2006

Development of a Software Application to Extract the Features of Normal Respiratory Sounds from the Lungs and the Trachea

Ranjani Sabarinathan
Virginia Commonwealth University

Development of a Software Application to Extract the Features of Normal Respiratory Sounds from the Lungs and the Trachea

A thesis submitted in partial fulfillment of the requirements for the degree of Master of Science at Virginia Commonwealth University.

By

Ranjani Sabarinathan

B.S. Electrical Engineering
Queen's University
Kingston, Ontario, Canada
May 2004

Director: Ding-Yu Fei, PhD.
Associate Professor
Department of Biomedical Engineering

Virginia Commonwealth University
School of Engineering
Richmond, Virginia
December, 2006

To

My family

ACKNOWLEDGEMENT

Firstly, I thank my advisor, Dr. Ding-Yu Fei for the great help and guidance he provided throughout my research. I want to give my thanks to my colleagues at the Bioinstrumentation and Medical Imaging Laboratory for their support in helping me complete this thesis.

I would like to thank Dr. Ronald Merrell, Dr. Azhar Rafiq and other faculty members in Medical Informatics and Technology Applications Consortium (MITAC), Department of Surgery for providing guidance and technical support to this research project. I also thank Dr. Martin Lenhardt for serving as my committee member and providing suggestions to my thesis.

I also give my thanks to the faculty and staff of Biomedical Engineering Department for giving me the opportunity to pursue my Master's Degree at Virginia Commonwealth University.

Finally, I want to extend my appreciation towards my family and friends for their support and encouragement during my research.

Table of Contents

List of Figures viii

ABSTRACT x

Chapter 1 1

Introduction 1

  1.1 Background and Purpose of Study 1

  1.2 Anatomy and Physiology of the Respiratory Tract 4

  1.3 Organization of the Thesis 6

Chapter 2 8

Literature Review 8

  2.1 Categories of Respiratory Sounds 8

    2.1.1 Normal Lung Sounds 8

    2.1.2 Normal Tracheal Sounds 9

    2.1.3 Adventious Breath Sounds 10

      2.1.3.1 Wheezes 10

      2.1.3.2 Crackles 10

      2.1.3.3 Stridor 11

      2.1.3.4 Squawks 11

  2.2 Devices used in the Auscultation of Respiratory Sounds 12

    2.2.1 Stethoscope 12

    2.2.2 Sensors 13

2.2.2.1 Contact Sensors 13

2.2.2.2 Air-coupled sensors 13

2.3 Signal Acquisition 14

2.4 Graphical Representation in the Time Domain 14

2.5 Noise and Interference 15

2.6 Digitization of Data for Recording Respiratory Sounds 16

2.7 Envelope Extraction 17

2.7.1 Hilbert Transform 17

2.8 Filtering Methods 18

2.8.1 Low Pass Filtering 18

2.8.2 High Pass Filtering 18

2.8.3 Band pass Filtering 18

2.9 Spectral Analysis 19

2.9.1 Windowing 21

2.9.1.1 Hanning 22

2.9.1.2 Hamming 22

2.10 Positioning for Respiratory Sound Recording 22

2.10.1 Body Position 22

2.10.2 Microphone Location 23

Chapter 3 24

Methods 24

3.1 LabView Application 24

3.2 Signal Processing 29

3.3 Experimental Setup 30

3.4 Experimental Procedure 33

Chapter 4 34

Results and Discussion 34

4.1 Time Domain Results 34

4.1.1 Left and Right Lung Sounds 35

4.1.2 Tracheal Sound 45

4.2.1 Lung Sound Power Spectrum 50

4.2.3 Trachea 57

4.3 Analysis 61

4.3.1 Time Domain 61

4.3.2 Power Spectrum 64

4.4 Statistical Analysis 67

4.4.1 Average Amplitudes in the Time Domain 68

4.4.1.1 Left-Right Lung 68

4.4.1.2 Left Lung-Trachea 68

4.4.1.3 Right Lung- Trachea 69

4.4.2 Power Spectrum of the Whole Signal 69

4.4.2.1 Left-Right Lung 70

4.4.2.2 Left Lung-Trachea 70

4.4.2.3 Right Lung- Trachea 70

4.4.3 Power Spectrum of the Signal in the Inspiration Phase 71
    4.4.3.1 Left-Right Lung 71
    4.4.3.2 Left Lung-Trachea 72
    4.4.3.3 Right Lung- Trachea 72
Chapter 5 73
Conclusions and Future Work 73
5.1 Conclusions 73
5.2 Suggestions for Future Work 75
5.2.1 Adventitious and Normal Breath Sounds 75
5.2.2 Simultaneous Recording of the Respiratory Sounds 76
5.2.3 Supine Position 76
REFERENCES 78
VITA 86

List of Figures

Figure number……………………………………………………………..Page

1.1 Image of the respiratory tract……………………………………………...5

3.1 Front panel of the LabView program………………………………………..27

3.2 An example of a block diagram in LabView……………………………...28

3.3 Schematic of data flow in the program……………………………………...29

3.4 Equipment setup of the experiment…………………………………………31

3.5 Electronic stethoscope……………………………………………………….32

3.6 A/D converter………………………………………………………………..33

4.1 Graphical representation of the signal from the left lung of subject R…...36

4.2 Graphical representation of the signal from the right lung of subject R….37

4.3 Graphical representation of one breath cycle from subject R's left lung…38

4.4 Graphical representation of one breath cycle from subject R's right lung..39

4.5 Inspiration phase of the sound signal from subject R's left lung…………..40

4.6 Inspiration phase of the sound signal from subject R's right lung…………40

4.7 Expiration phase of the sound signal from subject R's left lung…………...41

4.8 Expiration phase of the sound signal from subject R's right lung…………41

4.9 Signal envelope from subject R's left lung………………………………...42

4.10 Signal envelope from subject R's right lung……………………………...43

4.11 One breath cycle taken from subject R's left lung signal envelope………45

4.12 One breath cycle taken from subject R's right lung signal envelope........45

4.13 Graphical representation of signal taken from the trachea of subject R.....45

4.14 One breath cycle of subject R's tracheal signal...............................46

4.15 Inspiration phase of subject R's tracheal signal...............................47

4.16 Expiration phase of subject R's tracheal signal...............................48

4.17 Signal envelope from subject R's trachea...................................49

4.18 Tracheal sound signal envelope for one breath cycle from subject R.......50

4.19 Power spectrum of subject R's left lung signal...............................51

4.20 Power spectrum of subject R's right lung signal...............................52

4.21 Power spectrum of the inspiration phase of R's left lung signal.............53

4.22 Power spectrum of the expiration phase of R's left lung signal..............54

4.23 Power spectrum of the inspiration phase of R's right lung signal........... 56

4.24 Power spectrum of the expiration phase of R's right lung signal............56

4.25 Power spectrum of subject R's tracheal signal...............................58

4.26 Power spectrum of the inspiration phase of R's tracheal signal..............60

4.27 Power spectrum of the expiration phase of R's tracheal signal..............60

4.28 A bar chart displaying the average amplitudes of the lung and tracheal sound signals from all subjects...62

4.29 A bar chart showing the power distribution ratios from the whole signals of the left and right Lungs and the trachea..65

4.30 A bar chart showing the comparison of power distribution ratios from the inspiration phase of the left and right lungs and the trachea................66

ABSTRACT

DEVELOPMENT OF A SOFTWARE APPLICATION TO EXTRACT THE FEATURES OF NORMAL RESPIRATORY SOUNDS FROM THE LUNGS AND THE TRACHEA

By

Ranjani Sabarinathan

A thesis submitted in partial fulfillment of the requirements for the degree of Master of Science at Virginia Commonwealth University.

Virginia Commonwealth University, 2006.

Major Director: Ding-Yu Fei, Ph.D.
Associate Professor
Department of Biomedical Engineering

Auscultation has been widely regarded as one of the most important non-invasive diagnostic tools for clinical diagnosis of the respiratory tract. The purpose of this thesis was to develop a software application, capable of extracting the key features of respiratory sound signals from the lungs and trachea of healthy persons. The efficacy of the program was evaluated by the verification of the important features of the sound signals from the left and right lungs and the trachea such as 1) right and left lung symmetry and 2) dissimilarity between the trachea and both lungs.

The program was developed in LabView and was designed to capture the respiratory sound signals from the lungs and the trachea in real-time and process them in the time and frequency domains for further analysis. The features compared were 1) signal amplitude in the time domain and 2) power spectra in the frequency domain. Results of the study had shown that the program had been able to verify that 1) the key features of the breath sound signals from the left and right lungs were similar and 2) the features of the signals from the trachea and both lungs were different.

Chapter 1

Introduction

1.1 Background and Purpose of Study

Auscultation is one of the most important non-invasive and simple diagnostic tools for detecting disorders in the respiratory tract like lung diseases (Gross et. al., 2000). It is defined as the 'act of listening for sounds within the body, mainly for ascertaining the condition of the lungs, heart and other organs (Jones et. al., 1999; Dorland, 1981). Diseases such as asthma, tuberculosis can be identified with this method through the analysis of lung and tracheal sounds.

Auscultation has been used for over two centuries as part of clinical evaluations of the cardio respiratory system. An experienced physician would be able to identify respiratory conditions just by listening to the sounds through a stethoscope. Research on the diagnosis of respiratory pulmonary conditions like bronchitis, sleep apnea, asthma has established the utility of the stethoscope's acoustic signal in common day to day practice. However, despite their effectiveness, these instruments only provide a limited and subjective perception of the respiratory sounds. The drawbacks of using stethoscopes and listening to the sounds using the human ear are a) their inability to provide an objective study of the respiratory sounds detected, b) their lack of sufficient sensitivity and (c) the existence of the imperfect system of nomenclature (Dalmay et. al., 1995).

In the last few decades, improvements in electronic recording and the development of computer-based methods have made quantitative studies of lung and tracheal sound signals possible as well as overcome many limitations of human ear subjective auscultation. Modern digital processing techniques, along with advancements in computer analysis, have become an established research method for the investigation of respiratory sounds. However, there has been no standardized method for the nomenclature of lung sound signals (Pasterkamp et. al., 1997). There is also considerable variation in which the sounds are processed and analyzed, despite similarities in basic methodology (Earis et. al., 2000).

The two most common techniques used to understand the attributes of breath sounds and to increase the usefulness of auscultation in clinical diagnosis are the time and frequency domain analysis. From the review of literature, it can be observed that the use of spectral analysis is more prevalent than time domain analysis. The few studies reported in the literature for the time domain analysis have used recording of sounds in conventional or expanded time scales or direct listening of amplified sounds

(Charbonneau et. al, 2000; Gavriely et. al., 1981). However, these methods were found to be limited in their capacity and objectivity (Gavriely et. al, 1981).

The goal of this thesis is to develop a software application that can extract the key features of respiratory sound signals from the lungs and the trachea. For left and right lung sounds, the inspiratory phase of the signal will be mainly studied. The amplitude of the signal and the power distribution from the frequency spectrum would be used as a key parameter to analyze the features of the sound in the time and frequency domains respectively.

The efficacy of the program will be evaluated by performing the verification of the following commonly accepted features of sound signals from lungs and trachea namely;

(a) **Right and left lung sound signal symmetry**: It is expected and understood that the significant features of the sound signal recorded from the left and right side of the chest of the same person would be similar. In the time domain, the amplitude of the sound signal and the power spectra in the frequency domain for the left and right lungs are expected to be in similar ranges.

(b) **Tracheal and lung signal dissimilarity**: In the time domain, it is expected that the left and right lung sound signals would have significantly different amplitudes from the signal corresponding to the trachea. The tracheal sound's power spectrum will be different from the spectra of both lungs.

In the rest of this chapter a brief review of the anatomy and physiology of the respiratory tract would be provided followed by an outline of the rest of the thesis.

1.2 Anatomy and Physiology of the Respiratory Tract

The respiratory system's function involves gas exchange by delivering oxygen to the circulatory system for transport to all parts of the body and removing carbon dioxide, thereby preventing the build up of waste products in the tissues. Its organs are responsible for regulating the balance of acid and base in tissues, which is crucial for the normal functioning of cells. The respiratory system protects the body against disease-causing organisms and toxic substances inhaled with air. It contains the cells capable of detecting smell and producing sounds for speech.

The respiratory tract consists of the vocal tract and the subglottal airways, which have been studied extensively for acoustic investigations (Pasterkamp et. al., 1997). The branching airways in the thorax were made to assess the structural determinants of sound reflection and transmission measurements (Pasterkamp et. al., 1997; Van Den Berg, 1960). The upper respiratory tract consists of the nose, larynx, trachea, nasal cavity and extends from the nose to the throat. The lower respiratory tract is made up of lungs, alveoli, and the bronchi and they cover the area between the throat and the diaphragm.

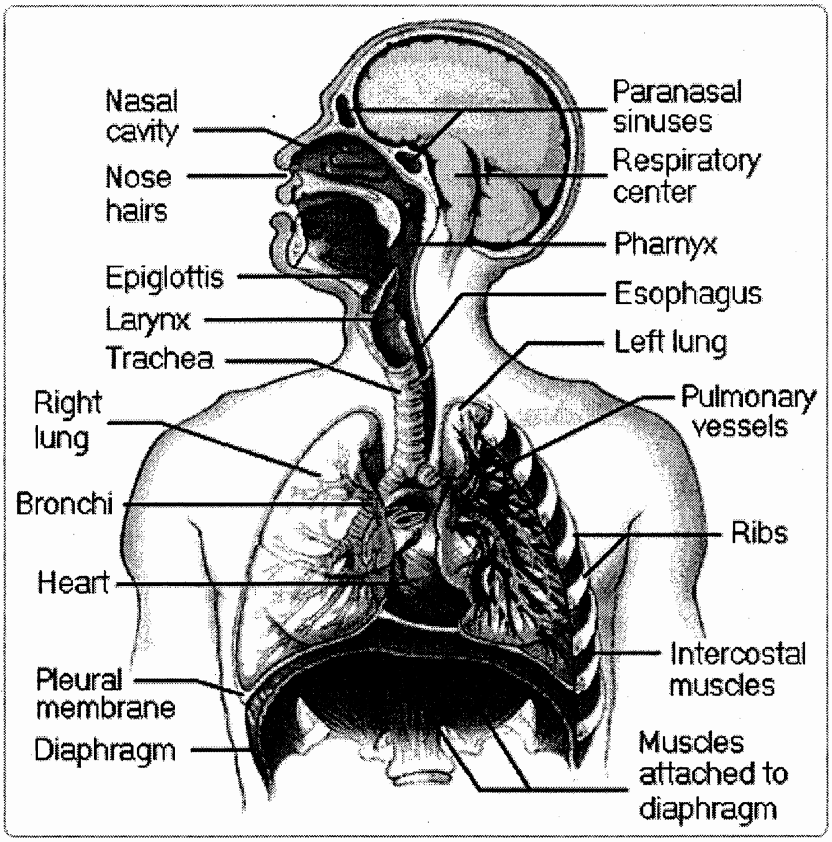

Figure 1.1: Image of the Respiratory System (AMA , http://www.ama-assn.org/ama/pub/category/7166.html)

The trachea filters the air and extends from the neck into the thorax, where it is divided into left and right bronchi. In the lungs, the bronchi divide themselves further into bronchioles. Gas exchange takes place in the lungs, where they pick up oxygen and get rid of carbon dioxide. The lungs are enveloped in a membrane known as the pleura and are separated from each other by the mediastinum, which contains the heart, oesophagus, trachea and thymus. The bronchioles end in air sacs called the alveoli. On the surface of each alveolus, there is a network of capillaries that carries blood from other parts of the body.

Breath sounds originate predominately in the larger airways and when they are heard on the chest wall, information about their origin and the way these signals have been transmitted through the lungs are studied (Earis et. al., 2000, Kraman et. al., 1990). Changes in lung structure, occurring in diseases, affect the amplitude and timing of sound transmission from the airways to the chest (Pasterkamp et. al., 1997).

1.3 Organization of the Thesis

In this thesis, the focus was on designing a software application for extracting the features of breath sound signals from the lungs and the trachea. The program was developed for the purposes of obtaining the signals in real-time and processing them in the time and frequency domains for further analysis. This section provides an overview of the following chapters.

Chapter 2 gives a comprehensive, literature review of the categories of respiratory sounds and the equipment used to record them. It also reviews previous studies in auscultation such as the research done on spectral analysis, body positioning and microphone locations that would make suitable sites for the sensors to record breath sound signals.

Chapter 3 covers a detailed description of the methodology, designed and used for this study. The software application for the study, written in the LabView environment, will be explained in detail as well as other information such as the set up of the experiment and the procedure.

Chapter 4 provides a detailed explanation of the results obtained from this study, while chapter 5 summarizes all the findings of this thesis and discusses suggestions for prospective work in this area.

Chapter 2

Literature Review

2.1 Categories of Respiratory Sounds

2.1.1 Normal Lung Sounds

The breathing sound heard from the chest of a healthy person is the normal lung sound or vesicular sound. Normally, sound intensity is higher during inspiration than expiration (Gavriely et. al., 1981). The most commonly used frequency bandwidth for recording lung sounds ranges from 60-100 Hz to 2 KHz (Vannuccini et. al., 2000). Heart sounds are noise for respiratory signals, particularly at low and medium frequencies (Pourazad et. al., 2005). Heart and muscle sounds overlap at frequencies below 100 Hz (Gross et. al., 2000, Charbonneau et. al., 1982), a factor that should be taken into consideration when making lung sound measurements.

Normal lung sounds peaks in frequencies below 100 Hz and drops off sharply between 100 and 200 Hz, but can still be detected beyond 1000 Hz with sensitive microphones in a quiet room (Pasterkamp et. al., 1997, Gavriely et. al., 1995). The power spectrum is not musical and is devoid of discrete peaks (Pasterkamp et. al., 1997).

2.1.2 Normal Tracheal Sounds

The breathing sound that can be listened from the trachea of a healthy person is called the normal tracheal sound. Tracheal sounds are acquired on a wider frequency range than the normal lung sounds (Pasterkamp et. al., 1997) and are large in amplitude (Golabbakhsh et. al., 2005). The frequency bandwidth generally used for recording tracheal sounds ranges from 60-100 Hz to 4 KHz (Vannuccini et. al., 2000).

Tracheal sounds are higher in intensity than the sounds captured from the chest (Earis et. al., 2000). The power spectra of tracheal signals contain peaks and troughs that are related to airway dimensions and are dependent on gas density (Pasterkamp et. al., 1997). Tracheal sound measurement has been able to provide valuable information about respiratory health, such as apnea monitoring, which has been successfully applied in both adults and children (Pasterkamp et. al., 1997, Krumpe et. al., 1990, Beckerman et. al., 1982).

2.1.3 Adventious Breath Sounds

Adventious breath sounds indicate the presence of a pulmonary disorder. The types of adventious respiratory sounds are continuous (wheezes) and discontinuous (crackles) as well as short sounds having features between continuous and discontinuous sounds (like squawks) (Sovijarvi et. al, 2000). The most commonly used bandwidth for recording abnormal sounds ranges from 60-100 Hz to 6 KHz (Vannuccini et. al., 1997).

2.1.3.1 Wheezes

Wheezes are pseudoperiodic signals that are characterized by their pitch and duration and their waveform resembles a sinusoidal sound (Charbonneau et. al., 2000). The duration of the wheeze is longer than 250 ms (Sourjavi et al., 2000). They can be high pitched or low pitched (sometimes referred to as rhonchi). Wheezes are common clinical signs for patients with asthma and other obstructive airways diseases.

2.1.3.2 Crackles

Crackles are discontinuous and explosive lung sounds. They are frequent in cardiopulmonary diseases (Piirila et. al., 1995). The two types of crackles are coarse and fine. Coarse crackles are loud, low pitched with a high amplitude and long duration, whereas fine crackles have a high pitch, low amplitude and a short duration (Sourjavi et al., 2000).

2.1.3.3 Stridor

Stridors are very loud and harsh low frequency wheeze, caused by obstructions in the larynx or trachea (Sourjavi et al., 2000). They appear in respiratory problems like whooping cough.

2.1.3.4 Squawks

Squawks are short, musical and inspiratory sounds, preceded by a crackle. They are occasionally found in patients with interstitial lung diseases. Their duration does not exceed 400 ms (Sourjavi et. al., 2000).

2.2 Devices used in the Auscultation of Respiratory Sounds

2.2.1 Stethoscope

A stethoscope is an acoustic medical device used by physicians to listen to respiratory sounds. It was first invented in France in1816 by Rene Laënnec, who published his treatise on auscultation in 1819, describing acoustic events generated by ventilation of the lungs and systematically correlating them with anatomical and pathological findings after autopsy (Dalmay et. al., 1995). Stethoscopes are found in two types: acoustic and electronic.

In this study, the electronic stethoscope is used to record the breath sounds. It can selectively amplify or attenuate sounds within the spectrum of clinical interest (Sovijarvi et. al., 2000). The electronic stethoscope converts acoustic sound waves into electrical signals that are processed for optimal listening.

2.2.2 Sensors

Sensors enable the vibrations emanating from the chest wall to be converted to electrical signals. Sounds acquired from the respiratory system are captured using a microphone or a contact sensor at the chest, mouth or neck.

2.2.2.1 Contact Sensors

A contact sensor is used when directly recording the sounds from the chest. For lung sound recording, the accelerometer and piezoelectric sensors are quite popular.

2.2.2.2 Air-coupled sensors

The air-coupled sensor involves recording the movement of the diaphragm, exposed to the pressure induced by the chest-wall movement. The Electret microphone with a coupling chamber is the transducer commonly used for lung sound recording and is similar to a stethoscope bell (Pasterkamp et. al., 1997).

2.3 Signal Acquisition

The study of respiratory sounds using more advanced digital signal processing techniques is now widely established amongst research scientists. Although similarities in the basic methodology are found, there are variations in the approach taken to process and analyze breath sound signals (Earis et. al., 2000).

The analogue processing system comprises of a sensor, amplifier and filters, which are responsible for conditioning the signal before A/D conversion and generally a combination of a low pass and high pass filters in cascade is used (Vannucini et. al., 2000). The respiratory sound signal is displayed as a function of time on a linear scale. Only one channel is usually used but many researchers have used two and more channels (Earis et. al., 2000).

2.4 Graphical Representation in the Time Domain

In a technique called TEWA (Time Expanded Waveform Analysis), normal breath sounds were described to have an irregular shape with no repetitive pattern or sudden rapid deflections (Charbonneau et. al., 2000). Wheezes and stridors have periodic waveforms, which is either sinusoidal or more complex, while crackles have sudden short deflections, followed by deflections with larger amplitudes (Charbonneau et. al., 2000).

2.5 Noise and Interference

The recording of respiratory sounds is affected by noise interference such as acoustic environmental noise like those generated by electronic devices, slamming doors, hard disks to name a few and non-respiratory sounds like heart murmurs, muscle sounds, bowel sounds, swallowing, speech, chest motion etc (Rossi et al., 2000). Heart and muscle sounds are reduced using filtering techniques (Vannuccini et. al., 2000).

According to Rossi et. al. (2000), most of the environmental noise can be avoided by the use of a soundproof room, where the respiratory sound recording takes place. Another way to reduce acoustic noise is to shield the sensors, where different types of sensors are known to have different sensitivities to the environmental noise reaching them directly (Rossi et. al, 2000).

2.6 Digitization of Data for Recording Respiratory Sounds

Digitization enables a high-quality representation of the sound to be obtained, which is more accurate and permanent than analogue recording (Cheetham et. al., 2000). It makes the signal data more accessible and compatible with other data.

In respiratory sound measurements, a wide range of different sampling rates, ranging from 4 KHz to 22.05 KHz are commonly used. The specifications of sampling frequency depend on the type of signal studied and the processing techniques used. Cheetham et. al. (2000) recommended the sampling frequency for a sound channel to be 11.025 KHz, even though values at 5.512 KHz, 22.05 KHz and 44.1 KHz could also be used, because the sampling frequency of 11.025 KHz allowed the adoption of a relaxed analogue, anti-aliasing filtering specification and would not place too much stress on the storage capacity of the channel.

Cheetham et. al. (2000) also recommended using an anti-aliasing analogue filter, which is a 4th order Butterworth low-pass filter with 3 dB cut off frequency at 2 KHz if the sampling frequency was 11.025 KHz.

2.7 Envelope Extraction

The signal envelope is the amplitude of the respiratory signal trace and the information contained in the amplitude is used to monitor the cardio-respiratory nervous system (Rezek et. al., 1998). The traditional method of extracting an envelope is done by the tracing of peaks in the signal and was found to be quite tedious, computationally expensive and very noise sensitive (Rezek et. al., 1998). Some known ways of extracting the envelope are the use of the Hilbert Transform, the amplitude modulation process and the homomorphic filtering technique.

2.7.1 Hilbert Transform

The Hilbert transform is used to construct a complex signal from a real valued signal without major changes to the signal characteristics and the implementation of this transform is based on the Fast Fourier Transform (Rezek et. al., 1998). The transform is named after the mathematician, David Hilbert. The Hilbert Transform of the signal is defined as the signal obtained by the convolution of the original signal with 1/ (πt). The Hilbert transform introduces a 90 degrees phase shift to every frequency of the real signal to obtain the imaginary part of the function (Rezek et. al., 1998).

2.8 Filtering Methods

2.8.1 Low Pass Filtering

Low pass filtering (LPF) is used to eliminate aliasing of the digitized signal, where frequencies above half the sampling frequency are removed (Vannuccini et.al., 2000). The cut off frequency must be above the upper frequency. Vannuccini et. al. (2000) recommended the LPF should have a -3 dB cut off at the upper frequency and provide 24 dB of attenuation at that frequency.

2.8.2 High Pass Filtering

Some studies generally use a high pass filter with a cut off frequency of 60 Hz to reduce low frequency distortion to the respiratory sound signal (Vannuccini et. al., 2000). High pass filter helps to reduce heart, muscle sounds and contact noises.

2.8.3 Band pass Filtering

A filter responsible for letting components within a specific band of frequencies to pass, while substantially attenuating or removing all lower and higher frequencies.

2.9 Spectral Analysis

Spectral Analysis of respiratory sounds involves the study of the signals in the frequency domain, where they are represented by their power spectrums. Frequency domain analysis has been conducted by many researchers, using different methods. The scale for a power spectrum is either linear or logarithmic for the intensity (amplitude or power distribution), when plotted against the frequency. FFT (Fast Fourier Transform) or DFT (Discrete Fourier Transform) are generally applied to the sound to obtain the frequency spectrum with signal block lengths of 256, 512 or 1024 samples commonly used (Earis et. al., 2000).

A study made on the averaged spectra computed on tracheal sounds have shown that the log amplitude response curve remained flat in the range 75-900 Hz, before falling away at higher frequencies (Charbonneau et. al., 2000, Pasterkamp et. al., 1989, Gavrieli et. al., 1981). Other authors had acquired spectra of lung sounds on a linear plot with maximum amplitudes at the 140-200 Hz range, which was followed by exponential decay to insignificant levels at approximately 400 Hz (Charbonneau et. al., 2000). Respiratory signals are not usually studied at a frequency range less than 50-60 Hz due to the interference caused by heart and muscle sounds. Normal Respiratory sounds contain components among which the most significant have a frequency of 50-1200 Hz (Charbonneau et. al., 2000).

Sound spectra are associated to respiratory airflow rate, where the sound intensity increases with air flow (Charbonneau et. al., 2000, Charbonneau et. al., 1987). The relationship between the air flow and sound depend on the subject and upper airways configuration. There is a strong correlation between airflow, sound intensity and frequency mean (Earis et. al., 2000, Charbonneau et. al., 1987).

There are few methods used to characterize a frequency spectrum. A very common way is to divide the spectrum into parts, where they form the percentiles of the total spectrum energy. For example, the quartiles divide the spectrum into four parts, where they are used to characterize major changes in breath sounds (Charbonneau et. al., 2000). Murphy et al. (1973) in their study on chest auscultation in the diagnosis of pulmonary asbestosis were the first to use FFT for spectral analysis of breath sounds of asbestos workers (Gavriely et. al., 1981, Murphy et. al., 1973).

The power spectra of normal lung sounds had been shown to have small variations between subjects, which Charbonneau et. al. (2000) believed to be due to the interest in low frequencies (Charbonneau et al., 2000, Ployosongsang et. al., 1991). Changes in lung sound spectra can be used to detect the early stages of airways disease (Earis et. al., 2000, Gavrieli et. al., 1994). The amplitudes at low frequencies to the sites overlying the right lung are found to be greater than that measured at the corresponding locations over the left lung, indicating the preferential coupling of sound to the right lung, due to the massive mediastinum, adjacent to the left side of major airways

(Pasterkamp et. al., 1997). Gavriely et al. (1981) in their study on the spectral characteristics of normal breath sounds found that the power of the signal from normal vesicular sounds decreased exponentially as frequency increased.

The spectral shape of tracheal sounds is greatly variable between persons but reproducible within the same person, thus reflecting the strong influence of individual anatomy (Pasterkamp et. al., 1997, Pasterkamp et. al., 1996). High tracheal frequencies influence the frequency spectrum, where the main energy of the sounds extend up to 850-1000 Hz, with a sharp decrease in power above that frequency range (Sovijarvi et. al., 2000, Pasterkamp et. al., 1984). Gavriely et al. (1981) discovered that breath sounds picked up over the trachea have a broad power spectrum with a sharp decrease in power at the cut off frequency, varying between 850 Hz and 1600 Hz (Gavriely et. al., 1981).

Some of the frequency domain analysis studies on breath sounds were conducted by extracting the inspiration and expiration components separately and comparing them.

2.9.1 Windowing

Windowing is a method commonly used in spectral analysis, whose functions help improve spectral estimates. When a signal is multiplied by a window function, the product is zero-valued outside the interval. The two windowing functions widely used in breath sound analysis are the Hanning and the Hamming windows.

2.9.1.1 Hanning

The Hanning window, also referred to as the Hann window, was named after the inventor Von Hann. Its shape is one cosine wave cycle with one added to it so as to make it positive all the time. The Hanning window forces both ends of the signal to become zeroes.

2.9.1.2 Hamming

The Hamming window is similar to a Hanning window, except that it has a disadvantage of being discontinuous at the edges, leading to signal contamination.

2.10 Positioning for Respiratory Sound Recording

2.10.1 Body Position

A study made by Jones et al. discussed the effect of positioning on lung sound intensities in persons without pulmonary disorders. In a sitting position, the inspiratory lung sound intensity was found to be higher over the left chest wall than the right, but the intensity during expiration was the same (Jones et. al., 1999). When taking a side lying position, Jones et. al. (1999) reported that the sound intensity recorded over the dependent lung was louder than that of the nondependent lung.

The body position for short-term recording of breath sounds is the sitting position, whereas for long-term recording, the posture commonly used is supine (Rossi et. al, 2000).

2.10.2 Microphone Location

In previous studies, researchers used different sites for microphone placement and had come up with some conclusions. The lung sound intensity, like amplitude or power distribution in the frequency spectrum, was found to vary at different locations. Respiratory sounds are not uniform over the lungs, thus the sound intensity differs in the location where they are recorded and also vary with the ventilatory cycle (Sovijarvi et. al., 2000). Investigators have deemed the need to identify sites that have proved to be the most suitable ones for listening to respiratory sounds.

Chapter 3

Methods

3.1 LabView Application

LabView was the programming platform used to develop the software application for the extraction and processing of the lung and tracheal sound signals. The application contains a user friendly graphical interface, developed to display the respiratory sound signals in both the time and frequency domains.

LabVIEW stands for Laboratory Virtual Instrument Engineering Workbench and is created by National Instruments. It is a programming environment that differs from traditional programming languages like C or Java by utilizing a powerful graphical programming language, known as G, instead of lines of text. A LabVIEW program is called a virtual instrument (VI), whose appearance and operation resemble actual physical systems or instruments. However, behind the scenes, they are analogous to main programs, functions, and subroutines from popular programming languages like C or Basic (Bishop, 2006).

A VI contains three main parts: front panel, block diagram and the icon. The front panel is the user interface of a VI that is interactive. The front panel designed for this study shows three graphical windows with each one having a particular function and the start and stop buttons (Figure 3.1). The raw respiratory signal acquired from the device is displayed on the first window in time domain. The second window presents the signal envelope extracted from the raw data by Hilbert Transform and the third window shows the power spectrum. When the start button is pressed, acquisition of the breath sound signal begins. Data collection ends when the stop button is clicked.

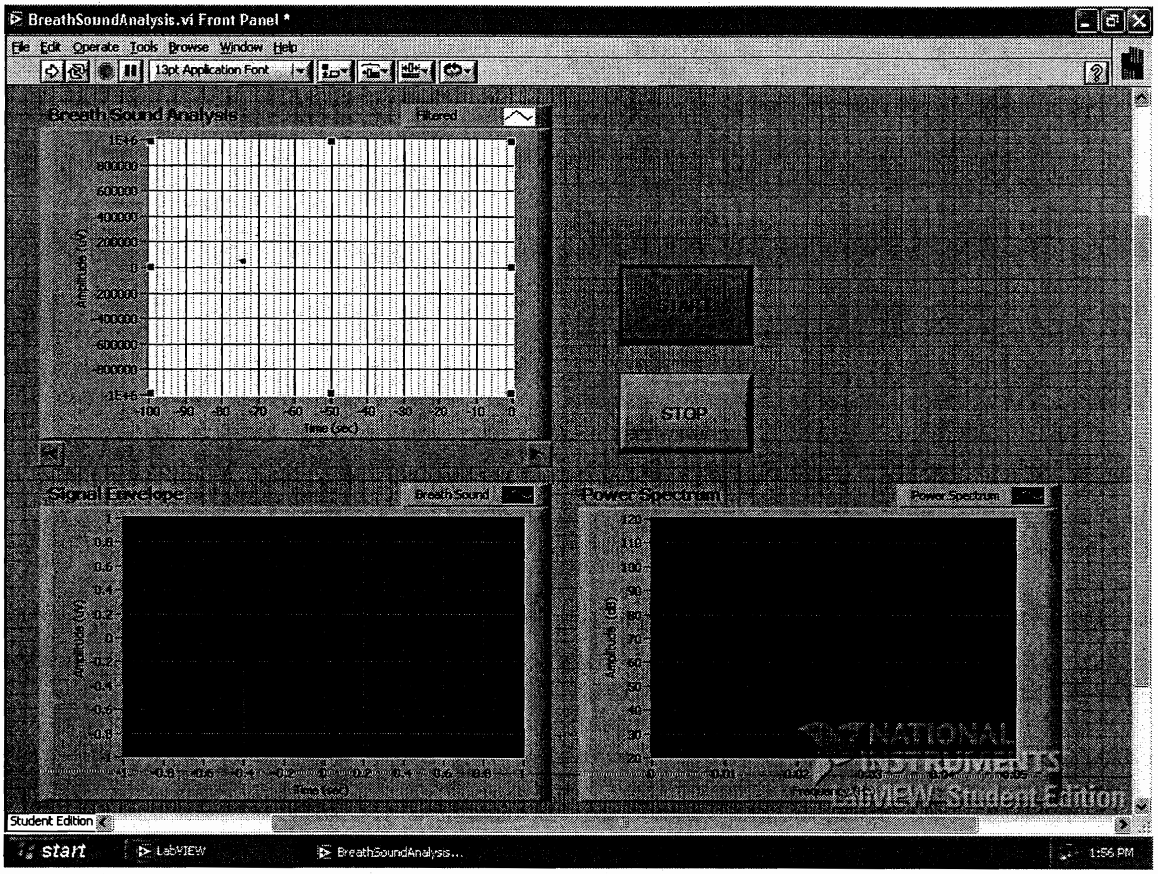

Figure 3.1: Front panel of the LabView program

The block diagram is the source code, which is made up of graphical icons and wires (Figure 3.2). The icons are the pictorial representation of VIs, program control structures and other built-in functions. The wires connect the icons to allow data flow throughout the program. A VI used by another VI is known as a SubVI and functions like a subroutine.

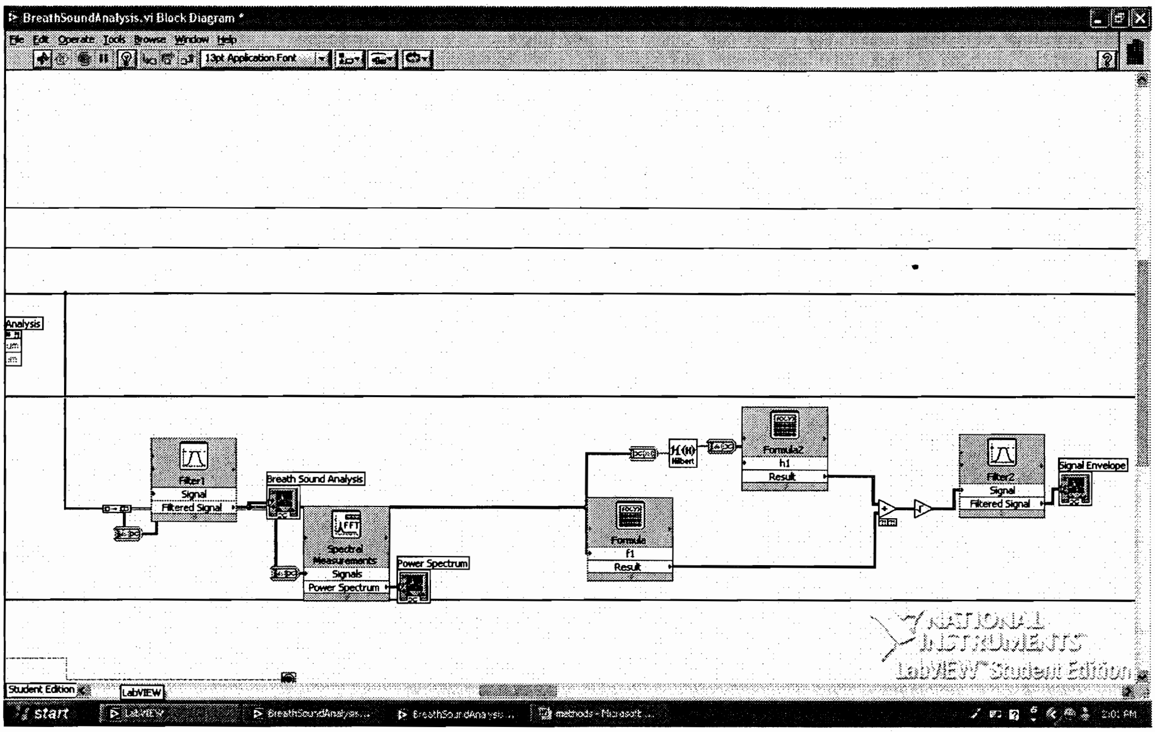

Figure 3.2: An example of a block diagram in LabView

Figure 3.3 shows the set up of the data flow through the program. When the program begins running, a subVI called DDLStartLogging is responsible for setting up parameters such as the gain, channel and the sample interval, which is set to 91 μs. The data travels from DDLStartLogging VI to the DDLGetLoggedData VI, the subVI used to log all data acquired from the device. From the DDLGetLoggedData VI, the raw signal flows to the built-in array function icons, which arranges the data into an array and then sends it to the Filter Express VI and the Spectral measurements VI.

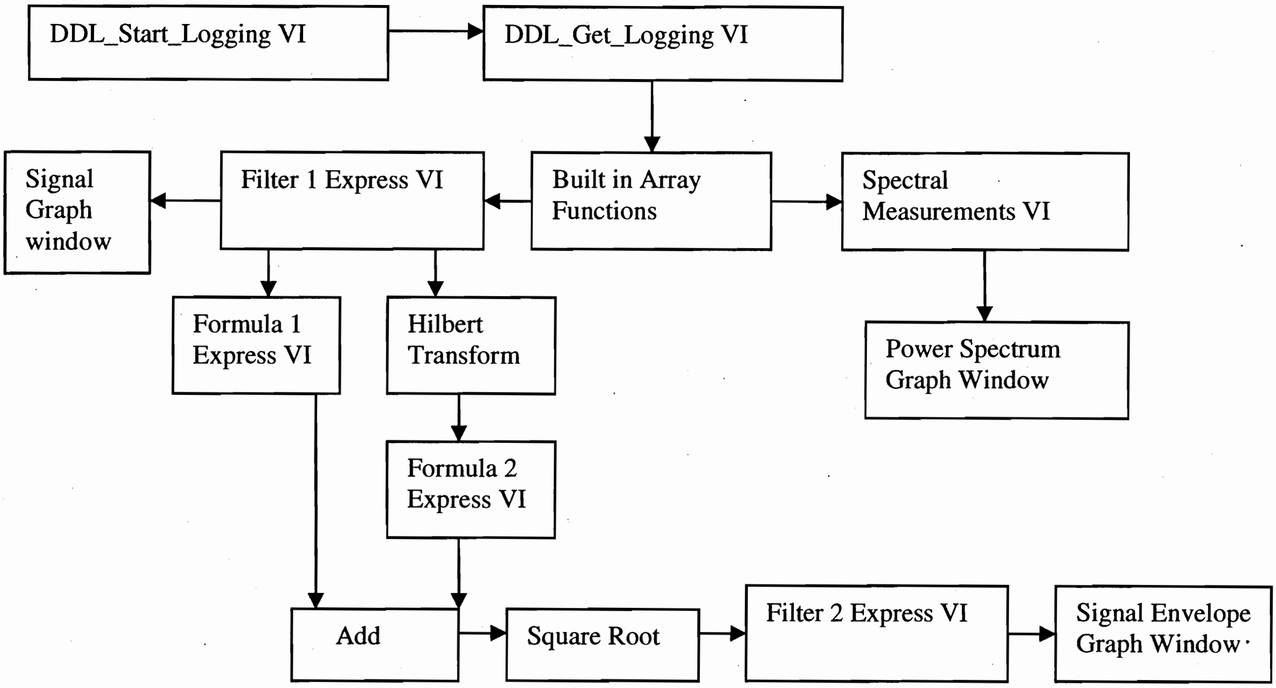

Figure 3.3: Schematic of data flow in the program

The filter express VI is responsible for filtering out the noise components from the signal. The spectral measurements express VI sets up the parameters for the power spectrum and is connected to a graph icon, a representation of the window displaying the power spectrum. The connections from the filter express VI branches off in three different directions: graph icon representing the window displaying the filtered signal in time domain, Hilbert transform icon and a formula express VI, set to double the y-component of the filtered data to improve sensitivity. The Hilbert transform icon is connected to another formula express VI, configured to increase the Hilbert transform of the signal by a factor of two to improve sensitivity. The formula express VIs are connected to the addition icon, which sums up both sets of data. The addition icon is wired to the square root icon, where the square root of the combined signal took place. The square root icon then led to another filter express VI. The second express VI is connected to the graph icon, a pictorial representation of the window showing the signal envelope.

3.2 Signal Processing

A sampling rate of 11.025 KHz of the A/D converter was used, which led to the division of the signal into 91 μs segments at a conversion resolution of 12 bits. This particular frequency was chosen, based on a recommendation made by Cheetham et al. (2000), as discussed in the previous section. The signal was filtered by an anti-aliasing low pass filter, which effectively removed high frequencies above the cut off frequency. The attributes of the filter were taken from LabView's user libraries. A 6th order low pass Butterworth filter was chosen for the acquisition of the lung sounds and a 4th order low pass Chebyschev filter for tracheal sounds, in order to remove the noise and enable the representation of a clearer signal.

The Hilbert's Transform was responsible for extracting the envelope from the signal. The envelope then got filtered with a 4th order low pass Butterworth filter to remove any background noise, thereby enabling the display of a smooth curve.

FFT was used to obtain the power spectra. Each segment of the digital data was windowed with a Hanning function. Variables recorded from the power spectrum for analysis were the peak intensity (total power of the spectral data), frequency at the peak intensity and in addition for tracheal sounds, all major peaks.

3.3 Experimental Setup

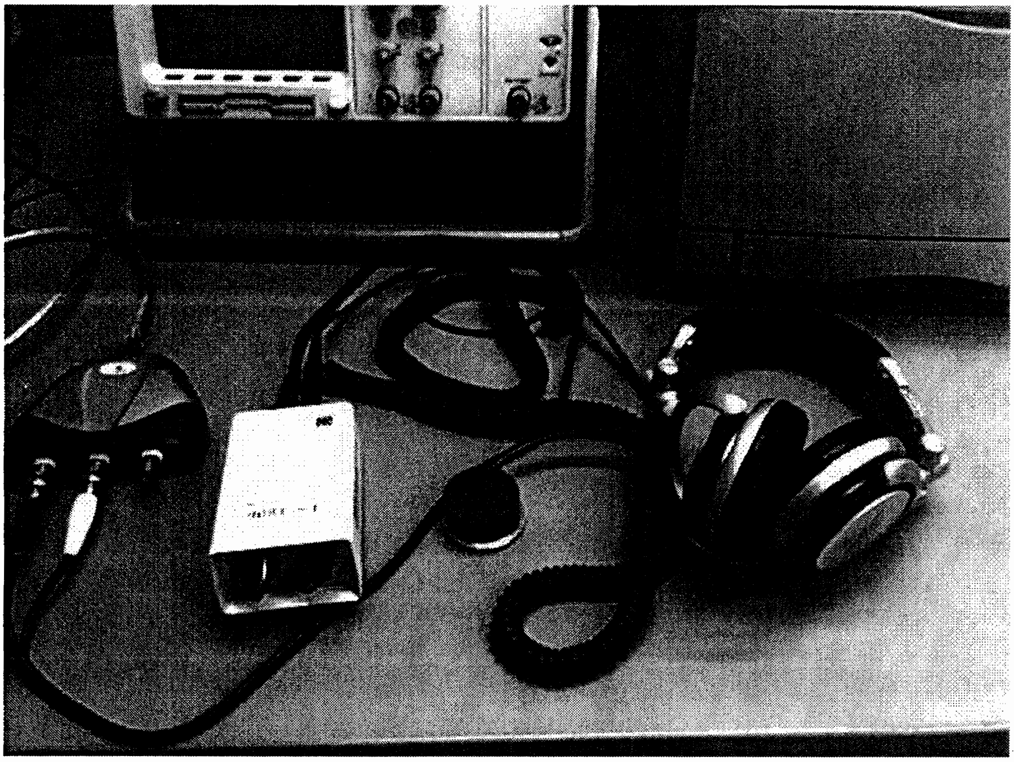

Figure 3.4: Equipment set up of the experiment

The set up of the experiment is shown in figure 3.4. The respiratory sounds would be acquired using an ES-1000 Electronic Stethoscope (CFE Inc. Medical

Products) with a contact sensor and headphones (Figure 3.5). The headphones were used to listen to the sounds. The sensor was placed on a chosen site on the chest or neck to record the respiratory sounds. The stethoscope was also responsible for filtering and amplifying the signal.

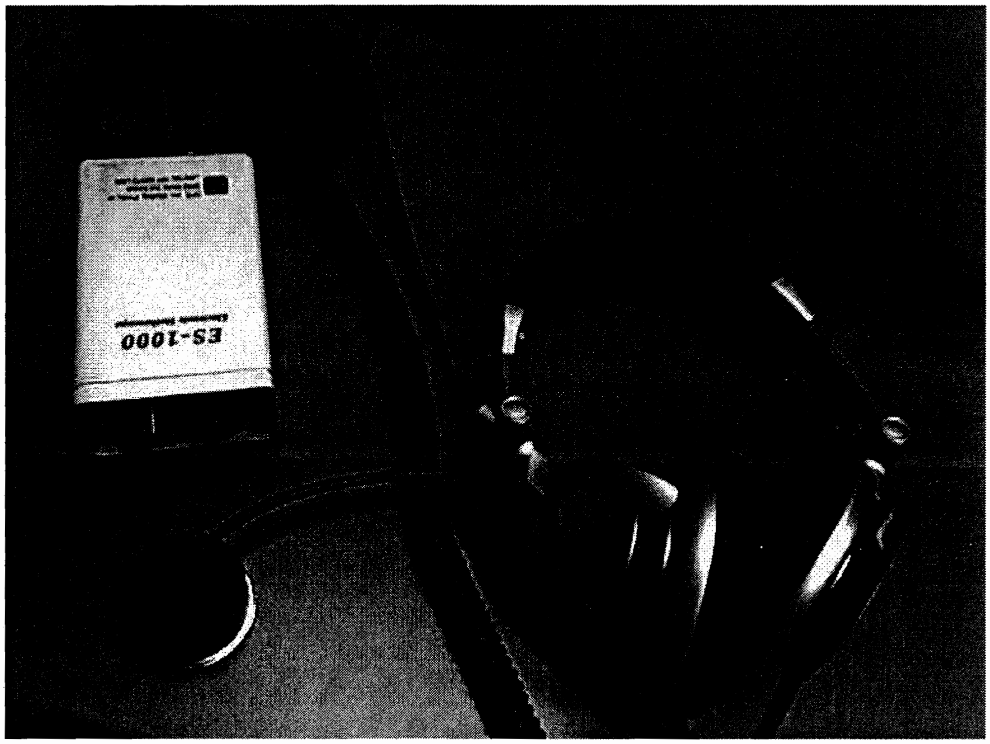

Figure 3.5: Electronic Stethoscope

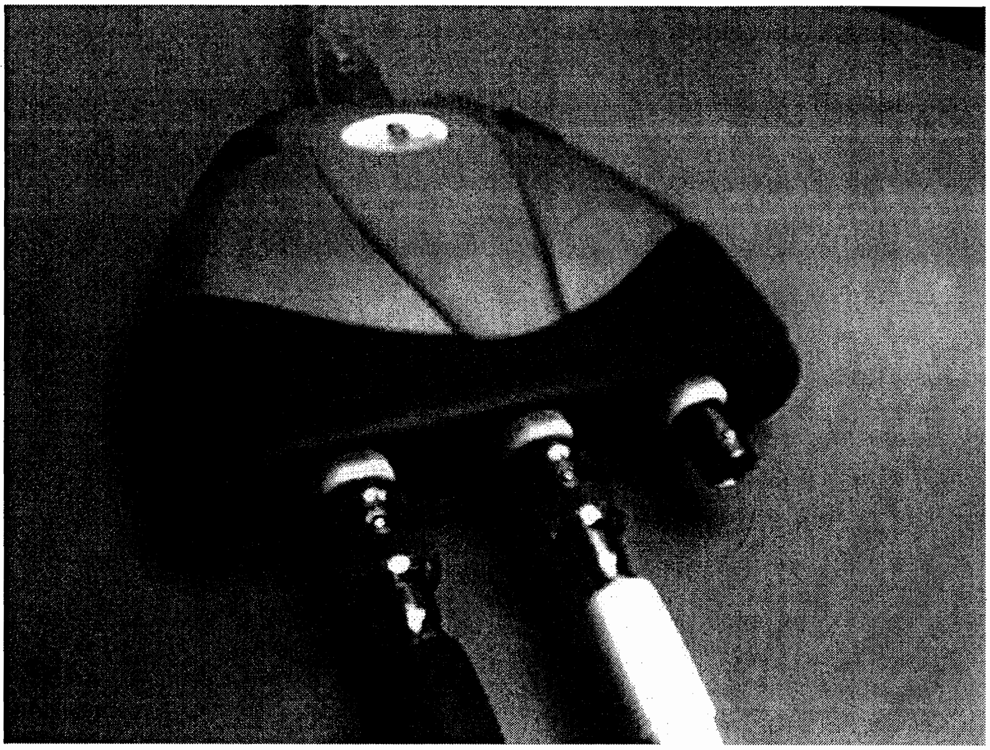

Figure 3.6: A/D Converter

An acquisition system (DS1M12 Stingray, USB Instruments) was used to convert the analog electrical signal to a digital signal (Figure 3.6). The A/D converter is connected to a desktop computer through an USB interface port. The LabView program, installed on a desktop computer, was used to extract the sound signal and then process the physiological information from the signal.

3.4 Experimental Procedure

In this experiment, six healthy volunteers participated in the study. The participants had no known history of respiratory illnesses. They comprised of two males and four females and their age range was 20-65 years old. The experiment took place in a normal room, filled with background noise originating from many sources like people movement, air conditioner sounds, doors slamming, equipment noise etc.

The subjects were in a sitting position for the entire duration of the experiment. The sensor was placed on the subject's neck for recording tracheal signals and for the left and right lungs, the respiratory sounds were recorded from the 2nd intercostal space on the chest. They were asked to produce regular, continuous breaths at 4 or 6 seconds interval for a total duration of 20-30 seconds. Two to four respiratory cycles were recorded where the sequence was inspiration for 2 or 3 seconds and expiration for another 2 or 3 seconds. Two sets of data for the left lung, right lung and the tracheal sounds were taken.

The raw signal, the signal envelope and the power spectrum were recorded for further analysis.

Chapter 4

Results and Discussion

4.1 Time Domain Results

The waveforms of the signal obtained in the time domain were plotted as amplitude (μV) against time (seconds). This section presents all the findings made on the raw signals in the time domain and their envelopes. Also the inspiration and expiration phases of the signals were studied.

4.1.1 Left and Right Lung Sounds

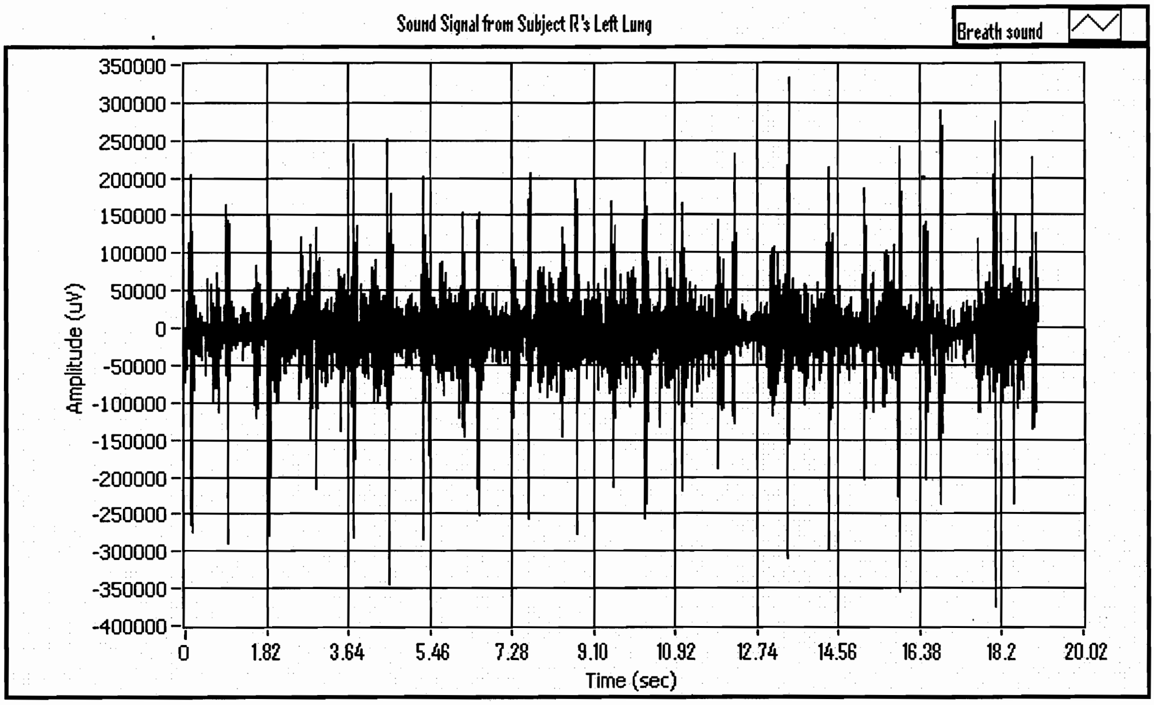

Figure 4.1: Graphical representation of the sound signal from the left lung of subject R

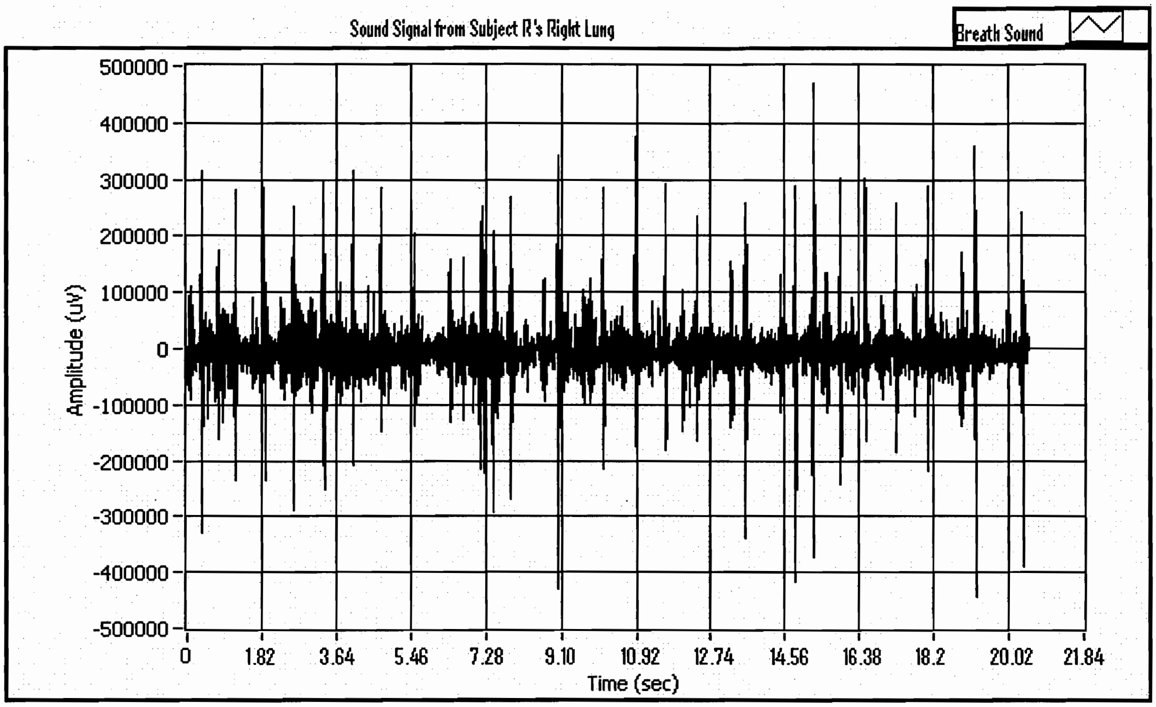

Figure 4.2: Graphical representation of the sound signal from the right lung of subject R.

Figures 4.1 and 4.2 show the left and right lung sound signals from subject R. The shape of the signals was irregular with no repetition pattern as found in previous studies such as the one by Charbonneu et al. (2000) with the TEWA technique, explained earlier in chapter 2. Spikes indicate the presence of heart sounds, which serve as noise interference and filtering techniques were used to remove them. The signal from subject R was acquired at a 6 second interval for one breath cycle. A breath cycle was comprised of the inspiratory and the expiratory components, where the time duration for each was 3 seconds. M was the only subject, who chose to breath at a 4 second interval for each cycle. Figures 4.3 and 4.4 present the breath cycles of the left and right lung signals.

There were variations in the shape and amplitude found in the waveforms of all the subjects. The inspiration phases of the left and right lung sound signals of subject R can be seen in figures 4.5 and 4.6. The expiration phases are shown in figures 4.7 and 4.8.

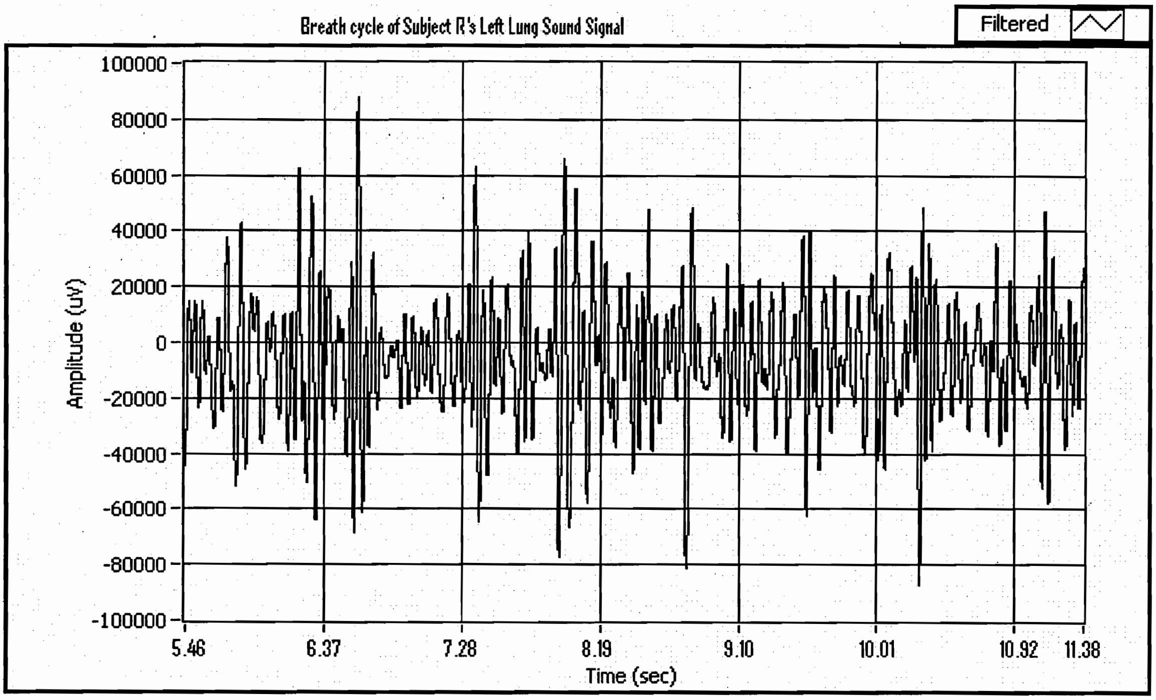

Figure 4.3: Graphical representation of one breath cycle from subject R's left lung

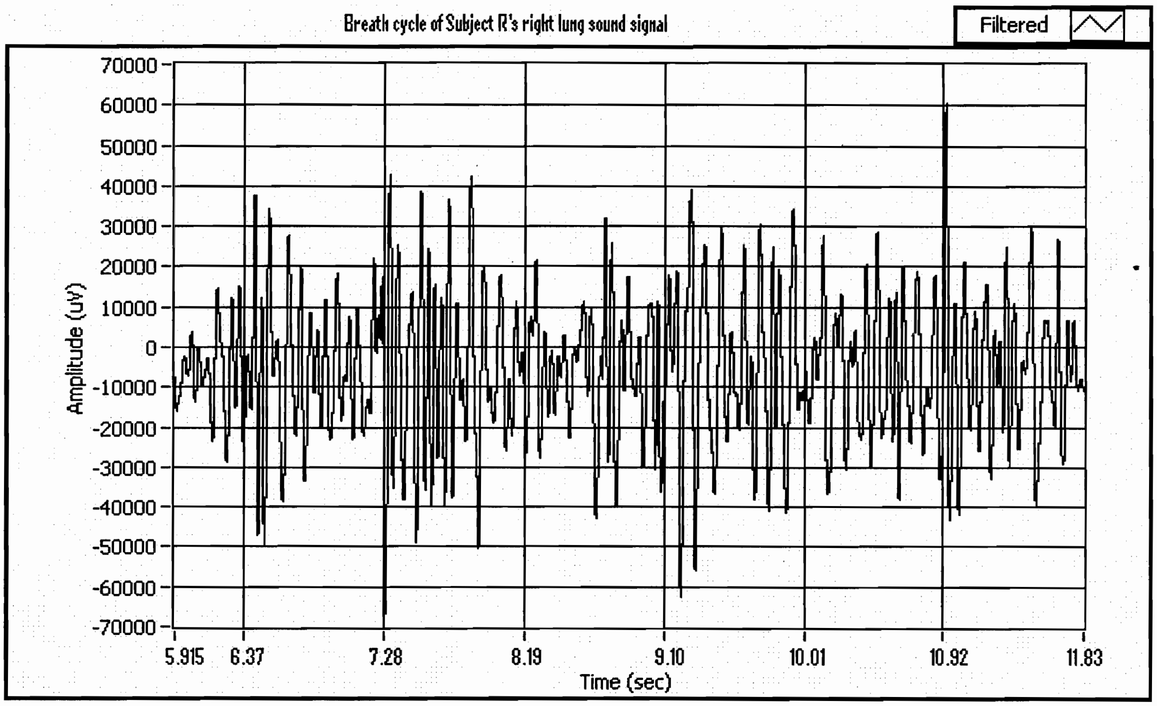

Figure 4.4: Graphical representation of one breach cycle from subject R's right lung

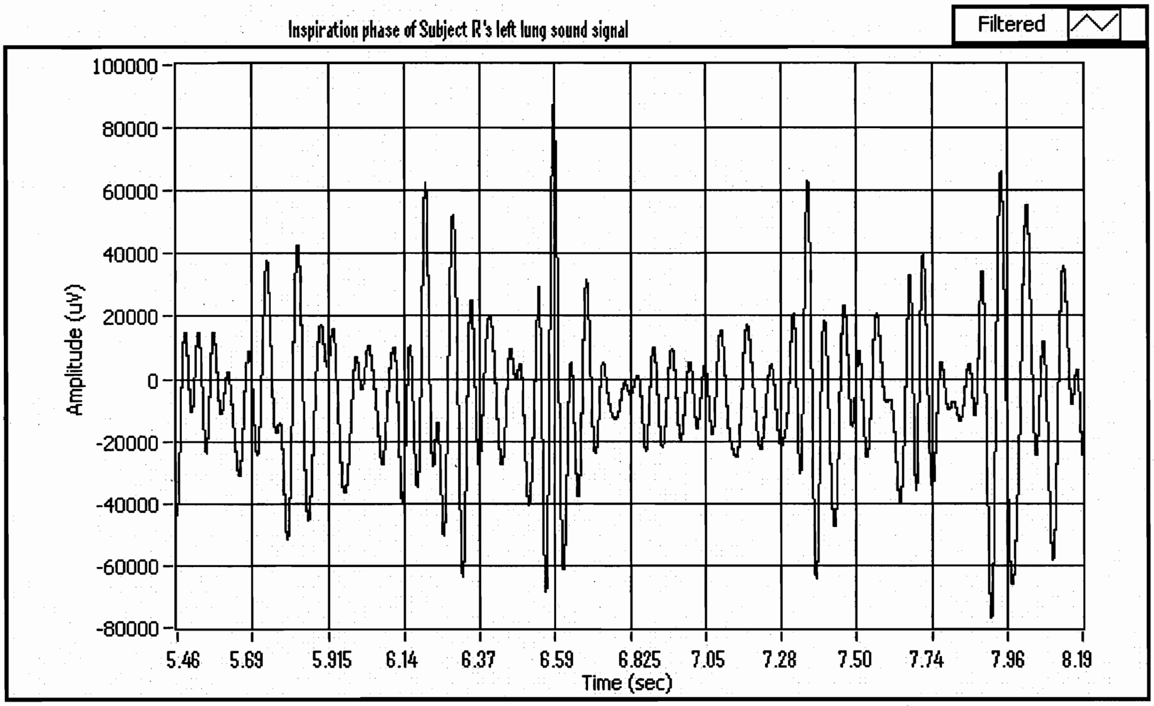

Figure 4.5: Inspiration phase of the sound signal from subject R's left lung

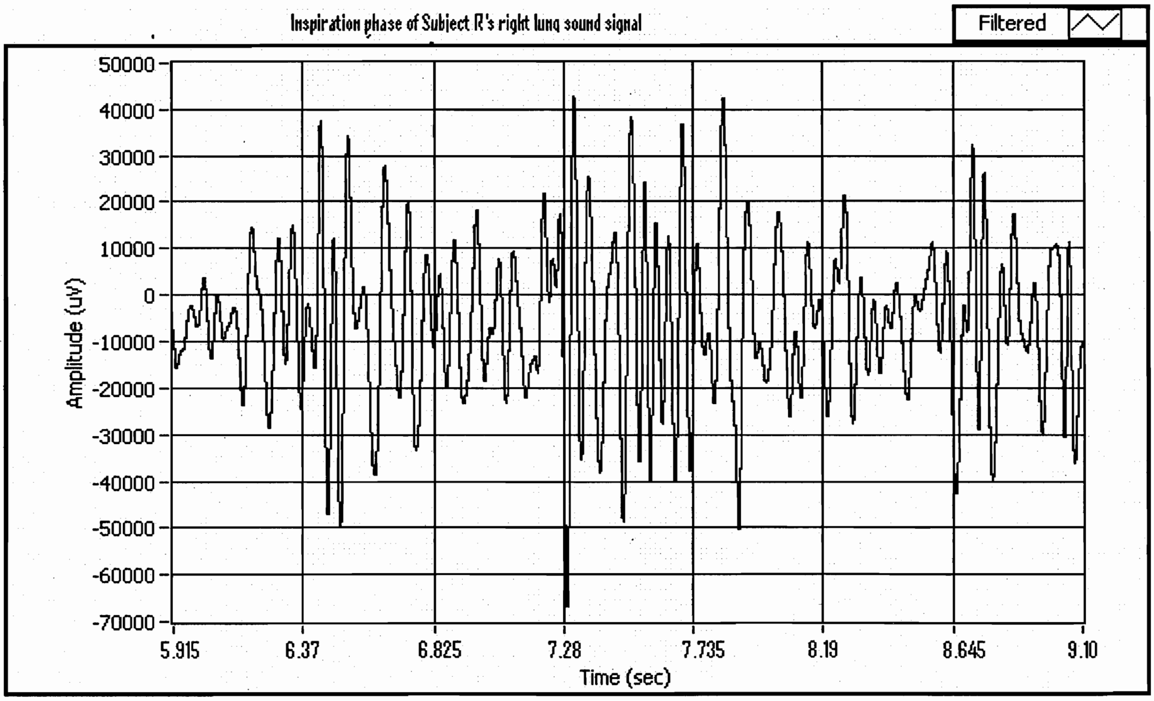

Figure 4.6: Inspiration phase of the sound signal from subject R's right lung

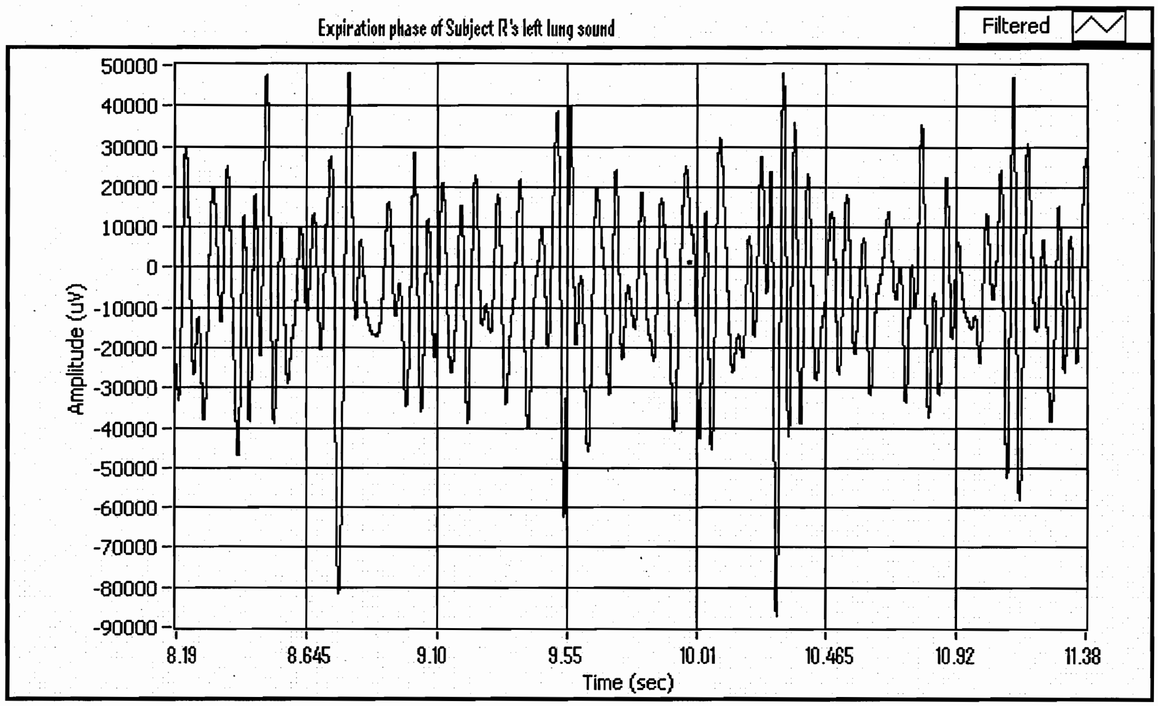

Figure 4.7: Expiration phase of the sound signal from R's left lung

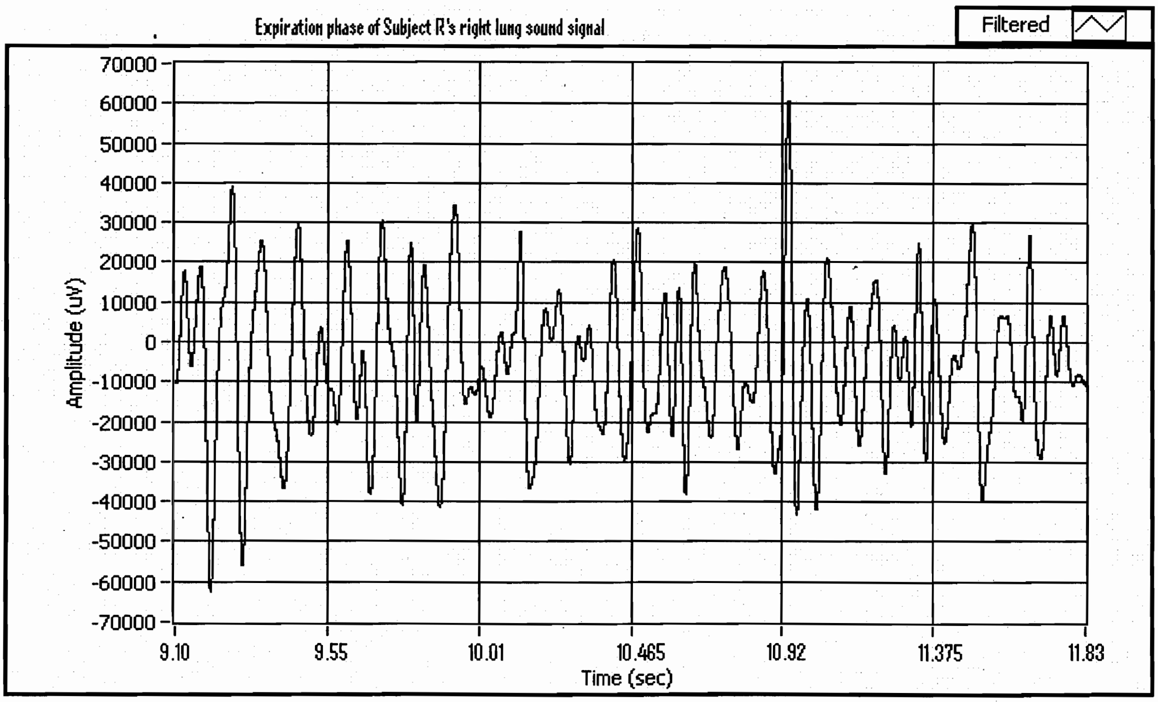

Figure 4.8: Expiration phase of the sound signal from R's right lung

To verify the similarities between the features of the lung sounds on the left and right side of the chest in the time domain, the signal amplitude had to be determined. In order to get the amplitude, an envelope had to be extracted from the raw signal. The signal envelope was obtained by applying the Hilbert Transform to the original signal. Both the Hilbert Transform of the signal and the original signal were increased by a factor of 2 and then added. The square root of the combined data was taken before the envelope was filtered to get rid of the noise interference.

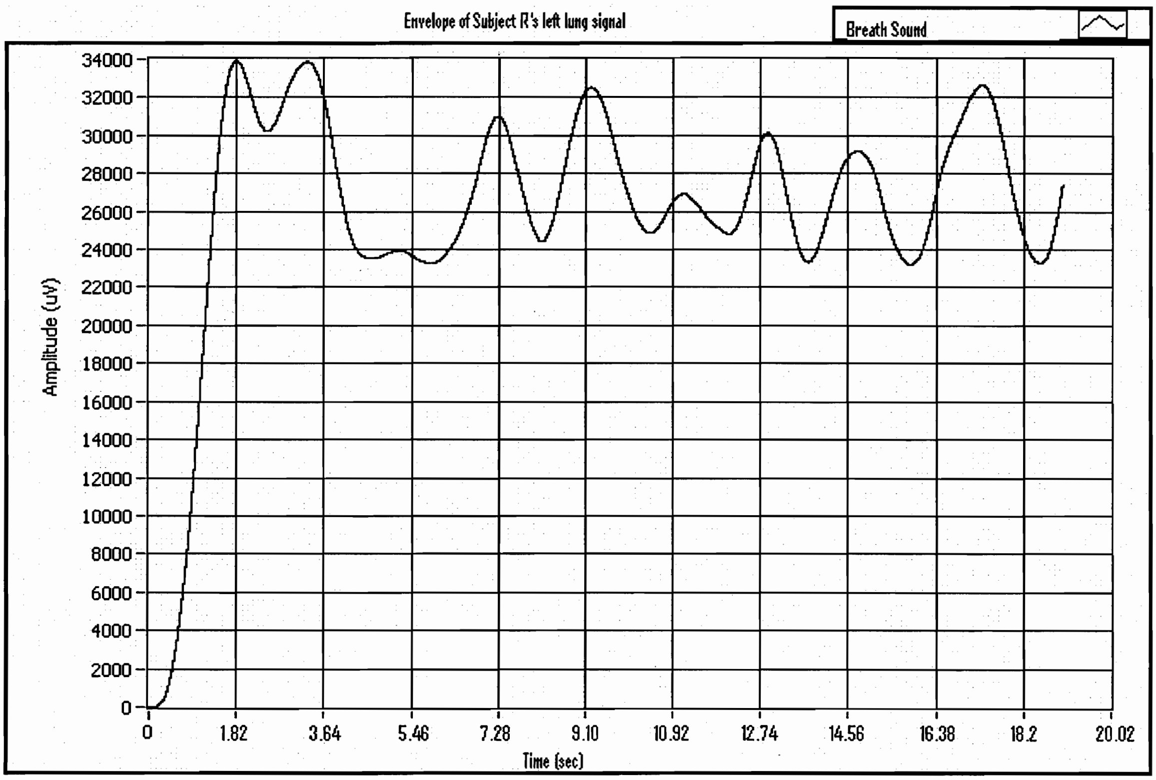

Figure 4.9: Signal envelope from subject R's left lung

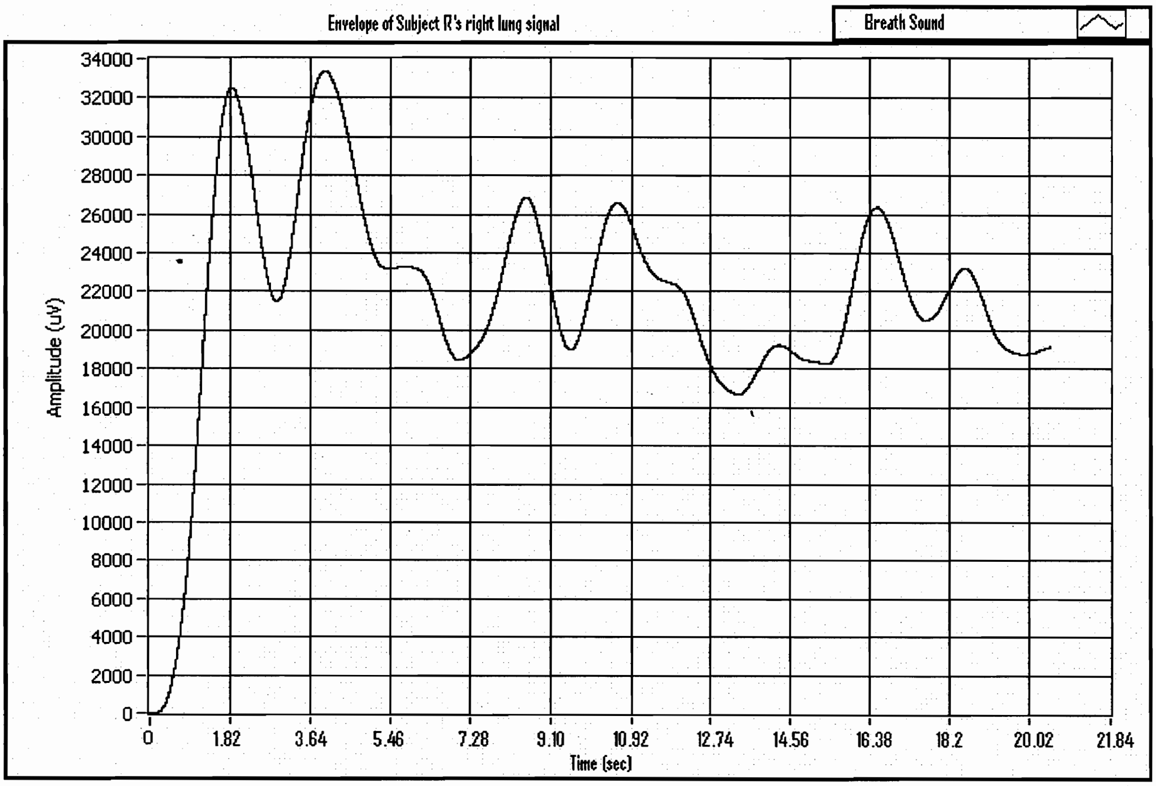

Figure 4.10: Signal envelope from subject R's right lung signal

The envelopes for the left and right lungs can be viewed in figures 4.9 and 4.10 respectively. For each subject, the peak amplitude in the inspiratory phase was taken for each breath cycle and then averaged over the total number of cycles, which was different for each subject. The average amplitude was the parameter used to compare the features of the left and right lung and the tracheal sounds. However in most subjects, either the first or the last cycle or both were ignored. This was because the LabView program failed to properly record the first or last cycle of the respiratory signal by either cutting off the inspiration part at the start or the expiration component towards the end.

The average amplitudes of the left and right lungs were found to be approximately close. This result indicated that there was a similarity between the features of the left and right lungs. Earlier in chapter two, it was mentioned that the sound intensity for inspiration would be generally higher than that of the expiration, thus the amplitude of the inspiratory phase would be greater than that of the expiratory phase. However in many signal envelopes, it was observed that the expiratory peak amplitude was higher in some of the breath cycles. This could be because of the way the subject was breathing, where he or she might have been putting more effort in the expiration than the inspiration at some point during the experiment. Figures 4.11 and 4.12 showed one breath cycle of the left and right lung signal envelopes respectively.

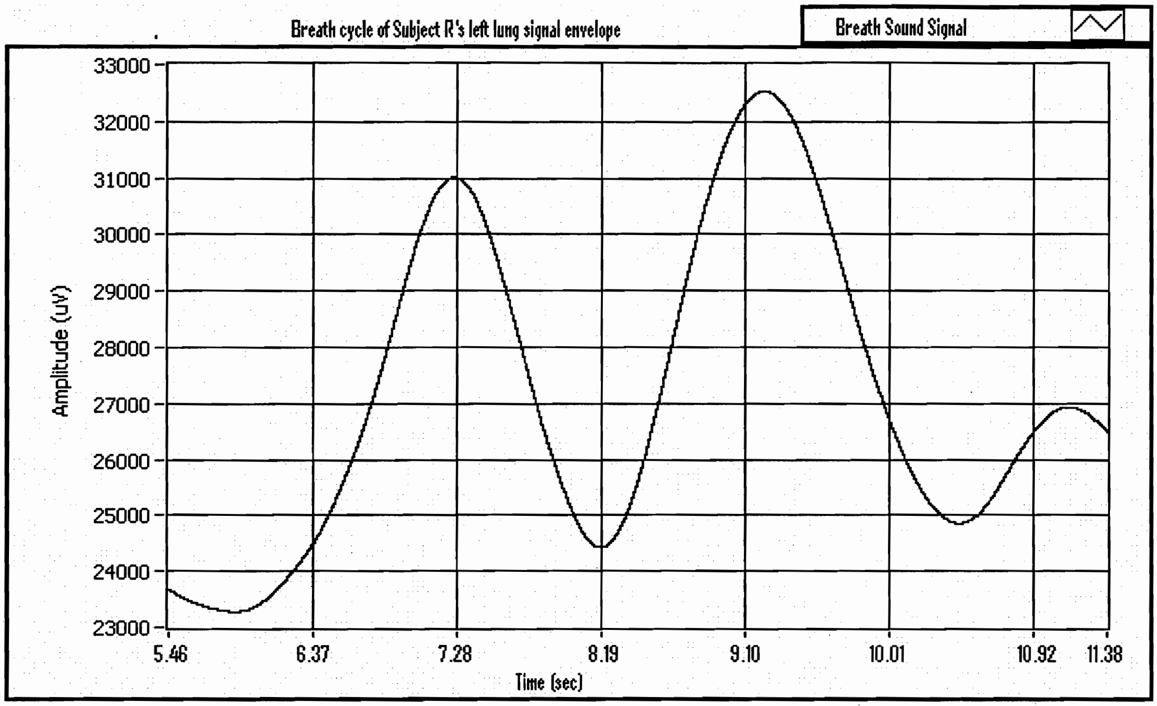

Figure 4.11: One breath cycle taken from subject R's left lung signal envelope

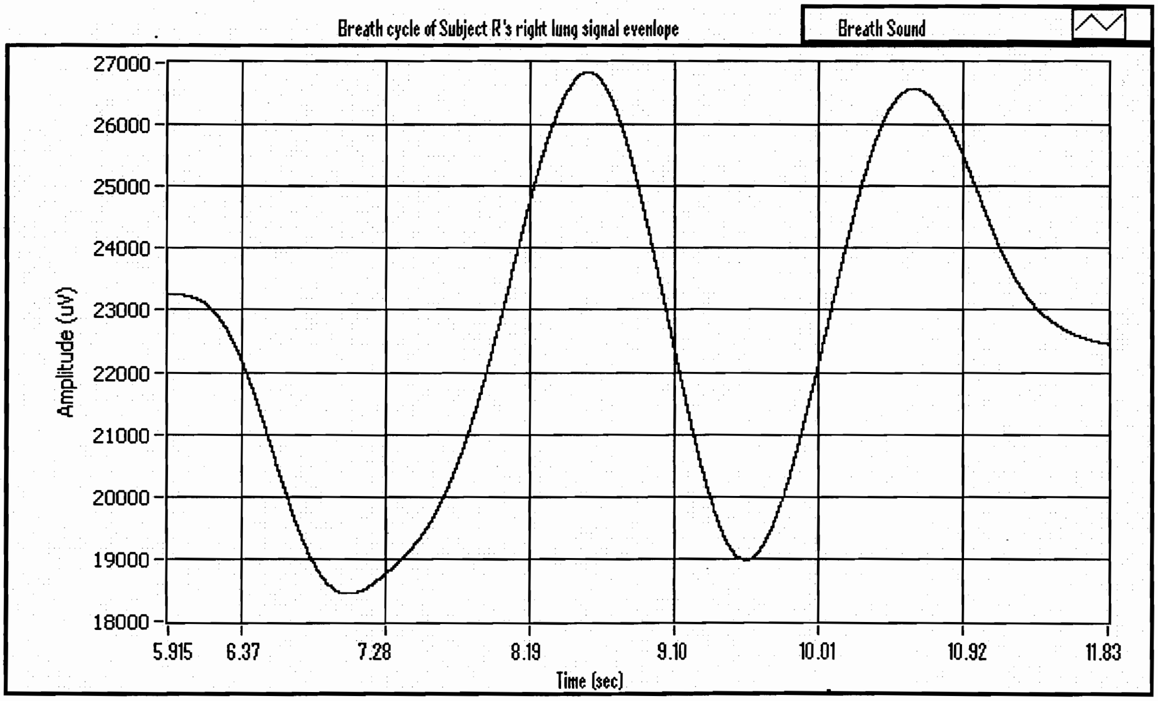

Figure 4.12: One breath cycle taken from subject R's right lung signal envelope

4.1.2 Tracheal Sound

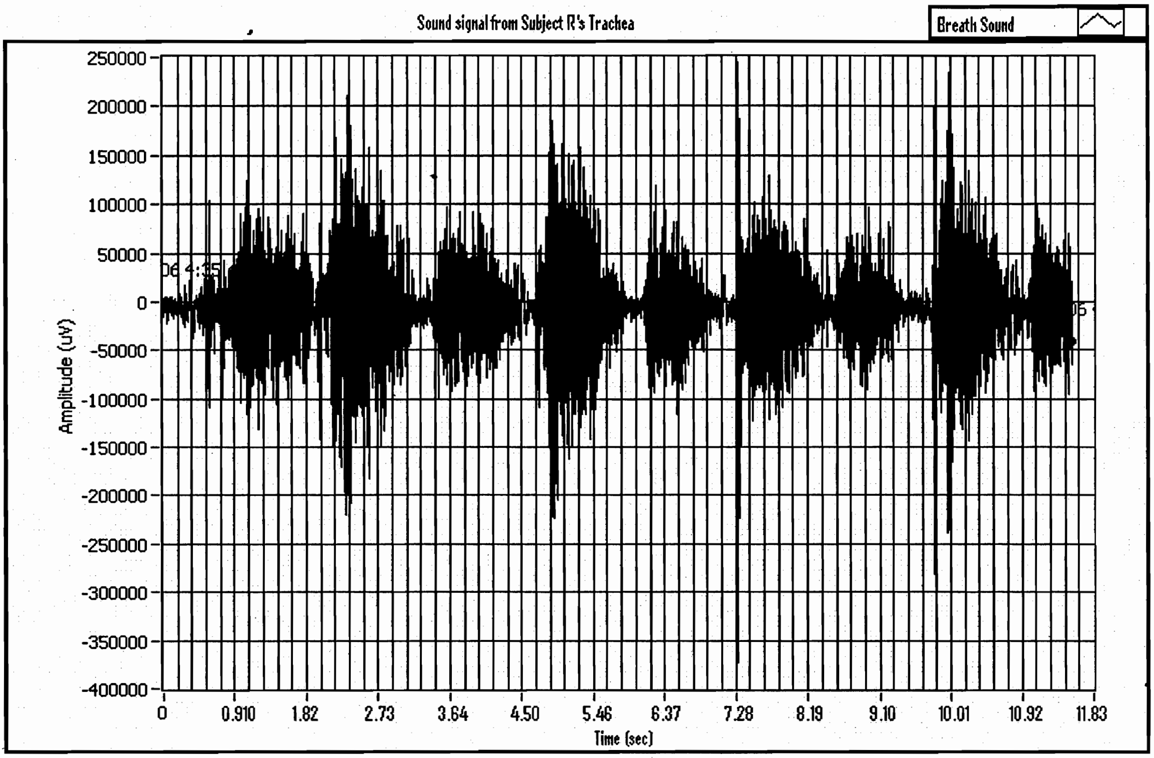

Figure 4.13: Graphical representation of signal taken from the trachea of subject R

Figure 4.13 shows the respiratory sound signal from the trachea of subject R. The breath cycle of a trachea sound cycle comprised of two short bursts, as seen in figure 4.14 and was also found in previous works such as a study conducted by Que et al. (2002). The general shape of the signal was found to be the same for all the subjects. The waveforms of the inspiration and expiration phases look almost identical to each other as seen in figures 4.15 and 4.16 respectively. The signal acquired from the throat was clearer to read than from the chest, because of less interference like heart sounds, muscle sounds, subject's clothing etc (Que et. al., 2002).

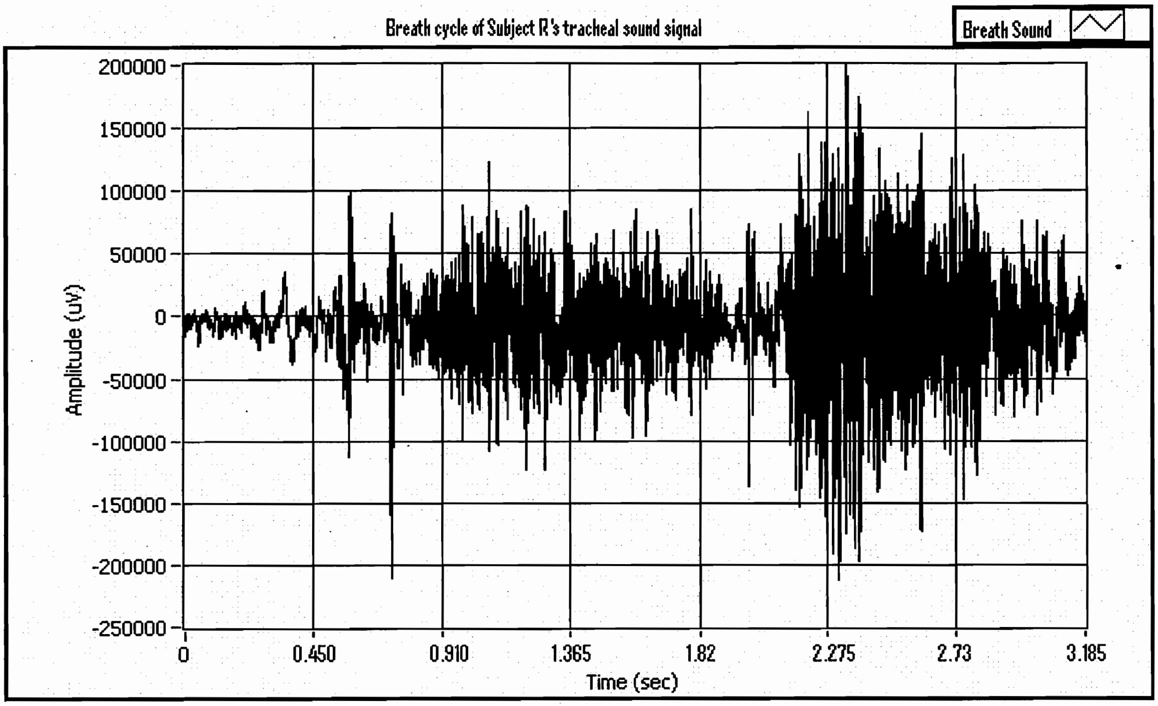

Figure 4.14: One breath cycle of subject R's tracheal signal

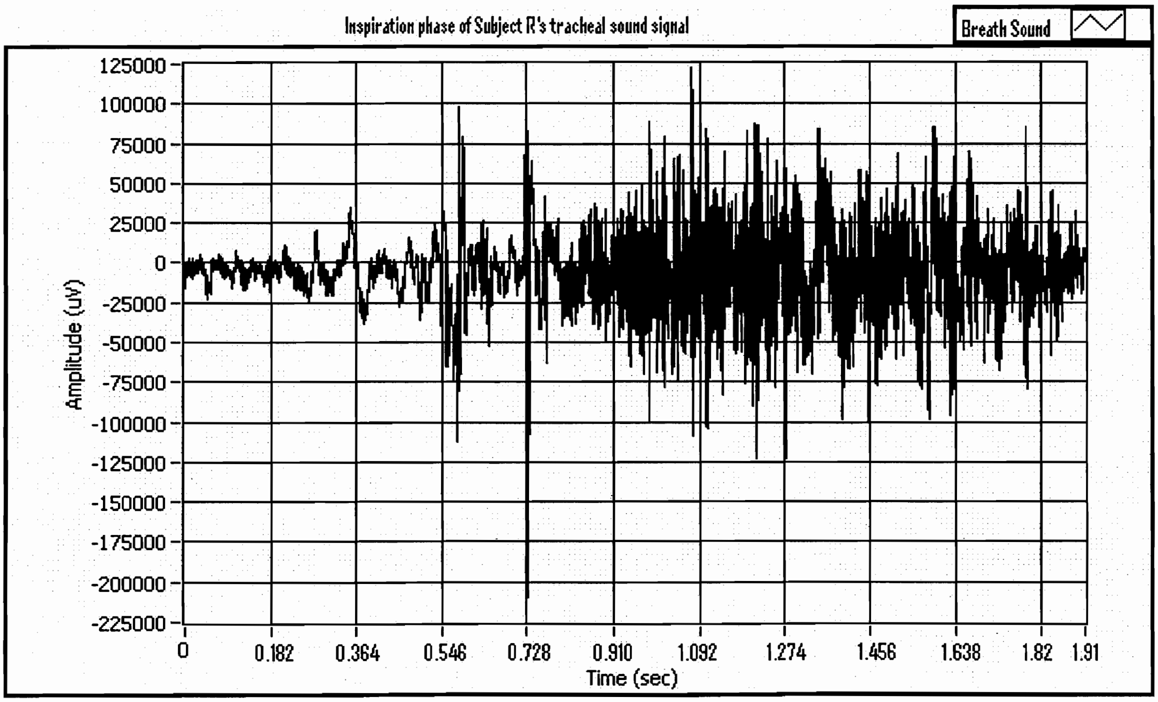

Figure 4.15: Inspiration phase of subject R's tracheal sound signal

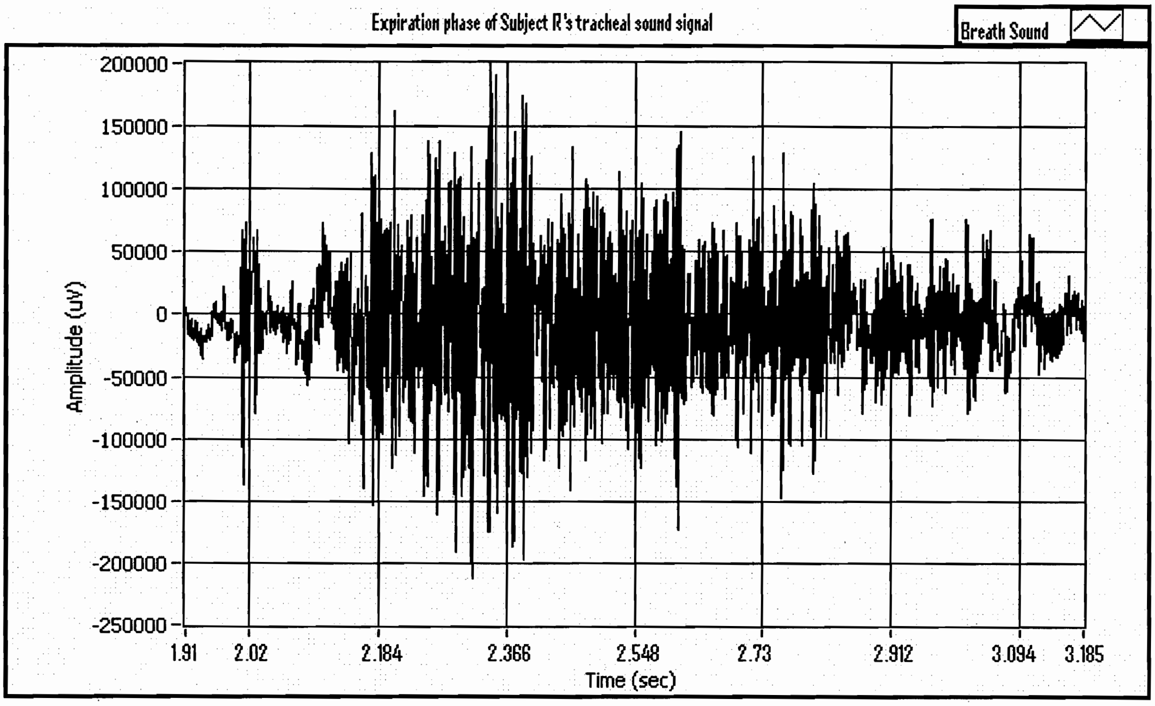

Figure 4.16: Expiration phase of subject R's tracheal sound signal

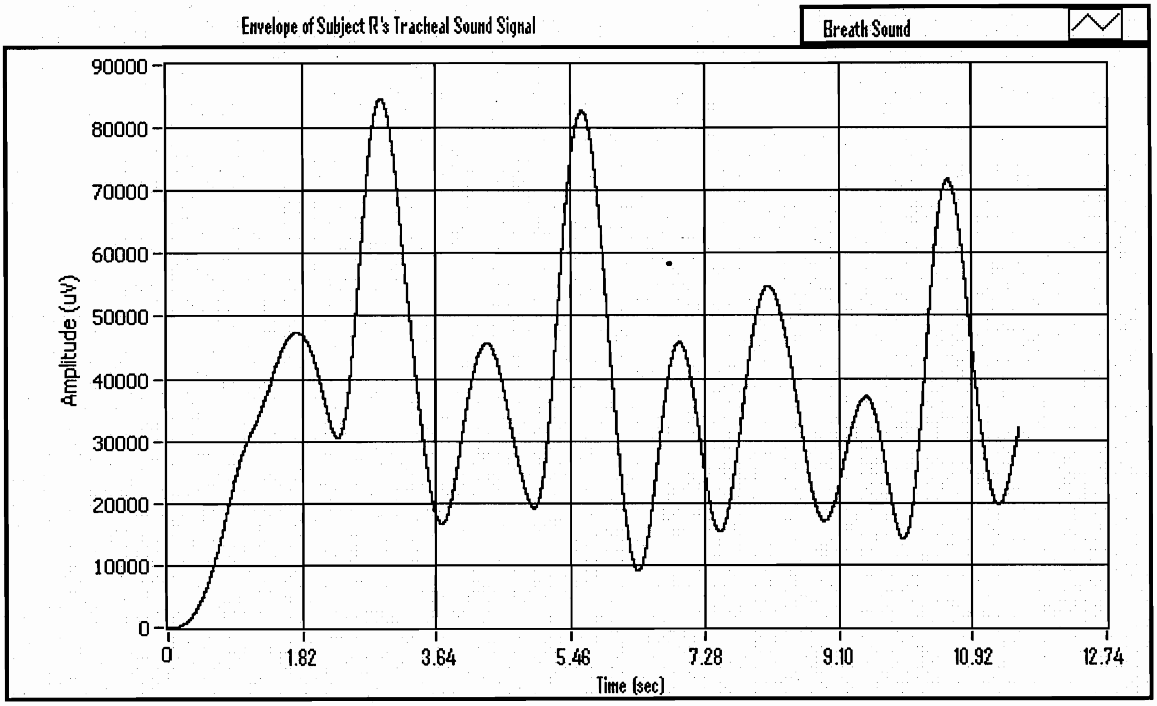

Figure 4.17: Signal envelope from subject R's trachea

In the signal envelopes of all subjects' tracheas, the amplitudes for the inspiration and expiration varied in each cycle, even though in some cycles, they were approximately close. The higher peak amplitude of each breath cycle was chosen, regardless of whether it was an inspiratory or an expiratory peak. That means if the peak amplitude of the inspiration phase was higher, then that value would be chosen for the calculation of the average amplitude.

The process of extracting the signal envelope from the trachea was the same as the lung sounds, where the average amplitude was determined. The average amplitudes of the tracheal signals were found to be different from the amplitudes of the lung sounds, which verified the program's capability to show one of the key features of the tracheal and lung sound signals are dissimilar. The amplitudes were also higher for the tracheal sound than that of the lung sounds and this is expected since the sound intensity when breathing from the trachea is greater than from the chest.

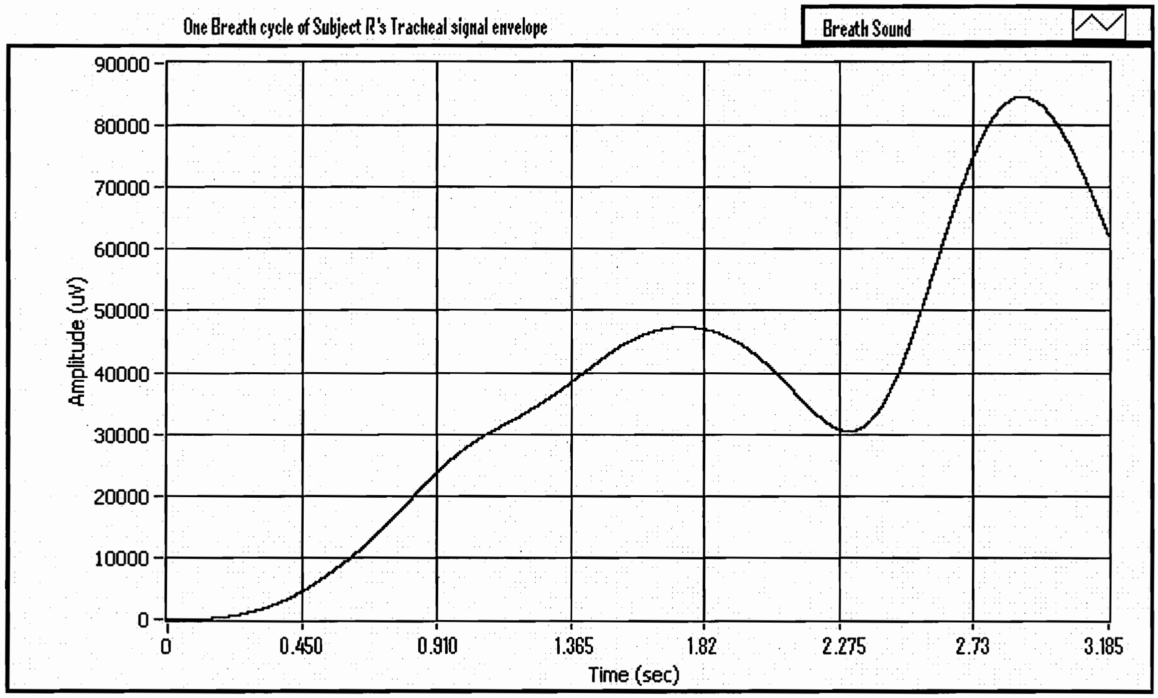

Figure 4.18: Tracheal sound signal envelope for one breath cycle from subject R's

4.2 Frequency Domain Results

The frequency spectrum was plotted as power (dB) against frequency (Hz). After applying FFT to the signal to extract the power spectra, the following parameters were recorded: maximum power (peak intensity), frequency ranges for the lung sounds and the highest intensity in that particular range and for the trachea, the second peak. This section summarizes the findings of the power spectra of the left and right lung sounds and the tracheal sounds.

4.2.1 Lung Sound Power Spectrum

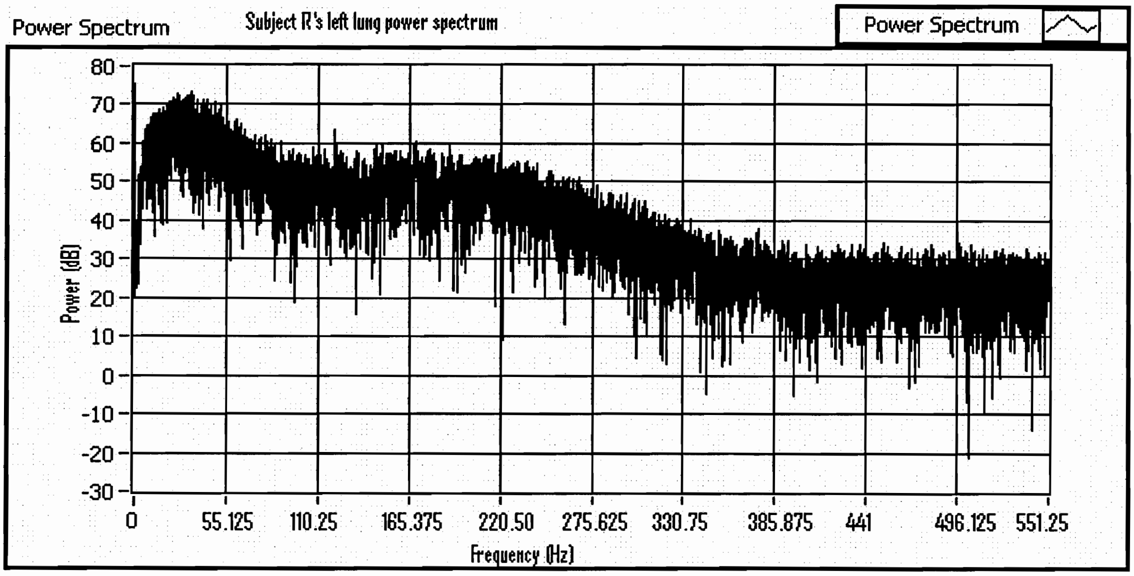

Figure 4.19: Power spectrum of subject R's left lung signal

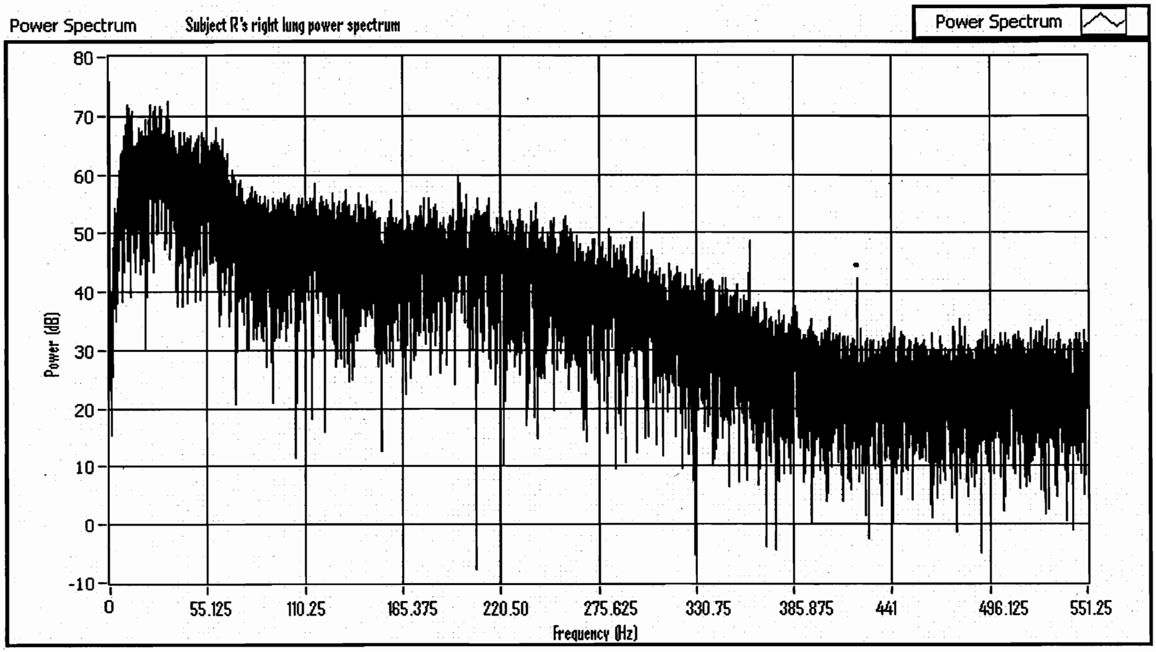

Figure 4.20: Power Spectrum of subject R's right lung signal

The power spectra of the left and right lung sound signals shared a common pattern with only a few variations in all the subjects. The maximum power (peak intensity) was high, ranging from 65 dB to 75 dB for both lungs. Subject R's left and right lung frequency spectra, as seen in figures, showed a gentle, downward slope from the peak. In Subject R's left lung, the frequency spectrum leveled off in a plateau in the frequency range, 100-220.50 Hz till it fell again. There was no plateau in subject R's right lung in that range. The power distribution intensities of both lungs decreased till they reached the frequency at 385.75 Hz and beyond where the spectra flattened at around 30 dB.

The common feature seen in both lungs for all subjects was the decrease in power distribution as the frequency increased and the spectra flattening out at around 30-40 dB in the frequency range, 330.75-441 Hz.

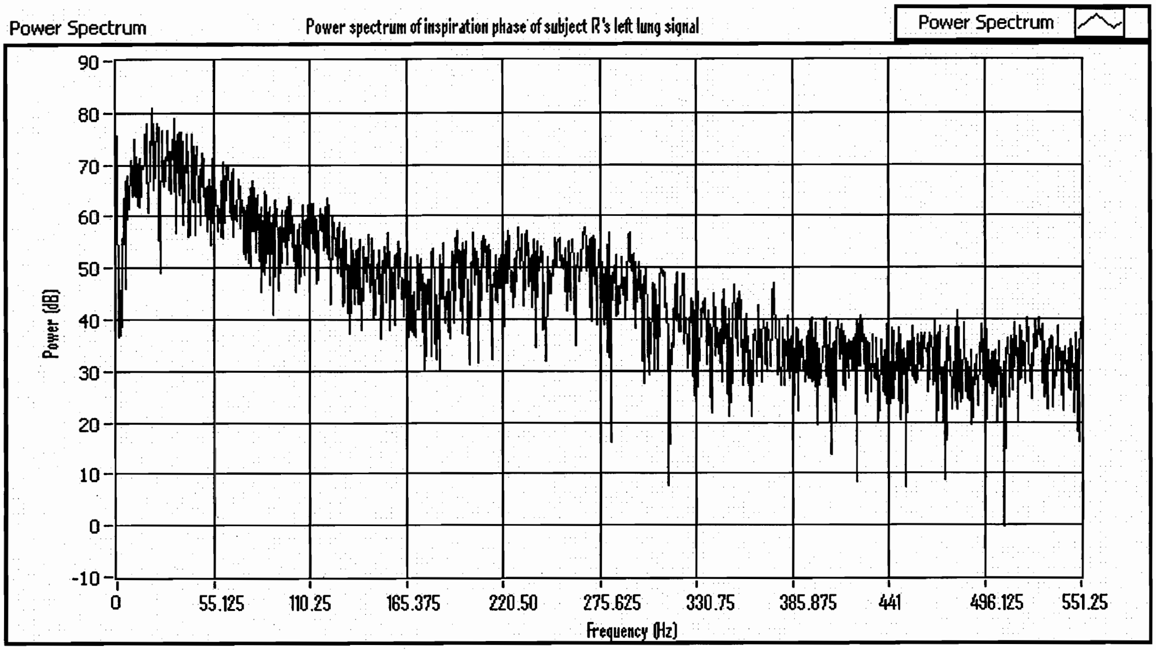

Figure 4.21: Power spectrum of the inspiration phase of R's left lung signal

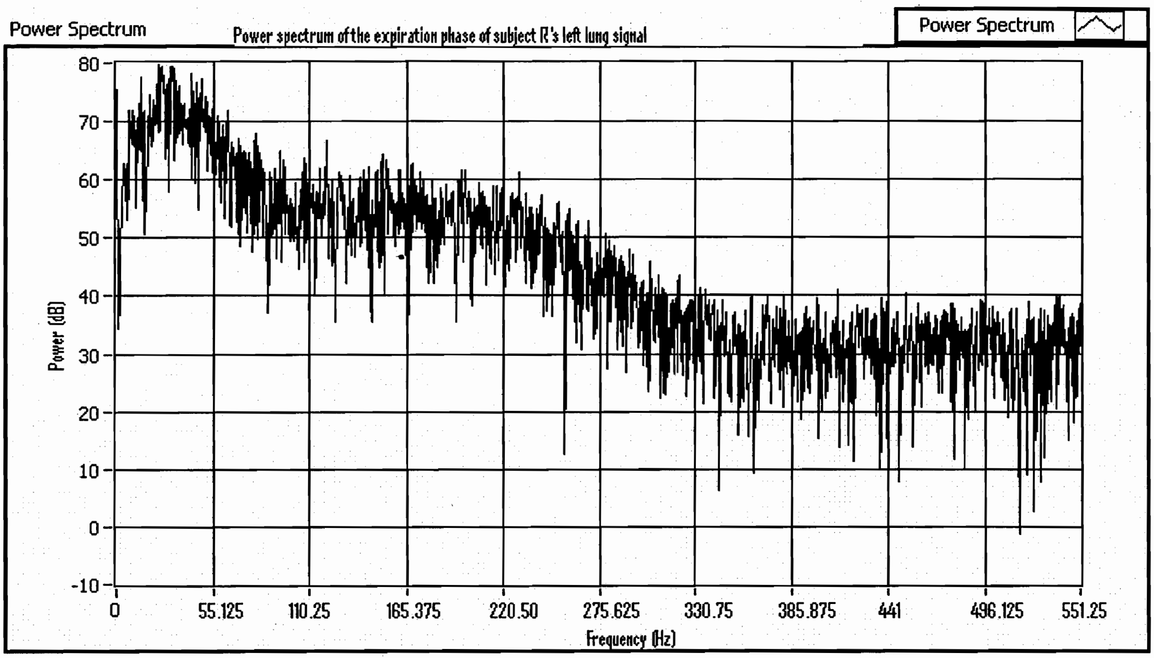

Figure 4.22: Power spectrum of the expiration phase of R's left lung signal

From figures 4.21 and 4.22, it could be observed that the power spectra of the signal in the inspiration and expiration phase of the left lung had similar features. However there were slight variations found in both spectra. In the frequency range, 30-165.375 Hz, the downward slope in the power spectrum of the inspiration phase was steeper than the slope in the expiration phase, which leveled off into a plateau at the range, 165.375 Hz- 220.50 Hz. But in the frequency range, 220.50 Hz - 330.75 Hz, it was the spectrum of the expiration phase that had the steeper downward slope than in the inspiration phase.

Both spectra flatten at 330.75 Hz and beyond. The peaks were approximately close at around 79 dB, which was higher than the peak in the whole signal.

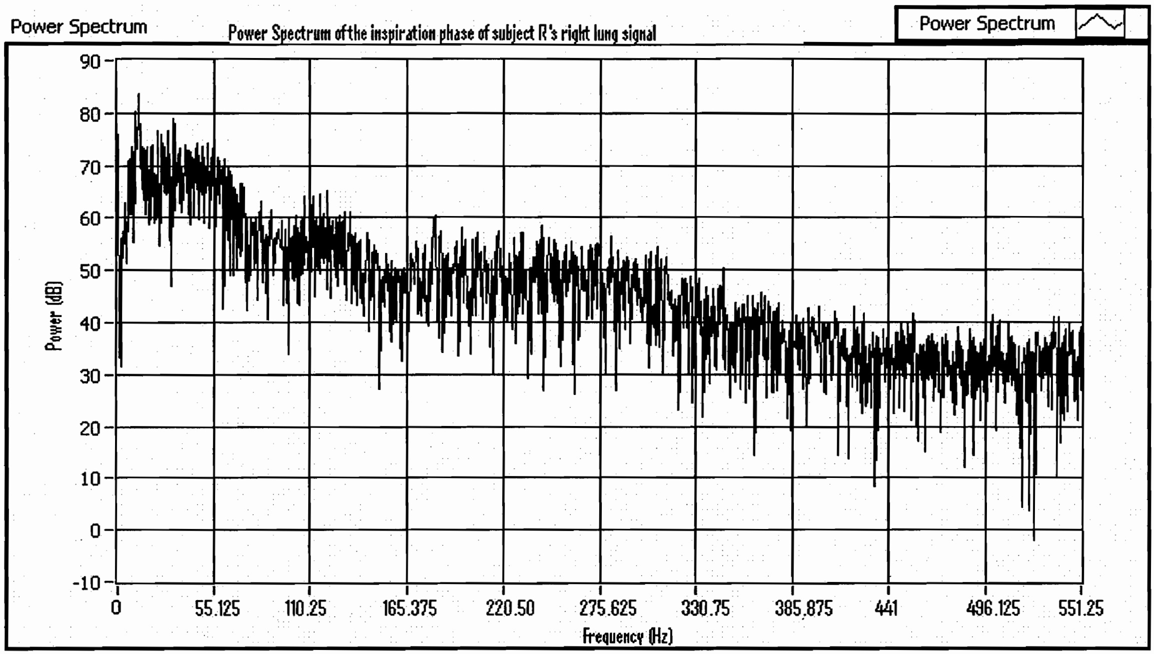

Figure 4.23: Power spectrum of the inspiration phase of Subject R's right lung

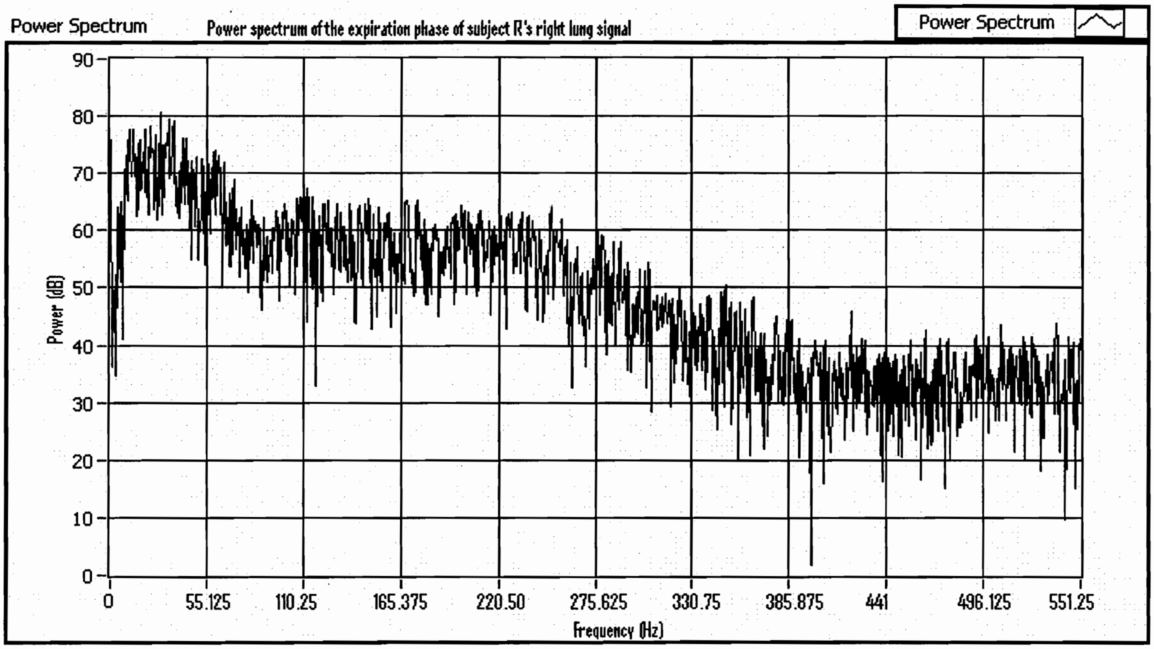

Figure 4.24: Power spectrum of the expiration phase of Subject R's right lung

Like the left lung, the power spectra of the right lung in the inspiration and the expiration phases also shared similar features (see figures 4.23 and 4.24). Tiny variations were seen in both spectra. The spectra of the inspiration phase in the left lung were found to be not different from the right lung for all subjects, which further verified the similarities between the features of the left and right lung sounds.

4.2.3 Trachea

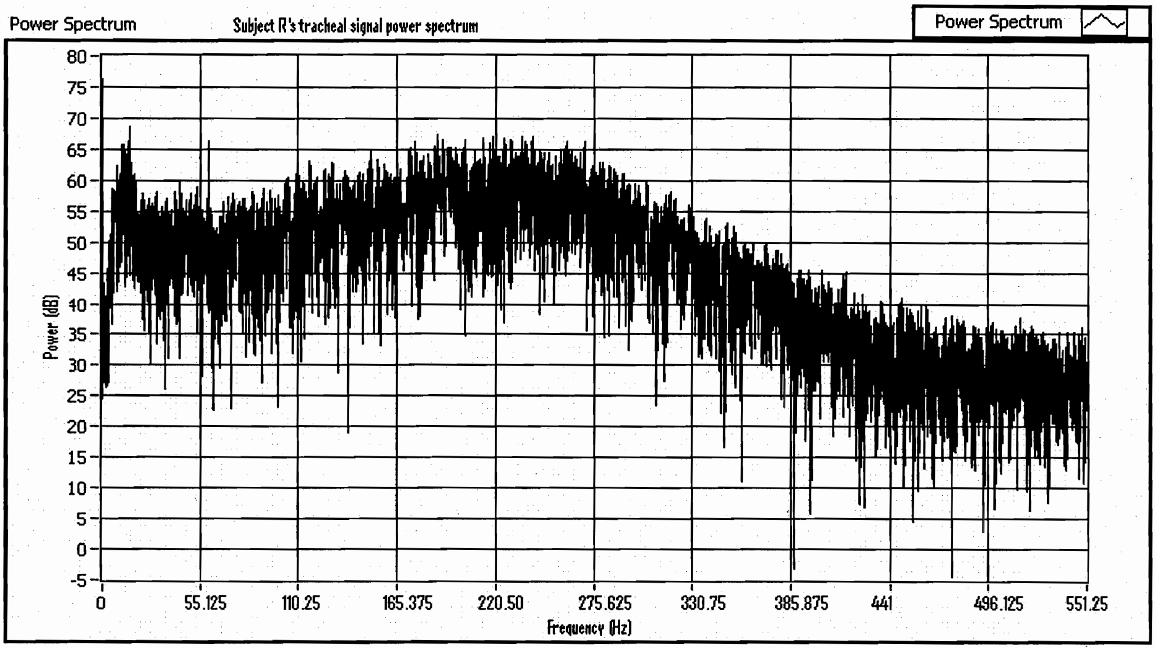

Figure 4.25: Power spectrum of subject R's tracheal signal

The power spectra of trachea signals was found to be the same for all subjects, where it displayed two peaks as confirmed by previous works in the frequency domain analysis of trachea sound signals (Gavrieli et. al., 1981). There was a difference between the power spectra of the lung and the tracheal sound signals. The power spectra of the lungs had only one peak.

The maximum power for each subject was found to be slightly lower than that of the lungs. For all subjects, there was a shallow slope from the maximum power that leveled off at the frequency of 55.125 Hz. From there, gradual rise began till the spectrum reached its second peak, whose value was found to be close to the first peak.

From the second peak onwards, the power decreased exponentially until it leveled off at around 30 dB in the frequency range, 385.875 -441 Hz.

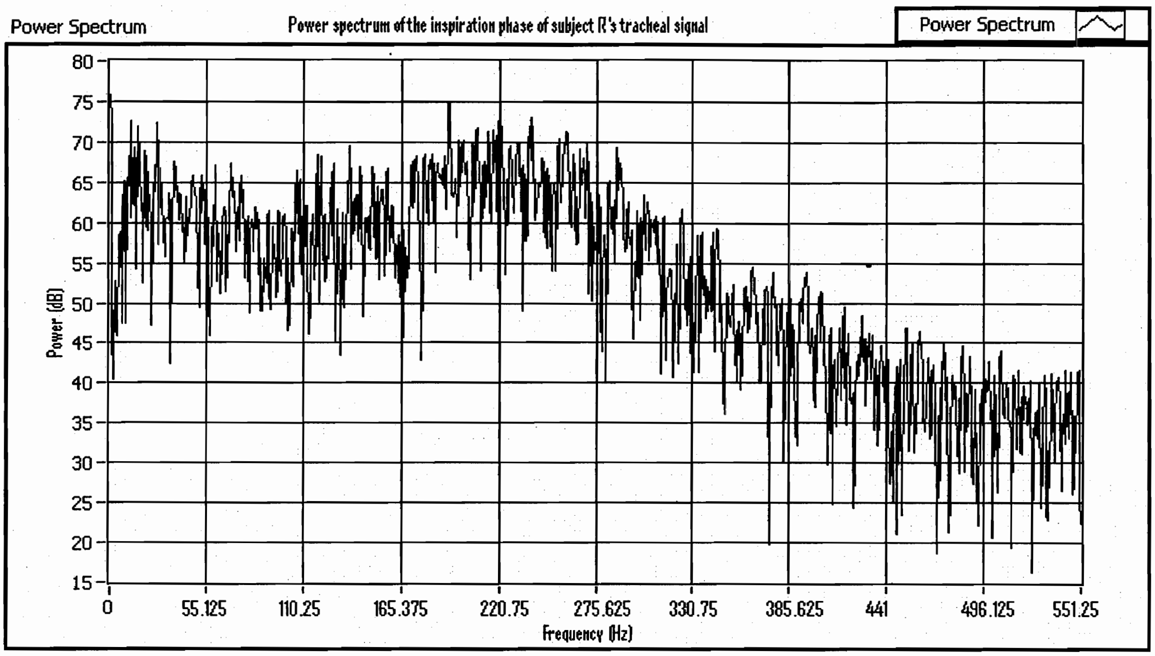

Figure 4.26: Power spectrum of the inspiration phase of R's tracheal signal

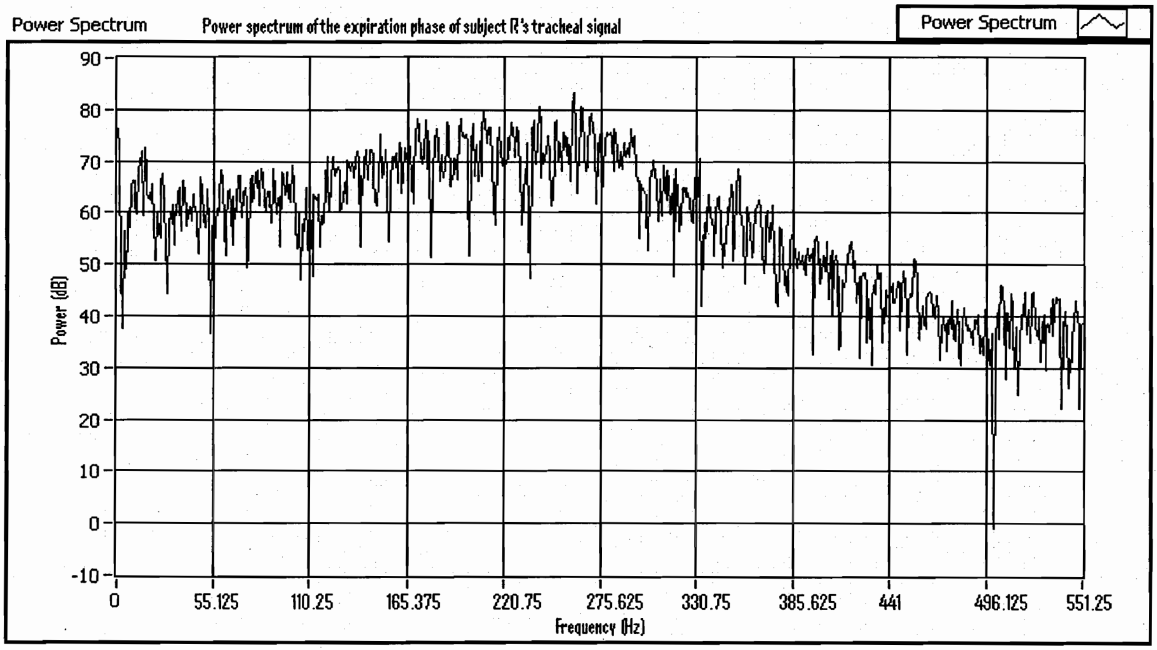

Figure 4.27: Power spectrum of the expiration phase of R's tracheal signal

Figures 4.26 and 4.27 showed that the power spectra of the inspiration and expiration phases of the trachea were similar in their appearance, where they had two peaks, just like in the power spectrum of the whole signal. There were differences evident between the two phases. The power spectrum had a smooth and gentle upward slope in the frequency range, 27 Hz to 220.50 Hz for the expiration phase and in the range, 110.25 Hz- 220.50 Hz for the inspiration phase.

4.3 Analysis

The analysis of the results from this study involves how the key features of the left and right lungs and the trachea are compared in all the subjects. The average amplitudes of the signals and the power distribution will be the chosen parameters for the time and frequency domains respectively.

4.3.1 Time Domain

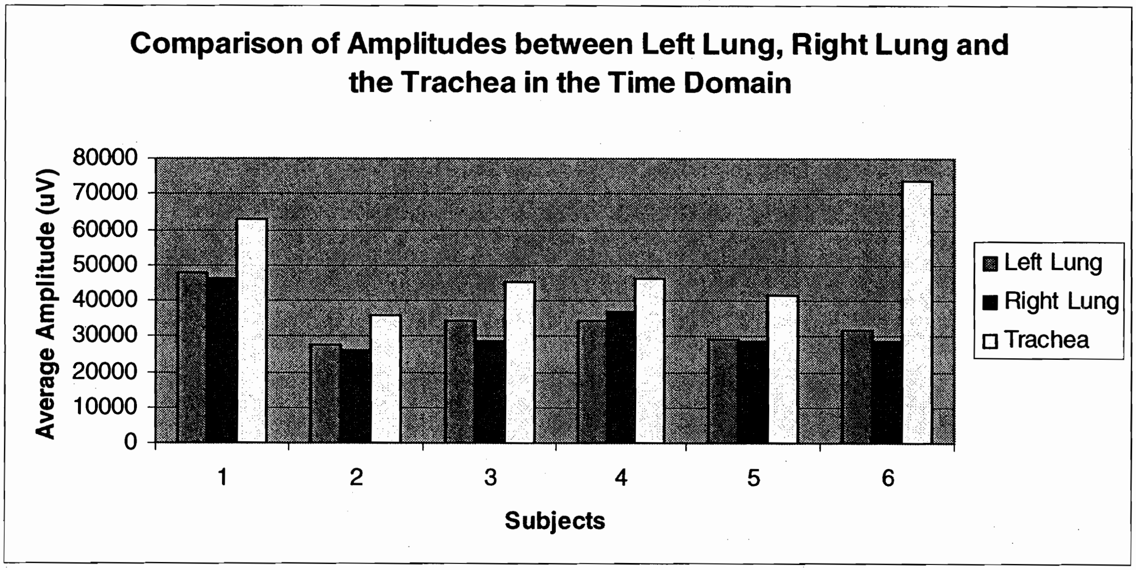

Figure 4.28: A bar chart displaying the average amplitudes of the lungs and tracheal sound signals from all subjects

The average amplitudes of the lungs and the trachea from all subjects were used to make a comparison between the features of the left and right lung and tracheal sounds in the time domain. The process of obtaining the amplitudes was explained earlier in section 4.1. A bar chart was generated by excel to show how the average amplitudes of the left lung, right lung and the tracheal sound signals compared with each other in all subjects.

From the bar chart displayed in figure 4.28, it was concluded that the average amplitudes of the left and right lung sounds were close as expected at the beginning of this thesis. The average amplitudes between the sounds from the lungs and the trachea were different. Also, the tracheal signal amplitudes were also found to be greater than that of both lung sounds and this was due to the higher sound intensity in the trachea than the lungs.

The amplitudes that were determined from the envelopes were found to be not consistent amongst the subjects. The shapes of the envelope were not the same for each subject and even within the same person, the extraction of the signal envelope turned out to be different every time. This was because the raw signal acquired from each person was different. Reasons for the variation were due to a number of factors such as site location, where as mentioned in previous works, the lung sound intensity varied for different sites. Other factors included age, gender, weight and height of the person and the intensity of the person's breathing. The above results show the LabView program can be used to verify the significant features of the breath sound signals from the lungs and trachea.

In chapter one, it was briefly discussed in previous works that the methods for time domain analysis had been limited in their capacity and objectivity. The above result in figure 4.26 confirmed that we could establish the features of the left and right lung sounds and the tracheal sound using the signal amplitude. The physician or surgeon would be able to identify whether the signal came from the lungs or the trachea by just looking at the signal envelopes and obtain the amplitude.

4.3.2 Power Spectrum

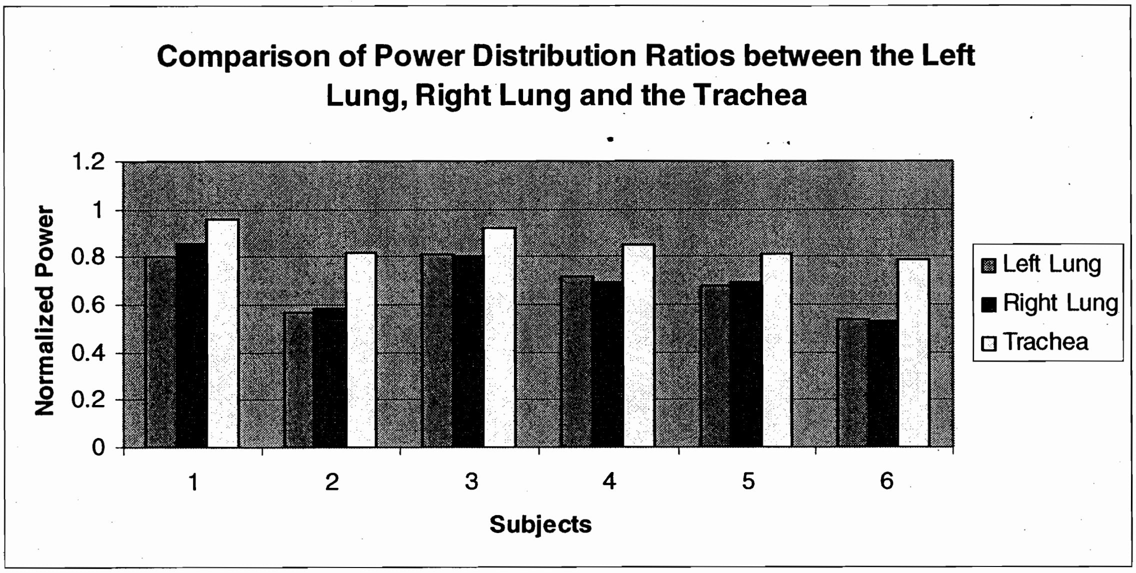

Figure 4.29: A bar chart showing the power distribution ratios from the whole signals of the left and right lungs and the trachea.

For each subject, the peak intensity of the spectrum was located. Then a frequency range was chosen depending on the subject's power spectrum and for both the left and right lungs, the range usually contained the spectrum's downward slope. The highest power intensity in that range was picked and normalized to the maximum power in the spectrum. The range was decided in a way to ensure the normalized power distribution ratios for both lungs were approximately close to each other. For the trachea, the second peak, found to be slightly smaller, was chosen and normalized to the first peak.

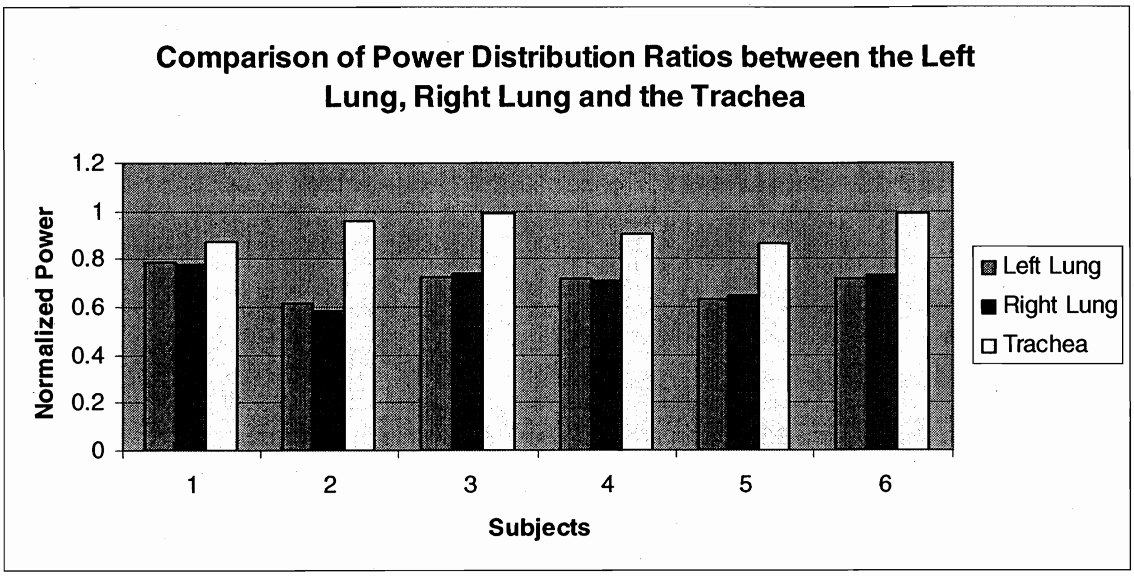

Figure 4.30: A bar chart showing the power distribution ratios, taken from the inspiration phase of the left and right lungs and the trachea.

Both bar charts were generated by Excel to compare the ratios taken from the frequency spectrum of the whole signals and the inspiration phases for all subjects. The chart showed the difference to be minimal between the left and right lung sounds for each person. However the difference was clearly evident between the signals from the trachea and both lungs. As expected, the ratios taken from the trachea were shown to be higher than that of both lungs. This was due to the higher sound intensities recorded over the trachea than the chest. For frequency domain analysis, either comparison can be used for studying the data and defining the features on the power spectra of the sound signals from the lungs and the trachea. These results had verified the efficiency of the software application to establish the key features of the breath sound signals from the lungs and the trachea.

From observing the results from the analysis, it would be better to use the comparison made on the average amplitudes between the lungs and tracheal sound signals. This is because the signal envelopes were extracted in real-time, where the parameters (time and amplitude) used would give an accurate picture of what is happening in the respiratory tract. With the power spectrum, the power distribution was plotted on the log scale and normalized to the highest amplitude in the signal after applying the windowing function.

4.4 Statistical Analysis

The statistical analysis of the result comprised of three tests, which utilized the paired t-test method. The goal of these tests was to make a statistical comparison between the left and right lungs and the trachea. There will be three sets of data compared for each test between the left and right lungs; left lung and trachea; right lung and trachea. The tests were performed separately on the average amplitudes in the time domain, normalized intensity ratios from the power spectra of the whole signal and the intensity ratios from the power spectra of the inspiration phase in the frequency domain. The null hypotheses set were that the difference between the means of the following pairs (left lung- right lung; left lung-trachea; right lung-trachea) would be zero.

The tests were made with the help of the JMP statistical software. JMP was first developed by the SAS Institute, Inc. at Cary, N.C. in 1989. The application is not part of the SAS system, but portions of JMP were adapted from the various routines in the SAS system, particularly for linear algebra and probability calculations (SAS Institute, Inc., 2005). JMP utilizes a very advanced graphical interface for the display and the analysis of data.

4.4.1 Average Amplitudes in the Time Domain

The test was set to give a 95% confidence interval (two-tailed test). If the p-value was found to be less than 0.05, the null hypotheses would be rejected, thus indicating a significant difference between the two means (Ott, 2001). The predictions made for this test were that the amplitudes of the left and right lung sounds would be statistically the same, but there would be a statistical difference between the signal amplitudes from both lungs and the trachea.

4.4.1.1 Left-Right Lung

The p-value was 0.1950, which was greater than 0.05. This result showed that there was no statistic difference between the means of the left and right lung data, which supported the null hypotheses. This meant that the average amplitudes between the left and right lung signals were found to not be statistically different.

4.4.1.2 Left Lung-Trachea

The p-value was calculated to be approximately 0.0225, found to be less than 0.05. This result rejected the null hypothesis and showed that there was a significant difference between the means of the left lung and the trachea data. The conclusion of this particular test showed that the average amplitudes between the left lung and tracheal sounds were statistically different from each other.

4.4.1.3 Right Lung- Trachea

The p-value here was found to be approximately 0.0205, which was less than 0.05. Thus the result ended up rejecting the null hypotheses, showing the difference between the means of the right lung and the trachea data were significant. This test concluded that there was a statistical difference between the signal amplitudes of the right lung and the trachea.

4.4.2 Power Spectrum of the Whole Signal

The test was set to give a 95 % confidence interval (two-tailed test). If the p-value were less than 0.05, the null hypotheses would be rejected, thus indicating a significant difference between the two means (Ott, 2001). The following predictions made for this test were that the power distribution of the left and right lung sounds would be statistically the same, but there would be a statistical difference between both lungs and the trachea.

4.4.2.1 Left-Right Lung

The p-value was 0.7498, which was greater than 0.05. The result showed there was no difference between the means of the left and right lung values, which supported the null hypotheses. This meant the power distribution of the left and right lung sounds were statistically no different from each other.

4.4.2.2 Left Lung-Trachea

The p-value was calculated to be approximately 0.0022, thereby rejecting the null hypotheses. This result showed that there was a significant difference between the means of the left lung and the trachea data. Thus the conclusion of this test was the features of the power distribution between the left lung and the tracheal signal were shown to be statistically different.

4.4.2.3 Right Lung- Trachea

Just like the Left Lung-Trachea pair, the p-value here was found to be very small at approximately 0.0047, thereby rejecting the null hypotheses. Thus the difference between the means of the right lung and the trachea data were significant. The conclusion to the test is that the power distribution between the left lung and trachea sound were found to be statistically different.

4.4.3 Power Spectrum of the Signal in the Inspiration Phase

The test was set to give a 95 % confidence interval (two-tailed test). If the p-value were less than 0.05, the null hypotheses would be rejected, thus indicating a significant difference between the two means (Ott, 2001). The predictions made for this test were that the power distribution in the inspiration phase of the left and right lung sounds would be statistically the same, but there would be a statistical difference between the power distribution of the inspiration component from both lungs and the trachea.

4.4.3.1 Left-Right Lung

The p-value was 0.9797, which showed there was no difference between the means of the left and right lung values and this supported the null hypotheses. The result concluded that there was no statistical difference between the power distribution in the inspiration phase of the left and right lung sounds.

4.4.3.2 Left Lung-Trachea

The p-value was calculated to be approximately very small at 0.0012 (less than 0.05), thereby rejecting the null hypotheses. This result concluded that the power distribution in the inspiration phase of the left lung and tracheal sounds were statistically different.

4.4.3.3 Right Lung- Trachea

Just like the Left Lung-Trachea pair, the p-value here was found to be very small at around 0.0016, thus rejecting the null hypotheses. The results proved there was a statistical difference between the power spectra in the inspiration phase of the right lung and tracheal sound signals.

Chapter 5

Conclusions and Future Work

5.1 Conclusions

The goal of the thesis was to develop a software application, capable of extracting the features of normal lung and tracheal sounds. The methodology comprised of a set up involving a program, developed in the LabView environment, an electronic stethoscope, and a USB acquisition system. The stethoscope was used to record the sounds and the acquisition system acted as an A/D converter. The LabView program extracted the sound signals in real-time and processed them in the time and frequency domains for further analysis.

The amplitude was used to verify the key features of the sound signal in the time domain and the power spectra for the frequency domain. For the left and right lung sounds, the inspiration component was mainly considered.

The average amplitudes of the left and right lung sounds were found to be approximately close, but different between both lungs and the trachea. The tracheal sound average amplitudes were greater than that of the lung sounds. The amplitude in the inspiration phase was supposed to be generally higher than that of the expiration phase. However in many of the signal envelopes, there were breath cycles, where the peak amplitude in the expiration phase was found to be higher than that of the inspiration phase.

The power spectra shared a similar pattern for the left and right lung sounds. The inspiration phases for both lungs were found to have minimal differences between each other. The power spectra of the tracheal signal were different from both lung sound signals. Two peaks were found in the frequency spectra of the trachea, whereas the lung sound spectra only had one peak.

In conclusion, the software application in LabView had been able to verify that the key features of the respiratory sound signals from 1) the left and right lungs were similar and 2) the trachea and both lungs were found to be different.

5.2 Suggestions for Future Work

In this section, suggestions for future work after the study are discussed. The suggestions involve using the same software application to extract the features of the different adventitious and normal respiratory sounds, finding a way to simultaneously record the signals from different sites using more than one channel and conducting the experiment with the subject in a supine position instead of sitting.

5.2.1 Adventitious and Normal Breath Sounds

The software application, developed for this project, can be used to extract the features of adventitious breath sounds and compare them with normal respiratory sounds. Data from patients, suffering from respiratory diseases, can be gathered along with healthy volunteers, using the same experimental set up as in this study. More subjects should be studied in order to get a more accurate result. The same parameters would be used here to establish the key features of the respiratory sound signals, with the amplitude for the time domain analysis and the power spectra in the frequency domain.

5.2.2 Simultaneous Recording of the Respiratory Sounds

In this thesis, only one channel was used for acquiring the signal. Here the same sensor was used for all three sites, which posed some problems in terms of getting accurate data. There are variations found in the signals taken at different locations over the chest. When recording was completed for the left lung, the sensor had to be placed on the corresponding site over the right lung. However with only one channel, the sensor's location on the right side of the chest could be off and the signal acquired could end up coming out differently from the other side.

With more than two channels, the sensors can be placed over both sides of the chest and the trachea at the same time. This would improve the accuracy of the signals, thereby ensuring a clearer representation of the waveforms and a better comparison of the breath sound signal features.

5.2.3 Supine Position

The subject was made to be in a sitting position for the entire duration of the experiment. The sensor was placed over the 2nd intercostal space to record the lung sound signals and it was the only site on both sides of the chest, where the amplitudes of the left and right lung sounds were found to be approximately the same. It would be interesting to see how the features of the sound signals would compare, if we recorded the data at the base of the lung like the 5th or 6th intercostal space. This experiment could be tried with the subjects placed in a supine position, where they would be either lying on their backs or at one side.

REFERENCES

REFERENCES

American Medical Association (AMA). http://www.ama-assn.org/ama/pub/category/7166.html

Beckerman, R.C., Wegmann, M.J., Waring, W.W. Tracheal breath sounds for detection of apnea in infants and children. Critical Care Medicine 1982; 10: 363-366.

Bishop, R.H. LabView 7 Student ed. Upper Saddle River, NJ, Prentice Hall 2006.

Charbonneau, G., Racineaux, J.L., Sudraud, M., Tuchais, E. An accurate recording system and its use in breath sounds spectral analysis. J. Appl Physiol 1982; 55: 1120-1127.

Charbonneau, G., Sudraud, M., Soufflet, G. Method for the evaluation of flow rate from pulmonary sounds. Bull Eur Physiopathol Respir 1987; 23: 265-270.

Charbonneau, G., Ademovic, E., Cheetham, B.M.G., Maimberg, L.P. Vanderschoot, J., Sovijarvi, A.R.A. Basic techniques for respiratory sound analysis. European Respiratory Review 2000; 10: 77, 625-635.

Cheetham, B.M.G., Charbonneau, G., Giordano, A., Helisto, P. Vanderschoot, J. Digitization of data for respiratory sound recordings. European Respiration Review 2000; 20: 77, 621-624.

Dalmay, F., Antonini, M.T., Marquet, P., Menier, R.. Acoustic properties of the normal chest. European Respiration Journal 1995; 8: 1761-1769.

Dorland's Illustrated Medical Dictionary. 26th ed. Philadelphia, Pa: WB Saunders Co; 1981.

Earis, J.E., Cheetham, B.M.G. Current methods used for computerized respiratory sound analysis. European Respiration Review 2000; 10: 77, 586-590.

Earis, J.E., Cheetham, B.M.G. Future perspectives for respiratory sound research. European Respiration Review 2000; 10: 77, 641-646.

Gavriely, N., Palti Y., Alroy, G. Spectral characteristics of normal breath sounds. J Appl Physiol 1981; 50: 307-314

Gavrieli, N., Nissan, M., Cugell, D., Rubin, A. Respiratory health screening using pulmonary function tests and lung sound analysis. European Respiration Journal 1994; 7: 35-42.

Gavriely, N., Nissan, M., Rubin, A.H.E., Cugell, D.W. Spectral characteristics of chest wall breath sounds in normal subjects. Thorax 1995; 50: 1292-1300.

Golabbakhsh, M., Moussavi, Z., Aboofazeli, M. Respiratory flow estimation from tracheal sounds by adaptive filters. Proceedings of the IEEE Engineering in Medicine and Biology 27th Annual Conference Shanghai, China, 2005.

Gross, V., Dittmar, A., Penzel, T., Schuttler, F., Von Wichert, P. The relationship between normal lung sounds, age and gender. American Journal of Respiratory and Critical Care Medicine 2000; 162: 905-909.

Jones, A., Jones, R.D., Kwong, K. Burns, Y. Effect of positioning on recorded lung sound intensities in subjects without pulmonary dysfunction. Physical Therapy 1999; 79: 7, 682-690.

Kraman, S., Wang, P. Airflow-generated sound in a hollow canine airway cast. Chest 1990; 97: 461-466.

Krumpe, P.E., Cummiskey, J.M. Use of laryngeal sound recordings to monitor apnea. A. Rev. Respir. Dis. 1980; 122: 797-801.

Murphy R.L. Jr, Sorensen, K. Chest auscultation in the diagnosis of pulmonary asbestosis. J. Occup Med 1973; 15: 272-276.

Ohio State University (OSU) Medical Center. http://medicalcenter.osu.edu/patientcare/healthinformation/diseasesandconditions/respiratory/about/anatomy/

Ott, R.L., Longnecker, M. An introduction to statistical methods and data analysis. Pacific Grove, CA, Duxbury 2001.

Pasterkamp, H., Fenton, R., Tal, A., Chernick, V. Tracheal vs lung sounds in acute asthma. Am Rev Respir Dis 1984; 129: 256A.

Pasterkamp, H., Carson, C., Daien, D., Oh, Y. Digital respirosonography. New images of lung sounds. Chest 1989; 96: 1405-1412.

Pasterkamp, H., Schafer, J., Wodicka, G.R. Posture dependent change of tracheal sounds at standardized flows in patients with obstructive sleep apnea. Chest 1996; 110: 1493-1498.

Pasterkamp, H., Kraman, S.S., Wodicka, G.R. Respiratory sounds: Advances beyond the stethoscope 1997. American Journal of Respiratory and Critical Care Medicine 1997; 156: 3, 974-987.

Piirila, P., Sovijarvi, A.R.A. Crackles: recording, analysis and clinical significance. European Respiration Journal 1995; 8: 2139-2148.

Pourazad, M.T., Moussavi, Z., Farahmand, F., Ward, R.K. Heart sounds separation from lung sounds using independent component analysis. Proceedings of the IEEE Engineering in Medicine and Biology 27th Annual Conference Shanghai, China 2005.

Ployosongsang, Y., Iyer, V.K., Ramamoorthy, P.A. Reproducibility of the vesicular breath sounds in normal subjects. Respiration 1991; 58: 158-162.

Que, C., Kolmaga, C., Durand, L.G, Kelly, S.M., Macklem, P.T. Phonospirometry for noninvasive measurement of ventilation: methodology and preliminary results. J. Appl Physiol 2002; 93: 1515-1526.

Rezek, I.A., Roberts, S.J. Envelope extraction via complex homomorphic filtering. TR-98-9 1998; 1-8.

Rossi, M., Sovijarvi, A.R.A, Piirila, P., Vannuccini, L., Dalmasso, F., Vanderschoot, J. Environmental and subject conditions and breathing maneuvers for respiratory sound recordings. European Respiration Review 2000; 10:77, 611-615.

SAS Institute Inc. JMP Introductory guide. Cary, N.C. 2005

Sovijarvi A.R.A, Malmberg, L.P, Charbonneau, G., Vandershoot, J., Dalmasso, F., Sacco, C., Rossi, M., Earis, J.E. Characteristics of breath sounds and adventitious respiratory sounds. European Respiration Review 2000; 10:77, 591-596

Sovijarvi, A.R.A., Vanderschoot, J., Earis, J.E. Standardization of computerized respiratory sound analysis. European Respiration Review 2000; 10: 77, 585.

Sovijarvi, A.R.A., Daimasso, F., Vanderschoot, J., Malmberg, L., Righini, G., Stoneman, S.A.T. Definitions and terms for applications of respiratory sounds. European Respiration Review 2000; 10: 77, 597-610.

Van Den Berg, J.W. An electrical analogue of the trachea, lungs and tissues. Acta Physiol. Pharmacol 1960; 9: 361-385.

Vannuccini, L., Earis, J.E., Helisto, P., Cheetham, B.M.G., Rossi, M., Sovijarvi, A.R.A, Vanderschoot, J. Capturing and preprocessing of respiratory sounds. European Respiratory Review 2000; 10: 77, 616-620.

VITA

Ranjani Sabarinathan was born on May 7, 1981 in Kuwait and is a Canadian citizen. She graduated from Halifax West High School, Halifax, Nova Scotia, Canada in 1999. She received her Bachelor of Science degree in Electrical Engineering from Queen's University, Kingston, Ontario, Canada in 2004. She is currently finishing her Master of Science in Biomedical Engineering in Virginia Commonwealth University.