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Multifield spectral analysis of human respiratory sounds

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MULTI-FIELD SPECTRAL ANALYSIS OF

HUMAN RESPIRATORY SOUNDS

by

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ABSTRACT

Computer analysis of human respiratory sounds (phonopneumography) has been attempted since the 1960s with poor results. Steady research efforts by numerous researchers have failed to yield a practical diagnostic tool to replace the physician and the stethoscope. This has been the result of the high degree of variability in proposed analysis parameters from subject to subject.

The purpose of this research is to examine a new approach to phonopneumography by comparing multiple lung fields of the same subject for the same breath in order to develop the diagnostic parameters. This approach has the advantage of allowing the subject to provide the norm for comparison. Such an approach is possible because lung pathologies rarely affect both lungs in the same location in the same way, unless the condition is so severe as to be easily diagnosed without auscultation.

Two pediatric subjects were recorded in four locations on the anterior torso consisting of two pairs of homologous lung fields. These recordings were analyzed for their spectral content, and parameters relating to the power distribution were calculated. These parameters were compared to seek similarities between homologous segments for each subject. In addition, the researcher compared these parameters for each pair of segments in the same lung to seek an additional parameter that might prove useful for

diagnosis. Finally, the parameters for each subject were compared to see if the similarity would extend beyond the individual subject.

A strong similarity was noted in homologous lung fields for each subject. Additionally, each subject showed a rising central frequency as the analysis moved to a higher location on the lung. No such similarity was noted when comparing the two subjects' data.

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CHAPTER 1

AN INTRODUCTION

A Problem Noted

For children with cystic fibrosis and adults with emphysema, quality and length of life are tied directly to the ability to keep the airways clear. By the time it becomes obvious that obstructions have occurred, it is often too late to reverse the damage caused. As a result, physicians must constantly monitor the state of respiratory health for such at-risk patients. The current state of medical arts provides two methods for monitoring: radiation imaging and auscultation.

Radiation imaging encompasses a broad class of methods. Whether the physician uses the radiograph (commonly called the x-ray) or newer methods such as the MRI or the CAT scan, these methods provide the physician and patient with a clear picture of the lungs and air passages, providing warning prior to the onset of damage. But these methods have significant drawbacks. Some are radioactive, and as such have a cumulative negative effect. Repeated x-rays put the patient at risk for development of cancers and growth disorders. All of these methods are expensive, with the safest methods being the most expensive. Thus such methods are rarely used for routine monitoring of at-risk patients. Their use is chiefly diagnostic, after a problem is suspected.

The second method, auscultation, is much less expensive than the first, and is the primary method currently used to monitor at-risk patients. Auscultation is the process of a trained medical professional listening to sounds through a stethoscope. Auscultation technically refers to any sound monitoring for diagnostic purposes, but for the purpose of this dissertation its meaning will be restricted to its use in respiratory diagnosis. Auscultation follows a regimented process, moving from a lung field on the right lung to its homologous field on the left lung, comparing the sounds made for differences in intensity and pitch, as well as for additional sounds (adventitious sounds) which should not be there. This method also has drawbacks. Auscultation is a highly subjective technique requiring a lengthy application of the procedure. While it is true that it is less expensive than the alternatives, it is not inexpensive given the hourly rate of physicians. Nor does it provide diagnostic information of the quality of the radiation methods. This subjective comparison of sounds being heard with sounds heard previously is a difficult art and, according to recent research, one poorly practiced (Mangione and Nieman 1997).

What is needed is a new method, a method that does not have the drawbacks of auscultation nor the expense and risks of the radiation methods. Such a method would allow more regular and objective monitoring of at-risk patients for respiratory problems before they become serious problems.

A Thirty Year Losing Battle

A promising method that has failed to achieve practical application is phonopneumography. This is a wedding of computer technology with the established techniques of auscultation and was pioneered by Paul Forgacs in the late 1960s. Dr.

Forgacs began to record the respiratory sounds of patients and used a computer to estimate the spectral information is these sounds (Forgacs, et al. 1971). In a series of articles, he noted various relationships between sound parameters such as intensity and frequency with respiratory parameters such as forced expiratory volume at one second (FEV_1), peak expiratory flow (PEF) and adventitious sounds (Forgacs, et al. 1971). He also made early attempts to correlate these sound parameters with various clinical conditions (Forgacs 1969).

Since this effort by Dr. Forgacs, many have attempted to extend his concept into a practical diagnostic tool. All have failed. Many researchers have been plagued by poor engineering methods. One such researcher used an inexpensive handheld tape recorder to record the sounds of respiration near the suprasternal notch. To gather his data, he held the microphone (which was included with the recorder) "close" to each patient's throat. The combination of poor coupling and variability in distance doomed his research long before his data analysis failed to show any significant conclusions. Many of the engineering difficulties were investigated and discussed in M. J. Mussell's paper entitled "The Need for Standards in Recording and Analyzing Respiratory Sounds" (1992).

As engineering problems were discovered and corrected, another problem arose, this one more difficult to deal with. Many of the parameters that the researchers had used to quantify breath sounds had a wide variability from patient to patient (Gavriely, et al. 1981). The current research suggests that there is no reliable standard for intensity, mid-power frequency, etc., which separate healthy breath sounds from abnormal breath sounds. As an example, the normal level of sound intensity of a young child would indicate lung consolidation in an older patient. Even normalizing by age, height and weight, there seems to be no parameter that has broad diagnostic potential when drawn from a wide population.

A New Approach

A better method has been suggested by limited research literature. Schreiber, et al. (1981), recorded respiratory sounds at two homologous lung segments, then sounds from non-homologous segments. The homologous segments were "remarkably similar" while the non-homologous segments "differed significantly." This similarity for homologous sections is what physicians are taught to listen for. Should one lung exhibit markedly different sound characteristics between two homologous sections, some clinical explanation is sought. But studying the differences between non-homologous segments also offers hope for an improvement over simply trying to mimic current auscultatory techniques. If a body of research could establish a parameter that relates a group of segments, new diagnostic possibilities would be possible.

The new method would record simultaneously at multiple locations on the thorax. By comparing homologous segments, this method would avoid the variability between patients. A patient's respiratory sounds would not be compared with the population norm; it would be compared with his/her own respiratory sounds. This method also avoids the problems noted in the literature of parametric differences due to different flow rates (Mussell, et al. 1990). A normal breath would be compared with the exact same breath throughout the lung. If the next breath doubled the volume, its parameters would still be compared with similar flow rate parameters.

Such analysis takes advantage of the fact that sound altering conditions frequently affect one lung before the other, and if both lungs are affected, different segments are affected. It is unusual for two homologous segments to be altered in precisely the same manner. However, if such were the case, a parameter that compares non-homologous segments in the same lung might be useful. Suppose, as some literature has suggested (Hildago 1992), that the central power frequency for each segment rises as the sound is monitored higher in the lung. If a straight line could model this relationship, then a slope would be a significant parameter to describe the entire lung. Any variation in the slope, or the change to a more complex curve, could suggest a problem in the segment that has deviated from the expected frequency.

The Eventual Hope

Many years ago, when the art of blood diagnosis was new, physicians kept microscopes in their laboratories to examine a patient's blood. Such a practice was necessary to provide state-of-the-art medical care. Now a microscope is no longer standard equipment for a physician. Instead patients are sent to the blood lab where their blood is drawn by medical technicians (phlebotomists) and analyzed by computer-controlled equipment. The analysis is reduced to parameters (cholesterol levels, etc.) and returned to the physician who uses such information to aid in a diagnosis.

The eventual hope of this research is to begin the process that will lead to development of the respiratory lab. Patients will be sent there to have their respiratory health monitored, returning to the physician with a parametric description of their current lung conditions. By reducing the cost and difficulty of such an analysis, it is hoped that respiratory monitoring will become the norm of medical care, that ailments such a histoplasmosis will be discovered before they become a serious threat to heath, and that the stethoscope will join the microscope in the historical archives of a physician's standard equipment.

CHAPTER 2

PRIOR ART

Auscultation of Breath Sounds

Introduction

Auscultation has a long and storied history. It begins with the writings of the Hippocratic school (circa 400 BC), which notes various sounds, including "creaking" and "bubbling" as emanating from the chest (Gavriely 1995, 1). The early method of listening to these quiet but informative sounds was for the physician to place his ear on the chest of the patient (Gavriely 1995, 1). Though generally effective for some patients, it was impractical for obese patients and socially unacceptable if the patient was a young woman.

The first major improvement in the practice of auscultation came in the late nineteenth century:

> In 1816 I was consulted by a young woman presenting general symptoms of disease of the heart. Owing to her stoutness, little information could be gathered by application of the hand and percussion. The patient's age and sex did not permit me to resort to the kind of examination I have just described (i.e., direct application of the ear to the chest). I recalled a well-known acoustic phenomenon: namely, if you place your ear against one end of a wooden beam the scratch of a pin at the other extremity is most distinctly audible. I occurred to me that this physical property might serve a useful purpose in the case with which I was then dealing. Taking a sheaf of paper I rolled it into a very tight roll, one end of which I placed over the praecordial region, whilst I put my ear to the other. I was both surprised and gratified at being able to hear the beating of the heart with much greater clearness and distinctness than I had ever done before by direct application of my ear.

> I at once saw that this means might become a useful method for studying, not only the beating of the heart, but likewise all movements capable of producing sound in the thoracic cavity, and that consequently it might serve for the investigation of respiration, the voice, râles and even possibly the movements of a liquid effused into the pleural cavity or pericardium (Laennec, 1819).

This passage introduces René Théophile Laennec's classic text De l'Auscultation Médiate, which began the drive to add technology to the art of diagnosis by auscultation. From this humble beginning in 1819 has grown a practice so universal as to make the stethoscope the universally recognized symbol of the physician.

Laennec's simple device was modified and improved over the years until it evolved into the currently recognized version, but that simple design has really been modified little over the intervening 180 years. Originally built of wood with a single earpiece, the basic design was modified to provide flexible tubing, input to both ears (the binaural stethoscope), and a chest piece for concentrating the sound. Later (in an innovation largely ignored) a second chest piece was added to provide differential input to the listener (the stereophonic stethoscope). Although technically superior, the stereophonic stethoscope is more difficult to use. Thus the binaural stethoscope has become the recognized standard for the medical community.

Currently, the stethoscope is made of a combination of a double-sided chest piece, flexible tubing (about 30 cm long), a binaural headset and eartips to fit the headset securely in the ear. The double-sided chest piece has a bell-shaped side for listening to low-pitched sounds and a diaphragmatic side for relatively high pitched sounds (it will be discussed later, but conventional wisdom indicates that most chest sounds are found at frequencies lower than 500 Hz, although some are found higher (Kramen 1983). Therefore the term "high pitched" sounds is a relative one).

Procedures for Stethoscope Use

Substantial training and practice are required to use a stethoscope accurately. A recent study (Mangione and Nieman 1997) has shown that a large number of newly-trained physicians are not proficient with the use of the instrument, and many physicians do not give a thorough auscultatory examination as a part of an office check-up. Obviously unusual sounds, unless very quiet, are usually heard, but other more subtle diagnostic clues evade many physicians who favor more expensive, time consuming and often invasive diagnostic techniques.

The patient should be sitting upright in a quiet, well-ventilated room (Levitzky, et al. 1990, 22-23). The procedure cannot take place over clothing, as the clothing will act as a filter, muffling and modifying the sounds, as well as adding extraneous sounds (Levitzky, et al. 1990, 22-23). Additionally, under the best of circumstances, extraneous sounds from muscle, joint, and tendon movement, as well as hair movement against the diaphragm, will provide extraneous noises which must be factored into any analysis of the lung sounds (Levitzky, et al. 1990, 22-23).

When listening to a patient's lungs, the physician is listening for several things to provide clues regarding health or disease. First of all, adventitious sounds (crackles and wheezes) are sought (Wilkins, et al. 1990, 41). These are the most obvious sign of some abnormal pathology present. Failing to hear such adventitious sounds, the physician is comparing the sounds heard with the memory of what "normal" sounds are (Waring 1980, 60). Also the homologous fields from the left and right lungs are compared for similarity (Waring 1980, 60). Both normal and abnormal lung sounds are discussed in the following sections.

Normal Breath Sounds

What qualifies as normal depends somewhat upon where the sound is heard. Three common locations are used by physicians to listen to breath sounds: at the mouth, at the trachea and through the chest wall.

Breath Sounds at the Mouth

At low levels of breathing, the sound of normal human breathing is virtually silent to the unaided ear (Forgacs 1973). As the rate of ventilation increases, as with exertion, this higher rate of breathing can become clearly audible (Forgacs 1973). Little diagnostic use is made of such sounds in most patients. However, for patients with airflow limitations, such as asthma, bronchitis, and emphysema, this sound becomes audible with much less exertion (Forgacs 1973). There is a strong correlation between the intensity of the sound and the degree of airflow limitation in such patients (Soufflet, et al. 1990).

Breath Sounds at the Trachea

Auscultation of the trachea is unusual in routine physical examinations, but it is used in a variety of clinical settings such as detection of apnea (Cummiskey, et al. 1982) and monitoring breathing during anesthesia (Hök, et al. 1988). In the area of phonopneumography, monitoring tracheal breath sounds provides several advantages in for gathering and subdividing data. When the sound is quite loud, its intensity is directly related to airflow, and there is a gap in the sound as the pulmonary system switches over from inspiration to expiration. Thus the sound at the trachea can be used to estimate relative flow rates and to provide a clear division between inspiration and expiration (Mussell, et al. 1990).

Breath Sounds through the Chest Wall

The sounds heard through the chest wall are commonly referred to as vesicular breath sounds. These sounds are generally broken into two distinct phases, the inspiratory and the expiratory phase. The expiratory sounds are typically lower in pitch, shorter in duration, and quieter than those of the inspiratory phase (Gavriely 1995, 102). In fact, with all other factors taken into account, the expiratory phase sound is only half the intensity of the inspiratory phase; the reason for this is unknown. Various factors affect, in complex ways, the intensity, pitch and duration of the breath sounds. Some of these changes have diagnostic value, such as the increased intensity of breath sounds over a section of consolidated lung tissue. Others are artifacts of the act of taking the measurements. For instance, it is well documented that intensity and frequency distribution of breath sounds correlates with airflow (Soufflet, et al. 1990), though again, the exact nature of the relationship is not known. The use of a flow meter to measure this airflow has been shown to alter the distribution of frequencies of the noise (Mussell 1990). It is almost impossible to extract correct frequency information from breath sounds without altering them with the flow meter. Other factors that modify breath sounds include changes in lung volume, direction of air flow, breathing patterns, whether the subject is breathing through the mouth or the nose, distribution and volume of upper torso body fat, and the location of sound pickups (Mussell 1992).

Abnormal Breath Sounds

Several different auditory phenomena are included in the concept of abnormal breath sounds. Absent or greatly reduced breath sounds make up one such group of sounds. Greatly increased sounds constitute another. Since intensity of breath sounds is closely correlated with airflow, these conditions are best diagnosed by comparing homologous segments of the lung. Otherwise, the concept of “greatly reduced” or “greatly increased” breath sounds become a subjective measure.

Another category of abnormal breath sounds which bear special discussion is the bronchial breath sound. This phenomenon describes breath sounds heard at the chest wall that greatly resemble breath sounds heard at the trachea (Gavriely 1995, 4). The sounds are louder and higher in pitch than normal breath sounds. Bronchial breath sounds are often heard in association with pneumonia due to the consolidation of the lung parenchyma (Gavriely 1995, 4), and are always a sign of lung disease of some type (Gavriely 1995, 4).

Other categories of abnormal breath sounds which bear mentioning are phase, pitch and amplitude heterophony (Waring 1993, 69). Using a stereophonic stethoscope, the examiner is able to directly compare the sounds of homologous lung segments. If the sound begins in one side prior to the onset of sound in the other side, this is phase heterophony. If the sounds differ in pitch, this is pitch heterophony. A third category, amplitude heterophony, is when one side’s breath sounds are louder than the other side’s breath sounds. Any of these conditions could indicate underlying disease. For example, phase heterophony is usually a sign of large airway obstruction on the delay side (Waring 1980, 64).

Adventitious Breath Sounds

By far, the most commonly understood phenomenon of abnormal breath sounds is grouped under the umbrella term of adventitious breath sounds. These are sounds not present in normal breath sounds. Most adventitious ("added on") breath sounds indicate a lung operating abnormally (Hilman 1993, 57).

Terminology. Describing such adventitious breath sounds has been a difficulty over the years. Anyone who has tried to describe an intermittent engine noise to a mechanic understands the difficulty the medical community faces when trying to describe the sounds in the literature of their profession. The currently accepted terminology of adventitious breath sounds dates back to 1957, to an essay by Robertson (1957). He divided adventitious breath sounds into two broad groups, and each of those groups into two subgroups. These sounds were either continuous or discontinuous. Continuous adventitious breath sounds were further divided into wheezes (if high pitched) or rhonchi (if low pitched). Discontinuous adventitious breath sounds were divided into fine crackles (high pitched) or coarse crackles (low pitched).

Despite the popularity of Robertson's work, the battle over adventitious breath sound terminology is far from settled. The Hippocratic school introduced the term *wheeze* and it continues to be used (Gavriely 1995, 1). Laennec (Gavriely 1995, 1) used the term *rales* to describe what this paper calls *crackle*, and that term still is widely used today. Adjectives appended to the term crackle, many of which date back to Laennec, include wet, dry, consonnating, and crepitating, though little agreement exists as to the exact meaning of such phrases.

Some literature also uses the acronym CALS (Continuous Adventitious Lung Sounds) for simplicity, although this is not yet standard.

Continuous Adventitious Breath Sounds

Although both are continuous in nature, there are obvious differences between the two categories of continuous adventitious breath sounds. Wheezes are higher pitched and have a musical quality, while rhonchi have no such melodic nature (Waring, et al. 1985). Both, however, are often indications that some lung pathology exists.

Wheezes. Wheezes come in a variety of flavors. They are clearly the result of one or more sinusoidal waves, which give them their pleasantly musical quality. The pitch can range from 60 Hz to 3.2 kHz (Kramen 1983). They can be monophonic (single frequency) or polyphonic (multiple frequencies), are rarely confused with any other adventitious sound, and may be heard at several sites on the chest wall (Waring, et al. 1985). Some wheezes can be clearly heard at a considerable distance from the subject (Waring, et al. 1985).

Early literature on the subject of wheezing speculated that it was the result of a similar mechanism as musical wind instruments. Other writings suggested that the cause was the oscillation of normal secretions of the sides of normal airways. Both of the hypotheses have been rejected in favor of the current thinking, which says that the wheeze is induced by airway wall oscillations caused by flow restriction (Gavriely, et al. 1987).

Easily identified and measured, wheezes have a number of diagnostic uses. The duration of wheezing is a useful measure of asthma severity (Baughman and Loudon 1984), while the presence of subtle wheezes has been suggested as an early sign of chronic bronchitis.

Rhonchi. Less is written about rhonchi. Less distinctive than wheezes, they are often confused with other adventitious sounds, such a pleural rubs, coarse crackles, and muscle noises. They are created by non-sinusoidal periodic waveforms with multiple frequencies. They have no musical quality and little diagnostic use.

Discontinuous Adventitious Breath Sounds

Crackles. Crackles usually occur when a group of small airways that were closed suddenly snap open (Forgacs 1978). This is not always a sign of disease. One study (Workum, et al. 1982) of 56 normal women, crackles were observed in 35 of them during one breathing maneuver (slow breathing from reserve volume). Nonetheless, the observation of crackles during normal breathing is often the sign of either pulmonary or cardiovascular disease. In cardiovascular disease, it is thought that the cause is a fluid imbalance.

Classification of crackles is a difficult process, usually requiring a time-expanded scale to differentiate between fine and coarse. Poor recording methods and improper filters can alter one to look like the other.

Timing and duration are often the important diagnostic distinctions for crackles. For instance, patients with bronchiectasis often have long-duration coarse crackles late in both the inspiratory and expiratory phases (Waring 1993, 70). Contrast this with patients suffering from chronic airway diseases, such as bronchitis. These crackles typically occur early in the inspiratory phase and are inconsistent in duration (Nath and Capel 1974). And patients with interstitial lung fibrosis, when they have crackles, tend to have fine crackles late in the inspiratory phase (Waring 1993, 70).

Pleural Sounds. The pleura is a thin serous membrane that envelops each lung and folds back to make a lining for the chest cavity. Normally no noise is associated with it. However inflammation can cause a rubbing together of two pleural surfaces. This sound is called a pleural rub. It has a "low-pitched, repetitive, ratchet-like sound that has been likened to the creaking sound produced by flexing an piece of old leather." It is significant only in that it is often confused with rhonchi or coarse crackles, depending upon the duration of the rub (Forgacs 1969).

Nonrespiratory Chest Sounds.

When analyzing breath sounds, it would be ideal if these were the only sounds present. Many undesired sounds, such as air conditioning and outside traffic, can be eliminated by the use of a sound-insulated booth. However, even with all of these sounds eliminated, there are sounds the body produces which will interfere with any analysis of breath sounds. They are heart sounds, muscle sounds and hair sounds.

Heart Sounds. When listening to heart sounds, the examiner will typically ask the subject to hold his breath so that the noise of breath sounds will not interfere with the analysis. The opposite cannot be achieved however. The subject cannot voluntarily suspend cardiac noise. Thus it is a virtual certainty that heart sounds will be present. The common "lub-dub" sound of a normal heartbeat has the advantage of being regular and low frequency. Both the first and second sounds are longer in duration than crackles, but shorter in duration than most continuous adventitious breath sounds. They can easily be separated from breath sounds (Gavriely 1995, 146). However, some patients will have irregular cardiac rhythms, loud blood flow sounds, or heart murmurs which are not so easily separated. Various signal-processing techniques, such as subtraction (Gavriely 1995, 146), blanking (Pasterkamp, et al. 1989) and filtration (Iyer, et al. 1986), have been employed with varying success. However, some murmurs also disappear during breath-holding maneuvers (Levine and Harvey 1949, 159), making such signal processing techniques of limited value in those situations.

Muscle Sounds. When a breath is taken, muscle movements are produced. These muscle movements produce low-intensity, low-frequency noise that can interfere with sound analysis. Normally stethoscopes don't pick up such noise, but other sensors designed to pick up and amplify noise (such as microphones) may hear the noise made by muscle movement. Muscle movement is typically eliminated by a low-pass filter.

Hair Noises. During auscultation, the diaphragm of the stethoscope is picked up and placed in a variety of locations on the chest wall. If placed over hair, the movement of that hair as the chest wall moves produces sound. Rarely noticed during such an auscultation, the noise is often picked up during breath sound recordings (Gavriely 1995, 24). The hair noises resemble fine crackles in their frequency content and duration, but their waveform has a different shape (Gavriely 1995, 24). Shaving the area prior to attaching the sound sensor, a common practice when attaching EKG sensors, can alleviate this problem.

Transmitted Chest Sounds. Vocal sounds and percussive sounds heard at the chest have diagnostic value. Often an examiner will ask the subject to speak certain words or letter ('a', 'e', and 'ninety-nine' are the most common) (Cohen and Berstein 1991). How these sounds are transmitted through the chest gives important diagnostic clues.

Voice sounds can easily be heard through the chest with a stethoscope, but due to the low-pass filtration of the chest wall the sounds are muffled and obscured (Cohen and Berstein 1991). If the sound is more muffled than usual, this can be an indication of pleural effusion, tumor, etc (Wilkins, et al. 1990, 46). If there is consolidation, pneumonia, etc., these sounds begin to come through with increased clarity and volume (Wilkins, et al. 1990, 53). This condition is known as *bronchophony*. A special case of bronchophony, *egophony*, causes the 'e' sound to be heard as an 'a' sound as it is transmitted through the chest wall.

A thumping of the chest, either directly or through a wooden plate, will cause a sound to be transmitted through the chest. This resonant tone is altered by a variety of pulmonary conditions. When the percussive thump is over an organ or a pleural effusion, the resonant quality will be diminished or eliminated, resulting in a duller, softer sound. Emphysema or a pneumothorax will cause the intensity of the sound to increase (Murray and Nadel 1988, 446).

Phonopneumography

Phonopneumography is the practice of recording and analyzing breath sounds using electronic equipment. William Waring, Professor Emeritus at the Tulane School of Medicine, has done considerable research in the area of phonopneumography. Dr. Waring describes the necessary equipment for phonopneumography as "(1) a recording environment that minimizes contamination of the sound signal by background noise (quiet room, sound chamber); (2) at least one microphone that for most applications can be coupled to the chest wall in order to entrain sound from the lungs; (3) a recording device, …and (4) signal processing equipment, which can include filters (to eliminate unwanted sound frequencies), [and] a spectral analyzer or computer for Fast Fourier transform (FFT)" (Waring 1993, 69).

For this dissertation project, the first requirement was only minimally met. No acoustic chamber was used to filter all unwanted sounds from the experiment. However, precautions were taken to minimize the effects of such noise. They are as follows:

1. Prior to all recording, a recording of the ambient noise was made. This will allow such noise to be considered in subsequent analysis. Additionally, a breath-holding recording was made of each subject to allow for subsequent subtraction of heart and blood flow sounds. This will also eliminate the effect of most stationary ambient noise.

2. Recordings were made in a relatively quiet room during a quiet time of the day. This reduced grossly non-stationary noises (traffic, shouting in the halls). Data samples obviously contaminated with such noises were not included in the final analysis data.

3. All observable stationary noises (air conditioning, noisy light ballasts, etc) were eliminated during recording.

The second requirement was met with four Sennheiser MKE-5 microphones. These microphones were selected for a combination of their low frequency response, overall flat frequency response, and low mass (one gram). They were coupled to the recording equipment with Sennheiser preamplifiers and a Yamaha MLA-7 Line-to-Mic adapter, as recommended by Sennheiser. The microphones were located in four chest wall locations as discussed in Chapter Three and attached to the chest wall by relatively flat conical couplers. The size and shape of the couplers was selected to optimize sound transmission (Wodicka, et al. 1994, Kramen et al. 1995).

The recording equipment selected was a TASCAM DA-88. It is an eight-channel digital recorder. It has linear response from 20 Hz to 20 kHz, offers an excellent signal-to-noise ratio, and is less vulnerable to mechanical instabilities introduced by tape motion in analog recorders. The objective was to "get to digital" as soon as possible. Recording in analog and converting to digital later simply adds another opportunity for error. The DA-88 samples at 44 kHz.

The fourth requirement was filled with an IBM PC compatible computer. The digital recording was converted into a PC file and analyzed using FFT methods. The information from the FFT output was then used as an input to develop the other parametric measures. Time domain measures were obtained from the original signal file.

Methods of Signal Analysis and Parametric Estimation

The data comes to the computer in the form of a standard WAV file. This format enables the sounds to be played by Windows playback devices such as sound recorder.exe. When read into an analysis program, the data becomes a vector of amplitude values which correspond to the sound pressure at the microphone. Even short segments such as a 2.5 second inspiration present the researcher with more than 110,000 sixteen-bit words of data. Such an overwhelming mass of data defies simple attempts to analyze. The traditional method is to reduce the data to parameters that describe in a meaningful way attributes of the data taken as a whole. But prior to such analysis, the data must be gathered, stored and preprocessed.

Data Gathering Issues

Some of the issues relating to the gathering of data have previously been discussed. The most important of these issues involve sampling rate, storage capacity, resonant characteristics of the equipment, and noise abatement.

Sampling of a stationary signal requires that it be sampled at a rate at least twice the highest component frequency of that signal. Breath sounds are not stationary, however, and higher sampling rates are necessary to get an accurate picture of such short events as crackles. Gavriely (1995, 56) recommends a sampling rate of 10,000 Hz to adequately reconstruct breath sounds. There is no penalty for higher sampling rates except for increased storage requirements. The equipment used in this project samples at 44,100 Hz, more than sufficient to meet the requirement.

Data storage is frequently mentioned as a consideration in designing phonopneumography experiments. The reason, as mentioned above, is that even short samples of sounds sampled at reasonable rates produces an enormous data file. The design of this project removes the concerns of data storage. The DA-88 digital record will store up to one hundred minutes of eight channels of sound input on a standard Hi-8 cassette. After transfer to the PC, a CD-ROM will hold seventy-four minutes of sound. Since most samples are shorter than one minute, data storage is not a problem.

Resonance means the non-linear amplification of certain ranges of frequencies. It can cause parametric analysis to shift falsely to higher or lower values, ultimately leading to incorrect diagnoses. All of the commercial equipment was selected for its linear amplification over normal audio frequencies. The exception to this was the acoustic coupling device. It showed a resonant characteristic around 2 kHz. This presented no problems in this project, as normal breath sounds are found at frequencies much lower than this. However, in future, this resonance will cause difficulties in identifying wheezes and crackles, both of which often provide audio power in frequencies higher than this resonant peak. Both subjects were carefully auscultated to eliminate the concern that such adventitious sounds were present.

Noise abatement has previously been discussed. Ideally the gathering of sound data is done in an acoustically-isolated chamber. This would minimize any background noises that might find their way into the data analysis. Failing that, recording should be done in a quiet environment, samples with clearly extraneous noise should be discarded (this happened with one of sample in this project), and breath-holding data should be gathered to allow for spectral subtraction prior to spectral analysis.

<u>Preprocessing of Data</u>

Three efforts must be undertaken prior to trying to extract useful parameters from the data. These are removal of DC, digital filtering to remove unrelated low and high frequency signals, and division of the entire data into separate inspirations and expirations.

Removal of DC is necessary because amplifiers and filters occasionally add a DC-level constant to the digitized signal. Though DC-related artifacts and inaccuracies are minimized by the design of this project, removal of DC is still a good first step to preprocessing. This is accomplished by subtracting the mean of the data from each sample.

Heart sounds, muscle sounds and other non-respiratory noises add low frequency noise to the data. Some, but not all, is eliminated by data subtraction. As noted earlier, the sound-concentrating coupler selectively amplifies higher frequencies, which gives them undue value in the analysis. Filtering the digital signal prior to analysis can minimize both of these problems. For this project, data was filtered by an eighth order Butterworth filter with a passband stretching between 100 Hz and 1600 Hz.

Finally, with unwanted frequency information removed, the signal must be divided into inspirations and expirations. It is not appropriate to consider these two phenomenons as the same. In fact, for older children and adults, it is common for the expiratory sound to be too faint to be properly analyzed. It is for this reason that most published analyses of respiratory data concentrate on inspiratory data. The original goal of this project was to automate the process of dividing the breath sounds into their constituent parts, but no reliable method was discovered. Other researchers have accomplished this by monitoring flow rate and using this data to determine the beginning and end of inspirations and expirations. But the use of flow meters alters the characteristics of the sounds that are to be analyzed. Therefore, it was desired to divide the breath sounds by comparing the intensity of breath parts to the intensity during the dead time between these parts. This effort will continue in later research, even without additional funding. To expedite the analysis and writing of this dissertation, the breath data was divided manually by the researcher, using the attempts at automatic division as a guide for the division where appropriate.

Spectral Analysis and Matlab

Matlab software (Mathworks, Incorporated, release 5.2.0), with its signal processing toolbox, was used for data analysis. This software provided many useful tools for preprocessing and spectral analysis. Specifically, Matlab provided the software to create digital filters for preprocessing, FFT routine for spectral analysis and several methods of power distribution analysis for that spectrum. In order to use this software, it was necessary to write a number of script files to automate certain processes. These script files are included in the appendix, along with brief descriptions of important Matlab functions as provided by the Matlab help files. Matlab also created the plots included in Chapter Four.

Parameters for Breath Sound Analysis

Time of Inspiration Start (T_{IS})

Prior to any other analysis or parameter estimation, the question of when sound begins must be answered. Individual samples may be positive, negative or zero but no single sample can be pointed to and called the start of the breath. Two methods can be utilized to make this determination. The first is a mechanical analysis of the sound by the experimenter. The playback would be monitored and marked digitally to note the onset of the first sound. This method also works for the remaining timing parameters. Appealing in its simplicity, this method should be rejected for its lack of automation (surely a goal of such research as this) and its relative subjectivity. Still, for the purposes of this project, this method was used to expedite the research.

The automated method is enabled by the fact that each sample is preceded by a breath-holding maneuver. The beginning of the sample can be expected to cause an overall increase in sound energy. By comparing the overall sound energy of small sample groups in a sliding window fashion, a point in time when the increase increases beyond a predetermined threshold will be found. The first sample in this group will be labeled as the Time of Inspiration Start (T_{IS}). Formally it will be when a group of 128 samples shows an increase of 25 percent in overall power over the previous sample.

Time of Inspiration End (T_{IE})

When the power drops back to the level of breath holding, it will indicate a cessation of respiratory sound. If this follows an inspiration, it will be the time that the respiratory system is changing over from inspiration to expiration. The first sample in the group which shows such a drop in power will be called the Time of Inspiration (T_{IE}). These two parameters will mark the beginning, end, and duration of inspiration.

Time of Expiration Start (T_{ES}) and End (T_{EE})

In the exact same fashion, the Time of Expiration Start (T_{ES}) and the Time of Expiration End (T_{EE}) will be found. These two parameters will mark the beginning, end, and duration of expiration.

Central Power Frequency (f_{50})

As noted above, the end result of spectral analysis is a series of values that represent the power contained in groups of frequencies centered about the enumerated frequency. Adding all of these power levels will give an overall power level, a value which was used in determining the beginning and the end of inspiration and expiration. There exists a frequency below which one-half of the power energy is contained. That would be the Central Power Frequency (f_{50}). In a perfect analysis, this value could be identified exactly. However, a real-world analysis requires a more relaxed definition. For this project, f_{50} will be the lowest enumerated frequency in a spectral analysis which has below it at least fifty percent of the total power energy.

Low Power Frequency (f_{10})

Another common frequency parameter in breath sound analysis is the Low Power Frequency (f_{10}). For this project, f_{10} will be the lowest enumerated frequency in a spectral analysis which has below it at least ten percent of the total power energy.

High Power Frequency (f_{90})

The final common frequency parameter used in this project is the High Power Frequency (f_{90}). For this project, f_{90} will be the lowest enumerated frequency in a spectral analysis which has below it at least 90 percent of the total power energy.

Power Frequency Ratios ($f_{90/10}$), ($f_{90/50}$), ($f_{50/10}$).

Introduced in this project is the concept of the ratio of the three power parameters. The experimenter believes that these ratios may prove useful in diagnosis. There is a variation between subjects of normal frequency responses. What may be a high f_{90} for one subject may be perfectly normal from another. This would usually be reflected in a higher f_{10} as well. These ratios would reflect reduced or enlarged ranges of frequencies that would be normalized to the particular subject. A much larger study with more subjects would be required to demonstrate the truth of this statement.

Homologous Lobe Relationships

As noted earlier in this chapter, the stereophonic stethoscope provided advantages in auscultatory diagnosis that have largely been abandoned. One of these advantages was the direct comparison of homologous lobes of the lungs. In each of the subjects' four samples, two paired homologous lobes were recorded. The analysis will be able to check for phase heterophony (one lobe starting or ending earlier than the other) by comparing the start and end times for comparable values. The analysis will be able to check for pitch heterophony by comparing the three power parameters. The analysis will be able to check for pitch heterophony by comparing the three power parameters. Finally, the analysis will be able to check for amplitude heterophony by comparing the overall power levels of each lobe. Amplitude analysis has proven to be of little use to past researchers due to the variability in amplitude between subjects and the different equipment and methods used in the various studies. This researcher believes that by comparing amplitudes of homologous lobes for the same breath or breath segment, the weaknesses of such analysis will be overcome. Any variability (usually loud or soft breathing) in a particular subject will be compensated for by the fact that the comparison is with that same subject. This analysis will also translate to louder environments, different sensors, and different recording methods, since such differences will be reflected equally in both lobes. This analysis is a major part of the uniqueness of this study.

Linear Lobe Relationships

Also unique to this study is the comparison of non-homologous lobes in the same lung. The relationship of the sounds in such lobes has only been studied one time. In a non-published study, Huberto Hidalgo (while a resident at the Tulane Medical School in New Orleans) used a two-microphone setup to study the sounds in two lobes of a lung. He reported in a poster session that the frequency response of the lower lobe was generally lower than the frequency response of the upper lobe. No further analysis of this phenomenon was done and no other researcher has considered such a relationship.

A lung with an obstruction in a lower lobe will not show this relationship. The obstruction will cause the overall frequency of that lobe to rise, inverting the relationship. This researcher believed that by plotting a curve of the Low, Central and High Frequencies of each lobe, a new parameter could be established. This project, due to limited equipment, was limited to two such lobes. This parameter will be the slope of the line connecting these frequencies. A better experiment with more microphones could be expected to find a more complex parameter relating these lobes.

Time Expanded Wave Analysis

Previously described parameters were either whole breath, inspiratory, or expiratory in nature. Such parameters are useful in their own way, but fail to properly diagnose many adventitious sounds that occupy only a portion of the total time of a breath. For instance, a short wheeze may only change the various power frequencies by a small amount. Lost in the enormity of the data, this vital clue would be overlooked. By sampling at such a high rate compared to the frequency of the actual sounds, it is possible to divide the data into small time samples to be analyzed separately. At 41.8 kHz sampling rate, each sample is 23.9 μS in length. This means that each 2090 samples represent 50 ms, while 836 samples represents 20 ms. By analyzing such smaller segments of data and plotting their power spectrum in a time dependent fashion, a short wheeze and even faint crackles can be clearly seen. But the output plots will not be necessary in most cases, as the adventitious sounds will create sections of dramatically changed central power parameters. A lung segment without adventitious sounds will show generally similar power parameters throughout the entire inspiration and expiration. Any temporary change in these power parameters would indicate adventitious sounds present in the sample.

CHAPTER 3

THE EXPERIMENT

Statement of the Problem

From the discussion in the section on prior art, it is clear that auscultation is an important diagnostic tool. Equally clear was that it was a tool with some important limitations. The most important of these are now listed and will be discussed in depth below.

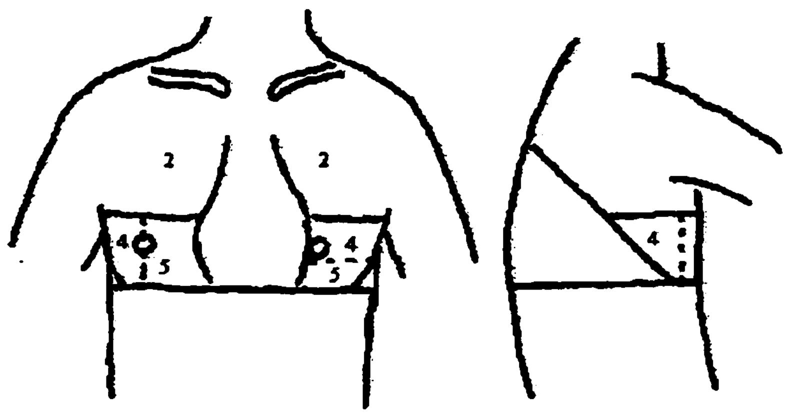
1) The process is an expensive procedure, requiring lengthy analysis by a highly paid practitioner, the medical doctor.
2) The process is a difficult one, so much so that the best equipment (the binaural stethoscope) has been abandoned in favor of the inferior but easier to use monaural version. Even with this simplification, research has shown that the physician often misses important clues from the examination.
3) The process is full of complex human judgements involving the comparison of a sound with the memory of a sound. The human mind, though a remarkable instrument, is not well suited for such analysis.
4) Auscultation is a highly subjective practice. The determination of abnormally high pitched or loud breath sounds depend upon the hearing and experience of the examiner.

Further it can be seen that phonopneumography, though not without promise, has not yielded significant improvements in auscultation. The purpose of this dissertation project is to investigate a computer assisted method of auscultation that will eliminate or improve upon the limitations listed above.

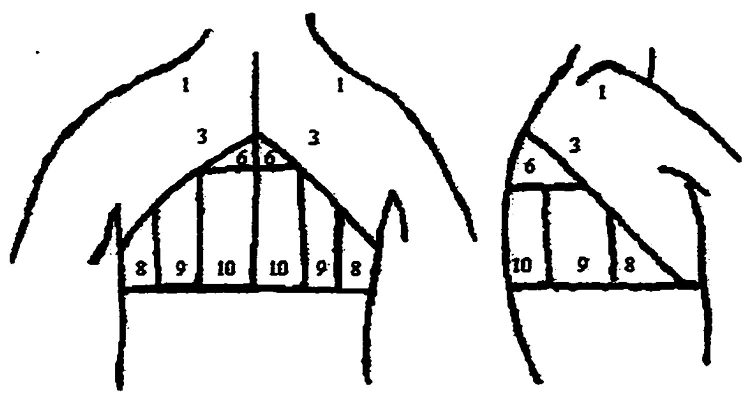
Design of the Experiment

The Ideal Experiment

The ideal experiment would be to gather simultaneous sound from a patient from nineteen locations as shown in Figure 3-1. These locations represent the projections of the lung segments on the torso. (It is noted that there exists a segment numbered seven which has no such projection and cannot be monitored directly by auscultation.) One microphone would monitor the sounds from the trachea (it would be ideally located at the suprasternal notch). The remaining eighteen microphones would be located about the patient's torso as shown in the figure. The exact locations would be determined with a qualified medical professional listening to the patient's lungs and determining the approximate middle of each lung segment. A twentieth microphone would remain unattached but near the patient to record ambient noise.



Lung Segment Projections
Anterior Chest



Lung Segment Projections
Posterior Chest

Figure 3-1
Projection of Lung Segments onto the Human Torso

The patient would be located in an acoustically-isolated room, seated in an upright and comfortable position. At the start of the test, the patient would be instructed to take a deep breath and hold it for a few seconds. This would allow the gathering of data without breath sound. It would allow the analysis to correct for heart sounds and other ambient noises. Then the patient would be instructed to breathe normally for several seconds, then deeply for several more. This would end the data-gathering portion of the analysis.

The data would be recorded on a digital tape recording system. The data would be transferred to a computer by a digital channel where the sample would be parsed into individual breaths. These individual breaths would then be divided into inspiration and expiration. Deep breathing samples would be analyzed separately from normal breathing samples

Prior to frequency analysis, the ambient channel must be examined to determine whether any sample should be discarded due to unusually high ambient noise, such as a door opening or closing, or the patient coughing or moving. After discarding any such tainted sample, each segment of data would be analyzed for frequency content using a Fast Fourier Transform (FFT). Additionally, the breath-holding data would be analyzed using an FFT algorithm and its spectrum would be subtracted from the spectrum of the breath sample. This would remove the contribution of stationary background noise from the data. Each type of data (deep breath inspiration, normal breath expiration, etc.) would then be averaged to arrive at a representative spectrum. The result will be a high-quality spectral analysis of the particular patient's lung sounds. Discussion of the value of such data and how this value is to be extracted will be delayed until Chapter Four.

Before discussing how the actual implementation of this dissertation project differs from the ideal, it is worthwhile to discuss how such an examination improves upon the limitations discussed earlier, assuming practical information can be derived from the data.

Limitation #1 (the high expense of the physician performing the auscultation). The reason a physician is required to perform the auscultation is the subjective nature of the analysis. Great experience is required to extract meaningful diagnostic information from such sound. The only training requirement placed on the experimenter in this scenario is the capacity to locate the lobes on the torso, a much easier task. Thus a lower-paid medical technician can reliably perform placement of the microphones. This situation compares favorably with the analysis of blood in modern practice. The modern physician no longer keeps a microscope handy to analyze blood samples. Instead patients are referred to a specialized clinic which performs the task less expensively and more reliably than the physician could. The physician is presented with a parametric analysis of the information to be found in the blood sample for his diagnosis.

Limitation #2 (difficulty of the task). As discussed earlier in the dissertation, the difficulty arises from two sources: the persistent use of easier but inferior stethoscopes and the requirement of concentration over a long period of time. This design not only takes advantage of the binaural stethoscope's strengths, it amplifies them. The analysis of homologous lobes with regards to amplitude, pitch and phase heterophony is made available. Further it is possible to compare the sounds of all the lobes simultaneously, an analysis never before possible due to the obvious limitation of the doctor's physiology. And since a machine gathers the data, concentration is not a problem. A side benefit of this procedure is a shorter time of examination for the patient. This should improve compliance, especially for pediatric patients for whom long procedures are trying.

Limitation #3 (difficulty of the analysis). The difficulty in analysis arises due to the requirement that the examiner listen to a lung segment, paying close attention to the sound, then to the homologous segment, comparing the new sound with the remembered sound. Both sounds must also be compared with the memory of the idealized lung sound. Such comparisons are difficult to do effectively.

Limitation #4 (lack of objectivity). Each examiner is limited by his/her own experience. This researcher has personally observed two physicians listening to the same patient and disagreeing on whether or not the sound was abnormal. This subjective analysis of the sound limits the diagnostic ability of each physician.

The Limited Experiment

The difficulty for the researcher in implementing the ideal experiment can be summed up in a single word: cost. Digital multi-channel recorders cost approximately $5,000 for each eight-channel unit. Additional hardware is required to synchronize multiple recorders so that data integrity is maintained. Microphones that meet the linearity requirement and are small enough to be attached easily to the torso of a patient are expensive as well. The units selected for this experiment (the Sennheiser MKE-5), along with the required preamplifier cost approximately \$600. Adding the cost of the microphone-to-line adapter and a rack capable of holding the units, the cost of the equipment is approximately \$34,600. A budget for setting up such an experiment is included in Chapter Six. Such cost was beyond the meager budget of the researcher.

Some funding to purchase equipment was obtained through LSU Medical School, through the cooperation of Dr. Bettina C. Hilman, Director of Allergy and Immunology. Her interest was fueled by her extensive practice with asthma and cystic fibrosis patients. The College of Engineering and Science provided some additional funding. The remainder of the cost was borne by the researcher. Thus the experiment was limited to four microphones.

Additional money would be required to soundproof a room for data gathering. This could be achieved by gathering data in a recording studio or an acoustic chamber such as those used by audiologists. In the end, it was decided that the benefit for this dissertation project would not be sufficient to justify the cost. Thus the data was gathered in an office with precautions taken to limit the negative effects of the location.

All other aspects of the ideal experiment were achieved.

Equipment Used

An important aspect of this research was the quality of the equipment used to record and store the data. It is important that all the electronic equipment be high quality to avoid the potential for "loss in [respiratory sound] quality" (Mussell 1992). What follows is a summary of the important pieces of electronic equipment used and the characteristics that guard against loss of quality.

Recorder

The recorder chosen was the Tascam DA-88. This is a studio-quality eight-channel digital tape recorder (DAT). It samples the incoming analog signal at either 44,100 Hz or 48,000 Hz (user selectable) and stores the resulting digital signals on an eight millimeter tape. Its frequency response is flat from 20 Hz to 20 kHz (± 0.5dB). It has a dynamic range of more than 92 dB with a total harmonic distortion of less than 0.00496%. Its channel separation is better than 90 dB at 1 kHz and its wow and flutter is too small to be measurable.

Microphones

The Sennheiser MKE-2 microphone is an omnidirectional microphone with a flat frequency response from 20 Hz to 20 kHz (± 3 dB). It has a diameter of six mm and weighs only one gram. The flat response coupled with the small size and weight made this an ideal choice for the project. It was coupled with the Sennheiser K-6 in-line preamplifier to provide a sufficient signal to the Yamaha Microphone to Line Adapter.

Microphone to Line Adapter

Sennheiser recommends that its microphones be coupled to recording or amplifying equipment using the Yamaha MLA7 8-Channel Microphone/Line Adapter. Its frequency response is relatively flat from 20 Hz to 20 kHz (+1, -3dB). It provides an amplification of up to 64 dB with a total harmonic distortion of less than 0.1%. Its channel separation is better than 70 dB at 1 kHz and its hum and noise is –128 dBμ equivalent input noise.

Acoustic Coupler

Figure 3-2 shows a photograph of the plastic acoustic coupler used to attach the microphone to the chest and concentrate the sound. A small hole was drilled in the side to prevent saturating the microphone. The cavity held the microphone approximately five millimeters from the skin. The coupler had a resonant frequency of approximately 2 kHz, limiting this experiment to normal breath sounds. Adventitious sounds would be distorted by this resonance. For the purpose of this dissertation project, this was acceptable. A better coupler, one with a significantly higher resonant frequency, will be needed to further pursue the research.

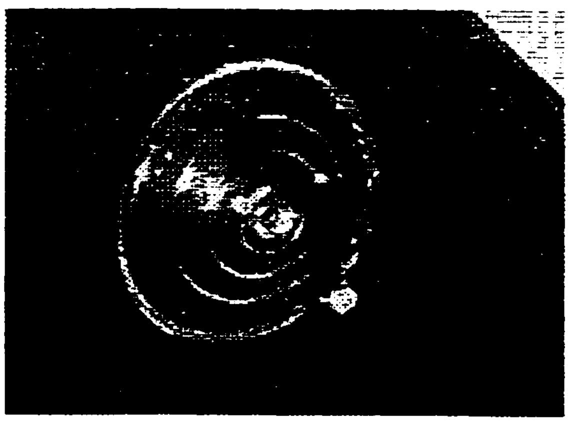


Figure 3-2
Microphone and Acoustic Coupler

Transfer of Audio Data to PC

It was originally hoped that the movement of data from the DAT to the PC would be accomplished by the purchase of a sound card that would accept the Teac Digital Interface Format (TDIF) output of the DAT. Although this is possible with the

Alesis ADAT, Tascam's major competitor, no such card exists for the TDIF format. The kind people at Apogee donated an FC-8 format converter to help, but it still required an Alesis ADAT to make the transfer possible. The actual transfer was accomplished by Carvil Avis of Soigné Studios in Monroe. He was able to borrow an Alesis ADAT and make the transfer for the project. This researcher is greatly indebted to both Apogee Electronics and Carvil Avis.

The complete system is shown in Figure 3-5 at the end of this chapter.

Collection of Data

With equipment in place through the kind patronage of Louisiana State University Medical School, Shreveport and the College of Engineering and Science, Louisiana Tech University, this researcher was prepared to go forward with data collection. The actual data will be discussed in Chapter Four and its significance in Chapter Five. The remainder of this chapter is devoted to issues and procedures surrounding the actual collection of data.

Statement Regarding Use of Human Experimentation

Human experimentation is a process that must be taken seriously. When that experimentation uses children as the subject, the requirements upon the researcher are more stringent, as the possibility of abuse of such children is strong, and the children cannot themselves give informed consent to a procedure.

This process has the advantage of being completely risk free. Perhaps an infection could occur from the adhesive used to attach the microphones, but this remote possibility is the only negative physical sequela that could be imagined by this researcher. Additional danger is posed by the undesired sharing of medical information, but no such information was gathered from either subject except for the fact that they breathed. Informed consent was obtained from the children's parents, both of whom clearly understood the procedure and the purpose of the data gathering.

To add further credibility to this researcher's process, it has been reviewed by the Institutional Review Board for Human Research (IRB) of Louisiana State University Medical School and approved. The Louisiana Tech University Human Use Committee also reviewed and approved the project.

Recording Procedure

During a quarter break, on a quiet afternoon, the researcher brought two subjects to his office. The choice of location was largely determined by the fact that that was where the recording equipment was located and it was difficult to move. One subject was escorted to an empty classroom down the hall and the air conditioning and lights were turned off to diminish background noise. The researcher then auscultated the first subject to determine the approximate center of lung segments two and five (refer back to Figure 1). The microphones were attached (as seen in Figure 2) using medical adhesive tape, but a lubricating jelly was used between the acoustic coupler and the skin to minimize friction noise. Then the subject was asked to hold her breath. This was to gather data on the background noise, both from the surroundings and internal to the subject. Then the subject was asked to take three deep breaths. Finally the subject was asked to breathe normally for three deep breaths. The researcher then listened to the recording to be certain that data were actually collected. This concluded the gathering of data for subject #1. Subject #1 was escorted to the empty classroom, while subject #2 was brought to the office, where the entire procedure of gathering data was repeated.

Discussion of the Subjects Used in Gathering Data

For the purpose of the project, data were gathered from two subjects. This limited first trial was to determine the efficacy of the process before continuing on to larger trials. After realizing the difficulty in transferring the data to the PC for analysis, a decision was made to hold off on further data gathering until a better-funded, fully equipped project could be undertaken. Such a project is discussed in Chapter Six.

Subject #1 was a seven-year-old female, approximately forty pounds. Her slight physical build meant no significant body fat would interfere with the sounds. Although she has been diagnosed with cystic fibrosis, she has no lung involvement to date and will be treated as a normal for the purposes of this dissertation. Multiple physicians check her lung status four times a year. In fact, she may be the most frequently auscultated subject this researcher has ever known, making her the ideal "normal" for data gathering purposes. Her data consists of three deep breaths (inspirations and expirations) and two normal breaths. A third normal breath was discarded due to a sudden sharp background noise that contaminated the data. Her picture, with the microphones attached, is shown as Figure 3.-3.

Subject #2 was a thirteen-year-old male, approximately eighty-five pounds. Although not quite as slight as subject #1, he also had very little body fat, allowing for maximum sound from his lungs. His data consists of three deep breaths and three normal breaths. His picture, with the microphones attached, is shown as Figure 3-4.

Figure 3-3
Subject #1 with Microphones Attached

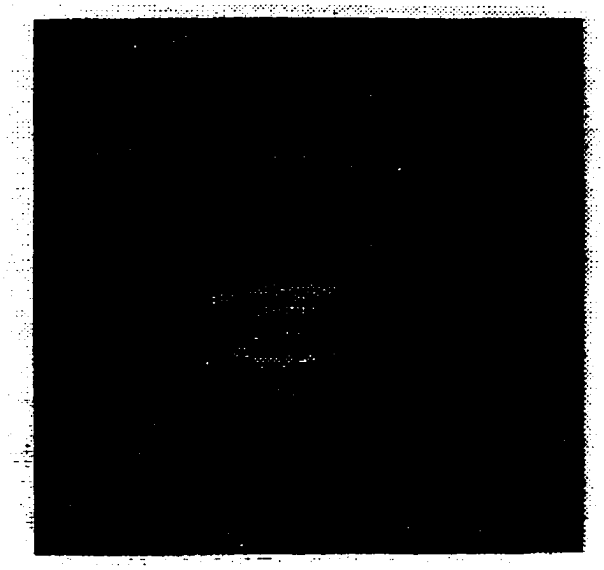


Figure 3-4
Subject #2 with Microphones Attached

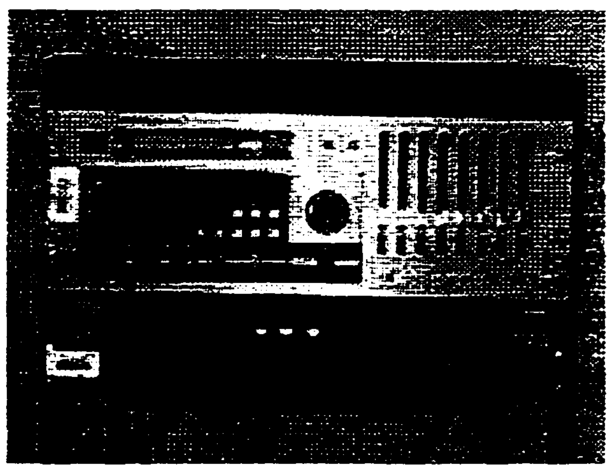


Figure 3-5
Recording System

CHAPTER 4

ANALYSIS OF THE DATA

Data Format as Gathered

As noted in Chapter Three, the data was recorded on a digital tape recorder and transferred to the PC by Soigné Studios. As presented to the researcher, the data was in the form of four files in the WAV format. This is a standard PC sound file format that is a special case of the Resource Interchange File Format (RIFF) file type. As such it is easily manipulated by the PC. Windows includes several utilities for playing and modifying WAV files. One of these, Sound Recorder, proved quite valuable in the process of subdividing the data into separate breath parts for analysis. Additionally, MatLab has a built-in command (wavread) which will input a WAV file as a vector for analysis. This routine will also supply the sampling rate as stored in the header of the file and the number of bits used to represent each sample of data. In the case of this data, it was sampled at 44,100 Hz and used sixteen bits per sample.

The original files, named by Carvil Avis, were Bill Ray2M.wav, Bill Ray3M.wav, Bill Ray4M.wav and Bill Ray5M.wav. Each file was 8,352,044 bytes long including the header and included slightly over one minute thirty-four seconds of sound data. The numbers in the names indicated the recording channel. Channel one recorded provided a recording channel for the researcher's voice and was unnecessary. Channel two recorded the upper segment of the right lung (segment 2R). Channel three recorded the upper segment of the left lung (segment 2L). Channel four recorded the lower inside segment of the right lung (segment 5R). Channel five recorded the lower inside segment of the left lung (segment 5L). Each file contained all the data for both subjects as well as the spoken words of the researcher that identified which portion of the data was to be associated with which subject and which type of data (breath holding, deep breathing, or normal breathing). The general layout of the original files can be seen in Figure 4-1.

Subdividing the Data

Dividing into Individual Subjects

The first step in data division was to remove the extraneous information at the beginning and end of the file and then to subdivide into subject #1 data and subject #2 data. This process was complicated by the fact that no tool was available to handle multiple channels simultaneously. Sound Recorder could be used to easily divide any of the original files, but then the other three files would have to be divided exactly the same to preserve the integrity of the multichannel breath samples. To accomplish this, Sound Recorder was used to remove the dead time and speech at the beginning of the file and the dead time at the end. Then the researcher wrote a MatLab script to extract the middle of a file using the first file as a template to make certain they were the same length. This script, called *ExtractMiddle.m*, is found in the appendix. The result is illustrated in Figure 4-2.

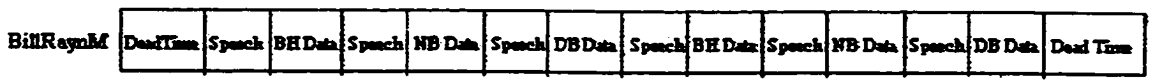


Figure 4-1
Layout of Original Data File

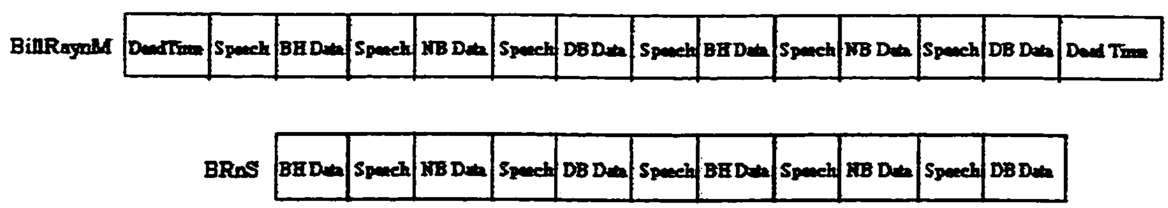


Figure 4-2
Initial Data File Reduction

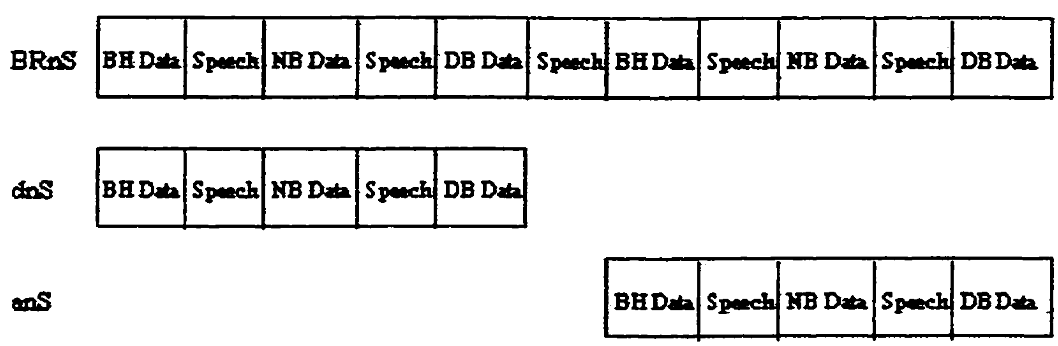


Figure 4-3
Subdividing of Data into Subject Files

The resulting files were named BR2S.wav through BR5S.wav and were considerable shorter (5,923,588 bytes, one minute, seven seconds long).

These files were then divided using the same method into subject #1 data (a2s.wav through a5s.wav) and subject #2 data (d2s.wav through d5s.wav). This division was accomplished in the same fashion (Sound Recorder then *ExtractMiddle*) and is illustrated in Figure 4-3.

Finally, for this portion of the data dividing process, each individual subject file was divided into a breath holding file (ABH2.wav, etc), a normal breathing file (ANB2.wav, etc.) and a deep breathing file (ADB2.wav, etc.). This process is illustrated for subject #1 in Figure 4-4. The file sizes for each of these files and their lengths are shown in Table 4-1.

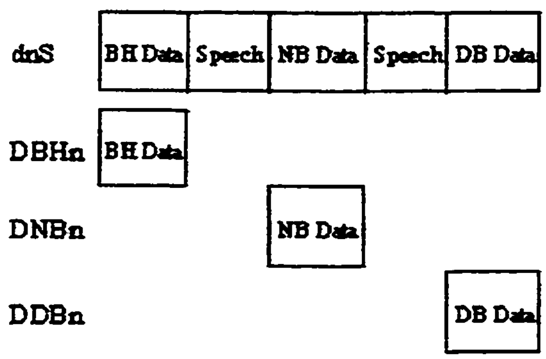


Figure 4-4
Division of Subject Files into Breath Files

Table 4-1
File Sizes for Larger Divisions

Filename	File size (bytes)	Sound length (seconds)
AnS.wav	2,014,448	22.84
ABHn.wav	220,556	2.50
ADBn.wav	645,754	7.32
ANBn.wav	695,054	7.88
DnS.wav	3,233,706	36.66
DBHn.wav	269,376	3.54
DDBn.wav	1,033,734	11.72
DNBn.wav	920,024	10.43

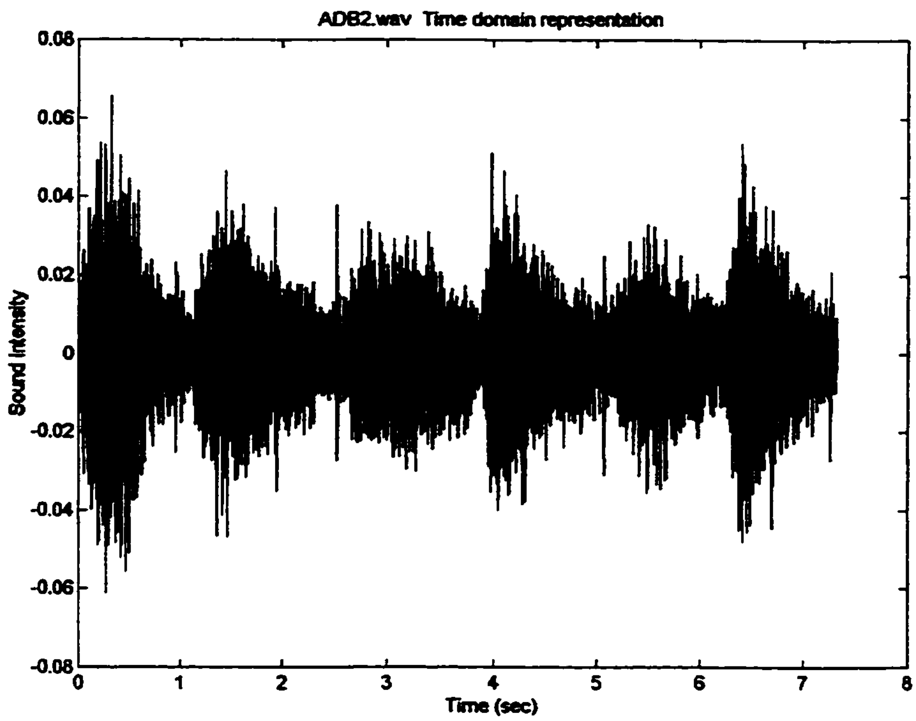


Figure 4-5
Time Domain Representation of a Sound File

Figure 4-5 shows one of the files, ADB2.wav, in time-domain representation. The three breaths can be clearly seen as the intensity of the sound increases and decreases. Further it can be noted that there is no clear norm for the intensity for this subject. This is typical of younger pediatric patients. Their lesser compliance is not a matter of stubbornness, it is simply one of less control.

Dividing into Individual Breaths

The original goal of this project was to devise a way to automatically divide each of the breath files (such as ADB2.wav) into individual inspirations and expirations. This was premised on the fact that there was dead time as the respiratory system reversed direction. However it can be seen in Figure 4-5 that the power levels in the down time was not consistent enough to accomplish this automatically. A MatLab script (*FindBreaks*) was developed to try this. It failed, but did provide some insight in dividing the files manually. The MatLab script *GetBreathParts* was developed to subdivide the breath files into individual inspirations and expirations. It required as an input the number of breaths in the file and the sample numbers to start and stop the inspiration or expiration. These inputs were requested interactively. It would then open each channel's contribution to that data and extract the exact same samples. This would create a hierarchy of file names such as ADB2E1 (subject #1 deep breathing, channel 2, expiration #1) based upon the original file name (ADB2.wav in the above case). Figure 4-6 shows a representative inspiration file in the time domain. The file sizes for each of these files and their lengths are shown in Table 4-2 below the figure. No attempt was made to standardize the individual files. Each simply represented a significant middle of the breath part. This manual extraction of the breath parts did mean that the time-oriented parameters could not be found. This will have to await the next phase of the research.

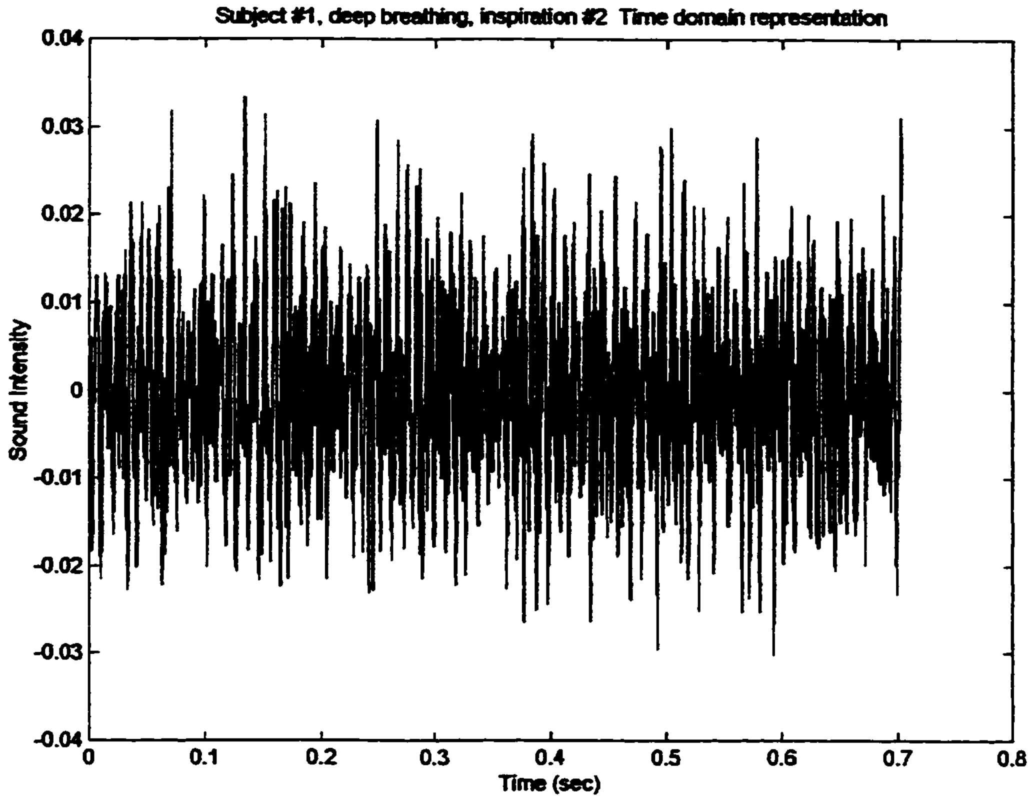


Figure 4-6
Time Domain Representation of an Inspiration

Parameter Extraction

With the individual inspirations and expirations in hand, the frequency parameters can now be extracted. This was done with a Matlab script called FindParameters. This routine calculated the frequency spectrum of each individual inspiration or expiration for each subject and for each type of breathing. Then these spectra were averaged and the spectrum of the breath-holding sample was subtracted from this average. The purpose of this was to remove the contribution of the ambient noise, both internal to the subject and external to the subject, from the analysis. Then a power density was calculated by squaring the magnitude of the resulting spectrum.

Table 4-2
File Sizes for Individual Inspirations and Expirations

File name	File size (bytes)	Sound length (seconds)
ADBnI1.wav	69,166	0.78
ADBnI2.wav	61,998	0.70
ADBnI3.wav	46,638	0.53
ADBnE1.wav	89,134	1.10
ADBnE2.wav	65,070	0.74
ADBnE3.wav	71,214	0.81
ANBnI1.wav	81,966	0.93
ANBnI2.wav	66,606	0.76
ANBnE1.wav	54,318	0.62
ANBnE2.wav	78,894	0.89
DDBnI1.wav	106,030	1.20
DDBnI2.wav	119,854	1.36
DDBnI3.wav	184,366	2.90
DDBnE1.wav	136,750	1.55
DDBnE2.wav	225,326	2.55
DDBnE3.wav	80,430	0.91
DNBnI1.wav	110,638	1.25
DNBnI2.wav	92,718	1.51
DNBnI2.wav	53,294	0.60
DNBnE1.wav	141,870	1.61
DNBnE2.wav	120,366	1.36
DNBnE3.wav	122,926	1.39

The total power density was then summed, resulting in a total power for the sample. The power values were then summed, seeking the three frequency parameters of interest. f_{10} is the first frequency which has ten percent of the total power found in frequencies below it. The same is true for f_{50} and f_{90}, except obviously for the percentage of power found below. These parameters were printed to the screen to be recorded by the researcher. Finally a graph of the power spectrum from the lowest value to 1500 Hz was created and saved to disk for further analysis. One such graph is included below in Figure 4-7.

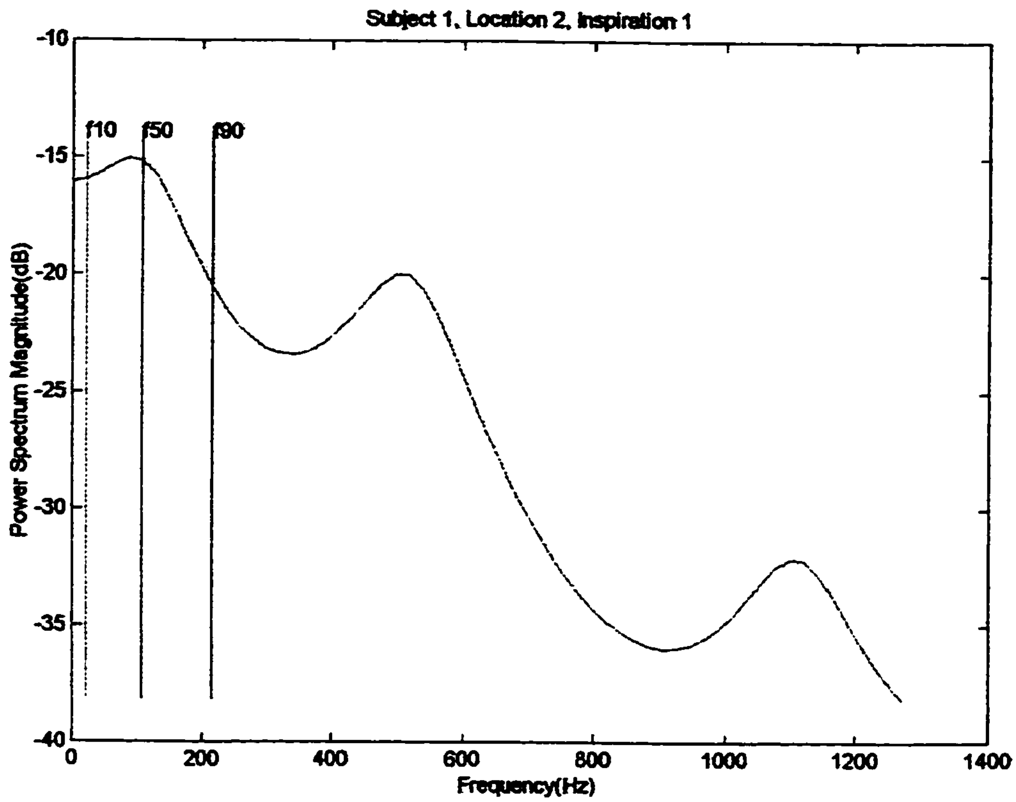


Figure 4-7
Power Spectrum Magnitude for Frequency Parameter Extraction

Parametric Results of Subject #1

Subject #1 was a seven-year-old female. As such she showed somewhat higher frequencies than did subject #2. This was expected from the literature review, which showed that the average spectral characteristics decreased in frequency as the subject grew taller and stouter. There is a clear correlation between the frequency characteristics of the left and right lung for homologous segments. In all cases, the fifty-percent power frequency was the same. The ninety-percent power frequency was slightly higher in one case (deep breathing, expiration), but closer examination of the data shows that this difference is more due to the histogram nature of the FFT. A slight shift of power in the left upper lung would have caused this value to have also been 215.3 Hz, making all homologous lung segments identical in all three parameters. Additionally, it should be noted that with the exception of the above noted case, an upward shift in frequency was noted when comparing the upper lung segment with the lower segment of the same lung. This upward shift could provide the basis for a new diagnostic parameter which is currently not being considered in the research literature and is beyond the capability of a physician to consider with a single-bell stethoscope.

The summary of the data for each breath type can be seen in Figures 4-8 through 4-11.

Parametric Results of Subject #2

Subject #2 showed the same basic parametric correlations as subject #1, but at lower frequencies, typically twenty to thirty hertz lower. However, the same basic truths applied, specifically that homologous segments tended to have the same frequencies and the frequency parameters tended to rise as analysis moved up the lung.

The summary of the data for each breath type can be seen in Figures 4-12 through 4-15.

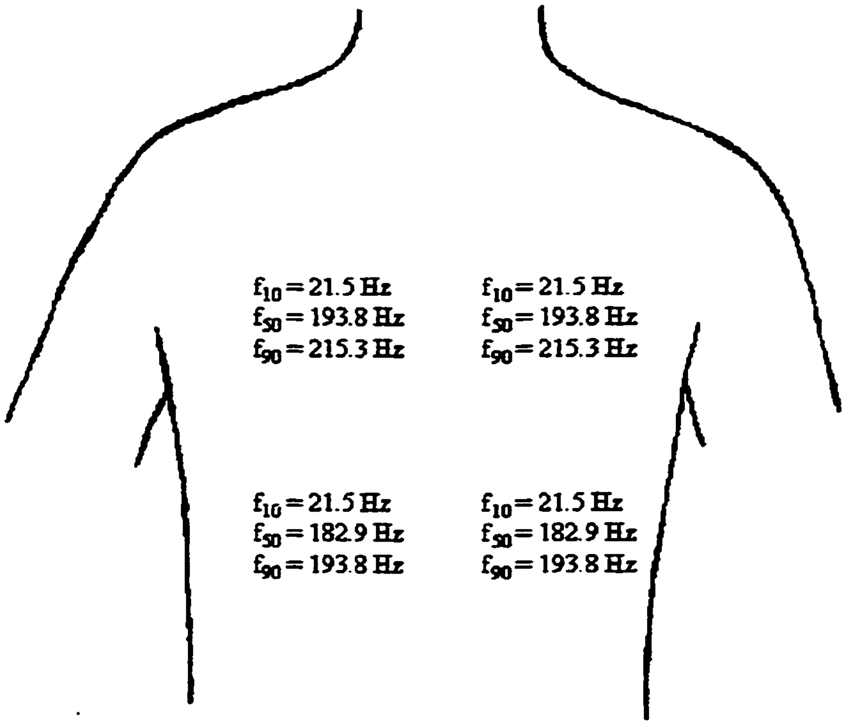


Figure 4-8
Subject #1, Deep Breathing Inspiration Parameters

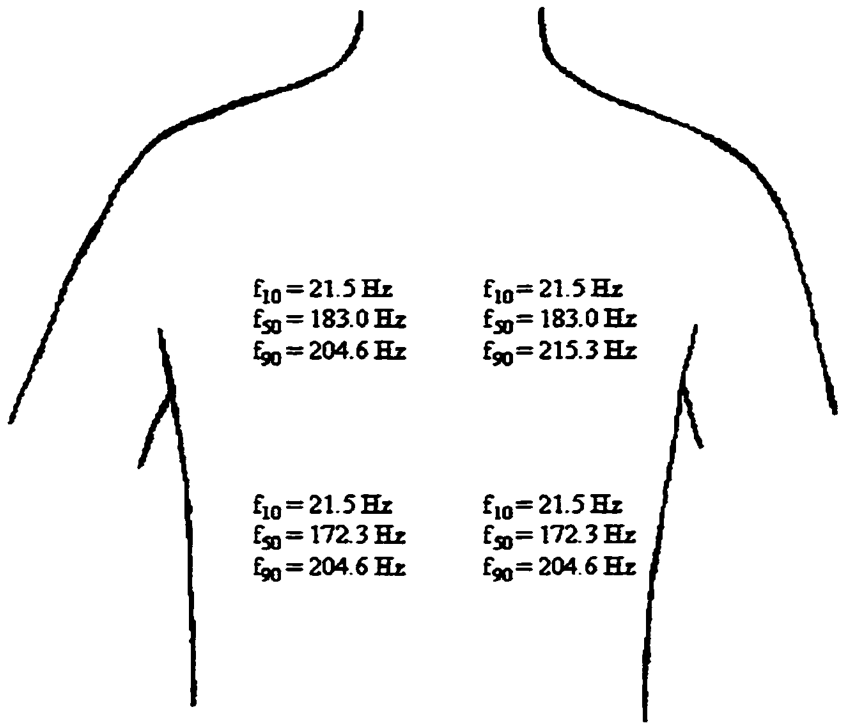


Figure 4-9
Subject #1, Deep Breathing Expiration Parameters

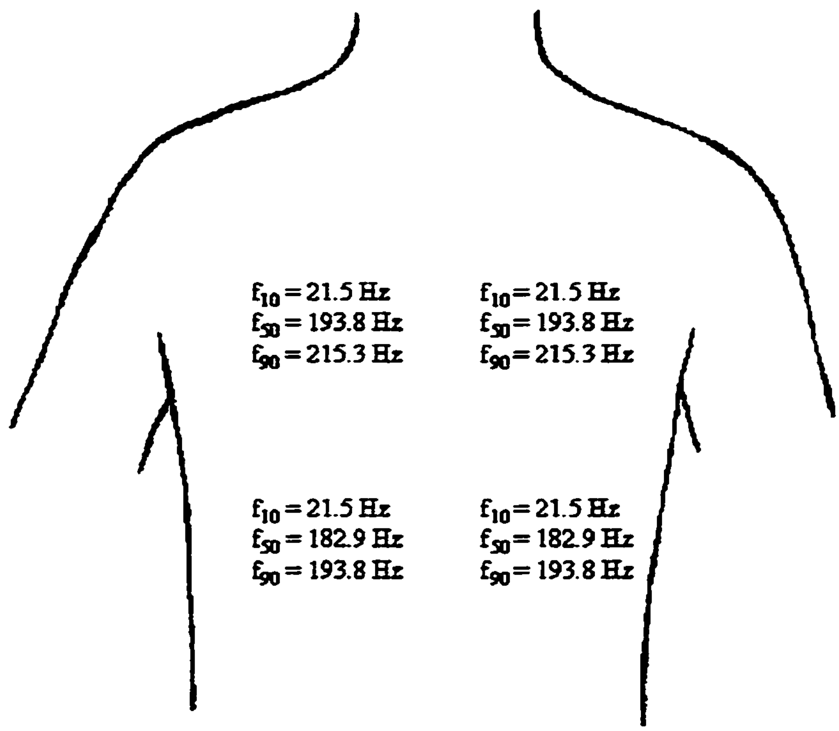


Figure 4-10
Subject #1, Normal Breathing Inspiration Parameters

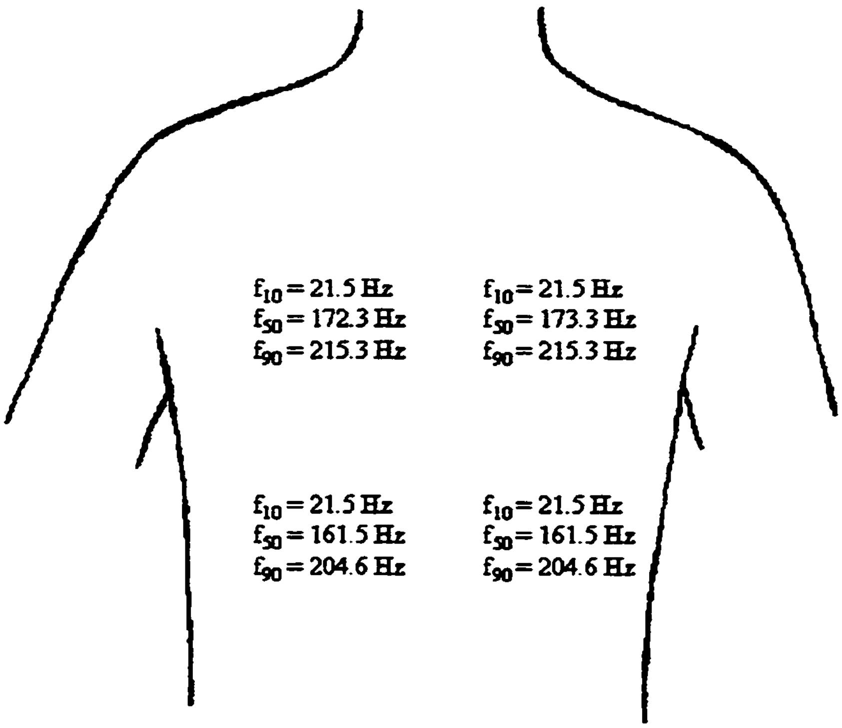


Figure 4-11
Subject #1, Normal Breathing Expiration Parameters

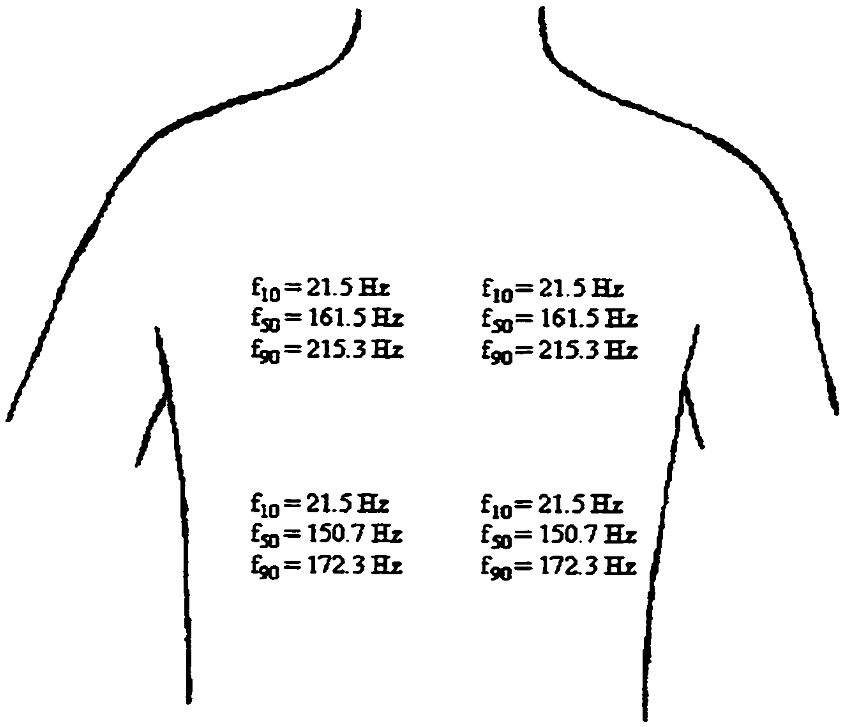


Figure 4-12
Subject #2, Deep Breathing Inspiration Parameters

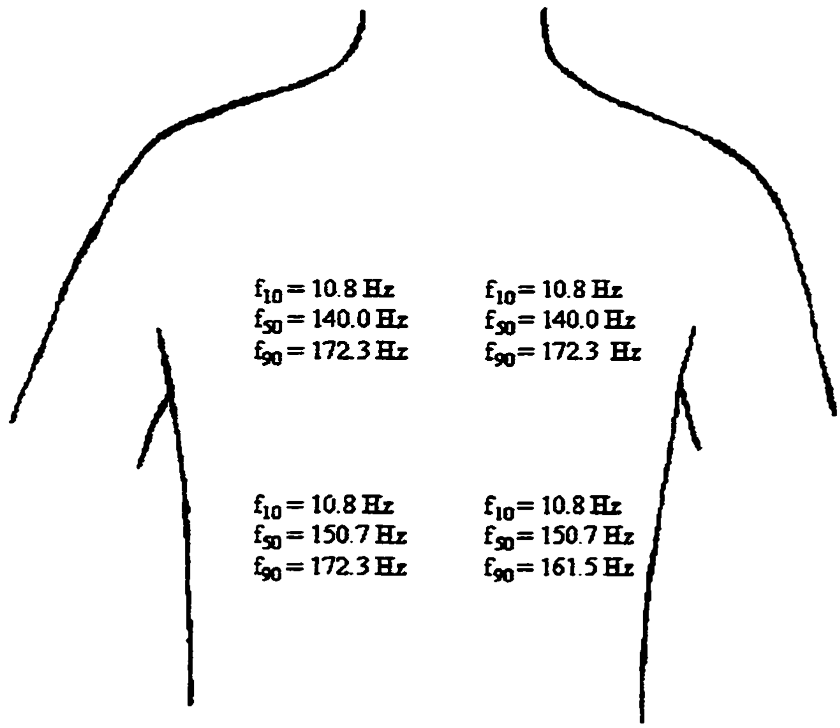


Figure 4-13
Subject #2, Deep Breathing Expiration Parameters

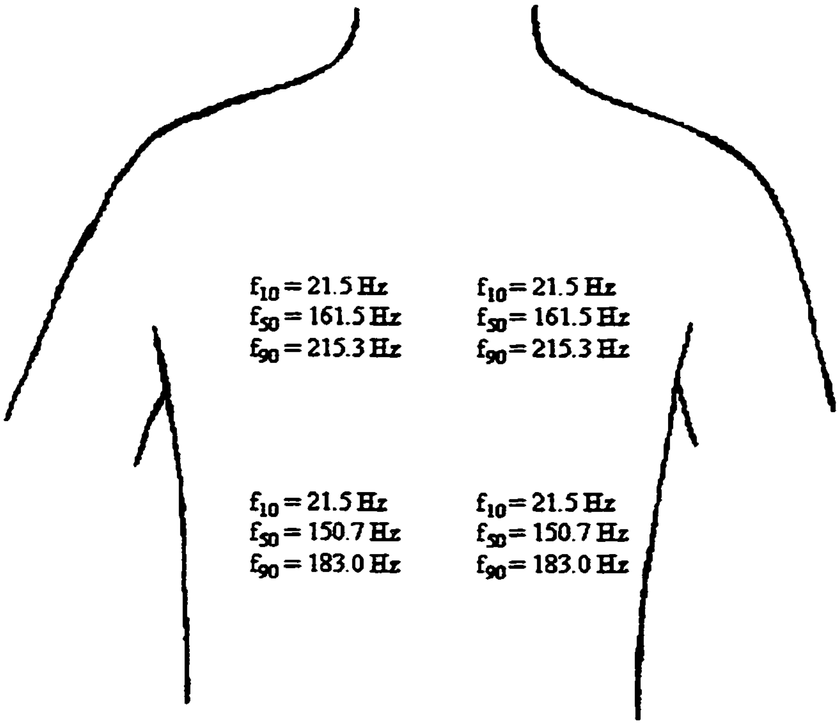


Figure 4-14
Subject #2, Normal Breathing Inspiration Parameters

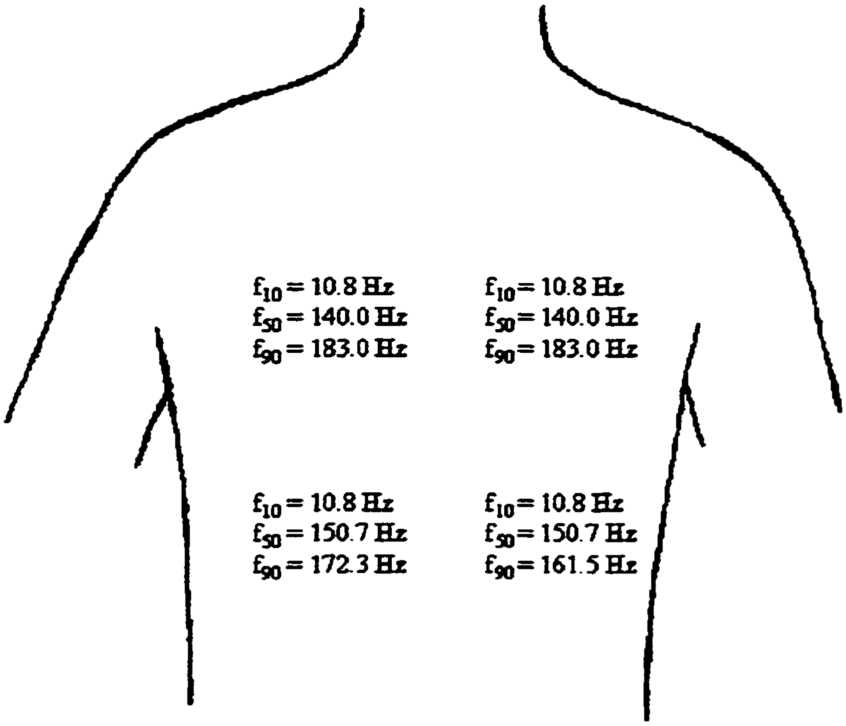


Figure 4-15
Subject #2, Normal Breathing Expiration Parameters

CHAPTER 5

CLAIMS AND CONCLUSIONS

When drawing conclusions from data, caution must be exercised to draw out only what is present and not overstate the value of limited data. While the number of subjects (two) in this study was small, it is not unusually small for a medical study. Studies of four, five and six subjects are common in medical literature, while case studies of a single individual presenting a unique challenge are also common. However no attempt to over-generalize will be made.

It is also noted that the number of sites is also small compared to the ideal. Although the four sites yielded many megabytes of data, care must also be taken to not overstate the value of these four sites which may or may not hold true throughout the lung.

This aside, this analysis improved upon the current state-of-the-art in three ways, and the project gives hope for significant improvements in auscultation technology in the future.

Comparison of Parametric Results for Both Subjects

Chapter Four provided some comparative analysis of the data, but it bears repeating here. Each subject showed similar characteristics when homologous segments were compared. Almost without exception, such lobes exhibited the same frequency parameters regardless of the age, height, weight, or sex of the subject and regardless of the rate of air flow. When a parameter did vary, it was by the minimal amount, and closer analysis of the data showed that the variance was insignificant.

Such similarity was not demonstrated when the same or homologous segments were compared between subjects. Subject #1 showed consistently higher frequency parameters in all segments and at all flow rates. While expected, this does demonstrate the difficulty of trying to do auscultatory diagnosis based upon population norms. These norms would have to be carefully segmented by a variety of factors. This segmentation of norms would be unnecessary by this method.

It is further noted that in all cases a shift upward in frequency was noted when comparing the upper segment of a lung with that same lung's lower segment. This consistent rise in frequency gives rise to the hope that a new parameter can be established which will further aid in diagnosis. While the recording of only two sites on each lung gives rise to a straight line, with the slope as the suggested parameter, more recording sites may yield more useful parameters.

Based upon the data presented, this researcher has arrived at three conclusions, which will be presented and explained in the following paragraphs.

Conclusion #1: The Method is Useful

The aim of phonopneumography is an objective analysis of lung function. While this specific research did not completely demonstrate the ultimate utility of this process, the consistency of the data seems to indicate that this process can yield objective data-based parameters which can be analyzed in absence of the actual respiratory sounds. As a further benefit, should some question of diagnosis arise from the parameters, the data, in the form of sound files can be examined by physicians independently or in groups. No longer will two physicians have to listen to separate breaths and discuss whether or not a subtle crackle or wheeze was present. The auscultatory sound will be available for review and discussion, even as an x-ray is available for such group discussion.

Conclusion #2: The Method Improves on Previous Attempts

As stated earlier, previous attempts at establishing phonopneumography concentrated on examining a single segment in multiple subjects and seeking to establish a population norm. These attempts have been a collective failure due to the wide variability in subject size and age, in air flow, and in recording procedures. This research improves upon these attempts by emulating the superior but often unpracticed method of differential auscultation. By allowing the subject to serve as his or her own norm, it can be said more objectively that the sounds heard are normal or abnormal.

This research failed to establish the concept of a parameter associated with phase heterophony, but it is the belief of this researcher that such a set of parameters can be established with continued effort.

Conclusion #3: New Parameters for Diagnosis Have Been Suggested

Consistent differences between the calculated parameters for non-homologous lung segments have been found in this research. While the actual parameter to describe these differences is beyond the scope of this research, its presence is certainly implied.

Such differences give rise to the hope that such pathologies as an obstruction caused by the thickening mucus in an airway can be observed by the more sensitive recording equipment and discovered prior to a true problem manifesting in a larger way.

CHAPTER 6

FUTURE HOPES

As noted earlier, this project was conducted with less than a full complement of equipment required to fully establish the promise of this method. This researcher does have plans to continue to seek funding to purchase a complete experimental laboratory setup and conduct further research. The purpose of this chapter is to outline the future plans and expectations of such research.

Building a Full Data Collection Laboratory

Three eight-channel digital tape recorders are required to collect data in all desired locations on the torso. This project used a TASCAM DA-88 recorder. While the recording quality was acceptable to achieve the desired results, it is difficult to get the data from the recorder to the PC. No manufacturer makes a card for a personal computer which allows the direct transmission of the data. To achieve the transmission, an Apogee FC-8 Format converter is required to synchronize the data transfer from the TASCAM to an Alesis ADAT and from there to a PC. The effort was assisted by Carvil Avis of Soigne Recording Studio in Monroe. However, since such transfers require the use of an Alesis ADAT, it only makes sense to change to the Alesis in the full laboratory arrangement. Nineteen microphones and cables will also be required, as well as nineteen acoustic couplers. Three Yamaha Microphone/Line adapters are required, because each only handles eight channels. A unit to synchronize the starts and stops of the three ADAT is needed. To conserve space, all these units must be rack-mounted. Finally, a sound card that accepts Alesis Light Pipe input to facilitate data transfer is needed. Table 1 shows the full equipment list and approximate costs for each line item, as well as an estimate of the cost of the full set of equipment.

Table 6-1
Recording Laboratory Equipment Estimates

Equipment (number required)	Approximate cost per unit	Total cost
Alesis ADAT (3)	$5,000	$15,000
Yamaha MLA7 Microphone/Line adapters(3)	$600	$1,800
Microphones/preamps/cables (20)	$600	$12,000
Alesis ADAT controller (1)	$1,500	$1,500
ADAT compatable sound card (1)	$700	$700
19" rack (1)	$900	$900
Cables, tapes, etc	$200	$200
PC for data gathering and analysis	$2,500	$2,500
	Total cost	$34,600

This cost clearly is more than this researcher is able to spend out of personal funds, so the first step in achieving the full laboratory setup is to seek funding for the equipment. Having achieved some experience through this dissertation project, the researcher believes that a BORSF proposal would be favorably received by the state. This proposal would allow the researcher to gather data as outlined in the two following sections. It would request money for equipment, release time, graduate student funding, money to pay subjects for participation in the gathering of data. The university would be expected to provide laboratory space in a quiet environment or an acoustic chamber. If funded, the research would be expected to lead to an NIH proposal for a broader research effort.

A Broad Collection of Data from Normals

The data collection would come from two broad groups. The first group would be the normal group. This would be comprised of individuals with no known respiratory problems to provide a strong set of baseline parameters for understanding what is normal variation and what is pathological. The data would initially be drawn from two populations. Louisiana Tech University would provide a large population of eighteen-to-twenty-two year old men and women for such data. For pediatric data, a local school (perhaps A.E. Phillips) would be tapped for potential subjects. All subjects would be recorded under a properly approved protocol for human experimentation which would assure confidentiality of medical information.

This population would need to encompass multiple racial groups, both sexes and would have adequate representation from all body types (tall to short, ectomorph, to obese).

This data collection would occur during the normal nine-month term.

A Broad Collection of Data from Abnormals

In additional to the normal population data, the research would want to gather data from individuals with known respiratory difficulties. Such a group of individuals is available through Dr. Bettina Hilman of LSU Medical School in Shreveport. She is the Chief of the Allergy/Immunology section of the Department of Pediatrics, as well as the Director of the Cystic Fibrosis (CF) Center in Shreveport. Her regular patient load of more than one hundred CF patients cycles through her office once every three months, and she has a strong practice in the treatment of asthma. Further, Dr. Hilman has a great interest in this research and has expressed a desire to assist in this further effort.

Each patient participating in the study will undergo the normal examination. X-rays and other normal diagnostic methodology will occur as needed. This normal routine will provide a known diagnosis, including severity of difficulty, location of problems, and the presence of crackles and wheezes. Such information will enable the researcher to improve software development to search for adventitious sounds and localizing respiratory difficulty to particular segments in the lung.

This data collection would occur during the normal nine-month term, and would be partially funded by the grant that purchased the equipment and partially by LSU Medical School.

A Correlation of Parameters and Diagnoses

The result of this study will be an enormous amount of sound data that will be tied logically to a particular study participant with a known diagnosis. The analysis of the normal participants' data will yield a better understanding of how the parameters relate to one another. It could answer many questions that are troublesome for this research: How much variability from one homologous lung segment to the other is acceptable before a pathology is suspected? How much should the frequency vary as the sound collection site moves up the lung? How do you determine start and stop times for inspirations and expirations, and how much variability is acceptable?

The analysis of the abnormal data will provide a wealth of information for diagnosis of lung pathology. First, it will establish the correctness of the answers to the questions asked above. If five percent variability in frequency parameters is acceptable from one homologous lung segment to the other, do obstructions, consolidations, etc. consistently provide more than a five percent difference? Which conditions yield increased frequencies and which yield decreased frequencies? When an obstruction or thick mucus delays the beginning of breath sounds in a segment, how much delay is expected for known severities of such conditions?

Secondly, it will allow the correlation of specific variances in parameters with specific conditions. It will automate the search for adventitious sounds, and as has been noted earlier, and will locate quieter and shorter sounds.

The Auscultation Lab

The eventual goal of this research would be the creation of an auscultation laboratory at hospitals and clinics. Instead of engaging in auscultation personally, a physician would send the patient to the lab for auscultation, just as would be done for blood work. Such a request would be made of a patient for a normal check-up or if the patient is complaining of respiratory problems.

At the lab, a respiratory technician would gather relevant physical data (height, weight, age, etc.) and place the microphones using a stethoscope. Since the lung segments are well known, and pinpoint accuracy is not required, such microphone placement should only take a few minutes. The technician would then request that the patient breathe deeply for several minutes while recording. (Sound levels should be monitored on each channel to ensure a full collection of data; perhaps this could be automated too). After the data collection, the data will be dumped to the PC for analysis. Finally, a printout of relevant information related to the data collection would be printed out and sent to the doctor. For check-up patients, the recordings would be kept for twenty-four hours to allow physicians to come and listen for themselves if desired. For a patient with respiratory difficulties, the recordings could be burned into a CD-ROM and attached to the patient's medical record, just as an x-ray would be. Thus the progression of a respiratory problem could be accurately followed, by comparing the respiratory sounds of the last visit with the ones of the current visit.

In this fashion, the researcher hopes eventually to raise the standard of care in respiratory medicine.

APPENDIX

MATLAB SCRIPTS

```
function ExtractMiddle(OrigName,NewName)
% Given templates, this function extracts the middle out of a file in
%       a repeated manner.  It is specifically designed to work on my dissertation
%       files, but can be modified to be more general.

% It expects that the user has taken OrigName.wav, and deleted the unwanted
%       end, saving it with the same name with a 'a' appended (Orignamea.wav).

% Then the undesired beginning was deleted and the resulting file was saved
%       with NewName.wav.  Each filename passed should end with a numerical digit,
%       (assumed here to be 2, my dissertation, remember;-) which is replaced
%       repeatedly with 3,4,5 and 6.

FullName=strcat(OrigName,'.wav');            % append the file type
LOrigName=length(OrigName);                  % needed later to locate digit
HoldName=strcat(OrigName,'a.wav');           % told you this was assumed
FullNew=strcat(NewName,'.wav');              % append the file type
LNewName=length(NewName);                    % also needed later

[y,fs,bits]=wavread(HoldName);               % read the intermediate file
e=length(y)                                  % and get its length
[y,fs,bits]=wavread(NewName);                % read the resultant file
b=e-length(y)+1                              % b and e define the middle

for i=3:6                                    % dissertation specific
   stri=num2str(i);                          % to make new names
   NextName=OrigName(1:LOrigName-1);         % Cut off digit
   NextName=strcat(NextName,stri,'.wav');    % and put new one on
   NextNew=NewName(1:LNewName-1);
   NextNew=strcat(NextNew,stri,'.wav');

   [y,fs,bits]=wavread(NextName);            % Read in the next original
   yy=y(b:e);                                % Extract the middle
   length(yy)
   wavwrite(yy,fs,NextNew);        % And save it
end
```

```
function[BreakTimes,b,p]=FindBreaks(wavefile,threshhold)

% This function takes a .wav file as an input and a power level threshhold
%       and searches the wav file for breaks between breaths.

% The threshhold is the actual break threshhold.

% In addition to the vector of power levels, the function prints out the minimum,
%       mean and maximum power level found throughout the file.

p=PowerLevel(wavefile);
n=length(p);
BreakTimes=zeros(n,1);
i=1;
b=1;
BreathSense=0;                          %boolean flag 0=outside breath, 1=inside breath

for j=1:n
    switch BreathSense
    case 0
        if p(j)>threshhold
            BreakTimes(b)=j;
            b=b+1;
            BreathSense=1;
        end
    case 1                  % if BreathSense==1
        if p(j)<threshhold
            BreakTimes(b)=j;
            b=b+1;
            BreathSense=0;
        end
    otherwise
        disp('BreathSense is confused')
    end
end

if b==1
    disp('No breaks found');
end
if BreathSense==1
    disp('quit in the middle of a breath');
end
if BreathSense==0
    if b>1
        disp('quit between breaths');
    end
end
```

```
function [p,Pxx,F]=FindParameters()
%==========================================================================
% This function extracts the final power parameters from an input file, subtracting
%      first the spectrum found in the breath-holding file. It then saves the graph as a
%      jpeg file.
%
% The three returned parameters are as follows:
%      p is the set of power frequency parameters f10, f50 and f90
%      Pxx is the power frequency spectrum
%      F is the array of frequencies for which power was calculated
%
% I ran out of time to generalize it, so I just edited the names and reran it each
%      time for each file.
%==========================================================================

nfft=2048;                          % set the FFT length to 2048
p=zeros(3,1);

[y1,fs,bits]=wavread('adb5I1.wav'); % read in the first inspiration file
                                    % fs is the sampling frequency
                                    % bits is the number of bits per sample

[b,a] = Butter(4, [100/fs 2000/fs]; % implement an 8th order Butterworth filter
F = (0:2047)*fs/nfft;               % vector of fft frequencies

y2 = wavread('adb5I2.wav');         % read in the second inspiration file
y3 = wavread('adb5I3.wav');         % read in the third inspiration file
ybh=wavread('abh5.wav');            % get the breath holding data for
                                    %      spectral subtraction

sy1 = fft(y1,nfft);                 % spectrum for the first inspiration
sy1 = RemoveDC(sy1);                % remove the DC component if any
sy1 = filter(b,a,sy1);              % apply the bandpass filter

% now repeat for the other three files

sy2 = fft(y2,nfft);
sy2 = RemoveDC(sy2);
sy2 = filter(b,a,sy2);

sy3 = fft(y3,nfft);
sy3 = RemoveDC(sy3);
sy3 = filter(b,a,sy3);

sybh = fft(ybh,nfft);
sybh = RemoveDC(sybh);
sybh = filter(b,a,sybh);
```

```
% subtract breath holding from average and calculate the power spectrum

sy = (sy1 + sy2 + sy3)/3 – sybh;
Pxx = abs(sy).^2;

pmax=0;                          % initialize power max and min
pmin=10;
f10=0;                           % initialize parameters to zero
f50=0;
f90=0;

plen=length(Pxx);
psum=0;                          % initialize the total power to zero
for i=8:60                       % ignore low and high frequency data
    psum=psum+Pxx(i);            % calculating the total power
    if Pxx(i) > pmax
        pmax = Pxx(i);
    end
    if Pxx(i) < pmin
        pmin=Pxx(i);
    end
end

p10=psum/10;                     % calculate the 10% power value
p50=psum/2;                      % calculate the 50% power value
p90=psum-p10;                    % calculate the 90% power value

ppart=0;                         % initialize the holding variable
for i=8:60
    ppart=ppart+Pxx(i);          % totaling power again
    if f10 == 0
        if ppart > p10
            f10=F(i);
        end
    end
    if f50 == 0
        if ppart > p50
            f50=F(i);
        end
    end
    if f90 == 0
        if ppart > p90
            f90=F(i);
        end
    end
end
```

```
p(1)=f10;
p(2)=f50;
p(3)=f90;

% Now output a graph

ydata=linspace(10*log10(pmin), 10*log10(pmax*1));
ylen=length(ydata);
x10=ones(ylen,1)*f10;
x50=ones(ylen,1)*f50;
x90=ones(ylen,1)*f90;

newplot;
plot(F(8:60), 10*log10(Pxx(8:60)));
xlabel('Frequency (Hz)'), ylabel('Power Spectrum Magnitude (db)');
title('Subject 1, Location 5, Inspiration');
hold on;
line(x10,ydata,'linestyle', '-');                % add a solid vertical line
line(x50,ydata,'linestyle', '-');
line(x90,ydata,'linestyle', '-');
text(f10*0.9,10*log10(pmax*1),'f 10');
text(f50,10*log10(pmax*1),'f 50');
text(f90,10*log10(pmax*1),'f 90');
hold off;

print -djpeg adb5I.jpg                           % save to a jpg file
```

```
function GetBreathParts(OrigName,b,e)
% Given templates, this function extracts the inspirations and expirations out of a
%     breath file in a repeated manner. It is specifically designed to work
%     on my dissertation files, but can be modified to be more general.

% It expects that the user has taken OrigName.wav and has determined the beginnings
%     and ends of each inspiration and expiration. These values are input, then the
%     original data file is parsed into its parts, each saved with the original name
%     plus I (or E, as appropriate) and the number of the inspiration (or expiration).

% The beginning and ending numbers (b and e) are assumed to have been extracted
%     from analysis of the power levels using a 256-byte sliding window. The
%     numbers, then, will be the number of the 256 byte segment of the breath file.

% It is assumed that the first part is an inspiration, that each breath has both an
%     inspiration and expiration, and that the user did a good job of creating b and e.

BreathNums=length(b)/2;                 % e would work too.

FullName=strcat(OrigName,'.wav');       % append the file type to open
LOrigName=length(OrigName);             % needed later to locate digit
InspName=strcat(OrigName,'I');          % root for inspiration file name
ExpName=strcat(OrigName,'E');           % root for expiration file name

% Open the original file and extract the parts
[y,fs,bits]=wavread(FullName);

for i=1:BreathNums
   k=2*i-1;
   yi=y(b(k):e(k));
   ye=y(b(k+1):e(k+1));
   stri=num2str(i);               % to make new names

   IName=strcat(InspName,stri,'.wav');    % Create full name for this inspiration
   EName=strcat(ExpName,stri,'.wav');     % Create full name for this expiration

   wavwrite(yi,fs,IName);                 % Save inspiration in a separate file
   wavwrite(ye,fs,EName);                 % Save expiration in a separate file
end
```

```
function[outvector]=RemoveDC(y)

% This function takes a sound vector as an input and removes the DC level. This
%       technique was described in Breath Sound Methodology, Chapter 5, page 61.

ylen=length(y)
Powersum=0;
for i=1:ylen
    Powersum=Powersum+y(i);
end

Powersum
Powersum=Powersum/ylen

outvector=y-Powersum;
```

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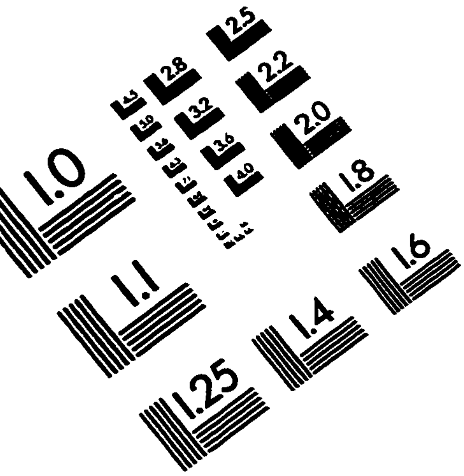
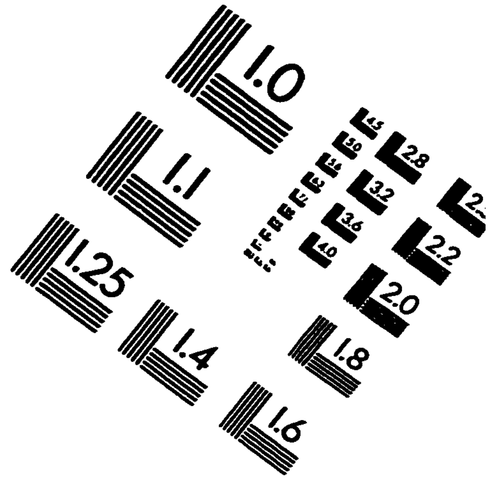
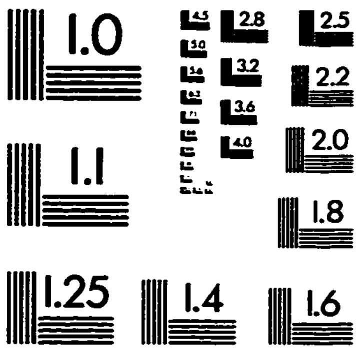
IMAGE EVALUATION TEST TARGET (QA-3)

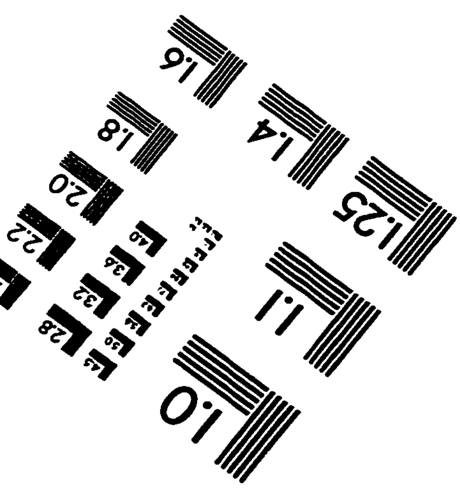


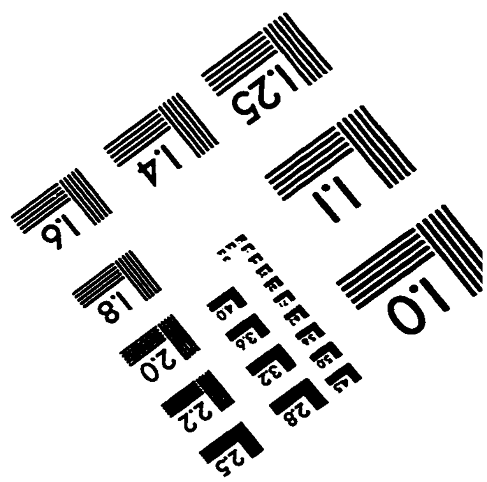




150mm

6″





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