Averaged and Time-Gated Spectral Analysis of Respiratory Sounds*

Repeatability of Spectral Parameters in Healthy Men and in Patients With Fibrosing Alveolitis

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Study objective: **To obtain a basis for assessment of changes in breath sound spectra in patients with pulmonary diseases, short-term and day-to-day repeatability of spectral parameters was studied.**
Design: **Breath sounds were recorded simultaneously from the trachea and from the chest twice at an interval of 15 min (short-term repeatability) and of 1 to 3 days (day-to-day repeatability). During recordings, air flow at the mouth was controlled, the target inspiratory and expiratory peak flow being 1.25 L/s. Inspiratory and expiratory breath sound spectra were averaged over 7 to 10 successive respiratory cycles. The repeatability of sound intensity (RMS), frequency of maximum intensity (Fmax), and median frequency (F50) was analyzed with analysis of variance.**
Participants: **Short-term repeatability was studied in 10 healthy nonsmoking men (age 25 to 44 years), and day-to-day repeatability was studied in 10 healthy nonsmoking men (age 23 to 41 years) and in 12 patients with clinically stable fibrosing alveolitis (age 35 to 82 years).**
Results: **Short-term coefficient of variation (CoV) of Fmax and F50 was 2.6 to 6.7% when recorded from the chest, and 6.2 to 8.7% when recorded from the trachea. Day-to-day CoV of Fmax and F50 in healthy subjects was 4.7 to 8.5% and 5.0 to 8.7% recorded from the chest or from the trachea, respectively. Inspiratory day-to-day variation in those parameters was higher in patients with fibrosing alveolitis. CoV of RMS was high, ranging from 18 to 47% in different subject groups and sampling situations.**
Conclusions: **Repeatability of F50 of averaged flow-controlled lung sound spectra is good both in healthy subjects and in patients with fibrosing alveolitis. Thus, F50 of respiratory sound spectra may be useful in monitoring of changes induced by respiratory diseases and interventions. These results emphasize the importance of standardization of recording conditions and of analyzing techniques.**
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Key words: fibrosing alveolitis; repeatability; respiratory sounds; spectral analysis

Abbreviations: CI=confidence interval; CoV=coefficient of variation; F25, F50 and F75=upper frequency limit for the first, second (median frequency), and third quartile of spectrum power, respectively; F50/1, F50/2, and F50/3=the early-, mid-, and end-inspiratory or expiratory F50, respectively; FFT=Fast Fourier Transformation; Fmax=frequency of maximum intensity; PTEF=peak tidal expiratory flow; PTIF=peak tidal inspiratory flow; RMS=root mean square value of sound amplitude; TE=duration of expiration; TI=duration of inspiration

Recent development in signal processing methods has improved the possibilities of extracting physiologically and clinically relevant information from respiratory sounds. Methods such as phonopneumography,[1] frequency spectral analysis with Fast Fourier Transformation (FFT),[2] and time-expanded waveform analysis[3] are at the moment commonly used. Advanced computerized analyses such as respirosonograms with illustrative displays in the time domain,[4] flow-standardized spectral analyses,[5] and calculation of background noise spectrum at zero-flow[6] have improved the possibilities of standardizing methods for recording and analysis of breath sounds, and enhanced their potential for characterizing changes in respiratory sounds occurring in pulmonary disorders of different pathophysiologic conditions. Several investigators have proposed clinical implications for respiratory sound anal-

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Table 1—*Mean (Range) of Anthropometric and Lung Function Data for Healthy Subjects and Patients With Fibrosing Alveolitis**

	Short-term Repeatability: Healthy Men	Day-to-day Repeatability	
		Healthy Men	Fibrosing Alveolitis
No.	10	10	12
Age, yr	37 (25-44)	33 (23-41)	62 (35-82)
Height, cm	180 (167-194)	178 (173-188)	172 (154-183)
Weight, kg	77 (61-115)	78 (65-106)	76 (64-97)
FEV_1, L	4.76 (4.04-5.50)	4.94 (3.96-5.73)	2.56 (1.65-4.33)
FEV_1, % predicted†	104 (87-140)	107 (89-128)	72 (56-95)
FVC, L	5.76 (5.08-6.87)	5.82 (4.47-7.05)	3.36 (2.01-5.78)
FVC, % predicted†	103 (82-129)	104 (83-134)	74 (55-100)
Dco, % predicted†	—	—	48.9 (34-97)

*Dco=diffusing capacity of lungs for carbon monoxide.
†Viljanen et al.[18]

ysis in their studies concerning crackling sounds in asbestosis[7] and some other pulmonary disorders,[8] respiratory health screening by use of lung sound analysis,[9] and spectral analysis of wheezing during challenge tests.[10]

Recently, interest has been focused on changes in frequency distribution of breath sounds in asthmatic patients with and without wheezing sounds[11-13] and without significant airways obstruction.[5] By use of time-gated and averaged spectral analysis, a significant correlation between changes in FEV_1 and frequency content of breath sound spectra in patients with asthma during histamine-induced bronchoconstriction has been demonstrated both in adults[13] and in children.[14] In the studies cited above, quartile frequencies introduced by Kraman[15] have served to characterize the frequency distribution of breath sounds.

Knowledge of the repeatability of commonly used spectral variables is essential when these are used to monitor the effects of respiratory diseases or interventions such as bronchodilator or challenge tests. However, at present, there are only a little data available on the repeatability of lung sound characteristics.[16,17] To our knowledge, no reports have been published concerning short-term or day-to-day variation in spectral parameters used in the investigations cited above among normal subjects or in patients with pulmonary diseases.

In this article, we present short-term and day-to-day repeatability of time-gated and averaged spectral parameters of air flow-controlled tracheal and lung sounds in healthy subjects and in patients with fibrosing alveolitis in order to obtain a basis for their use in monitoring changes in respiratory sounds.

Materials and Methods

The short-term repeatability of the frequency spectral variables of breath sounds was studied in ten healthy nonsmoking men. Their anthropometric and lung function data are presented in Table 1. Two recordings of breath sounds were made for each subject at an interval of 15 min. The day-to-day repeatability of the breath sound variables was studied in an other group of 10 healthy nonsmoking men and in 12 patients with stable fibrosing alveolitis (Table 1). Recordings were performed twice at an interval of 1 to 3 days at the same time of day. On all occasions, breath sounds from the trachea and the chest and air flow at the mouth for 7 to 10 respiratory cycles were recorded simultaneously with two microphones.

The healthy nonsmoking subjects were volunteers from among the personnel of Helsinki (Finland) University Central Hospital, having no history of any cardiorespiratory disease or atopy. At the time of the study, they showed no symptoms or signs of any illness. Conventional auscultation revealed no abnormality in their lung sounds, and results from flow-volume spirometry were normal for all of them (Table 1). They were taking no medication of any kind.

The diagnosis of the patients with fibrosing alveolitis was based on clinical history, findings on physical examination, chest radiograph, flow-volume spirometry, and pulmonary diffusing capacity test, and on analysis of BAL fluid. Eleven patients (92%) were receiving peroral glucocorticoid treatment and one used azathioprine. The severity of the disease was regarded as slight or moderate. For 11 of them, the pulmonary diffusing capacity for carbon monoxide was decreased (below 75% of the predicted[18]) and 7 showed a restrictive change in static spirometry with the helium dilution method (total lung capacity below 80% of the predicted[18]). All patients had both inspiratory and expiratory crackles in the recorded lung sound samples; on average, 8.8 crackles were detected during an inspiration and 4.3 crackles during an expiration with an automatic crackle counter (see below).

Informed consent was obtained from all subjects and patients and the study was approved by the ethics committee of the Department of Pulmonary Medicine at Helsinki University Central Hospital.

A schematic block diagram of the recording and analyzing system is presented in Figure 1. This method has been described in detail elsewhere.[19] The recordings were performed in an ordinary quiet laboratory room with the subject sitting in a chair. Before recordings, the subjects rested in the laboratory for at least 15 min. For lung sound recording at the chest, an air-coupled condenser microphone (B&K 4134; Bruel & Kjaer; Copenhagen, Denmark) with a preamplifier (B&K 2619) was used. The microphone was encased in the cup of a hearing shelter to reduce external noise and attached with a rubber belt to the subject's chest. The dimensions of the slightly conical coupling cavity (diameter 17 mm, depth 5 mm) conforms to international standards,[20] ensuring that the resonance frequency is above the measuring frequency band. The microphone was placed on the right lower lobe area about 10 cm below the inferior margin of the scapula, and about 15 cm to the right of the spine where the sound signal was usually best heard. A

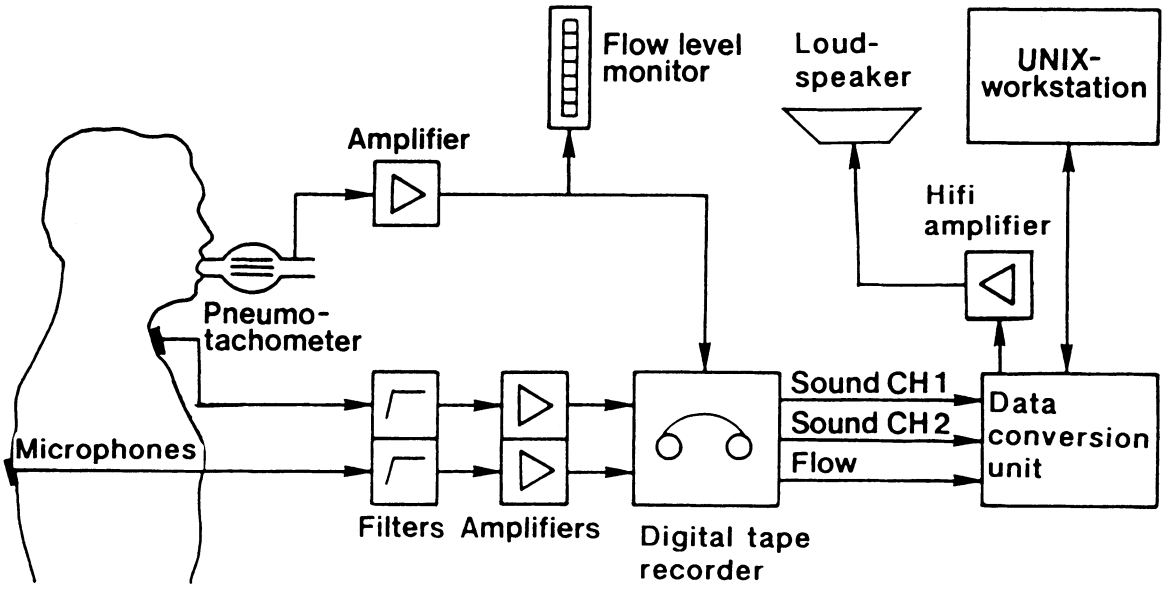

FIGURE 1. Block diagram of recording and analysis system for lung sounds in the present study.

small piezoelectric contact sensor (PPG Sensor No. 102; Technion; Haifa, Israel) was used for the recording of tracheal sounds. This microphone was firmly held by hand beside the cricothyroid cartilage, about 2 cm to the right of the midline. The outer surface of the sensor was covered with noise-reducing plastic material and a thin layer of sound transmission gel (Aquasonic) was used to prevent friction noise. The sound signal was prefiltered with a passive third-order high-pass filter with a cutoff frequency of 50 Hz to prevent the amplifiers being saturated by low-frequency noise.

The location of the microphone was marked on the skin for the repeated recordings. During the breath sound recording, air flow at the mouth was also recorded and monitored with a pneumotachograph (Medikro MF S202; Medikro Oy; Kuopio, Finland). The monitoring was accomplished with a center-zero analog monitor in front of the subject; the subjects were asked to breathe with similarly repeating pattern and to keep their peak tidal expiratory and inspiratory flows at the target flow level of 1.25 L/s by visual feedback. The mouthpiece was held tightly in the mouth, and the nose was closed with a clamp.

The sound and flow signals were recorded with an eight-channel recorder (DAT recorder; Teac RD-111T). The analog output signal of the recorder was digitized in a data acquisition and control unit (HP 3852A) with a 13-bit analog-to-digital conversion; the sampling rate was 12 kHz for sound and 100 Hz for the air flow. An analog low-pass Bessel filter with a cutoff frequency of 4 kHz (24 dB/oct) was used to prevent aliasing. The data from flow and sound signals were stored on a magneto-optical disk of a work station (Unix) (HP 9000/330C). The sound signal was high-pass filtered (digital Kaiser-FIR filtering) with a cutoff frequency of 100 Hz (24dB/oct).

The software developed for spectral analysis[19] is interactive, menu driven, and mouse controlled with automatic and semiauto-

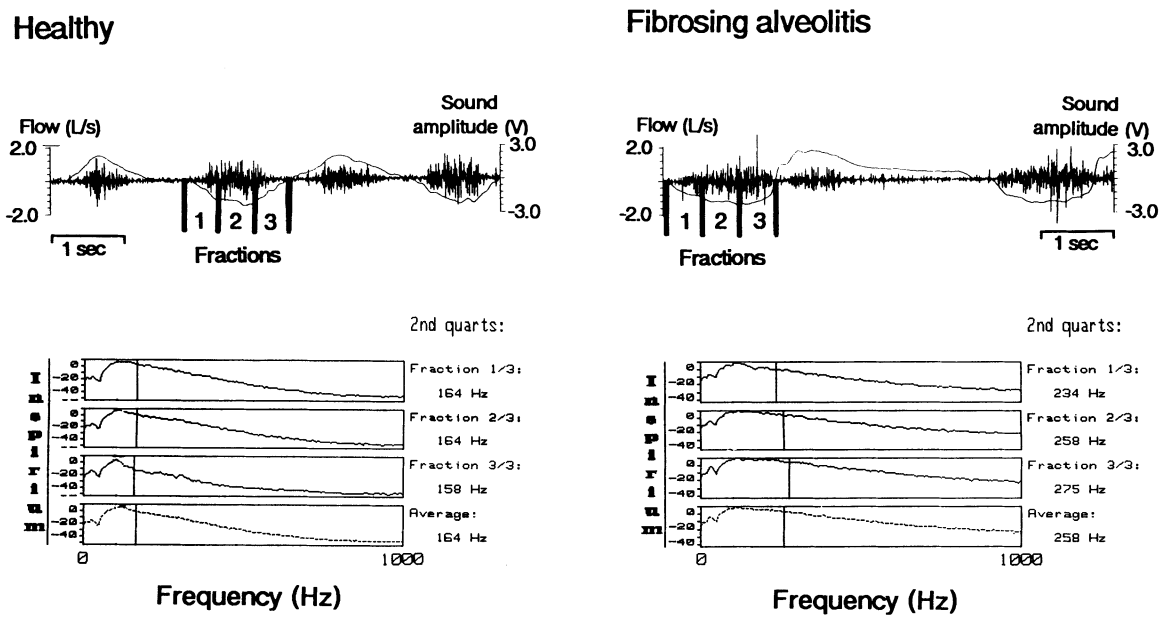

FIGURE 2. Examples of phonopneumograms and averaged fractional FFT spectra of a healthy subject and a patient with fibrosing alveolitis. Inspiratory fractions are indicated in phonopneumograms and in FFT partial spectra, respectively.

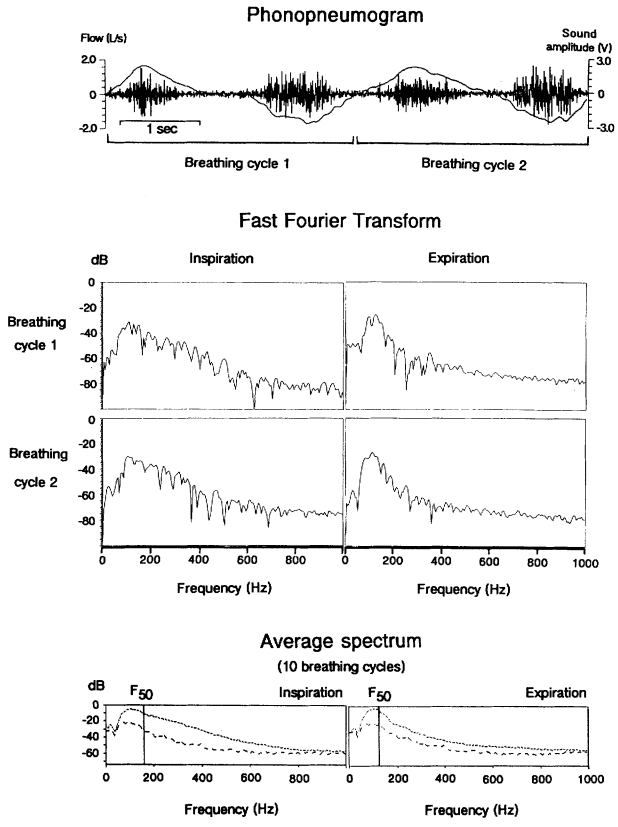

FIGURE 3. Phonopneumogram (two breathing cycles), midinspiratory and midexpiratory FFT spectra of corresponding cycles and averaged spectra from ten midinspiratory and midexpiratory breathing cycles with indication of background spectra (dotted curves) for a healthy subject.

matic analyzing possibilities. The basic displays for analysis are the phonopneumogram (sound amplitude and air flow rate in the time domain), expanded waveform, fast Fourier transformation (FFT) of a selected sound sample, serial FFT spectrum of a respiratory cycle, and the sonagram (frequency-spectrum in the time domain). Automatic detection and counting of crackles are also available.[21] This method was used to count inspiratory and expiratory crackles in our patients with fibrosing alveolitis; crackles were counted from the first five expiratory and inspiratory breathing cycles of the lung sound samples used for spectral analysis (chest recording). Semiautomatic measurement of waveform characteristics of single crackles is possible as well but was not used for the present analysis.

To obtain a more detailed analysis of frequency spectra and pattern changes during the respiratory cycles and to minimize the effect of biological and random variation on the spectra, software to calculate time-gated and averaged frequency spectra was developed. In this analysis, the sound signal was first divided into inspiratory and expiratory phases discriminated by the zero crossing points of the air flow curve. The inspiratory and expiratory phases were subdivided into three fractions or "gates" of equal time lengths corresponding to early-, mid-, and end-inspiratory or -expiratory fractions (Fig 2). Based on 2048-point FFT with a Hanning window, power spectra of each respiratory phase and fraction were computed by use of the overlapped-segment (50% overlapping) method introduced by Welch[22] and further averaged over the corresponding phases and fractions of the whole sound sample of 7 to 10 inspiratory and expiratory cycles (Fig 3).

The parameters automatically calculated for characterizing the average spectra in each phase included the upper frequency limits for quartiles of spectrum energy (F25, F50 or the median frequency, and F75). To limit the amount of data in this article, only the variation in F50 is presented. The median frequencies of the early-, mid-, and end-inspiratory or -expiratory fractions, F50/1, F50/2, F50/3 (Fig 2), respectively, were also calculated, as well as the frequency of maximum intensity (Fmax). For the spectral estimates, the measuring band of 100 Hz to 2 kHz was used, the spectral resolution being 5.86 Hz. The intensity of the sound was estimated by use of the root mean square (RMS) value of the sound signal. The average values of peak tidal inspiratory flow (PTIF) and peak expiratory air flow (PTEF) and the duration of inspiration (T_I) and expiration (T_E) over the respiratory cycles during recordings were also calculated.

The effect of averaging on breath sound spectra of a healthy subject is demonstrated in Figure 3. Two midinspiratory and midexpiratory FFT spectra (length, 100 ms) from successive respiratory cycles are presented: variation from breath to breath is evident. In contrast, an average spectrum of ten midinspiratory and midexpiratory fractions from ten successive cycles contains much less frequency-by-frequency variation. The spectra are smooth in shape and the average frequency distribution can be estimated easily. The differences in sound spectra calculated from early-, mid-, and end-inspiratory fractions are demonstrated in a healthy subject and in a patient with fibrosing alveolitis in Figure 2.

The intraindividual variance (σ^2) in paired measurements (x_1 and x_2) of RMS, Fmax, F50 and F50/1, F50/2, and F50/3 from expiratory and inspiratory sounds and that of PTIF, PTEF, T_I, and T_E of simultaneously recorded air flow signals was assessed by analysis of variance:

$$\sigma^2 = \sum_{i=1}^{n} (x_{1i}-x_{2i})^2/2n,$$ where n is the number of subjects in the group.

The results were expressed as the coefficient of variation (CoV) by the following formula: CoV=(σ/mean)*100, where the mean is the overall average value of the measurements in the group. The 95% confidence intervals (CI) of CoV were also determined.[23] The CoV values were regarded as significantly different if the mean value of

Table 2—*Overall Mean (SD) and Repeatability (CoV) of PTIF, PTEF, T_I, and T_E During Breath Sound Recordings in Healthy Subjects (n=10) and in Patients With Fibrosing Alveolitis (n=12)

	PTIF		PTEF		T_I		T_E	
	Mean (SD), L/s	CoV, %	Mean (SD), L/s	CoV, %	Mean (SD), s	CoV, %	Mean (SD), s	CoV, %
Healthy (short-term)	1.42 (0.06)	3.76	1.45 (0.08)	3.43	1.50 (0.91)	13.8	1.98 (1.15)	14.9
Healthy (day-to-day)	1.46 (0.12)	4.46	1.41 (0.16)	5.10	1.61 (0.65)	9.33	1.53 (0.58)	8.72
Fibrosing alveolitis (day-to-day)	1.37 (0.11)	8.39	1.47 (0.12)	12.1	1.42 (0.43)	9.67	1.77 (0.64)	12.3

*Intraindividual variation (CoV) was assessed from duplicate measurements at an interval of 15 min (short-term) and of 1 to 3 days (day-to-day).

Table 3—*Overall Mean (SD) and Short-term Variation (CoV) of RMS, Fmax, F50, F50/1, F50/2, and F50/3 in Healthy Men (n=10)**

	RMS	Fmax	F50	F50/1	F50/2	F50/3
Overall mean (SD), Hz						
Trachea/insp	297 (238)	99 (8)	524 (152)	494 (153)	542 (166)	413 (137)
Trachea/exp	257 (171)	96 (11)	576 (128)	551 (147)	601 (136)	389 (141)
Chest/insp	244 (159)	109 (11)	171 (23)	174 (22)	173 (23)	164 (28)
Chest/exp	144 (89)	103 (14)	146 (18)	152 (23)	141 (17)	151 (20)
CoV (CI), %						
Trachea/insp	25 (12-38)	9.1 (4.4-14)	8.7 (4.2-13)	14 (6.9-22)	13 (6.5-20)	19 (9.0-28)
Trachea/exp	35 (17-53)	7.4 (3.6-11)	6.2 (3.0-9.5)	6.2 (3.0-9.5)	4.5 (2.1-6.8)	19 (9.1-29)
Chest/insp	18 (8.4-27)	4.9 (2.4-7.5)	2.6 (1.2-39)	3.8 (1.5-4.7)	3.1 (1.5-4.7)	2.5 (1.2-3.7)
Chest/exp	22 (10-33)	6.7 (3.3-10)	4.4 (2.1-6.6)	9.3 (4.5-14)	6.1 (2.9-9.3)	6.8 (3.3-10)

*Intraindividual variation (CoV) of the parameters was assessed from duplicate measurements (interval, 15 min). Insp=inspiration; exp=expiration.

CoV in a group of parameters was beyond the 95% CI of that parameter in another group.

Results

The overall mean values and intraindividual variation of peak tidal air flow values and the duration of respiratory cycles during breath sound recordings are shown in Table 2. The CoV of PTIF and PTEF in healthy subjects was around 4% and that in patients with fibrosing alveolitis, 8 to 12%. The mean PTIF and PTEF during recordings varied between 1.37 L/s and 1.47 L/s. Compared with healthy individuals, the patients with fibrosing alveolitis had slightly lower PTIF values during the recordings (p=0.02). The CoVs of the duration of breathing cycles were higher (9 to 15%) than those of the tidal peak flow values (3 to 12%).

The short-term and day-to-day variations in RMS, Fmax, and F50 in successive inspiratory and expiratory breathing cycles and their fractions in healthy men and in patients with fibrosing alveolitis are presented in Tables 3 and 4 and in Figure 4. The intraindividual variation in the spectral parameters (Fmax and F50) was small; recorded at the chest, the short-term CoV of F50 varied from 2.6 to 4.4% and the day-to-day CoV from 5.0 to 8.5% in healthy subjects. Inspiratory day-to-day variation in Fmax and F50 in patients with fibrosing alveolitis was higher than that in healthy subjects when recorded at the chest (Table 4, Fig 4).

The short-term variation in all the parameters for sounds recorded at the trachea tended to be higher than for those captured at the chest. Usually, the CoVs of F50 determined from fractions of respiratory cycles tended to be higher than those from the corresponding parameters obtained from whole inspiratory or expiratory cycles (Table 3).

The upper limit of the 95% CI of CoV for short-term and day-to-day repeatability of F50 in healthy subjects was markedly small, up to 13% and only slightly higher in patients with fibrosing alveolitis, up to 15% (Table 4).

The intraindividual variation in intensity (RMS) for healthy men was high both in short-term (CoV, 18 to 25%) and day-to-day recordings (CoV, 27 to 47%). The day-to-day variation in RMS was almost similar for healthy people and for patients with fibrosing alveolitis (CoV, 23 to 47%). The variation in RMS recorded from the chest was smaller in short-term (CoV, 18 to 22%) than in day-to-day (CoV, 41 to 47%) recordings. In tracheal recordings, however, the short-term and day-to-day CoV of RMS was almost similar (25 to 35%). In patients with fibrosing alveolitis, the intraindividual CoV of the number of inspiratory crackles in repeated recordings was 27%.

Table 4—*Intraindividual Day-to-Day Variation (CoV) in Duplicate Measurements (Interval, 1 to 3 Days) of RMS, Fmax, and F50 in Healthy Men (H) and in Patients With Fibrosing Alveolitis (FA)*

	RMS		Fmax		F50	
	H	FA	H	FA	H	FA
CoV (CI)						
Trachea/inspiration	27 (13-42)	35 (19-51)	8.7 (4.2-13)	8.1 (4.4-12)	6.5 (3.1-9.9)	10 (5.5-15)
Trachea/expiration	33 (16-50)	23 (13-34)	8.2 (3.9-12)	6.7 (3.7-9.8)	5.0 (3.3-10)	8.2 (4.4-12)
Chest/inspiration	41 (20-63)	47 (25-68)	5.7 (2.7-8.6)	15 (8.1-22)	5.0 (2.4-7.5)	10 (5.5-15)
Chest/expiration	47 (23-72)	35 (19-50)	4.7 (2.3-7.2)	8.5 (4.6-12)	8.5 (4.1-13)	5.3 (2.9-7.7)

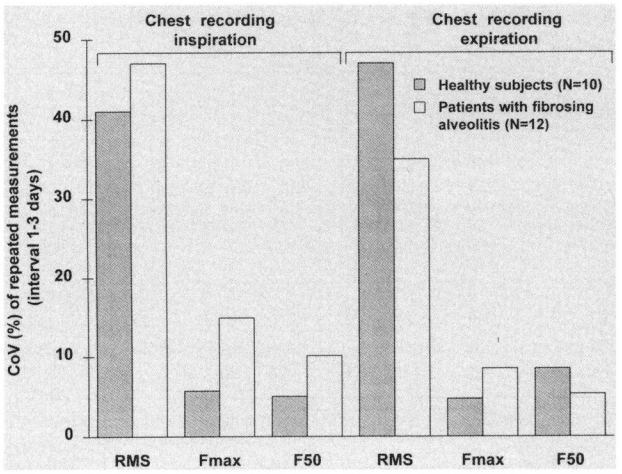

FIGURE 4. Graphic representation of coefficient of variation (CoV%) of RMS, Fmax, and F50 in repeated (intervals, 1 to 3 days) measurements of breath sounds from the chest during inspiration and expiration in healthy subjects and in patients with fibrosing alveolitis.

DISCUSSION

Knowledge of the repeatability of breath sound parameters is essential if such measurements are used on two or more occasions to monitor the effects of a respiratory disease, its treatment, or interventions such as bronchodilator or challenge tests. A test variable with a good discrimination property shows small within-subject variation both during the short term and long term. Variability in respiratory sound parameters obtained by computerized analyses derives from many sources. The respiratory sound signal itself has a nonstationary character during breathing. Variation in breathing pattern, air flow, and volume have an influence on the variability of respiratory sounds. The repeatability of spectral variables is further affected by many technical factors: variation in the location and attachment of microphones on the body, acoustical and electrical properties of the microphones, shielding and elimination of external and internal noise, filtering and amplification of the sound signal, sampling rate, signal processing and analysis methods, and by parameters selected for the analysis. All these items should be standardized to reduce variation and thus to make results of breath sound analyses comparable within and between laboratories.[24,25] There are as yet no international guidelines for standards of computerized lung sound analysis, but recently, a European scientific group sponsored by the European Community (CORSA project, BIOMED 1) has started work on this task.[26]

In the present study, several steps in the signal capturing, recording, and analysis chain were standardized to reduce the variation. The location of the microphones was defined and attempts made to keep it unaltered between the repeated recordings. The chest microphone was fixed at its location during the two recordings 15 min apart (short term) but not in day-to-day recordings. The location of the tracheal microphone may have been slightly changed between the repeated recordings, both in the short term and day to day, due to manual replacement, although the microphone location was marked on the skin. This fact may explain the slightly higher short-term CoV for tracheal recordings than for chest recordings in our study. This possible source of variability may also partly explain the slightly higher CoV for day-to-day than for short-term recordings.

Tracheal sounds were found to have a wider frequency band than sounds recorded from the chest, a finding in accordance with previous studies.[27] A detailed comparison of frequency characteristics and intensity of breath sounds over trachea vs chest, however, is not possible due to the different microphones used. An air-coupled microphone with shieldings was used at the chest, since lung sounds, particularly expiratory, have low intensity, requiring a sensitive sensor with high signal to-noise ratio. For tracheal sounds, the piezoelectric contact sensor was regarded as more suitable, due to smaller size and better frequency response characteristics at high frequencies compared with those of air-coupled microphones.[6,28] The variability in spectral parameters tended to be larger at the trachea, but due to the differences regarding the attachment and the type of microphone at the two locations, we are not able to draw any conclusions as to whether these differences are physiologic or technical in origin. At a given location, however, the same microphones with the same air coupling and shieldings for external noise were used in the repeated measurements. Thus, the microphones themselves may not have induced any significant variation in the spectral parameters tested.

The averaging of breath sound spectra over several expiratory and inspiratory breathing cycles has probably a most marked beneficial effect in reducing variation in spectral parameters (Fig 3). By averaging with an overlapped segment method, random breath-by-breath variation in spectra could be minimized. The number of breathing cycles averaged (7-10) was chosen based on our own earlier experiments. This amount of data ensures the stabilization of the spectral pattern and is reasonable regarding the memory requirements of the computer, as observed also by others.[6,9]

By high-pass filtering at 100 Hz, the variation induced by low-frequency noise, *eg*, from the heart or respiratory muscles,[16,29,30] could be attenuated. The measuring band ranged up to 2 kHz, a commonly used upper limit in spectral analysis.[6,17] In the present study, the sampling rate was high enough (12 kHz) to guarantee the statistical reliability of the intensity and spectral estimates of the sound.

The breathing pattern, flow, and volume during and between the recordings was kept as constant as possible by visual autofeedback; the subject was asked to reach the target peak expiratory and inspiratory flow (1.25 L/s) during the recording of sounds by following the flow monitor. The intraindividual CoV of peak tidal air flow values during the recordings was relatively small (3 to 5%) in healthy subjects but higher in patients with fibrosing alveolitis (8 to 12%). This may be one reason for the higher CoV values of spectral parameters in these patients. However, the CoV of the duration of breathing cycles in healthy subjects and in the patients was about the same. The averaging method has probably also attenuated the effect of breath-by-breath variation in air flow.

Lung sound spectra have been demonstrated to be quite stable in shape at flow rates above 1.0 L/s[15,31] but appear to be dependent on flow at lower levels.[5,32] In the present sound recordings, the measured peak tidal flow values were around 1.4 L/s, and with the tidal breathing pattern used, the averaged spectra were predominantly affected by the sound during high air flow rates. Controlling the air flow during recordings probably had a beneficial effect on variation in frequency distribution of breath sounds in each individual. The present method of flow-controlled respiratory sound analysis is the same as that used in earlier studies for assessing the effect on spectral parameters of induced bronchoconstriction.[13,14]

It has been shown previously that breath sound intensity is critically dependent on air flow.[33,34] In our study, the variation of intensity (RMS) was markedly higher (CoV around 20 to 40%) than that of the parameters of frequency spectra (5 to 14%). The reason could be partly the variation found—although not large—in the air flow from one recording session to the other. Flow- and volume-standardized analysis methods[35] could reduce this kind of variation, but data for this are not yet available. It must be pointed out that RMS was calculated from the time domain signal containing energy also beyond the 100-Hz to 2-kHz band used for calculation of the spectral analysis.

The present method enabled us to present background or zero-flow spectra for quality control, as suggested by Pasterkamp et al.[6] Due to efficient ambient noise reduction by the shielding of microphones, subtraction of the background spectra was not considered necessary for the present study to assess intraindividual variation in sound spectral parameters by the method previously used. However, since interindividual differences in background noises may exist, the background should be regarded in interindividual comparisons of breath sound spectra. Subtraction of background spectra, the frequency response of the microphones[6] and their air-coupling cavity dimensions,[28] amplification, filtration, analog-to-digital conversion, and the algorithms used for calculation of the power spectrum all have an effect on breath sound spectra.[25] The present results concerning actual mean values of spectral parameters are therefore not comparable to those obtained with methods utilizing a different kind of equipment or signal processing.[15,16,36]

The CoV of Fmax and F50 was very low in healthy subjects (4 to 9%) and only slightly higher in the patients with fibrosing alveolitis (5 to 15%). The higher variation among patients may be due to adventitious sounds, crackles, with high variation of occurrence, and to differences in the breathing pattern described above. The crackles occurred mostly during inspiration as was also found earlier in patients with fibrosing alveolitis.[8] Therefore, higher CoV was found in inspiratory spectral parameters. Variation in the process of the disease itself may also have contributed to the repeatability by increasing CoV. However, the disease of our patients with fibrosing alveolitis was stable in clinical terms.

The variation in F50 calculated from fractions of the respiratory cycles tended to be larger than that of averaged F50 for the whole cycles. Reasons for this may be the smaller number of samples during fractions, cycle-by-cycle changes in breathing pattern, and lower sound energy within fractions of expiratory cycles in healthy subjects.

Compared with that of a recent study by Mahagnah and Gavriely,[17] the CoV in our data is smaller. However, direct comparison is irrelevant, since in that study, the variation of point by point (frequency by frequency) of averaged spectra was calculated with no assessment of any single spectral parameter. Our CoV of F50 in healthy subjects was about the same as that found in the breath-by-breath study by Ploysongsang et al.[16] The CoV of F50 in healthy subjects in the short term was only slightly higher than that of FEV_1, the most repeatable spirometric variable.[37] In the study of Malmberg et al,[13] the within-subject variability in F50 for repeated breath sound samples during histamine provocation test on healthy nonresponding subjects was close to that observed in the present article.

In the present study, repeatability was determined both in healthy volunteers and in patients with stable fibrosing alveolitis. The latter group was chosen, because breath sounds in these patients are usually abnormal but relatively stable within short periods of time, and are not affected by changes in bronchial tone, which is often the object of monitoring. The variability in breath sound parameters for patient populations is probably better characterized by the CoV in patients with fibrosing alveolitis than by that of the healthy controls.

Repeatability of a set of breath sound parameters was studied to obtain a basis for the previous and fu-

ture studies concerning changes of respiratory sounds in pathologic conditions and experiments. Recent observations[11-14] about the relationship between FEV_1 and F50 of breath sounds during challenge tests in asthma suggest that respiratory sound analysis could serve as an alternative method for monitoring of bronchial obstruction in patients with limited cooperation in spirometry. For the development of such clinical applications, knowledge of the repeatability of lung sound measurements is particularly important. According to our results, short-term changes in F50 by 6% and 17% in inspiratory sounds recorded at the chest or at the trachea, respectively, can be considered significant (1.96 times CoV).

The present results indicate that variability in median frequency of the averaged lung sound spectra both in the short term and day to day in healthy people and in patients with fibrosing alveolitis is small enough for monitoring of changes of lung sound spectra when recorded as flow controlled. These estimates of averaged respiratory sound spectra may be useful in studies concerning effects on breath sounds of respiratory diseases, their treatment, and interventions, and in possible future applications of respiratory sound analysis in clinical settings.

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