The 17th International Conference on Lung Sounds
Helsinki, Finland
Program

Monday, August 24

Workshop

15.00 – 18.00	Needs for standardization of lung sound analyzing methods Prof. Robert G. Loudon and Dr. Raymond L. H. Murphy, Chairmen
19.30	**Transportation to Welcome party (from Market Place)**

Tuesday, August 25

8.45 **Welcome address – Dr. Anssi R. A. Sovijärvi**

Session A
Lung Sounds and Bronchial Provocation
Drs. David Cugell and Hans Pasterkamp, Chairmen

9.00 – 9.20	Variability of lung sound frequency spectra during histamine challenge test in healthy adults	Malmberg
9.20 – 9.40	The influence of methacholine challenge on lung sounds in normal and asthmatic subjects using airflow-standardized phonopneumography	Schreur
9.40 – 10.00	The relationship between wheeze, airflow and lung volume during methacholine induced bronchoconstriction in asthmatic subjects	Spence
10.00 – 10.20	Variable response to histamine in children with cystic fibrosis	Beck
10.20 – 10.40	Bronchial provocation and forced expiratory wheeze	Ishikawa
10.40 – 10.55	*Coffee break*	

Session B
Tracheal Sounds
Drs. Filiberto Dalmasso and John Earis, Chairmen

10.55 – 11.15	Tracheal vs. glottic and supraglottic stridor	Schaefer
11.15 – 11.35	Tracheal sound spectra depend on body height	Sanchez
11.35 – 11.55	Effect of gas density and breathing apparatus on tracheal sounds	Pasterkamp
11.55 – 12.15	Variation in wheeze characteristics over multiple chest sites	Davidson
12.15	Poster viewing	
12.45	*Photo*	
13.00	*Lunch*	

14.30 – 15.15 **Invited lecture A**

Chairman – Dr. Wilmot Ball

Title – Mechanisms of wheezing
Presenter – Prof. Noam Gavriely

Session C
Crackles and Squeaks
Prof. Toivo Katila and Dr. Päivi Piirilä, Chairmen

15.15 – 15.35	Frequency analysis of crackling lung sounds recorded through a stethoscope	Kiyokawa
15.35 – 15.55	Crackle sound intensity in the auscultation of crackling sounds in fibrosing alveolitis and heart failure	Piirilä
15.55 – 16.15	Importance of auscultatory crackles in lymphangitic pulmonary metastases	Lower
16.15 – 16.35	Breath sounds analysis in pulmonary screening at work	Gavriely
16.35 – 16.50	*Coffee break*	
16.50 – 17.10	Squeaks as a diagnostic finding in lung disease	Baughman
17.10 – 17.30	Squeaklets in BOOP	Loudon
17.30 – 18.30	**Business meeting**	
19.30	**Transportation to Banquet (from hotels)**	

Session D
Signal Capture and Processing
Prof. Gerard Charbonneau and Dr. Steward Stoneman, Chairmen

8.30 – 8.50	Polynomial autoregressive models of normal lung sounds	Vanderschoot
8.50 – 9.10	Computational methods for asymmetry analysis of lung sounds	Sovijärvi
9.10 – 9.30	An algorithm for the detection of respiratory sound crackles	Gueler
9.30 – 9.50	Mathematical formulae for pulmonary diagnosis using lung sounds	Murphy
9.50 – 10.10	Performance of air-coupled and contact sensors in lung sound measurements	Pasterkamp
10.10 – 10.25	*Coffee break*	
10.25 – 10.45	The bronchofiberscope used as an acoustic probe for the endobronchial sounds	Dalmasso
10.45 – 11.05	Speed of sound propagation in inflated sheeps lungs and normal men	Gavriely
11.05 – 11.25	Feasability and utility of nocturnal monitoring of breath sounds and pulse oximetry	Postiaux
11.25 – 11.45	A comparison of computer-based learning (CBL) and small group teaching of pulmonary auscultation to first year medical students	Mangione
11.45 – 12.30	**Invited lecture B** **Chairman – Dr. Anssi R. A. Sovijärvi** **Title – New approach to pathophysiology of asthma** **Presenter – Prof. Lauri A. Laitinen, Helsinki**	
12.30	*Lunch*	

Scientific session E
Pathophysiology of Respiratory Sounds
Prof. Noam Gavriely and Dr. Shoji Kudoh, Chairmen

14.00 – 14.20	Lack of low frequency power in lung sounds of newborn infants compared to adults	Powell
14.20 – 14.40	The acoustical analysis of coughs	Stoneman
14.40 – 15.00	The spectral characteristics of sound transmission in the human respiratory system during HFJV	Chen
15.00 – 15.20	The characteristics of tracheal breath sounds during HFJV	Chen
15.20 – 15.40	*Coffee Break*	

Poster Session

Wednesday, August 26

15.40 – 17.15
Space for posters is 90 cm width x 150 cm height.
Five minutes will be allocated for each presentation, with a maximum of 2 slides.

Dr. Raphael Beck and Dr. Emile Lens – Chairmen

P1	Auscultation of fine crackles using a stereophonic stethoscope	Matsui
P2	Comparison of lung sound parameters between normal and asthmatic subjects using airflow-standardized phonopneumography	Schreur
P3	Snoring sounds recorded by air-coupled microphone and by a new piezo-electric sensor	Schaefer
P4	Comparison of spectral density of breath sounds by fast Fourier transform and Burg's estimator	Gonzales
P5	The frequency content of snores in man	Spence
P6	Noise reduction in respiratory sounds by adaptive filtering in time and frequency domains	Aljama

Dr. Sadamu Ishikawa and Dr. Juergen Schaefer – Chairmen

P7	Lung sound classification experiments by using self-organizing maps	Kallio
P8	Comparison of methods for automatic classification of respiratory sounds	Rajala
P9	The 'Ultrafine' crackles	Dalmasso
P10	Respiratory sounds during metacholine challenge	Giordano
P11	Prevalence of bedmate reported snoring and sleep apnea in a male population aged 40 – 65 yrs	Sacco
P12	A computerized system for lung sound analysis with versatile spectral analyses and automatic crackle detection	Sovijärvi
P13	On the sound transfer function of respiratory system in human	Xu
P14	A new method for analysis of lung sounds	Xu
P15	A study on wheezes models	Xu

17.30 **Closing remarks and Dr. Dalmasso's award of the best poster – Chairmen of the Steering Committee: Prof. Robert G. Loudon and Dr. Raymond L. H. Murphy, Jr.**

ABSTRACTS

VARIABILITY OF LUNG SOUND FREQUENCY SPECTRA DURING HISTAMINE CHALLENGE TEST IN HEALTHY ADULTS

Malmberg P*, Sovijärvi A*, Paajanen E†, Piirilä P*, Kallio K†, Katila T†

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† Laboratory of Biomedical Engineering, Helsinki University Hospital, 02150 Espoo, Finland

Breath sounds of 6 healthy non-smoking adults were recorded simultaneously at the trachea and at the right lower lung lobe area during a dosimetric histamine challenge test with four dose steps. Four to six respiratory cycles were recorded at each step, and the average frequency spectra of breath sounds corresponding inspiratory and expiratory phase were calculated using an overlapped segment averaging method introduced by Welch. The following spectrum variables were determined: the breath sound intensity (RMS), the frequency of maximum intensity (Fmax) and the upper frequency limits for first (F25), second (F50) and third (F75) quartile of spectrum energy. None of the subjects responded significantly to histamine: the median decrease of FEV1 was -6.5 % (range from -1 to -9 %). The within-subject variability of each frequency spectrum variable was determined by one-way analysis of variance (30 observations). Expressed as coefficient of variation (CV), the variability of F50 at the chest was 2.5 % (inspiration) and 2.8 % (expiration), and at the trachea 8.7 % and 6.8 %, respectively. RMS was the most and F75 the least variable parameter. All the frequency spectrum parameters were more variable when recorded at the trachea than at the chest. By multiplying the observed CVs with 1.65, 95 % confidence limits for normal variability of frequency spectrum parameters during histamine challenge test can be obtained; these can be applied as limits for a significant change during a challenge test when adult asthmatic patients are investigated using lung sound spectra.

THE INFLUENCE OF METHACHOLINE CHALLENGE ON LUNG SOUNDS IN NORMAL AND ASTHMATIC SUBJECTS USING AIR-FLOW-STANDARDIZED PHONOPNEUMOGRAPHY. HJW Schreur, MC Timmers, R Schot, J Vanderschoot#, JH Dijkman, PJ Sterk. Dept. of Pulmonology, University Hospital of Leiden, #Dept. of Medical Informatics, University of Leiden, The Netherlands.

There are morphological differences of the airway walls between normal and asthmatic subjects. However, it is still unclear whether these morphologic properties are of importance in the generation of lung sounds during airway narrowing in asthma. Therefore, we examined the relationship between induced airway narrowing and lung sound changes in 9 normal and 9 asthmatic subjects. All subjects underwent phonopneumography at baseline and at $\geq 20\%$ fall of FEV_1 by methacholine challenge. Together with airflow and lung volume changes, lung sounds were recorded at 3 locations during standardized quiet breathing, and during maximal forced manoeuvres. Airflow-dependent power spectrums were computed using FFT, and submitted to analysis of lung sound intensity (**LSI**), quartile power points (Q_1-Q_3), and the ratio of wheezy spectrums to the total number of spectrums ($W_{\%}$). The results were statistically analyzed by ANOVA. **LSI** was weaker in asthmatic subjects than in controls ($p \leq .001$), and increased with increasing airways obstruction ($p \leq .010$). Q_1-Q_3 were higher in asthmatic than in normals for the two microphones over the lung base ($p \leq .022$). During airway narrowing, at quiet breathing, $W_{\%}$ was higher in asthmatics than in normals ($p \leq .026$), and increased with the level of obstruction ($p \leq .017$). During forced expirations $W_{\%}$ was not dependent on subject group or level of obstruction ($p \geq 0.111$).

In conclusion, the intensity and pitch of lung sounds, and the extent of wheezing are different between normal and asthmatic subjects at given levels of induced airway narrowing. This suggests that airflow standardized phonopneumography reflects differences in airway morphology between asthmatics and controls.

Supported by a grant of the Netherlands Asthma Foundation.

THE RELATIONSHIP BETWEEN WHEEZE, AIRFLOW AND LUNG VOLUME DURING METHACHOLINE INDUCED BRONCHOCONSTRICTION IN ASTHMATIC SUBJECTS.

DPS SPENCE, DR GRAHAM, G JAMIESON, PMA CALVERLEY AND JE EARIS
Aintree Chest Centre, Fazakerley Hospital, Liverpool

Wheeze is a frequently sought sign of airflow obstruction, yet little quantitative data is available about its relationship to lung volume and airflow obstruction. We have analysed breath sounds in 8 stable wheeze-free asthmatics [FEV1(SE) 66(22)% pred] during an extended methacholine challenge continued until audible wheeze occured or significant symptoms developed. FEV1, Raw and FRC were measured at each methacholine concentration in a body plethysmograph, symptoms by a modified Borg scale and breath sounds by a microphone over the left upper chest for off line analysis. Breath sound power spectra were derived using a Fast Fourier Transform technique, and were displayed with flow and volume on an IBM compatable PC. During the challenge FEV1 fell by 51(14)%, Raw increased by 119(50)% and FRC by 34(15)% from baseline. Audible wheeze developed abruptly late in the challenge, and was present during tidal breathing in 5 cases and confined to inspiration in one. Tidal expiratory wheeze began at 769(97)Hz, the frequency falling to 575(108)Hz by end expiration. Wheezing frequency was reproducible within subjects (CV starting frequency 4.25%, ending frequency 12%). Inspiratory wheezes occured around peak inspiratory tidal flows whereas expiratory wheezes developed after flow had peaked (58% peak tidal flow). Expiratory wheeze began 0.8(0.38-0.81) and ended 0.18(0-0.4)L above FRC whilst inspiratory wheezes started at 0.21-0.61 and ended 0.86-1.1L above FRC. Subjects percieved chest tightness before wheezing began mean fall in FEV1 at the onset of symptoms being 17(19)% compared to 53(7)% for wheezing. These data demonstrate that wheeze frequency is very reproducible but in asthmatics severe bronchocostriction can occur before wheezes develop. Perception of chest tightness is a more sensitive pointer to worsening lung function. This has important implications for clinical assessment.

Variable response to histamine in children with cystic fibrosis

Raphael Beck, M.D., Mark D. Montgomery, M.D., Ian Mitchell, M.B.
Dept. of Pediatrics, U of Calgary; Alberta, Children's Hospital, Calgary, Canada

Background: Airways hyperreactivity (AHR) in CF patients has a reported incidence of 24-68%. A number of studies suggest that the mechanism of AHR might be different in CF and asthma. To further investigate this question, we studied the lung sounds pattern of CF patients in response to histamine.

Method: Children with CF who were able to perform spirometry consistently, underwent histamine challenge by the tidal breathing of continuously nebulized histamine method, adapted from D. Cockroft. Patients inhaled histamine for 2 1/2 minutes, followed by spirometry and lung sounds recording. Histamine concentration was doubled every 6 minutes, starting from 0.03 mg/ml, to a maximum of 16 mg/ml. The challenge was terminated when FEV_1 fell by > 20% or when symptoms appeared.

Lung Sounds: Were recorded from 3 sites: Trachea, right upper anterior chest (RUL), and right lower posterior chest (RLL), using a HP21050A contact sensor. Sounds were amplified (x10 - x1000), band-filtered (50 - 2000 Hz) and digitized (8 bit A/D converter) at 5.5 KHz into a Macintosh SE computer, using SoundWave software. Expanded wave-form and spectral analysis was performed on the recorded sounds off-time.

Results: Six male CF patients with mild (2) or moderate (4) lung disease, were studied (table). Three (50%) showed severe or moderate AHR, two (33%) had mild AHR and one (17%) had no AHR by PC_{20} criteria (table).

Age/Sex	FVC (% pred)	FEV_1 (% pred)	FEF_{25-75} (% pred)	PC_{20} (mg/ml)
7/m	87	95	94	> 16.0
10/m	77	69	42	0.06
11/m	59	58	44	8.0
12/m	70	56	23	8.0
15/m	71	58	30	2.0
16/m	90	82	62	2.0

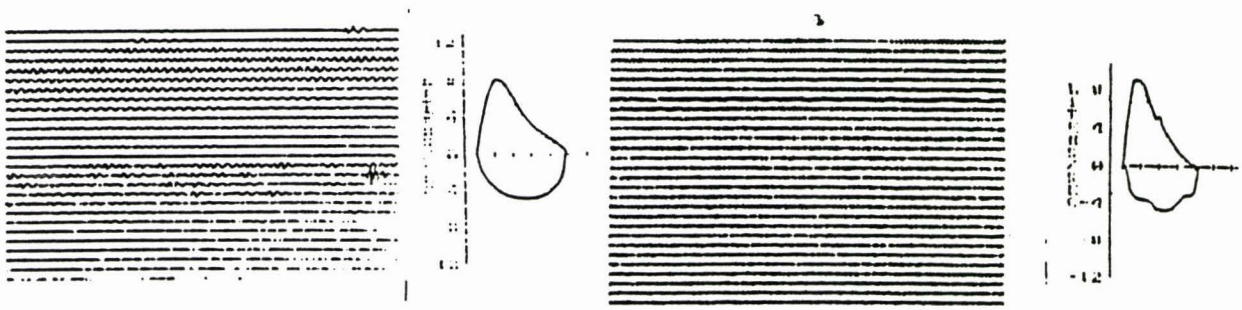
Of the five subjects who reacted to histamine, two had wheezing, two crackles and one became tachypneic, but with no detectable lung sound abnormality on expanded wave-form and spectral analysis.

Speculation: Different mechanisms of reactivity to histamine may exist among CF patients. Some have a typical asthmatic reaction producing wheezing, while others respond with excessive secretions resulting in crackles. The tachypnea could be a result of V/Q mismatching.

BRONCHIAL PROVOCATION AND FORCED EXPIRATORY WHEEZE

S. ISHIKAWA, H. BEAUCHAMP, L. KENNEY, R. ARANSON, J. ALLARD, S. McNALTY, M. KRASHIM & K. F. MACDONNELL
TUFTS LUNG STATION, ST ELIZABETH'S HOSPITAL OF BOSTON AND DEPT. OF MEDICINE, TUFTS UNIV. SCHOOL OF MEDICINE, BOSTON MA.

PREDICTABILITY OF THE BRONCHIAL HYPERREACTIVITY WAS TESTED IN 30 CONSECUTIVE METHACHOLINE CHALLENGE TESTS IN ADULTS. WE PERFORMED SPIROMETRY (FLOW VOLUME CURVE)AT BASELINE AND EACH DOUBLING DOSE OF METHACHOLINE UNTIL THE CONCENTRATION PRODUCED MORE THAN 20 % FALL IN FEV1 OR THE END POINT (25MG/ML) WAS REACHED. TRACHEAL AND LUNG SOUNDS WERE RECORDED SIMULTANEOUSLY DURING EACH SPIROMETRIC MANEUVOR. SOUND SIGNAL WAS DIGITIZED AND REAL TIME SPECTROGRAPH DISPLAYED USING A FAST FOURIER TRANSFORM SPECTRUM ANALYZER.
PRESENCE OF FORCED EXPIRATORY WHEEZE(F.E.W.) AT BASELINE DID NOT PREDICT LEVEL OF PD20 OR THE PRESENCE OF BROCHIAL HYPERREACTIVITY. THOSE WHO HAD EARLY APPEARANCE AND HIGHER MEAN POWER AND FREQUENCY OF F.E.W. TENDED TO HAVE LOWER PD20. THERE WAS NO CORRELATION NOTED BETWEEN LEVEL OF FEV1 AND DEVELOPEMENT OF F.E.W. OR WHEEZING IN THE CHEST.
THE SOUND SPECTRA AND AIRWAY CALIBER RELATIONSHIP REMAINS ELUSIVE.



TRACHEAL VS. GLOTTIC AND SUPRAGLOTTIC STRIDOR

J.W. Schäfer, M. Stojcic, P. Bonfils

ENT-clinic Universit of Ulm, Prittwitzstr. 43, D-7900 Germany

According to Cotton inspiratory stridor generally is supposed to be of supraglottic origin, in- and exspiratory stridor is due to extrathoracical tracheal origin and exspiratory stridor has its origin in the intrathoracical trachea. To test this hypothesis recordings of stridor in 11 patients presenting with stridor of tracheal origin, in six patients of glottic origin and in six patients of supraglottic origin were made using an electret condenser microphone at the suprasternal notch attached to a Sony WM 6 DC tape recorder. All recordings were digitized and analysed with an A/D-converter in a 486 Personal Computer using commercially available digital signal processing software (Stemmer, ILS). Obstructive sites in all patients were verified either by endoscopy or by CT scans or both. Signal analysis was in the time and frequency domain.

Independent of obstructive site, inspiration in most patients was louder than exspiration. Timing of inspiration vs. exspiration showed a large variation in all three groups. Frequency spectra of in- vs. exspiration showed only small differences in patients with tracheal lesions while spectral differences were larger in patients with stridor of supraglottic origin.

We conclude that the respiratory phase of stridor is not a good indicator for the obstructive site.

Tracheal sound spectra depend on body height.

Sanchez I, Powell RE, Pasterkamp H. Dept. of Pediatrics and Child Health, University of Manitoba, Winnipeg, Canada

We had previously reported an upward frequency shift of tracheal sounds during helium breathing in healthy adults (Pasterkamp et al., ARRD 1990;141:A709). Our findings suggested an effect of resonances in upper and central airways on tracheal sounds. Since tracheal dimensions are a function of body height, we hypothesized that children have higher frequency components than adults. To test this hypothesis we recorded tracheal sounds at standardized air flows in 21 healthy children, ages 9.1 y ± 0.6 (mean ± S.E.) and in 24 healthy adults, ages 30.2 y ± 0.8. A contact sensor (EMT25C, Siemens) was attached at the supra-sternal notch of the sitting subject, and airflow was measured at the mouth with a calibrated pneumotachograph. Tracheal sounds were low-pass filtered at 2.4 kHz and digitized at 10 kHz. A 2048 point FFT was applied at a successive 100 ms intervals, using a Hanning data window. Resulting spectra were normalized to a reference power of $0.1(mV)^2/5Hz$. We applied a gating algorithm to extract sounds at inspiratory flows of 1 l/s (±10% tolerance) and computed average power spectra from the collected samples. We calculated the average spectral power (P_{avg}), the quartile frequencies below which 25% (Q_1), 50% (Q_2) and 75% (Q_3) of the power in the 50-2000 Hz range was contained, and the spectral edge frequency (SE_{95}) below which 95% of the power was found. We also calculated the frequency where spectral power rolled off sharply (F_{cut}). We found that children had significantly louder sounds (P_{avg} 27±1 vs 17±1 dB) and higher frequencies (Q_1 365±10 vs 196±11 Hz, Q_2 747±18 vs 431±23 Hz, Q_3 1127±22 vs 711±33 Hz, SE_{95} 1627±27 vs 1028±51 Hz). Figure 1 shows the relation of F_{cut} to body height and Fig.2 displays average spectra for subjects grouped by body height. We conclude that this strong correlation of spectral characteristics and body height is further evidence that tracheal sounds are influenced by resonances in central airways.

Figure 1

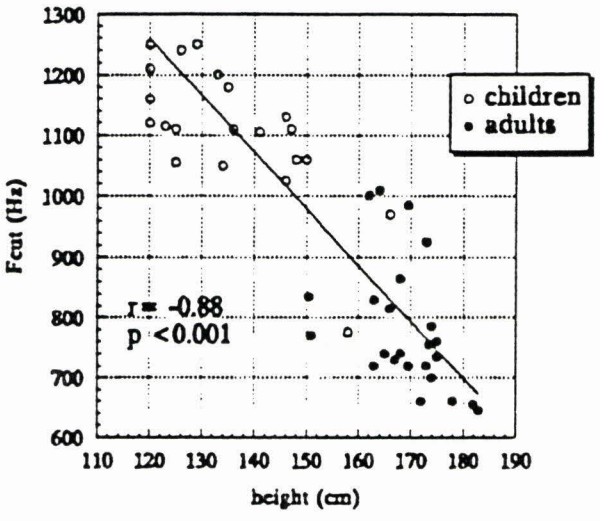
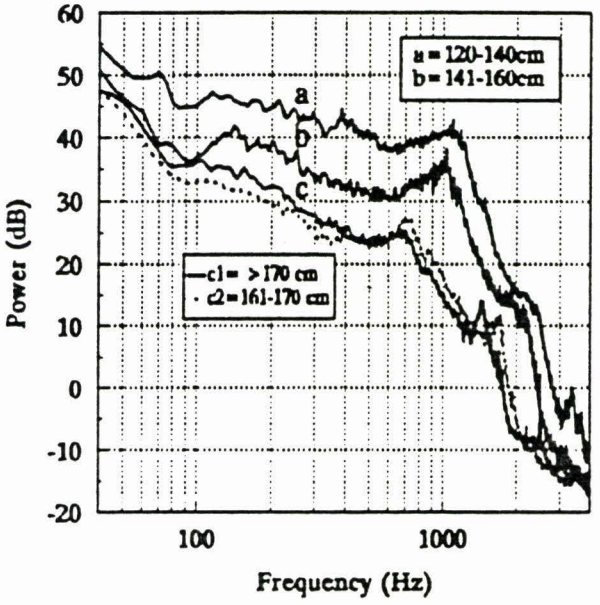


Figure 2



Effect of gas density and breathing apparatus on tracheal sounds.

Pasterkamp H, Sanchez I, Oh Y. Dept. of Pediatrics and Child Health, University of Manitoba, Winnipeg, Canada

We have previously reported an upward shift of frequencies in tracheal sounds on heliox (80% helium/20% oxygen) and we have considered an effect of resonances in central airways on tracheal sounds. Since formant frequencies of speech are approx. ½ octave higher in heliox (Sergeant RL, Aerospace Med 1963;24:826), we predicted a similar 50% shift of spectral peaks in tracheal sounds. To test this hypothesis, we studied 6 healthy male subjects, ages 30 to 33 y. They first breathed air at target flows of 2 l/s (±10% tolerance) through a calibrated pneumotach (Fleisch #3), and then breathed air followed by heliox at the same flows from a 12 l rolling seal spirometer. Sounds were recorded at the suprasternal notch with contact sensors (EMT25C, Siemens and PPG sensor, Technion, Haifa) and low-pass filtered at 2.4 kHz to avoid aliasing. Flow and sound signals were digitized at 10 kHz with a 12 bit A/D board (Data Translation 2831g). A 2048 point FFT was applied at 100 ms intervals, using a Hanning window. Resulting power spectra were averaged from samples within target flow range. Figure 1 shows average signal (solid = 2 l/s) and noise (dotted = zero flow) spectra from one subject breathing through the pneumotach. The two sound sensors gave similar results. Peaks at approx. 340, 700, 1500 and 2350 Hz are apparent. Figure 2 shows signal spectra from the same subject, breathing from the spirometer. A peak at approx. 180 Hz appears (a) and is approx. ½ octave higher on heliox. The peak at 700 Hz (Fig.1) now has an adjacent peak at 600 Hz (b). Both shift approx. ½ octave higher on heliox. The peak at approx. 1.5 kHz was most consistent and found in all subjects (mean 1.5 kHz ± 0.1 S.D.), shifting on heliox by 47% ± 10 (to 2.2 kHz ± 0.1). We conclude that tracheal sounds show resonances that likely originate within central airways. However, effects of breathing apparatus are obvious and have to be considered in studies of tracheal sounds.

Figure 1

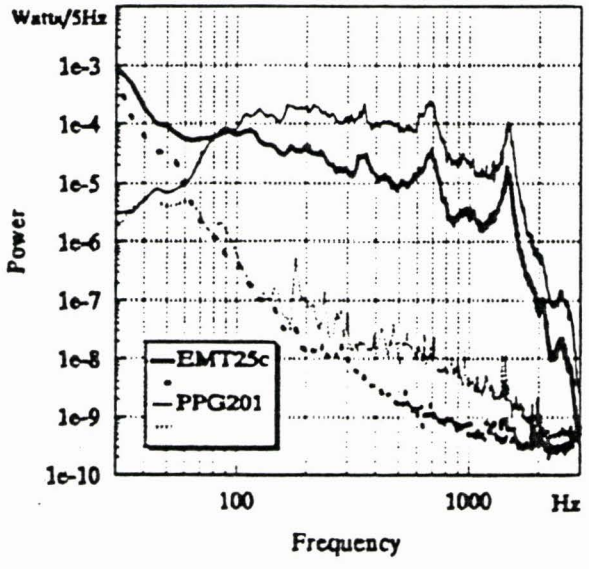
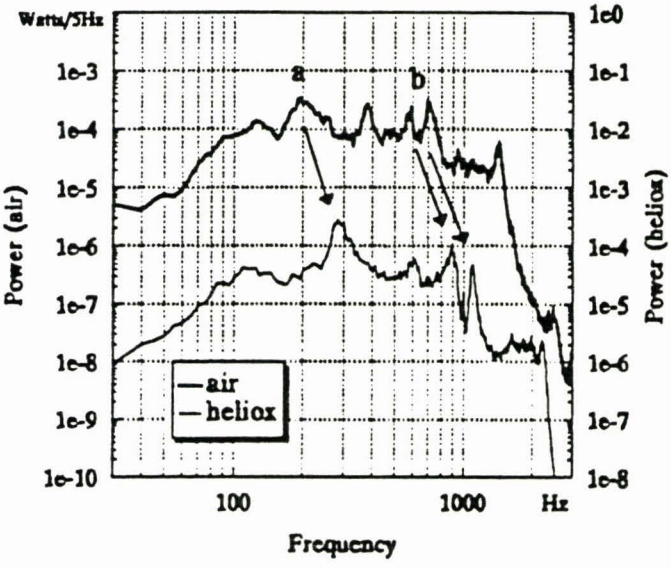


Figure 2



VARIATION IN WHEEZE CHARACTERISTICS OVER MULTIPLE CHEST SITES.

Davidson F.and Murphy R.

Little information exists on the waveform characteristics in the time domain and the frequency characteristics of wheezes in different areas over the chest wall. To study this, we simultaneously recorded lung sounds using 4 microphones placed in a variety of configurations over the chest wall of a patient with wheezy bronchitis. For the first set of recordings, a center microphone was placed at the right base posteriorly with 3 microphones placed radially 1" from this center microphone. Subsequent recordings were made after moving the peripheral microphones 2", 3" and 4" from the center. In addition, we placed 6 microphones around the chest in the following locations: right and left upper and lower chest posteriorly and right and left lateral chest. Simultaneous recordings were also made at these sites.

We examined the waveform patterns, frequency characteristics and time of differences of these patterns in tracings from each of the microphone sites. At a separation of 1", an early expiratory loud wheeze and a lower intensity end expiratory wheeze were similar in each microphone. As the microphones were placed farther apart, it became more difficult to identify these patterns. The differences in these parameters will be presented together with speculations as to the possible underlying mechanisms.

New approach to pathophysiology of asthma

Lauri A. Laitinen

FREQUENCY ANALYSIS OF CRACKLING LUNG SOUNDS RECORDED THROUGH A STETHOSCOPE

Hiroshi Kiyokawa, Makoto Yonemaru, Shinobu Horie, Yuichi Ichinose, Keisuke Toyama
First Department of Medicine, Tokyo Medical College, Tokyo, Japan.

Despite progress in analytical measures for the lung sounds recorded through a microphone, most physicians use stethoscopes to detect abnormalities in lung sounds and differentiate fine from coarse crackles. This study aims to characterize the crackles recorded through a commonly used stethoscope. For the frequency analysis of the crackles, we used the fast Fourier transform (FFT) and maximum entropy method (MEM). A microphone was attached to the ear piece of a stethoscope (Littmann® Cardiology Stethoscope). The lung sounds and signals for the respiratory phase were recorded on DAT. These signals were digitalized by a 12-bit analog to digital converter. All crackles detected in one inspiration period were subjected to the frequency analysis with a range from 0 to 2 kHz. The peaks in the MEM spectrum estimated from a crackle were determined and classified into two groups: Type I; peak>800 Hz (-), Type II; peak>800 Hz (+). Coarse crackles by auscultation were recorded from four patients with bronchiectasis. The percentage of Type I:Type II were 92%:8%, 78%:22%, 86%:14%, and 100%:0%, respectively. Fine crackles by auscultation were recorded from four patients with interstitial pneumonitis. The results were 0%:100%, 24%:76%, 38%:62%, and 23%:77%, respectively. These results seem compatible with our previous report that the fine crackles contain 1 kHz spectral peak, but coarse crackles do not (1). It is speculated that the characteristic frequency components for discrimination between coarse and fine crackles remain in the crackles recorded through a stethoscope. Due to stethoscope's prevalence, the application of stethoscope as a pickup may be relevant for ordinary purpose in lung sound analysis.

1. Yonemaru, M. et al. Proc. of 10th International Conference on Lung Sounds. 1985; 15.

Crackle sound intensity in the auscultation of crackling sounds in fibrosing alveolitis and heart failure

Piirilä, P,MD, Kallio K, BSc, Katila T, DSc, Paajanen E, Rajala, H-M, MSc Rosqvist T, MSc, Sovijärvi ARA, MD.

When crackling lung sounds are auscultated in heart failure, they often give the impression of fine crackles, although in sound waveform analysis they usually are the most coarse ones. The auscultatory finding in pulmonary diseases is influenced by the subjective properties of hearing, the properties of the stethoscope and the properties of the sound to be auscultated. The aim of this study was to compare the intensity of the crackle sounds in fibrosing alveolitis (N=10), which typically shows fine crackles, and in heart failure (N=10), in which the author has found the crackles to be the most coarse in crackle sound analysis.

The crackles were recorded at the back basal region of the lungs with the most numerous auscultatory crackle findings. The lung sound signal was prefiltered with a 50 Hz (-3dB) passive high-pass filter. The signals were digitized with a 12 bit AD-converter, and the sound signal was filtered (high-pass 95 Hz/24 dB/oct and low-pass 5kHz/24dB/oct.). According to the crackle parameters, the crackles in cryptogenic fibrosing alveolitis were typically fine crackles (IDW 1.3 ± 0.2ms, 2CD 7.7 ± 1.3 ms, LDW 1.9 ±0.3 ms) and in heart failure coarse ones (2.1 ± 0.3 ms , 11.8±1.3 ms and 2.9 ±0.3 ms, respectively). The differences in these parameters between the patient groups were highly significant ($p<0.001$). The RMS values of the same crackles were 1.12 ± 0.5 V and 0.8 ± 0.35 V, respectively; the difference was not statistically significant.

Two observers (general practitioner, GP, and pulmonary specialist, PS) listened to the patient recordings, and had to define the crackles as fine or coarse. According to GP seven patients with fibrosing alveolitis were considered to have coarse crackles, and eight patients with heart failure to have fine crackles. According to PS nine patients with fibrosing alveolitis had coarse crackles and six patients had fine crackles. The observer agreement was 60%. The crackles considered by GP to be fine had a mean RMS value of 0.8 ± 0.35 V, and as coarse a value of 1.07 ± 0.59 V, though the difference was not statistical. The crackles considered by PS as fine had a mean RMS value of 0.90 ± 0.41 V and coarse a value of 0.98 ± 0.41 V. Both observers considered the same six recordings of patients with fibrosing alveolitis to have coarse crackles, five recordings of heart failure patients to have fine crackles, and one recording with heart failure to have coarse crackles. The mean RMS value of the crackles considered as fine was 1.24 ± 0.54 V, and that of the ones considered coarse was 0.84 ± 0.48 V, respectively, though the difference was not significant.

The results show considerable observer error, but they are in concordance with previous studies on observer variability. In the present study the impression of coarseness or fineness heard in the recorded lung sound was often the opposite to that in the waveform analysis. There was some indication that the crackles that were fainter in intensity could be heard as finer than those with louder intensity. These observations were made on the recorded sound signal; with the stethoscope the sound intensity difference could be suspected to influence even more the subjective impression of crackling lung sounds. Factors influencing the auscultation of pulmonary sounds need more attention.

IMPORTANCE OF AUSCULTATORY CRACKLES IN LYMPHANGITIC PULMONARY METASTASES.

Elyse E. Lower and Robert P. Baughman
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In order to identify the significance of auscultatory findings in patients with lymphangitic pulmonary metastases, we studied patients with pathologically confirmed lymphangitic pulmonary spread. The presence of crackles was compared to underlying histologic disease, patient symptoms, and chest roentgenograms. Since we had previously shown that increased lymphocytes in the bronchoalveolar lavage (BAL) of breast cancer patients with lymphangitic metastases had a better prognosis, we also compared auscultation to BAL findings. Ten patients with lymphangitic pulmonary metastases were identified. Four of ten patients had lymphangitic spread from breast cancer, three patients had unknown primary carcinomas, and one patient each had bladder cancer, prostate cancer, or melanoma. All patients experienced dyspnea, and nine of ten patients noted cough. Chest roentgenogram findings revealed localized infiltrates in four patients and diffuse infiltrates in six patients. Six patients had >10% lymphocytes noted in the BAL fluid, with normal BAL lymphocytes <10% in our laboratory. Auscultatory crackles were detected in six of ten patients. There was no difference in symptomatology or underlying disease between the patients with crackles and those without crackles. All four patients without crackles had increased BAL lymphocytes, while only two of six patients with crackles had increased BAL lymphocytes (Fisher exact test, upper tail $p<0.01$). We conclude that a significant percentage of patients with lymphangitic pulmonary metastases do not have crackles.

BREATH SOUNDS ANALYSIS IN PULMONARY SCREENING AT WORK

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Pulmonary function tests (PFT) are often the sole objective measurements used in screening for lung diseases. Phonopneumography (PPG) offers another complimentary, objective method.

This report summarizes the results of 3 years of PPG research. A portable micro-computer based system was developed that samples lung sounds from 4 transducers and analyzed them using FFT in conjunction with PFT and a computerized questionnaire. Comparing the PFT and PPG results to known standards and adding the questionnaire findings provided primary detection of lung abnormalities.

We obtained complete results in 493 male and female subjects. 79% of them were found normal by all criteria. PFT identified 70% of the 104 patients. Lung sound analysis identified only 53 of the patients, but detected abnormal lung sounds in 17 patients with chronic bronchitis not identified from their PFT. The use of both lung sounds plus PFT increased objective test sensitivity from 70% to 87% ($p<0.001$).

SQUEAKS AS A DIAGNOSTIC FINDING IN LUNG DISEASE

Robert P. Baughman, Elyse E. Lower

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Squeaks are a short (<250 msec), continuous lung sound, usually of frequency >400 Hz. In order to determine the prevalence of squeaks and its relationship to lung disease, we noted all patients who presented with squeaks over an eight month period. During that time, 16 patients with squeaks were encountered. Underlying diseases included idiopathic pulmonary fibrosis (IPF) in 6 cases, sarcoidosis in 5 cases, bronchiectasis in 2 cases, and asthma in one case. Two patients had no underlying lung disease. Four of the five sarcoidosis patients had advanced disease with evidence of bronchiectasis by CT scan. The six patients with IPF had persistent squeaking throughout the period of examination, several had squeaks documented for more than two years. The six patients with bronchiectasis (including 4 sarcoid patients) had intermittent squeaks associated with increased sputum production. The squeaks resolved with antibiotic therapy, but recurred with recurrent infection. Of the remaining four cases, including one patient with sarcoid, there was an abrupt onset of symptoms of cough and dyspnea. In two of these cases, bronchiolitis obliterans with obstructing pneumonia (BOOP) was diagnosed and treated with steroids. The other two cases were treated with antibiotics. All four patients had resolution of squeaks and symptoms with therapy. We conclude that squeaks are often associated with chronic lung diseases, either IPF or bronchiectasis. In patients without IPF, squeaks are an important diagnostic finding since therapy directed towards infection or steroids (in those patients with BOOP) may resolve symptoms.

SQUEAKLETS IN BOOP

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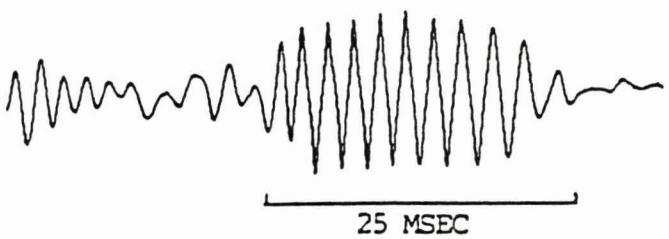
The majority of adventitious lung sounds can be classified as crackles (rales) or wheezes (rhonchi). Crackles are commonly described as less than 20 msecs in duration, and wheezes as more than 250 msecs. Intermediate or transitional forms are not as common, but have been described as squeaks or squawks (Earis, Thorax 1982;37:923-6) or as "crackle-wheezes" (Forgacs, Lancet 1967;2:203-5). The differences between the two major categories lie not only in their duration, but also in their morphology and in their mechanism of production. Wheezes show a periodic or quasi-periodic sinusoidal pattern, whereas crackles show only a few cycles, changing rapidly in their amplitude and zero-crossing interval. On auscultation the distinction can readily be made between the showers of crackles, brief sounds without perceptible pitch, and the more continuous musical sounds of wheezing.

A patient with bronchiolitis obliterans and organizing pneumonia (BOOP) presented with profuse, varied adventitious sounds all over her chest. Sounds were recorded at six sites on six successive days, during which she improved rapidly. Crackles and wheezes were apparent to ear and to eye, but very brief squeaks were also audible in the earlier recordings. They were readily identified from breath to breath, occurring at the same point in the respiratory cycle over a series of breaths like the "index crackles" of Nath and Capel (Thorax 1974;29:695-8). They ranged from 16 to 36 msecs in duration, with consistent zero-crossing times within each "squeaklet". They were not regularly preceded by a crackle, although they lived in a jungle of crackles. Their envelope varied in shape more than their frequency content. They seem likely to have been produced by a different mechanism from that which produces crackles. Presumably they originated in bronchioles excited to produce sound only briefly.



POLYNOMIAL AUTOREGRESSIVE MODELS OF NORMAL LUNG SOUNDS

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Earlier research of our group has shown that autoregressive (AR) parameters of lung sounds clearly depend on airflow and lungvolume. In the research reported here, it is shown that 3rd degree polynomials in flow and volume can be used to describe AR models of lung sounds. These polynomial models produce linear prediction errors comparable to those of AR models directly fitted to the original data. It is shown that the coefficients of the polynomials are influenced by induced airway narrowing. Therefore, these coefficients may become a diagnostic tool for lung diseases that affect the lung sounds. Moreover, only 30 coefficients describe completely the polynomial model. The result is a considerable datareduction, compared to the original number of lung sound samples.

COMPUTATIONAL METHODS FOR ASYMMETRY ANALYSIS OF LUNG SOUNDS

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Assessment of differences of lung sound characteristics of left and right lung is needed in clinical situations where lung function is asymmetrical, e.g. in pneumothorax, endobronchial tumor, and in all other situations with unilateral pulmonary pathology. Real-time asymmetry analysis of lung sound recordings can be employed to monitor too deep location of intubation tube during anesthesia or in intensive care. Furthermore, dual-channel analysis of recordings performed with two sound sensors located apart of each other can be used to extract knowledge about the penetration of the sound signal in the chest, which in turn provides information of the sound transmission properties of the underlying tissue in the presence or absence of a disease.

We have developed computer software for analysis of asymmetry in lung sounds. The methods provide the following asymmetry indices or functions: (i) energy ratio index of RMS values of the time-domain signals; (ii) energy ratio index of RMS values of FFT spectra, (iii) similarity function of the FFT spectra (spectral fit), including an index of the similarity over the considered frequency range, (iv) coherence function and index, and (v) cross-correlation function of the signals. Furthermore, (vi) the upper frequencies of power quartiles of the spectra of signals from both channels are calculated to quantize differences in the form of the signal spectra.

The FFT-spectra in methods (ii-vi) are computed with the Welch overlapped-segment averaging method[1], using a selectable window function. The frequency range considered can be varied.

The asymmetry functions and indices provide quantitative information about the differences in the two sound signals. Straightforward comparison of RMS signal energy (i, ii) or indices of differences in the shape of the spectra (iii, vi) are of value in diagnosis and monitoring. Coherence and cross-correlation functions might be utilized in localization of the generator of the sound or in the design of registration[2].

References:

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AN ALGORITHM FOR THE DETECTION OF RESPIRATORY SOUND CRACKLES

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In this work, a method using two-level adaptive thresholding of the energy envelopes of the respiratory sounds is suggested for crackle detection. The detection algorithm is tested on data from ten patients having restrictive and obstructive pulmonary diseases. In the suggested algorithm, the recorded sound signal is first segmented into inspiration and expiration phases, then each segment is normalized to its maximum value and the energy envelope of the normalized signal is calculated. To reduce the effect of the background vesicular sound on the energy envelope and to aid in the determination of start and end of crackles, the signal is set to zero below a threshold, T_1. A peak detector is used to locate the peaks in the resultant signal and if a peak value exceeds an adaptively adjusted threshold T_2, a decision is made towards the existence of a crackle. The number of automatically detected crackles is compared to the number of crackles counted from time-expanded waveforms by two observers to test the performance of the detector. The performance analysis shows that the sensitivity and the positive predictivity of our detector is comparable to performance of the detector recently introduced by KAISLA, *et. al.* (1991).

Mathematical Formulae for Pulmonary Diagnosis Using Lung Sounds

Murphy R., Spiegelman D., DelBono E., Hertzmark E. and Bettencourt P.

Utilizing data from a study reported at the 1990 ILSA conference, we developed multiple logistic regression models to aid in the classification of common lung diseases.

The previous study included observations utilizing lung sound mapping with an acoustic stethoscope and time-expanded waveform analysis (TEWA) on 20 patients with each of the following diagnoses: interstitial pulmonary fibrosis (IPF), chronic obstructive pulmonary disease (COPD), congestive heart failure (CHF) and pneumonia (Pn) as well as 15 subjects without evidence of lung disease (CNTL). Two logistic regression models were developed to predict, separately, patient probability of being in each of the five diagnostic categories as a function of lung sound and demographic variables. The first model included all available variables, the second all variables without TEWA. Receiver operating Curves (ROC) were constructed for each model. The simultaneous diagnostic prediction rule, using these models, was applied to classify each patient into the disease category for which the highest predicted probabilty was obtained. Using Model 1, 90% of IPF, 90% of Pn, 80% of CHF, and 55% of COPD patients were correctly classified. Using Model 2, 85% of IPF, 75% of CHF, 65% of Pn, and 35% of COPD were correctly classified. Eighty-seven percent of controls were correctly classified as controls. AROC ranged from 0.96 for IPF and CHF to 0.80 for COPD. Using a bootstrap procedure to simulate performance of this system with new data, the results were similar: Model 1 correctly classified 78% (CI=67,90) and Model 2 correctly classified 68% (95% CI=58,79). These observations support the hypothesis that lung sound patterns differ among common lung diseases and this has implications for noninvasive diagnosis.

-from the Pumonary Services of the Faulkner and Lemuel Shattuck Hospitals, and the Division of Biometry, Department of Community Health, Tufts University School of Medicine

Performance of air-coupled and contact sensors in lung sound measurements.

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Acoustic sensors are a fundamental component of any system used in the measurement of lung sounds. However, their use suffers from a lack of standardization that is also apparent in other aspects of lung sound research. We therefore decided to define the performance of a representative selection of lung sound sensors under standardized conditions. We gathered 3 air-coupled electret microphones: Sony ECM 155, Sony ECM 77, and Radio Shack #33-1052; and 4 contact accelerometers: Hewlett-Packard 21050, Siemens EMT 25C, PPG sensor (Technion, Haifa), and FYSPac2 (Univ. of Brussels, Belgium). Lung sounds were recorded from three of the investigators who were non-smokers. The best recording site on the posterior lower chest was chosen by auscultation (left in SSK, right in PDD and GRW). The sensors were tested in sequence and the recording site remained fixed. The subjects sat in an anechoic chamber and breathed through a Fleisch No.2 pneumotachograph. Lung sounds were band-pass filtered between 0.1 and 2 kHz and then sampled together with air flow at 10 kHz, using 12 bit A/D conversion. A 2048 point FFT was applied at successive 100 ms intervals, using a Hanning window. We averaged the resulting power spectra for sounds at flows of 2 l/s ± 0.5 l/s and plotted them against background noise detected at zero flow. To describe sensor performance we used the maximum signal-to-noise ratio (dB) and the slope of the lung sound spectra from 0.3 to 1 kHz on a log frequency/log power plot (dB/octave). Results were as follows:

	Subj. 1		Subj. 2		Subj. 3	
	S/N	slope	S/N	slope	S/N	slope
SONY 155	41.4	-26.5	29.8	-23.7	24.6	-19.9
SONY 77	42.2	-27.9	28.3	-24.4	23.1	-19.2
Radio Shack	38.6	-18.7	25.3	-19.2	23.3	-18.0
HP 21050	42.0	-15.7	24.5	-16.0	28.9	-12.3
Siemens EMT	33.0	-16.2	16.4	-12.8	22.3	-14.2
PPG (Haifa)	41.1	-16.2	25.0	-16.1	26.2	-15.0
FYSPac2	40.3	-11.4	26.0	-10.7	25.5	-12.4

Both air-coupled and contact sensors were comparable with regard to S/N ratio, with the Siemens accelerometer showing the lowest sensitivity. Unexpectedly, we observed a steeper drop of the spectral curve and therefore less sensitivity at high frequencies in the air-coupled devices. Our findings illustrate the need for exact characterization of sensor performance in studies of lung sounds. As a minimum requirement we suggest to average spectra over several breaths within a range of known air flows, and to present all measurements relative to sounds at zero flow.

The Bronchofiberscope used as an acoustic probe for the endobronchial sounds

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Endo bronchial detection of breath sounds has been performed using a normal bronchofiberscope (Olympus BF type 1T10) as probe microphone. The inner channel of insertion tube acts a sound pipe and is acoustically coupled,at the upper end,with an 1/4 inch electret condenser microphone (Sony ECM-144). The little chamber which normally houses the manoeuvrable accessory heads has proved luckily suited for this purpose. The acoustic transfer function of a cylindrical pipe is theoretically characterized by resonance occuring at regularly spaced frequencies, easily calculable in the ypothesis of small transmission losses, consistent with the inner channel diameter (2.6 mm) and the wall stiffness of the fiberscope.The experimental results have fully confirmed the theoretical forecast. As a matter of fact,we can use the measured transfer function of the probe microphone (Fig.1 respectively phase and magnitude) to equalize,once for all, its response before the signal processing.
During normal sessions of bronchoscopy we have tried to detect breath sounds in the trachea and in the bronchial tree by our probe microphone and simultaneously on the chest wall by an air coupled microphone (Sony ECM-144).The signal were processed by a two channel FFT analyser (Ono Sokki CF-940) . Fig.2 shows typical spectra of the sounds detected on the chest (a) and in trachea (b):comparing them, it may easy evidentiated the strong attenuation of the high frequencies of the breath sounds detected on chest wall due to the low-pass filtering action of the surrounding tissues. Therefore picking up the breath sounds closer as possible to the generation site seems to be very useful both for physiological and clinical studies, as their intrinsic features are more accurately preserved.

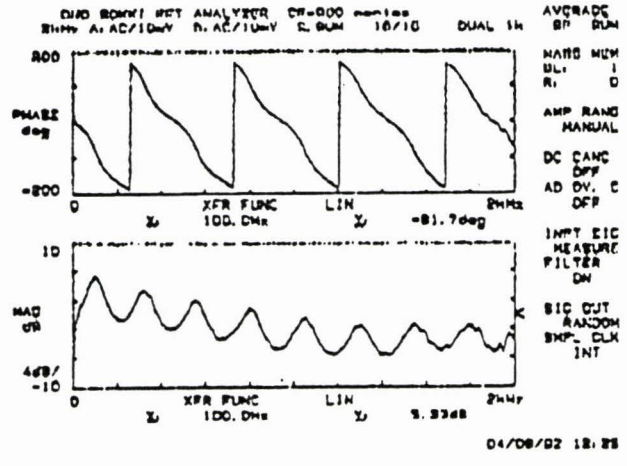

FIG 1

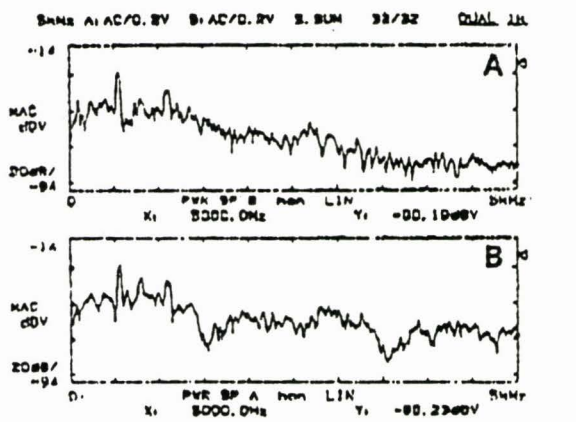

FIG 2

Speed of sound propagation in inflated sheep lungs and normal men

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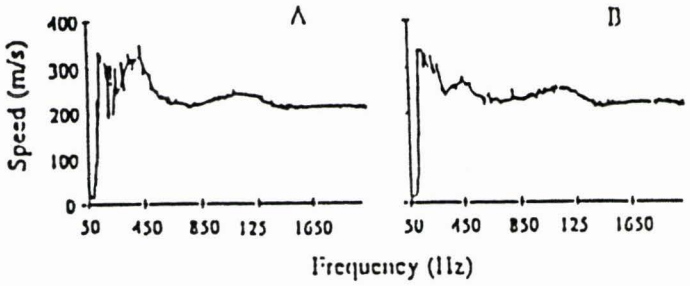
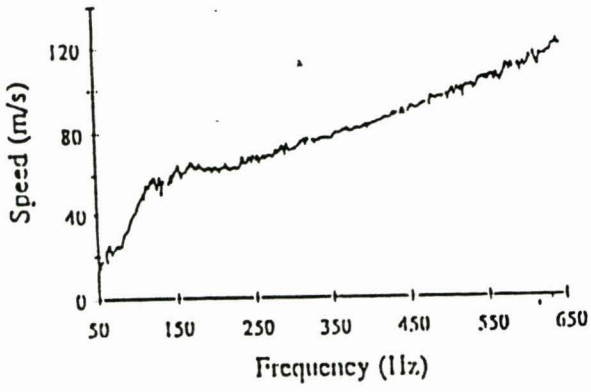
In order to understand how sounds propagate in the respiratory system, the transfer function phase and coherence of sound signals were measured from the trachea and different sites on the pleural surface of inflated sheep lungs and on the chest wall of normal men. Random noise in the range of 50 - 2000 Hz (AudioSource EQ ten Noise Generator - Equalizer) was applied to the trachea of 5 excised adult sheep lungs and to the mouth of 6 young healthy men. The signals were picked up simultaneously by contact sensor (built in our laboratory), amplified, filtered and directly converted to digital form. Using a computer program that we wrote to handle the data, 0.5 second segments of data were digitized at 4000 samples per second from each of the sensors during open glottis quasi-breathold in the men or in the sheep lungs which were inflated using external air-source. The Fourier transform of the digitized signals was calculated by the FFT and the magnitude, phase (Φ) and the coherence of the transfer function were calculated. The speed of sound propagation (c) between the two sensors was calculated from $c = 2\Pi fl/\Phi$, where f is the frequency and l is the distance between the sensors. The calculated parameters of each segment were averaged with those of previous segments until a stable, low noise pattern was obtained. A speed of approximately 250 m/s was found for each of the lungs studied and in several pick up locations on each lung. Using simple algebra and previously published values of 350 m/s and about 50 m/s for sound propagation through airways and lung parenchyma we estimated that only approximately 1 cm of the total trachea - pleural surface acoustic path is through the parenchyma and the rest is via airways. Very little dispersion(i. e., frequency dependence of sound speed) was noted in the sheep lungs (see figure below). The speed of the sound propagation in the respiratory system of young men is 40 - 120 m/s, much lower than the speed in the excised sheep lungs.and a very significant dispersion is observed (see figure below). The mechanisms of this features are not well understood and may be related to species differences, to the effect of the chest wall or to frequency dependence of acoustic path.



SHEEP LUNG



HUMAN, intact

FEASABILITY AND UTILITY OF NOCTURNAL MONITORING OF BREATH SOUNDS AND PULSE OXIMETRY.

Postiaux G. (*), Lens E. (*) and Physicians of Courcelles (Belgium).

Setting: during 3 months, 32 night records on 30 consecutive patients followed at home by 8/35 physicians in a town of 25.000 inhabitants.
Intervention: continuous respiratory sounds analysis of sonospirogram(R) (normopnea, hypopnea, apnea, snoring, cough), wheezing and pulse rate + SaO2 (pulse oximetry). Portable device: SLEEPSOUND-ELENS.DSA.II.
Subjects: 30: 20 males, 10 females; mean age: 55 years, range: 0,5-84 years.
Results:
A. Utility: 1) 40 physician's motives: wheezing: 18 times, apneas-hypopneas, snoring: 16 times, SaO2-BPCO: 4 times, cough: 2 times. 2) 22 contribution versus expected signs: wheezing: 13 times, apneas-hypopneas, snoring: 5 times, SaO2-BPCO: 3 times, cough: 1 time. 3) 32 unexpected discovered signs where one expected sign was associated with one or several others. Thus utility may be appreciated by the score of 54 signs disclosed versus 40 expected.
B. Feasibilty: 1/32 technical failures, 1/32 SaO2 probe untolerated. Suspended questions are comparative cost versus hospital's sleeplab, technical management: training of physician or technician.
Conclusion: we estimate that these objectivation's possibilities are usefull and ought to be performed in standard practise. Feasability needs further comparative evaluation. Moreover we propose to associate so far as needed oesophageal pHmetry, holter and arterial tension monitoring.

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A COMPARISON OF COMPUTER-BASED LEARNING (CBL) AND SMALL GROUP TEACHING OF PULMONARY AUSCULTATION TO FIRST YEAR MEDICAL STUDENTS.

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Bedside auscultation is suffering from declining interest and proficiency, caused by competing diagnostic technology and inadequate training of physicians. We need novel methods to teach and revive this time-honored art. Computer Based Learning (CBL) supporting graphics and digitized sound could be ideally suited for this task. To evaluate this premise we randomized 53 first-year medical students to 1 hour of seminar teaching plus the use of audiotapes and a textbook (group 1) or the self-use of a MacIntosh-based CBL (group 2). All students took a pre and post-test consisting of 9 pre-recorded respiratory events and a multiple-choice questionnaire on pulmonary auscultation. They were also evaluated for Computer Anxiety and preference of conventional teaching versus computer-based learning. Both groups significantly and equally improved in both sound recognition and knowledge. There were no differences in the time spent on the learning material between the two groups. Computer anxiety among students with poor prior academic performance was negatively associated with improvement. We conclude that CBL is as effective as a seminar in teaching pulmonary auscultation. Students with poor prior academic performance and low computer anxiety appear to benefit the most from the use of courseware.

Mechanisms of Wheeze Generation

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Wheezing lung sounds are the predominant objective sign of obstructive lung diseases. There are very few obstructive airway diseases in which wheezes are not present. (E.g. bronchiectasis, the "silent lung" syndrome). Wheezes are continuous adventitious lung sounds that have a musical character on auscultation. They are produced during both expiration and inspiration and may be heard over the chest wall, the trachea, or at the mouth. Wheezing may be monophonic or polyphonic and their duration may vary from 50 ms to a whole respiratory phase. In the time domain, wheezes have a continuous and repetitive (periodic) waveform that may be simple (i.e. sine wave), or complex and superimposed on the random fluctuations of the basic lung sounds. The intensity of wheezes may be substantially greater than that of the basic sounds, but may also be lesser than the background. In the frequency domain, a wheeze is represented by a single sharp and narrow peak of power within a wide frequency range (80-2000 Hz over the chest wall). A wheeze may have harmonics (i.e. multiples of the basal frequency) and may be shifted in frequency during the course of a breath. Multiple peaks of power within a single power spectrum are due to either simultaneous production of several wheezes, or to sequential generation of single wheezes during the time segment represented in the power spectrum.

Recent studies on the mechanisms of wheeze generation provided experimental and theoretical information that may explain how wheezes are produced. First, The spectral pattern of wheezes, namely sharp and narrow peaks, is clearly incompatible with a resonator mechanism (e.g. Helmholtz resonator), still proposed in some physical examination texts as the mechanism of wheezes. Second, experiments in self-supporting thick-wall collapsible tubes, and in excised animal airways and lungs show that wheeze-like sounds are generated when a specific set of conditions is met: 1) The airway is buckled but not completely collapsed, so that the opposing airway walls are close to each other but do not touch. Such buckling is always associated with production of sufficiently negative transmural pressure[1]. 2) The flow - driving pressure[2] relationships clearly indicate the presence of airway flow limitation[3]. Furthermore, the air speed inside the airway is equal to, or greater than, a specific threshold, called the "critical flutter velocity", found to be greater than the air speed needed for the onset of flow limitation. 3) The release of acoustic energy coincides with oscillatory motion of the airway walls. The maximum amplitude of the oscillations is located downstream from the starting point of buckling.

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[1] Transmural pressure: pressure inside the airway minus the pressure outside.

[2] Driving pressure: pressure gradient along an airway.

[3] Airway flow limitation is when flow rate does not increase when driving pressure is raised.

Experiments in normal subjects who produced wheezes by exhaling forcefully - forced expiratory wheezes (FEW), provided additional information that support the observations outlined above. In particular, it was found that FEW are generated only when the flow is limited, and only if a critical transpulmonary pressure (P_{tp})[4] has been attained. It should be noted that while the critical P_{tp} for onset of flow limitation was only -6.5±1.2 cmH_2O, the threshold P_{tp} for onset of wheezes was -33±12 cmH_2O. This means that not only flow limitation must be present in order for wheezes to be generated, but the airways downstream from the choke point should be sufficiently collapsed. This collapse serves two purposes: it brings the opposing airway walls closer together, and, for a given flow rate, increases the air velocity. Both effects increase the Bernoulli forces and the likelihood of onset of airway wall oscillations. The importance of airway wall oscillations as opposed to gas-phase oscillations was clearly demonstrated in a study that showed that there is no gas density effect on the frequency content of FEW.

Once the critical conditions for wheeze production have been reached, airway wall oscillations emerge. The onset of these oscillations is the outcome, according to Grotberg's mathematical model, of interactions between several forces: the forces exerted by the flowing gas, the static transmural pressure, and the restoring elastic forces of the airway wall - the tissue tethering, the elastic properties of the wall structure, and the longitudinal tension along the airway. Instability is caused by a delay between the local pressure fluctuations inside the airway and the actual motion of the airway wall, which provide energy to sustain the oscillations. This wall oscillation is predicted by the theory to be the source of the acoustic energy of wheezes.

There are several unresolved issues regarding the mechanisms of wheeze generation. Wheezing patients often produce wheezes while not necessarily breathing on their maximal expiratory curve and therefore are not globally flow limited. Subsequently, the concept of regional flow limitation must be invoked to account for these wheezes and yet maintain a unified mechanism for wheeze production. Similarly, inspiratory wheezes, frequently found in asthma, indicate that inspiratory flow limitation may exist in some airways. These new concepts are yet to be evaluated experimentally. Some wheezes, especially in young infants, are different from the adult ones in their time domain waveform - periodic but non-sinusoidal - and in their spectra (Pasterkamp et al.). It is not clear whether these wheezes are produced by the same mechanism as in adults and what is the physiological and clinical significance of these differences.

In conclusion, current data indicate that wheezes are produced in narrowed airways that are flow limited, when air speed exceeds a certain critical value. Interactions between the dynamic forces exerted on the airway walls by the flowing air, the static transmural pressure and the restoring elastic forces of the wall, generate oscillatory wall motion that is the source of acoustic energy of wheezes.

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[4] Transpulmonary pressure is defined as airway opening pressure minus pleural pressure.

Lack of low frequency power in lung sounds of newborn infants compared to adults.

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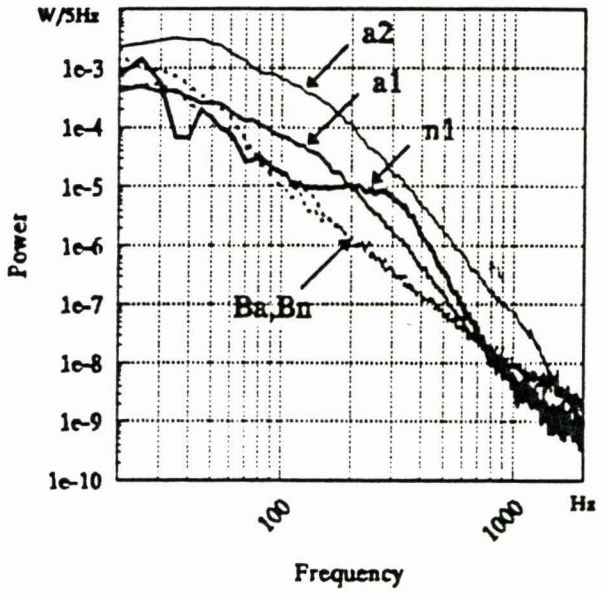
On auscultation, lung sounds in newborns appear different from those in adults. Higher median frequencies have been described in infants, but measurements have not been made at standardized air flows. We therefore studied 10 healthy term newborns (N) on their 2nd or 3rd day of life, and compared them to 10 normal adults (A), ages 25 to 37 y. Using a contact sensor (EMT25C, Siemens), we recorded lung sounds over the posterior right lower lobe of the prone subjects. Flow was recorded from calibrated pneumotachographs. Both signals were digitized at 10 kHz, and sounds were analyzed by 1024 point FFTs, applied at successive 100 ms intervals, using a Hanning data window. At inspiratory flows of 0.015 $l/s \cdot kg^{-1}$ (± 20% tolerance), we sampled 8.1 s ± 0.7 (mean ± S.D.) of lung sounds from each N (total recording 52 s ± 3), and 8.6 s ±1 from each A (total recording 57 s ± 2). Power spectra during inspiration were plotted against background noise (see Figure). Lung sounds in N reached background noise (< 3 dB difference) at 131 ±15 Hz as the lower limit and at 635 ± 42 Hz as the upper limit. Respective values in A were 73 ± 20 ($p < 0.05$) and 481 ± 55 (NS). There was greater spectral power from 100 to 300 Hz in A (31 dB ± 2) than in N (27 dB ± 1; $p < 0.05$), while N had greater power from 300 to 600 Hz (21 dB ± 1 vs 16 dB ± 1 in A; $p < 0.01$). Roll-off in spectral power above 300 Hz was similar (16.8 dB/octave ± 3.0 in N vs 14.7 dB/octave ± 2.4 in A). The frequency at maximum signal-to-noise (Fmax) was 307 ± 17 Hz in N and 142 ± 20 Hz in A ($p < 0.001$). The power at Fmax was 14 ± 2 dB in N and 11 ± 2 dB in A (NS). Thus it appears that newborn lung sounds at given air flows are characterized by a relative lack of low frequency power compared to adults, while sound attenuation at higher frequencies is similar. We speculate that resonance characteristics of the thoracic cavity in newborns may explain some of the observed differences.

Figure

Average spectra of lung sounds from adults (a1) and newborns (n1) at inspiratory flows of 0.015 $l/s \cdot kg^{-1}$, and from adults (a2) at 0.030 $l/s \cdot kg^{-1}$. Sound spectra are plotted with background (sound at zero airflow) subtracted. Background spectra are shown as dotted curves (Ba and Bn). Note the relative lack of power from 100 to 200 Hz in the newborn infants.



The Acoustical Analysis of Coughs

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Cough can be induced in the laboratory by the inhalation of several tussive agents such as low chloride solutions, capsaicin and prostaglandins. It is not clear whether all these stimuli induce cough through the same receptor, such as the irritant receptor in the larynx and large airways or through separate receptors. While performing cough challenges with low chloride solutions, capsaicin and $PGF_{2\alpha}$ it was found possible to differentiate between the responses induced by the various tussive agents since they sounded different to the ear. In order to investigate this further, cough sounds were recorded and then subjected to spectral analysis, to investigate whether different tussive agents produce coughs with distinctive spectral profiles.

This paper will (a) present the spectrographic method used, (b) introduce a new set of parameters which can be used to analyse and characterise cough responses and (c) present the analysis of spectral profiles of cough challenges performed on nine normal subjects, provoked by the administration of NaCl, capsaicin and $PGF_{2\alpha}$.

It will be shown that (i) there was significant reproducibility of spectral profiles for the same subject, provoked by the same tussive agent, when repeated over time, (ii) the spectral profiles tended to be specific to a tussive agent, (iii) the spectral profiles of coughs clearly show that there was reproducible and significant levels of acoustic energy up to at least 5 kHz, although the peak sound pressure levels were at frequencies less than 1.5 kHz, (iv) the spectral profiles of capsaicin suggest it provokes a fast cough response (i.e. the period of time from the administration of the tussive agent to the onset of cough is short), however, its effects are short-lived, (v) $PGF_{2\alpha}$ appears to have long initial time to cough onset but it is a persistent irritant, and (vi) repeating the capsaicin tests after $PGF_{2\alpha}$ showed an apparently more severe response in terms of the established cough count parameter and the new parameters proposed.

Possible future research applications will be discussed.

THE SPECTRAL CHARACTERISTICS OF SOUND TRANSMISSION IN THE HUMAN RESPIRATORY SYSTEM DURING HFJV

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High frequency jet ventilation (HFJV) is a open-system with high frequency and low tidal volumn,it is different from normal ventilation in the principle.

In order to delineate the pathway taken by sound induced into the airway during HFJV,we measured the propagation time of sound in respiratory system of neurosurgical patients.Spectral characteristics of transmission were also estimated.

Five points over the thorax were chosed as representative pickup locations:trachea-anterior cervical midline (TR);right andleft antrior chest(CR,CL),2nd intercostal space,in the midclavicalar line;right and left base of lung,5th intercostal space in midaxillary line(BR,BL).

Trough reserch,we found that:

1.The power spectrum of lung sound at TR,at CR & CL and at BR &BL are similar curves.When the point of sound pickup from centra to periphery of lung (TR--CR,CL--BR,BL),the frequency range gradally contracts,and the dominant frequency shift from 115Hz to 98Hz to 85Hz (Figure 1).

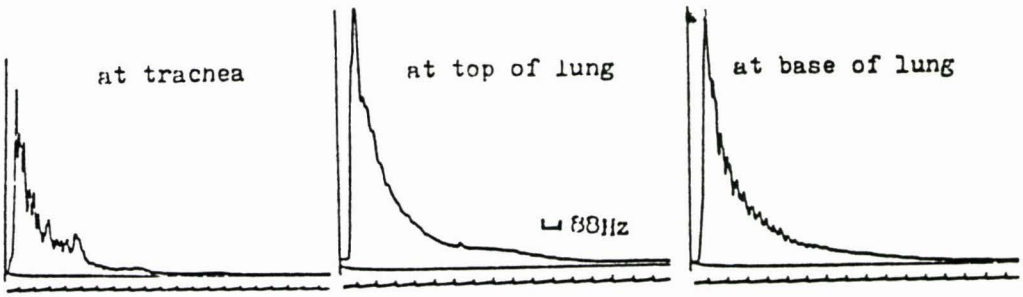

Figure 1:The power spectrum of lung sound at TR,CR and BL during HFJV.

2.By comaried,the power spectrum of lung sound on the dominant frequncy and character of the amplitude spectrum during HFJV is same as the power spectrum during normal respiration. Dependent upon these similar points,we suggest that principe and place of producting lung sound during HFJV would be similar with normal respiration in essence.

THE CHARACTERISTICS OF TRACHEAL BREATH SOUNDS DURING HFJV

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High frequency jet ventilation (HFJV) is a open-system with high frequncy and low tidal volumn, is is difficulty to be measured the tidal volumn under every driving pressure (DP) during HFJV. But we found that DP has a certain correlation with lung sound intnsity.

To regulate HFJV on accurate and quntity, we reseached the correlationship between DP and lung sound at trachea during DP from 0.06 to 0.14 MPa. Subjects were all neurosurgical patients. The point of sound pickup was chosen at trachea-anterior cervical midline. Tracheal breath sound recorded on magetic tape was replaied and passed through a low-pass filter (100-2000Hz), filted sound was analysised by mode 7T08-5 Signal Processor (A/D-10bit, Sample rate-1024points/second, response range-DC to 50kHz, etc).

Results are:

1. On frequncy domain, the power spectrum of tracheal breath sound is from 100 Hz to 1300 Hz, and has a dominant frequncy at 110-120 Hz and a peak frequncy at 400-490Hz. By analysis of the peak frequncy, we found that the wave of the peak frequncy was a wheeze's (Figure 1)

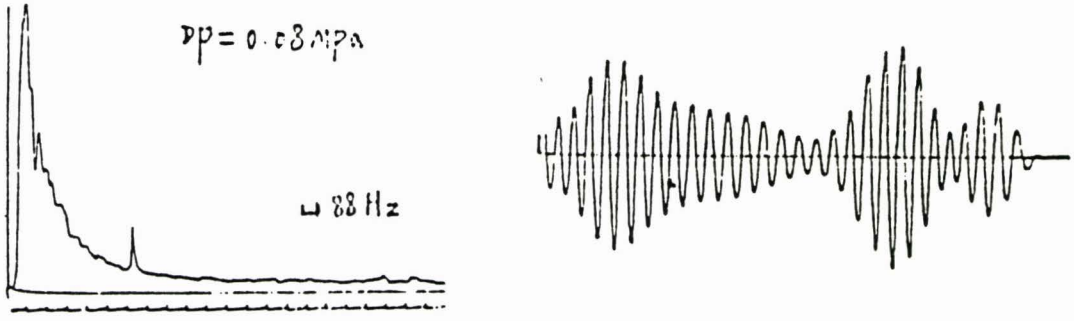
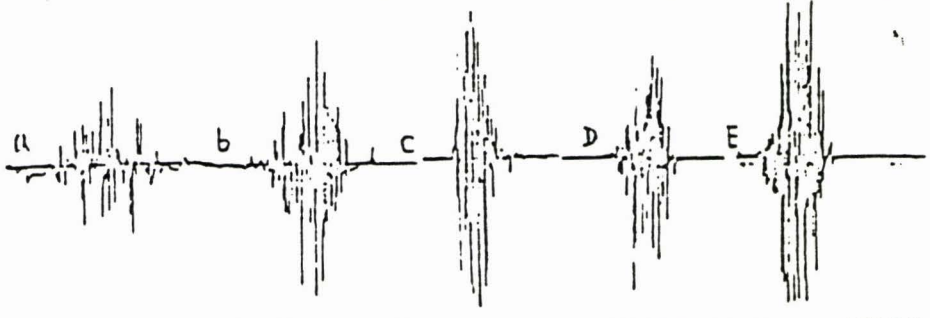


Figure 1: The power spectrum of tracheal breath sound and wave of the peak frequncy.

2. On time domain the envelop of inspiration of lung sound at trachea near linearly increased when DP 0.06→0.14MPa. The is a positiv coefference between them ($r=0.997$). The envelop may be used non-invasely to estimate the DP level during HFJV. (Figure II)



(a) DP=0.08MPa Y=16 (b) DP=0.10MPa Y=32 (c) DP=0.12MPa Y=64 (D) DP=0.14MPa Y=128 (E) DP=0.16MPa Y=128

Figure II: Change of the envelop of tracheal breath sound duing DP 0.06→0.16MPa.

AUSCULTATION OF FINE CRACKLES USING A STEREOPHONIC STETHOSCOPE

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Using a newly available stereophonic stethoscope (Kenz Medico Co.,Ltd., Stereophonette), fine crackles were analyzed on time and intensity difference between the sounds separately recorded from the right and the left chambers of the stethoscope. Acoustic characteristics of the stethoscope were also studied. The stethoscope has a single chest piece housing paired open-bell and diaphragm type of semicircular chambers. A pair of earpieces are separately connected to the right and the left chambers with a single tube containing two independent sound ducts.

First, frequency characteristics of the stethoscope, diaphragm type, was studied using white noise. No conduction difference of sounds was found between the two chambers.

Second, the chestpiece was put on a parallel or perpendicularly to the direction of the sound transmission from the source in the free air field. In this way, the two chambers were estimated as two independent pick-ups 2.8cm apart from each other.

Third, fine crackles were examined in eight patients with idiopathic pulmonary fibrosis. Time and intensity differences were analyzed on each crackle of 21 fine clackles recorded from both chambers. The mean time difference between both sides was 199 microseconds (49-440 microseconds), which was considered enough to identify the laterality of the sound source in acoustic sence. However, intensity difference was less reliable because the sound intensity was affected by attachment conditions of the cnest piece to the skin.

In clinical auscutation, fine crackles were heard from the multiple direction, although the exact direction of each crackle could not be distinguished because of the complexity and transitoriness of fine crackles.

COMPARISON OF LUNG SOUND PARAMETERS BETWEEN NORMAL AND ASTHMATIC SUBJECTS USING AIRFLOW-STANDARDIZED PHONOPNEUMOGRAPHY. H.J.W. Schreur, M.C. Timmers, R. Schot, J. Vanderschoot*, J.H. Dijkman, P.J. Sterk.
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The difference of lung sound characteristics between normal and stable asthmatic subjects has not been studied in a airflow-standardized way yet. We compared lung sounds in 9 normal and 9 mildly asthmatic subjects with normal lung function. All subjects underwent simultaneous recordings of airflow and lung volume changes, and lung sounds at 3 locations during standardized quiet breathing, and during maximal forced manoeuvres. Airflow-dependent power spectra were computed using FFT. These spectrums were submitted to wheeze detection, and to analysis of the ratio of wheezy spectrums to the total number of spectrums (W%). Subsequently, lung sound intensity (LSI) and the quartile power points (Q_1-Q_3) were determined for each spectrum. The results were analyzed by ANOVA. LSI was lower in asthmatic than in normal subjects ($p \leq .001$), and LSI was strongly dependent on airflow ($p \leq .001$). Q_1-Q_3 were either higher or lower in asthmatic than in normal subjects, depending on the breathing manoeuvre ($p \leq .049$). Q_1-Q_3 were strongly dependent on airflow ($p \leq .001$). $W_{\%}$ was not different between asthmatic and normal subjects ($p \geq 0.146$). In conclusion, lung sounds are louder in asthmatic subjects with a normal lung function than in normals, and the pitch of the sounds is different between the subject groups (asthma/control). This implies that in stable asthmatics the generation or transmission of lung sounds is different from that in normal subjects, and that airflow standardized phonopneumography reflects differences in airway morphology between asthmatics and controls.

Supported by a grant of the Netherlands Asthma Foundation.

Snoring sounds recorded by air-coupled microphone and by a new piezo-electric sensor

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In the literature different methods of recording snoring sounds have been described e.g. using larynx-microphones positioned at the cricoid notch or at the supra-sternal notch. Apart from this the sound-channel in our sleep laboratory consists of a professional quality condenser microphone (Sennheiser MKE-2-3) suspended 60 cm above the patients head resulting in a quasi-nearfield recording situation. To compare the different recording techniques we made simultaneous 2-channel-recordings with the air-coupled microphone and a piezo-electric transducer with preamplifier and antialiasing-filter developped by the Fachhochschule Ulm. Recordings were all-night in ten patients with different severity of snoring and obstructive sleep-apnea-syndrome. The piezo-electric tansducer was attached to the suprasternal notch. Recordings were on a SVHS HiFi-Videorecorder Panasonic 100 NV. Power spectra were calculated from simultaneous parts of the records for comparison.
Results. Large differences were seen in the frequency-response and sensitivity of the two devices and setups. The larger bandwidth was seen in the quasi-nearfield recordings, with a frequency range from about 25 to more than 10 kHz. Frequency response with the piezo-electric senosor was restricted to about 3 kHz. In the region about 1,25 kHz the sensitivity of the piezo-electric transducer was at least 20 dB better than with the air-coupled microphone. As a first conclusion we can say that for recordings of tracheal breath sounds the piezo-electric transducer is superior. For recordings of snoring the advantages are not as clear.

COMPARISON OF SPECTRAL DENSITY OF TRACHEAL BREATH SOUNDS BY FAST FOURIER TRANSFORM AND BURG'S ESTIMATOR.

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Most of the actual studies on spectrum analysis of breath sounds have employed an FFT approach, although it has known limitations in frequency resolution and distortion due to windowing. Autoregressive models have been proposed as an alternative to reduce FFT restraints, and because they present advantages such as data compression for recognition purpose. The goal of this study was to compare spectral densities of tracheal breath sounds estimated by FFT and Burg's algorithm. Separated inspiratory and expiratory sounds (for a total of 78 records) were processed from 10 subjects breathing at flow rates from 0.5 to 1.5 l/sec. Statistical comparisons of spectral values showed a good correlation ($r>0.973$) and no-bias between both techniques. Highest mean relative error was found for the frequency at the maximum amplitude (3.4%) and lowest for the frequency at 99% of the spectral energy (0.37%). Limits of agreement (mean of differences$\pm$2SD) were 1.87 ± 0.53 Hz for mean frequency, -1.0 ± 8.1 Hz for frequency at 99% of the spectral energy, -3.75 ± 23.6 Hz for frequency at 10% of maximum amplitude and -1.7 ± 30.5 Hz for frequency of maximum amplitude. Since a good agreement between FFT and Burg approaches was proved, we suggest employing the later in the spectrum analysis of breath sounds, because in addition it provides an easier identification of spectral parameters.

THE FREQUENCY CONTENT OF SNORES IN MAN

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Heavy snoring is a cardinal symptom of obstructive sleep apnoea and a source of considerable embarrassment in non-apnoeic snorers. However, there is little objective data about how the pitch of the snore varies between breaths, whether it differs through the night, is influenced by sleep stage or severity of disease. We have addressed these problems in a study of 7 patients (all male) mean age 50(17 undergoing standard polysomnography. Upper airways sounds were detected by an air coupled microphone at the manubrium sterni and recorded on to video tape. Sleep was staged using standard criteria with a computer assisted manual system and respiratory variables were verified manually. Snores were identified in each sleep stage (if present) and 30 second periods were digitised and analysed for median frequency by an FFT technique. Sleep quality was variable (TST 334(55)mins sleep efficiency 83(11)% with a wide range of apnoea and hypopnoea values (0-50 events per hr). Median frequency of the snores varied in each sleep stage to a similar degree (CV13%-15%) with no difference between the same stage whether it occurred early or late in the night. The snores did not differ in frequency with sleep stage but frequency was higher in stage 2 sleep in the 4 patients with AHI >15 mean 26) (333 SE 34 Hz) than in the other 3 (Mean A+HI 3.1) (219 SE 35). These initial data suggest median frequency is relatively reproducible and may differ with the nature of upper airway obstruction during sleep.

NOISE REDUCTION IN RESPIRATORY SOUNDS BY ADAPTIVE FILTERING IN TIME AND FREQUENCY DOMAINS.

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Auscultation of breathing sounds have been widely used as diagnostic tool in the clinical area where lung sound analysis have been done on the basis of trained physician's ear. However, in order to standardize clinical results between different observers it is important to analyzed the temporal and spectral characteristics of these sounds. This goal requires the reduction of interference signals like the heart signal noise. In this work, two adaptive filtering schemes are proposed, in the time and frequency domains, in order to minimize the interference signal. The spectral density is obtained by the use of adaptive prediction's coefficients of the related transversal filter. The schemes were tested on records of tracheal breath sounds from five healthy subjects, in the inspiration and expiration phases. One advantage of these filtering schemes is that they use a ten order for the transversal filter in compararison to the higher order proposed in other works. The main difference between the two proposed schemes is that the second one is independent of the temporal variations of heart noise between each cardiac cycle. Results showed that schemes can reduce the heart signal about 80%. Difficulties with the schemes for noise reduction are discussed.

LUNG SOUND CLASSIFICATION EXPERIMENTS BY USING SELF-ORGANIZING MAPS

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We have carried out new experiments to apply the method of self-organizing mapping (SOM) [1] to the classification of lung sounds [2, 3]. One recognition test and two classification experiments were performed.

The SOM method was tested on lung sound samples recorded from ten subjects to recognize the difference between the samples recorded with different sensors at lower lung regions and at the trachea. The samples were recorded with a condenser microphone from the right lung and with a piezo-electric contact sensor at the trachea. The SOM method could make clear difference between the two sample groups forming two regions on the map and all the twenty samples were recognized correctly.

In the first classification experiment the SOM method was applied to classify two phases of the histamine challenge test, the basic and the provocated phase: the lung sound samples were recorded from five patients. Three pairs of the five sample pairs were used for the teaching of the map and two pairs were used for the recognition in the respective map. The procedure was repeated six times by choosing different samples for teaching and for recognition. 68 % of the samples were classified correctly. An additional experiment was done by choosing three sample pairs classified best in the first experiment for a new round of teaching and recognition. All the six samples (100 %) of the three patients were classified correctly.

In the second classification experiment the SOM method was applied to classify the crackling lung sounds of 39 patients in four groups, bronchi ectasis (BE), fibrosing alveolitis (FA), chronic obstructive pulmonary disease (COPD) and heart failure (HF). The method where the features were based on FFT-spectra could not make a clear difference between COPD and HF in their spectral contents. The difference between BE and FA was more marked. When combining two groups as pairs (BE&FA and CO&HF) the result was 82.1 % . Respectively, when using a combination of three groups (BE, FA and CO&HF) the result was 69.2 % of the samples classified correctly and of four groups (BE, FA, CO and HF) 53.8 %.

The results are promising for clinical use at classification by using self-organizing maps.

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COMPARISON OF METHODS FOR AUTOMATIC CLASSIFICATION OF RESPIRATORY SOUNDS

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Automatic classification of lung sounds might be clinically useful. The classification methods used in the present study were *discrimination analysis* and *self-organizing feature maps*. For the self-organizing feature maps the pattern vectors were formed with the fast Fourier transform (FFT). For the discrimination analysis the pattern vectors were formed with the FFT or autoregressive (AR) modelling.

The teaching material consisted of 28 patients having crackles, 4 patients having wheezes, and 6 healthy persons. The diagnosis of the patients with crackles was fibrosing alveolitis (FA), bronchiectasis (BE), chronic obstructive pulmonary disease (COPD), or heart failure (HF). The test material consisted of 6 patients with crackles (FA, BE, or CO) and and 1 healthy person (HP); the persons were not the same as in teaching material. From each patient's recording 5 representative sections were searched visually. From the healthy persons' recordings the five sections were chosen both from inspiration and expiration. Samples of lengths 10 ms, 15 ms, 20 ms, and 25 ms were gathered from the chosen sections.

Discrimination analysis with pattern vectors that were formed from AR model's parameters proved to be the best classification method with every tested sample length. The best classification results (percentages test material samples classified correctly) obtained were: BE 60 %, CO 30 %, FA 60 %, HP insp. 60 %, HP exp. 100 %. The results seem promising for future development of automatic classification of respiratory sounds. It is likely that the classification results for patients' respiratory sounds will be improved by adding parameters that contain information of the timing of the adventitious sounds in the respiratory cycle to the pattern vectors.

The "Ultrafine" Crackles

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We applied the most used parameters for individual crackles classification to the high pitched crackling sounds heard on the chest wall in some pulmonary deseases:

Subcutaneous Emphysema (SE)	4 sites	4 cases
Extrinsic Allergic Alveolitis(EAA)	2x2 sites	3 cases
Lung Granulomatous Disorder (LGD)	2x2 sites	1 case

Breath sounds recorded by an air coupled electret microphone simultaneously with flow at the mouth, were analysed for timing,time-expanded waveform (IDW,2CD,LDW) and spectral analysis (Sonagram,FFT) using a PC with data acquisition card Metrabyte DAS16.

			IDW msec	2CD msec	LDWmsec
		n	m +/- SD	m +/- SD	
SE	a	15	0.38+/-0.15	1.93+/-0.40	0.51+/-0.04
	b	15	0.36+/-0.19	1.71+/-0.43	0.53+/-0.05
EAA	a-b	20	0.74+/-0.11	3+/-0.16	0.91+/-0.12
LGD	a	15	0.52+/-0.17	2.71+/-0.7	0.79+/-0.17
	b	15	0.49+/-0.06	2.16+/-0.14	0.57+/-0.04

The spectra are quite flat with maximum energy in the frequency range of 500-700Hz.The values were compared with the values of the same parameters of 'fine' crackles measured in Criptogenetic Fibrosing Alveolitis(CFA).In SE-EAA-LGD the IDW was significantly shorter($p<0.001$-$p<0.05$-$p<0.001$ respectively)than in crackles of CFA;2CD and LWD were shorter ($p<0.001$)than in crackles of CFA.The cluster of this crackles lies in unexplored area of IDW-2CD phase domain.The data support and suggest to call this high pitched crackles which appears in some pulmonary disorder "Ultrafine".

RESPIRATORY SOUNDS DURING METACHOLINE CHALLENGE.

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The aim of this study is to quantify the correlation between respiratory sounds and respiratory flow, and to yield data in order to build an "automatic wheeze detector", wich will be employed in several clinical studies.
Lung sounds were detected in .02-2560 Hz frequency band using a contact sensor placed at the sternal notch and recorded on a digital tape with the respiratory flow signal and a PCG trace. Ten normal subjects underwent lung sounds recording both during tidal volume (at least 10 respiratory cycles) and during FVC manoeuvre. Ten subjects with bronchial hyperreattivity experienced the same procedure. These patients were forbidden from ingesting bronchodilating drugs for at least 48 hours before the test. All the subjects underwent metacholine bronchoprovocation test (cumulative method) starting from 20 and following every 4 minutes cumulative doses up to 800 . Metacholine challenge was stopped when a FEV1 drop ≥20% of basal occurred. Respiratory sounds were recorded during every metacholine challenge, both during tidal volume and FVC manoeuvre. Functional respiratory data were used as standard to indicate the presence of bronchial obstruction.
Stored signals were acquired off-line by a personal computer and subsequently split in segments and analized in the frequency and time domain. The parameter evaluated in the first phase are:
Time domain parameters
a) duration of sound event
b) duration of expiratory and inspiratory phases
c) ratio between a and b
d) sound crest factor
Frequency domain parameters (after computing of the power spectrum)
a) power spectrum bandwidth
b) frequency of maximum amplitude peak
c) peak in the high frequency zone with amplitude in excess to a predefined percentage of the mean power.

PREVALENCE OF BEDMATE REPORTED SNORING AND SLEEP APNEA IN A MALE POPULATION AGED 40-65 YRS. Sacco C, Braghiroli A, Donner CF. Clinica del lavoro Foundation, Institute of Care and Research, Division of Pulmonary Disease, Medical Center of Rehabilitation, I-28010 Veruno (NO), Italy.

Several epidemiological studies investigated the prevalence of respiratory sleep disorders in different populations showing discrepancies due to anthropometric and habit factors. Aim of the present study was to evaluate the prevalence of snoring (as main symptom of OSAS) in male patients aged 40-65 yrs living in our geographic area. A telephone poll of about 2750 persons representative of a population of 250.000 inhabitants, performed by one of the most important Italian agencies, selected a sample of 800 males. The telephone interviews were based on a questionnaire previously validated by a one year experience in our department; only the answers confirmed by a bedmate (93.1% of the sample) were used for the analysis. Sample characteristics were (mean): age (53 yrs), weight (75 Kg), height (172 cm). A score from 3 to 12 was assigned according to the importance of snoring, apneas or excessive nocturnal body movements. 0 are males sleeping alone.

Sample characteristics	Nr subjects	score 7-12	6-5	4-3	0
40-44 yrs	124	13.3%	38.2%	44.8%	3.7%
45-49 yrs	121	20.1%	35.7%	42.1%	2.1%
50-54 yrs	105	20.2%	36.1%	40.4%	3.3%
55-59 yrs	98	18.6%	40%	34.2%	7.2%
60-65 yrs	89	12.7%	33%	45.6%	8.7%
snoring	222	36.5%	63.5%	0%	0%
apneas	43	95.9%	4.1%	0%	0%
sleepiness	41	23.2%	32%	32.4%	12.4%
	prevalence of risk factors				
BMI >29 Kg/cm^2	142	65.9%	24.8%	23.7%	
smokers	202	45%	41.1%	31.8%	
wine drinkers	124	29.6%	22.8%	21%	
spirit drinkers	105	21.9%	21.8%	17.9%	
hypertension	86	21.9%	14.7%	13.9%	

Score correlates with BMI ($p<0.006$), smoking ($p<0.008$), wine consumption ($p<0.002$) and hypertension ($p<0.05$).

A COMPUTERIZED SYSTEM FOR LUNG SOUND ANALYSIS WITH VERSATILE SPECTRAL ANALYSES AND AUTOMATIC CRACKLE DETECTION

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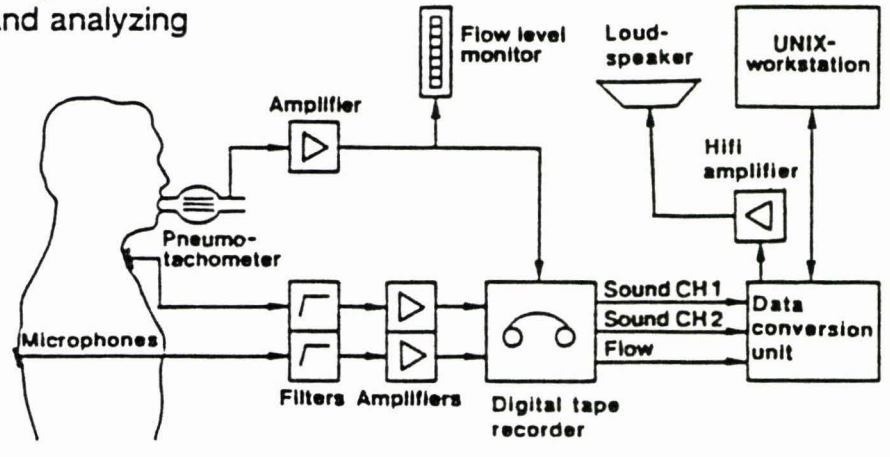
We have developed a versatile computerized method with automatic and interactive features for analyzing respiratory sounds. A block diagram of the recording and analyzing system is illustrated below (Fig. 1). Respiratory sounds are captured with two air coupled condenser microphones or with piezo-electric sensors. Air flow at the mouth is monitored with a pneumotachograph. The signals are recorded with a four channel digital tape recorder (TAEC RD-111T) connected with an antialiasing filter unit (TEAC T2-312 FA) with a frequency range from DC to 5 kHz. Sampling of the data is carried out with a HP-3852 data-aquisition unit with 12 bit D/A and A/D conversion. Sampling rate is 12 kHz for sounds and 100 Hz for air flow. A 100 Hz high-pass and 4 kHz low- pass filtration is used. Digitized signals are processed and analyzed with a HP-9000 workstation using a HP-UNIX operating system and a high-resolution monitor. The software is interactive, menu-driven and mouse-controlled with automatic and semiautomatic possibilites for sound analysis. The basic displays are: 1) phonopneumography, 2) expanded waveform, 3) FFT-analysis of user defined sample, 4) serial FFT- spectrum of successive fractions of lung sound cycles 5) phase-gated and averaged FFT of selected fractions of inspiratory or expiratory lung sound cycles, 6) serial FFT of different stages of a bronchial challenge or bronchodilation test, 7) sonagraphic frequency spectra (frequency-time display), 8) automatic crackle detection and calculation, 9) semiautomatic determination of single-crackle parameters (IDW, 2CD, LDW). The parameters from the FFT-spectra automatically calculated are: 1) frequency of maximal intensity, 2) the highest frequency at intensity level of - 20dB from the maximum intensity, 3) the upper frequency limit of the 1st, 2nd, 3 rd and 4th quartile of sound energy. Also RMS intensity of the sound and the average peak flow rate of the sample are automatically determined. The method has been used with good benefit for analyzing patients with crackling sounds and with obstructive diseases.

Grants: Paulo Foundation, Finnish Foundation Against Tuberculosis.

Fig. 1. A block diagram of the recording and analyzing system.



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On the sound transfer function of respiratory system in human

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Understanding the details of Acoustic transmission characteristics through the respiratory system is important in clarifying the possibilities of their clinical application. To estimate the sound transfer function of respiratory system, cepstrum analysis was applied.

The property of the cepstrum, which gives rise to a large number of applications, is its ability to separate sources and transmission path effects, i.e. to effect a "deconvolution". The relation shops between the input (the sources of Lung sounds) s(t) and output signal (the Lung sounds) y(t) of respiratory system can be expressed as:

$$y(t) = s(t) * h(t)$$

see(1): Lung sounds signals is the convolution of the sources of Lung sounds with the pules response of respiratory system.
or in the frequency domain.

$$Y(f) = s(f)\ H(f)$$

where h(t) is the pulse response of respiratory system, H(f) is the sound transfer function of respiratory system. Based on the cepstrum of Lung sounds, by using short pass lifter to separate sources (the sources of Lung sounds) and transmission path effects (the sound transfer function of respiratory system).

Fig.1. Show the sound transfer function of normal respiratory system and abnormal respiratory system. The experimental results show that Lung-thorax system could be simulated as a sound Low-pass filter, and that thorax system have a sound Low-pass characteristics, and that the sound transfer function of abnormal Lung-thorax system was significantly different from the sound transfer function of normal Lung thorax system.

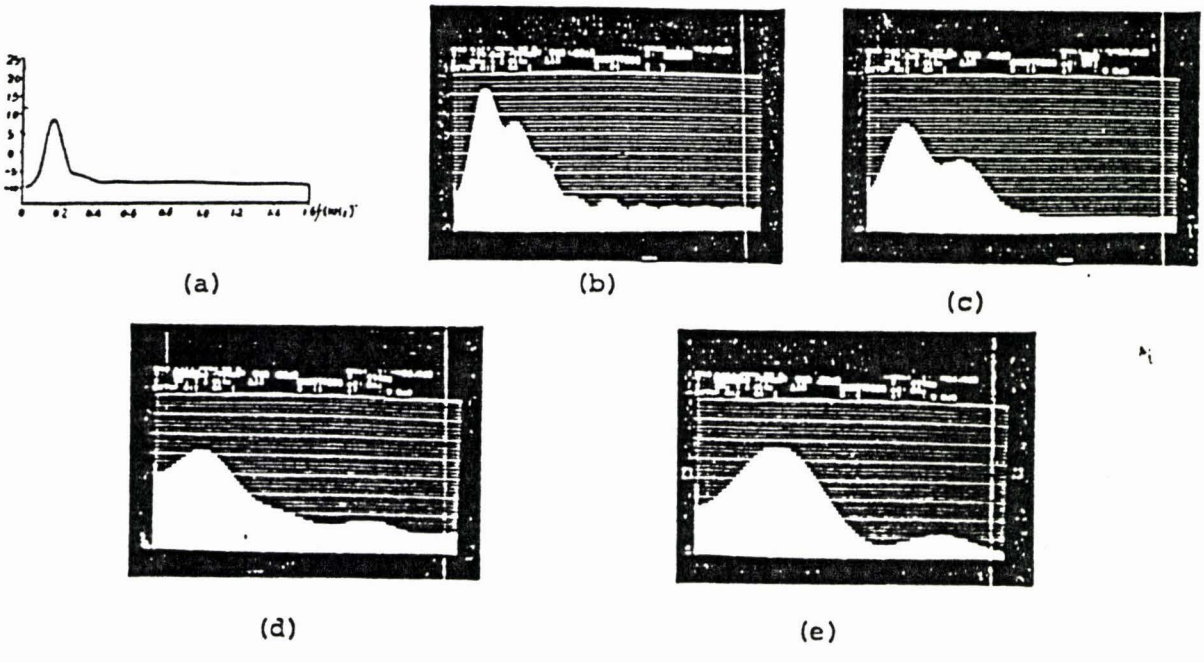

Fig.1: The sound transfer function of respiratory system. (Y axis: amplitude, 1dB/div, x axis, frequency, o.2KH$_Z$/div)
a. normal Lung-thorax system. b. bronchi-trachea-thorax system.
c. bronchi-trachea system. d.e. abnormal Lung-thorax system (crackles,wheezes).