weight loss, lung volume-related changes in pharyngeal area (i.e. the difference in pharyngeal area between FRC and RV) were significantly smaller after weight loss, which may imply an increase in pharyngeal stability. Some of the observations regarding pharyngeal structure and function obtained using this technique are still controversial, and are presently being verified either by independent studies employing different methods, or by the studies using the acoustic technique, but employing much larger sample populations.

Glottis

This anatomical region has not been as extensively studied with the acoustic method as the pharynx. There are only two studies dealing with the glottis. In one study Rubinstein and co-workers [27] examined glottic movements during breathing in normal subjects and found that the conventionally accepted inspiratory descent of the glottis accompanied by increase in glottic cross-sectional area is not present in all individuals. Some subjects exhibit the opposite, i.e. descent of the glottis during expiration and little change in cross-sectional area.

The second study was prompted by the observation of Rivlin et al. [9] that glottic area is reduced in patients with sleep apnoea. Rubinstein and co-workers [28] hypothesized that not only the structure, but also the function of the glottis may be abnormal in patients with OSA. The authors studied patients with severe OSA and seemingly normal pharyngeal function, and found that these patients have paradoxical inspiratory narrowing of the glottis, which, if also present during sleep, might predispose these patients to development of complete airway occlusion at the level of the larynx.

Trachea

Tracheal properties have been a subject of many investigations, most of which have examined tracheal mechanics, i.e. changes in tracheal area with lung volume, applied pressure, and exercise [29–31]. One of the results of these studies was constructing the pressure-area curves of the trachea (which could subsequently be used in modelling of flow limitation), and demonstration that trachea exhibits hysteresis [32]. This latter property is related to the tracheal tone; recent study of Katz et al. [33] showed that tracheal hysteresis is non-uniform and varies along the length of trachea, implying that the tracheal tone itself may be non-uniform throughout the tracheobronchial tree. This unexpected finding was in agreement with the previous finding that the airways exhibit a dichotomous response to exercise [31]. Following exercise, the trachea dilates, while the peripheral airways constrict.

The acoustic technique also contributed to the study of the relationship between the tracheal size and lung size, confirming the concept of dysanapsis proposed by
MEAD [34]. It was found that the lung and the airways grow unequally and differently in men and women [29, 35].

The only clinical application of the acoustic method so far has been a study of six patients with tracheal stenosis [10]. In these patients the acoustic technique correctly determined the location of stenosis, the extent of the stenotic segment, and the minimum cross-sectional area of the stenotic segment. This technique appears to be more sensitive than the conventionally used flow-volume loop; in five out of six patients with tracheal stenosis proven by the acoustic measurements and confirmed by direct fiberoptic endoscopy, the flow-volume loops were normal.

Conclusions and future directions

It is clear from the already accumulated evidence that the acoustic technique may become a valuable tool for studying the clinical and physiological properties of the upper airway. So far this technique is the only one which allows non-invasive, accurate, reproducible and inexpensive measurements of the upper airway area. This technique is suitable not only for the measurements of static properties, but by virtue of its ability to measure airway areas up to 5 times the stenosis, it is capable of providing the dynamic characteristics of the upper airway. This technique can be employed in adults as well as children, and requires minimum patient co-operation.

The major problem with the technique so far, which prevents its wider application, is that it is not yet commercially available; it still exists only as a research tool available to a few investigators. This situation will hopefully be remedied in the near future, when the apparatus for the acoustic measurements will become available in the same easy manner as any other pulmonary function testing equipment.

The future directions and possibilities that can be explored using this technique (once it is widely accepted and validated) are numerous. Possible directions will include further studies of airway physiology such as changes in upper airway structure and function with provocation challenges (exercise, cold air, pharmacological manipulations). Another possibility, as yet unexplored, is measurement of airway area during maximum expiratory flow. The acoustic technique may be used in conjunction with other measurements to explore the individual influences of the factors that determine airway area, such as neural, mechanical, and hormonal influences. In addition, combining this technique (which is best for central airways) with a technique which assesses both central and peripheral airways, such as for example the forced oscillations technique, may enable us to subtract the effect of the central airways and to obtain a better test of peripheral airways.

Clinical applications might include pre-intubation assessment of patients with expected difficult intubation, such as patients with known upper airway or neck pathology, or study of post-intubation complications, such as stenosis or strictures. If the apparatus can be simplified to be used during sleep, this technique will become useful in assessing upper airway collapse in patients with sleep apnoea.

A new and exciting possibility is presently being pioneered by ILBERG et al. [36], acoustic rhinometry. Making measurements of upper airway area via the nose, rather than the mouth, will obviate many of the technical problems such as mouthpiece construction, air leaks etc., and will allow us to assess the entire airway - from the nasopharynx to the subcarinal airways.

It is clear that having a tool for a non-invasive direct look into the airway will ultimately prove to be beneficial in extending our understanding of the structure and function of the upper airway in health and disease.

References