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Standardization of computerized respiratory sound analysis

A.R.A. Sovijärvi, J. Vanderschoot, J.E. Earis

Pulmonary disease is a major cause of ill-health throughout the world. In Europe, chronic obstructive pulmonary disease (COPD) and asthma have been estimated to affect between 10 and 25% of the adult population. Pulmonary infections such as acute bronchitis and pneumonia are common, and interstitial lung disease is increasing in incidence. The diagnosis of these common chest diseases is facilitated by pulmonary auscultation using a stethoscope. This device, invented in 1821 by the French Physician, Laennec, is still the commonest diagnostic tool used by doctors.

Auscultation with a stethoscope has many limitations. It is a subjective process that depends on the individual's own hearing, experience and ability to differentiate between different sound patterns. It is not easy to produce quantitative measurements or make a permanent record of an examination in documentary form. Long-term monitoring or correlation of respiratory sound with other physiological signals is also difficult. Moreover, the stethoscope has a frequency response that attenuates frequency components of the lung sound signal above about 120 Hz [1], and the human ear is not very sensitive to the lower frequency band that remains.

Over the last 30 yrs, computerized methods for the recording and analysis of respiratory sounds have overcome many limitations of simple auscultation. Respiratory acoustic analysis can now quantify changes in lung sounds, make permanent records of the measurements made and produce graphical representations that help with the diagnosis and management of patients suffering from chest diseases.

Over recent years, the scientific activity within the field of respiratory acoustics has increased markedly. However, a lack of guidelines for data acquisition, storage, signal processing and analysis of the lung sound signal has made it difficult to compare results from different laboratories and has hampered the commercial development of respiratory sound analysis equipment. Several efforts have been undertaken to solve these problems [2–4].

The European Community has financed a BIOMED 1 Concerted Action project entitled Computerized Respiratory Sound Analysis (CORSA). This collaboration, which was also a task force of the European Respiratory Society, involved research workers in seven European Countries (Belgium, Britain, Finland, France, Germany Italy and the Netherlands). The main objective of the participating centres was to develop guidelines for research and clinical practice in the field of respiratory sound analysis. This issue of The European Respiratory Review includes the agreed consensus of the CORSA project group. As an introduction, a survey of current clinical practice and research initiatives in Europe is presented. Because the definitions of terms including lung sound nomenclature used in the field are variable both within and between countries, a paper presenting both established and new definitions of medical and technical terms used in pulmonary acoustics is included. This paper is deliberately comprehensive in order to provide easily accessible definitions to all workers involved in this field. Another paper deals with the environmental conditions required and patient management procedures to be adopted. Further papers deal with the acquisition, pre-processing and digitization and analysis of lung sounds. Guidelines for publishing the results of research and clinical trials are given so they can be more easily related to other findings. Finally, a perspective on the future of respiratory sound analysis is given.

We would like to express our cordial thanks to the whole CORSA group, which includes over 20 scientists, for their valuable and intensive work resulting in the papers in this special issue. We hope that this issue of the European Respiratory Review will facilitate the development of standardized lung sound analysis equipment and promote research into the understanding of respiratory sounds. This will inevitably lead to better and new clinical applications.

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Current methods used for computerized respiratory sound analysis

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Current methods used for computerized respiratory sound analysis J.E. Earis, B.M.G. Cheetham. ©ERS Journals Ltd 2000.
ABSTRACT: The study of respiratory sounds by computer has a considerable history, which spans a time of rapidly evolving technology and changing perceptions of analogue and digital signal processing. Much of the knowledge gained in recent years has resulted from the use of a wide variety of data acquisition, processing and analysis techniques. Details of the techniques used in published research emanating from European and other world wide centres over the past 10 yrs are surveyed in this paper. The survey reveals the range of clinical conditions studied, the type of analysis equipment used and the extent to which the engineering parameters of the various equipment used were similar and/or different. It is clear that, in addition to the well-established analysis of adventitious sounds, there is increasing interest in the analysis of breath sounds as a measure of regional physiology. In addition, over 60% of published studies over the 10 yrs involve upper airway sounds. Although marked similarities in the basic methodology were found, there was considerable variation in the way that sound was processed and analysed. It is concluded that there is a need for the development of guidelines for the recording, processing and analysis of respiratory sounds in order to facilitate the easy exchange of data and to enable a meaningful comparison of results between research centres.
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With the use of modern digital signal processing techniques, the analysis of wave forms by computer has become an established research technique for the investigation of respiratory sounds [1, 2]. As research in this area has developed, many different types of equipment and techniques have been used. The aims of this paper are to analyse the activities, in this field, in European and other main world-wide centres, to assess the degree to which equipment is standardized and to identify the techniques used to analyse respiratory sounds. To enable comparisons to be made between the various types of equipment and techniques used currently, and recently, for the computerized analysis of respiratory sounds, the authors have undertaken a survey of this area of research.

Methods

The survey was carried out in the main European and world centres by addressing the published literature, making site visits and holding meetings with workers active in the field. The study was undertaken as part of the concerted action project, CORSA (Computerized Respiratory Sound Analysis), included in the BIOMED 1 programme of the European Union.

A study of the world literature over the 10 yrs from January 1986 to January 1996 was undertaken using Index Medicus and the ILSA (International Lung Sounds Association) bibliography (http://www.ilsa.cc/referenc.htm) as current in January 1996. The 1,672 papers referring to upper and lower respiratory sounds published world-wide were analysed. Among these, 163 papers were identified that used some form of electronic wave-form analysis techniques (12.6% of the total number of papers). The ILSA bibliography includes all papers that use form of electronic analysis published since 1970. The analysis techniques used in each of the main European, North American and Israeli centres were identified and catalogued according to the criteria presented in table 1. A detailed analysis of all papers in the survey originating in Europe between 1991 and 1996 was undertaken in order to provide a detailed catalogue of the equipment and techniques used.

Survey of respiratory sounds research

The survey revealed that within Europe, the most active contributors were from the CORSA groups in Finland, England, France, Italy and The Netherlands who were responsible for about 25% of the total published world-wide papers on respiratory sounds. Table 2, produced from the literature survey, gives an overall indication of the type of work being undertaken in European centres.

Table 1. – Headings used for the survey

Name of institution:
Name of researchers

Condition under study
(*eg*. asthma, COPD)

Type of sound studied:
a) Wheeze
b) Stridor
c) Crackles
d) Snoring
e) Other

Other signals recorded,
e.g. flow, volume

Method of sound capture:
a) Microphone (type, attachment attributes)
b) Other

Storage of respiratory sounds
a) Audio cassette tape recorder
b) DAT recorder
c) Direct digitization
d) Other

Analogue pre-filtering and frequencies
a) High pass
b) Low pass
c) Overall band width of signal
d) Type of filter

Method of digitization
a) Commercial PC card (*e.g.* Soundblaster
b) Commercial A-to-D converter
c) Specially designed converter
d) Other
e) Sampling rate
f) Number of bits per sample

Type of computer used
a) PC and series (*i.e.* 486)
b) Unix work station
c) Other

Signal processing
1) Spectral analysis (type *e.g.* FFT; wavelet)
a) Number of points
b) Overlap
c) Windowing

2) Time domain expanded wave-form
3) Modelling and type
4) Neural networks
5) Digital filtering techniques
6) Other

Analysis Techniques
a) Frequency
b) RMS
c) Log power
d) Quartiles
e) Wave-form measurements (IDW *etc.*)
f) Other

Procedures applied:
a) Event identification
b) Segmentation
c) Other measurement

Display
Number of signals displayed
(*i.e.* sound and flow)
Time domain display

Spectrograph
a) Quasi-three dimensional waterfall
b) Colour spectrograph
c) Power spectra
d) Flow spectra
e) Other

Other
Graphs software used:
a) Harvard
b) Stanford
c) Corel Draw
d) MATLAB graphics
e) Other

Development of dedicated software
a) Name
b) Language used
c) What does it do?
d) What is it for?
e) Is it commercially available?
f) Is it used by others?
g) Has it been published?
h) References

Papers concerned with the study of upper airway sounds (snoring, cough and stridor) were found to make up 63% of the total. The analysis of wheeze produced 26% of the total, and the study of a variety of other respiratory sounds (*e.g.* hoarseness of voice) made up the balance. Bibliographies reviewing the overall literature may be found in the following papers on snoring [3, 4], cough [5–7], stridor [8], and wheeze [9, 10]. Studies involving an objective analysis of respiratory sounds show that lower respiratory sound analysis accounted for 55% of the total and snoring for the remaining 45%. Overall, 50% of the papers were written by CORSA participants. Over the 5 yrs of this study, the evaluation of breath sounds has produced almost as many publications as the more traditional analysis of adventitious lung sounds. Good bibliographies on breath sound analysis may be found in papers by MALMBERG *et al.* [11], GAVRIELY *et al.* [12] and SCHREUR *et al.* [13].

Table 2. – Type of sound analysed in European countries 1991–1996

Application area	Subjective analysis	Electronic analysis	Total
Snoring	190	15	205 (55%)
Stridor	31	0	31 (8%)
Cough	6	1	7 (2%)
Wheeze	97	3	100 (26%)
Crackles	0	7	7 (2%)
Breath sounds	0	8	8 (3%)
Other	13	1	14 (4%)
Totals	330	34	364

Although there is a limited amount of commercial equipment specifically designed for the analysis of respiratory sounds (for example, the "PNP Fonopneumografo" system from the Carex European Group S.P.A, Arezzo, Italy; The Sleep Sound ELENS-DSA from B.E.A. Medical, Orban, Belgium; The Helsinki Lung Sounds Analyser HELSA from Pulmer Ltd Helsinki; The RALE system from PixSoft Inc, Winnipeg, Canada and the PulmoTrack system from Karmel Medical Acoustic Technologies, Israel), most research has been carried out by using customdesigned electronic systems. Various common practices have emerged, though substantial differences in approach were also apparent. The survey reported in this paper revealed the following similarities and differences, which have been characterized under the headings presented below.

Signal acquisition methods

In all applications, sounds recorded from the respiratory system were captured by microphones or contact sensors situated at the mouth (free field), on the chest or elsewhere. Other physiological signals were commonly captured alongside the sound for example, airflow, change of lung volume and/or intra-thoracic pressure and oxygen saturation. All but two European centres and one North American Centre have published work where flow, measured with a pneumotachograph, was recorded simultaneously with the sound recording. Typically, one chest-wall sound channel was used, but in many papers, two and occasionally multiple channels were used [14, 15]. Upper airway sounds such as snoring and cough were often captured by microphones used in free field at a set distance from the patient's mouth [7]. Adventitious and breath sounds originating from the lower airways were captured from the chest wall using two types of microphone: 1) electret air-coupled microphones and 2) contact sensors (accelerometers) [1, 16]. Air-coupled microphones were used in all European Centres, but the size, shape and dimensions of the air cavity between the microphone and chest varied from centre to centre [17]. Microphone housings were generally designed and custom-made by individual centres according to particular theories and ideas. In North America and Israel, a variety of commercially available and custom-made contact sensors and accelerometers, attached on to the chest wall with either adhesive rings or a rubber belt, were employed [18].

Analogue prefiltering and storage

The analogue filtering applied to the captured sound signal varied from centre to centre according to established practice, available technology and the particular application. Most researchers employed a high-pass filter [18, 19] with a cut-off frequency chosen somewhere in the range from 30–150 Hz, the norm being around 50–60 Hz [10, 20, 21]. A low-pass filter was always used in the capture of lower airway sounds with the cut-off frequency set between ~1600 and 3000 Hz [9, 19–21]. Upper airway sounds were generally processed with higher cut-off frequencies [7]. Until 1990, normal practice was to store sound and flow signals on analogue magnetic recording tape, for subsequent digitization off line (flow signals were usually recorded using FM tape recorders). In recent years, DAT tape recorders have been used for both sound and flow, though normal practice is now direct digitization and acquisition by computer [20, 22].

Digitization protocols

Analogue-to-digital converters are used with word lengths of nominally 12, 14 or 16 bits per sample [9, 13, 20] A wide range of different sampling rates are in common use, the lowest being around 4 kHz and the highest being 22.05 kHz. Three centres used standard multi-media sound cards *e.g.* "Soundblaster" cards [22], and several others used other commercial multi-channel signal acquisition cards.

Signal processing

The spectral analysis of respiratory sounds using the discrete Fourier transform (DFT), invariably making use of a Fast Fourier Transform (FFT) algorithm. [12, 18–20], was universal. The duration of each analysis segment was typically between 20 and 50 ms, which means that with a sampling rate of around 10 kHz, signal block lengths of 256, 512 or 1024 samples were commonly used. Zero-padding and overlapping of analysis segments techniques were commonly used [9, 20, 23], and windowing was usually by a Hamming, Hann or other universally accepted type nonrectangular window. The survey revealed that newer highly advanced spectral analysis and digital signal processing techniques were being increasingly used, these included autoregressive analysis [24, 25], wavelets [26], Pronys method [27], neural networks [28] and higher-order spectra [29]. The analysis of the signal usually involved some of the following elements [18, 19, 30]: short-term power and power spectral density, spectrographs; averaged power spectra; estimation of spectral energy distribution (*e.g.* quartiles); flow representation (sometime flow gating or flow-standardized spectra) [19, 20, 30, 31]; wheeze detection [9, 18, 32–34]; crackle detection [14, 35–38]; cough detection [5–7]; snoring detection [3, 4, 39, 40], and a variety of other techniques [16, 19, 28, 30, 33, 41, 42].

Displays

Graphical representations of results were usually custom-written. Some of the commoner forms of display were power plots in the time-domain, three-dimensional spectrographs and airflow, plot of averaged power spectra and time expanded waveforms [35–37, 43,]. Many other more specialized types of display including real-time spectro–graphs were employed specific to individual centres [44]. The graphics and programming software used to produce such displays were very variable but there is increasing usage of the graphic facilities offered by versions of C++ and MATLAB.

Discussion

The results of this study show that in earlier years the analysis of lower airway sounds was directed mainly towards adventitious sounds such as wheezes and crackles and in recent years there has been an emphasis on the analysis of breath sounds [11–13]. The main reason for this is the increasing evidence that analysis of breath sounds provides clinically useful information about regional ventilation within the lungs [11, 15] (see *Future Perspectives* in this issue). In order for such measurements to be meaningful breath sounds must be normalised for air-flow [19, 20, 30]. Approximately 65 percent of the total peer review papers during this survey were concerned with upper airway sounds particularly snoring. This highlights the interest in the possible use of sound as a non-invasive way to monitor and diagnose common sleep related breathing disorders.

Sound was most commonly recorded with an air-coupled microphone, but there appears to be little uniformity in chamber size and method of attachment [16, 17]. Thus, the frequency response of the data-acquisition equipment is likely to vary significantly between centres, even when the analogue electronics are carefully controlled. Although, currently, there appears to be no standardized way of measuring the frequency response of a microphone when it is attached to the surface of the human body, PASTERKAMP *et al*. [16, 17] have now published optimum dimensions for air coupled microphones.

Previously, sound was usually recorded on to magnetic tape, using either a cassette recorder or an FM recorder (if flow was also recorded). More recently, with the ready availability of cheap computer memory, direct storage on a hard disc, read/write CD ROM or other similar media has become the norm in most centres. A considerable variation in the analogue pre-processing was revealed by this survey. A high-pass filter to eliminate muscle, heart and other low-frequency sounds which could otherwise overload the input amplifier was generally used with a cut-off frequency range between 50 and 150 Hz, but in Europe, the cut-off frequency was usually between 50 and 60 Hz [10, 20, 21]. Low-pass filtering varies according to the application.

A wide range of commercial and custom-designed equipment is used to digitize the sound and flow signals. Such equipment is now readily available and as "multi-channel cards", though there is also a trend towards the use of multi-media sound cards primarily designed for the digitizaton of speech and music [22]. There is a trend towards higher levels of over-sampling and subsequent decimation to reduce the sampling rate after digitization and digital filtering. This use of higher sampling rates also allows simpler, lower-order analogue low-pass filters to be used with higher cut-off frequencies [22, 26, 27, 44]. There is strong feeling and some disagreement about the bandwidths that need to be considered when analysing different types of respiratory sounds. Bandwidths of less than 1 kHz with sampling rates of around 2 kHz have been used in some studies of wheeze, though this is now widely believed to be insufficient, and modern technology allows much higher bandwidths and sampling rates to be readily accommodated.

Signal processing usually involves some form of time-domain measurement, *e.g.* crackle waveform characterization, and spectral analysis generally using the fast Fourier transform (FFT). The order of the discrete Fourier transform (DFT) must be dependent on the sampling frequency, the spectral resolution required and the degree of non-stationarity in the signal. Zero-padding and overlapping of analysis segments techniques are commonly used to improve displays. Real-time displays of digitized signals and their spectra are well within the capacity of current technology and have been implemented in some centres [44]. The real time aspect makes a spectrograph easier to use as signal capturing and processing parameters can be rapidly modified and useful data immediately identified (*e.g.* during overnight monitoring of asthma, or the monitoring of respiratory patients in intensive care units).

Parametric representation of respiratory sounds using techniques such as autoregressive analysis, linear prediction and wavelets has been investigated in some centres [18, 24–27]. These modelling techniques are well established in the analysis of speech and are proving useful in the analysis of respiratory sound. Specific applications include autoregressive and wavelet modelling of lung sounds [24, 29, 45], adaptive filtering techniques for crackle analysis and filtering of heart sounds [42] and linear prediction for upper airway sounds [4, 22]. The development of neural networks for lung sound research is in its early stages but offers the prospect of sophisticated automatic pattern recognition associated with specific disease states.

Various forms of display have been used, usually designed in-house, and these had some form of time-domain representation, power spectral display and a quasi-three-dimensional spectrographic display [35–37]. Different programming languages, *e.g*. "C" and s graphics packages, *e.g.* Harvard graphics, have been used to develop the software. MATLAB is now used as a development tool to try out new analysis techniques before incorporating them into new programmes. There is now some commercial interest in developing equipment based on these programmes, which would enable more centres to undertake the analysis of respiratory sounds.

Conclusions

This study shows that there are marked similarities in the basic methodology of respiratory analysis in the main world centres. However, there are many variations in the details of sound capture and analysis techniques between researchers, which make comparison of results from the different centres difficult. A clear description of methodology is essential, and this study has confirmed the necessity for the development of guidelines for the recording, processing and analysis of respiratory sounds, [18, 46], which are being developed from the results of the CORSA project. This development will facilitate the easy exchange of data, and enable meaningful comparison of results between research centres.

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Characteristics of breath sounds and adventitious respiratory sounds

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Characteristics of breath sounds and adventitious respiratory sounds. A.R.A. Sovijärvi, L.P. Malmberg, G. Charbonneau, J. Vanderschoot, F. Dalmasso, C. Sacco, M. Rossi, J.E. Earis. ©ERS Journals Ltd 2000.

ABSTRACT: Respiratory sounds contain significant information on physiology and pathology of the lungs and the airways. The frequency spectrum and the amplitude of sounds, *i.e.* tracheal or lung sounds without adventitious sound components (crackles or wheezes), may reflect airway dimension and their pathologic changes (*e.g.* airway obstruction) or pathologic changes in the pulmonary tissue. Characteristics of crackles, their timing in a respiratory cycle and their waveform, are significantly different in pulmonary disorders. Also, the wheezes may have acoustic features indicating not only the presence of abnormality in the respiratory system but also the severity and locations of airway obstruction most frequently found in asthma and large-airways stenosis.

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Sounds generated in healthy lungs and airways by normal breathing, differ according to the location where they are recorded and vary with the ventilatory cycle [1, 2]. The origin of sounds generated by ventilation is not completely clear. Probably, multiple mechanisms are involved. The lung itself cannot generate sound if there is no airflow; pressure differences between structures within the thorax or different lung volume levels cannot by themselves induce sounds in the absence of airflow. Breath sounds are probably induced by turbulence of the air at the level of lobar or segmental bronchi. In smaller bronchi, the gas velocity decreases and becomes less than the critical velocity needed to induce turbulence. Therefore, the airflow in smaller airways is believed to be laminar and silent.

The resulting noise, coming from the larger airways, has a wide frequency spectrum. It is transmitted to the skin, after filtering by the lungs and the chest wall, which act acoustically as a low-pass filter. Therefore, the nominal breath sounds recorded over the lungs have their main frequency band up to 200–250 Hz. Unfortunately, this frequency band also contains components from respiratory muscles and the heart. Above 250 Hz, there is a rapid decline in energy (fig. 1). When recorded over the trachea, the sound is not (or less) filtered. Therefore, and also due to resonance phenomena, the frequency spectrum contains higher frequency components as high as 1200 Hz (fig. 2). The frequency spectrum is also influenced by rather high tracheal resonance frequencies [3]. The main energy of tracheal sounds extends up to 850–1000 Hz, with a sharp decrease in power above that frequency (fig. 2). All types of breath sounds are dependent on turbulence induced by airflow-rate. The resultant waveform of normal breath sound is disorganized (*i.e.* contains many different frequencies) as shown in figure 3.

Breath sounds heard close to the chest wall were described by LAËNNEC [4] as "a distinct murmur corresponding to the flow of air into and out of air cells". Normal breath sound was previously called "vesicular sound" [5]. That term is no longer recommended because, at the alveolar ("vesicular") level, airflow is assumed to be zero with no possibility of generating breath sounds. Normal breath sounds have acoustically a soft character. The inspiratory phase is longer than the expiratory phase, with a ratio inspiration/expiration of about 2/1 during tidal breathing. Expiration is nearly silent.

Breath sounds are not uniform over the lungs. There are regional variations in sound intensity. At the apex, the sound is less intense during an inspiration performed from residual volume. Conversely, at the base, the sound is less intense at the beginning of the inspiration, then the intensity gradually increases and reaches its maximum at about 50% of the vital capacity [6].

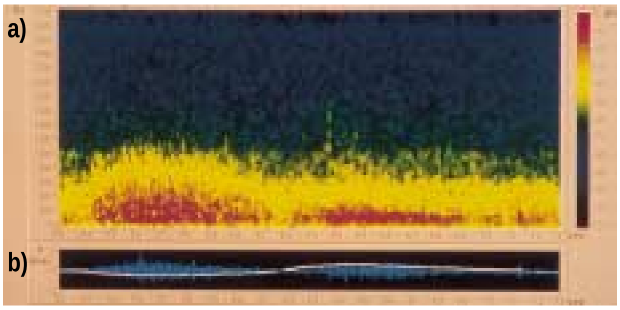

Fig. 1. – Frequency analysis in time domain: a) sonogram; and b) phonopneumogram of the lung sound sample of a healthy male (36 yrs old). In the phonopneumogram, the white curve indicates the airflow at the mouth, and the blue colour the sound.

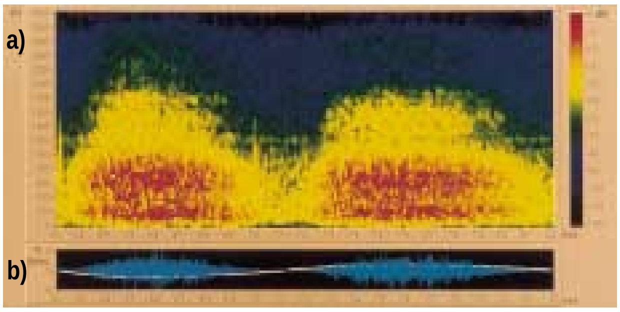

Fig. 2. – Tracheal sound sample of the same healthy man as in figure 1: a) sonogram; and b) phonopneumogram (see explanations in legend of figure 1).

Abnormal breath sounds

Breath sounds may be abnormal in certain pathological conditions of the airways or lungs. Bronchial obstruction, *e.g.* in asthma, induces an increase of higher frequency components of the sound spectrum without the appearance of wheezing [7, 8]; during bronchodilatation, the sound energy moves back to lower frequencies. In asthma, a significant association was found between the level of bronchoconstriction assessed in spirometric variables and the median frequency of breath sounds recorded over the trachea or on the chest in bronchial challenge tests [7]. Even in asthmatic patients with a normal ventilatory function, the median frequency of the breath sounds may be elevated [9]. Thus, it is probable that the allergic inflammation in the airways in asthma may induce certain changes in the mucosal or the submucosal part of the bronchi, which can induce changes in the airflow dynamics, including turbulence, during breathing.

Breath sounds with abnormally high frequencies and intensity, and with a prolonged and loud expiratory phase are typical in many diseases with airway obstruction, like in asthma and in chronic bronchitis. These abnormal breath sounds have also been called bronchial sounds. They have frequency components up to 600–1,000 Hz recorded over the posterior chest wall. In chronic obstructive lung disease (COPD) with an emphysematic component, two phenomena are often observed. Firstly, the breath sound intensity is often reduced, which has been attributed to a

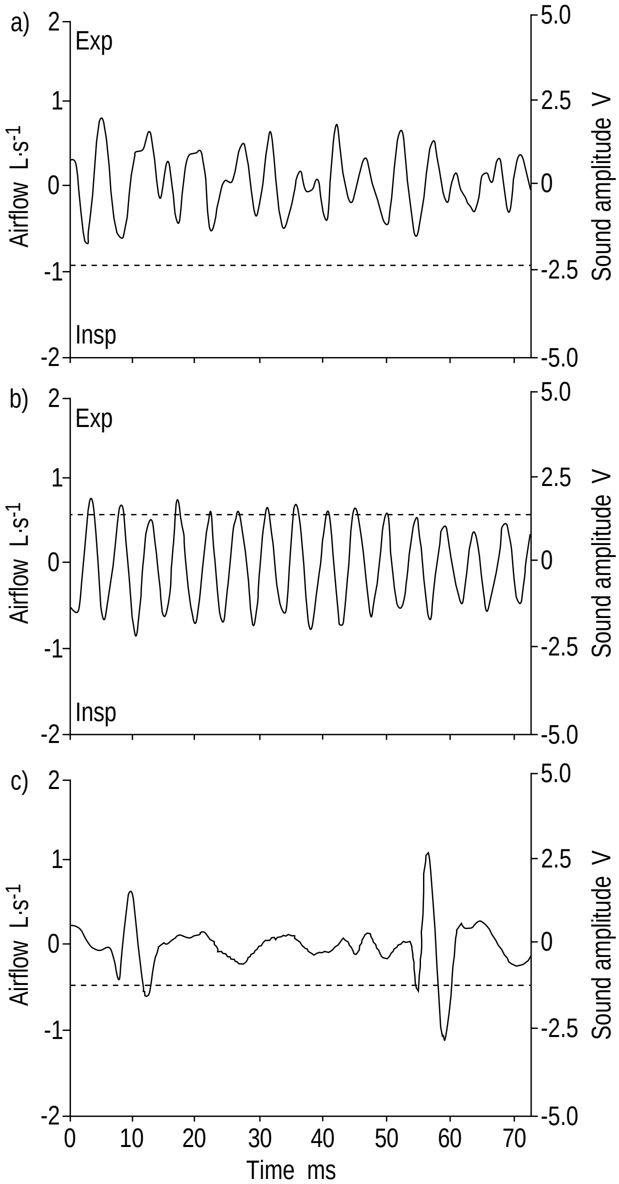

Fig. 3. – Examples of phonopneumograms of lung sounds in: a) a healthy subject; b) a patient with asthma with a wheezing sound; and c) a patient with alveolitis with crackles. —: sound amplitude; - - - : airflow. Insp: inspiration; Exp: expiration.

reduced airflow [10]. Secondly, the, values of frequency variables may be within normal limits or lowered [7], which has been attributed to an increase in the low-pass filtering effect of the damaged pulmonary tissue in pulmonary emphysema.

Adventitious sounds

Crackles

Crackles are discontinuous adventitious lung sounds [11, 12]. explosive and transient in character, and occur frequently in cardiorespiratory diseases [13]. Their duration is less than 20 ms, and their frequency content typically is wide, ranging from 100 to 2000 Hz or even higher

[14, 15]. An example of the waveform of a crackle, in a patient with alveolitis, is shown in figure 3. The frequency spectrum in time domain of a crackling sound in a patient with lung fibrosis is shown in figure 4. Two types of crackles may be distinguished: coarse and fine. The acoustical basis for this classification is well presented in the literature [12].

Crackles are assumed to originate from the acoustic energy generated by pressure equalization [16] or a change in elastic stress [17] after a sudden opening of abnormally closed airways. Crackles may sometimes occur in healthy subjects, during a deep inspiration [18], as a result of segmental reopening of dependent lung units. In those cardiorespiratory disorders where crackles are frequently found, abnormal closure of the small airways may result from increased elastic recoil pressure (*e.g.* in pulmonary fibrosis) or from a stiffening of small airways caused by accumulation of exudated fluid (*e.g.* in heart failure) or infiltrative cells (*e.g.* pneumonitis, alveolitis).

The mechanisms of generation of the crackling sounds in chronic bronchitis and emphysema are incompletely understood, but, a source in the large airways has been suggested [19]. Bubbling of air through secretions is one possible mechanism but does not account for all the crackling phenomena in these patients. In patients with chronic obstructive lung disease, the loss of elastic recoil and bronchial support [20] may predispose to collapse and subsequent reopening of the lobar bronchi [21–23].

When present, crackling sounds in patients with lung fibrosis are typically fine, repetitive, and end inspiratory, whereas those associated with chronic airways obstruction *(e.g.* COPD, emphysema or bronchiectasis) are coarse, less repeatable, and occur early in inspiration [22, 23]. Patients with airways obstruction may also have expiratory crackles, and, unlike in patients with pulmonary fibrosis, the crackles may be audible at the mouth; in addition, these crackles may change or disappear after coughing [24]. In heart failure, the crackles tend to occur from the mid to late inspiratory cycle, and they are coarse in character [25]. Mathematical models and experiments predict that crackles originating from smaller airways are shorter in duration (fine in character), and those originating from larger airways are more coarse [17].

The appearance of crackles may be an early sign of respiratory disease, *e.g.* in asbestosis 5 [13, 26]. Since the closure of small airways is gravity-dependent, crackles tend to occur first in the basal areas of the lungs, and later, when the disease progresses, also in the upper zones of the lungs. When present, the number of crackles per breath is associated with the severity of the disease in patients with interstitial lung disorders [27]. Moreover, the waveform and timing of crackles may have clinical significance in differential diagnosis of cardiorespiratory disorders [13, 24].

Squawks

0ccasionally, in patients with interstitial lung diseases, crackles may be followed by short inspiratory musical sounds; these are called squawks [19, 28, 29]. In extrinsic allergic alveolitis, squawks have been found to be shorter in duration and higher in pitch than in pulmonary fibroses due to other causes [29]. Their duration rarely exceeds 400 ms. An example of squawk is shown in figure 5. Squawks are assumed to originate from oscillation of small airways after sudden opening, and their timing seems to depend on the transpulmonary pressure in a similar manner as in crackles. Thus, the basic mechanisms of their origin probably differ from that of wheezes in asthma. Therefore, we suggest that the term "squawk" should be limited to inspiratory short wheezes in patients with interstitial lung disorders that involve small airways; otherwise, short musical sounds may be called simply "short wheezes". The basic methods of respiratory sound analysis for squawks are the same as for wheezes.

Wheezes

Wheezes are continuous adventitious lung sounds, which are superimposed on the normal breath sounds. The waveform of a wheezing sound resembles that of a sinusoidal sound (fig. 3). According to the earlier definition of the American Thoracic Society (ATS), the word "continuous" means that the duration of a wheeze is longer than 250 ms. The ATS also defines wheezes as high-pitched continuous sounds and qualifies low-pitched continuous sounds as rhonchi. The ATS nomenclature specifies that a wheeze contains a dominant frequency of 400 Hz or more, while rhonchi are characterized as low-pitched continuous sounds with a dominant frequency of about 200 Hz or less. However, investigators have not always agreed with those features. For instance, wheezes produce highly variable frequencies ranging from 80 to 1600 Hz according to GAVRIELY *et al.* [30] and from 350 to 950 Hz according to PASTERKAMP and co-workers [31].

According to the new definitions of the present CORSA guidelines, the dominant frequency of a wheeze is usually >100 Hz and the duration >100 ms [5]. Wheezes, which are louder than the underlying breath sounds, are often audible at the patient's open mouth or by auscultation by the larynx. They can be monophonic, when only one pitch is heard, or polyphonic when multiple frequencies are simultaneously perceived. Frequency spectra of wheezing sounds in asthma are presented in figures 6 and 7.

The transmission of wheezing sound through the airways is better than transmission through the lung to the surface of the chest wall. The higher-frequency sounds are more clearly detected over the trachea than at the chest [32, 33]. The high-frequency components of breath sounds are absorbed mainly by the lung tissue [34]. The highest frequency of wheezes observed by BAUGHMAN and LOUDON [35, 36], who recorded lung sounds over the chest wall, was 710 Hz. FENTON *et al.* [33] have studied the frequency spectra of wheezy lung sounds recorded simultaneously over the neck and the chest. Peaks at 870 and 940 Hz detected over the trachea were almost absent on the chest, as a result of the low-pass filtering effect of the lungs. These observations emphasize the importance of tracheal auscultation and sound recording in asthma [3, 37].

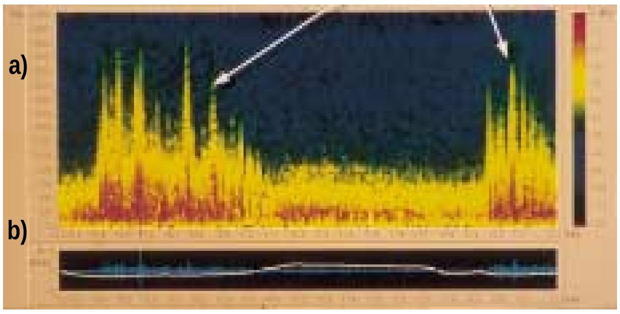

Fig. 4. – Lung sound sample (recorded in right lower lobe area) of a patients (45 yrs) with fibrosing alveolitis: a) sonogram; and b) phonopneumogram. Several inspiratory crackles can be seen in the sound sample (peaks; some indicated by arrows).

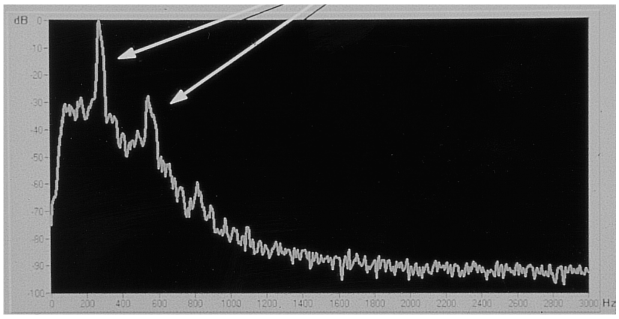

Fig. 6. – Fast Fourier power spectrum (averaged over 40 s) of an expiratory lung sound sample (recorded in right lower lobe area) of an asthma patient (female; 38 yrs) with expiratory wheezes (peaks; indicated by arrows).

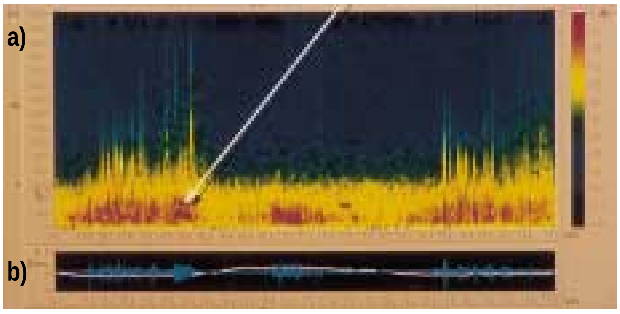

Fig. 5. – Lung sound sample (recorded in right lower lobe area) of a patient (45 yrs) with fibrosing alveolitis : a) sonogram; and b) phonopneumogram. An inspiratory squeak (arrow) and several inspiratory crackles can be seen in the sound sample.

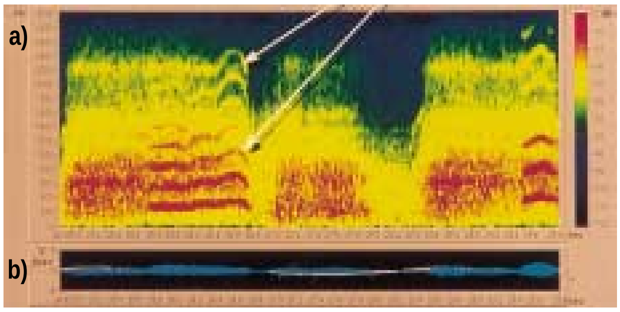

Fig. 7. – Tracheal sound sample of an asthma patient (female; 38 yrs): a) sonogram; and b) phonopneumogram. The figure indicates expiratory wheezes with harmonic components (arrows).

The mechanism of wheeze production was first compared to a toy trumpet, whose sound is produced by a vibrating reed [15]. More recently, GROTBERG and GAVRIELY [39] proposed a model in which wheezes are produced by fluttering of the airways. The oscillations begin when the airflow velocity reaches a critical value, called flutter velocity. This model shows that wheezes are always accompanied by flow limitation but that flow limitation is not necessarily accompanied by wheezes [38, 40].

Wheezes can be heard in several diseases, not only in asthma [40, 41]. They are common clinical signs in patients with obstructive airways diseases, and particularly during acute episodes of asthma. An association between the degree of bronchial obstruction and the presence and characteristics of wheezes has been demonstrated in several studies [42–44]. The strongest association has been obtained when the degree of bronchial obstruction is compared to the proportion of the respiratory cycle occupied by wheezing (t_w/t_{tot}) [35]. However, the association is too variable to predict the forced expiratory volume in one second (FEV_1) from the duration wheezing. There is no relationship between the pitch of wheezes and the pulmonary function. The appearance and quantification of wheezes have also been used for the assessment of bronchial hyperresponsiveness in bronchial-challenge tests [45].

Nocturnal asthma or asthma worsening during the night is a common complaint of asthmatic patients [46–48]. Wheezes are very often reported by patients who wake up at night with nocturnal asthma symptoms [48]. Thus, a non-invasive monitoring of wheezes has been proposed to assess changes in airways obstruction during sleep, without disturbing the patient. Different studies [49, 50] showed that monitoring wheezes during sleep in asthmatic patients provides more information on the changes in airways obstruction than measurements of pulmonary function indices during spontaneous awakening induced by symptoms of asthma.

Snores

Snores are noises commonly heard during the sleep. It is suggested that a snore is produced by vibrations in the walls of the oropharynx [51]. However, it is possible that also other structures could be put in vibration and participate to the snores. Snoring is frequently associated with the obstructive sleep apnoea syndrome and with cardiovascular diseases [52].

The snore is an inspiratory sound, although expiratory components can appear in obstructive sleep apnoea. It can occur during the whole inspiration or at the end of the inspiration. Snores are loud sounds with an intensity higher than 50 dB(A). This intensity depends on the recording technique, but mean energies as high as 85–90 dB have been reported [53–56]. The snore contains periodic components, having a fundamental frequency between 30 and 250 Hz [56, 57]. The fundamental frequency varies during the same snore or from a snore to

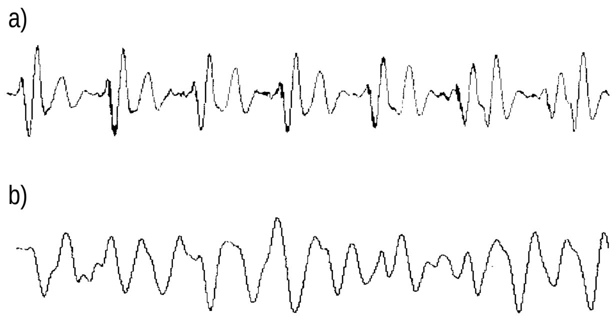

Fig. 8. – Example of waveforms of snoring sounds. a) A time domain complex of snoring waveforms consisting of repeating elements each of which is made up of four or five deflections. Each repeating element is thought to represent an individual closure of the upper airway. b) A simple snoring periodic waveform in the time domain. This pattern is thought to occur with airway vibrations without complete airway closure.

another [58]. The snore is associated with an inspiratory flow limitation, as well as an increase in airways resistance. An example of a snoring sound is shown in figure 8.

Stridors

Stridors are very loud wheezes, which are the consequence of a morphologic or dynamic obstruction in larynx or trachea. This sound can be heard near the patient without a stethoscope. The ear of a trained examiner may recognize the source of the noises: supraglottic, glottic, subglottic or tracheal [59]. Different terms are used to compare them to known noises: "cluck of turkey", "whistle of snake", "foghorn". The stridor usually occurs during inspiration when it is extrathoracic and during expiration when it is intrathoracic unless the obstruction is fixed, in which case, stridor may appear in both phases of respiration. The principal aetiology of the supraglottic stridor is suctioning of ary-epiglottic folds onto the lumen of the airways during inspiration. These phenomenoma occur because of an excess of supraglottic tissue (anatomic hypothesis). In the glottic area, the main aetiology of stridor is vocal cord paralysis. Stridor is common in infants and in babies, since the dimensions of the supraglottic area are small. However, the obstruction in babies is most often due to a subglottic viral inflammation (laryngitis).

Stridor is usually characterized by a prominent peak at about 1,000 Hz in its frequency spectrum. This component is called the pitch. The envelope of the pitch and the complexity of the spectrum (*i.e.* number of peaks or harmonics) is dependent on the disease, the site of obstruction, the airflow and the volume. Moreover, the elasticity of the obstruction and the surrounding tissues influence the sound generation. A fixed obstruction will generate a constant pitch, and a dynamic obstruction will modulate the pitch in frequency as in the case of a laryngomalacia.

Conclusion

There is a large variety of normal and abnormal respiratory sounds with characteristics, which may be typical for a disease or for a certain pathological change in the respiratory system. The ability to analyse the acoustic patterns of these breathing-induced phenomena will improve the knowledge of the physiology and pathophysiology of respiratory disorders that can be used in clinical assessment.

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Definition of terms for applications of respiratory sounds

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Definition of terms for applications of respiratory sounds. A.R.A. Sovijärvi, F. Dalmasso, J. Vanderschoot, L.P. Malmberg, G. Righini, S.A.T. Stoneman. ©ERS Journals Ltd 2000.
ABSTRACT: It has been a clear demand, based on new knowledge, to collect and update definitions of terms covering the field of respiratory sounds and computerized respiratory sound analysis for medical doctors and engineers. Also, new relevant terms should be defined. The present paper contains an alphabetical list of 162 terms and their short definitions. The terms defined have been used systematically in the present Computerized Respiratory Sound Analysis (CORSA) guidelines, and can also be applied for conventional pulmonary auscultation. Respiratory sounds for which new definitions are given include wheeze, rhonchus, squawk, breath sound and lung sound.
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Computerized lung sound analysis is a multidisciplinary field. Work in this field would be easier if the terms to be used were clearly defined. This chapter is meant for doctors, engineers and other persons who are working with respiratory sound analysis.

The terms collected in this paper of the Computerized Respiratory Sound Analysis (CORSA) guidelines include terms of respiratory diseases, pulmonary physiology, acoustics, automatic data handling and instrumentation. The list includes terms that have not yet been defined clearly in the literature, or defined in a controversial way. Some terms included have been defined elsewhere, *e.g.* the terms of lung function variables, but were regarded useful to be included in this selection in order to improve the readability. Some general terms have also been included in cases where the common sense meaning of that term is different from the specific meaning in the discipline of respiratory sound research.

Each definition of a term includes, as a rule, a short formal definition, some additional characteristics and references if available. Official guidelines (European Respiratory Society Guidelines for Lung Function Studies, American Thoracic Society publications) position papers, statements and reports of international societies as well as original papers and review articles in scientific international journals have been used as references. Handbooks have also been used.

The terms defined are listed in alphabetical order, and have been used systematically in the CORSA guidelines.

For the definition of different kinds of sounds, an attempt has been made to make clear distinctions between adjectives indicating locations (like lung, trachea, *etc.*),

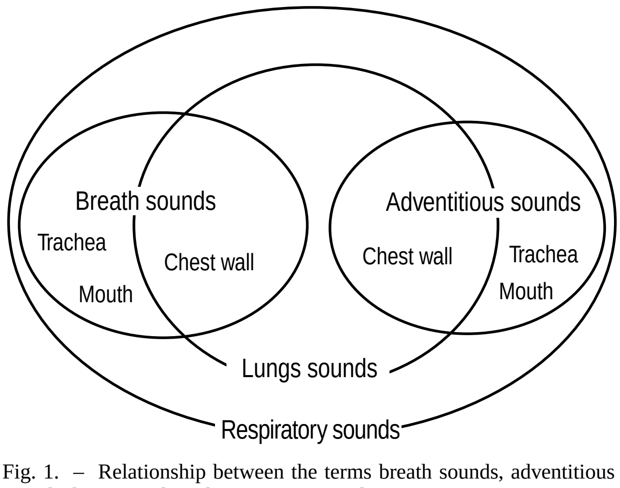

Fig. 1. – Relationship between the terms breath sounds, adventitious sounds, lung sounds and respiratory sounds.

respiratory manoeuvres (like breathing, coughing, *etc.*) and perceptions (like crackling, wheezing, *etc.*). Figure 1 indicates the basis for the present classification for the terminology of respiratory sounds. The dimensions used are the anatomical location of sound recording and the composition of the sound, either without adventitious components (breath sounds) or with them, *i.e.* wheezes or crackles (adventitious sounds). New criteria in the definitions for the terms wheeze, rhonchus, squawk, breath sound and lung sound have been applied.

List of terms

Accelerometer

A transducer which, when attached to an object, converts the acceleration of this object along the axis of the accelerometer into an electrical signal. (Two or three axis accelerometers are required if the direction of acceleration is not previously known. Angular accelerometers are used for rotary motion [1].)

Acoustic coupler

A cup-shaped transducer, of which the bottom is a microphone diaphragm. It forms a closed air chamber when attached firmly to the surface of an object.

Analogue-to-digital converter resolution

The length in bits of the binary number that represents the number of distinct and different values of the input voltage that the analogue-to-digital (ADC) converter is able to distinguish: for example 12-bit resolution corresponds to a range of 4,096 (2^{12}) values. When the maximum input voltage is known, then on the basis of ADC resolution, the sensitivity (*i.e.* the minimum change of the input voltage that can be detected) of the converter is also known. [2]

Adventitious sound

Additional respiratory sound superimposed on breath sounds. There are continuous (wheezes) and discontinuous (crackles) adventitious sounds [3–6]. There are also relatively short, adventitious sounds having features between continuous and discontinuous sounds (squawks). The presence of adventitious sounds usually indicates a pulmonary disorder.

Air-coupled microphone

A microphone coupled to the acoustic radiating surface by a properly shaped air chamber (acoustic coupler) [7, 8].

Airflow limitation

A state in the respiratory system where a further increase (during expiration) or decrease (during inspiration) of transpulmonary pressure by breathing effort does not increase airflow at the mouth [9]. Airway narrowing predisposes to airflow limitation.

Airway hyperresponsiveness

An exaggerated response of the airways to bronchoconstrictive stimuli [10]. A synonym is bronchial hyperresponsiveness. Specifically, "hypersensitivity" refers to a leftward shift and "hyperreactivity" to an increase of the slope of the dose–response curve during a bronchial challenge test with a bronchoconstrictive agent. Airway hyperresponsiveness is associated with inflammatory disorders of the airways, such as asthma.

Aliasing

The effect that, after sampling, a harmonic function appears to manifest another frequency. This occurs if the frequency of the original continuous harmonic signal is higher than half the sampling rate. The apparent frequency is equal to the smallest distance of the original frequency to any integer multiple of the sampling rate. For example, if the sampling rate is 1 kHz, a sampled harmonic signal of 800 Hz will appear to have a frequency of 200 Hz; a sampled harmonic signal of 1,000 Hz will appear to have a frequency of 0 Hz (a constant value); a sampled harmonic signal of 5,100 Hz will appear to have a frequency of 100 Hz. In general, for arbitrary signals, the spectrum should be zero above half the sampling rate. All frequency components above this frequency (the Nyquist frequency) will be "aliased", and this corrupts the actual original components in the base band [11].

Amplification

Proportional increase of the magnitude of a physical quantity (measurand) in order to better measure or observe it. [8, 12]

Amplifier

A device capable of amplification.

Amplifier gain

The ratio of output magnitude to input magnitude of an amplifier, generally expressed in decibels.

Analogue signal

Sequence of values of a quantity, which can be represented by a mathematical function having real numbers (time) as independent variable and real numbers (value of the quantity) as the dependent variable. This is the most common form of output from a transducer. [8]. An analogue signal, even from a limited time interval, cannot be stored by digital equipment since it contains an infinite number of values.

Analogue-to-digital conversion

The process of sampling an analogue signal at a finite number of points in time, and converting the level at each time into an integer number. Usually, uniform sampling (*i.e.* equidistant in time) is applied [8].

Analogue-to-digital converter

Equipment which performs analogue to digital conversions. [8].

Anti-aliasing filter

A low-pass filter with a cut-off at or below the Nyquist frequency of the digitizing process to remove those frequency components that would generate aliasing. [13, 14].

Apnoea

Cessation of breathing lasting for ≥10 s [15].

Apnoea index

The number of apnoeas per hour during sleep [15].

Apnoea/hypopnoea index

Number of breathing events per hour during sleep characterized by an abnormally low level of ventilation [15].

Autocorrelation function

The mean of the product of a signal (×) at time (*t*) with that at time $t+\tau$:

$$\rho(\tau) = \lim_{T\to\infty} \left[\frac{1}{2T} \int_{-T}^{T} x(t)x(t+\tau)dt \right], \qquad (1)$$

Where τ is the time delay between two different instances of × (t) and T is the time period of the signal. The autocorrelation is a function of τ and is used to investigate the degree of periodicity/randomness of a signal. Moreover, it provides a method for establishing the power spectral density [16].

Autoregressive process

A stochastic signal (x), for which

$$x(i) = -\sum_{k=1}^{p} a_k x(i-k) + u(i), \qquad (2)$$

where i is the consecutive number of the observatory p is the order of the model, u is a white noise process and a_k is the autoregressive (AR) parameter of the process.

x can be considered to result from the application of an all-pole filter (defined by a_k) to u.

Autoregressive moving average process

A stochastic signal (x), for which

$$x(i) = -\sum_{k=1}^{p} a_k x(i-k) + \sum_{k=0}^{q} b_k u(i-k), \qquad (3)$$

where p is the AR order of the model, q is the moving average (MA) order of the model, u is a white noise process, a_k is the AR parameter of the process and b_k is the MA parameter of the process.

x can be considered to result from the application of a filter with a rational transfer function (defined by a_k and b_k) to u.

Axillary lines

The anterior axillary line descends from the angle constituted by the anterior border of the pectoralis major with the chest wall. The posterior line descends from the angle constituted by the latissimus dorsi with the chest wall and is parallel to the anterior axillary line. The middle axillary line descends parallel to, and between, the other two.

Band-pass filter

A filter that allows components within a specific band of frequencies to pass while substantially attenuating or stopping all lower and higher frequency components.

Band-stop filter

A filter that attenuates or stops components within a specific band of frequencies while allowing all lower- and higher-frequency components to pass.

Bandwidth

The spacing between frequencies at which a band pass filter attenuates the signal by 3 dB.

Body mass index

The body mass index (BMI) is the ratio between the weight (kg) and the squared height (m^2). An acceptable measure of body fatness [17]. An international classification of the obesity (World Health Organization) is based on the following arbitrary ranges of BMI: below 20 (below normal), 20–24 (normal), 25–29 (slight obesity), 30–40 (moderate obesity) and above 40 (morbid obesity) [18].

Breath sound

The sound arising from breathing excluding adventitious sounds, heard or recorded over the chest wall, the trachea or at the mouth. The generation of breath sounds is related to airflow in the respiratory tract. Acoustically, they are characterized by broad spectrum noise with a frequency range depending on the pick-up location.

Bronchial sound

This term has been used with two different meanings. 1) Normal breath sounds detected at the upper anterior chest wall. They have approximately similar intensities during inspiratory and expiratory phases. 2) Abnormal breath sound detected at the posterior chest wall, containing higher-frequency components and a higher intensity than that of normal breath sounds at the same location. The change from normal to abnormal breath sounds is due to lung disorders. This abnormal breath sound also has an approximately similar intensity during the inspiratory and expiratory phases. Use of this term is not recommended since it can be confusing.

Central sleep apnoea

Apnoea caused by a decreased respiratory centre output characterized by the absence of both ribcage and abdominal movements. It is the least common form of sleep apnoea syndrome (SAS) [19].

Coarse crackle

Crackle that is low pitched and with a high amplitude and long duration. Its total duration (two-cycle duration (2CD)) is >10 ms [20, 21]. (See also *Crackle* section)

Computerized stethoscope

A portable device to monitor, record and analyse body sounds.

Condenser microphone

A microphone in which the metallic diaphragm is positioned in parallel to a fixed electrode, in such a way that an electric condenser is formed. The incident sound wave displaces the diaphragm and thereby causes electrical capacitance variations. These capacitance variations may be converted to voltage variations. For example, by applying a direct current (DC) voltage to the electrodes through a high resistance. [7]

Contact microphone

A microphone that senses directly the vibration of the radiating surface. [1]

Correlation coefficient

If n pairs of measurements (x_i, y_i) of two random variables, X and Y, are made the correlation coefficient r is given by:

$$r = \frac{\Sigma(x_i - \bar{x})\,(y_i - \bar{y})}{\sqrt{}\,[\Sigma(x_i - \bar{x})^2]\,[\Sigma(y_i - \bar{y})^2]\,\pi} \quad . \qquad (4)$$

This is a measure of any relationship between x and y. A value approaching 1.0 or -1.0 indicates a high probability that there is a linear (or nearly linear) relationship between the variables, while values close to 0.0 indicate no significant relationship [16].

Cough

A respiratory reflex characterized by sudden expulsion of air at a high velocity accompanied by a transient sound of varying pitch and intensity. Cough is caused by irritating stimuli in the airways or elsewhere in the body. One single cough consists of an inspiratory phase followed by an expiratory effort with the glottis closed (compressive phase) and the sudden opening of the glottis with rapid expiratory airflows (expulsive phase). [22, 23]. Chronic cough usually indicates a disease in the airways or in the pulmonary tissue.

Cough sound

Transient sound induced by the cough reflex with a frequency content between 50 and 3,000 Hz. The characteristics of cough sounds are different in different pulmonary diseases. Cough sounds containing wheezes are typical in asthma.

Crackle

Adventitious, discontinuous, explosive sound occurring usually during inspiration. It can be characterized on the basis of its waveform and duration, and timing in a respiratory cycle. A crackle can be classified, on the basis of its total duration, as fine (short duration) or coarse (long duration); see *Fine crackle*, and *Coarse crackle* sections. Occurrences of crackles in lung sounds usually reflect a pathological process in pulmonary tissue or airways [20, 21, 24–27].

Crackling period

Elapsed time between the beginning of the first crackle in a respiratory phase (inspiration or expiration) and the end of the last crackle within the same phase.

Cross-correlation function

The mean product of a signal at time *t* with that of another signal at time $t + \tau$:

$$\rho xy(\tau) = \lim_{T \to \infty} \left[\frac{1}{2T} \int_{-T}^{T} x(t)y(t+\tau)\,dt \right] \quad . \qquad (5)$$

The cross-correlation is a function of lag τ and provides a means of investigating the relationship between two signals [16, 28].

Cut-off frequency

The frequency at which the frequency response of a filter (or other circuit) is 3 dB below the maximal value of the frequency response.

Data base

A collection of data values, usually organized into records and tables following a predefined structure. A database may also contain facilities for querying and reporting.

Decibel

The ratio of two sound levels a and b is expressed in decibels as

$$20 \log_{10} P_a/P_b$$

where P_a and P_b are the respective sound pressure levels, or, equivalently, as

$$10 \log10 W_a W_b,$$

where W_a and W_b are the respective sound intensities [29]. Normally, the notional threshold of hearing (datum level) is used as a reference by taking P_b as 20 µPa or, equivalently, W_b as 10^{-12} W·m^{-2}.

Deterministic signal

The deterministic signal is a signal, the amplitude of which can be described as a mathematical function of time. This term is frequently used as the opposite of stochastic signal.

Digital signal

Sequence of values of a quantity, which can be represented by a mathematical function having integral values (time or sample number) as independent variable and integral values (value of the quantity) as dependent variable. Unlike analogue signals, digital signals can be stored and processed by computers (due to their finite nature).

Digital-to-analogue conversion

Digital-to-analogue conversion (DAC) is the process of converting a digital signal into an analogue signal. This is done by first defining the relative timing of the digital time index. Then, a continuous interpolation (*e.g.* zeroth order) between successive values is determined. Finally, some low-pass filtering is applied to remove higher-frequency components (due to the inaccurate interpolation) [30].

Discrete Fourier series

A series of numbers representing the average level in each of a series of narrow frequency bands ("bins"), *i.e.* a stepwise representation of a spectrum [31].

Discrete Fourier transform

A version of the Fourier transform (FT) applicable to a discrete time series (finite sequence of signal samples) [31, 32].

Discrete time series

A series of numbers usually proportional to the values of an analogue signal at a series of times (normally equally spaced) [33].

Distortion

A deformation of the input signal to a system, detected at its output due to nonlinear behaviour of some system component, sometimes induced by strong interference signal (intermodulation distortion). The distortion can be measured in the time domain as lack of linearity or in the complex domain as harmonic distortion and/or phase distortion [34].

Dynamic range

The range from the lowest to the highest values of the physical input to a transducer or of an analogue or digital signal at which equipment will function correctly. [8].

Electret material

A material in which, by applying a voltage across two faces, a permanent charge can be induced. This is an electrostatic analogy of a permanent magnet. When the material is strained, the voltage across the faces changes in proportion to the strain.

Electret microphone

A transducer in which incident sound is applied to a diaphragm made of an electret material sandwiched between conducting metal electrodes. When combined with a suitable transistor circuit and an energizing battery, it forms a small, highly sensitive microphone. The output commonly includes a DC offset voltage. [8, 35].

Electrodynamic microphone

A microphone in which incident sound causes a coil to move within a magnetic field so that flux through the coil varies. An electrical voltage related to the velocity is generated across the coil and delivered as an output signal.

Envelope (curve)

A curve, connecting the local extrema (either maxima or minima) of another curve.

Equivalent sound level

The equivalent sound level (L_{eq}) of a given time variable sound (noise) is the level expressed in decibels of a hypothetical constant noise, which, if substituted for the real noise for the same time interval, would involve the total quantity of sound energy. L_{eq} is calculated from

$$L_{eq} = 10 \log_{10} \frac{1}{T} \int_0^T \left\{ \frac{P_A(t)}{P_0} \right\}^2 dt, \qquad (6)$$

where T is the total measurement time, $P_A(t)$ the instantaneous A-weighted sound pressure and P_0 the reference acoustic pressure of 2×10^{-5} N·m^2. The use of the equivalent level allows the characterization of a noise [36, 37].

Expiratory reserve volume

The expiratory reserve volume (ERV) is the volume of lungs that can be maximally expired from the level of the functional residual capacity (FRC) [38].

Fast Fourier transform

The fast Fourier transform (FFT) is a very efficient algorithm (numerical process) used for calculating the discrete FTs [39–41].

Filter

A device that transforms a signal at its input into a signal at its output. Usually, the transformation aims to remove unwanted components [31]. Filters can be classified in analogue filters (*e.g.* implemented by operational amplifiers, resistors and capacitors) and digital filters (*e.g.* implemented by programmable digital hardware).

Fine crackle

Crackle that has a high pitch, low amplitude and short duration. Its total duration (2CD) is <10 ms [20, 21]. See also *Crackle* section.

Forced expiratory time

The forced expiratory time (FET_b) is the time required to exhale a specified fraction (b) of the forced vital capacity (FVC) [38].

Forced expiratory volume in one second

The forced expirtory volume in one second (FEV_1) is the maximal lung volume expired during the first second starting from full inspiration. [38].

Forced inspiratory vital capacity

The forced inspiratory vital capacity (FIVC) is the maximal lung volume that can be inspired during forced inspiration from a position of full expiration [38].

Forced vital capacity

The FVC is the volume of pulmonary gas delivered during an expiration made as forcefully and completely as possible starting from full inspiration [38].

Formant (formants structure)

Term used in phonetics to denote spectrum peak shaped owing to resonances in the voice generating system (vocal tract). A formant is characterized by the frequency of the peak, the resonance factor and the relative amplitude level [42, 43].

Fourier transform

A mathematical operation for decomposing a time function into its frequency components (amplitude and phase). The process is reversible and the signal can be reconstructed from its Fourier components [39, 44].

Free field

An acoustic space free from reflections. For example, for monitoring of snoring, stridors and cough, a microphone is commonly placed near the mouth (5–20 cm).

Frequency

The number of periods in 1 s of a periodic signal x. The unit is the Hertz (one cycle per second). An example of a periodic signal is $x(t) = \sin(\omega t)$, where ω is the angular velocity, which has a period of $T = 2\pi/\omega$, and hence a frequency (f) of $f = \omega/2\pi$ [45].

Frequency domain

The space of the variable "frequency" associated to the space of the variable "time" by an FT. In the frequency domain, a signal is described by its spectrum [45].

Frequency resolution

The width (frequency range) of the bands corresponding to each value in a discrete Fourier series (often referred to as the "bin" width of the transform) [45].

Frequency response

The response of a system to harmonic (sinusoidal) excitation of varying frequency. For linear time invariant dynamic systems, if a harmonic signal is presented at the input, a harmonic signal of the same frequency occurs at the output (after the decay of a transient phenomenon). Therefore, the steady-state response for every frequency can be represented by two numbers: the input-to-output amplitude ratio and the input-to-output phase difference [46, 47].

Functional residual capacity

The functional residual capacity (FRC) is the volume of gas present in the lungs and airways at tidal end-expiratory level. It is the sum of expiratory reserve volume and residual volume [38].

High-pass filter

A filter that allows components above a specific frequency to pass while attenuating or stepping all lower-frequency components.

Hypopnoea

Episode with an abnormally low level of pulmonary ventilation.

Impedance

The specific acoustic impedance (Z_s) is the ratio between the complex representation of the acoustic pressure (P) in a point of the sound wave and the complex representation of the particle velocity (V) in that point:

$$Z_s = P/V. \quad (7)$$

The acoustic impedance (Z_a) is the ratio between the complex representation of the mean P in an oscillating surface (S) and the complex representation of the sound flow (Q'; $=VS$) through this surface:

$$Z_a = P/Q' = Z_s/S. \quad (8)$$

The mechanical impedance (Z_m), is the ratio between the complex representation of the force that acts on an S of a mechanical system and the complex representation of the v in that S (or in that point) in the direction of the force:

$$Z_m = PS/V = Z_s S \quad (9)$$

[47, 48].

Impulse response

The response of a system to all impulsive excitation. If the system is linear time-invariant and if the impulse is the ideal delta function (Dirac function), the impulse response and the transfer function are related by an FT.

Inductive microphone

A transducer in which incident sound causes the electrical inductance of a coil to change, generally by varying the coupling between the coil and a magnetic circuit.

Initial deflection width

The initial deflection width (IDW) is the duration of the first deflection in a crackle waveform [20, 21].

Intensity (of a sound wave)

The mean rate of flow of acoustic energy through a unit area normal to the direction of propagation ($W \cdot m^{-2}$) [36].

Interference

A deterministic signal, emanating from an external source or phenomenon, and disturbing the signal of interest. Two kinds of interference are very relevant to respiratory sounds. 1) Acoustic interference, *e.g.* heart sounds, flow transduction, noisy wards, slamming doors and distance talking [48]. 2) Electromagnetic interference, *e.g.* power distribution (50 Hz) and cross-talk.

Internal field

A space of a microphone inside the body (*e.g.* oesophagus, trachea or bronchi). Lung sounds detected within the body may contain special information on the site and generation of the sound.

Jitter

A phenomenon that accompanies the digital signals all along their life, from the conversion analogue/digital, through all the transfer and processing phases, up to the conversion digital/analogue. It is constituted by all the continuous (and generally small) temporal oscillations of the instants of switching around their mean values. In sound analysis, it measures pitch perturbation, *i.e.* the variation in fundamental frequency (F_0) [49].

Largest deflection width

The largest deflection width (LDW) is the duration of the deflection of the largest amplitude in a crackle waveform [21].

Linear time invariant system

A system that can be described by linear differential or difference equations with constant parameters. These systems are relatively easy to deal with, both theoretical and practical. For example, a linear time invariant amplifier exited by a harmonic input signal will deliver at its output a harmonic signal with the same frequency and with an amplitude proportional to the input amplitude.

Low-pass filter

A filter that allows components below a specific frequency to pass while attenuating or stopping all higher-frequency components.

Lung sound

All respiratory sounds heard or detected over the chest wall or within the chest [5, 6] including breath sounds and adventitous sounds detected at this location.

Maximal expiratory flow at a specified lung volume

Maximal instantaneous flow achieved at the designated lung volume during forced expiratory maneouvre starting from full inspiration [38]. Usually, the flow is obtained when a given percentage of FVC remains to be expired.

Maximal inspiratory flow at a specified volume

Maximal instantaneous flow observed when a specified percentage of the FIVC has been inhaled [38].

Maximal mid-expiratory flow

The maximal mid-expiratory flow (MMEF) is the mean maximal expiratory flow during the middle half of the FVC [38].

Microphone

A transducer that converts incident sound waves into an analogue signal (usually an electrical voltage). [46, 47].

Mixed sleep apnoea

A type of sleep apnoea characterized by central apnoea early in the apnoea followed by obstructive apnoea. [15].

Moving average process

A stochastic signal (x), for which

$$x(i) = \sum_{k=0}^{q} b_k \, u(i - k), \qquad (10)$$

where i is the consecutive number of the observaqtion, q the MA order of the model, u a white noise process and b_k is the MA parameter of the process.

The signal (x), can be considered to result from the application of a all-zero filter (defined by the b_k) to u.

Muscle sounds

Sounds generated by skeletal muscle contraction. They are generally of a low frequency (<20 Hz) and weak intensity. The sound amplitude and the frequency are related to the contraction force [50–52].

Noise

A stochastic signal, emanating from an external source or phenomenon and disturbing the signal of interest.

Normal breath sound

A breath sound detected on 1) the chest wall, where it is characterized by a quiet low-frequency noisy sound during inspiration, and where it is hardly audible during expiration; and 2) the trachea, where it is characterized by a broader spectrum of noise *(i.e.* containing higher-frequency components) than the normal breath sounds from the chest wall, and where it is audible both during inspiration and expiration.

Nyquist frequency

The highest frequency component of a signal that can be correctly measured owing to the aliasing phenomenon:

if f_s is the sampling frequency, the Nyquist frequency is equal to $f_s/2$ [16].

Obstructive sleep apnoea

Obstructive sleep apnoea (OSA) is a type of sleep apnoea due to upper airway obstruction despite persistent ventilatory movements. It is the most common type of SAS [15].

Obstructive ventilatory defect

A state in the lungs with a pathological airflow limitation during breathing. It is characterized by, a decrease in forced expiratory volume, out of proportion to any decrease in vital capacity (VC), and by low airflows at the mouth during forced breathing maneouvres. It occurrs commonly in patients with chronic obstructive pulmonary disease and asthma [38].

Oral airflow

Airflow measured from the mouth during breathing with or without the nose closed. [53, 54].

Pack-year

A measure of the amount of smoking. It is the number of cigarettes smoked per day multiplied by the number of years of smoking divided by the number of cigarettes in a packet (usually 20).

Peak expiratory flow

The peak expiratory flow (PEF) is the maximal instantaneous air flow during a forced expiratory manoeuvre starting from full inspiration, usually reached at the beginning of expiration. [38, 55, 56].

Peak inspiratory flow

The peak inspiratory flow (PIF) is the maximal instantaneous airflow achieved during an FIVC manoeuvre starting from full expiration [38, 55].

Peak expiratory flow, diurnal variability

The diurnal variability of PEF is calculated (as a percentage) according to the formula [56]:

$$\text{Diurnal variability} = 100 \frac{PEF_{max} - PEF_{min}}{0.5(PEF_{max}+PEF_{min})}, \qquad (11)$$

where PEF_{max} is the highest value and PEF_{min} the lowest value in a day.

Percussion sounds

Percussion sounds of the chest are resonant sounds induced by a direct "thump" on the chest or indirectly by striking one finger upon another finger applied firmly to the chest. They have the greatest content of energy in the range of 150–200 Hz. The character of the percussion sounds may change due to pathological processes of the lung. [57, 58].

Periodic signal

Any signal (x(t)) for which there exists a duration (*T*) such that x(*t*) = x(*t* +*T*), in other words, if a part of the waveform is repeated over and over again. If such a *T* exists for a given signal, then also x(t) = x(t + *2T*), *etc*. However, there is always a smallest possible *T*; this value is called the period of the signal.

Piezoelectric material

Material which, when strained (distorted) by the action of external forces, becomes electrically polarized and produces voltages linearly related to the mechanical strain. Electrically, it behaves as it capacitance of which the charge varies with the imposed strain. Piezoelectric transducers are used to measure many variables, including acceleration, force and sound [47].

Piezoelectric microphone

A microphone in which incident sound acts on a piezoelectric material to generate an output voltage [47].

Pleural friction sound

Coarse crackles arising from parietal and visceral pleura rubbing against each other. Their presence indicates inflammatory processes of the pleura. They usually precede the beginning of pleural effusion and disappear when the fluid is formed. They are synchronous with breathing and are modified by posture and breathing pattern. [5, 59].

Phonopneumogram

Simultaneous and overlapped display of sound signal and air flow in time domain during breathing [60].

Physical examination

A general examination of a subject performed by a physician including inspection, palpation, percussion and auscultation.

Polysomnography

Polygraphic recording from a subject during sleep of following measures: electroencephalogram, electrocardiogram, electro-oculogram, electromyogram, nasal and oral airflow, thoracic breathing movements (by strain gauge or by inductive plethysmography) arterial oxygen saturation [61].

Posture

Body position, like supine, sitting and standing.

Power

The rate at which energy is generated/radiated/transmitted, *etc.* ($W=J \cdot s^{1}$) [45].

Power spectral density

Mean square value of the signal power in a band 1 Hz wide [31].

Power spectrum

Frequency domain data representing the power distribution of a sound with respect to frequency [31].

Quantization

Mapping of the amplitude values of an analogue signal into a series of binary numbers with a given ADC resolution. [31].

Residual volume

The residual volume (RV) is the volume of gas remaining in the lungs at the end of full expiration [38].

Resistance of airways

The resistance of the airways (R_{aw}) is the ratio of driving pressure (P) to airflow (V') ($R_{aw}=P/V'$) [38]. It can be measured with a body plethysmograph and is inversely proportional to the fourth power of airway diameter. It is increased in patients with airways obstruction.

Respiratory manoeuvre

The sequence of lung volume (V) levels during an interval of time. This sequence can be considered as a function of time ($V(t)$), of which the derivative dV/dt (or V') is the airflow at the mouth. The airflow at the mouth alone is not sufficient to characterize a certain manoeuvre, unless the lung volume at one instant is also given. Examples of generic manoeuvres are: quiet tidal breathing; forced expiratory or inspiratory effort; tidal breathing with target peak airflows, switching between inspiration and expiration at free lung volumes; and breathing with target airflows, switching between inspiration and expiration at certain target lung volumes.

Although $V(t)$ completely characterizes the manoeuvre, it is useful to represent manoeuvre by a curve in the flow/volume plane.

Respiratory sounds

All sounds related to respiration including breath sounds, adventitious sounds, cough sounds, snoring sounds, sneezing sounds, and sounds from the respiratory muscles. Voiced sounds during breathing are not included in respiratory sounds.

Restrictive ventilatory defect

Reduction of lung volumes, best characterized by a reduction in total lung capacity (TLC) [38]. It occurs commonly in disorders that restrict lung expansion such as lung fibrosis, disorders of the pleural space, chest wall neuromuscular diseases, and obesity.

Relative crackling period

Crackling period related to the duration of the containing phase.

Rhonchus

A low-pitched wheeze containing rapidly damping periodic waveforms with a duration of >100 ms and frequency of <300 Hz. Rhonchi can be found, for example, in patients with secretions or narrowing in large airways and with abnormal airway collapsibility.

Roll off

The rate at which the signal at the output of a filter (or other system) is attenuated with changes in frequency, generally expressed in decibels per octave (frequency ratio of 2) or decade (frequency ratio of 10).

Sampling frequency

The repetition rate (number of times per second) at which the level of the analogue signal is measured and converted to a digital value [31].

Segmentation

The selection of time intervals with their corresponding waveforms from a given signal. In spectral density estimation, it is usual to allow for overlapping intervals. [8].

Signal preprocessing

Modification of an analogue signal by amplification, attenuation, filtering, *etc.* prior to recording, digitizing, displaying, undergoing analysis, *etc*.

Signal processing

Signal data manipulation and analysis, generally by a digital computer, in accordance with a given algorithm. The aim of the signal processing can be diverse: *e.g.* to change the form, to reduce the amount of data, to extract meaningful information, to reduce noise and interference levels, or to calculate features for pattern recognition.

Skin fold

A measure of the double thickness of the epidermis, underlying fascia, and subcutaneous adipose tissue. [62, 63].

Sleep apnoea syndrome

The sleep apnoea syndrome (SAS) is a syndrome with repeatedly occurring apnoeas during sleep resulting from upper-airway obstruction (obstructive apnoea), or a lack of respiratory muscle activity (central apnoea), or a combination of these factors (mixed apnoea). More than 30 apnoeas per 7 h during sleep or greater are diagnostic of SAS [15].

Snoring sound (snore)

A respiratory low-frequency noisy sound with periodic components (fundamental frequency 30–250 Hz) detected usually during sleep induced by abnormal vibrations in the walls of the oropharynx. It is a typical inspiratory sound but a small expiratory component can appear especially in patients with OSA.

Socioeconomic profile

A collection of subject data including occupational activity, usual daily activity, hobbies and interest, alimentary customs, education and economic situation.

Sonogram

A particular three-dimensional presentation of the results of the sound spectral analysis, which shows how the spectrum varies with time. In the sonogram, the increasing time is associated with the x-axis, the increasing frequency to the y-axis and the amplitude of the spectral components to the hidden z-axis represented by a grayscale or colour palette. A more common term in signal processing is spectrogram (64, 65].

Sound power level

A measure of sound power, defined as:

$$L_W' = 10 \log_{10} \frac{W'}{W'_0}, \qquad (12)$$

where W' is the root-mean-square (RMS) of sound power, and W'_0 is equal to 1 pW [36].

Sound pressure level

A measure of sound pressure, defined as:

$$L_P = 20 \log_{10} \frac{P}{P_0} \qquad (13)$$

where P is the RMS of the sound pressure, and P_0 is equal to 20 μPa [36].

Specific airways conductance

The specific airways conductance (sG_{aw}) is the inverse of R_{aw} divided by thoracic gas volume (TGV) ($sG_{aw} = R_{aw}^{-1} \cdot TGV^{-1}$) [38]. It is decreased in patients with airways obstruction.

Spectrum

The Cartesian plot of the amplitude of the spectral components of a signal *versus* frequency. In sound analysis, the spectral components are generally obtained from the Fourier transform of the signal [3l].

Spirometry

A lung-function test, that measures lung volumes and their changes during inspiration and expiration. Airflow at mouth or the rate at which the lung volume is changing may also be measured by spirometry. Spirometry includes effort-dependent manoeuvres that require good co-ordination and cooperation by the subject.

Squawk

Relatively short inspiratory adventitious sound having a musical character, occasionally found in patients with interstitial lung disorders. Acoustically, its waveform may resemble that of short wheezes, but they are often preceded by a crackle. The duration of squawks may vary between 50 and 400 ms [66]. The basic mechanisms of their origin probably differ from those of wheezes in obstructive lung diseases.

Sternal notch

Small anatomical area just above the upper margin of sternum. This site can be used to monitor breath sounds.

Stochastic signal

Signal of which the amplitude level cannot be described as a mathematical function of time. A stochastic signal can be characterized by means of probability density functions. Also called the stochastic process.

Stridor

Very loud low-frequency wheeze originating in the larynx or trachea. It appears most frequently during inspiration. It can be audible at the mouth, at the trachea and over the chest wall. Stridor can appear, for example, in whooping cough, and in laryngeal or tracheal stenosis [67].

Thoracic gas volume

The thoracic gas volume (TGV) is the volume of gas in the thorax at any point in time and any level of thoracic compression during breathing. It is usually measured by the whole-body plethysmograph method from the level of tidal end-expiration [38].

Tidal volume

The tidal volume (V_T) is the volume of gas that is inspired or expired during a respiratory cycle. It is dynamically changing and commonly measured at the mouth [38].

Timed forced expiratory volume

The volume of gas exhaled in a specified time from the start of the FVC manoeuvre. Conventionally, the time used is 1 s (FEV_1) [38].

Timed forced inspiratory volume

The volume of air inhaled in a specified time, usually 1 s (FIV_1), during the performance of the FIVC manoeuvre [38].

Time domain

The natural space in which the analogue signal is represented as instantaneous amplitude *versus* time *i.e.* by its waveform [45].

Time-expanded waveform

The time-expanded waveform (TEW) is the display of a respiratory sound signal with a time scale of ≥800 $mm \cdot s^{-1}$. From a visual inspection of such a display, it is possible to study the waveforms of normal breath sounds, tracheal sounds and adventitious sounds (crackles, wheezes) and to distinguish them from each other [20].

Total lung capacity

The total lung capacity (TLC) is the volume of gas in the lungs at the end of a full inspiration [38].

Tracheal sound

Sound heard or detected over the extrathoracic part of the trachea.

Transducer

A device that converts a measurand (*i.e.* a physical input such as sound, pressure, temperature, *etc.*) into an electrical signal. The output may be in the form of analogue or digital data [47].

Transfer function

Mathematically, the FT of the impulse response of a linear time-invariant system. Usually, the transfer function has the form of a fraction of polynomials of the complex frequency variable. The zeroes of the denominator are known as poles of the transfer function. The frequency response of the system can be derived from the transfer function by substitution of the complex frequency variable by the purely imaginary $j\omega = j2\pi f$ [31], where j is an imaginary unit, ω angular velocity and *f* frequency.

Triangle of auscultation

An area free of (intervening) muscle fibres of the rhomboid muscle where breath (or adventitious) sounds can clearly be detected. This triangle is formed by the lateral border of the trapezius, the upper border of the latissimus dorsi and the medial border of the scapula.

Two-cycle duration

The two cycle duration (2CD) is the time from the beginning of the initial deflection of a crackle to the point where the waveform of the crackle has completed two cycles [20, 21].

Ventilation

Gas exchange between lungs and the surroundings of the body.

Vesicular sound

Normal breath sound detected over the chest wall. The term is based on a fallacy that normal breath sounds orginates in the alveoli (vesicles) [68]. Its use is no longer encouraged and should be substituted by the term normal breath sound.

Vital capacity

The vital capacity (VC) is the lung volume change between the positions of full inspiration and complete expiration by using a slow breathing manoeuvre [38].

Voiced sound

Sound generated in vocal cords, like speech and singing sounds.

Waveform

The curve of the instantaneous amplitude values of a signal *versus* increasing time [45].

Wavelet transform

The wavelet transform is a linear transformation that consists of decomposing an arbitrary signal into elementary contributions. Those contributions (called wavelets) are generated by dilation and translation of an analysing function called the mother wavelet [69–71].

Wheeze

Adventitious, continuous sound having a musical character. Acoustically, it is characterized by periodic waveforms [20] with a dominant frequency usually over 100 Hz [72] and with a duration of ≥100 ms; hence, the sound must include at least 10 successive vibrations. Wheezes are usually associated with airways obstruction due to various causes. If the wheeze contains essentially a single frequency, the wheeze is called monophonic. If it contains several frequencies, it is termed a polyphonic wheeze.

White noise

Stochastic signal of which the power spectral density function is constant.

Window function

A series of factors by which the elements of a discrete time series are multiplied to minimize any errors in the Fourier transform of the data that arise because the beginning and end of the time series are generally arbitrarily related to the periodic variations of the variable concerned. There are several different functions (Bartlett, Blackman, Hamming, Hanning, Gaussian, Kaiser) that may be used in various circumstances. The feature common to most is that they taper away to zero at each end. A "square window" is a series of ones and leaves the data unchanged [31].

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Environmental and subject conditions and breathing manoeuvres for respiratory sound recordings

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Environmental and subject conditions and breathing manoeuvres for respiratory sound recordings. M. Rossi, A.R.A. Sovijärvi, P. Piirilä, L. Vannuccini, F. Dalmasso, J. Vanderschoot. ©ERS Journals Ltd 2000.

ABSTRACT: This paper gives recommendations for experimental conditions to be followed during recording of respiratory sounds, including environmental conditions, subject conditions and interventions (*e.g.* bronchial challenge tests).

The optimal experimental conditions and procedures are dependent on the type of the respiratory sound to be recorded (such as breath sounds, cough or snoring), the indication for the recording (*e.g.* diagnostic, assessment of therapy, monitoring), the age of the subject (baby, child, adult), and the method and application of the recording (*e.g.* free field, endobronchial microphone).

Short-term recording and analysis for diagnostic, therapeutic and follow-up purposes have different requirements to those for long-term monitoring and analysis of respiratory sounds (*e.g.* nocturnal wheezing, cough or snoring).

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Environmental and subject conditions, and breathing manoeuvres, may have a marked influence on different variables of respiratory sounds. Therefore, basic standards are needed for these items to be able to compare the results between different tests and laboratories. The present guidelines refer to earlier guidelines for lung function testing [11 and for sleep studies [2].

Experimental applications of respiratory sounds recording

Sleep studies require methods for simultaneous monitoring of sleep stage, respiratory pattern (both effort and flow), oxygen saturation, electrocardiogram and snoring. Standardized techniques for the monitoring of sleep stage are now well established. However, optimal techniques for monitoring of respiration are still uncertain. For monitoring of mouth and nasal air flow, various types of devices have been tested. These include rapid response CO_2 analysers, thermistors, laryngeal and tracheal microphones and impedance pneumography. There have been approaches to simplify the method to assess respiration during polysomnography. These include tracheal sound recording and analysis with automated computer techniques. It has been shown that sound monitoring can support the detection of apnoeas and hypopnoeas [3, 4].

Respiratory sounds can be recorded continuously, and analysed on-line to monitor sleep apnoea, nocturnal changes of bronchial obstruction in asthma (*e.g*. wheezing time) [5], ventilation during anaesthesia [6] and regional distribution of ventilation [7].

Respiratory sounds recording and analysis can be used when assessing the response to bronchodilators and to bronchoconstrictors [8, 9] or the variations of airflow obstruction during acute bronchial challenge tests in children. Respiratory sounds can also be applied to monitoring and analysing the bronchial response to inhaled nonspecific bronchoconstrictive agents like methacholine or histamine both in children [10] and in adults [11–13]. During methacholine challenge, the appearance of wheezing detected over the trachea has been found to correlate closely with the concentration of methacholine inducing a fall of 20% in forced expiratory volume in one second in children. Moreover, changes in breath sound frequency distribution, in terms of the median frequency, have been shown to reflect the airway changes during histamine challenge tests in adults and children with asthma [9, 14]. Other authors have studied the behaviour of breath sounds during exercise-induced airway obstruction in children with asthma [15].

Treatment of acute heart failure especially can be assessed by means of lung sound analysis. The effects of breathing a gas mixture different from air (*e.g.* He 80%–O_2 20%) can be applied to study the mechanisms of generation and transmission of normal and pathological breath sounds [16, 17].

Endobronchial and oesophageal microphones can be used for research as well as multimicrophone recordings. Regional distribution of ventilation [18, 19], subtraction intensity index [20] and phase-angle test (apex to base intensity ratio) [21] are applications of multimicrophone analysis.

Monitoring of ventilation during anaesthesia in emergency departments has special needs for microphones and their locations. However, in such circumstances, there is poor control of the environmental noise.

In some cases, it could be useful to use methods for long-term recording of cough using filtered acoustic signals [22, 23]. Body movements related to the cough can be recorded by a static charge sensitive bed or by sensors giving similar information. The patient can be studied lying or sitting with no transducers or electrodes attached [23].

Environmental conditions

Satisfactory environmental conditions for short-term respiratory sound recordings, are most often achievable in laboratories of lung function or clinical physiology, where it is also possible to combine sound recordings with other physiological measurements [24].

Environmental conditions when monitoring and recording respiratory sounds in intensive care units and bedside recording at home and inpatient departments have specific features and must be adapted to the recording procedures to avoid artefacts and bias.

Noise interference

Any sound not directly induced by breathing is regarded as background noise (BN). There are two types of BN: 1) environmental noise: a) continuous noise: generated by motors, hard disks, air-conditioning ducts, fluorescent light bulbs, transformers, fans in electronic equipment and computers, traffic in the street, *etc*.; and b) transient noise: speech, music, noise from airplanes, trains, cars, slamming doors, furniture squeaks, phone rings, alarms from monitors or other electronic devices; and 2) nonrespiratory sounds and body sounds: a) related to breathing: chest motion, respiratory muscle sounds, skin friction, sounds induced by airflow in devices, and tubes and valves for monitoring of airflow and volume; and b) not related to breathing: heart sounds and murmurs, vascular sounds, swallowing, burping, bowel sounds, joint crackles, speech or other noises from the subjects.

Many methods have been proposed to eliminate environmental noises. Most of the environmental noises can be avoided by using a soundproof room. An acoustic chamber can reduce ambient background noise by up to 30 dB [25], but most frequently, it is not available for clinical respiratory sound recordings. For practical purposes, it is recommended to have a room that is free from transient noises and to have lung function devices and computers as silent as possible. The background noise level in the room where respiratory sounds are recorded should be below 45 db (A) or 60 dB (linear). Respiratory sound recording in a body plethysmograph could be a good choice to reduce ambient noise.

Shielding of the sensors with sound isolation materials can be helpful to eliminate environmental noise [26]. Different types of sensors (condenser microphone, piezoelectric sensor) have different sensitivities to environmental noise reaching the microphones directly. Therefore, this sensitivity may be a factor in the selection of a microphone.

Transient noises are much more difficult to eliminate than continuous noise. One method is to identify the sections of a sound recording that contain significant noise and discard them from the analysis. For ambient noise, this could be done automatically during the recording by means of an additional microphone.

The noise at zero airflow (breath holding) picked up on the chest wall for assessing BN in the frequency domain [27] should be measured in order to assess the quality of the recording. In spectral analysis, it can even be used to subtract the noise spectrum.

The noise generated by the turbulent vortex in flow transducers while recording the flow at the mouth can influence the power spectrum density of the sounds measured at the trachea. This phenomenon is induced even by the low-flow-resistance transducers [28]. There is no evidence that sounds picked up on the chest wall are influenced by the use of the flow transducer.

Background noise in specific circumstances

There are specific and varying requirements for the environmental shielding in sleep studies, nocturnal monitoring, bedside recordings, home monitoring, emergency departments and intensive care units, as well as during anaesthesia. However, the principles are the same as those during conventional short-term respiratory sound recordings.

Other environmental conditions

The temperature, humidity, lighting and ventilation of the room must be comfortable.

Subject conditions and procedures

Preparation of the subject before sound recording

The guidelines concerning preparation of the subjects before short-term recording of breath sounds are the same as those recommended by the European Respiratory Society [1] to be used before lung function tests.

Body position and location of microphones

The posture of the subject to be recommended for short-term recording of respiratory sounds is a sitting position. The subject should support the hands on the thighs to avoid contact of the arms with the axillary areas. During long-term recordings, the most common body position is supine.

The number of microphone pick-up locations depends on the application. Because a large number of sites has been adopted (>50) by the investigators, there is a need to define anatomically the microphone locations that have proved to be most relevant: trachea: on the sternal notch; chest, posterior: right and left bases: 5 cm from the paravertebral line and 7 cm below the scapular angle for both sides; chest, anterior: right and left anterior chest, second intercostal space, mid-clavicular line (optional); and chest, lateral: right and left axillary, fourth to fifth intercostal space, mid-axillary line (optional).

The minimal pick-up locations recommended are: trachea, and left and right posterior base of the lungs. Any variation from the standard location must be defined in the reports. Recordings from these sites can be undertaken either sequentially or simultaneously.

There are special needs of microphone location for long-term recording of respiratory sounds like cough, snoring and adventitious sounds (crackles and wheezes). Free-field re-cording may be used for snoring and cough sound analysis [29].

Respiratory manoeuvres

Short-term recording. A variety of respiratory manoeuvres have been used during respiratory sounds recording in scientific studies. Since the airflow and volume have a fundamental influence on the respiratory sound, basic standards of respiratory manoeuvres for short-term respiratory sound studies are needed. The flow rate has an influence on respiratory sound amplitude and the frequency spectrum of the sound in healthy subjects and in patients with pulmonary diseases [30–33]. Lung volume may have an effect on the spectral contents of normal breath sounds [34, 35]. However, lung volume has a strong effect on adventitious lung sounds. Wheezes theoretically appear when a flow limitation condition has been achieved [36]. The effect of lung volume on that condition is indirect. The occurrences of crackles are strongly related to lung volume [37]. For instance, crackles have been shown to appear early in the inspiration in patients with chronic obstructive pulmonary disease and late in the inspiration in patients with fibrosing alveolitis [38–40]. In the early stage of some disorders,

Table 1. – Summary of recommendations

Environmental conditions	
Acoustic noise from environment	Background noise level preferably <45 dB (A) and <60 dB (linear) Minimum ambient noise from the environment Take into account the sounds induced by airflow through the flow transducer
Nonrespiratory sounds from the subject	Minimized generation of nonrespiratory sounds including voiced sounds
Other room conditions	Comfortable room temperature, humidity, lighting and ventilation
Subject conditions and procedures	
Preparation of the subject	For short-term recordings, the guidelines for preparation of the subject are the same as those recommended by the ERS for lung function tests [1]
Body posture	
Short-term recording	The sitting position of the subject is preferred
Long-term recording	The supine position of the subject is preferred for most purposes
Microphone locations	
Short-term recording	The following locations on the body are recommended (successive or simultaneous recordings) depending on the position of the subject and the application: trachea: on the trachea at the sternal notch; and chest: right and left posterior and basal area of the chest usually 5 cm laterally from the paravertebral line and 7 cm below the scapular angle (in adults); right and left anterior area of the chest at the second intercostal space on the mid-clavicular line (optional); right and left lateral area of the chest at the fourth or fifth intercostal space on the mid-axillary line (optional)
Long-term recording	The preferred locations of the microphones are the same as in short-term recordings but optional
Respiratory manoeuvres during sound recording	
Short-term recording	Tidal breathing, 7–10 respiratory cycles, with a peak expiratory and inspiratory flow of 1–1.5 $L \cdot s^{-1}$ (adults) or 10–15% of the predicted maximum peak flow and tidal volume of 1.0 L or 15–20% of predicted vital capacity (adults) Slow vital capacity manoeuvre, two to three times successively For special purposes: forced vital capacity manoeuvres (two to three times) Breath-holding for 10 s at FRC
Long-term recording	Usually, tidal breathing without any voluntary effort
Monitoring and recording of respiratory manoeuvres	
Airflow, volume or flow/volume display in front of the subject	
For babies and young children, chest movement monitoring by strain gauge bands, pneumatic belts or chest straps	
Proper calibration of the flow-transducer, according to ERS guidelines [1]	

ERS: European Respiratory Society

crackles appear only at the end of deep inspiration (*e.g.* asbestosis, collagen diseases, extrinsic allergic alveolitis) [41].

The basic standard to be recommended in adults is to use tidal breathing, with an expiratory and inspiratory peak flow of 1.0–1.5 $L \cdot s^{-1}$ or 10–15% of the predicted maximum peak expiratory flow for 7–10 respiratory cycles. The volume of tidal breathing recommended is 1.0 L or 15–20% of predicted vital capacity in adults.

The route of breathing (nose or mouth) has an effect on overall intensity of breath sounds. Mouth breathing is recommended for two reasons: flow in the nose breathing is more difficult to standardize and sufficiently high flow levels are difficult to obtain [42]. During mouth breathing, a nose clip is recommended.

Deep breathing with slow vital capacity manoeuvres can increase the sensitivity of the appearance of adventitious sounds like crackles and wheezes in certain diseases; two to three such manoeuvres successively are usually enough for the analysis. For special purposes, forced respiratory manoeuvres (two to three times successively) are indicated.

Long-term recording. During long-term recording and monitoring, voluntary respiratory manoeuvres are not usually needed or not possible (*e.g.* during sleep) for assessment of respiratory sounds. The dominant type of breathing is tidal breathing.

Monitoring of respiratory manoeuvres. The respiratory manoeuvre must be monitored and controlled by recording the airflow. For scientific purposes, monitoring and recording of flow-volume loops are recommended. The best way is a flow or flow-volume display in front of the subject or a flow-targeting system; the simplest device is a large centre-zero analogue meter driven by the flow transducer [43].

Flow-gated sampling of breath sounds for the analysis (*e.g.* flow-standardized) is likely to reduce the effect of flow and has been used in many scientific papers [44].

Chest movements during breathing manoeuvres can be recorded by chest- strap pneumographs, pneumatic belt or strain gauge, but they give only a qualitative signal that allows the phases of the respiratory cycles to be distinguished. These methods can be applied to poorly co-operating patients and to children and to babies. Flow thermistors can also be used, but their signal is semiquantitative.

Summary of recommendations

A summary of recommendations is presented in table 1.

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Capturing and preprocessing of respiratory sounds

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ABSTRACT: This paper provides a detailed description of the analogue part of the sound acquisition chain, *i.e.* sensors, noise and interference, electronic signal conditioning devices, electrical safety and calibration techniques. Guidelines for sound capturing and preprocessing are given where possible. Otherwise, critical aspects of the acquisition chain that cannot currently be standardized are identified.
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The most commonly used bandwidth for breath sounds is from 60–100 Hz to 2 kHz when recorded on the chest (lung sounds) and from 60–100 Hz to 4 kHz when recorded over the trachea. For adventitious sounds on the chest, it is from 60–100 Hz to 6 kHz.

The analogue processing system consists of a sensor, an amplifier and filters that condition the signal prior to analogue/digital (AD) conversion. A combination of low-pass filters (LPF) and high-pass filters (HPF) in cascade is usually applied. The purpose of using a HPF is to reduce the heart, muscle and contact noises. The LPF is needed to eliminate aliasing. The amplifier is used to increase the amplitude of the captured signal so that the full AD converter range can be optimally used, and sometimes to adjust the impedance of the sensor.

The goals of this paper are: 1) to describe in detail the analogue sound acquisition chain, 2) to provide, where possible, guidelines for the operation of the acquisition chain and 3) to identify those critical aspects of the acquisition chain that cannot currently be standardized.

Sensors

Listening to respiratory sounds (by means of a stethoscope) involves several physical phenomena: vibrations of the chest wall are converted into pressure variations of the air in the stethoscope, and these pressure variations are then transmitted to the diaphragm of the ear. Devices for the recording of sound are generally called microphones. Whatever the type of microphone, it always has a diaphragm, like in the human ear, and the movement of the diaphragm is converted into an electrical signal. There are two major microphone approaches: 1) a kinematic approach, which involves the direct recording of chest-wall movement. In this case, the term "contact sensor" is used; and 2) an acoustic approach, which involves the recording of the movement of a diaphragm exposed to the pressure wave induced by the chest-wall movement. In this approach, the term "air-coupled sensor" is used. The chest-wall movements are so weak that a free-field recording is not possible; it is essential to couple the diaphragm acoustically with the chest wall through a closed air cavity.

Whatever the approach, kinematic or acoustic, vibrations have to be converted into electrical signals. For that, three major basic transduction principles have been applied: 1) electromagnetic induction: movement of a coil in a magnetic field induces an electric current through the coil; 2) condenser principle: changing the distance between the two plates of a charged capacitor induces a voltage fluctuation; and 3) piezoelectric effect: if a crystal (rod, foil) is bent, an electric charge on the surface is induced.

A dynamic microphone applies electromagnetic induction in such a way that its output voltage is proportional to the velocity of its diaphragm. This type of microphone is not recommended because it has a narrow bandwidth and may introduce distortion due to resonances of the moving coil. An electret or condenser microphone applies the condenser principle in such a way that its output voltage is proportional to the displacement of its diaphragm.

A piezoelectric microphone applies the piezoelectric effect in such a way that the output voltage is proportional to the displacement of its diaphragm with respect to its housing. Remarkably enough, this sensor is mostly used as a contact sensor: the diaphragm is directly applied to the skin, without any intervening air cavity. Due to the difference in inertia between the diaphragm and the housing, the diaphragm moves with respect to the housing, and therefore deforms.

A piezoelectric accelerometer applies the piezoelectric effect in such a way that the output voltage is proportional to the acceleration of the whole sensor. Early applications used heavy-weight sensors; these resulted in a high sensitivity and a good signal-to-noise ratio. The disadvantages due to its heavy mass are mechanical loading of the chest wall, difficulties with attachment, and a low resonance frequency (well within the band of interest). A very-low-mass (1 g) piezoelectric accelerometer, has been applied successfully in the past, yet it may be so fragile that routine clinical applications may be difficult.

Summarizing, it can be recommended that either a condenser microphone coupled to the chest wall with an air cavity (condenser microphone), or a piezoelectric microphone is applied as a contact sensor (piezoelectric contact sensor). Further support for this recommendation can be found in Refs. [1–6]. Comparisons have been made between these two types of sensors. DRUZGALSKY *et al.* [7] found that the piezoelectric sensors have a higher sensitivity and are less influenced by the ambient noise than condenser microphones, which show a larger frequency response in a free-field measurement. In a more recent study, PASTERKAMP *et al.* [8] found a slightly different result concerning the frequency response of lung sounds from the chest. This is probably due to the use of a bell to attach the condenser sensor. It has been found [8] that air-coupled and contact sensors, in the frequency range of respiratory signals, have a similar signal-to-noise ratio and spectral performance. Although both condenser microphones and piezoelectric contact sensors are displacement receivers, the waveforms that they deliver are different due to the coupling differences. The criteria for selection of a device might also include: size, average lifetime and maintenance cost. The main disadvantages of both sensors are: 1) piezoelectric sensors are very sensitive to movement artefacts, for example by the connecting wire, and their characteristics depend on the static pressure against the body surface [2], and they are brittle; and 2) condenser microphones need mounting elements that change the overall characteristics of the sound transduction.

Specifications

In this section, the minimal characteristics for a sound sensor are listed. IEC 179 is the standard for precision sound level meters, but it refers to free-field measurement only. A set of recommendations exists for sensors in phonocardiography [9]. General definitions of frequency response (FR), dynamic range (DR) and distortions have been given elsewhere in this issue [10]. These definitions are applicable to acoustic sensors also.

Frequency response

The frequency response (FR) must be flat in a frequency range that includes at least the actual band of respiratory sounds. A maximum value for deviation can be 6 dB. Microphones with a wide frequency range and small deviation are easy to find.

Dynamic range

Dynamic range (DR) is the difference between the greatest and lowest acoustic pressures that a microphone can capture. A high DR is required because of the shape of the power spectrum of respiratory sounds. DRUZGALSKY [7] suggested that the DR of microphones used for respiratory sounds should be greater than, or equal to, 40–50 dB; 50 dB can be considered as the minimum DR. Moreover, DR of the sensor should match the dynamics of the AD converter being used, see Ref. [11]; therefore, we recommend 60 dB as the dynamic range for the acoustic sensor.

Sensitivity

Sensitivity (S) is defined as the voltage generated by the microphone when the input sound pressure is 0.1 Pa (1 μbar) at a frequency of 1 kHz. S must be high when capturing low-intensity sounds. The higher it is, the more noiseless the recording can be. The effective sensitivity is determined by the signal-to-noise ratio. The stability of S is more important than its numerical value; according to the statement of the International Electrotechnical Commission (IEC) no.179, S must be steady for varying frequencies and be independent on variations of static pressure or sound direction. A minimum value cannot be given. A typical value for S of a sound sensor is 1 mV/Pa.

Signal-to-noise ratio

The signal-to-noise ratio (SNR) is the ratio between the output voltage of the microphone receiving a signal of 1 Pa at 1 kHz and the noise, *i.e.* the output voltage when there is no input signal. The SNR of the sensor cannot be lower than that of the other parts of the signal chain. During lung sound capture, the microphone is not used in its nominal condition attached to the chest, its characteristics are modified [8]. The recommendation is to use microphones with the highest possible SNR [7, 12]. A microphone with a SNR=60 dB, with S=1 mV·Pa^{-1} on a load of 200 Ω is acceptable for general purposes. Such a microphone would introduce a noise of about 1 μV. Microphones with higher SNRs are recommended for lung-sound capturing. Microphones of a higher quality reach SNR values below 70 dB.

PASTERKAMP *et al.* [8] have performed noise measurements *in situ*, with different microphones attached to the chest. The noise was determined when the airflow was "zero" (0–0.1 $L \cdot s^{-1}$). They have found that the SNR of the whole system was lower than the free-field specifications provided by the manufacturer (from 16.4–42 dB) and that it varies from subject to subject. SNR in an "*in situ*" measurement should not be under 20 dB in the frequency range of 100–500 Hz. The recommendation is to report the spectral characteristics of the noise at zero flow.

Harmonic distortion

Harmonic distortion is due to the non-linearity of the transducer. If the input signal is a sinusoidal wave at frequency *f*, then, with a linear transducer, a sinusoidal electrical signal at the same frequency is obtained. If the transducer is not linear, sinusoids with frequencies that are multiples of *f* (harmonics) are obtained. The amplitude of harmonics generally decreases with increasing frequency. The total harmonic distortion (THD) is a measure of how much the input sinusoid is affected by harmonic distortion. It is defined as

$$\mathrm{THD}\sqrt{= D_2^2 + D_3^2 + \ldots} \qquad (1)$$

where D_2, D_3, ... are the relative amplitudes of each harmonic. The following is the relation between the output power (P) and the power of the fundamental harmonic (P_1):

$$P = (1 + \mathrm{THD}^2)\, P_1 \qquad (2)$$

If THD of the fundamental harmonic is 10%, then P = 1.01 times P_1, which shows that the power output is only 1% higher than the fundamental. In this case, the contribution of higher harmonics to the spectrum is negligible. Data sheets do not always give this specification of the microphone; they often give the sound pressure value for which distortion is greater than a prefixed value (generally 0.5%).

Attachment

Coupling and fixing methods

The coupling depends on the kind of device used. In particular, condenser microphones are air-coupled, and piezoelectric transducers are contact sensors.

Air-coupled sensors. These sensors are coupled to the body surface by a closed cavity in which sounds propagate. In this way, the vibrations can move the membrane, and the sound can be captured. The presence of this cavity affects the frequency characteristics of the transmission between the body and the microphones. In particular, a resonance can disturb the flatness of the frequency response. These effects depend on the size, shape and material of the sensor. The cavity can be designed to make sure that its resonance is out of the frequency range of the signal [12–15]. Some work has been done on the effects of the depth, width and shape of the chamber on the frequency response [14, 15]. It has been found that the chamber must be shallow (a few millimetres); the width has no effects under 500 Hz. Chambers with 10 and 15 mm of diameter are optional between 500 and 1,000 Hz. Conical couplers provide 5–10 dB more sensitivity than cylindrical couplers.

A vent in the chamber can be used to equalize the internal pressure to the ambient one, but the vent transmits ambient noise into the microphonic chamber. The vent must have a high acoustic impedance by being narrow and long.

The recommended couplers have a depth of 2.5–5 mm and a conical shape 10–15 mm in diameter at the skin. They can be vented with a tube no wider than 0.35 mm. The sensors can be fixed to the body in different ways. The force of application can alter the characteristic of the air in the chamber, but this effect is minimized by the vent. Sensors can be attached by hand, by an elastic belt or by a self-adhesive coupler. In this last case, the vent is less useful. The belt cannot always be used, for example, during tracheal sound capturing.

Contact sensors

Piezoelectric contact sensors are designed to be used directly on the chest without a compressible air buffer between the skin and the sensor. They do not introduce the distortion due to the coupling chamber, but they have other kinds of problems, mainly the dependence on the force of application on to the chest wall. This should be kept constant among different measurements to allow a comparison of data and also during a single acquisition. This condition can not be easily maintained by hand. Also, the use of the elastic belt introduces variations due to the movement of the thorax during respiration. The use of an adhesive ring may be a solution to the problem.

Analogue signal processing

Noise and interference

The recording of respiratory sounds is affected by environmental interference (*e.g.* acoustic and electromagnetic noises) and by chest-wall movement, muscle sounds and heart sounds. Environmental noise is a particular problem with air-coupled sensors and can be minimized by room insulation and by shielding caps. However, little can be done to reduce environmental acoustic noise that reaches the sensor through the body. Muscle and heart noise can be reduced by filtering techniques, as discussed later on. The recording equipment should be as quiet as possible in order to achieve a SNR of at least 20 dB in the frequency band of interest during breath holding.

Immunity to electromagnetic fields

Shielding from electrical interference can be considered as part of the analogue signal processing as it may affect signal quality. A particular problem is the 50–60 Hz alternating current power line. In order to minimize this interference, shielded twisted pair or coaxial cables are recommended, while ground loops should be avoided. However, coaxial cable might induce signal distortion due to a piezoelectric effect.

Amplifier

The amplifier is described by its frequency bandwidth, noise and dynamic range; they should have the following properties: 1) the noise of the amplifier should be below that of the sound sensor; 2) the SNR of the amplified signal should be approximately 60 dB in the frequency range of interest (SNR of lung sound measurement is usually around 40 dB); 3) the amplifier must be used in accordance with its specified maximal voltage or current output (*i.e.* must not be overloaded by excessive amplification or capacitive loading).

Instrumentation amplifiers are recommended for respiratory sounds as they have high input resistance (>100 MΩ) and extremely high common mode rejection ratio (of the order of 100 dB). The latter characteristic is particularly important in the amplification of small differential signals because the common mode signal can be several orders of magnitude greater than the differential mode signal.

Filtering

High pass filtering (HPF). The HPF with a cut-off frequency of 60 Hz and a slope greater than 18 dB·oct^{-1} is generally used in respiratory sound analysis to reduce low-frequency distortion to the signal. Such distortions are typically produced by changes in the contact pressure of the sensor cup due to patient or sensor motion, by cardiovascular sounds, muscle noise, and external low-frequency noise. Without HPF, the dynamic range of the respiratory sound signal may be reduced due to saturation of the AD converter by low-frequency heart and muscle sounds. Pass- band ripple is eliminated by the choice of gain responses. Moreover, it is recommended that linear phase or delay equalized frequency responses be used, as incorrect high-pass filtering can be particularly damaging to waveforms containing crackles or other transient type signals with a wide bandwidth and short duration [16]. An approximately linear phase response can be obtained by using Bessel filters or by cascading the HPF with a delay equalization stage (allpass filter). In this case, the HPF should be a Butterworth filter because of its flat frequency response (*i.e.* without ripple). These analogue filters must be implemented as active circuits, and a particular hybrid technology can be used to implement these devices: switched-capacitor filters. These filters are analogue sampled-data systems, which are recommended because they exist in monolithic form and are programmable.

Low pass filtering (LPF). LPF is used to eliminate aliasing of the digitized signal. Aliasing occurs when the signal contains frequencies above half the sampling frequency, f_s (Nyquist frequency) [11]. The recommended LPF must have a -3 dB cut-off at the upper frequency of the signal and provide at least 24 dB of attenuation at that frequency producing aliasing in the band of the signal (see the example in Ref. [11]). Moreover, no bandpass ripple is allowed (*e.g.* Butterworth filter). The minimum for the order of the filter depends on f_s: the higher the f_s, the smaller the order can be. Details on f_s and the order can be found in Ref. [11]. Sometimes, anti-aliasing filtering is not needed in the analogue part of the chain because the AD board has its own filter. However, the behaviour of the filter is often unknown, and the testing of the frequency response is recommended using the procedure described in Ref. [17]. If the analogue signal processing chain is not known accurately or if there is any doubt that the signal may be distorted, the whole system should be calibrated by the procedure described in Ref. [11].

Electrical safety

Although no specific standard for safety in biomedical sounds recording exists, devices for respiratory sounds capturing must conform to EN 60 601-1, concerning safety in electromedical devices. The whole chain of acquisition (sensor, amplifier filters, AD converter, computer) must be designed to prevent the risk of direct and indirect contacts during normal use and under first damage conditions [18]. The sensor, brought into contact with the patient, is in "applied part", and it can be a critical aspect for safety. The recommended sensors do conform to this norm since they are passive transducers, *i.e.* they do not absorb current from an external source. The biasing voltage is low and comes from a battery (the "low safety voltage" must be less than 60 V d.c.); the condenser, if present, must have a very low capacity (from 20 to 100 pF). The only current that might reach the patient is a dispersion current coming from the amplifier, which is avoided if the amplifier conforms to the TEC guideline [18].

Calibration procedures and system response

Although the absolute magnitudes of respiratory sounds are not very important, the relative magnitude and the shape of the frequency components are of great importance, and a method for calibration and verification of the system response is needed. However, there is no ideal reference sensor or reference input signal available to calibrate the system. GAVRIELY [13] has developed a system to calibrate contact sensors, which is very complicated and is not easily used in a clinical setting. Another approach is to send a reference signal to the sensor through a simulated chest.

Summary of recommendations

A summary of recommendations for piezoelectric-or-condenser-type sensors is presented in table 1.

Table 1. – Summary of recommendations*

Sensor specifications	
Frequency response	Flat in the frequency range of the sound Maximum deviation allowed 6 dB
Dynamic range	>60 dB
Sensitivity	Must be independent of: frequency; static pressure; and sound direction
Signal-to-noise ratio	>60 dB (*S*=1 mV·Pa^{-1})
Directional characteristic	Omnidirectional
Coupling	
Piezoelectric contact	
Condenser air-coupled	Shape: conical Depth: 2.5–5 mm Diameter at skin: 10–25 mm Vented
Fixing methods	
Piezoelectric	Adhesive ring
Condenser	Either elastic belt or adhesive ring
Noise and interferences	
Acoustic	Shielded microphones Protection from mechanical vibrations
Electromagnetic	Shielded twisted pair or coaxial cable
Amplifier	
Frequency response	Constant gain and linear phase in the band of interest
Dynamic range	>60 dB
Noise	Less than that introduced by the sensor
High-pass filtering	Cut-off frequency 60 Hz Roll-off >18 dB·octave^{-1} Phase as linear as possible Minimized ripple
Low-pass filtering	Cut-off frequency above the upper frequency of the signal Roll-off >24 dB·octave^{-1} Minimized ripple

*: applied for piezoelectric or condenser sensors

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Digitization of data for respiratory sound recordings

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Digitization of data for respiratory sound recordings. B.M.G. Cheetham, G. Charbonneau, A. Giordano, P. Helistö, J. Vanderschoot. ©ERS Journals Ltd 2000.
ABSTRACT: This paper provides for a detailed discussion of the issues concerning digitization of respiratory sound recordings. Most important are guidelines for the selection of sampling frequency, filtering, analogue/digital resolution, analogue/digital input range, test and calibration procedures. It is not the goal of this paper to present standards for the data exchange format.
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The purpose of this paper is to discuss the digitization of particular types of sound, *i.e.* lung sounds as would be conventionally monitored by doctors using the stethoscope. Related problems of data and file formats for the digital storage and the transmission of these sounds over telecommunication networks will not be addressed in detail. Background material for this chapter is comprehensively covered [1–4]. The advantage of representing pulmonary signals in digital form is worth restating so that we can take care to preserve these when proposing standards.

Advantages of digitization

1) A high-quality representation of the sound is obtainable, which is potentially more accurate than an analogue recording.
2) A digital representation is more permanent than a tape-recording, for example.
3) Data stored digitally can be easily accessed.
4) Digital records may be conveniently documented and labelled.
5) Data may be duplicated without loss of quality.
6) Signal data are compatible with other data, *e.g.* patient records.
7) Data are easily communicated from one establishment to another.
8) The characteristics of a recording may be precisely defined (recording levels, bandwidth, *etc*.)
9) Digitized signals are compatible with digital signal processing techniques for analysis, making comparisons, statistical surveys, drawing graphs, *etc*.

Sampling frequency

A sampling frequency specification must depend on the type of signal being studied and the characteristics of the analogue pre-processing circuitry. Perhaps the most important consideration is the bandwidth of the signal.

A survey of the literature [5] reveals that in the study of adventitious sounds, much information about wheezes has been found within frequency bands below about 2 kHz. The upper limit has often been placed well below 2 kHz. A bandwidth extending up to 2 kHz would therefore appear to have been sufficient for many studies of wheeze, though there is evidence that useful information may be found at frequencies higher than 2 kHz. If a 2-kHz maximum frequency is taken as a working assumption, then the sampling rate should be above 4 kHz.

Although a 2-kHz bandwidth may also be reasonable for many studies of other adventitious sounds, such as crackles, the frequency content may extend beyond 2 kHz. There is likely to be useful information in crackles beyond 2 kHz, and upper airway sounds such as cough, snore and speech-like sounds certainly contain significant energy beyond 4 kHz and even beyond 10 kHz.

Clearly, a range of different recommendations for the sampling frequency must be considered in view of the different applications. Since there are no other theoretical constraints, there is a broad range in which the sampling frequency can be chosen for a given application. There appears to be no theoretical reason for preferring the sampling frequency used by any particular researcher. Therefore, it is useful to investigate the applicability of existing commercial standards.

A standardized sampling frequency applied in many industry standard sound facilities, for example for CD formats, is 44.1 kHz. In principle, this accommodates a signal bandwidth of 20 kHz, which is rather high for pulmonary sounds. Application of a universal 44.1 kHz sampling rate standard for the digitization of pulmonary sounds could result in a waste of digital storage and, perhaps more importantly, processing capacity for many applications. However, there are clear advantages in adopting sub-multiples of 44.1 kHz, *i.e.* 22.05, 11.025 or 5.5125 kHz. Many researchers have been using such standards [5].

When the sampling rate is 11.025 kHz, any input signal content above 9.025 kHz will cause aliasing in the frequency band 0 Hz to 2 kHz. This signal content must be effectively removed by an "anti-aliasing" low-pass filter before digitization. If the anti-aliasing filter is an analogue low-pass filter with 3-dB cut-off frequency at 2 kHz, and at least 24-dB attenuation is required for any aliasing, the order of the filter can be as low as two. A fourth-order Butterworth low-pass filter with a cut-off frequency of 2 kHz would give 52-dB attenuation at 9.025 kHz.

When the sampling rate is 5.125 kHz, any input signal content above 3.125 kHz will cause aliasing distortion in the frequency band 0 Hz to 2 kHz. If the analogue low-pass filter with 3-dB cut-off frequency at 2 kHz is required to have at least 24-dB attenuation at 3.125 kHz and above, the order of the filter must be at least six. A fourth-order Butterworth low-pass filter with a cut-off frequency of 2 kHz would give only about 16-dB attenuation at 3.125 kHz.

For current studies, there appear to be many advantages in adopting a standard of 11.025 kHz as the preferred sampling frequency. Among these advantages are the fact that this sampling frequency is accommodated by an extremely wide range of commercial and relatively inexpensive equipment, it allows a relatively relaxed analogue anti-aliasing filtering specification to be adopted, and it does not make excessive demands on digital storage capacity. Where storage capacity is at a premium, or possibly transmission capacity (over the internet for example), a standardized decimation strategy for reducing the sampling rate, digitally, to say 5.125 kHz could be used.

It is possible that researchers may wish to extend the bandwidth of interest to 4 kHz or even higher for studying lower-airways phenomena not so far widely recognized. Such bandwidths will clearly be of interest in the study of upper-airways sound. In these cases, it is recommended that sampling rates of 22.05 and 44.1 kHz be adopted. For a 4-kHz bandwidth with a sampling rate of 22.05 kHz, a second-order Butterworth low-pass filter (with 3-dB cut-off frequency at 4 kHz) would attenuate any aliased components by at least 24 dB; fourth order would give at least 52 dB attenuation. For an 8-kHz bandwidth, it may be best to use a sampling rate of 44.1 kHz with second-order (at least 24 dB attenuation of aliasing) or fourth-order (at least 52 dB) Butterworth low-pass analogue pre-filters.

Digitization of airflow-signal

For practical reasons, the airflow-signal during breathing will normally be digitized at the same sampling frequency as the sound. However, the analogue pre-processing must be different in that a high-pass filter should not be applied to the flow signal whose frequency content may be assumed to be band-limited well below 50 Hz. The 3-dB cut-off frequency of the analogue low-pass anti-aliasing filter can therefore be as low as 50 Hz. Alternatively, and perhaps more conveniently, the same low-pass filter specification used for the sound signal may be used, the required low-pass anti-aliasing filtering being carried out digitally. Once digitized at say 11.025 kHz, the sampling rate of the flow signal may be reduced by decimation to say 551.25 Hz by omitting 19 samples out of every 20. If the signal contains noise above 200 kHz, it may be reduced by low-pass filtering prior to the decimation process. The signal should be digitally low-pass-filtered with a 3-dB point at 20 Hz and decimated to a sampling rate of 110.25 Hz by omitting four samples out of every five. This sampling rate has been recommended as it is close to the 100 Hz used by many researchers and is a sub-multiple of 11.025 kHz.

In some systems, it may be difficult or undesirable to eliminate a high-pass filter from the channel sampling the flow signal. In this case, the flow-signal may be amplitude modulated (large carrier) on to say a 3-kHz carrier. A simple envelope detection algorithm may be used to digitally demodulate the flow signal, which is then filtered and decimated as described above.

Analogue-to-digital converter resolution (quantization error)

An ADC resolution of at least 12 bits is normally recommended for the digitization of lung sounds. In principle, this gives a dynamic range of about 35 dB for Gaussian-like signals with a signal-to-quantization noise ratio (SQNR) of at least 30 dB. (This is the ratio of the largest signal power that can be accommodated without overflow to the smallest signal power with a signal-to-quantization ratio of at least 30 dB.) Uniform quantization is assumed.

There are now strong reasons for adopting a standard of 16 bits for the ADC resolution. With highly accurate equipment, this allows a 59-dB dynamic range (with at least 30-dB SQNR), though in practice, less expensive commercial equipment will not achieve this accuracy. Nevertheless, the adoption of a 16-bit ADC word length conforms to a common commercial standard for uniform quantization and should make the achievement of true 12-bit resolution, and a little more, readily achievable in practice.

It is recommended that the 16-bit samples be stored as follows: two bytes per sample, each pair of bytes representing a 16-bit signed integer in the range -32,768 to 32,767. The least significant byte is first, followed by the most significant byte (large endian) 16-bit "two's complement" form, is used for the 16-bit signed integer.

Note that 16-bit two's complement can be converted to 16-bit "offset binary" (another common standard) or vice versa simply by negating the most significant (or "sign" bit). Offset binary is a representation which maps the range (-32,768 to 32,767) to (0–65,535) by adding 32,768 to all integers.

The use of professional multi-channel data acquisition cards with 12-bit analogue-to-digital conversion (ADC) is widespread and quite satisfactory as the quality of these cards is generally very high. It is recommended that the 12-bit integers obtained be converted to 16-bit integers in 16-bit two's complement form occupying the most significant 12 bits of the available 16, the least significant four bits being made zero.

Input gain control characteristics

If the range of the ADC is assumed to be $-V$ to $+V$, and the input signal is assumed approximately Gaussian with zero mean and RMS value, S, overflow may be considered sufficiently unlikely if S is less than about $V/4$ (four standard deviations away from the norm are considered "sufficiently unlikely" here). Hence, the maximum allowed signal power should be about $V^2/16$, and the maximum signal-to-quantitation noise ratio (SQNR) will be about 65 dB. The input amplifier should therefore be set to produce an ADC input with a maximum power of $V^2/16$, and a minimum power of 35 dB less than this.

Inter-channel delay

This depends on the sampling equipment used and should not cause any difficulty as long as it is known. The delay can be made zero using a "fractional sampling interval delay" method, accomplished with a Finite Impulse Response (FIR) digital filter to each delayed channel.

Verification (test) procedures

Procedures are suggested for verifying that the recommended standards are being adhered to. The use of a 600-Hz (say) sinusoidal test signal is recommended with a test program running on the computer. The test program should be capable of displaying a section of the captured time-domain waveform, performing a 1,024-point Hanning windowed FFT (Fast Fourier Transformation) and displaying the resulting power spectrum. An average of a large number of such power spectra should also be a display option. A sinusoidal amplitude of up to V should not produce any visible overload distortion (clipping or wraparound). In principle, the average of a number of power spectra with a sinusoid of amplitude say $V/128$ (minimum to give a SQNR of about 30 dB) should produce a characteristic power spectrum with a noise floor that is flat and about 60 dB lower than the frequency sample at 600 Hz. The graph obtained in practice should be recorded as an indication of how close to the ideal the system is.

It is also recommended that a range of higher-frequency sine-wave measurements be made to evaluate and verify the effect of the anti-aliasing input filters.

Calibration procedures

Interesting problems arise in relation to the characterization of the frequency response of the analogue pre-filters that precede the digitization process. This filtering may be realised partly by components on a commercial acquisition card and partly by custom-designed external analogue equipment. For various reasons, the precise details of the pre-filters may not be known and may even be variable due, for example, to temperature variations, impedance matching and programmability. The low-pass anti-aliasing filter characteristics are not likely to be too critical, especially when a sampling rate is used that is high enough to allow a relaxed specification. However, it is the practice, particularly when analysing adventitious lung sounds, to include a high-pass analogue filter, with a cut-off frequency typically between 50 and 100 Hz, to remove heart sounds and extraneous microphone pick-up that could otherwise cause non-linearities due to overload and amplifier clipping. Further, it is not unreasonable for some experimental work to use commercial "sound cards" for lung-sound digitization, and such cards, being primarily designed for speech and music, generally include a high-pass filter whose frequency response may not be readily ascertained.

The effect of the high-pass filter may be considerable. The gain response will strongly affect spectrographs, for example. Also, non-linearities in the phase response will affect the time-domain wave shapes, and this may affect measurements made of crackles, for example. The lung-sound researcher, therefore, needs a means of measuring and documenting the frequency response, particularly of the high-pass filter. A calibration procedure is required that is easily carried out using laboratory equipment readily available to all researchers. The frequency response must be determined solely from the response of the pre-filters to some agreed analogue test signal. This pre-filter response will be stored in a computer file in digitized form. The test signal itself will be assumed to be unavailable because digitizing it would require a second channel that would also have a partially unknown analogue filter. Further, it is assumed that the test signal must be capable of being generated from an unsophisticated standard laboratory function generator so that all researchers can easily carry out the test procedure. The test signal can therefore be specified only in general terms. The use of a Gaussian white noise source as an analogue test signal is ruled out by the above specification if the phase response of the input filter is to be measured as well as the gain response. This is because an exact version of the source would be needed for the phase response analysis.

Four possible solutions to this problem have been investigated and are described in [6]. It was concluded that the "square wave response method" [6] is satisfactory for the measurements required. This method is based on the discrete Fourier transform (DFT) analysis of the response of the pre-filter to a complete cycle of a 1 Hz square wave. Full details will be published separately.

Summary of recommendations

1) Sampling frequency for sound channel:

main 11.025 kHz;
subsidiary 5.5125, 22.05, 44.1 kHz.
2) Anti-aliasing analogue filter for main recommendation: fourth-order Butterworth low-pass filter with 3-dB cut-off frequency at 2 kHz. The cut-off frequency must be increased to 4 or 8 kHz for subsidiary recommendations of 22.05 or 44.1 kHz, respectively. Digitally low-pass filtering to 2 kHz or below before decimating to 5.5125 kHz.
3) Sampling frequency for flow channel: as for sound channel.
4) Filtering and decimation for flow signal: digitally low-pass filtering to 20 Hz and decimation to achieve a 110.25-Hz sampling frequency.
5) Word length: 16 bits per channel, two's complement, large endian.
6) Given the ADC voltage range $-V$ to $+V$, the maximum true signal power should be about $V^2/16$ squared volts, or $10\log_{10}(V/4)$ dB as measured by a "true RMS" meter.
7) The inter-channel delay should be known.
8) Verification procedures should be applied to check that clipping/overload will not occur, that channel noise or pick-up is not excessive, and that the anti-aliasing filters are working correctly.
9) Calibration procedures are available for determining the gain and phase responses of analogue pre-conditioning filters and should be used.

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Basic techniques for respiratory sound analysis

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Basic techniques for respiratory sound analysis. G. Charbonneau, E. Ademovic, B.M.G. Cheetham, L.P. Malmberg, J. Vanderschoot, A.R.A. Sovijärvi. ©ERS Journals Ltd 2000.
ABSTRACT: This paper presents a detailed discussion and guidelines for the application of basic analysis techniques to respiratory sounds. More specifically, expanded time waveform, classical spectral analysis and parametric spectral analysis are presented briefly. Detailed specifications for the application of these techniques to the different types of respiratory sounds are then presented.
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This paper forms part of the publications of the CORSA project supported by the EU BIOMED I programme (contract BMH1- CT94-0928) and by the European Respiratory Society (ERS task force).

The purpose of this paper is to give recommendations on the application of basic techniques that are commonly used in computerized respiratory sound analysis. Thus, this is not an exhaustive review of all possible techniques that could be used in lung-sounds processing. For instance, the recent wavelet analysis could certainly be applied to respiratory sounds, but so far, it has not been commonly used, and for this reason, it is not discussed here.

A compact disc of normal and pathological lung sounds has been recorded at the Angers hospital by J-L. Racineux. These sounds are available as examples on the Internet [1]. There are three other Internet sites where sound samples are also available [2–4].

Signal-processing methods

Expanded time

In phonopneumography, the sound amplitude signal is displayed simultaneously with the airflow (or volume) signal as a function of time. With this mode of representation, temporal relation of respiratory sound events or amplitude with respiratory cycle and airflow (or volume) is possible, as well as the identification of several charac teristic waveforms such as wheezes and crackles. Crackling sounds can usually be visualized as transient peaks in the sound signal in time domain. This mode of display may serve as a basis in respiratory sound analysis [5, 6], in order to provide an overview of the recorded signal of possible artefacts. For extracting detailed information of sound waveforms, recording of breath sounds onto, magnetic tape with high tape speeds, and using a slower play-back mode and a conventional chart recorder for analysis has been applied [7]. This technique, called time-expanded waveform analysis (TEWA), may be accomplished easily by the use of computerized digital signal processing at the desired time-base resolution. TEWA has been found to be particularly useful in characterizing waveforms of crackles. For this purpose, a resolution of 3,000 $mm \cdot s^{-1}$ is recommended.

In TEWA, normal breath sounds are characterized as an irregular signal shape, without any repetitive pattern or sudden rapid deflections. In contrast, adventitious continuous sounds such as wheezes and stridor have a periodic waveform, either sinusoidal or more complex. Also, snoring sounds may appear as repetitive sound patterns, even if they are not musical in character. Crackles have a distinctive appearance in TEWA: a sudden short deflection followed by deflections with a greater amplitude. Arbitrary criteria for the amplitude of the deflections have been used [8]. The initial deflection of an expiratory crackle is usually opposite to that of inspiratory one [5, 9, 10]. Depending on the high-pass filtering, the initial deflection may be followed by one or several other deflections. The original waveform of a crackle is simply monopolar or bipolar without any further deflections [11]. When the characteristics of high-pass filtration have been standardized as described in [12], the crackle waveform may be subjected to further analysis by determining several morphological features, that correlate with the clinical classification of fine and coarse crackle (see below). Figure 1 depicts an example of a crackle waveform in TEWA.

Classical spectral analysis

The representation of a phenomenon in the time domain may hide some important characteristics and thereby make any comparison difficult. The Fourier transform (FT) is a mathematical tool that decomposes a time signal in another representation introducing the concept of frequency. This idea, first introduced by J.B. Fourier, is very useful because it can be easily related to physical phenomena (time, vibrations in mechanics or acoustics, lightwaves). FT has become an outstandingly useful technique to analyse and detect hidden periodicities. Nowadays, nonparametric analysis is still used widely, even though new techniques, *e.g.* modelling or wavelets, have been developed. This analysis is very simple to set up and does not require much knowledge on the statistics of the observed phenomena. Because fast computational algorithms have been developed, it is used in real-time systems. The basis of the FT will be reviewed and its application to respiratory sound analysis presented. An extension of this technique, the short-time fourier transform (STFT), is also discussed.

Definition and limitations. The FT ($X(f)$) of a finite energy function ($x(t)$) is defined as its decomposition on to a set of trigonometric basis functions ($\psi(t, f)$) [13]:

$$\psi_f(t)=e^{j2\pi ft},\ (t,f)\in \mathrm{R}^2, \text{ and} \tag{1}$$

$$X(f)=\int x(t)\overline{\psi}_f(t)\mathrm{d}t, \tag{2}$$

where $\overline{\psi}$ (.) denotes the complex conjugate of ψ (.), t denotes the time, f the frequency and R the real numbers ensemble.

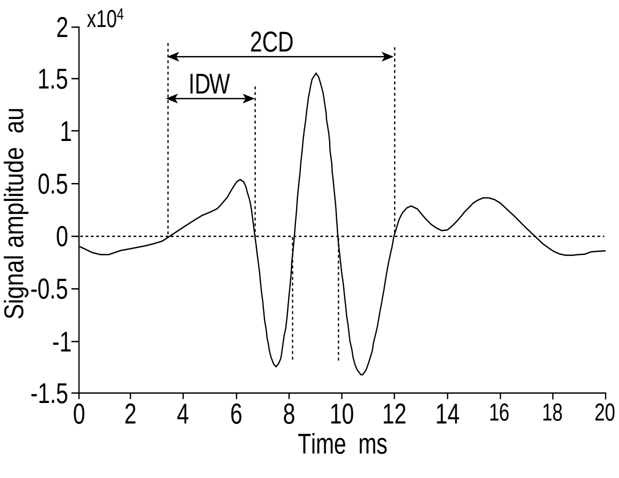

Fig. 1. – Crackle in the time domain. 2CD: two-cycle duration; IDW: initial deflection width.

A major property of this transform relies on the Parseval theorem, which insures that the energy is the same in both representations:

$$||x||^2 = ||X||^2. \tag{3}$$

From Equation 2 it can be seen that computation of the FT requires the whole scope (past and future) of a signal. The signal is decomposed as a linear combination of continuous pure waves, without any idea of occurrence in time. All the tones are simultaneously produced with a constant amplitude ($X(f)$), and, because of the interference combinations (destructions or reinforcements), the reconstruction is made perfect [13]. Such a presentation has no physical sense.

Another intrinsic limitation of the Fourier analysis concerns the resolution. It can be demonstrated that the expected resolution is controlled by the Gabor-Heisenberg inequality [4]:

$$\Delta_f \Delta_t \geq \frac{1}{4\pi}, \tag{4}$$

where Δ_f and Δ_τ are the dispersion measures of $X(f)$ and $x(t)$. This expresses the impossibility of obtaining any reasonable resolution simultaneously in time and in frequency. A fine analysis in frequency needs observation over a long period of time, and only a coarse analysis is possible on a short-period observation.

Discrete Fourier transform. Continuous transformations of analogue recordings cannot be calculated, so data have to be presented in discrete form in order to estimate the FT on computers. Instead of dealing with $x(t)$, a finite series of the signal is worked with if N equally spaced samples of the continuous signal are considered:

$$x(nT)=x[n]=x_n,\ 0\leq n<N, \tag{5}$$

and the discrete basis, expressed as:

$$\psi_k^N[n]=e^{j2\pi\frac{kn}{N}} \text{ where } k\in[0, \ldots, N), \text{ and} \tag{6}$$

$$X_k=\sum_{n=0}^{N-1} x_n\overline{\psi}_k^N[n]. \tag{7}$$

The discrete Fourier transform (DFT) (Equation 7) can be found from Equation 2 by a discretization in both time and frequency.

Subsequently, this discretization imposes constraints on the sampling frequency (F_s): $F_s = 1/T$ (8). This is known as the Shanon-Nyquist theorem (see section *Digitisation of data*). The signal has to be band-limited, and the upper boundary must $\leq F_s/2$ to avoid aliasing phenomena.

In conclusion, the discrete spectrum is computed on the same number of points comprising the time series, and only half of them are relevant for the spectral analysis due to the symmetry of the trigonometric analysis. The resulting spectrum describes the analysed sequence on the bandwidth $F_s/2$ corresponding to the discrete indices [0; $N/2$ Hence, we can express the resolution (R) as:

$$R=\frac{F_s/2}{N/2}=\frac{F_s}{N}. \tag{9}$$

Estimation of the power spectral density. Let x be a discrete time stochastic process composed of a deterministic signal (y) disturbed with an additive noise (w):

$$x_n=y_n+w_n\,. \tag{10}$$

If it is supposed that this process is stationary up to order 2 and ergodic, then the power spectral density (PSD) of x is defined as the FT of the correlation function (ρ_x) (Wiener-Khintchine theorem) [14]:

$$S_x\,[k]=\sum_n \rho_x\,[n]\,\overline{\psi}_k^N\,[n]=\lim_{N\to\infty}\mathrm{E}\left\{\frac{1}{N}\,|X_k|^2\right\}. \tag{11}$$

Ergodicity is a property that is difficult to verify. Thus, it is generally assumed to be true. Stationariness of the signal is difficult to check too, but this problem can be alleviated by paying attention to the choice of the duration of observation. Thus it is very easy to respect this condition if a small time window is chosen. The theorem of Wiener-Khintchine (Equation 11) ensures that the FT of the analysed series X_k converges toward the PSD of the process.

Periodogram. This PSD estimator (periodogram Π_x) is directly inspired by Equation 11.

$$\Pi_x\,[k]=\frac{1}{N}\,|X_k|^2. \tag{12}$$

It can be seen that its implementation requires only the computation of the FT on the time series without considering the auto correlation function [15–17].

This estimator is asymptotically unbiased (see Equation 12). Because of its very important variance, several methods to improve the periodogram have been proposed. An example of a wheeze periodogram is shown in Figure 2a.

Window influence. Computing the FT on the raw signal, *i.e.* a limited time series, is equivalent to applying a rectangular window on to the data. This windowing introduces discontinuities on the boundaries of the analysed sequence (side-effects), leading to artefacts in the decomposition coefficients. In order to reduce these boundary effects, a window can be applied on the analysed time series. This operation is not always relevant. Moreover, it can easily be demonstrated that no gain can be observed on the variance of the estimator and that, in any case, the resolution is lowered since this operation corresponds to a convolution. Windows are also used for other reasons, for instance in methods where analysed segments overlap (Welsh method) or to introduce a time dependence in the analysis, *e.g.* STFT (see [18]). Generally, the Hanning window (Equation 14) is commonly used in such an estimation. Figure 2b illustrates the use of Hanning's window applied to the same signal as in figure 2a.

$$x_w\,[n]=x\,[n]w[n] \tag{13}$$

$$w\,[n]=0.5\left[1-\cos\left(2\pi\frac{n+1}{N+1}\right)\right],\ n=0,\ \dots,\ N-1. \tag{14}$$

In [19], a presentation of the most used windows is given with a comparison of their performances in the spectral analysis.

Averaged periodogram. Instead of performing the estimation on the whole scope, the sequence is divided into P subsequences of length, M (the sequences may overlap). On each segment, a periodogram is computed, and the P periodograms are averaged. The variance is reduced by a ratio equal to P. Improvements brought by such averaging is demonstrated in figure 3, using the same signal as in the previous figures.

$$\Pi_x^{\mathrm{avg}}\,[k]=\frac{1}{P}\sum_{j=1}^{P}\Pi_{x(j)}\,[k]. \tag{15}$$

Conclusion of the non-classical spectral analysis. Applying one of the enhanced periodogram techniques can reduce the variance of the estimated value, at the expense of frequency resolution. Therefore, application requirements

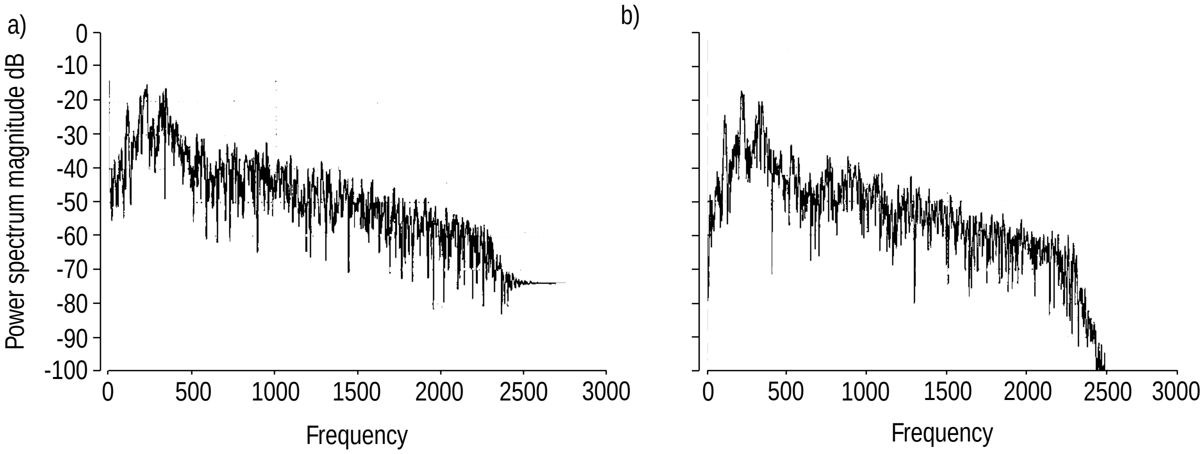

Fig. 2. – Power spectral density estimate of a wheeze with: a) the periodogram technique; and b) the effect of Hanning window preprocessing.

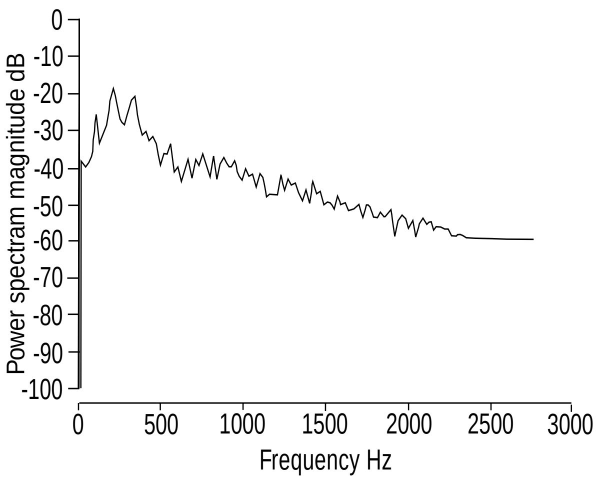

Fig. 3. – Power spectral density estimate with averaged periodogram.

have to be taken into account when deciding to perform the raw periodogram or rather an enhanced periodogram with its inherent trade-off between resolution and variance.

These techniques are the first step of an analyzing method and are mainly devoted to stationary signal analysis. Thus they are well adapted to wheeze and upper-airways sounds detection. Applied to these classes of respiratory sound, they require in addition, to build up a specific treatment to highlight the presence or lack of these phenomena in the analysed sequence. Such algorithms have been fully described for wheezes in several papers, *e.g.* [20–22]. Usually, efficient detectors should take care of signal statistics to pinpoint wheezes with great accuracy.

Short-time Fourier transform

The FT induces the idea that the frequencies are not localized in time and present on the whole scope of analysis. Physically, this representation has no sense: a frequency is produced at time, t and has a duration, τ.

The Fourier analysis resolves enough to highlight the different tones involved in a signal and is inefficient in describing their respective occurrences.

A method of analysis dedicated to the time/frequency plane has been developed from Fourier analysis: the STFT. By introducing a window centred in time around θ to weight the basis ψ (t, f), a new family of analysing functions is defined: also known as time–frequency atoms (Equation 16) [14]. A local analysis can be achieved with this new basis (Equation 17).

$$\Phi_{\theta,f}(t)=w(t-\theta)\;\psi(t,f) \qquad (16)$$

$$T_x(\theta,f)=\langle x(t), \phi_{\theta,f}(t)\rangle. \qquad (17)$$

where w is defined by equation 14.

The modulus $|T_x(\theta, f)|^2$ is named the sonogram, spectrogram or respirosonograph, and is a description of the energy in the TF plane. This representation is well known in speech processing [15] and in respiratory sound analysis [23–26]. The consecutive spectra can be computed with or without an overlap.

The advantage of such a representation is the ability to reintroduce the notion of time. The signal is no longer characterized by a mean spectrum. The evolution of its "instantaneous" and successive spectra is observed [27, 28]. It represents, in a three-dimensional coordinate system, the energy of a signal *versus* time and frequency. In fact, the time/frequency plane is tiled with rectangular bins, because of the Gabor-Heisenberg inequality (Equation 4).

This mapping can be displayed either in three dimensions (*e.g.* waterfalls) or in two dimensions (*e.g.* spectrogram/sonogram).

Parametric spectral analysis

If some additional knowledge is available for a given stochastic signal, it may be possible to "simplify" the spectral estimation problem. In this paragraph, the approach is illustrated for the autoregressive (AR) signal. Given a stochastic signal, for which there is the additional knowledge that it is AR, the spectral estimation problem is "simplified" as follows.

For convenience, the definition of the AR process is repeated, as given in [29]. A stochastic signal, x, is called an AR signal if, and only if,

$$\forall\, i \quad x(i)=-\sum_{k=1}^{p} a_k x(i-k)+u(i). \qquad (18)$$

where p is the order of the model, u is the white noise process and a_k are the AR parameters of the process. x can be considered to result from the application of an all-pole filter (defined by the a_k) to the u. If all the poles have a magnitude <1, then the AR process is stationary. It can be shown that the power spectral density of the AR signal is given by

$$P_x(f)=\sigma^2\,\frac{1}{|1+\sum_{k=1}^{p} a_k e^{j2\pi fk}|^2}, \qquad (19)$$

where σ^2 is the variance of the u. It can be seen from the equation that the PSD is completely defined by σ, F and the AR parameter a_k. In other words, only certain "forms" of PSD are possible.

For practical and theoretical purposes, it is useful to introduce the concept of linear prediction. Suppose, in some way, estimates $\hat{a}_k$ of the p AR parameters of a given AR signal have been found. Then, for any i, the linear combination, $\sum_{k=1}^{p} \hat{a}_k x(i-k)$, of p past samples can be calculated. This linear combination is called the linear prediction $\hat{x}$ of x for the following reason. If the estimated AR parameters are equal to the actual AR parameters, then the linear combination $\hat{x}$ is as close as possible to the actual x in the mean square sense. Any other linear combination would have a larger prediction error variance.

An outline of the practical approach is as follows. 1) Select a set of N samples, say N=50. 2) Make an educated guess of p. 3) Estimate the p AR parameters. This can be done by several techniques, as explained below. 4) If needed, adjust p, and go to step 3. This is really the order estimation problem, explained below. 5) If needed, adjust the N, and go to step 2. This is the sample size problem, explained below. 6) Verify that the signal is indeed an AR signal. This is the model validation problem. From this outline, one can see that several (closely related) subproblems have to be solved.

Estimation of the AR parameters, given the p and a set of N samples, is the simplest problem to tackle. Depending on the signal (*e.g.* normally distributed or not), or the context of the application (*e.g.* on-line or off-line), different algorithms can be opted for. Theoretically, all these methods are based on the minimization of the prediction error over all possible models with the given p. Available methods are, for example, the autocorrelation method, the covariance method, the modified covariance method, and the Burg method.

The order estimation problem is difficult and controversial. Suppose we have calculated for each of the possible p, the best AR model in the prediction error sense. A naive approach would be to select the order that results in the smallest prediction error variance. The caveat is that the prediction error variance will always decrease if the order is increased. The result would be the selection of N, as the "best" order. Several statistical order selection techniques have been proposed in the past. They all introduce a penalty for overestimation of the order.

The required sample size determination is not so difficult once the order is known. Indeed, basic rules of thumb can be applied for the statistical estimation of $p + 1$ parameters. For example, a sample size of N=50 for a third-order AR model would be quite sufficient.

The model validation problem involves verifying that the actual signal is indeed an AR signal. If it is not, the calculated results make no sense whatsoever. The verification can be based on the fact that the resulting prediction error signal should be a white noise (or purely random) signal. In other words, the autocorrelation function of the prediction error should be a delta function, and its PSD should be constant.

Specifications for respiratory sound analysis

Analysis of breath sounds

Somewhat varying frequency bands have been obtained in studies on normal breath sounds. Averaged spectra computed on tracheal sounds (inspiration and expiration phases) have shown that the log amplitude response curve remained approximately flat in the range 75–~900 Hz, before rapidly falling away at higher frequencies [25, 30]. Several other authors measured spectra of lung sounds on a linear plot with maximum amplitudes of 140 –200 Hz, followed by an exponential decay to insignificant levels at ~400 Hz [31]. Such differences are partly the result of using different representation of data, a linear scale being more likely to overaccentuate high-amplitude responses and underestimate weaker signals.

However, even though some investigators have measured an upper limit frequency as high as 3,000 Hz for tracheal sounds [32], it is commonly admitted that normal respiratory sounds contain components among which the most significant have a frequency of 50–1,200 Hz. The frequency spectra of tracheal sounds decline rapidly at >850–900 Hz. Due to muscle sounds and heart sounds [33], respiratory sounds are not usually studied at <50–60 Hz and the range 0–60 Hz should be filtered by a high-pass filter. Due to the dependence of breath sounds on airflow rate, respiratory sound spectra should be reported at a known airflow. Moreover, the frequency spectra at zero flow should be given in order to determine the background noise [34].

Relationship with flow rate. Sound spectra are clearly linked to respiratory airflow rate. Sound increases with the air flow, engendering a marked upward shift in frequencies [35] in the spectra. Other recent studies revealed that flow modifies both the intensity and the frequency distribution in the spectrum [32, 36, 37]. The relationship between flow and sound depends on many factors, including upper-airways configuration and, especially, the subject themselves. However, for every subject, the mean amplitude and the mean frequency are increasing functions of the flow [35, 38]. For this reason, it is possible to evaluate the flow approximately from the tracheal sound [38, 39].

Spectral parameters and their repeatability. A very common way to characterize a frequency spectrum is to divide it into parts, so that each part represents the same amount of energy. The fractions can be halves (medians), quarters (quartiles) or any percentage (percentiles) of the total spectrum energy [40]. The percentile, i, is determined in computing the frequency (f_i) limiting the specified portion. For instance, the median frequency (f_{50}) is the frequency dividing the power spectrum into two parts of equal energy. The quartiles f_{25}, f_{50} and f_{75}, which divide the spectrum into four equal parts are the most often used by investigators. They give an evaluation of how the spectrum is balanced between low and high frequencies, and, for this reason, can be used to characterize global changes in breath sounds.

The repeatability of the mean f_{50} obtained from one short-term recording session has been shown to be good [40]. The intra-individual coefficient of variation of f_{50} in healthy subjects of recordings with an interval of 15 min was 2.6–4.4%, and, with an interval of 1–3 days, was 5.0–8.5%. In patients with fibrosing alveolitis, the results were similar. The variation of power spectra of normal lung sounds has been shown to be small, even between subjects [41], although this might be due to the fact that this study was mainly interested in low frequencies.

Another tool for evaluating the power spectrum shape of breath sounds is to calculate regression lines in a log/log coordinates presentation. This technique is mainly used in average spectra of chest wall breath sounds for evaluating the decay of the spectrum at high frequencies. One or two regression lines can fit the decay of the spectrum in normal people. Spectral shapes displaying slope irregularities are often associated with lung pathologies. The slope of regression lines is thus an interesting frequency spectral feature. Usual values of the slope of regression lines are 12–13.5 dB·octave-1 for healthy people.

Crackles

Analysis. Since the bandwidth of the commonly encountered crackles is 100–2,000 Hz, a sampling rate of 5,512 Hz provides a sufficient frequency range (*i.e.* 0–2,700 Hz). However, the study of several fine crackles may require a wider range of analysis as they exhibit high-frequency components. Therefore, in this case, the use of a sampling rate of ≥11,025 Hz is recommended [18].

Visually, the timing of crackles in relation to the respiratory phase is conveniently illustrated using a condensed time-domain presentation "phonopneumogram" [42, 43]. Quantatively, this relationship may be characterized by calculating the start and end point of crackles as a percentage of the respiratory phase [44]. Mean values from at least five respiratory cycles should be reported. In patients with lung fibroses, the timing of crackles is well repeatable from breath to breath. To characterize waveforms of crackles (see fig. 1) the following measurements may be applied [7, 45]: 1) the initial deflection width (IDW), *i.e.* the duration of the first deflection of the crackle; 2) the two-cycle duration (2CD), *i.e.* the duration of the first two cycles of the crackle; and 3) the largest deflection width (LDW), *i.e.* the width of the largest deflection of the crackles.

In each of the parameters, low values correspond to the clinical concept of fine crackles, whereas large values correspond to coarse crackles. According to suggestions made by the American Thoracic Society (ATS) [46], the mean durations of IDW and 2CD of fine crackles are 0.7 and 5 ms, and those of coarse crackles 1.5 and 10 ms, respectively. This recommendation did not provide any description of the methods on which these figures were based. The present CORSA (Computerized Respiratory Sound Analysis) standards for characteristic waveforms of crackles assume that a standardized technique has been used according to CORSA guidelines [12]. According to CORSA definitions, the 2CD of fine crackles is <10 ms, and that of coarse crackles, >10 ms [29].

By using the timing and waveform characteristics of crackles, a two-dimensional discriminant analysis has been applied [47]. This approach may be useful when different lung diseases presenting with crackles are to be distinguished from each other.

Automatic crackle detection. The number of crackles has been associated with the severity of the pulmonary disease in fibrosing disorders [48]. Because of the limitations of human hearing in distinguishing between individual crackles, automatic counting methods based on computer-assisted analysis have been developed. The basic methods used in automatic crackle recognition can be divided into those based on TEWA, spectral analysis and AR models.

Murphy *et al.* [8] used criteria based on TEWA. The basic assumptions were that crackle complexes contain three to 16 baseline crossings, their amplitude is greater, than twice that of the background signal, the beginning of the event has a sharp deflection and crossings of the baseline after the initial deflection are progressively wider. It should be noted that some of the above criteria are dependent on the high-pass filtration used. Another approach to detect crackles in the time domain is to use the first derivative of the lung sound signal [49]. The method described by Kaisla *et al.* [50] was based on spectral stationariness in a sonogram matrix. In sonographic (time/frequency) presentation, crackles can be detected as narrow peaks. The sensitivity of this crackle detector was reported to be 80–89%. Examples of parameter estimation based methods used in crackle detection include adaptive nonlinear filters [51] and wavelet transformation [52].

Wheezes

Wheezes are pseudoperiodic signals characterized mainly by their pitch and duration.

Range of pitch. The pitch of a wheeze should be considered as the fundamental frequency of a pseudoperiodic signal. However, a wheeze is very often approximately a pure sinusoidal signal, so the loudest component is the fundamental frequency with only few harmonics, if any. The old ATS nomenclature defined a wheeze as differing from a rhonchus in the sense that the pitch of the former is >400 Hz, whereas the pitch of the latter is below that frequency. According to recent definitions [53] and the present CORSA definitions, the dominant frequency of a wheeze is ≥80–100 Hz, and that of a rhonchus is ≤300 Hz.

From a signal-processing point of view, these limits seem insignificant and arbitrary. In automated techniques for detecting "wheezes", the authors suggest, as in studies of musical tones, that sounds which have a determined pitch, and that are of near-periodic signal, are distinguished from those which do not have a defined pitch. There is no reason to try to set an upper limit for the pitch. So far, no wheeze has been reported with a pitch of >1600 Hz [54]. It is recommended, however, when studying wheezes, to use a sampling rate of ≥5,000 Hz [18].

For instance, figure 4 shows the Fourier analysis performed with a raw periodograrn estimator over 512 points. The signal is sampled at a rate of 11,025 Hz and decimated by a factor of 2, namely a sample over 2 is ignored. Thus, the effective sampling frequency is 5,512.5 Hz.

Duration of a wheeze. In the past, a minimum duration of 250 ms has been set by the ATS for a wheeze [46]. In automated systems of wheeze detection, some researchers think that wheezes duration could be as short as 30 ms. Even if it were possible that wheezing conditions could

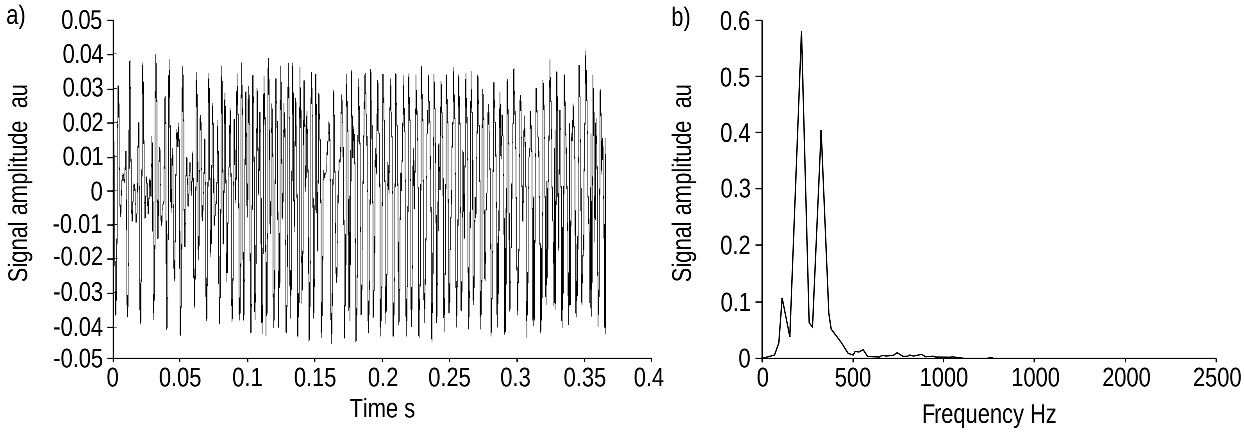

Fig. 4. – Example of : a) a wheeze segment; and b) its periodogram.

occur for such a short time, there is no evident interest in considering such an event as a wheeze. However, 250 ms seems too long, and, in systems of automatic recognition, shorter events can be detected and subjectively heard as wheezes. Therefore, the present CORSA guidelines define the shortest acceptable duration of a wheeze as 100 ms [29]. Similar minimum duration for a wheeze (80 ms) has been suggested recently [53] (see [29]).

Automatic wheeze recognition. In addition to the pitch and duration, it is necessary to determine an amplitude criterion to make possible the automatic detection of a wheeze. Automated analyses of wheezes are usually based on their spectral appearance, and they rely on identification of the presence of at least a peak with a sufficiently strong amplitude. In order for a peak to be accepted or rejected, it must be in the appropriate frequency range, as defined above, and have an amplitude greater than a threshold value, which can be either absolute or relative.

An absolute threshold needs a calibration phase of the automated system before a monitoring phase is started. Unfortunately, any calibration depends on both the patient and the operator particularly the latter. Conversely, a relative threshold can be determined with parameters computed simultaneously and is thus independent of the operator and more flexible. The authors recommend use of the latter. The total energy of the spectrum computed for a given duration is a good starting point for defining a reliable threshold.

Relevant measurements. Automated systems of wheeze recognition may bring the physician interesting statistical results that cannot otherwise be assessed. Among various measurements for quantification of wheezing, several researchers [55, 56] have proposed calculation of the duration of wheezes as a percentage of the respiratory cycle duration. Additional measurements can also be relevant. For instance, it is recommended that the number of wheezing episodes be computed in long-term recordings. A wheezing episode is an interval of time in which a wheeze is present over consecutive respiratory cycles. Computing the mean frequency of the wheezes, and labelling them respectively to the respiratory phase, *e.g.* the flow/volume curve [57], may be interesting as well.

Snores

Numerous studies concerning snores have been carried out, but are generally related to nocturnal breathing disorders rather than to the analysis of the snore itself. However, researchers, have already pointed out the most significant features and difficulties encountered with this type of sound. The first observation concerns the intensity of this signal. The snore can be much louder than any other respiratory sounds, and can also be weak in certain circumstances. Thus in order to avoid possible saturations, it is necessary to use sensors with a very large dynamic intensity scale. Sensors with a linear intensity response of ≥100 dB are recommended.

The frequency spectrum of snores also has specific characteristics compared to the spectra of other respiratory sounds. The snore is usually a pseudoperiodic signal with energy mainly concentrated in a fundamental frequency and its harmonics. The fundamental has been reported to be as low as 30 Hz and may be >250 Hz in some cases [58]. Thus in the high-pass filtering recommended for most lung sounds, the cut-off frequency of 60 Hz should be lowered to 25 Hz, or even lower, but the heart noise may then interfere with the snore signal.

The snore is produced in the upper airways, and the sound is modified by the resonating cavities of the open nose and mouth, which act as pass-band filters. For this reason, the frequency spectrum of a snore depends on whether the patient demonstrates nasal or oronasal snoring. In nasal snoring, the upper limit of the spectrum defined as the last peak maximum frequency (F_{max}) with a power >3% of the peak power is reported to be ~550 Hz [59]. For oronasal snoring, the same peak F_{max} is ~850 Hz. These limits can be increased in the case of obstructive sleep apnoea. When studying frequency spectra of snoring sounds, it is recommended that the range 30–2,000 Hz be considered, even if there is only a small amount of energy >1,200 Hz. The sonogram and the sound pressure level (SPL) should be reported.

Stridors

Many techniques have been used to study laryngo-tracheal sounds, taking into account the developments in signal processing. These works are of great interest but reveal the difficulties in analysing respiratory sounds. In 1980, HIRSCHBERG [23] used a spectrograph analyser to characterize various infant sounds. Thereafter, GRAY *et al.* [60] performed the first estimation of a computerized stridor spectra using fast FT algorithms, and several modelling methods were recently used by LEIBERMAN *et al.* [61] and SLAWINSKI and JAMIESON [62].

Fourier analysis. The technique of analysis used by GRAY *et al.* [60] is based on the combination of averaged and smoothed periodograms (see above). Eight consecutive stridor sequences selected in inspiration were computed to obtain their Fourier spectra. The spectra were then averaged, and the resulting spectrum was smoothed with a rectangular window (*i.e.* moving average). The signal was sampled at a frequency of 5 kHz, setting the analysed band to 0–2,500 Hz. The FTs were performed on sequences of 2,048 points (~0.4 s). Therefore, the resolution was about ~2.45 Hz·point^{-1}. With this analysis, a new tool is provided with which to analyse and identify respiratory sounds characterized by their spectral shape (number of peaks, peak features).

This usage of the FT should be viewed with caution because it makes assumptions regarding important characteristics of the signal: 1) reproducibility; and 2) stationariness during the 400-ms observation time.

Location of the stridor peaks. The commonly observed range for the pitch is 600–1,300 Hz [60, 61], usually ~1,000 Hz. In adults, the pitch of the stridor is usually much lower and is <200 Hz. Although it is quite difficult to provide information on the other peaks, it seems that they are much more flow/volume-dependent than the main peak. Thus, a sampling frequency of ≥5,512 Hz is recommended when studying the main peak of the stridor. In addition, if the interest is in the estimation of obstruction parameters, this frequency should be ≥11,025 Hz, according to the CORSA recommendation [18].

Duration. Stridor phenomena have variable characteristics, including the duration. The shortest duration may be observed after a cough.

Another variable important for the analysis is the adjustment of minimal duration, in order to reach stationariness of the observed signal. A duration of <30 ms can be sufficient, but the most accurate estimate of this variable can be obtained with a time/frequency method or a segmentation technique [63].

Spectrography. This method of analysis relies on the STFT. It displays spectra computed on consecutive segments to constitute a two-dimensional image or three-dimensional plot of the energy of a signal. With this method, phenomena can be analysed with greater accuracy. Complex dynamics of an event can be traced and several hidden properties highlighted. HIRSCHBERG [23] applied several spectrosonographic techniques to investigate infant respiratory sounds. In this work, spectrograms of cries and stridors were described. Nowadays, the same method can be executed in real time [24]. It is a useful tool by which to study qualitatively the properties of a signal.

An example of analysis of a stridor is given with two kinds of presentation in figure 5. The sampling frequency was 11,025 Hz. The spectrogram is displayed with a logarithmic scale for the amplitude. It is computed on segments of 256 points (of 23 ms), with an overlapping factor of 50% and a Hanning window. A different way to present the data is to plot the consecutive spectrograms in a waterfall presentation (fig. 5b). These maps of the analysis of a stridor in the time/frequency plane illustrate the problems encountered in PSD estimation and the limitation of the Fourier analysis.

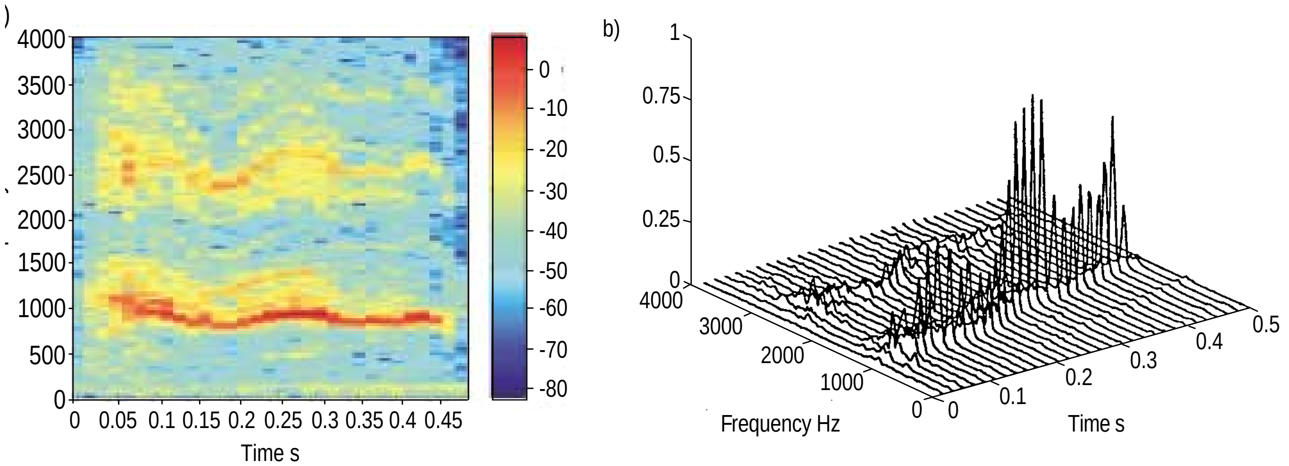

Fig. 5. – Analysis of a stridor event: a) spectrogram; and b) waterfall.

Table 1. – Summary of recommendations

Breath sounds	
	Methods: Periodograms, autoregressive models
Features:	Spectral slopes, quartile frequencies, octave band analysis
Presentation of result:	PSD plot, tables of parameters
Crackles	
Methods:	Time-expanded waveform analysis
Features:	Time parameters (IDW, 2CD, LDW), number of crackles and timing
Presentation of results:	Time plots, tables of parameters
Wheezes	
Methods:	Periodograms, STFT
Features:	Pitch location in frequency, wheeze duration and timing, histogram of wheezing episodes, mean frequency, balance between inspiratory and expiratory wheezes
Presentation of results:	PSD plot, sonogram plot, tables of parameters
Snores	
Methods:	Time-expanded waveform analysis, periodograms
Features:	Amplitude in time domain, main peak location and energy in frequency
Presentation of results:	PSD plot, tables of parameters, sound-pressure level, sonogram
Stridors	
Methods:	Periodograms: STFT autoregressive models
Features:	Pitch location, duration of the event, number and location of high-frequency peaks
Presentation of results:	PSD, plots, sonogram plots, tables of parameters

PSD: power spectral density; IDW: initial deflection width; 2CD: two-cycle duration; LDW: largest deflection width; STFT: short-time Fourier transform.

Presentation of the results

Graphical representations

In the time domain, respiratory sounds have to be presented graphically, amplitude *versus* time, on a linear scale and, if possible, with the flow [64]. The choice of time scale is not critical, except in the case of crackles, as indicated before.

In the frequency domain, the sound is represented by its average power spectrum (intensity *versus* frequency). The scale can be linear or logarithmic for the amplitude and linear or more rarely, logarithmic for the frequency, depending on what is being looked for. Because respiratory sound is a continuously varying signal, it is often useful to display consecutive short-time power spectra (obtained with or without overlapping) *versus* time in a three-dimensional coordinate system. In a sonogram, the representation is in the plane frequency *versus* time, the intensity being simulated by the blackness or the colour of the dots (see, for instance, fig. 5). Another way to represent the three dimensions intensity, frequency and time is the so-called "waterfall" display in which the successive power spectra are plotted in perspective (see fig. 5).

Presentation of measured and calculated parameters

For numerical values, the presentation of the results should appear as a table containing the items indicated in table 1. A column could be omitted *(i.e.* is optional) based on the relevance of the corresponding item. For instance, squawks are only inspiratory sounds, so only one column is needed to indicate a parameter.

The acronyms should be explicitly listed below the table, for example "COPD: chronic obstructive pulmonary disease". The authors recommend that only acronyms adopted by the ATS are used.

The authors also recommend that the way in which the confidence interval has been obtained is specified.

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Reporting results of respiratory sound analysis

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ABSTRACT: Breath sound studies performed in different laboratories have a variety of methodological differences in respiratory manoeuvres used as well as in the signal capture and processing techniques that are employed. It is a fundamental tenet of reporting scientific work that sufficient information about the techniques used and the assumptions made are provided in such a way that the work could be repeated in another research centre. Therefore, accurate and detailed reporting of the experimental procedures used for measurement and analysis of breath sounds are necessary both in scientific reports and in clinical documents. This paper gives recommendations for the contents of scientific and clinical reports, respectively.
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The check-list of items to be included in a scientific report is at the end of the paper.

Subjects and environmental conditions

Patients and subjects

The number of subjects, their sex and the selection criteria adopted have to be defined. The smoking history of the subjects must be described.

Environment

The environmental conditions, such as acoustic properties, frequency spectrum and intensity of background noise of the room must be indicated, especially if the signal-of-interest-to-noise ratio is less than 15 dB. If a soundproof room or some special sound insulation is used, those have to be specified in the report.

If noise-shielding of the microphones is applied, it has to be described.

The subject background noise-level spectrum during breath-holding should be reported, especially if the signal-of-interest-to-noise ratio is less than 15 dB. In cases where constant frequency peaks of background noise due to noise from external sources (*e.g.* from computer, air conditioning device, or lighting) have not been avoided, their frequencies have to be reported.

Experimental procedure

Body posture, the status of vigilance (sleep studies), and route of breathing during recording have to be defined. If mechanical ventilation is used, details of this have to be reported.

The breathing manoeuvre during sound recording must be described in detail. The feedback procedures of breathing pattern, *e.g.* monitoring of air flow or volume using a visual signal seen by the subject must be explained. Peak inspiratory and expiratory flow during recording must be reported.

The data collection has to be explained in detail including information on the duration of the samples or recording periods, as well as the number of inspirations and expirations recorded, and the actual time. The method of data collection, *i.e.* continuous or segmental, has to be reported, and the selection criteria for segmental sound samples have to be indicated. It should be mentioned whether the analysis has been made on-line or off-line.

Simultaneous physiological measurements

Air-flow and volume measurements. The method of measuring air flow and volume during sound recording must be described. Define the measuring equipment, type of the flow sensor and spirometer. It is important that the method of digitization of the signal is described. Also, the flow calibration procedure must be indicated. If a window for air flow is used, its characteristics have to be described.

Mouth pressure. The measuring equipment and the signal digitization must be explained.

Oesophageal pressure measurement. The method of recording and digitization must be explained.

Electrocardiography. The equipment and method of possible electrocardiographic gating must be reported.

Sound capturing

Microphone location and attachment

The location of microphones during sound recording has to be defined. The use of anatomic description of the pick-up locations is recommended [1]. Sometimes, it is reasonable to record the lung sounds on the location where the breath sounds or adventitious lung sounds are best heard. If the exact recording location has not been specified in the study protocol, the measurement of the microphone location during every single recording has to be defined. For example, indicate the distance of the microphone from the vertebral column, and from the shoulder or lower line of the scapula.

The mode of attachment of the microphones or sensors on or in the body has to be explained, *e.g.* adhesive tape fixation, belt fixation or hand-held sensor. Any attachment settings need to be defined.

Pick-up sensors

The type of sensor, *i.e.* contact sensor (accelerometer or deformable piezoelectric elements) or air-coupled acoustic sensor (condenser or piezoelectric microphone), as well as the manufacturer of the sensor have to be defined [2, 3]. Also, the sensitivity, dynamic range, frequency range and signal-to-noise ratio of the sensor must be indicated [1].

When an air-coupled sensor is used, the dimensions of the coupling volume, the free area on the chest wall under the air-coupling space, and the distance of the sensor from body surface must be reported.

The calibration procedure of the sensor and the frequency response in actual use have to be indicated, if possible.

If more than one sensor is used, a balancing procedure between the sensors has to be indicated.

Report also whether any analogue noise compensation within the microphone is used.

Sound recording

Analogue recording

The method of the magnetic tape recording (amplitude modulation, frequency modulation or other) and the linearity of the recorder have to be defined.

Tape speed during the recording and its frequency response range and linearity have to be indicated. Also, the low-frequency range should be described because some recorders may act as a high-pass filter [4]. The stability of the tape speed and the signal-to-noise ratio should also be indicated.

Report also which simultaneous physiological measurements are recorded: air flow, inspired and expired volume, mouth pressure or electrocardiogram.

The method of analogue-to-digital conversion, the sampling rate and period, and the dynamic range (in dB) have to be defined.

Digital sound recording

The sampling rate and amplitude resolution (dynamic range) of digitization should be indicated.

The digitization of parallel physiological data as air flow, mouth pressure, inspired or expired volume should be indicated.

Report the method of signal storage. Specify which data were stored and the mode of digital data storage. Specify the mode of signal compression, if used.

Amplification

Report the frequency range and linearity of response of the amplifier and the gain of the microphones. If more than one sensor is used, report how the amplifiers were balanced.

Sound filtering

The cut-off frequency of high-pass or low-pass filtering, *i.e.* the frequency at which the signal is reduced by 3 dB (the so-called 3 dB point), and the roll-off or the slope of filtering (in $dB \cdot octave^{-1}$) as well as the type of the filter have to be defined. Explain the properties of analogue (*e.g.* Butterworth, passive, *etc.*) or digital filters used in detail because the filtering type may significantly influence even the sound morphology concerning short-duration sounds [4].

Sound analysis

Data selection

Selection of the sound data for analysis must be reported, *e.g.* whole recorded sound or only a subset of sound samples. If a subset of sound samples are used, report the selection criteria, *e.g.* fractions of inspiratory or expiratory sound selected by some definite flow gate.

Timing of respiratory sounds

In breath-sound analysis, determination of timing of adventitious sounds is important. It should be reported how the time domain analysis is performed. Are the respiratory events counted and timed related to the whole duration of the respiratory cycle [5], related to the inspiratory or expiratory volume [6] or oesophageal pressure? If the occurrence of adventitious sound events is defined on the flow-volume plane, specify the method.

Sound waveform morphology analysis

Report the time scale. The direction of the initial deflection of a crackle may be the opposite in inspiration and in expiration [4, 7]; therefore, the direction of increasing and decreasing sound pressure must be specified at least in the figures.

Report also the criteria used in waveform morphology analysis. What are the criteria used for crackles [8] and wheezing sounds [9]? Define the quantitative parameters used.

Report the criteria for the onset [8] of a crackle or a wheeze. If automatic detection is used, which criteria and methods are used to detect crackles or wheezes?

Frequency domain analysis

Define the window used, *e.g.* Hanning, Hamming, rectangular, Blackman, as well as number of points used in the fast Fourier transform analysis. The sample size and the possible averaging of spectra have to be explained.

Presentation

Presentation of results has been dealt with in [10].

Graphical presentation

The graphical monitoring during recording and analysis have to be defined. The resolution of time, frequency and amplitude presentation of the graphics used in scientific reporting have to be defined. The expansion used in the time-expanded wave-form display has to be indicated.

Parametric analysis

The parameters and their measured values from the lung sound analysis should be reported as related to the anatomy and location. Also, average values and variations of the parameters have to be calculated. It has been shown that average values of power spectral parameters are very repeatable [11].

Normal chest wall sounds. Maximal frequency (f_{max}) in Hz (the frequency of maximal amplitude in the spectrum), total power (P_{total}) (the intensity of the sound of interest) in dB [12], the quartile frequencies in the power spectra (F75, F50, F25) [12, 13]. If slope and deflection point [12] are measured, report them.

In addition for tracheal sounds. The frequency range and major peaks have to be measured.

In addition for wheezes. The number of wheezing peaks, their maximal frequencies, duration, timing in respiratory cycle and location of recording on the chest wall or trachea should be indicated if possible.

In addition for crackles. The following durations of the time-expanded waveform of crackling sounds have to be indicated: the initial deflection width [14]; two-cycle duration [14] and largest deflection width [15]. In addition, the timing of crackles related to the respiratory cycle, their number, distribution, and possibly effect of posture, coughing or deep inspirations on their occurrence have to be defined.

Quality assurance

Refer to the standard publications or references. Details of specific analysis methods or statistical methods have to be indicated. The variances of measured variables have to be given, *e.g.* the variances of averaged spectra.

Clinical report

The results of lung sound data recorded and analysed with respect to an individual patient are given as a clinical report containing both graphical and numerical data. It should contain the information that is needed for patient identification, as well as the recording and analysis set-up needed for a clinician. The clinical report may also serve as the basis for collection of scientific data.

Patient data

In every single recording, anthropometric data must be known about each patient.

The preparation of the patients before the test should be indicated: smoking, eating, drinking and especially drug intake. Which drugs are used, and when was the last dose taken? The detailed knowledge of treatments with asthma drugs is of great importance.

Experimental procedure

The essentials about sound recording have to be defined, *e.g.* the recording site and the position of the subject (supine/sitting).

The gain of the microphones has to be indicated if the gain is going to be fitted to each patient recording individually.

The breathing patterns and the maximum flows and volumes during breathing have to be defined.

Table 1. – Check-list for a lung sound recording and analysis report for research

Subjects	Number of subjects Sex Diagnostic criteria Selection criteria Anthropometric data Drugs and smoking
Environment	Spectrum of background noise level of room, free field Subject background noise, spectrum during breath-holding
Experimental procedure	Body posture Route of breathing Sensor location Breathing manoeuvres Number of respiratory cycles recorded
Air flow and volume	Sensor Equipment Calibration Digitization
Sound signal capture	Sensor type Sensitivity Frequency response Dimensions of air-coupling volume Kind of venting of sensor Calibration of the sensor
Analogue recording	Magnetic tape (FM/DM) Linearity of recorder Tape speed Frequency range
Digital recording	Sampling rate Dynamic range Sample size Method of signal storage Number of bits
Signal processing	Frequency range of amplifier Gain of amplifier Amplifier balancing Filter type applied in sound channel High-pass cut-off frequency Low-pass cut-off frequency Roll-off of filtering (in $dB \cdot Oct^{1}$)
Sound analysis	Data selection Conditions of analysis
Graphical presentation	Scales with units of time, frequency and amplitude axes
Numerical presentation	Maximal frequency Sound-pressure level Presence of adventitious sounds (number per respiratory cycle) Quartile frequencies or slope Wheezing: number of wheezing peaks, duration, timing, frequencies Crackles: number per cycle, timing, distribution, crackle duration, IDW, 2CD, LDW (ms).

FM: frequency modulation; DM: direct mode; IDW: initial deflection width; 2CD: two-cycle duration; LDW: largest deflection width.

Graphical presentation

At least some samples of the phonopneumogram (sound amplitude, flow display) should be presented graphically. This gives a visual impression of the signal-to-noise ratio of the recording and the presence or lack of adventitious sounds. Frequency-time presentation (sonogram) of sound is informative, especially on the adventitious sounds. If crackles are present, examples of crackle waveforms should also be presented.

Numerical results

The maximum frequency of the power spectrum of the sound, frequency of wheezing and the timing of adventitious sounds should be reported. If quartiles [12, 13], slope or deflection points [12] of the spectrum are measured, report them. The (mean) duration of inspiratory and expiratory phases and the maximum inspiratory and expiratory flow must be indicated. If flow-volume recording is used, the flow-volume curve should be described. Report also the duration of crackle indices, IDW, 2CD, LDW, duration of crackling and the period of timing of crackling by indicating the beginning and end point of crackling as a percentage of the duration of the whole respiratory cycle.

Table 2. – Check-list for lung sound recording and analysis for a clinical report

Patient data	Identification code Age, sex, height, weight Previous lung function study results Clinical diagnosis or clinical problem Indication for lung-sound analysis
Preparation of the patient for the test	Smoking, eating, drinking Drug intake
Experimental procedure	Microphone locations Posture of the patient Breathing manoeuvres Target inspiratory and expiratory flow and volume Gain of the sound channel
Graphical presentation of results	Examples of representative phonopneumograms Examples of frequency/time presentations (sonogram) Examples of power spectra Examples of single crackles Flow-volume curve
Numerical presentation	Mean quartile frequencies of inspiratory and expiratory phases Mean durations and number of inspiratory and expiratory phases Mean maximum inspiratory and expiratory flows Mean sound pressure level (RMS) in expiratory and inspiratory phases Mean and SD of crackle variables (IDW, 2CD, LDW) Mean and SD of the timing variables of adventitious sounds Mean relative crackling and wheezing periods Frequency of wheezing peaks Spectral slope data (optional)
Clinical interpretation	

RMS: root mean square of sound pressure; IDW: initial deflection width; 2CD: two-cycle duration; FM: frequency modulation; DM: direct mode; IDW: initial deflection width; 2CD: two-cycle duration; LDW: largest deflection width

Summary of recommendations

In reporting scientific work on lung sounds, the description of the experimental procedure and methods for recording and analysis is important because of methodological differences in different laboratories. Essential data to be reported in scientific work as well as clinical work are dealt with and collected as check lists (tables 1 and 2).

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Future perspectives for respiratory sound research

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Future perspectives for respiratory sound research. J.E. Earis, B.M.G. Cheetham. ©ERS Journals Ltd 2000.
ABSTRACT: Modern technology may now be used to great advantage for the capture, storage, analysis and communication of sounds that are normally heard through a stethoscope. State-of-the-art digital signal processing equipment is already being used for such purposes in clinical medicine, but the advantages of computerized sound analysis remain to be fully explored. This paper aims to give some perspective on how modern digital signal processing technology is likely to be exploited for clinical use in the future. Applications, including diagnostic evaluation, remote monitoring and data exchange by Internet, are considered. It is concluded that the limiting factor is no longer the power, storage capacity, capability or speed of readily available personal computers. The further development of sound acquisition techniques, signal processing algorithms and software packages that exploit existing technology effectively will be the key to further progress.
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Although respiratory sound analysis by computer has seen major innovations over the past 30 yrs [1–4] it has not, as yet, found a major place in clinical medicine. It is self-evident that modern technology offers immense advantages with respect to the capture, storage, analysis and communication of sounds that are normally heard through a stethoscope, but it remains to be conclusively established how these facilities can routinely and effectively be employed to aid the day-to-day diagnosis and management of patients with respiratory diseases.

Among the commonly reported applications of computerized respiratory sound analysis are the graphical presentation of features of importance [4–6], the making of permanent records of such features, correlation of respiratory sound with other physiological signals (particularly airflow at the mouth) [5, 7], comparison of data obtained at different times during the progression of respiratory diseases or their treatment [8], long-term monitoring of asthma [9, 10], monitoring of breathing patterns of infants [11], monitoring of respiratory sounds of adults in critical care settings [12] and detection of features and patterns that are not easily recognized by the human ear [13–15]. A perspective on the future for computerized sound analysis can be gained by considering its potential for these applications.

Technical perspectives

One of the problems that is limiting the wider use of lung sounds is the technical difficulty of capturing sound from the surface of the body. No sensor is ideal, and it is only in recent years that the responses of microphones when attached to the chest in various ways have been extensively analysed [16]. Small air-coupled electret microphones in light-weight housings are commonly used, particularly in Europe. Contact accelerometers are used more often in North American centres (see Current Methods used for Computerized Respiratory Sound Analysis in this issue). Ideal sizes and shapes for air-coupled microphone housings have recently been proposed [17], taking into consideration that the frequency responses obtained will be very different from what would be obtained from the same microphones in free field. It is clear that the sound will be strongly attenuated at higher frequencies, *i.e.* above about 1 kHz [5], and the loss of this higher frequency information is considered by some researchers to be a considerable disadvantage. Contact sensors are fragile, have a limited bandwidth and exhibit internal resonances at frequencies within the bandwidth of interest for lung sounds. Many technical problems of signal capture from the surface of the body remain to be solved. An ideal sensor should be of minimal mass, as wide a bandwidth as possible and be relatively insensitive to environmental noise. Arrays of such sensors may be connected together to allow acoustic mapping of the chest wall. Innovations such as laser microphones, piezo-electric membranes and highly directional microphone elements should, in future, provide more efficient methods of capturing lung sounds.

The problem of environmental background noise is of particular importance in normal clinical environments. Clearly, its effect must be reduced as far as possible by acoustic shielding and the use of suitable sensors. Newer digital signal processing techniques, employing adaptive digital filtering by computer or microprocessor, are being successfully applied [18] to further reduce the effect of background noise and other interfering signals. The reduction of interfering sound from intercostal muscles, chest-wall movement and the heart can be more difficult to achieve without significantly disturbing the waveforms that are of interest. This interference is predominantly at low frequencies, *i.e.* below about 100 Hz, and is often so high in amplitude that saturation of the analogue-to-digital conversion will occur resulting in non-linearity of the amplification process. A high-pass analogue filter, with a cut-off frequency of between about 70 and 100 Hz is commonly used to attenuate this interference before the microphone signal is amplified and digitized [19]. Alternative strategies for reducing the heart signal include ECG gating where sounds are only recorded during diastole [20]. Once the signal has been digitized without amplifier saturation, "active" cancellation of various types of interference or artefacts, *e.g.* heart sounds, may be achieved by means of adaptive digital filters [18]. Artefact recognition and cancellation may in future become an in-built and routine part of the sound capture and digitization process.

A survey of the technology and methods used over recent years for the capture, pre-filtering, digitization and analysis of lung sounds, undertaken as part of the CORSA project, reveals that a variety of approaches have been adopted with relatively few comparisons or discussions given of advantages and/or disadvantages of the different approaches (see "Current Methods of Lung Sound Analysis" in this journal). Recommendations have been made by CORSA for acquisition and digitization techniques for respiratory sounds, which will not only guarantee signals with an acceptable standard of fidelity for research work within an individual centre, but also enable data from different centres to be freely exchanged, pooled and compared. Acoustic analysis applied to lung sounds is a rapidly developing field, and rigid standardization must not be imposed on the research community in such a way that the development of new techniques is inhibited. However, it is vital for the development of the research area that acceptable standards of documentation and repeatability be maintained in reports and research publications.

The processing power of readily available equipment is now so great that it is rarely a limiting factor in the field of lung sounds analysis. Computations that may have taken minutes or hours on desktop computers some years ago are now performed apparently instantaneously. For example, spectrographs can be readily produced in real time for respiratory sounds [21] in a frequency range up to, and beyond, 4 kHz. In future, even highly sophisticated analysis and monitoring procedures will most likely be implemented in real time, eliminating the tedious process of recording signal files and processing selected portions in batch mode.

Current analysis techniques have recently been reviewed by Sovijarvi *et al*. (see "Characteristics of breath sounds and adventitious sounds" in this issue) and others [4–6]. One important development is the widespread use of flexible signal processing tools, such as those provided by the MATLAB software package, allowing signal processing algorithms to be conveniently tried out before they are incorporated into dedicated analysis packages or hardware. The capacity and convenience of storage media have increased rapidly in parallel with the huge increases in processing power. Inexpensive magnetic disks are now capable of storing data representing enormous quantities of respiratory sound, to an extent unimagined by most researchers a few years ago. Limitations of available storage capacity are unlikely to be major factors in the future development of respiratory sound analysis.

The impact of the Internet on future developments in respiratory sounds analysis should not be underestimated. It is a vehicle for the exchange of software, databases and sound and video files. It is also a platform for remote monitoring and a powerful educational tool. Already, websites exist that provide information about respiratory sounds. Two such websites are the International Lung Sounds Association (ILSA) site (http://www.ilsa.cc/) and the R.A.L.E. site (http://www.rale.ca/), which supply details of current world-wide research in lung sounds and related areas, a comprehensive database of published papers, details of how to contact others interested in this field and links to other research sites involved in acoustic analysis. The R.A.L.E. site also exploits multi-media communications technology to provide examples of respiratory sound signals that may be downloaded, listened to by any computer user and used by researchers as representative data. This development surely signposts immense possibilities in the future for the use of Internet technology.

One goal of current technological developments is to combine processing power, storage, miniaturization of components and analysis programmes into a small hand-held computerized stethoscope that will provide the clinician with much more useful information than the current simple mechanical stethoscope.

Clinical applications

Analysis by computer is more likely to be successful in clinical applications when it is based on mathematical models of the underlying physical mechanisms of respiratory sound production. Such models allow the interaction of mechanical forces, airflow and sound transmission within the respiratory tract to be understood and related to different disease processes The following clinical applications have benefited from this approach.

Diagnosis based on breath sounds

There is much evidence that breath sounds originate predominantly in larger airways (inspiration in lobar airways and expiration in more proximal airways) [22] and thus, when heard on the chest wall, they contain information about their origin and the way they have been transmitted through the lung. The fact that, even in normal individuals, their amplitudes and frequency spectra vary between different chest wall sites [23, 24] is further evidence that they contain regional information. Moreover, it is now known from phase-delay measurements and air-density studies that the sound transmission at lower frequencies (below 300 Hz) is mainly due to parenchyma wave propagation, while higher frequency sounds are propagated more through airways [25]. Thus, newer instrumentation capable of acquiring higher-frequency sound may provide more information about different structures within.the lungs than was possible previously, using more restricted bandwidths. Clearly, the intensity and frequency spectrum of breath sound are related to the airflow measured at the mouth [23, 26]. A series of recent experiments, using bronchial challenge, showed that even in the absence of wheeze, significant changes in the intensity and frequency spectra of breath sounds occur with fairly small changes in forced expiratory volume in one second (FEV_1), *i.e.* changes of as little as 10% [13, 15]. In a population study, Gavrielly *et al*. [27] showed that changes in lung sounds spectra can be used as a means of detecting the early stages of airways disease. The future holds the prospect of using acoustic mapping of the surface of the chest wall to noninvasively measure regional ventilation and airflow obstruction within the lungs. This prospect is of particular importance in the treatment of children and the evaluation of the pulmonary functions of patients in critical care settings. Monitoring abnormal breath sounds may also reveal a variety of other applications; for example, bronchial breathing could be monitored in the treatment of pneumonia.

Breath sounds acquired at the trachea are higher in intensity, observable over a wider frequency range and easier to capture than sounds from the chest wall [4, 5]. There is a strong correlation between airflow, sound intensity and mean frequency [28]. Measurements of intensity and spectral shape have already been used to provide a noninvasive indication of airflow in sleep studies [8] and as the basis of apnoea monitoring [29]. There is also evidence that such measurements may be of value in diagnosing and monitoring tracheal obstruction, and it should be possible to use tracheal breath-sound analysis as a means of indicating the occurrence of structural and dynamic changes in the upper airway [30]. Finally, the analysis of breath sounds detected simultaneously at both the trachea and the chest wall should be further investigated as the two signals will contain different, but complimentary, information that may be useful in studying underlying pathophysiology in the lung.

Diagnosis based on adventitious lung sounds

When acoustic analysis was first applied electronically to lung sounds, wheezes and crackles seemed to be the phenomena most likely to yield useful clinical information [1, 3, 4]. Wheezes had been the most common type of adventitious sounds investigated for diagnostic purposes using the stethoscope, and it was natural to seek ways of applying computer analysis to improve our understanding of these sounds and their significance. Fundamental studies of the underlying mechanisms of wheeze suggested that "flutter" in bronchi is largely responsible, although there is still debate about some aspects of this underlying mechanism [31, 32]. Computational models based on the theoretical equations governing airflow in floppy tubes have been proposed and implemented in software as a means of increasing and testing our understanding of likely sound production mechanisms.

Clearly, audible wheezes can be easily detected and analysed by computer, and procedures such as event recognition, spectral analysis, classification and counting require only straightforward [5] software to be implemented. This software becomes slightly more complicated when quieter wheezes need to be extracted from the accompanying breath sounds. The software can be used for long-term monitoring of asthmatic patients, particularly at night, to assess the severity of symptoms and the efficacy of treatments. Wheeze analysis can provide information about the degree of airflow obstruction, *e.g.* by determining what proportion of each expiration cycle contains wheeze [9]. Such analysis may give early warning of the "silent chest" in acute severe asthma. The monitoring of adventitious wheezing sounds has a place in the treatment of adults and is likely to be of even greater use in the assessment of babies and infants who cannot perform respiratory function tests such as peak-flow measurements.

Crackles, the other main type of adventitious pulmonary sound, have also been subjected to extensive investigation [3, 4, 33] by various means. Using a stethoscope, these sounds can be detected, categorized as "fine" or "coarse" and observed as occurring at particular phases of the inspiratory or expiratory [34] cycle. These attributes of crackles have been found to correlate with different disease processes and are routinely used to assist in clinical diagnosis [35]. With modern technology, it is now possible to analyse crackles to determine their attributes more accurately and objectively. Identifying crackles and counting the average number of crackles within each breath cycle to a reasonable degree of accuracy [36, 37] is an ideal task for a computer. The use of linear and non-linear digital filters, artificial neural networks and newer methods of parametric spectral analysis should enable even more accurate automatic recognition and characterization of crackles [38]. Crackles heard at the surface of the chest are likely to come from lung tissue adjacent to the microphone and hence represent local pathology involving the opening of smaller airways [4]. Using mapping procedures similar to those employed in the science of seismology, the sources of crackles may be localized within the lung parenchyma. In asbestosis, the number of crackles per breath cycle has been found to be indicative of the severity of disease [39]. Similar relationships may also be present in other interstitial processes. Thus, there may be scope for monitoring interstitial disease, particularly if surface mapping becomes a practical proposition. Such monitoring could be used, for example, for assessing the effect of treatment on fibrosing alveolitis and monitoring the progression of asbestosis. In hospitals, monitoring of heart failure and pneumonia by crackle counting [40] and mapping is also on the horizon, though the problem of reducing background noise must be solved to make this practical. Similar approaches may be adopted for monitoring the development of pulmonary oedema or interstitial lung disease in critical care settings. New algorithms for identifying and measuring adventitious lung sounds are being developed in various research centres and will affect the future of medical practice in many ways.

Diagnosis based on upper-airway sounds

The analysis by computer of upper airway sounds for medical diagnostic purposes has been an area of active research for many years and has been the subject of many publications over the past five years [41]. These sounds, which include speech and speech-like signals, snoring, stridor and cough, are amenable not only to acoustic analysis as used traditionally in medicine for lung sounds but also to advanced signal processing techniques, as applied in speech technology (recognition, compression, modelling, *etc.*) applied to the rapidly developing fields of telecommunications [30].

Sleep apnoea syndromes occur in up to 5% of the population, and diagnosis depends on monitoring multiple physiological signals (polysomnography). It is expensive to fully investigate a sufferer, and alternative diagnostic techniques are needed. Acoustic differences between post-apnoeic and simple snoring sounds have been identified electronically [42], and this may be useful for assessing patients and studying the fundamental nature of these phenomena.

Clinically relevant acoustic analysis of snoring has been confined mainly to measurements of duration and intensity and is possible using several types of commercially available monitoring equipment [43]. Wave-form analysis using sonograms, has been difficult to relate to clinical observations [44]. It is known that upper airway surgery as a treatment for simple snoring alters the frequency spectrum of the sound produced but is not always successful in reducing its incidence or loudness, especially in the long-term. Theoretical models of upper airway function have been proposed for understanding the sound generation mechanism [45]. Research in this area may be expected to lead to routine methods of using acoustic analysis to diagnose sleep-related syndromes and monitor the effect of treatment such as surgery and continuous positive airway pressure (CPAP).

The characteristics of the sound produced by different forms of cough have been studied by many researchers [46–48] and instrumentation measuring the rate of coughing has been proposed for monitoring the effectiveness of treatments, particularly in children [49]. There is evidence that differences in the characteristics of coughing from one pathological condition to another will be measurable and show a useful degree of consistency [50]. The clinical value of such observations needs further evaluation.

The analysis of speech and speech-like sounds has long been used as a means of detecting disease or abnormalities in the vocal cords. Abnormalities caused by acute infective laryngitis, chronic non-specific laryngitis, benign and malignant tumours, vocal-cord paralysis and laryngeal myopathy secondary to inhaled steroids may be investigated and monitored in this way. Such analysis has also been proposed as a convenient means of assessing the susceptibility of patients to sleep apnoea and snoring. Fortunately, such analysis benefits from a vast amount of knowledge and processing expertise gained from applications in the speech technology and telecommunications industry.

Remote monitoring

Remote monitoring in medicine is an active and expanding field. Telemetry is commonly used to monitor cardiac rhythm disturbances in ambulatory patients recovering from myocardial infarctions. Remote monitoring of other physiological signals in patients at home or at work has many clinical, economic and social benefits. It is especially applicable to conditions such as sleep apnoea and "brittle" asthma. Simple sound acquisition equipment and a means of transmitting data *via* fixed or mobile telephone, possibly *via* the Internet, has many possible uses in this area. Unintelligent collection of signals could result in vast amounts of useless data, so intelligence is needed to decide what to preserve and when to transmit it. The intelligence could be based simply on sound intensity measurements though more sophisticated decisions would probably be needed, based, for example, on wheeze detection, breath-sound quality as an indication of airflow obstruction, episodes of snoring, coughing and the noninvasive detection apnoea episodes. Once a segment of sound has been captured, the remote intelligence can have the capability of carrying out a further analysis in non-real time to decide whether the data really do have only features of interest.

Portable equipment for remote monitoring would use the technology developed for portable computers, the Internet and mobile phones [50] and can now be mass-produced at reasonable cost. The cost of transmission can be made relatively cheap for fixed telephone links and is falling rapidly in the mobile field. The more extensive use of respiratory sound monitoring, especially within critical care environments, is therefore becoming more likely with a greater understanding of the information within these sounds.

Education

There is great potential for the results of research and development discussed in this paper to be used in the education of medical students and trainees. There is equal scope for the continuing education of qualified staff who use auscultation in the day-to-day evaluation of patients. Remote teaching and learning, virtual universities and continuing education, *via* modern communication media such as the Internet, are becoming major initiatives in many spheres of activity, including medicine, nursing and allied professions. The universal availability of powerful personal computers with high-quality multi-media equipment linked to the Internet [52, 53] (and private intranet systems as being proposed for health services) makes it straightforward for educational establishments to distribute educational material in the form of sounds, pictures, animation and software as well as accessible supporting text. As electronic commerce in all fields develops, the commercial marketing of educational packages will become more and more routine. On-line tutorials, "virtual seminars", coursework, assessment and examinations will be offered *via* the Internet or similar electronic means. Markets for such educational material will be worldwide, thus giving great incentive for educational establishments to develop high-quality, stimulating and exciting courses. In the future, the majority of higher educational establishments in most disciplines, including medicine, may serve mainly to provide supportive environments, whereby students study courses prepared by a small number of establishment specializing in course preparation. Courses using multi-media facilities to demonstrate the principles of auscultation would be a small, but interesting, part of this overall scenario. The advantages of such an approach to the study of lung sounds have already been demonstrated by the material made available on the Internet by ILSA (http://www.ilsa.cc/) and R.A.L.E. (http://www.rale.ca/).

Conclusions

It is clear that digital signal processing techniques and modern communications may be applied to study lung and upper-airway sounds as an aid to clinical diagnosis. Compared with the ubiquitous acoustic stethoscope, a computer equipped with sound acquisition equipment offers subjectivity, quantitative results in graphical form, long-term storage, instant communications and many other advantages. Disadvantages such as increased cost and physical size are disappearing with the reducing cost of computer hardware, portable computing and mobile telephony. Useful investigations may be performed using standard personal computers, as installed in most doctors' surgeries, and can therefore be of particular value to primary care physicians who do not have easy access to sophisticated diagnostic equipment. There may be extra advantages in devising computer-based examination procedures specifically for children and babies who may be uncooperative with standard procedures. Computerized analysis is ideally suited to the long-term monitoring of patients either in hospital or in the community, and should also be of value in less-developed countries and remote communities. This paper mentions some of the areas where the computer analysis of respiratory sounds is known to be of clinical value. The most likely areas for early exploitation are the study of upper-airway problems in patients with sleep apnoea and heavy snoring, cough and wheeze monitoring, the evaluation of lower-airway dynamics including the recognition of early manifestations of airway disease and the use of remote monitoring of interstitial and airway disease (particularly in a critical care environment). An exciting prospect for the future would be the routine availability of a miniaturized portable apparatus with the ability to capture both sound and airflow, implement simple and clinically useful analysis packages and, when necessary, communicate data *via* mobile telephony to a specialist centre in a local hospital. This could be mass-produced as a multipurpose "computerized stethoscope" and may replace the current acoustic stethoscope as a basic tool for future doctors.

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Appendix: Abbreviations and acronyms for terms involved in computerized respiratory sound studies

A.R.A. Sovijärvi, F. Dalmasso, J. Vanderschoot, L.P. Malmberg, G. Righini, S.A.T. Stoneman

A

A	ampere (unit of electric current intensity)
AC	alternating current
ADC	Analogue-to-digital converter
AI	apnoea index
AM	amplitude modulation
ANSI	American National Standards Institute
AR	autoregressive
ARMA	autoregressive moving average
ASCII	American Standard Code for Information Interchange
ATP	ambient temperature and barometric pressure
ATPS	Ambient temperature and barometric pressure saturated (with water vapour)

B

BMI	body mass index
BPT	bronchial provocation test

C

CD	compact disc
CDR	compact disc recordable
CD-ROM	compact disc-read only memory
COPD	chronic obstructive pulmonary disease
CPAP	continuous positive airway pressure
CSA	central sleep apnoea
CT	computed tomography

D

DAC	digital-to-analogue converter
DAT	digital audio tape
dB	decibel
DC	direct current
DFT	discrete Fourier transform
DSP	digital signal processor

E

ECG	electrocardiogram
EEG	electroencephalogram
EMG	electromyogram
EOG	electro-oculogram
EPP	equal pressure point
ERS	European Respiratory Society
ERV	expiratory reserve volume
$ETCO_2$	end-tidal carbon dioxide

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F

f	frequency
FEV_1	forced expiratory volume in one second
FET_b	forced expiratory time at a specified portion b of the FVC
FEW	forced expiratory wheeze
FFT	fast Fourier transform
F_{I,O_2}	inspiratory oxygen fraction
FIR	finite impulse response
FIV_1	forced inspiratory volume in one second
FIVC	forced inspiratory vital capacity
FM	frequency modulation
FRC	functional residual capacity
FVC	forced vital capacity

H

H	height

I

IC	inspiratory capacity
IDW	initial deflection width (crackles)
I:E	inspiratory-to-expiratory ratio
IFT	inverse Fourier transform
IIR	infinite impulse response
I/O	input/output

K

kg	kilogram (unit of mass)

L

LDW	Largest deflection width (crackles)
L_{eq}	equivalent sound level
LPC	linear prediction code
LSA	lung sound analysis

M

M	(m) mass
m	meter (unit of length)
MA	moving average
MDU	metered-dose inhaler
MMEF	maximal mid-expiratory flow
modem	modulator/demodulator
mol	mole (unit of quantity of substance)
MSA	mixed sleep apnoea
MVV	maximal voluntary ventilation

N

N	newton (unit of force)

O

OSA	obstructive sleep apnoea
OSAS	obstructive sleep apnoea syndrome

P

P_{A,O_2}	alveolar oxygen tension
P_{a,O_2}	arterial oxygen tension
P_{a,CO_2}	arterial carbon dioxide tension
PEEP	positive end-expiratory pressure
PEF	peak expiratory flow
PEFR	peak expiratory flow rate
PIF	peak inspiratory flow
P_{oes}	oesophageal pressure
PPG	phonopneumography
PSD	power spectral density

R

RAM	random-access memory
Raw	airway resistance

REM rapid eye movement

ROM read-only memory

RMS root-mean-square

RV residual volume

S

s second (unit of time)

SA sleep apnoea

S_{a,O_2} arterial oxygen saturation

SAS sleep apnoea syndrome

sGaw specific airways conductance

SI International System of Unity

SNR (S/N) signal-to-noise ratio

SOM self-organizing map

SPL Sound pressure level

S_{p,O_2} arterial oxygen saturation measured by pulse oximetry

STPD standard temperature pressure dry

STFT short-time Fourier transform

T

$T_{L,CO}$ transfer factor of the lung for carbon monoxide

TEW time-expanded waveform

TGV thoracic gas volume

TLC total lung capacity

TV tidal volume

2CD two-cycle duration (crackles)

V

VC vital capacity

W

W watt (unit of power)

W weight

Z

Z impedance