This is to certify that the

thesis entitled

AN INSTRUMENTATION SYSTEM FOR QUANTITATIVE STUDIES OF RESPIRATORY SOUNDS

presented by

Sui-Ming Huang

has been accepted towards fulfillment of the requirements for

Master degree in Science

Major professor

Date Apr 29, 1980

AN INSTRUMENTATION SYSTEM FOR QUANTITATIVE STUDIES OF RESPIRATORY SOUNDS

By

Sui-Ming Huang

A THESIS

Submitted to
Michigan State University
in partial fulfillment of the requirements
for the degree of

MASTER OF SCIENCE

Department of Electrical Engineering and System Science

1980

ABSTRACT

AN INSTRUMENTATION SYSTEM FOR QUANTITATIVE
STUDIES OF RESPIRATORY SOUNDS

By

Sui-Ming Huang

This thesis is concerned with the development and testing of an instrumentation system for quantitative and qualitative studies of respiratory sounds produced during a respiratory cycle. With this system it is possible to record sounds from the chest wall or trachea and correlate them with physical variables such as air flow rate and lung volume. Breath sounds were detected by a microphone, passed through a lowpass filter, digitized, and finally displayed in a comprehensive form using a x-y plotter. Digital data processing was conducted on a Hewlett Packard HP9825A calculator using a fast Fourier transform algorithm. By dividing a respiratory cycle into many small time segments, the system can provide detailed information about frequency, intensity, air flow and timing of acoustic events at any point in time during inspiration or expiration.

ACKNOWLEDGEMENTS

I wish to express my gratitude to Dr. Richard Hallgren for his effort and assistance in connection with my studies. His support and initial confidence in me has given me the opportunity to gain invaluable experience.

I would also like to thank my parents and my wife for encouraging me to undertake this project.

TABLE OF CONTENTS

LIST OF FIGURES

CHAPTER I

INTRODUCTION

Bronchial asthma, emphysema or chronic bronchitis are generally described as obstructive lung diseases, and constitute a rapidly growing problem in contemporary medicine. Since 1960, the death rate for these diseases has increased 43 percent, and this is the highest increase in the death rate for any major disease. For comparison, the death rate associated with cirrhosis of the liver increased 41 percent, with cancer 12 percent and with diabetes 9 percent. [(28)]

Present diagnostic methods of detecting pulmonary diseases are based upon roentgenoscopy, auscultation and function testing with an increasing emphasis on roentgenoscopic methodology during the last four decades. This tread is related to the continuing progress in this area and increased precision in x-ray diagnosis. However, since obstructive diseases affect the airways, they are not easily diagnosed using x-ray techniques, and thus interest in auscultation has increased.

In spite of the fact that auscultation is used in routine clinical diagnosis, there have been no major developments in this method since the invention of the stethoscope by Laennec in 1816. Auscultation is presently based upon non-uniform and subjective phraseology rather than on quantitative data describing energy and frequency distribution during the respiratory cycle.

The sounds generated in the lungs have the potential for providing valuable information about abnormal or normal pulmonary functions. Thus a modified quantitative auscultation method may become a basic, simple, non-invasive diagnostic technique in pulmonary medicine. Respiratory sounds have their origin in the mechanical events which occur in the airways during the respiratory cycle. The nature, timing and distribution of these sounds depend on the flow rate, regional volume distribution, and on mechanical characteristics of the airways and the surrounding tissue. However, very little is known about the mechanism of sound generation and transmission in the lungs, tracheobronchial tree, and surrounding tissue.

The purpose of this thesis was to develop an instrumentation system that would record, digitize and process respiratory sounds for quantitative and qualitative studies. The utility of the system has been demonstrated by collecting data from several healthy male subjects. Future studies will be concerned with the correlation of the characteristic sound data to air flow rate, microphone position, age, weight, and height.

CHAPTER II

RESPIRATORY SOUNDS

2.1 General Discussion

Sounds recorded over the chest contain complex waveforms associate with cardiovascular, respiratory and muscular events. Sounds generated in the lungs and tracheobronchial tree have been described as breath sounds, lung sounds, respiratory sounds, pulmonary sounds or respiratory noises. Recently, the method of recording and analyzing of sounds associate with the respiratory cycle has been named phononeumography [10] or phonorespirography. [9]

Acoustic phenomena of the respiratory system have been studied fragmentarily since the invention of the stethoscope by the French physician Laennec in 1816, who provided the basic terminology and description of the pathology of tuberculosis, and William Strokes (1837), who used a stethoscope as a diagnostic instrument. [5] As a result of these classic works, auscultation became the basic diagnostic method used in characterizing pulmonary diseases until the development of x-ray techniques. In spite of the rapid development of radiographic techniques there have been continued efforts to record and characterize breath sounds and to correlate them with normal conditions and with diseases. The remaining sections of this chapter summarize these results.

2.2 Classification of Respiratory Sound

R. Murphy [1] has described one of the most organized classifications of lung sounds. He has separated lung sounds into normal breathing sounds that occur when no respiratory problems exist and abnormal or adventitious sounds. Normal breath sounds can be divided into two types. The first type is tracheal or bronchial breath sounds which can be heard over the trachea or near the large airways. The second type is vesicular breath sounds which can be heard in the chest of most people as they breathe normally. Bronchial sounds have their origin in large airways which are greater than 4 mm in diameter. [3] The intensity of these sounds are directly related to an increase in the vital capacity and are the loudest at the end of expiration. Vesicular sounds are a function of the air flow in small airways and alveoli. In particular, vesicular murmurs are caused by turbulance of air flow entering the alveoli. Intensity of vesicular sounds are lower during expiration than during inspiration. Visual examples of typical recorded data in normal subjects are shown in Figures 1, 2 and 3. [6]

Bullar has conducted extensive experimental work to determine the origin of the respiratory sounds. [2] He has

Figure 1. Bronchial breathing (trachea). Expiration is louder, higher pitched and longer than inspiration. In this and other figures intensity is displayed on the Y-axis and time on the X-axis.

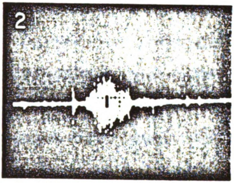

Figure 2. Bronchovesicular sounds (middle lobe). Intensity of expiration is markedly reduced with a duration equal to inspiration.

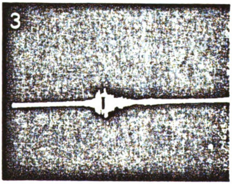

Figure 3. Vesicular sounds (base). Inspiration is louder, of greater duration, and higher pitched. From Weiss and Carlson.[6]

investigated three basic theories concerning the origin of respiratory sounds: 1) Sounds are generated as a result of the friction of air against walls of airways; 2) sounds are produced at the glottis; 3) sounds result from air flow through airway narrowings. After testing these theories he has concluded that vesicular sounds are produced in the lung but not in the glottis or trachea.

Adventitious sounds can also be divided into two types. The first type is rales or crackles, they are present for very brief periods of time, usually less than 10 milliseconds and thus called "discontinuous." The second type is rhonchi or sibilant, in contrast, they are present for much longer periods of time, usually more than ¼ second and are called "continuous." Rales are further divided by their quality into fine, medium and coarse. Rhonchi are further divided by their pitch into sibilant or high pitched and sonorous or low pitched sounds.

Bethke and Seireg [4] have defined adventitious sounds as "abnormal sounds produced by secretions or exudates in the lung or by structural changes, such as cavities or plural roughness." Usually, the presence of adventitious sounds is an indication of the presence of lung disease. [5]

2.3 Frequency Spectrum of Respiratory Sounds

There is much controversy concerning the maximum frequency component of breath sounds. Fahr [7] has reported that respiratory sounds are in the range 80-800 Hz and in some cases can have components up to 1,000 Hz.

Cabot and Dodge [8] in 1925 determined the frequency characteristics of heart and lung sounds using band-pass filter analysis. They reported that lung sounds consist of higher frequency components than heart sounds, and they identified the characteristic frequency components of certain types of breath sounds. According to their results, rales were found to contain frequency components in the range 120-1,000 Hz with coarse rales containing frequencies above 660 Hz, amphoric breathing had characteristic frequencies in the range 240-660 Hz, and bronchial breathing contained frequency components between 240 and 1,000 Hz. Thus, their total frequency range was 120-1,000 Hz, but they stated that sounds above 660 Hz had no significant diagnostic value.

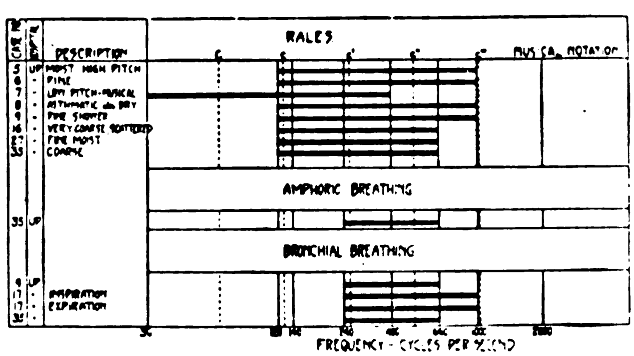

Figure 4. Respiratory sounds frequencies from Cabot and Dodge. [8]

Nachman et al., [3] while studying two subjects, concluded that respiratory sounds consist of irregular vibrations which place respiratory sounds in the category of noise. It is worth noting that they presented spectrograms of breath sounds which include frequencies in the range between 50 and 4,000 Hz, although most authors indicate 1,000 Hz as an upper frequency limit.

According to Meyer et al., [9] inspiratory breath sounds have characteristic frequencies up to 700 Hz which during expiration can reach 800 Hz. Bethke and Seireg [4] determined that the upper frequency limit of vesicular sounds is 1,250 Hz and have shown that an increase in intensity of sound with increasing flow rate is associated with frequencies between 125 and 500 Hz. Banaszek [10] found that most of the noticeable breath sounds have frequency components below 500 Hz. Fahr [7] has found that "A not considerable portion of the sound energy consisted of vibrations between 660-1,000 Hz." He also determined that the fundamental frequency of bronchial sounds is in the range of 300-500 Hz, and some overtones consist of frequencies up to 1,000 Hz. Wooten and Waring (1972) conducted a spectral analysis of respiratory sounds in pediatric patients. [11] In normal subjects they observed frequency components below 1,000 Hz with intensity varying in the range from 30 to 60 dB, while above 1,000 Hz the sound pressure level was below 35 dB.

As a result of the above discussing, the frequency range for this experiment was restricted between 120-1,000 Hz.

2.4 Correlation of Respiratory Sounds With Pulmonary Variables

Investigation of breath sounds should include correlation of their intensity and frequency spectrum with pulmonary variables and parameters. Most common studies include correlation of breath sounds with expiratory or inspiratory flow rate, lung volume, and airway resistance or compliance. The intensity of breath sounds is related to flow rate [4,10,12] and maximum intensities are probably associated with maximal alveolar filling which occurs with maximum inspiratory flow rates. [9] Higher frequency components of breath sounds have been recorded in subjects with higher respiratory flow rates. [4]

Differences in intensity and quality of breath sounds at different respiratory flow rates are associated with the character of the flow. Air flow in upper airways is turbulent at higher flow rates and this corresponds to respiratory sounds of higher intensity. Low inspiratory flow rates are associated with laminar air flow, and thus result in lower intensities. [7]

Higher intensities of sounds, recorded in patients with asthma and chronic bronchitis, for a given change in flow rate are attributed to narrowing of airways and an increase in turbulence of air flow. [12] Diagnostic attempts using breath sounds have utilized both the inspiratory and expiratory phases. Most studies have been concerned with inspiratory breath sounds because expiratory breath sounds are the result of bifurcation in the larger airways and are poorly transmitted to the surface of the thorax.[3] There are different opinions concerning the relationship between intensity of respiratory sounds and the phase of the respiratory cycle. Some investigators have found that sound intensity is greater during expiration than during inspiration,[7,9] while others have found the opposite to be true.[4,12]

Lablanc et al.[13] correlated sound intensity and frequency with various flow rates and lung volumes for different body positions. They reported that lung volume and body position influenced breath sounds, and that the intensity of sound increased with increasing flow rates. They found that for a given inspiratory flow and a given body position, the maximum relative intensity of breath sounds was always recorded at low lung volumes over superior regions of the lungs and at larger lung volumes over inferior regions. In contrast to Nachman et al.[3] their studies showed that maximum sound intensity over the upper lung field during inspiration was associated with the beginning of inspiration and was independent of body position. They have also found that breath sounds related to expiration have higher intensities than those related to inspiration.

Banaszak[10] correlated sound intensity and spectral content with respiratory flow and volume. He has shown that the intensity of respiratory sounds recorded over the thorax varies with the flow rate in normal subjects. The recorded breath sounds contained frequencies up to 500 Hz and the intensity of these sounds decreased with increased lung inflation.

Inflation of the lung is usually associated with an increase of intensity of sound at all frequencies during inspiration and at higher frequencies during expiration. [12]

2.5 Respiratory Sounds in Disease

Breath sound analysis can provide information related to obstruction of airways associated with bronchial asthma, chronic bronchitis, pulmonary emphysema and other airways disease. No precise correlation between specific disease and the associated sounds has been established, and there is a good deal of controversy related to the description of acoustical phenomena associated with specific pulmonary diseases.

Most of the studies have been performed upon individuals with emphysema because the associated changes occur in vesicular sounds are louder and more easily detected. Most investigators have agreed that the sound intensity in patients with emphysema is higher during expiration than during inspiration and these sounds generally have higher intensity and higher frequency components, than sounds from normal subjects. [4,12,14] The intensity of breath sounds in patients with emphysema is distributed throughout the total respiratory cycle, and is not specifically associated with maximum flow as it is in the normal subjects. [15]

Bethke and Seireg [4] studied vesicular lung sounds in normal subjects and those with emphysema and correlated sound power with respiratory flow rate. This work included classification of respiratory sounds. Using auto-correlation they found clear differences in the power density spectra between the normal and the emphysematous group. In subjects with higher respiratory flow rates there was a shift of power to higher frequencies particularly during inspiration. They also noticed a difference in the rate of change in sound power during inspiration and expiration in normal subjects and subjects with emphysema. The relationship between sound power and respiratory flow rate was linear on a logarithmic scale. They [4] suggested that the higher intensity of emphysematous sounds may be associated with better transmission because these patients are usually slim. In contrast, Forgacs [14] has found that lower intensity of breath sounds as heard through the chest wall is often listed among the signs of emphysema, and is generally attributed to poor transmission. Forgacs et al. [12] have not noticed any significant changes in the breath sound intensity in emphysematous patients.

Meyer [9] found that in asthmatics breath sounds in the range 200-400 Hz have higher intensity. Asthma and chronic bronchitis can be characterized by noisy respiration which is related to the narrowing of bronchi. [12] Forgacs et al. [12] related inspiratory breath sounds at the mouth to asthma and chronic bronchitis. They found that quiet inspiration usually is an indicator of emphysema. They correlated intensity of breath sounds and respiratory flow rates and found that patients with chronic bronchitis and asthma had higher increases in sound intensity for a given change in respiratory flow rate than did normal subjects. The relationship between flow rate and intensity of sound was nonlinear at low flow rates and linear at higher flow rates.

Nairn and Turner-Warwick [16] investigated breath sounds in emphysema. Specifically, they correlated breath sounds intensity quantified using a grading scale with the regional ventilation as determined by radioactive Xenon. It was found that the absence of breath sounds corresponded to regions of poor ventilation, which is a characteristic for emphysematous lungs.

Nemerovskii and Zelivyyanskaya [17] have studied obstructions of the respiratory airways by applying a 80 Hz acoustic signal at the mouth and recording transmitted sound over the thorax. In a preliminary study they compared the changes in transmission of sound in normal subjects and subjects with pulmonary diseases. They described only the method of investigation and had not included results.

CHAPTER III

SYSTEM ORGANIZATION

3.1 Characteristics of the System

The developed system records and analyzes breath sounds in humans. The basic feature of this system is related to the objective study of breath sound intensity, frequency content and timing. Characteristics of acoustical signals, which can be determined in the frequency as well as in the time domain, were correlated with physical variables such as respiratory air flow and lung volume. For this study, the frequency response of the overall recording system was from 120 to 1,000 Hz. Undistorted recording is of great importance in quantitative studies. For this reason, the dynamic performance of the specific components used for data recording, storage and analysis have been carefully analyzed to insure that they do not modify the signal characteristics.

A block diagram of the system is shown in Figure 5. The breath sounds at specific locations were detected by high sensitivity microphones selected so that they would not alter the spectral characteristics of the sounds or limit respiratory maneuvers. A pneumotach, with associated flow transducer, was used to record respiratory air flow

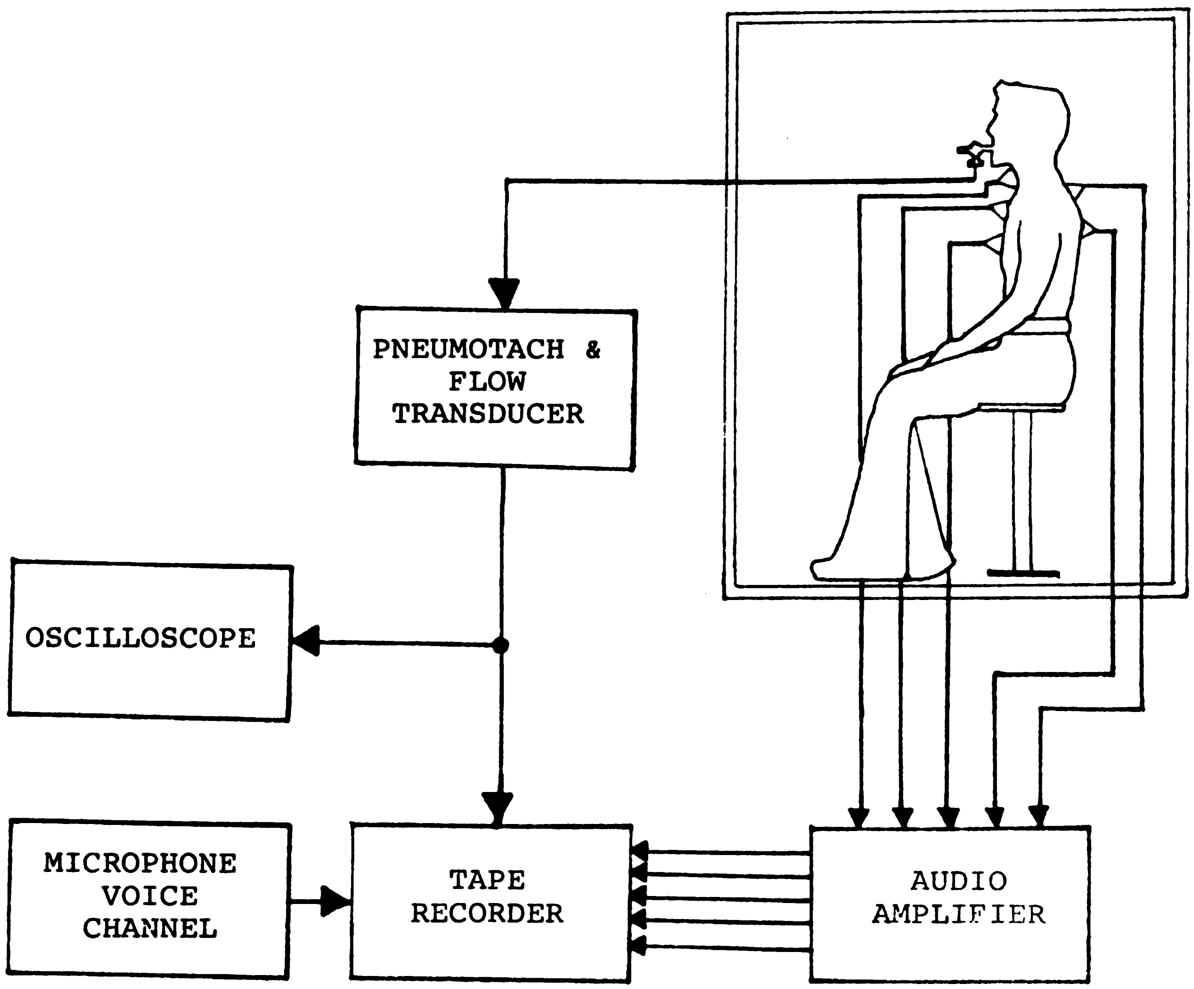

Figure 5. System for simultaneous recording of breath sounds and air flow rate.

signal. A voice channel on the tape recorder was utilized to identify the subject and the type of data being recorded. Since acoustical measurement should be made in a low noise environment, noise from fans, air conditioners and other equipment was reduced to as low a value as was practical.

3.2 Microphone and Attachment

The microphone is the most critical element in a system for recording respiratory sounds. Contact or air-coupled microphones, and nonacoustic transducers like accelerometers can be used. However, microphones which are heavy or require firm contact have limited application in the recording of breath sounds because they can produce distortion by altering the sound transmission characteristics of the thorax or trachea. Crystal microphones can be coupled directly to the chest wall, or indirect coupling can be achieved by using a dynamic air-coupled microphone. Both crystal contact and dynamic air-coupled microphones are used in recording of cardiovascular sounds and both have also been applied to respiratory sound studies. The crystal sensing element, usually is applied directly to the body surface, has a higher sensitivity with less susceptability to ambient noise. Air-coupled dynamic microphones are less dependent on the static force of attachment and have a wide frequency response.

Performance properties of the microphone can be characterized by frequency response, dynamic range, directional characteristics and electrical impendence. In addition, the response characteristics of a microphone are affected by temperature, aging, and humidity. For respiratory sound studies, the frequency response of a microphone is not usually a limiting factor. Contact crystal microphones operate in the frequency range from 20 to 2,000 Hz and air-coupled microphones operate from 20 to 14,000 Hz. The dynamic range of the microphone should be higher than 40-50 dB, and both contact crystal and air-coupled microphones meet this requirement.

Sensitivity is another important factor that describes the characteristics of a microphone and is specified by the ratio of the output signal to a specific sound intensity. However, sensitivity represents only a relative quality of the microphone and its threshold is influenced by ambient noise. Signal-to-noise ratio determines the effective sensitivity and should be large for high fidelity recordings.

Many microphones were compared, and as a result of these comparisons the electret condenser microphone ("Realistic" Cat. No. 33-1056A) was selected. It is a small and light weight microphone with relatively high sensitivity and can be used effectively in respiratory or cardiovascular studies. The manufacturer of this microphone provides the following specifications: frequency response 20-12,000 Hz, output impedance 600 ohms, sensitivity -70 $\pm$ 3 dB (0 dB = 1 V/u bar, 1 kHz), a weight of 4 g, a diameter of 12 mm and a length of 21.6 mm. Figure 6 shows a plot of sensitivity as a function of frequency for the "Realistic" microphone as reported by the manufacturer. The frequency response is relatively flat within the range of 20-12,000 Hz. Since the frequency range of interest in this study is between 120-1,000 Hz, this microphone is an adequate tool for breath sound recording.

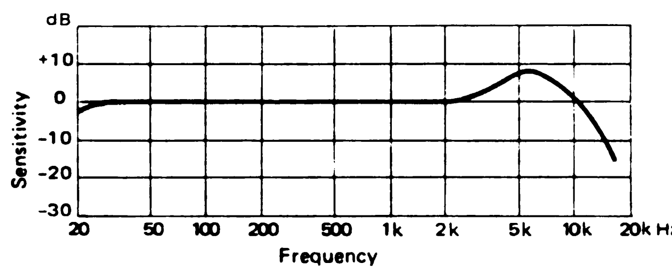

Figure 6. Frequency characteristic of the "Realistic" Cat. No. 33-1056A as reported by manufacturer.

The method of attaching the microphone to the thorax or larynx can affect the sensitivity and fidelity of the recordings. Unpredictable variations in the attachment are one of the basic reasons of the irreproducibility of many measurements. Typical attachments used in phonocardiography include rubber belts, vacuum rings and adhesive tape. In this experiment, an elastic belt was used to attach microphones onto the chest wall. Variations in the tension of the belt holding the microphone can cause changes in the microphone characteristics and affect the acoustical properties of the skin interface. Higher tension in the skin can cause formation of a pseudo-diaphragm and result in progressive attenuation at low frequencies.

In order to maintain a constant condition for measurement, the length of the belt is adjusted to fit the chest without stretching. Hair on the chest can produce noise but it was found that attaching the microphone firmly against the chest eliminates hair noise.

The microphone was surrounded by a rubber ring, and then fitted in a copper cup, 22 mm in diameter with a relatively air tight rubber contact edge to minimize ambient and surface noise. Because such units are very sensitive, surface movement was minimized. The method of attaching the microphone to the thorax using a cylindrical shaped cavity is shown in Figure 7. The coupling cavity has a height of 10 mm and its diameter is 22 mm. Transmission characteristics of a cylindrical air cavity can be expressed in mathematical terms.[18]

$$f_{axial} = \frac{16,640}{L} \quad \text{Hz} \qquad (1)$$

$$f_{cross} = \frac{19,520}{D} \quad \text{Hz} \qquad (2)$$

where: L = length in cm.

D = diameter in cm.

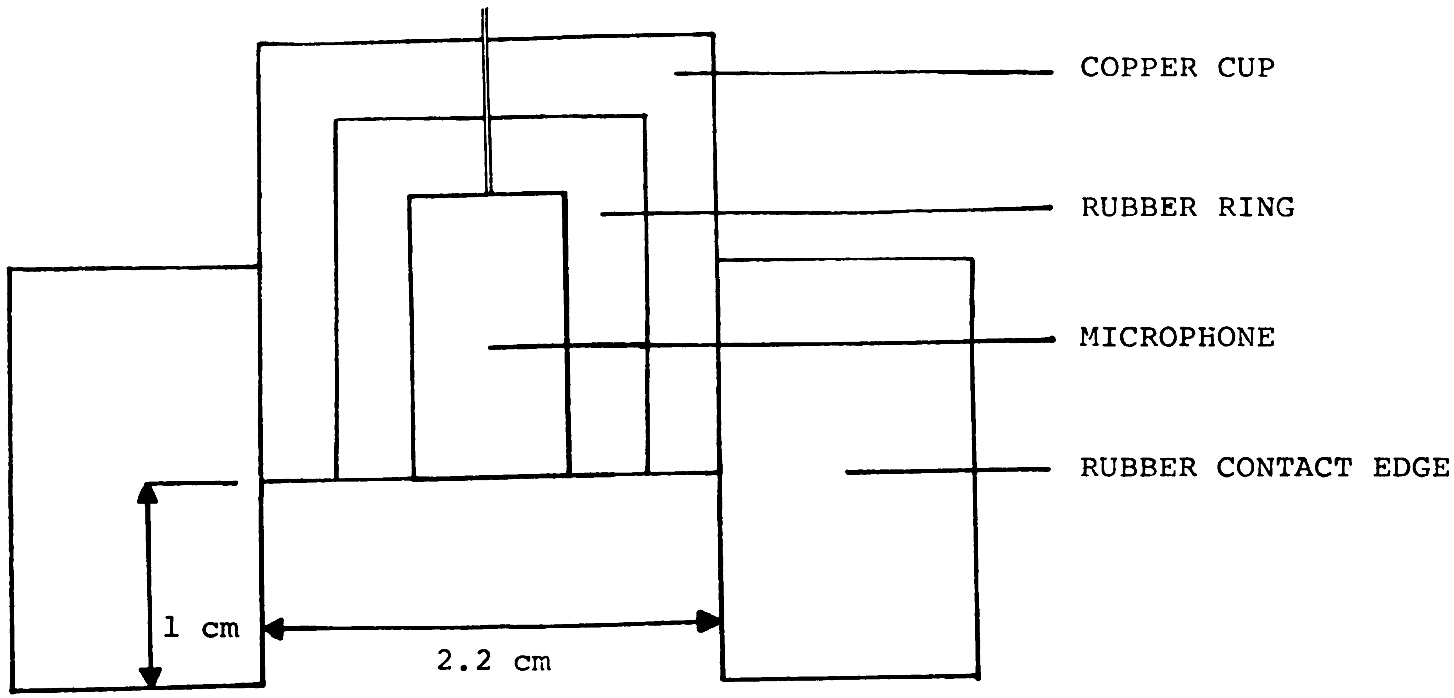

Figure 7. The system of attaching the microphone to the thorax and trachea.

These are expressions for the axial and cross mode resonance frequencies, respectively. Above coefficients are valid for room temperature. Dimensions of the coupling system affect acoustic performance and this influence can be determined from these equations. For the above dimensions the resonant frequencies were:

$$f_{axial} = \frac{16,640}{2.2\ \text{cm}} = 7,564\ \text{Hz.}$$

$$f_{cross} = \frac{19,520}{1.0\ \text{cm}} = 19,520\ \text{Hz.}$$

These simplified calculations show that all resonant frequencies exceed 7 KHz and are above the upper frequency limit of the recording signals, and do not affect them. Every microphone has a similar arrangement.

3.3 Signal Processing and Storage

Storage of breath sounds and flow data for subsequent analysis was accomplished by using an FM magnetic tape recorder. With the eight channel FM tape (Hewlett Packard HP3968A), five channels of breath sound information and one channel of respiratory air flow signal can be recorded simultaneously. Data is recorded in the FM mode at a speed of 15 ips, in which case the frequency response of the tape recorder is from DC to 5,000 Hz, which satisfies the frequency range (120-1,000 Hz) of interest. Frequency response over the passband is $\pm$ 1.0 dB, input level is from 1 to 30 V continuously adjustable, and tape speed accuracy is within $\pm$ 0.2%. VU meters indicate the magnitude of the input signals and are useful in setting the controls so as to insure that excessive input signals are not present. Low noise tape is used for recording (memorex).

The maximum sampling rate of the A/D converter (HP 47310A) is 200 samples per second, therefore, according to the Shannon's sampling theorem, [19,22] the maximum frequency that can be processed is 100 Hz. By slowing down the play back speed of the tape recorder to 1/16 of the recording speed, the recorded frequencies are reduced by a factor of 16. This given an effective band width from 7.2 to 62.5 Hz. At this speed (15/16 ips) the frequency response specified by the tape recorder manufacturer is from DC to 312 Hz which still covers our desired frequency range. For the same reason, the cut off frequency of a constructed low-pass filter was set at 62.5 Hz. Under this condition, the guard band for this system is 200 - (2*62.5) = 75 Hz. The low-pass filter had a cut off frequency at 62.5 Hz, stop-band edge frequency at 100 Hz with at most 2 dB attenuation in the passband and at least 20 dB of attenuation in the transition band. Chebyshev approximation and a standard Sallen & Key low-pass circuit was used to realize the desired circuit response. [20,21] Figure 8 shows the filter circuit diagram.

The microphone provides a typical output voltage signal in the range of 10-150 mV for typical intensities of breath

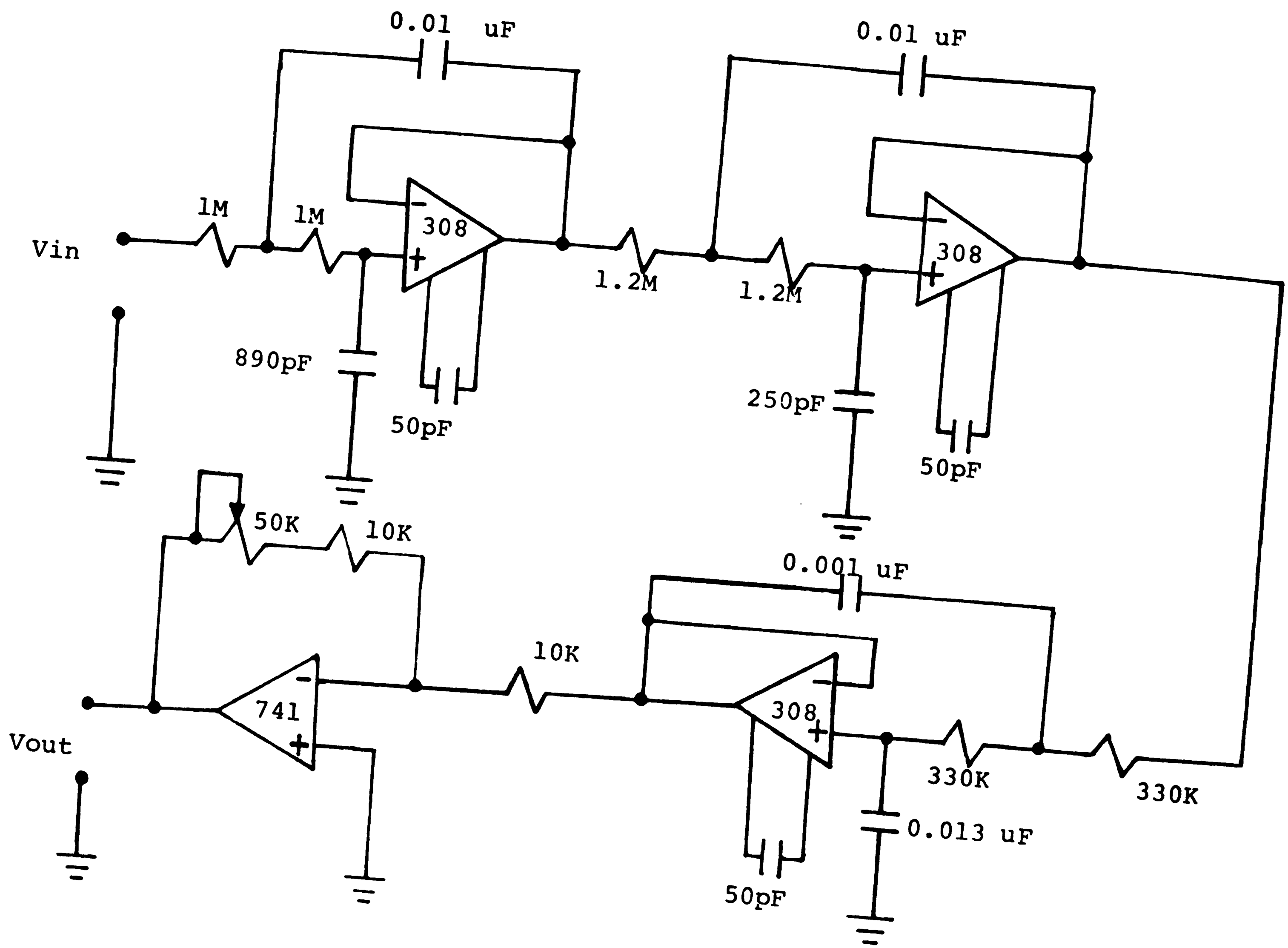

Figure 8. Filter circuit diagram

sounds. In order to meet the input level of the tape recorder, the voltage gain of the amplifier was set at approximately 200. A single operation amplifier is used to amplify the signal. Figure 9 shows the schematic diagram of the amplifier.

3.4 Physical Variables Recording

In addition to recording of breath sounds, it was also necessary to monitor respiratory flow as a function of time. Respiration is related to gas exchange and is a function of inflow and outflow of air. Monitoring air flow is one of the basic measurements in breath sound studies because the generation of sounds is directly related to the air flow. Usually the air flow change is too slow for direct magnetic tape recording, therefore, an FM tape recorder is necessary.

The respiratory air flow is detected with a pneumotach which consists of a stainless steel mesh screen and a sensitive flow transducer. The screen is heated to 95-110^{o} F to prevent the accumulation of moisture which could affect its accuracy. The Hewlett Packard HP47304A respiratory flow transducer permits a wide range of flow measurements by using an interchangeable plug-in pneumotach. This device produces a voltage that is proportional to the air flow through the pneumotach. The largest pneumotach, HP21703A, is suited to adult flow rates and has a sensitivity of 2.50 liters/second/volt with a full scale flow rate of

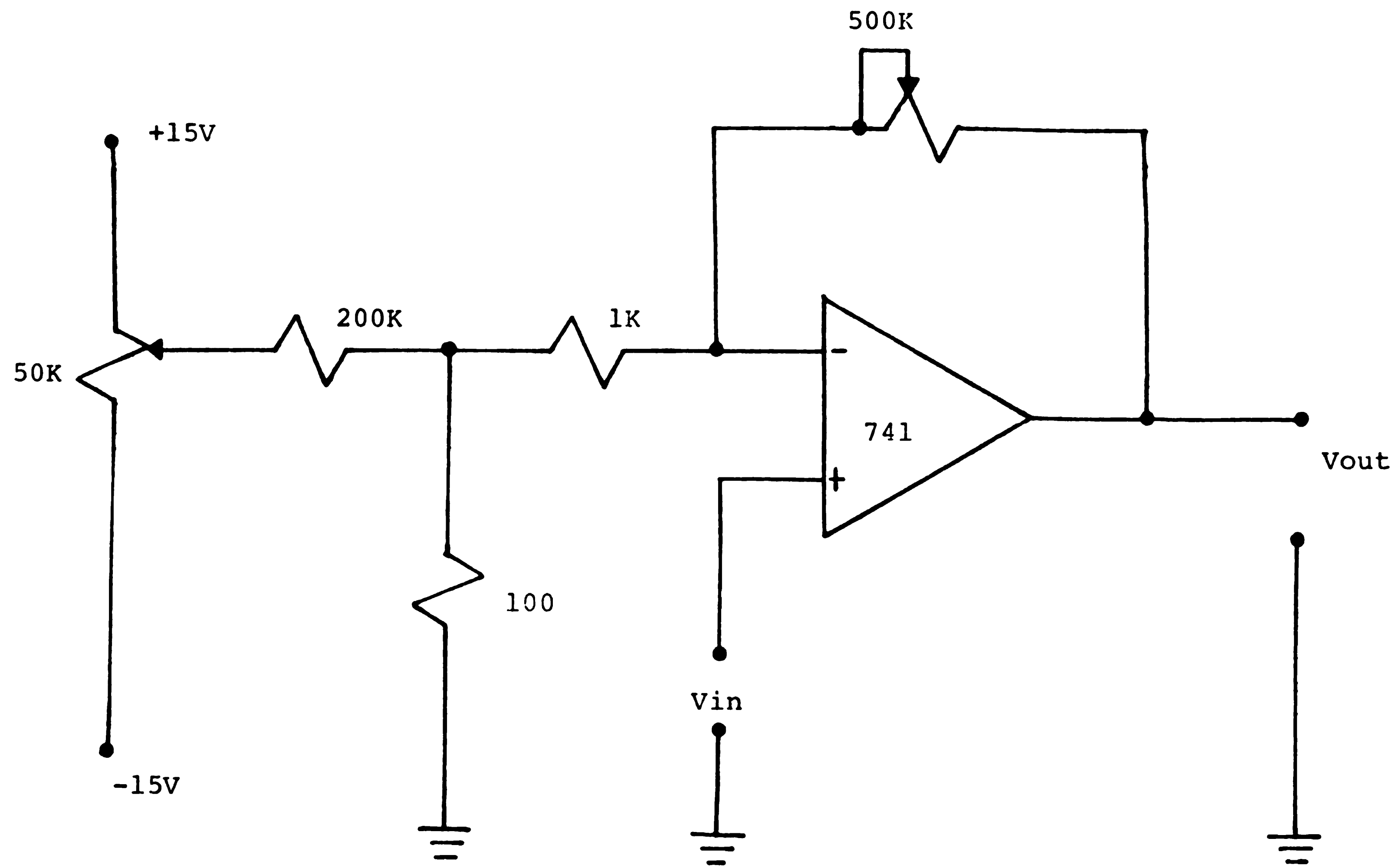

Figure 9. Amplifier circuit diagram.

10 liters/second. [24]

The pneumotach is a bi-directional, laminar flow device that requires nonturbulent air for accurate operation. This is assured by providing a smooth, straight path into the pneumotach. While the tube walls need not be perfectly smooth, there should be no abrupt changes in diameter near the pneumotach input. The inside diameter (ID) of the tubing should be the same as the ID of the pneumotach for a distance equal to at least five times the pneumotach ID.

Signals representing tidal volume and inspiratory or expiratory volumes can be obtained by integrating the flow signal. Instead of using an integrator in hardware to convert air flow into a signal proportional to volume, a software program (Appendix B) using Simpson's rule [26] to integrate flow signal data was used to get lung volume data.

3.5 Standardization and Calibration of the System

Standardization and calibration must be performed on the recording system as well as the instruments used for measuring other physical variables such as respiratory air flow and lung volume.

The sound recording system can be calibrated by using either a relative or an absolute standardization method. A relative standardization technique using arbitrary sound level has been used for calibration of the recording system. This method does not allow the determination of the sound intensity in absolute physical units, but it can be used for comparable studies and the sound intensity can be expressed in arbitrary units. This method is based upon similarity in the frequency characteristics of sound recording channels consisting of identical microphones and amplifiers. Equal amplitudes at the amplifier output are used as the criteria for standardization. Gain of breath sounds channels was calibrated at a single frequency of 500 Hz. It was sufficient due to the relative constant frequency response in the entire range of interest.

Calibration of the instrument system for recording physical variable involved the use of standard method. Connection of a variable air flow source through a standard rotameter to the pneumotach provided a calibration between values of air flow and amplitude of the output signal. [25]

3.6 Digital Data Processing

Digital data processing was conducted on a HP9825A calculator system having 32K bytes memory. Several ROMs (Read Only Memories) are available for the calculator, each provides additional language capabilities to perform specific tasks, such as plotting, controlling peripherials or extending the programming capabilities of the calculator. This system contains four peripherial devices: HP9871A printer, HP9885M flexible disk drive, HP47310A A/D converter and HP9872A plotter. All those peripherial devices are connected to the calculator via an HP-IB interface and all are controlled by the calculator with specific ROMs.

The signals, composed of respiratory sounds data and respiratory air flow were digitized at a sampling rate of 200 samples per second using the HP47310A A/D converter. The A/D conversion routine, using a system ROM card, allowed storage of a data in digital form on a floppy disk. The first step of data processing is to digitize all the data stored on the magnetic tape and store it on the floppy disk for future analysis. The digitized data then can be processed in the calculator and plotted out on the digital plotter in the form of a discrete amplitude plot.

The block diagram of the system for digital data processing is shown in Figure 10. The highest frequency component of the air flow signal is very low, therefore, no analog filter is needed before sampling. But for breath sound signals, in order to avoid aliasing phenomenon, a low-pass filter must be added before the A/D converter.

The next step involved a fast Fourier transform (FFT) analysis on a specific segment of the respiratory sound data and its correlation with air flow signal. Without the air flow signal, it is difficult to find the starting point of inspiration or expiration from the breath sound signal for a forced respiratory cycle. Figure 11 shows a typical pattern of the air flow signal during a forced respiratory cycle.[27] In this figure, the air flow rate is displayed on the Y-axis and time in seconds on the X-axis. Positive

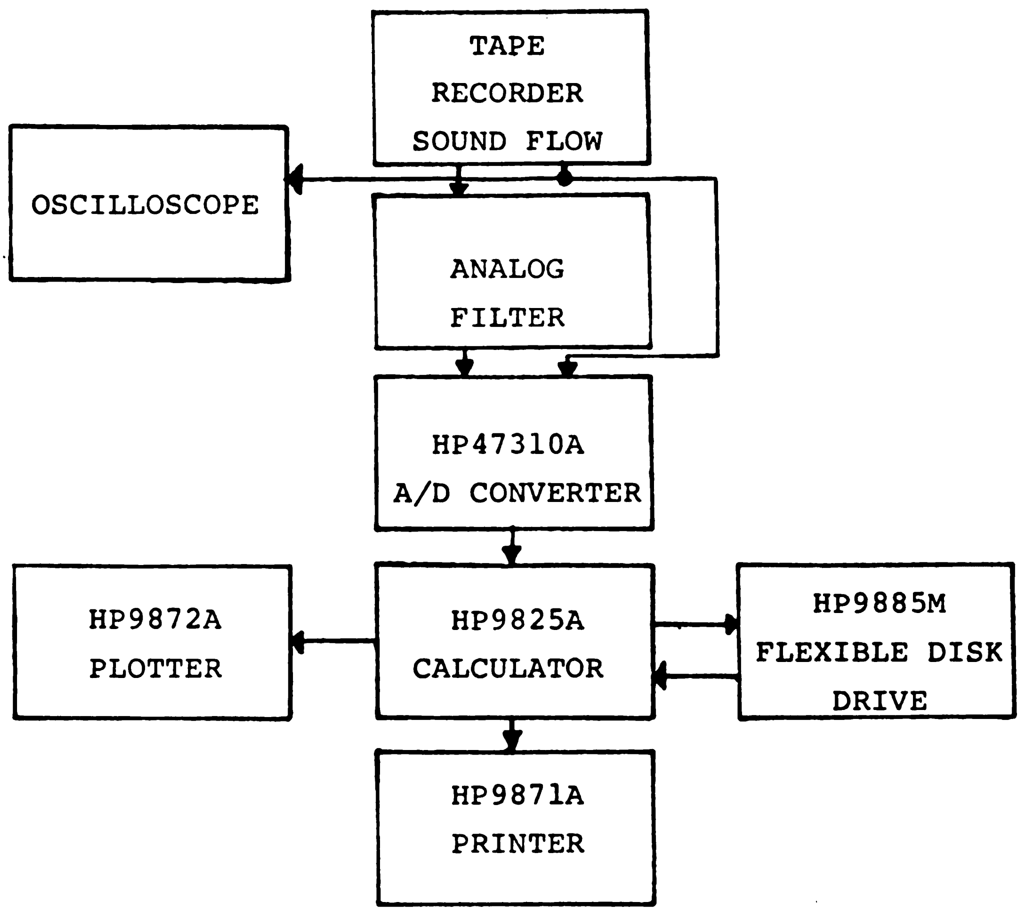

Figure 10. A system for respiratory sounds data processing.

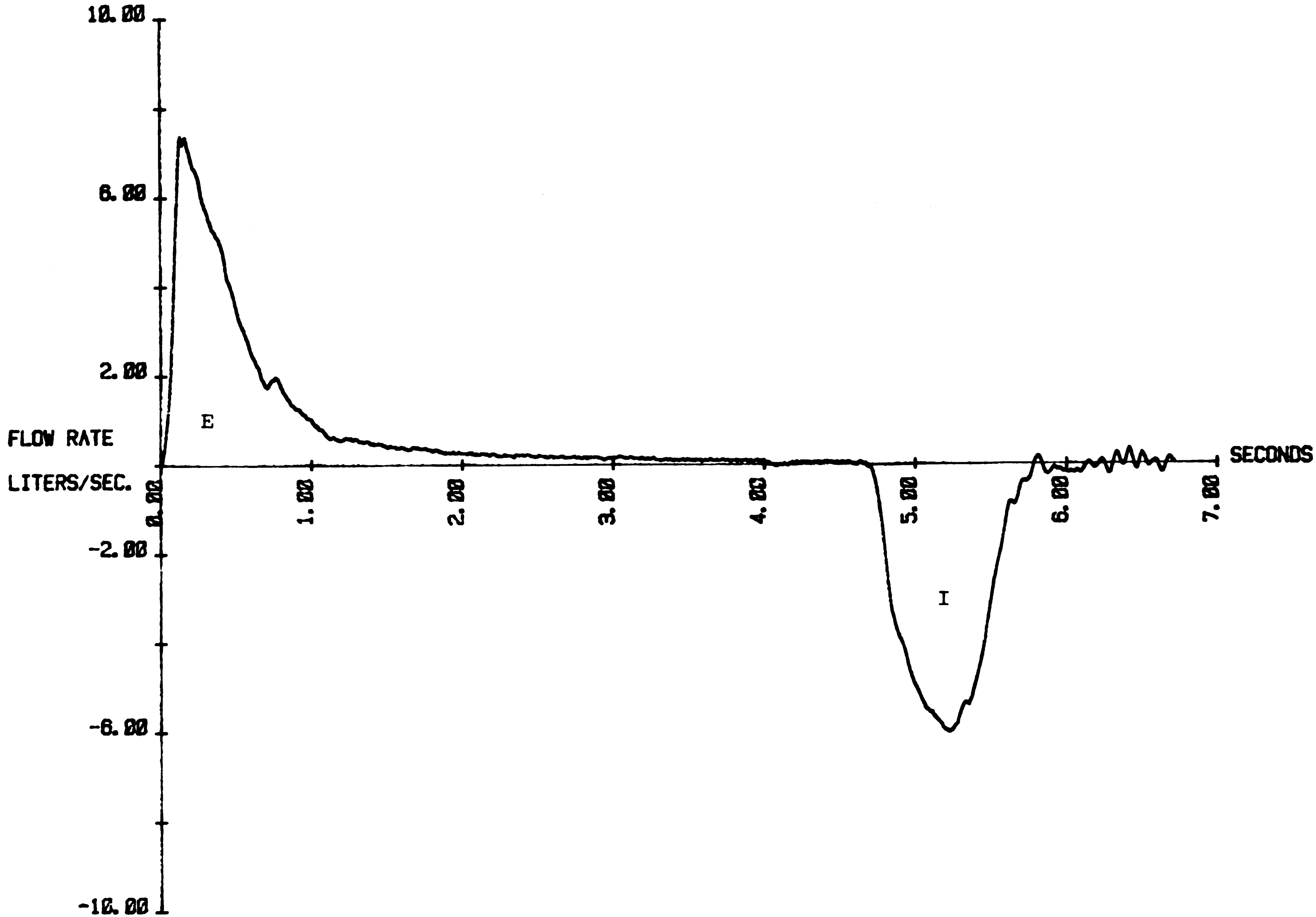

Figure 11. A typical pattern of the air flow signal during a forced respiratory cycle, E means expiration and I means inspiration.

air flow rate indicates expiration and negative air flow rate indicates inspiration. Since air flow signal and respiratory sounds were recorded simultaneously, by digitizing both signals from the same point of the magnetic tape and by taking advantage of this typical pattern, a computer program was used to find the starting point of inspiration and expiration from the digitized air flow data (Appendix C). Those starting points for the air flow signal are the same for the breath sound signals.

The FFT analysis was conducted on a block of data and the length of the data block was limited by the available memory space in the calculator system. In this small HP calculator, only $2^{10} = 1024$ data point pairs can be handled in a time. With 200 samples per second and the speed factor 16 of the tape recorder, the block of data can be analyzed in real time is $1024 \div 200 \div 16 = 0.32$ second. The analysis of breath sound signals and the air flow signal was based on this time segment. The computer program can easily find the starting point of the succeeding segment of data.

The output level of the tape recorder is from 0 to 5 V and the output voltage of the flow transducer is from 0 to 4 V, therefore by using a 1 Hz, 8 V(P - P) sinusodial wave as the calibrating signal and by adjusting the input level and the output level of the tape recorder, the magnitude of the output signal can be made the same as the input signal. Also, by adjusting the gain and the zero of the A/D converter, the absolute magnitude of the air flow rate can be determined. From the air flow rate signal, by using a computer program, both FVC and FEV1 value can be evaluated. Where FVC is forced vital capacity, the maximum volume of air expired following a maximal inspiration and using a maximal effort to expel the air, and FEV1 is timed forced expiratory volume, the volume of air expired in the time specified by one second from the start of expiration.

The computer program for computing the FFT by the successive-doubling method is shown in Appendix A. [22] The HP9825A calculator system cannot deal with a complex array, therefore, two different arrays, F and Y, are used to represent a complex array, F array is for the real part of the data and Y array is for the imaginary part of the data. Since the input data is a real function, the imaginary array Y must be set to zero at the input stage. On input, F and Y are arrays whose transform are desired. On output, F and Y contain the Fourier transform.

The following simple relationships are key for baseband analysis:

$$t = \frac{1}{2\ f_h}$$

$$T = Nt$$

$$f = \frac{1}{T}$$

where f_h = highest possible frequency in spectrum = 100 Hz.

t = the time domain sample spacing = 1/200 second.

N = the number of time domain samples = 1024.

T = the total time record length = 0.32 second.

f = the frequency domain sample spacing.

The increment between spectral components f can also be thought of as the frequency resolution. In this system, the frequency resolution is:

$$f = \frac{1}{T} * 16 = \frac{1}{Nt} * 16 = \frac{1}{1024 * \frac{1}{200}} * 16 = 3.125 \text{ Hz} \tag{19}$$

where the factor 16 is speed factor of the tape recorder.

CHAPTER IV

EXPERIMENTAL PROTOCOL

Quantitative analysis of respiratory sounds was conducted on male normal subjects. All the recordings were taken in a hospital pulmonary laboratory. Instrumentation was calibrated and standardized as discussed in section 3.5 and set as shown in Figure 5. Respiratory sounds were recorded over the trachea, upper and lower lung fields, on both anterior and posterior side, while the subject was sitting. Sound data were recorded while the subject was performing a standard forced vital capacity maneuver (FVC). The FVC maneuver is a forced expiration from maximum inspiration to residue volume, followed by an inspiration to total lung capacity. Subject's age, height, weight and smoking habits were also taken as reference data. Figure 12 indicates locations to place microphones, for both anterior and posterior view. [1]

Four microphones were fixed on the anterior of the chest wall and one microphone was fixed on the trachea. After the subject had performed the FVC maneuver twice, all four microphones on the anterior sites were moved to the posterior positions as shown in Figure 12, then the subject performed the same FVC maneuver twice again. All the data were recorded while the subject was performing the FVC maneuver. Although two FVC maneuvers were recorded for each case, only one was chosen for data analysis. Maximal FVC + FEV1 values were the criterion used to decide which one was a better maneuver.[27] From the digitized air flow rate data, a computer program (Appendix B) using Simpson's rule [26] was used to find which FVC maneuver had larger FVC + FEV1 values.

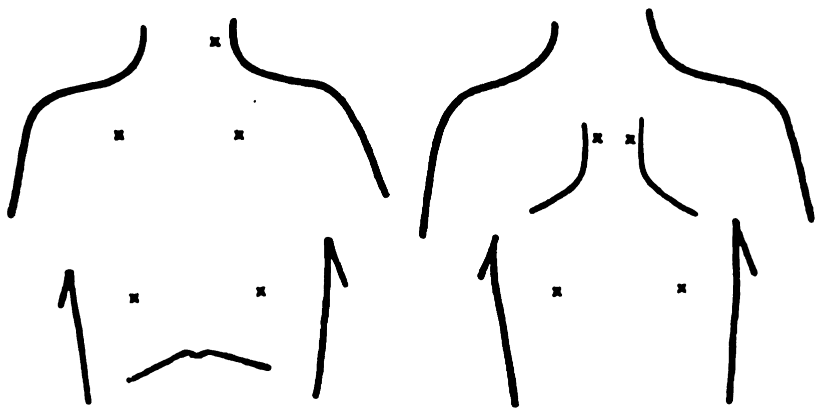

Figure 12. Recommended sites to place microphones for recording respiratory sounds.[1]

CHAPTER V

CONCLUSIONS AND AREAS FOR FURTHER RESEARCH

Figure 11, shown in section 3.6, was obtained from a normal subject performing a standard FVC maneuver. Time is displayed on the X-axis. From this figure, it can be seen that expiration has longer duration than inspiration. By dividing this complete cycle into 0.32 second time segments, the system can provide detailed information about frequency, intensity, air flow and timing of acoustical events at any point in time during expiration or inspiration. The advantage of dividing the respiratory cycle into small time segments is that the correlation between air flow rate and breath sounds can be easily found and detailed information at any time point during the whole respiratory cycle can be obtained.

Figures 13 through 18 show some typical data obtained from the microphone placed upon the trachea. In each figure, the upper part shows the air flow signal, the middle part shows the breath sound signal in the time domain, and the lower part shows the breath sound signal in the frequency domain. All data corresponds to the same time segment. Median frequency was calculated for every figure. FVC and FEV1 values were only shown in the first figure that represents the beginning of expiration of a FVC maneuver. The magnitude of air flow rate is expressed in absolute magnitude while the magnitude of breath sound signals are expressed as a relative magnitude as discussed in section 3.6.

Figures 13, 14 and 15 show the first three sets of data from the beginning of a FVC maneuver. The starting point of expiration can be seen in Figure 13 on both the air flow plot and the breath sound plot in the time domain. There is a strong correlation between breath sound intensity and air flow rate, and the intensity of sound decreases with decreasing flow rate. Figures 16, 17 and 18 show the first three sets of data from the beginning of inspiration. Again, the starting point of inspiration can be seen in Figure 16 and the intensity of the sound can be seen to increase with increasing air flow rate and then decreases with decreasing air flow rate. The increase in intensity of sound with increasing flow rate is associated with frequencies between 120 and 400 Hz. Breath sounds related to inspiration have higher intensities than those related to expiration. From these frequency spectrum plots, for both expiration and inspiration, it can be seen that most of the noticeable bronchial sounds have frequencies below 400 Hz and frequencies above 600 Hz have negligible intensity. High frequency components, especially between 400 and 600 Hz, are greater during expiration than during inspiration. The upper frequency limit of 1,000 Hz for this experiment

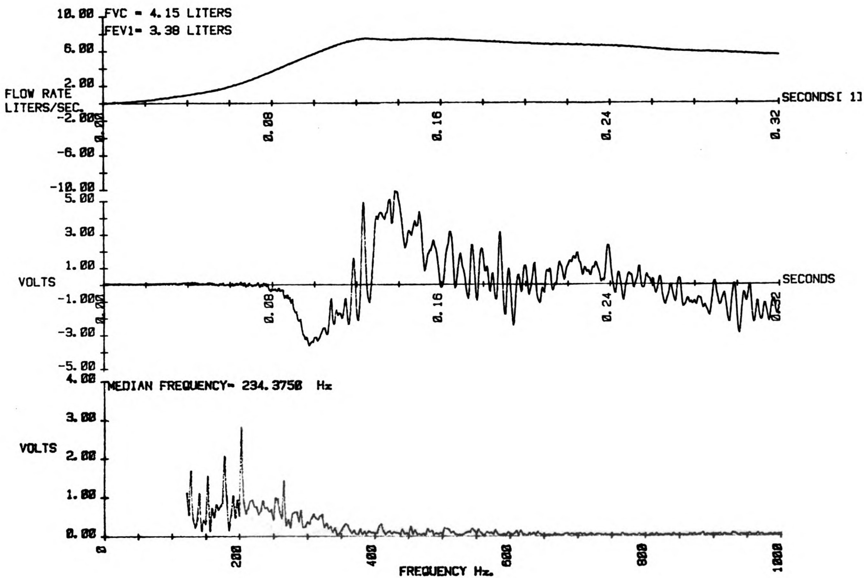

Figure 13. Data obtained from a microphone placed upon the trachea (Expiration 1).

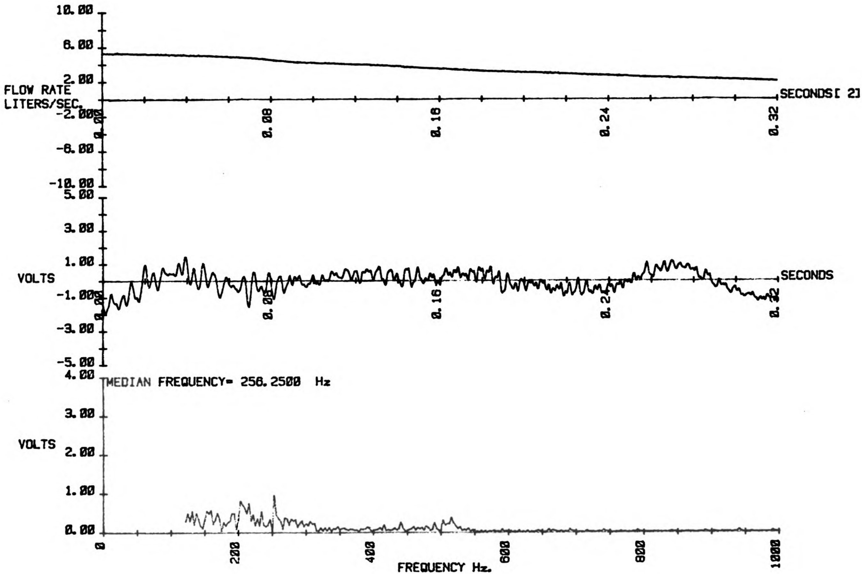

Figure 14. Data obtained from a microphone placed upon the trachea (Expiration 2).

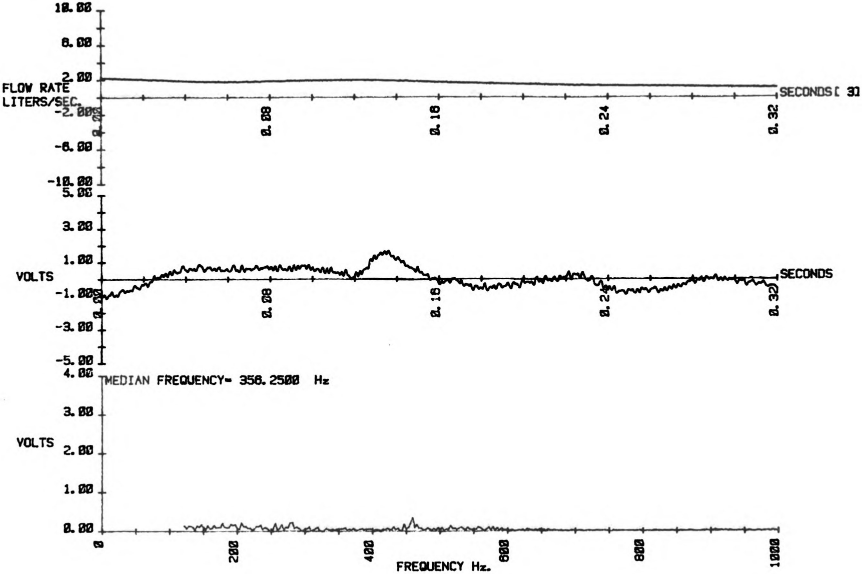

Figure 15. Data obtained from a microphone placed upon the trachea (Expiration 3).

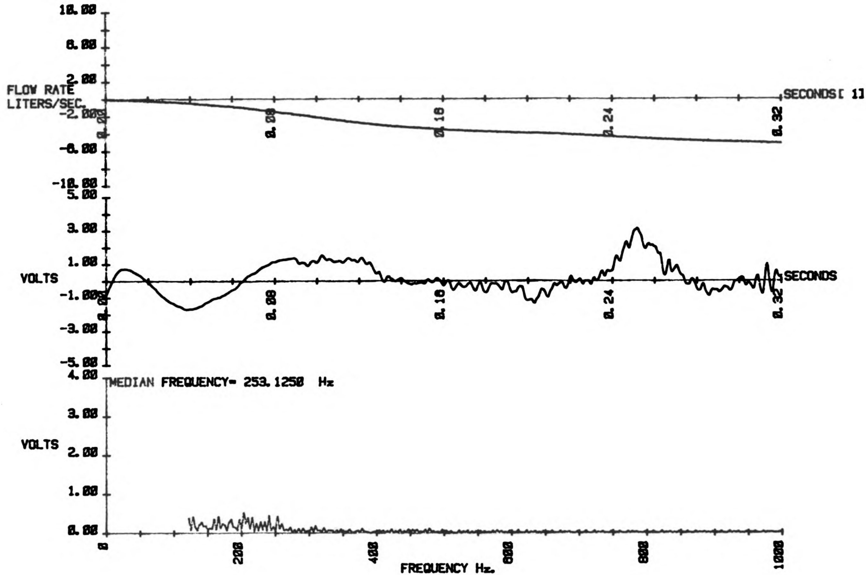

Figure 16. Data obtained from a microphone placed upon the trachea (Inspiration 1).

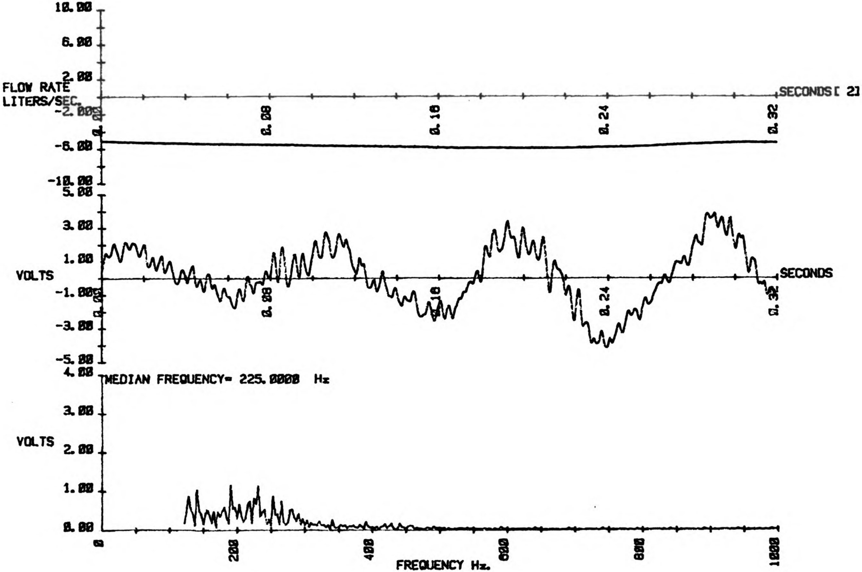

Figure 17. Data obtained from a microphone placed upon the trachea (Inspiration 2).

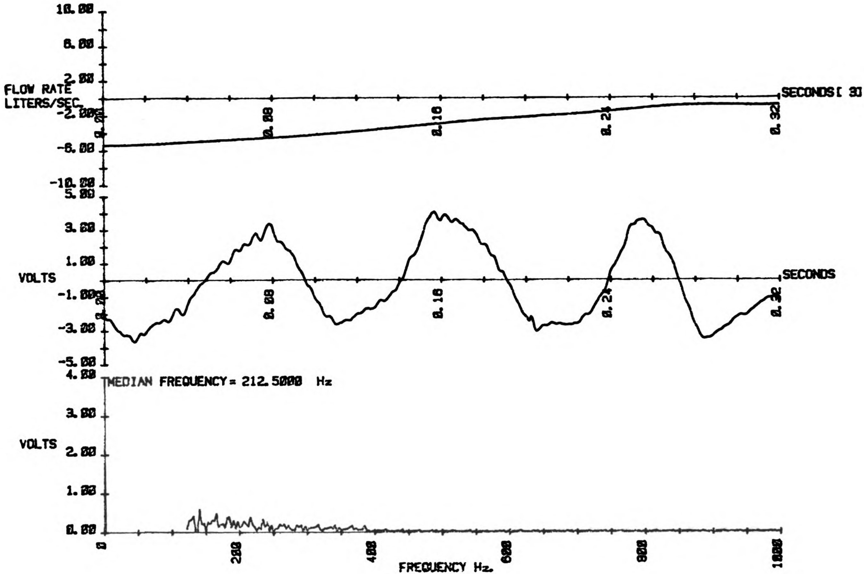

Figure 18. Data obtained from a microphone placed upon the trachea (Inspiration 3).

was therefore a very reasonable assumption.

The frequency resolution for this system is 3.125 Hz. This value is limited by the available memory space of the calculator system and the sampling rate of the A/D converter. It is therefore possible that a major peak of a particular component could lie between two of the discrete transform points. For a given upper frequency range, the frequency resolution can be increased either by using a larger computer system or by increasing the sampling rate of the A/D converter. With a larger computer system, the length of each time segment can also be increased.

Quantitative analysis of respiratory sounds requires precise definition of cardiovascular sounds which are higher in intensity and constitute a main source of noise. Cardiovascular sounds dominate a recording of respiratory sounds particularly during spontaneous breathing or other low flow maneuver. For these reasons, a variety of studies are required before the phonopneumographic methods can find clinical application in routine diagnosis of pulmonary diseases. Extensive studies are required to correlate respiratory sounds with pulmonary variables in normal subjects and subjects with different kinds of pulmonary diseases. A large number of subjects must be studied in order to get statistically valuable data as well as to define variations in the characteristics of sound associated with age, sex, height, weight, and body structure.

Determination of basic characteristics of respiratory sounds, such as distribution of sound intensity over specific frequency bands or the correlation of the sound intensity with respiratory flow requires sophisticated data analysis. This can be performed using a digital computer or it can be done using band pass filters or frequency analyzers. Respiratory sound analysis can also be applied to artificial sounds generated in the chest or applied to the airways. The performance of these so-called transmission studies can provide additional information about physics of the total phenomena and the characteristics of the thorax as a transmission medium. These studies should include transmission in normal subjects and those with diseases. The transmission characteristics is of great importance because respiratory sounds analysis should include determination of variation in cardiovascular sounds during respiratory cycle.

Until now the characteristics of oral respiratory sounds and those recorded over the chest has been studied separately. The aspect of oral respiratory sounds should be further explored, although a number of researchers believe that oral respiratory sounds have limited application in characterization of the lung. The relatively high intensity of the oral respiratory sounds, in comparison with those recorded over the chest is in many cases a principal reason to analyze these signals. These studies may provide some basis for correlation of oral respiratory sounds and sounds recorded over the chest.

Furthermore, respiratory sounds recorded over the left and right lung fields may be used to study asymmetry of the lung. This study, and future studies included with recording over different locations of the same lung field, may provide information concerning pulmonary ventilation. Experimental work may include the affect of various gases of different density, as well as numerous pharmacological agents, affecting respiratory airways.

Contact acoustical sensors have limited application because of difficulties in maintaining the constant conditions of measurements and elimination of interaction between a sensor and sound source. The developed system, which is described in this thesis, eliminates problems associated with the use of contact sensors, and meets requirements associated with data collection for quantitative respiratory sound analysis. Although the described system is adequate, new acoustical sensors should be tested. Internal placement of acoustical sensor would allow the study of sounds which are closer to the original patterns of respiratory sounds. Esophageal microphones can be used in recording of internal respiratory sounds which are less affected by environmental noise.

Dr. Murphy pointed out that, "the chest roentgenogram is superior to the stethoscope in diagnosing tuberculosis, but is inferior in the detection of obstructive diseases such as asthma, emphysema, and chronic bronchitis." [29] Roentgenographic techniques were developed in the 19th and the beginning of the 20th century. At the same time, tuberculosis was the main cause of pulmonary disorder, and one of the most common diseases of mankind in general. The development of auscultatory techniques has been neglected, and Laennec's work, conducted over 150 years ago with limited technological basis, is still valid. Tuberculosis is a disease of the past, but chronic obstructive lung disease has become a common clinical problem. For this reason, it is desirable to investigate the application of acoustic techniques in the detection of pulmonary abnormalities.

APPENDICES

APPENDIX A

COMPUTER PROGRAM FOR COMPUTING THE FFT

```
0: scl 0,1000,0,13.5;fxd 0
1: 1→P;rad
2: enp "HOW MANY SKIP STRINGS?",A
3: enp "STARTING POINT?",V
4: enp "WHICH SET OF DATA?",S
5: xax 0,50,0,1000,4;fxd 2;yax 0,1,0,4,1
6: 10→M
7: 2↑M→N
8: 2int(N/64)+2→G
9: dim A$[G,80],B$[1,350]
10: dim F$[N,4],Y$[N,4]
11: for I=1 to N
12: 0→X;fts (X)→Y$[I]
13: next I
14: files breath
15: for I=1 to A;sread 1,B$[1];next I
16: for I=1 to G
17: sread 1,A$[I]
18: next I
19: dsp "THANKS GOD"
20: for I=1 to G
21: if I=1;V→U;gto "A"
22: 1→U
23: "A":for J=U to 80 by 2
24: on err "GO"
25: if num(A$[I,J,J])>3;gto "DO"
26: num(A$[I,J,J])*256+num(A$[I,J+1,J+1])→X
27: fts (X)→F$[P];P+1→P;if P>N;gto "DONE"
28: next J
29: "DO":-((255-num(A$[I,J,J]))*256+(256-num(A$[I,J+1,J+1])))→X
30: fts (X)→F$[P];P+1→P;if P>N;gto "DONE"
31: next J
32: "GO":next I
33: "DONE":J+2→V;I+A-1→A;S+1→S
34: prt "NEXT STRING=",A;prt "NEXT POINT=",V;prt "NEXT S=",S
35: N/2→L
36: N-1→K
37: 1→J
38: for I=1 to K
39: if I =J;gto "JUNG"
40: stf(F$[J])→X
```

```
41: F$[I]→F$[J]
42: fts (X)→F$[I]
43: "JUNG":L→B
44: "HUA":if B>=J;gto "MING"
45: J-B→J
46: int(B/2)→B;gto "HUA"
47: "MING":J+B→J
48: next I
49: for A=1 to M
50: 2↑A→P
51: P/2→K
52: 1→U;0→V;fxd 4
53: cos(π/K)→W;-sin(π/K)→Q
54: for X=1 to K
55: for I=X to N by P
56: I+K→S
57: stf(F$[S])→B;stf(Y$[S])→G
58: B*U-G*V→T;B*V+G*U→C
59: fts (stf(F$[I])-T)→F$[S];fts (stf(Y$[I])-C)→Y$[S]
60: fts (stf(F$[I])+T)→F$[I];fts (stf(Y$[I])+C)→Y$[I]
61: next I
62: U*W-V*Q→U;U*Q+V*W→V
63: next X
64: next A
65: scl 0,320,0,80;0→C;0→D
66: for I=39 to 320
67: stf(F$[I])/N→A;stf(Y$[I])/N→B
68: √(A*A+B*B)→X
69: X/2+C→C
70: plt I,X
71: next I
72: for I=39 to 320
73: stf(F$[I])/N→A;stf(Y$[I])/N→B
74: √(A*A+B*B)→X;X+D→D
75: if D>C;gto "MED"
76: next I
77: "MED":I*3.125→I
78: scl 0,320,0,13;csiz
79: plt 140,-1,3;lbl "FREQUENCY Hz."
80: plt -40,2.2,3;lbl "VOLTS"
81: plt 2,3.5,3
82: lbl "MEDIAN FREQUENCY=",I," Hz"
83: end
*25433
```

APPENDIX B

COMPUTER PROGRAM FOR COMPUTING THE LUNG VOLUME

```
0: 1→G;100*G→N;2*G+2→W
1: enp "HOW MANY SKIP STRINGS?",F
2: dim A$[1,80],B$[1,350],C[20],D[12],3[12],F[2000],S[3]
3: files flow;2→P;0→C[1];0→A;fxd 1
4: for I=1 to F;sread 1,B$[1];next I
5: for I=1 to N
6: sread 1,A$[1]
7: for J=1 to 80 by W
8: on err "GO"
9: if num(A$[1,J,J])>3;gto "DO"
10: num(A$[1,J,J])*256+num(A$[1,J+1,J+1])→C[P];gto "TEST"
11: "DO":-((255-num(A$[1,J,J]))*256+(256-num(A$[1,J+1,J+1])))→C[P]
12: "TEST":if C[P]<0;2→P;0→A;gto "I"
13: if C[P]>C[P-1];A+1→A;gto "COM"
14: 2→P;0→A;gto "I"
15: "COM":J→D[P];I→E[P];P+1→P;if A>8;E[2]+F-1→E;gto "J"
16: "I":next J
17: "GO":next I
18: dsp "NO STARTING POINT";stp
19: "J":spc ;prt "I=",E;prt "J=",D[2]-2;E[2]→I;1→A
20: files flow;for 0=1 to E;sread 1,B$[1];next 0
21: for 0=1 to N
22: sread 1,A$[1]
23: if 0=1;D[2]-2→J;gto "B"
24: 1→J
25: "B":for P=J to 80 by 2
26: on err "GE"
27: if num(A$[1,P,P])>3;gto "DE"
28: num(A$[1,P,P])*256+num(A$[1,P+1,P+1])→F[A];gto "TES"
29: "DE":-((255-num(A$[1,P,P]))*256+(256-num(A$[1,P+1,P+1])))→F[A]
30: "TES":if A<80;gto "EN"
31: if F[A]<0;gto "END"
32: "EN":1+A→A
33: next P
34: "GE":next 0
35: dsp "NO ENDING POINT";stp
36: "END":spc ;prt "0=",0+E-1;prt "P=",P;prt "A=",A-2
37: if int(A/2)=A/2;A+1→A
38: fxd 8;1/(200*G)→D;A→Z;1→Y
39: "Z":gsb "SIMPEQ"
40: if flg2 or flg3;stp
```

```
41: spc 2;prt "INTEGRAL=",S[Y];spc 4;1+Y→Y;if Y=3;gto "H"
42: 198*G+1→Z;gto "Z"
43: "SIMPEQ":cfg 2,3;if Z<3;sfg 2;dsp "MORE POINTS";ret
44: if int(Z/2)=Z/2;dsp "ODD NO. OF PTS";sfg 3;ret
45: F[1]+4F[2]+F[Z]→S[Y]
46: 1→I
47: if (I+2→I)>Z-1;S[Y]D/3→S[Y];ret
48: S[Y]+2F[I]+4F[I+1]→S[Y];gto -1
49: "H":S[1]*2.5/150→S[1];S[2]*2.5/150→S[2];plt 5,850,1;fxd 2
50: lbl "FVC =",S[1],"  LITERS"
51: plt 5,800,1;lbl "FEV1=",S[2],"  LITERS"
52: end
*30659
```

APPENDIX C

COMPUTER PROGRAM FOR FINDING THE STARTING POINT OF A FORCED RESPIRATORY CYCLE

```
0: 16→G;500→N
1: 2→F
2: dim A$[1,80],B$[1,350],C[20],D[12],E[12]
3: files FLOW;2→P;0→C[1];0→A;fxd 1
4: for I=1 to F;sread 1,B$[1];next I
5: for I=1 to N
6: sread 1,A$[1]
7: for J=1 to 80 by 20
8: on err "GO"
9: if num(A$[1,J,J])>3;gto "GO"
10: num(A$[1,J,J])*256+num(A$[1,J+1,J+1])→C[P];gto "TEST"
11: "DO":-((255-num(A$[1,J,J]))*256+(256-num(A$[1,J+1,J+1])))→C[P]
12: "TEST":if C[P]→0;2→P;0→A;gto "I"
13: if C[P]>C[P-1];A+1→A;gto "COM"
14: 2→P;0→A;gto "I"
15: "COM":J→D[P];I→E[P];P+1→P;if A>10;E[2]+F-1→E;gto "J"
16: "I":next J
17: "GO":next I
18: dsp "NO STARTING POINT";stp
19: "J":spc ;prt "I=",E;prt "J=",D[2]-20
20: end
*24277
```

REFERENCES

REFERENCES

1. Raymond L. H. Murphy, Jr., "A Simplified Introduction to Lung Sounds," _Pulmonary Service_, Lemuel Shattuck and Faulkner Hospitals, 1977.
2. J. F. Bullar, "Experiments to Determine the Origin of Respiratory Sounds," _Proc. Roy. Soc._, London 37: 411-422, 1884.
3. R. Nachman, M. Nachman, L. Vinaru, "Recording and Analysis of Sounds Produced by Human Lungs During Respiration," _Experientia_, 25/10.
4. R. J. Bethke, A. Seireg, "Analysis of Vesicular Lung Sounds in Normal and Emphysematous Subjects," _Report of the American Society of Mechanical Engineers_, 1973.
5. V. A. Mckusick, "Cardiovascular Sound in Health and Disease," _The Williams & Wilkins Co._, 1958.
6. E. B. Weiss, C. J. Carlson, "Recording of Breath Sounds," _Amer. Rev. Resp. Dis._, Vol. 105, 1972.
7. G. Fahr, "The Acoustics of the Bronchial Breath Sounds," _Archives of Internal Medicine_, 1927.
8. R. C. Cabot, H. F. Dodge, "Frequency Characteristics of Heart and Lung Sounds," _The Journal of the American Medical Association_, Vol. 84, No. 24, 1925.

9. L. D. Meyer, N. R. McCullough, P. M. Griffen, "Acoustic Pulmonary Diagnostic Techniques," General Electric Company, Report NIH 69-2338-4, Ch. 3.

10. E. Banaszak, C. Ross, "Phonopneumography," American Review of Respiratory Disease, Vol. 8, No. 2, 1974.

11. Wooten, F. T., Waring, W. W., "Spectral Analysis of Respiratory Sounds," Medical Instrumentation, Vol. 8, No. 2, 1974.

12. P. Forgacs, A. R. Nathoo, H. D. Richardson, "Breath Sounds," Thorax, Vol. 26, No. 3, 1971.

13. P. Leblanc, P. T. Macklem, W. R. D. Ross, "Breath Sounds and Distribution of Pulmonary Ventilation," American Review of Respiratory Disease, Vol. 102, 1970.

14. P. Forgacs, "The Functional Significance of Clinical Signs in Diffuse Airway Obstruction," Brit. J. Dis. Chest, Vol. 170, 1971.

15. G. A. Feruglio, "An Intracardiac Sound Generator for the Study of the Transmission of Heart Murmurs in Man," American Heart Journal, February, 1962.

16. J. R. Nairn, M. Turner-Warwick, "Breath Sounds in Emphysema," Brit. J. Dis. Chest, Vol. 63, 1969.

17. L. I. Nemerovskii, L. I. Zelivyanskaya, "Apparatus for Local Examination of the Lungs (Report 1)," Pul'Mofon-3, Meditsinskaya Teknika, No. 4, 1959.

18. P. M. Griffen, R. B. Tatge, D. MacCanon, "Design Considerations for an Air-Coupled Phonocardiac Microphone," General Electric, Report No. 69-C-258, 1969.

19. W. D. Stanley, Digital Signal Processing, Reston Publishing Company, Inc., 1975.

20. G. Daryanani, Principles of Active Network Synthesis and Design, John Wiley & Sons, Inc., 1976.

21. A. Budak, Passive and Active Network Analysis and Synthesis, Houghton Mifflin Company, 1974.

22. L. R. Rabiner, B. Gold, Theory and Application of Digital Signal Processing, Prentice-Hall, Inc., 1975.

23. J. Millman, C. C. Halkias, Integrated Electronics, McGraw-Hill Book Co., 1972.

24. Preliminary Manual, Hewlett Packard Model 47304A Respiratory Flow Transducer.

25. Operating and Service Manual, Hewlett Packard Model 47310A Analog to Digital Converter.

26. E. Kreyszing, Advanced Engineering Mathematics, John Wiley & Sons, Inc., 1972.

27. System Manual, Hewlett Packard 47804A Pulmonary Calculator System.

28. U. S. News & World Report, Vol. 78, No. 24, June 1975.

29. Murphy, R. L., Human Factors in Chest Auscultation, Human Factors in Health Care, R. M. Pictett and T. J. Triggs, Lexington Books.