SPECTRAL ANALYSIS OF ACOUSTIC RESPIRATORY SIGNAL WITH A VIEW TO DEVELOPING AN APNOEA MONITOR

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Publications and Abstracts

During the course of my study the following paper has been presented at a learned society:

An oral presentation "Spectral Analysis of Acoustic Respiratory Signal with a view to developing an Apnoea Monitor."

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Australasian College of Physical Scientists and Engineers in Medicine and Biomedical Engineering Conference, Sept. 20-23 1993, Melbourne.

Declaration

I declare that this thesis is a record of original work and that it contains no material which has been accepted for the award of any other degree or diploma at any University.

To the best of my knowledge and belief, this thesis contains no previously published or written work by any other person, except where due reference is given in the text of the thesis.

I consent to this thesis being made available for photocopying or loan.

Ámit Ajmani

1993

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SUMMARY

Apnoea monitors detect the cessation of breathing in infants and adults who are at risk of respiratory failure. An apnoea monitor serves to alert an attendant when life threatening episodes occur. Current monitors employ one or a combination of methods to detect breathing and the absence of breathing.

The aim of this thesis is to develop a reliable apnoea monitor of respiration using acoustic signals acquired with a small transducer from the body surface. In the present study, acoustic signals were acquired from the upper neck position adjacent to the larynx with the aid of a transducer. An appropriate band pass filter was developed after analyzing the frequency distribution of the respiratory signal, eliminating the high and low frequency components (noise) of the signal. The filtered and the raw signals were recorded in analogue form on an FM tape-recorder. Subsequently, the signals were digitized and then sampled and stored on a hard disk. The digitized real-time signals were transformed into frequency domain using the fast Fourier Transform (FFT) algorithm. Spectral analysis of these acoustic signals was conducted to study the signal characteristics. Further, the study was extended to analyze the interference of other body noises such as speech, swallowing and snoring in the conventional approach.

An apnoea monitor was designed and developed based on the discerned spectral parameters. The monitor has been tested on a range of adult subjects with simulated apnoea in low noise conditions at the Bio-engineering laboratory of the Department of Electrical and Electronic Engineering.

The following is a brief synopsis of the thesis:

CHAPTER 1 This gives an introduction to the clinical significance and the scope of research.

CHAPTER 2 In this chapter, signals and their types has been discussed.

CHAPTER 3 This chapter discusses the types of transducers used and the instrumentation system used for the research.

CHAPTER 4 In this chapter, spectral analysis of respiratory signals and some body noises has been discussed.

CHAPTER 5 The design and operation of an Apnoea monitor are discussed in detail in this chapter.

CHAPTER 6 Test results and scope of further research are delineated in this final chapter. This chapter also contains conclusions of the research done in the thesis.

The work of the thesis was completed in eighteen months on the basis of two-third thesis and one-third coursework.

CHAPTER 1

GENERAL INTRODUCTION

1.1 Apnoea

Apnoea can be defined as cessation of breathing or the state of being unable to draw breath (Butterworth's Medical Dictionary). Apnoea can be classified as central, obstructive, and mixed. Central apnoea occurs when the brain stops sending signals to the respiratory muscles, even though the patient apparently has normal physiologic respiratory mechanisms. Obstructive apnoea occurs when airflow is cut-off because of physiologically induced upper-airway obstruction. Mixed apnoea has the elements of both patterns. Although there are definable patient groups which are at risk of respiratory failure, no clear underlying pathology regarding irregular breathing disorders has been discovered.

1.2 Sudden Infant Death Syndrome

Sudden infant death syndrome (SIDS) has been defined in 1989 by the United States' National Institute of Health as "the sudden unexpected death of any infant under one year of age whose death remained unexplained after a complete post-mortem examination, including investigation of the death scene and review of the case history."

Epidemiological studies have found certain maternal, infant and environmental characteristics to be associated with a higher risk of an infant dying of SIDS. Some of the following factors are outlined as the characteristics associated with a higher risk of SIDS (Ponsonby, 1992).

a) Maternal factors

b) Infant factors

- Low birth infants
- Premature births
- Prone sleeping position

c) Environmental factors

1.3 Role of Apnoea in SIDS

Evidence supporting a role for abnormal apnoea in sudden infant deaths is predominantly circumstantial. Investigations of infants having cyanotic episodes have shown four mechanisms for abnormal apnoea. The most frequent, prolonged expiratory apnoea, is particularly dangerous because it is associated with the rapid onset and progression of hypoxemia and other features, suggesting an alveolar ventilatory-perfusion mismatch. Seizure-induced apnoea, maternally imposed obstructive apnoea, and sleep-related upper airway obstructive apnoea may also produce severe hypoxemia. Attempts to identify infants at risk of sudden death using measurement of cardiorespiratory variables have to date had limited success. Nevertheless, techniques for more precise, yet non-invasive, monitoring of the respiratory function, including oxygenation, have recently been developed. Such attempts to identify "at risk" infants must continue. Epidemiologic and pathologic studies have provided considerable support for the "abnormal apnoea hypothesis" and need to be integrated with studies on the physiology of living infants. On the basis of the presently available evidence

concerning prolonged apnoea, it may be possible to prevent a significant proportion of sudden infant deaths by reducing the exposure of young infants to respiratory tract infections and by improving the prenatal environment.

(a) Absence of Inspiratory Efforts

Short pauses in inspiratory efforts occur in all healthy infants. Sometimes these pauses are isolated and sometimes they are a part of periodic breathing. They occur most frequently during rapid eye movement sleep.

It is important to be aware of the length and frequencies of such pauses in healthy infants. Knowledge of postnatal age is crucial in determining normal values. The study of an adequate number of infants at each age is also important because the frequency distribution of pauses is highly skewed. Although the frequency and length of pauses are positively correlated in normal infants, no study has yet shown that individual infants with abnormally prolonged pauses have increased numbers of short pauses or increased quantities of periodic breathing.

(b) Partial or Complete Airway Obstruction

Severe upper airway obstruction occurs during sleep in otherwise apparently healthy infants. This obstructive apnoea may be complete or partial, although in either case it is being accompanied by hypoxemia and arousal. Obstructive events usually occur during active sleep (Southhall, 1988).

Medical attention for the life-threatening events is rarely sought for infants with obstructive sleep. Most often they are diagnosed for inspiratory strider, chest wall recession, sleep disruption, and failure to thrive. These readily apparent clinical features are rarely identified from retrospective inquiries following sudden expected infant deaths (Southhall, 1988). It is unlikely, therefore, that these disorders are responsible for many instances of SIDS.

1.4 Periodic/Abnormal Breathing

An abnormality of respiration called periodic breathing occurs in a number of different disease conditions. In periodic breathing, the person breathes deeply for a short interval of time and then breathes slightly or not at all for an additional interval. This cycle then repeats itself (Guyton, 1991).

The most common type of periodic breathing, Cheyne stokes breathing, is characterized by slowly waxing and waning respiration, which reoccurs approximately every forty to sixty seconds.

Factors that affect respiration are as follows:

(a) Voluntary Control of Respiration

Respiration can be controlled voluntarily, and one can hyperventilate or hypoventilate to such an extent that serious dearrangements in carbon dioxide pressure (Pco_2), pH and arterial blood oxygen pressure (Po_2) can occur in the blood.

(b) Anaesthesia

The most prevalent cause of respiratory depressions and respiratory arrest is overdosage of anaesthetics or narcotics.

(c) Effect of Brain Edema

The activity of the respiratory centre may be depressed or even totally inactivated by acute brain edema resulting from acute concussion. Respiration depression thereafter may result.

1.5 Patient Monitoring

Monitoring provides a continuous watch over the vital characteristics and parameters of critically ill patients. Hospitals' intensive care units have saved lives in recent years

because of the careful and accurate monitoring afforded by monitoring equipment. With the growing dependency on monitors, there is a need for a reliable monitoring system. Recent advances in technology may very well be able to provide the reliability necessary for effective monitoring systems.

Essentially, patients are monitored because their body systems are imbalanced. By continual monitoring, the patient's problems can be detected as they occur and remedies taken before these problems become acute.

1.6 The Apnoea Monitor and Its Functions

An apnoea monitor is a device used to detect either the cessation of breathing or prolonged pauses while breathing. The use of apnoea monitors has been advocated as one way of preventing SIDS in an infant deemed to be at 'increased risk'. There is no doubt that an apnoea monitor both provides some parents with a sense of security and also gives them the feeling that should their baby die, at least they have done all they could to prevent the death.

Unfortunately, effective monitors are expensive items that are complicated to use. Parents who purchase monitors also must spend time to learn their correct use. To date there are no simple, comfortable, reliable, non-invasive breathing monitors. Current monitors employ one method or a combination of methods to detect breathing and the absence of breathing. The most commonly employed non-invasive techniques include impedance pneumography, pressure pads, pneumatic abdominal sensors, thermistors, proximal airway pressure sensors and carbon dioxide sensors (Penzel et al., 1991; Lue et al., 1992).

Most monitors either do not detect all the episodes of apnoea (especially obstructive apnoea) or give false indications of apnoea when it is not occuring. Most monitors therefore, are not reliable, giving both false positive and false negative signals. Sleep

laboratories, which exist primarily to monitor the cause, frequency, duration and severity of apnoeic episodes during sleep, therefore employ a multitude of techniques with which to monitor a child's breathing. These techniques include remote visual observation by closed circuit television, listening for snoring via microphones, constant measurement of inspired and expired air temperature by thermistors placed beneath the nostrils and over the mouth, and impedance pneumography. Even in such an environment, continual observation may be required to provide a reliable report. The requirement of an easy to use, reliable, and inexpensive apnoea monitor has led to the scope of this thesis.

CHAPTER 2

BIOLOGICAL SIGNALS

2.1 Signals

In signal processing, a signal refers to physical properties that change with time, such as electromagnetic waves. These signals are converted by a transducer into another form (usually electrical) that can more easily be manipulated or processed. An example of a transducer is a microphone, which converts sound-pressure variations into a voltage that varies proportionately with the sound pressure.

The two basic parameters for signals are the frequency and the amplitude. The frequency of a signal refers to the number of times that the signal varies per second (measured in Hz). Amplitude is a measure of the strength of a signal. A signal characteristic can change depending on the frequency and amplitude. A signal can consist of many separate components that propagate at different frequency and amplitudes. Such a signal is called a composite signal. Another important signal parameter is bandwidth, which describes the range in frequencies of a complex signal.

(a) Digital and Analogue Signals

Historically, most electronic signal processing has been done on analog signals with analogue components (e.g. transistors, transformers, and capacitors). However, it is

difficult and costly to create analog signal-processing components with precise values that do not change significantly with temporal and environmental fluctuations. Digital electronics allows us to compute the effect of components on a signal using mathematical operations. Digital signal processing is a precise method unaffected by temporal or environmental changes.

Digital processing speeds have increased with the advent of digital signal processing chips. However, these chips are not practical in an analog domain. This is due to a number of factors. First, unlike normal electrical components, digital components operate precisely as expected, with no losses, distortion, or other negative physical effects to the accuracy of the numeric representation used. Second, mathematical operations representing component functions that are not physically realizable are easily computed. Finally, because digital components are programmable, one can change the signal processing function by re-programming it rather than by resorting to a soldering gun.

(b) Sampling and Quantization

Before any digital computer can process signals, the analog signal must be converted into a digital signal. A digital signal has both discrete time and discrete amplitude. The first step of the conversion is accomplished by a process called sampling, which converts a continuous signal to a discrete time signal. The second step in converting an analog signal to a digital signal is called quantization, which converts a continuous amplitude value into a discrete amplitude value.

The most common type of quantization is called uniform quantization. Uniform quantization simply places the analog value of each sample into one of a set of possible bins. Quantizing the signal loses information and introduces quantization noise into the signal. The greater the number of bins, however the less noise is introduced. The result is that a 16 bit audio system sounds significantly better than an 8 bit system. The

process of sampling and quantization is called analogue to digital (A/D) conversion. The converter normally operates at a fixed sampling rate, measuring and quantizing the value of the signal once per sample period. These values are passed onto the digital system for storage and processing.

When designing a digital signal-processing system, an important consideration is collecting the sampling rate. For example, in a band-limited signal of frequencies below "f Hz", the input signal can be reconstructed from the sampled signal, provided the input signal is sampled at least "2f" times per second. Frequency "f" is called the Nyquist frequency. Sampling the signal below the Nyquist rate (sub-Nyquist sampling), results in aliasing: the sample points do not contain enough information to reconstruct the original signal. Aliasing causes frequencies in the input signal above the Nyquist frequency to generate undesirable frequencies in the digital signal. These frequencies form a mirror image around the Nyquist frequency. To ensure that aliasing does not occur, one must filter the signal to remove any components above the Nyquist frequency before it is converted to the digital domain.

2.2 Biological Signals

The living biological system governed by biochemical, physical, and chemical laws presently is not well understood. In particular, many aspects of the complex hierarchical control, the neural information transfer and processing, and other systems are still under extensive investigation. Very often we use prior information concerning the system to generate the signal of interest in order to assist in the analysis and processing procedures.

The complexity of the biological system often introduces difficulties in the measurement and processing procedures. Unlike physical systems, the biological system often cannot be uncoupled in such a way that subsystems can be monitored and investigated individually. Because of the complex hierarchical control linkages among

subsystems, and due to many feedback paths not always well understood, the biological system under investigation must remain in its natural environment during observation. The signals produced by the system are thus influenced directly by the activity of the surrounding systems. The signal is also inherently contaminated by noise produced by the neighbouring systems.

In carrying out their various functions, certain systems of the body generate their own monitoring signals, which convey useful information about the function they represent. These signals are the bioelectric potentials associated with functions as nerve conduction, brain activity, heartbeat and muscle activity.

Biological signals can be divided into groups according to inherent characteristics.

I. Bioelectric Signals

Types of bioelectric signals are as follows:

a) Action Potential

This is the potential generated by the excitable membrane of a nerve or muscle cell.

b) Electroencephalogram (EEG)

The recording of electrical activity (both spontaneous and evoked) of the brain is known as electroencephalography.

c) Electromyography (EMG)

Electromyography is the recording of electrical potential generated by the muscle.

d) Electrocardiography (ECG, EKG)

An ECG is the recording of the electric activity of the heart. Bioelectric potential is generated at the sino artrial (SA) of the right artrium which is transmitted to the heart muscles.

II. Impedance

There are two types of impedance that are commonly considered.

a) Impedance Plethysmography

The use of impedance changes for the recording of peripheral volume pulses is known as impedance plethysmography.

b) Impedance Pneumography

This method involves measurement of transthoractic impedance for monitoring respiration on the surface of the chest.

III. Acoustic Signals

Acoustic signals can be classified as follows:

a) Phonocardiography

Phonocardiography is the recording of sounds generated by heart/great vessels using non-invasive techniques.

b) Auscultation

The monitoring of sounds heard over the chest wall is known as auscultation. During respiration, gases flow through the various airways emitting acoustical energy.

c) Voice

Speech is produced by expelling air from the lungs through the trachea to the vocal cords.

Apart from these three types of biological signals, one may also consider:

IV. Mechanical Signals

V. Biomagnetic Signals

VI. Biochemical Signals

VII. Two- Dimensional Signals

2.3 Respiratory Signals

Air enters the lungs through air passages which include the nasal cavity, pharynx, larynx, trachea, bronchi and bronchioles. The movement of air through the respiratory tract may be utilized as an indication of respiratory signals. The respiratory signals can

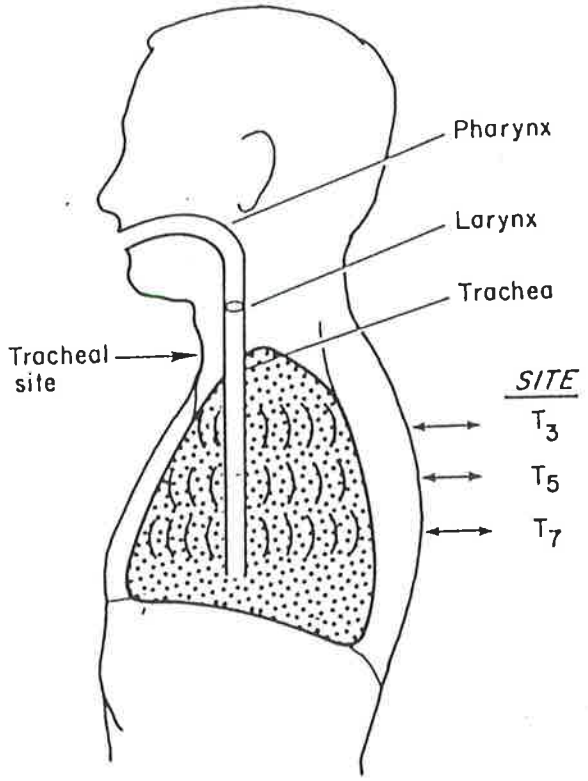


Fig. 2.0 Diagram of the model respiratory system. The respiratory tract is represented as a single effective tube which acts as a cylindrical sound source.

non-invasively be acquired as acoustic signals, as a change of impedance (transthoractic or peripheral volume), or a change of temperature of the exhaled air.

Respiratory acoustic signals may be acquired from different positions on the body. For example, they may be taken over the larynx (voice box), the trachea (windpipe), or over the upper (appical) or lower (basilar) chest. Figure 2.0 shows some of the locations that may be chosen for acquiring acoustical signals from the human body.

CHAPTER 3

INSTRUMENTATION AND SIGNAL ACQUISITION

3.1 Sensors

Data from the outside world is usually acquired through sensors. Depending upon how accurately one wants the system to be gauged, the sophistication of the sensors increases and the relationship is usually not linear or simple. Sensors therefore are an important part of the acquisition system and, in providing an interface between the system and the outside world, have to comply with the requirements of both environments. This places stringent requirements on the chemical, mechanical, and electrical characteristics of the sensors, especially in corrosive, harsh, or delicate environments. These conditions are encountered in industrial and biomedical applications. Hence, a reliable and systematic characterization of the sensor is of utmost importance.

Reliability and long-term stability of transducers are becoming increasingly important factors as transducers are widely employed in various applications. The use of transducers in biomedical applications make their reliability and long-term stability especially important (Gopel et al., 1989). Sensors can be broadly characterized by their

static and dynamic characteristics. The static characteristics include accuracy, precision, resolution, sensitivity, threshold, nonlinearity, distortion, conformance, hysterisis, noise, isolation and output impedance. The dynamic characteristics are transfer function, frequency response, impulse response, and step response.

3.2 Transducer Choice

From a functional stand-point, transducers may be classified as passive or active, according to whether the output signal energy is supplied by the input. Thus, an active transducer is one in which the output energy is supplied primarily by an auxiliary power source controlled by the input signal. In contrast, a passive transducer is one in which the output signal is independent of input energy.

In designing a biomedical measurement that uses a transducer, one basic decision concerns the choice of either invasive or non-invasive techniques. When this decision has been made, it is possible to consider the factors entering into the choice and design of a suitable transducer. Although some of these factors are difficult to isolate and categorize, most may be lumped under the three headings shown in the Figure 3.1 (Cobbold, 1974). These headings are economic, signal, and environmental factors. On the basis of system cost, mean time between failure (MTBT) and the size, a somewhat inferior transducer may in fact prove to be the best from the overall viewpoint, even though it may result in some loss of information.

3.3 Sensor Resolution

Resolution of a sensor measures its ability to detect a change in the sensed quantity, and is usually quoted in terms of the smallest change that can be detected (Sinclair, 1988). An infinite resolution would mean that a small change in the sensed quantity will

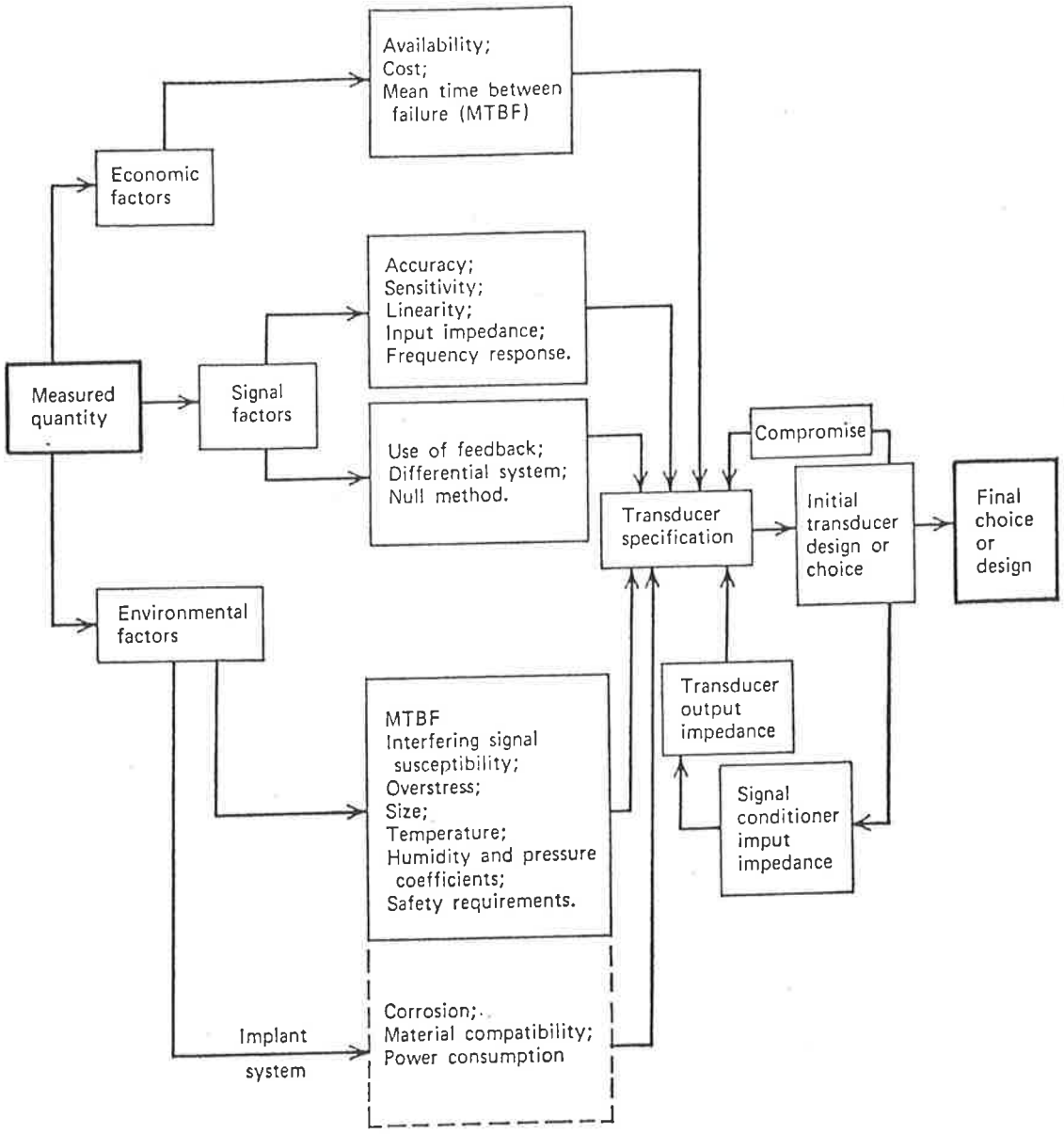


Fig. 3.1 Factors influencing the choice and design of an input transducer for biomedical measurements (Cobbold, 1974).

cause a small change in the electrical output. These changes are detectable to the limits of our measuring capabilities. Though few sensing methods provide a direct digital output, most digital outputs are obtained by converting from analogue quantities. In other words, the limits of resolution are determined by the A/D conversion circuits rather than by the sensor itself.

The sensing of any quantity is liable to error, and the errors may be static or dynamic. Although static error is caused by reading problems, dynamic error is caused by the difference between the actual quantity and the amount that is measured. This difference caused by the loading of the measuring instrument itself. All forms of sensors are liable to dynamic errors if they are used only for sensing. Two measurable quantities, the responsivity and the detectivity, may be quoted in connection with any sensor or transducer. The responsivity is defined as:

$$\frac{\text{Output Signal}}{\text{Input Signal}}$$

which will be a measure of transducing efficiency. Similarly, the detectivity is defined as :

$$\frac{\text{S/N of Output Signal}}{\text{Size of Input Signal}}$$

where S/N is the signal-to-noise ratio.

3.4 Audio to Electrical Transducers

The conversion of sound to an electrical energy transducer is done via the microphone. Microphone types are classified by the transducer they use. In addition to the transducer, a microphone will use acoustic filters and acoustic passages. The shape and dimensions of these acoustic passages modify the response of the overall system. Acoustic filters and acoustic passages are needed because each transducer will have its own response, which is determined by resonances in the materials as well as by the

transducing principle itself. Any correction to a more uniform response must be made by means of the acoustic passages in the microphone housing.

The characteristics of a microphone are both acoustic and electrical. The overall sensitivity is expressed either as millivolts or microvolts of the electrical output per unit intensity of sound wave, or in terms of the acceleration produced by the sound wave. In addition, the impedance of the microphone is of considerable importance. A microphone with high impedance usually has a high electrical output, but the high impedance makes it very susceptible to hum pick-up, either magnetically or electrostatically coupled. A low impedance is usually associated with very low output, but hum pick-up is almost negligible.

Another factor of importance is whether the microphone is directional or omni-directional. If the operating principle of the microphone is the sensing of the sound wave, then the microphone will be omni-directional, picking up sound arriving from any direction. In contrast, if the microphone responds to the velocity of the sound wave, then it is a directional microphone. In case of signal acquisition from a surface in contact with the diaphragm of the microphone, the microphone primarily responds to the sound waves coming from that surface, depending upon the sound's intensity. However, if the microphone is left exposed to the environment, the microphone will be susceptible to pick-up from all directions. The type of transducer does not necessarily determine the operating principle as velocity or pressure, because the acoustic construction of the microphone is usually a more important factor.

Condenser Microphones

The outline of a condenser microphone is illustrated in Figure 3.2. The amount of electric charge between the two surfaces is fixed, and one of the surfaces is a diaphragm which can be vibrated by a sound wave. The vibration causes a variation of capacitance which, because of the fixed charge, causes in turn a voltage wave. The output

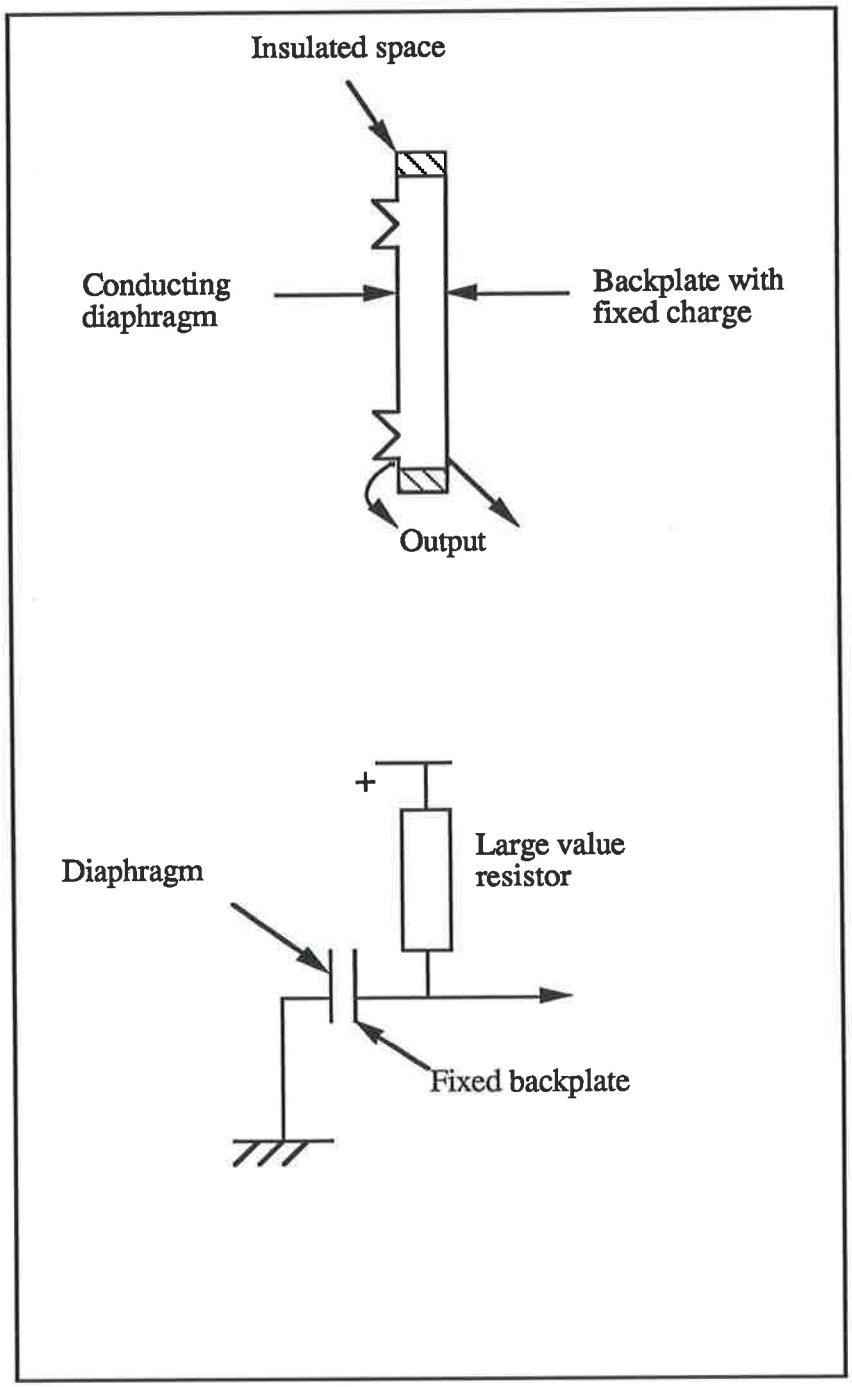


Fig. 3.2 The capacitor microphone principle

impedance is very high, and the amount of output depends on the normal spacing between the plates. The smaller this spacing, the greater the output for a given amplitude of sound wave. The construction of the microphone ensures that it is always pressure operated. The capacitor microphone can be linear in operation and therefore can provide good quality audio signals without the need for elaborate constructional techniques.

A block diagram of the total instrumentation involved in signal acquisition and signal analysis is shown in Figure 3.3. The signals acquired from the body are amplified and filtered to the desired frequency range. These analogue signals can be viewed on a cathode ray oscilloscope (CRO) in real time or can be recorded on an FM tape recorder for analysis. The selected recording segments of the signals can be digitized and stored as files, or can be processed to study their frequency characteristics.

3.5 Transducer

Acoustic signals are acquired from the surface of the body with the aid of the diaphragm of a condenser microphone. The condenser microphone originally was a part of an electronic stethoscope having a flat frequency response up to 1KHz. A condenser microphone was chosen primarily because a piezo-electric crystal is not suitable for detecting very low intensity signals such as breath sounds. The piezo-electric is not suitable because of the internal resistance of the crystal which results in poor signal-to-noise ratio (Heemels et al., 1986). A microphone from the electronic stethoscope also was chosen because it had a suitable coupling with the skin and good sensitivity (Charbonneau et al., 1983). The microphone is 25mm in diameter and light in weight, making it suitable for recording adult subjects.

A secondary condenser microphone (which essentially is for noise cancellation) was suspended in air and placed next to the principle microphone. A schematic diagram is shown in the Figure 3.4. The microphones are driven by a small (1mA) current source.

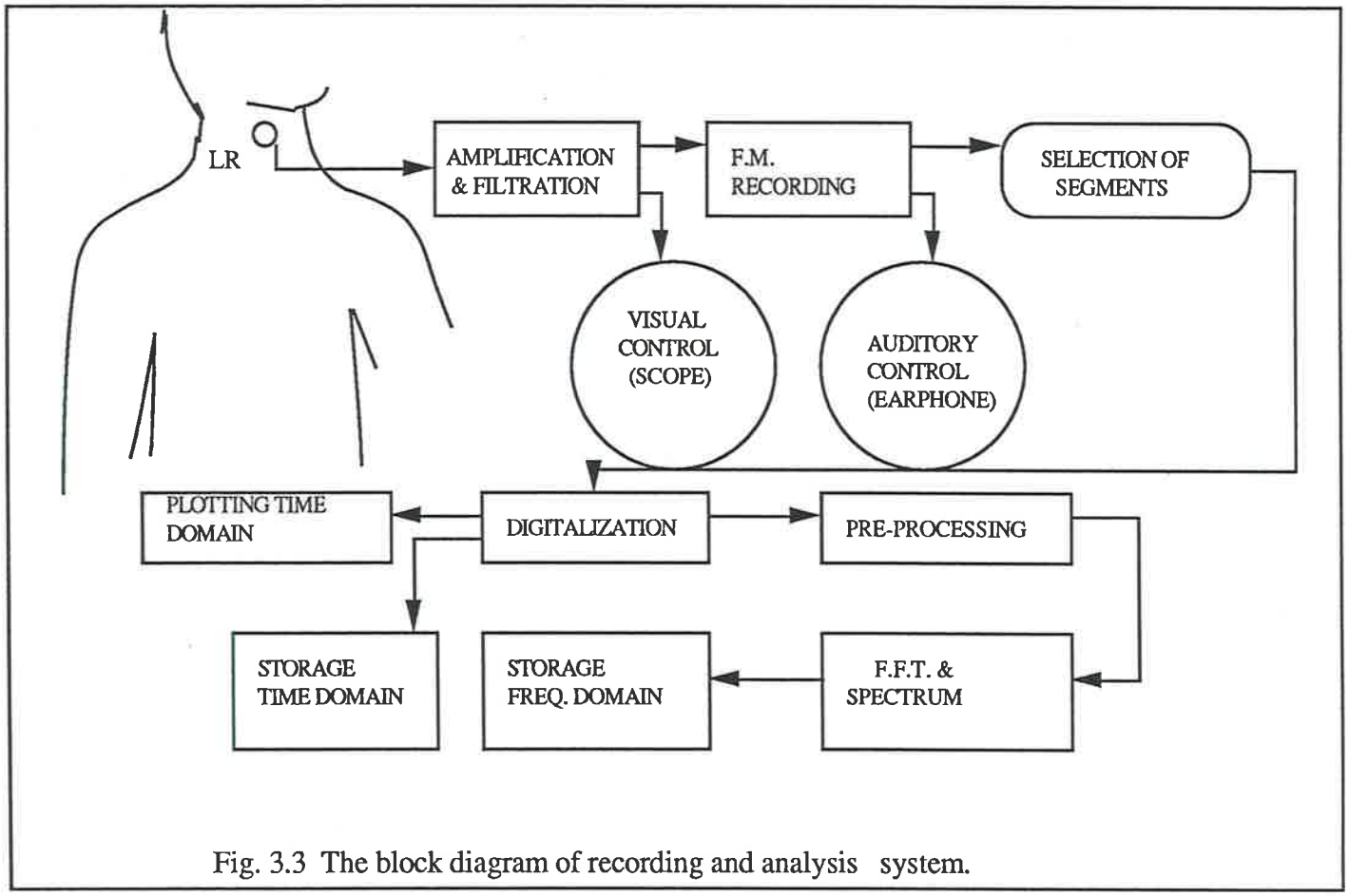


Fig. 3.3 The block diagram of recording and analysis system.

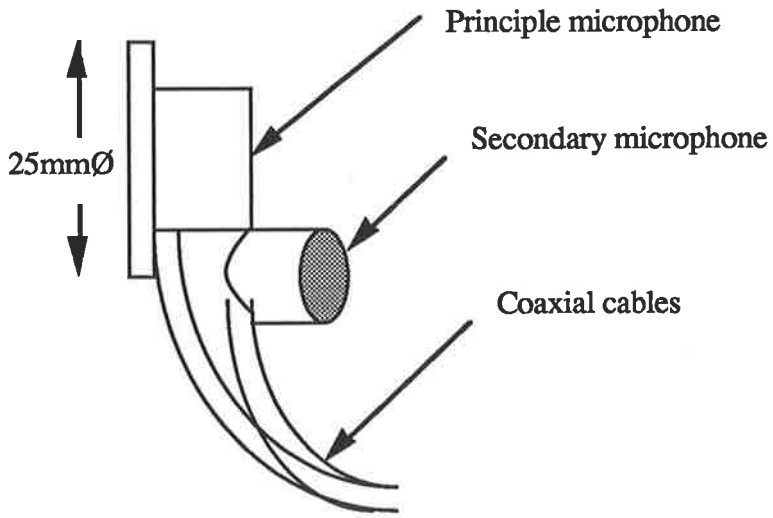


Fig. 3.4 Principle and secondary microphones

Coaxial leads of optimum length (to avoid noise pick-up) carry the output of the microphone to the amplifier circuit.

3.6 Amplifier

The acquired signal is transmitted by a twin core coaxial cable to the amplifier circuit. The signals (millivolts) from the two microphones are fed to a precision instrumentation amplifier LM363H, and the output is transmitted further for post-amplification. Figure 3.5 shows a schematic circuit diagram of the precision amplifier where inputs V_{in1} and V_{in2} are the signals from the two microphones and where the output V_2 is the amplifier output. The output from the precision amplifier is post-amplified using LF373 Opamp with a stage gain. The schematic circuit of the post-amplifier is shown in the Figure 3.6. The amplifier circuit has a frequency response in the range of 2 Hz - 2 KHz, making it suitable for the required application.

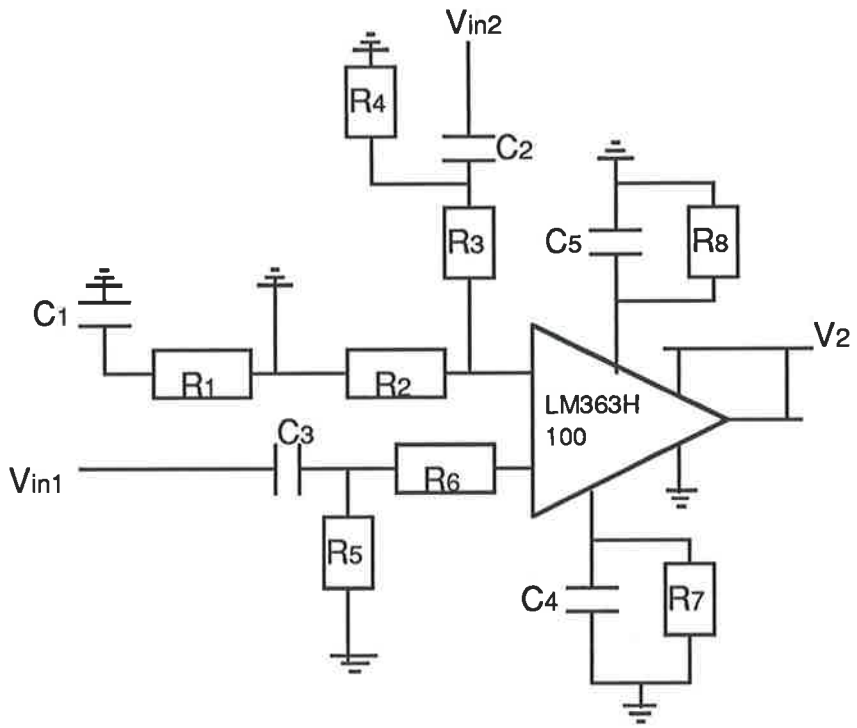


Fig. 3.5 Precision Instrumentation Amplifier

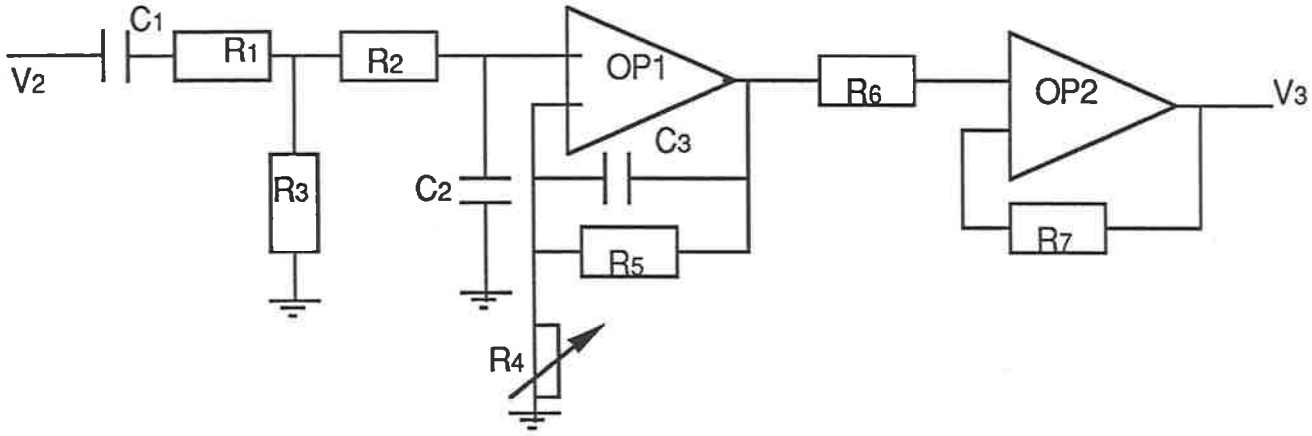


Fig. 3.6 Post- Amplifier Circuit

3.7 Filter

A band-pass filter passes the entire signal lying within a band between a lower and an upper frequency limit and essentially rejects all other frequencies that are outside this specified band. A generalized band-pass response curve is shown in Figure 3.7. The bandwidth is defined as the difference between the upper critical frequency (fc_2) and the lower critical frequency (fc_1).

$$BW = fc_2 - fc_1$$

The critical frequencies are the points at which the response curve is 70.7 percent of its maximum. The frequency about which the pass band is centred is called the centre frequency f_0, defined as the geometric mean of the critical frequencies.

$$f_0 = \sqrt{fc_1 fc_2}$$

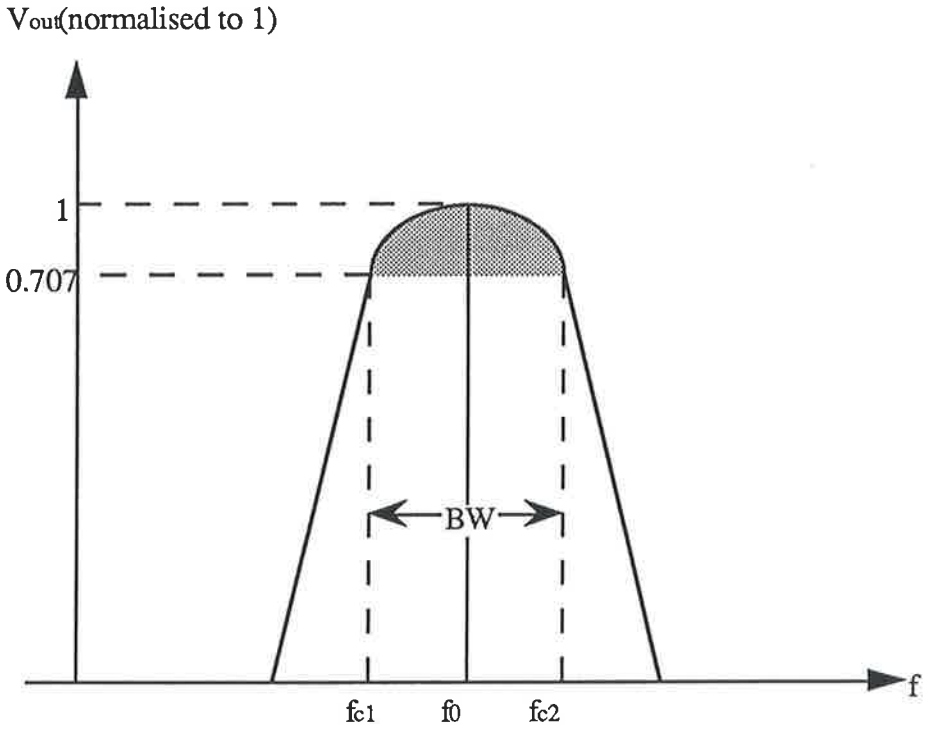


Fig. 3.7 General band-pass response curve

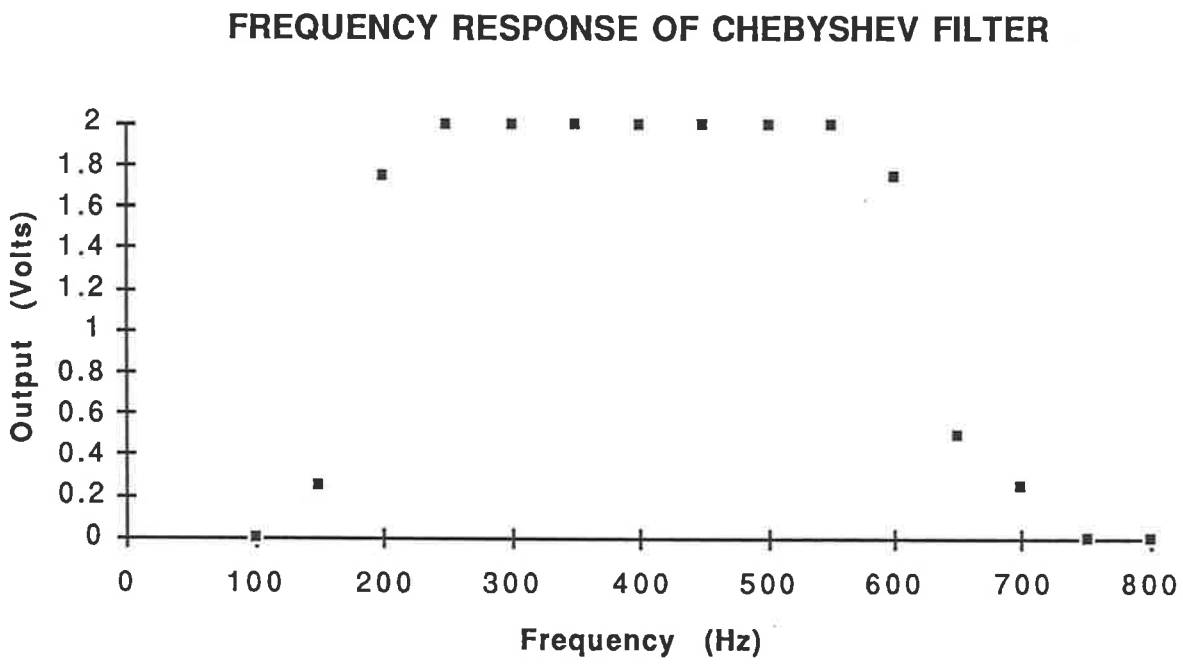

FREQUENCY RESPONSE OF CHEBYSHEV FILTER
Output (Volts)
2
1.8
1.6
1.4
1.2
1
0.8
0.6
0.4
0.2
0
0
100
200
300
400
500
600
700
800
Frequency (Hz)

Fig. 3.8

An analogue Chebyshev band-pass filter was developed with a frequency response of 150 Hz - 600 Hz and an attenuation of about 40dB/decade in the cutoff region. The filter response, though quite consistent peaks at approximately 400 Hz. Figure 3.8 shows the frequency response of the Chebyshev filter as tested with a signal generator.

The above filter is a fourth order Chebyshev band-pass filter with a 0.5 ripple factor. The design of the filter (pass band range) removes any component above the Nyquist frequencies in order to avoid aliasing. Figure 3.9 shows the schematic circuit of the Chebyshev filter, where the input V_i corresponds to V_3 (the output from the post-amplifier). The filter output V_o is transmitted further for full wave rectification. The output signal from various stages (amplified, filtered and rectified) could be utilized for recording and processing signals for signal analysis.

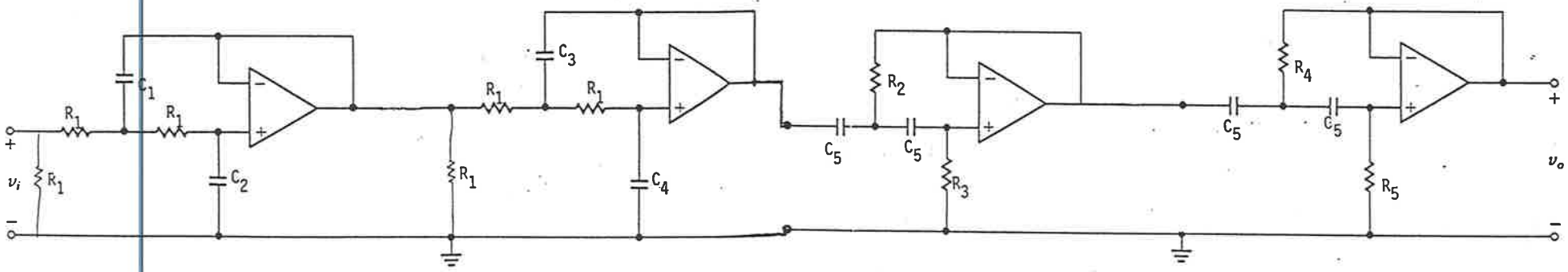


Fig. 3.9 Fourth order chebyshev band-pass filter with 0.5 dB ripple.

3.8 Tape Recorder

A frequency modulation record/replay system is shown in Figure 3.10. The input signal is applied to a voltage/controlled oscillator to which the recording head is connected. The signal recovered on play-back usually is attenuated and must be amplified before being presented to the demodulation circuit. A low pass filter follows the demodulator in order to remove the carrier frequency and side band frequencies generated during the recording process.

Recording at a low speed using a proportional, reduced frequency range, will increase the recording time available from a given magnetic tape length. This makes it possible to record at one tape speed and reproduce at an entirely different tape speed. The change in signal time base obtained in this way represents one of the most important characteristics of the FM recording processes.

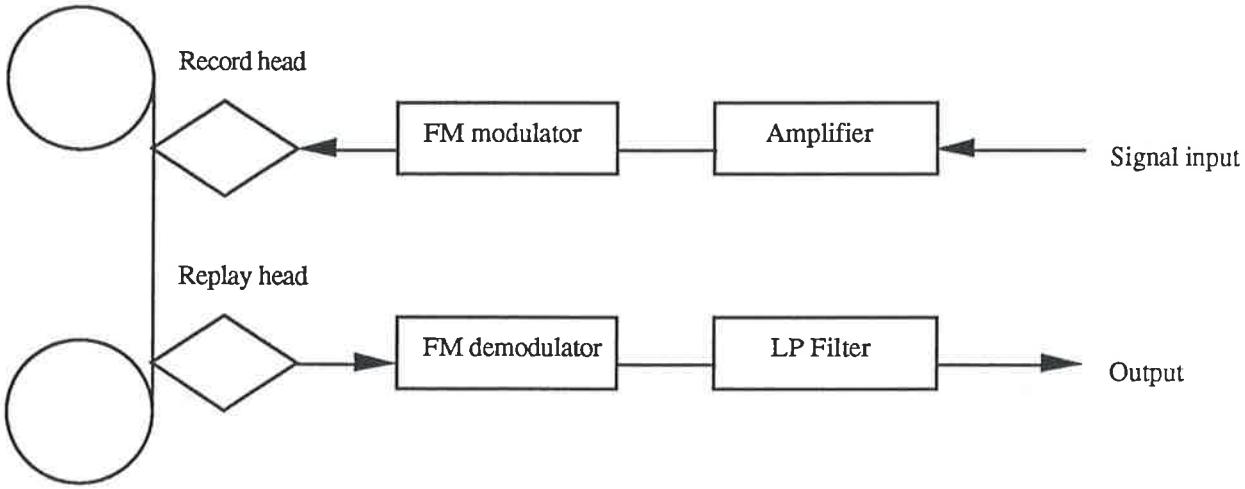


Fig. 3.10 A frequency modulation record/replay system.

A Hewlett Packard 3964A FM tape recorder is used, having bandwidth of 1250 Hz at 3.75 inch-per-second tape speed. The FM tape recorder has a four channel recording/play-back capability which enables one to carry out a comparative study. Three channels are utilized to record the unfiltered, filtered and rectified forms of the acquired signal, while the fourth channel contains the speech describing the signal being recorded. The tape recorder has four variable speeds for recording/play-back, providing flexible means with which to analyze the frequency characteristic of the signal.

3.9 Computer Hardware

An IBM PC230 computer with a PC ADDA-14 FPC-011 converter card for A/D and digital to analogue (D/A) conversion initially was used for signal processing of respiratory signals, providing a 14 bit resolution in the digitized signal.

The later part of the analysis was performed on a Macintosh II computer equipped with a sound acquisition board (Audiomedia) enabling 16 bit resolution and providing flexibility to work with the digitized signal files. This enhanced the capability of editing files and processing different components of the acquired signal.

3.10 Plotter

An analogue plotter (Yokogawa 3036 X-Y recorder) was utilized to plot the frequency spectra computed on the PC230 computer. The plotting time was chosen as 100 seconds, to enable proper plotting of the peaks in the spectra.

3.11 Signal Acquisition

Various aspects of signal acquisition system can be described as follows:

Problems Encountered in Measuring a Living System

One of the greatest problems in attempting measurements from a living system is the difficulty in gaining access to the variable being measured. Sometimes the problem stems from the required physical size of the transducer as compared to the space available for the measurement. Where a variable is inaccessible for the measurement, an attempt is often made to perform an indirect measurement. The problems encountered with signal acquisition are more pronounced in the non-invasive techniques, compromising the variables to be measured. This process involves the measurement of some other related variables which may under certain conditions make a usable estimate of the inaccessible variable.

Variability of Data

Physiological variables never can be viewed as strictly deterministic values, but must be represented by some kind of statistical or probabilistic distribution. In other words, measurements taken under a fixed set of conditions at one time will not necessarily be the same as similar measurements made under the same conditions at another time. The variability of these two measurements will change, and statistical methods must be employed to estimate the relationship among variables.

Effect of the Transducers

The presence of the transducer under normal measurements may affect the response of that system or some other system. The non-invasive techniques employed for measurement have the advantage of minimal effect on the system response.

Artifacts

Artifacts refer to any component of a signal that is extraneous to the variable represented by the signal. Thus, random noise generated within the measuring

instrument, electrical interference (including 50 Hz pick-up), cross-talk, and all other unwanted variations in the signal are considered artifacts. A major source of artifacts in measurement of a living system is the movement of the measuring device. Sometimes these variations are indistinguishable from the measured variable. At other times these variations may be sufficient to obscure the desired information completely.

CHAPTER 4

SIGNAL PROCESSING AND SPECTRAL ANALYSIS

4.1 The Fourier Transform Method

Interest in the Fourier transform technique in biomedical signal analysis has increased significantly, especially after the publication of the 'Fast Fourier Transform' (FFT) algorithm (Cooley and Tukey, 1965). The Fourier transform is a frequency domain representation of a signal in a continuous system, and is described mathematically by the relationship:

$$S(f) = \int_{-\infty}^{\infty} s(t)\, e^{-j2\pi ft}\, dt, \quad \text{where} \qquad (1)$$

- s(t) is the signal in time domain
- S(f) is the Fourier transform of s(t),
  i.e., the signal in frequency domain.

The inverse transform is given by the formula:

$$s(t) = \int_{-\infty}^{\infty} S(f)\, e^{j2\pi ft}\, df \qquad (2)$$

Equation (2) allows the recovery of the function in the time domain from its Fourier transform. For computer implementation of the Fourier transform, the signal must be represented in discrete form. Accordingly, the discrete Fourier transform of a time series s_n having N number of samples is given by :

$$S_k = \frac{1}{N}\sum_{n=0}^{N-1} s_n \, e^{-j2\pi nk/N} \qquad (3)$$

$$\text{for } k = 0, 1, 2,.....N-1$$

where S_k is the k^{th} coefficient of the discrete Fourier transform.

In order to compute the finite discrete Fourier transform of a series of N complex data points, N^2 operations are required. However, with the introduction of FFT, the number of operations has reduced to approximately $N \log_2 N$. Therefore the FFT is simply an algorithm that can compute the discrete Fourier transform much more rapidly than other available algorithms. The algorithm works by first rearranging the input elements into bit-reverse order, and then building up the output transform in $\log_2 N$ iterations.

If we take an N-point sample of the function s(t) at equal intervals and use the FFT to compute its discrete Fourier transform then the periodogram estimate of the power spectrum is defined at N/2+1 frequencies as:

$$P(0) = P(f_0) = \frac{1}{N^2}|S(0)|^2$$

$$P(f_k) = \frac{1}{N^2}[|S_k|^2 + |S_{N-k}|^2] \qquad (4)$$

$$k = 1, 2,.....(N/2-1)$$

$$P(f_c) = P(f_{N/2}) = \frac{1}{N^2}|S_{N/2}|^2$$

where f_c is the Nyquist frequency

and f_k is defined only for zero and positive frequencies.

$f_k = 2f_c k/N \qquad k = 0, 1, 2,.....N/2$

By Parseval's theorem

$$\sum_{k=0}^{N-1} |S_k|^2 = \frac{1}{N}\sum_{n=0}^{N-1} |s_n|^2$$

it is evident that equation (4) is normalized so that the sum of the N/2+1 values of P is equal to the mean squared amplitude of the function s_n.

The programs utilized for carrying out the spectral analysis are attached in Appendix A. The "four1", "realft" and utility routines (array memory allocation) are based on the routines by the same name published in Numerical Recipes in C (Press et al., 1988).

4.2 The Fourier Transform in Respiratory Signal Analysis

The application of FFT in the analysis of biomedical data has been reviewed by Mussel and the fast Fourier transform has found extensive application in the frequency analysis of acoustic respiratory signal (Mussel, 1992; Penzel, 1990; Charbonneau, 1983; Chowdhury, 1981).

In the present respiratory sound analysis procedure, the analog signals were first stored on an FM tape recorder. The signals subsequently were selected, digitized, and analyzed using the FFT algorithm.

The sampled data was windowed by a Hamming type window (0.54 + 0.46 Cosθ). Windows are weighting functions applied to data to reduce the spectral leakage and suppress the unwanted sidelobes associated with finite observation intervals (Harris, 1978). By applying a window to a data file, the data is brought to zero at the boundaries. The windowed data file is then analyzed by FFT analysis, yielding the spectrum. There are various FFT window functions, but the most common ones used are the Hanning window and the Hamming window (Lessard & Jones, 1988).

There are three main FFT variables. These are the sample frequency, the number of FFT points, and the window function. An increased number of FFT points will enhance the spectral resolution for a fixed sampling frequency. The varying degrees of spectral resolution show various amounts of spectral information. Unless there is a valid reason for transforming only a small section of the respiratory sound signal, the single FFT approach is not useful and merely limits spectral resolution (Mussel, 1992).

The following aspects of the application of FFT to signal analysis should be mentioned:

Three-dimensional Plots

To overcome the disadvantage of transforming a portion of the respiratory sound signal and thus to assist 'feature extraction', multiple Fourier transforming time-shifted signal blocks have been used for displaying the frequency components of long lengths of respiratory signal. Three-dimensional (3-D) plots enable important spectral events to be identified for further analysis. Short, overlapping segments of the respiratory signal are separately Fourier transformed and displayed on the 3-D plot.

Time-shifted Multiple FFT Averaging

Time-shifted, multiple FFT averaging can be used to generate a net average spectrum of a long length of respiratory signal recording in the same manner of producing 3-D plots. Successively overlapping FFTs were performed along the length of a respiratory signal recording and all resulting spectra were then averaged. Several investigators have used this method, including Charbonneau, who used 1024-point FFT and 128 point shift (Charbonneau et al., 1982).

The Fourier transform method has been applied to analyse the spectral components of respiratory sound. Unfortunately there is no information on the precise correlation of the time of the ausculatory signal with the spectral resonance.

The sampled signal is processed to remove any d.c. component and normalized with respect to the RMS value. The analysis of the analog signal is carried out as per the block diagram indicated in the Figure 4.1.

Three-dimensional Spectral Representation of Respiratory Signal

The spectral peaks can be examined as they develop with time and also by the appearance and disappearance behaviour of the various frequency components in the time domain. Instead of displaying spectral data obtained from the window analysis (as indicated in Figure 4.2, where the averaged spectra has been displayed on a two-dimensional axis), one could generate information more resourcefully in a 3-D space of amplitude, frequency and time. Figure 4.3 shows the three dimensional plot of a typical respiratory signal taken from a normal adult. A program was developed to generate the spectrogram for the 3-D display.

4.3 FFT Analysis Procedure

The analogue signal is played back from the FM tape recorder and the selection of an artifact-free segment or segment of interest of the respiratory signal is accomplished by visual inspection of the time signal on the analogue chart recorder. The next step is to sample the signal data for 256/512/1024/2048 points on a PC 230.

The digitized and demeaned and normalized signal is windowed by a Hamming window and a 2048 point DFT is calculated. The spectral output is displayed on the screen. This output indicates the scaled amplitude and the frequency. It also indicates the average amplitude of the spectra. As the sampling length is limited by the processing capability and the spectral resolution, multiple FFT are computed and an averaged spectra is computed (Charbonneau et al., 1982). Here there is no overlapping, although overlapping can be achieved.

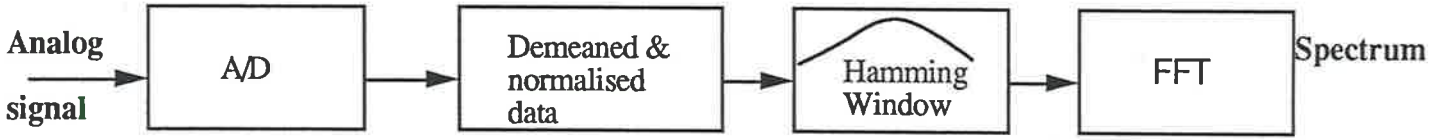
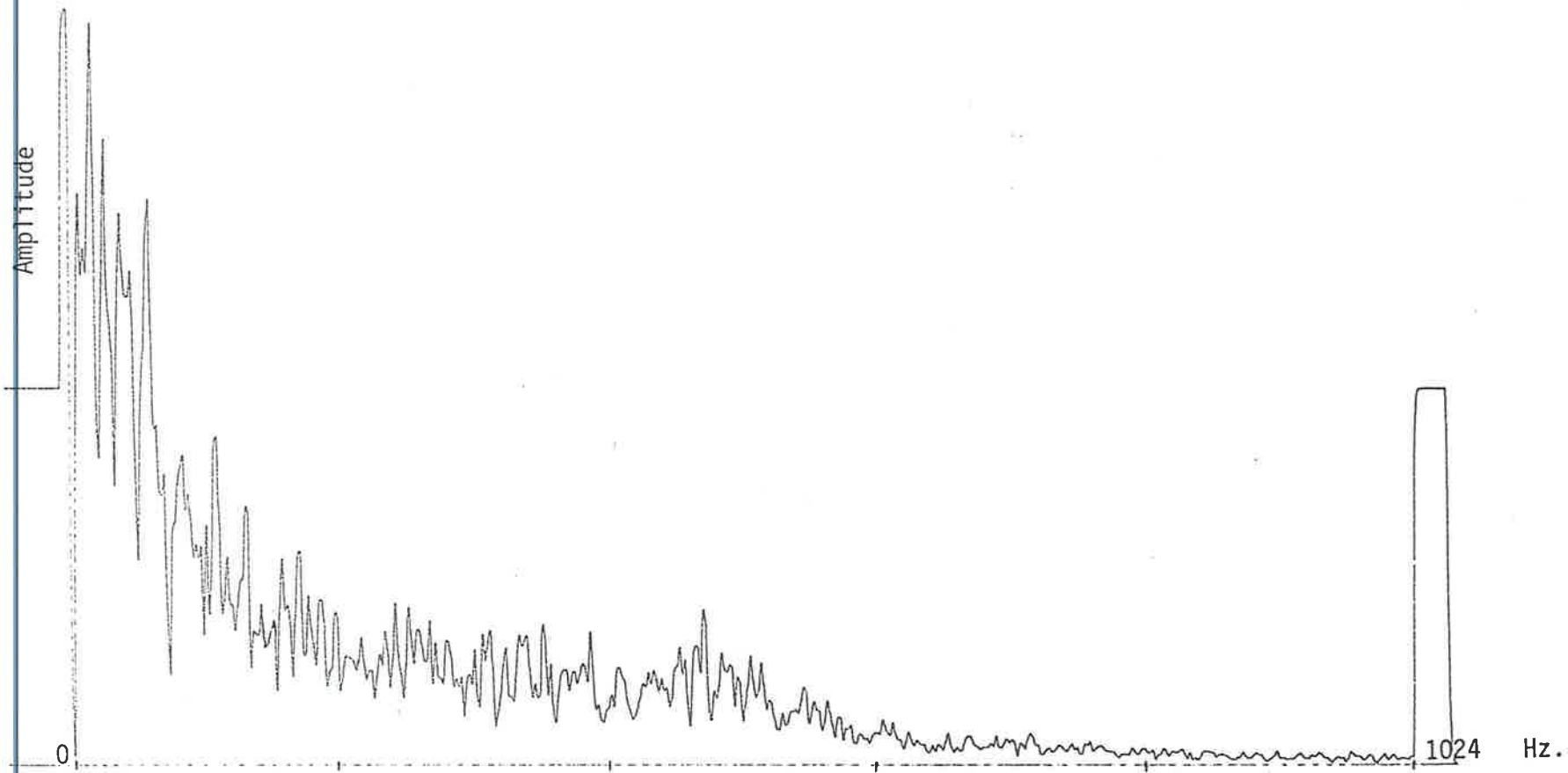


Fig. 4.1 Block diagram of the FFT analysis

Fig. 4.2 A typical FFT spectrum of an unfiltered respiratory signal acquired from an adult.



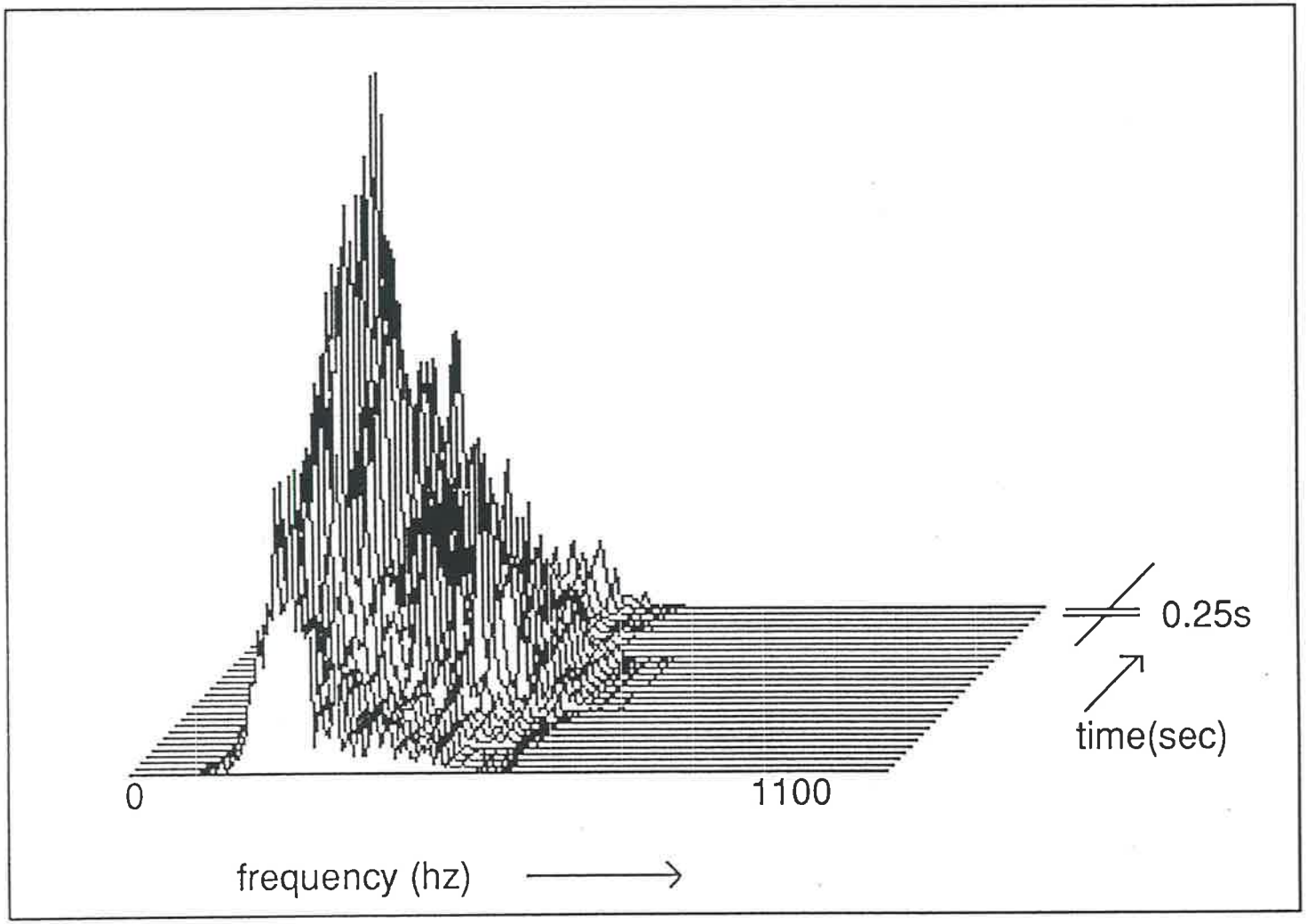


Fig. 4.3 A typical power spectra of respiratory signals acquired from a 40 yr. old adult.

An interactive Turbo Pascal program was used to perform the FFT analysis. The program performs the FFT analysis for each amplitude spectra and averages these spectra.

A typical respiratory signal (with cardiac noise) of a normal adult as displayed after the FFT analysis is plotted in Figure 4.2 after D/A conversion. The FFT was performed on 1024 points, and in order to cover a longer length sample, the signal was played back at half the actual speed, and thus the FFT plot has as its highest frequency 1024 Hz.

4.4 SPECTRAL OBSERVATIONS

Unfiltered Signal:

The unfiltered signals were acquired from the upper neck position and analyzed after amplification. The analog signal from the FM tape recorder was converted into digital signal and sampled at 2048 Hz (f_S). The spectral plots obtained after a fast Fourier transform of the unfiltered signal from a 40 year old female adult is shown in Figures 4.4 and 4.5. The plots are scaled to peak value in the spectrum. The maximum frequency on the plot is $f_S/2$ i.e. 1024 Hz. Figure 4.4 corresponds to the frequency spectra of a respiration signal whereas Figure 4.5 represents spectra of no respiration.

The spectral plot indicates dominance of low frequency signals up to 150 Hz and subsequent energy in the frequency range of 150 Hz to 600 Hz (Figure 4.4). Also, upon analyzing Figure 4.5, it is evident that the spectra is dominated only by low frequency signals. These low frequency signals of high magnitude are cardiac signals which are picked up by the microphone from the surface of the skin. The primary difference between the two plots is the absence of respiratory signals (assuming the other body noises remain the same) in the Figure 4.5 plot. Super-imposing the two plots gives the viewer an idea of the approximate frequency distribution of the respiratory signal.

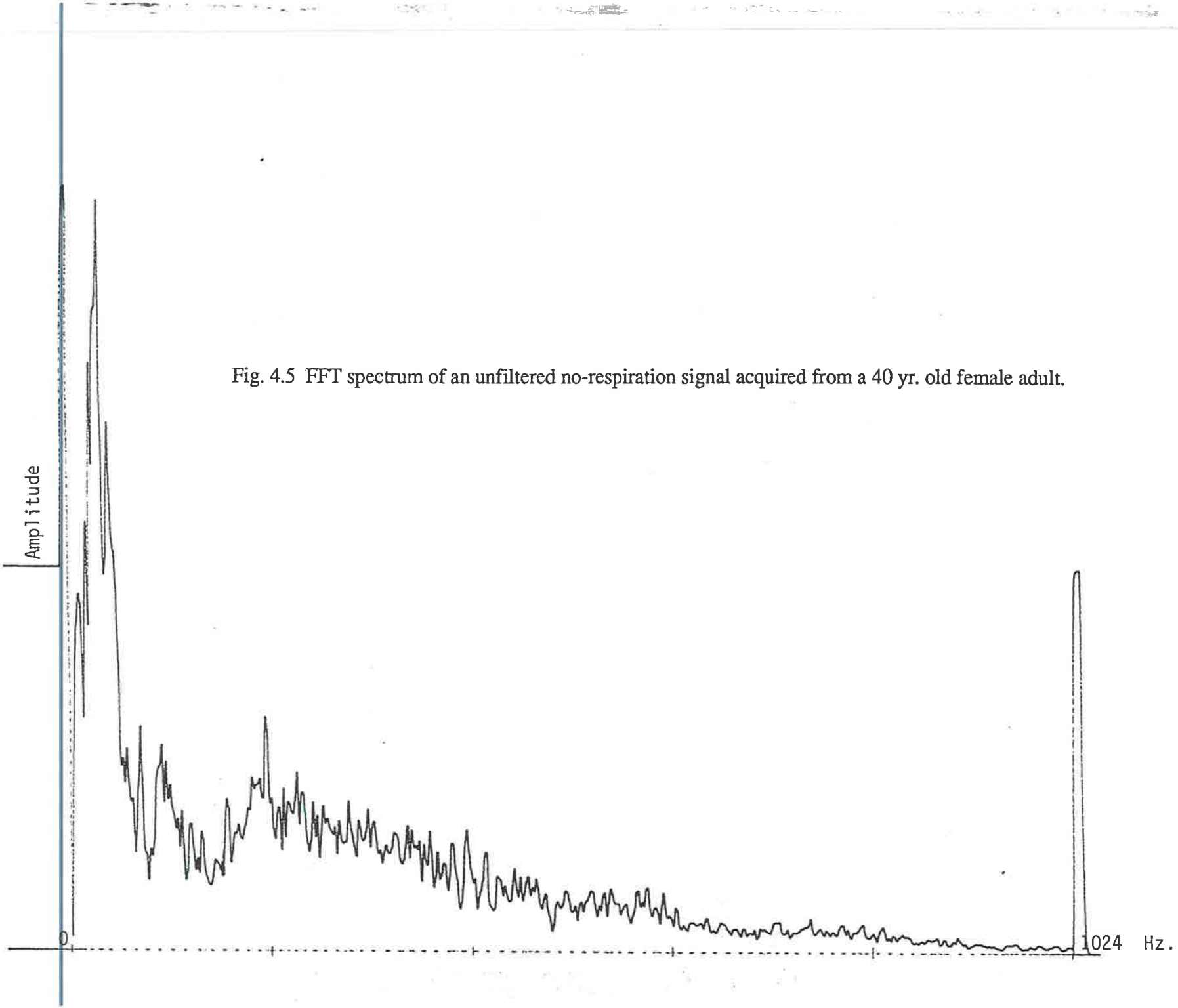
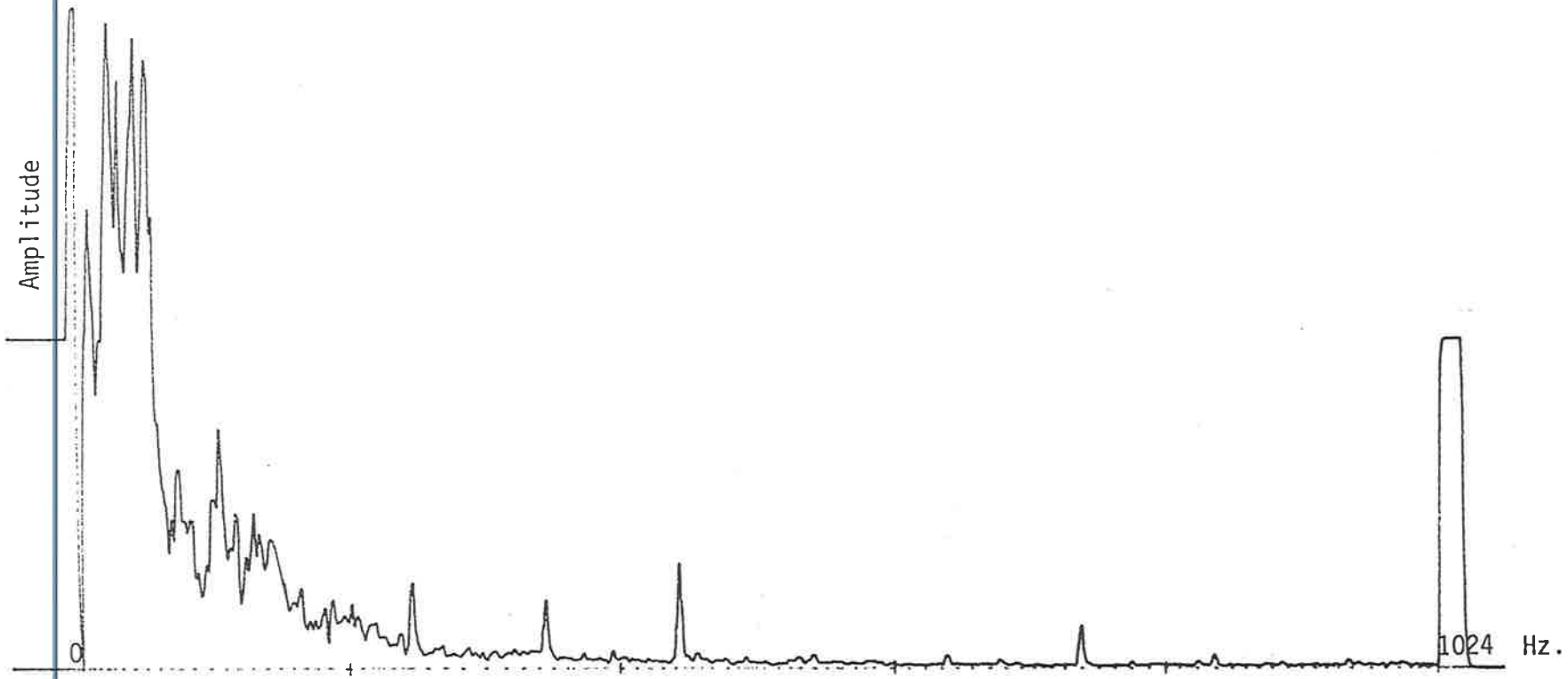


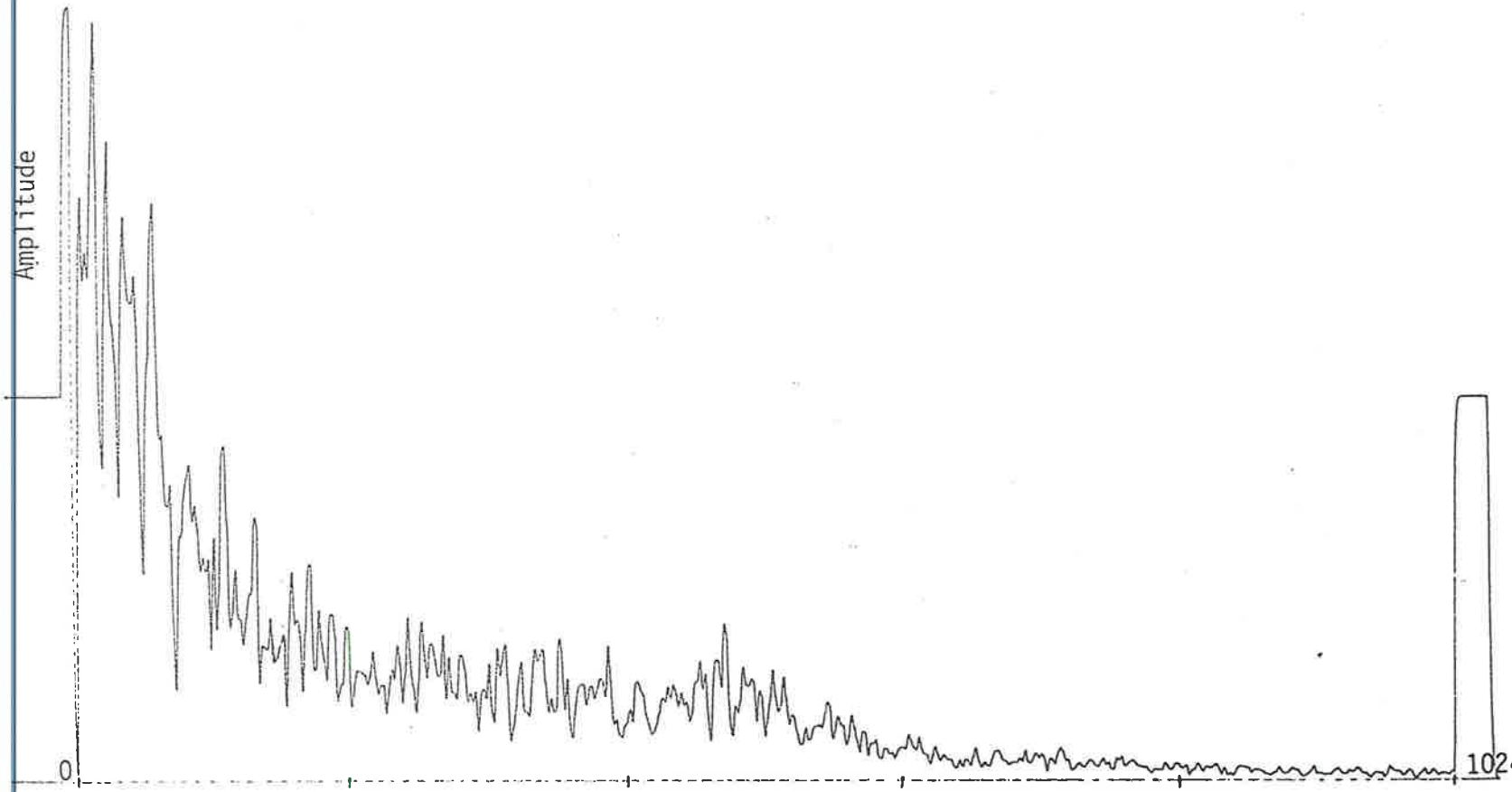
Fig. 4.5 FFT spectrum of an unfiltered no-respiration signal acquired from a 40 yr. old female adult.

Fig. 4.5 FFT spectrum of an unfiltered no-respiration signal acquired from a 40 yr. old female adult.



Figures 4.6 and 4.7 also can be analyzed with the same analogy. The spectra correspond to the respiration and no respiration signal from a 32 year old adult. The cardiac signals form the lower frequency band (up to 150 Hz) of the spectra whereas the respiration signals continue from 150 Hz to 600 Hz.

Fig. 4.6 FFT spectrum of an unfiltered respiratory signal acquired from a 32 yr. old adult.



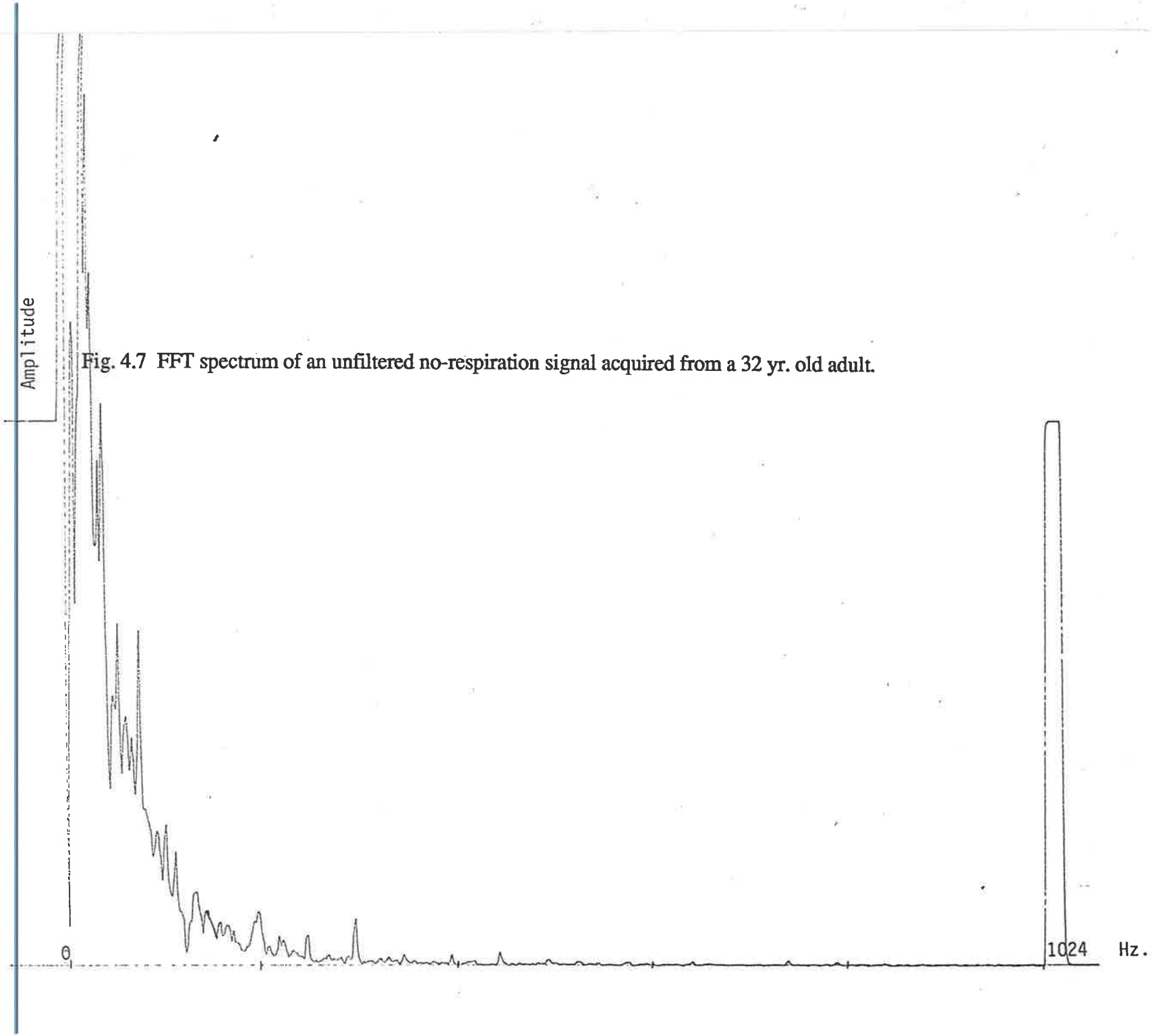


Fig. 4.7 FFT spectrum of an unfiltered no-respiration signal acquired from a 32 yr. old adult.

A Chebyshev band pass filter of frequency response 20-200 Hz was tested to ascertain the presence of any respiratory signal. Figure 4.8 corresponds to the filtered respiratory signal from an adult. Figure 4.9 corresponds to the filtered no respiration signal. The study was extended to five subjects. The comparison indicated the presence of some respiratory energy in the 150-200 Hz range. The remaining energy consisted of essentially low frequency cardiac signals.

The study of the presence of respiratory signals was extended to thirteen subjects in different age groups and sexes (table 4.0).

S.No.	Sex	Age(yrs)	Frequency Range (Hz)
1.	M	60	150 - 600
2.	F	40	150 - 700
3.	M	26	200 - 650
4.	F	44	200 - 650
5.	M(child)	4.5	150 - 700
6.	M	32	150 - 600
7.	M	33	200 - 700
8.	M(child)	2.5	200 - 650
9.	M	26	150 - 650
10.	M	24	200 - 600
11.	M(Baby)	0.3	250 - 750
12.	F	25	150 - 700
13.	F	29	150 - 600

Table 4.0 Frequency Distribution of Respiratory Signal in Different Subjects

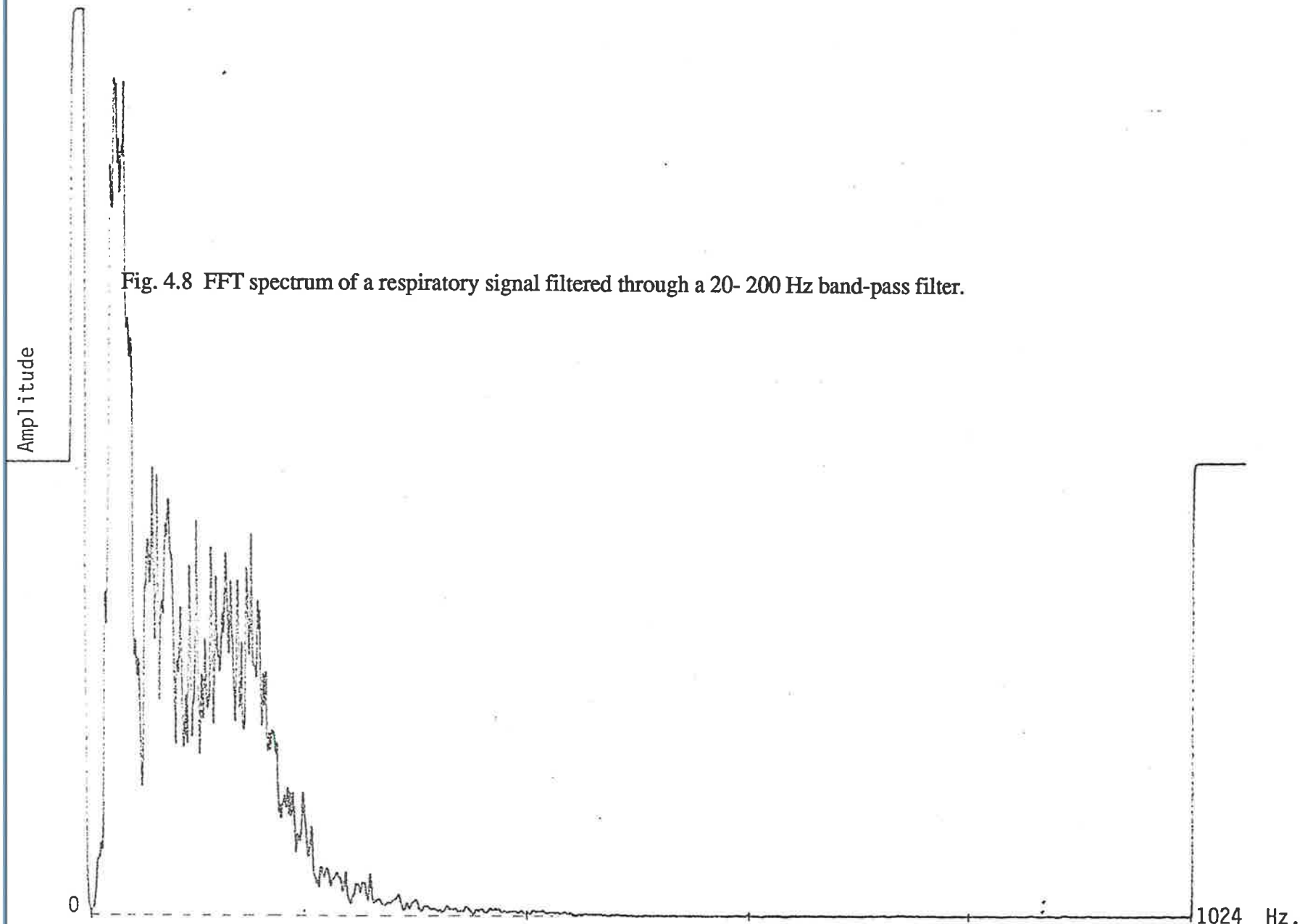
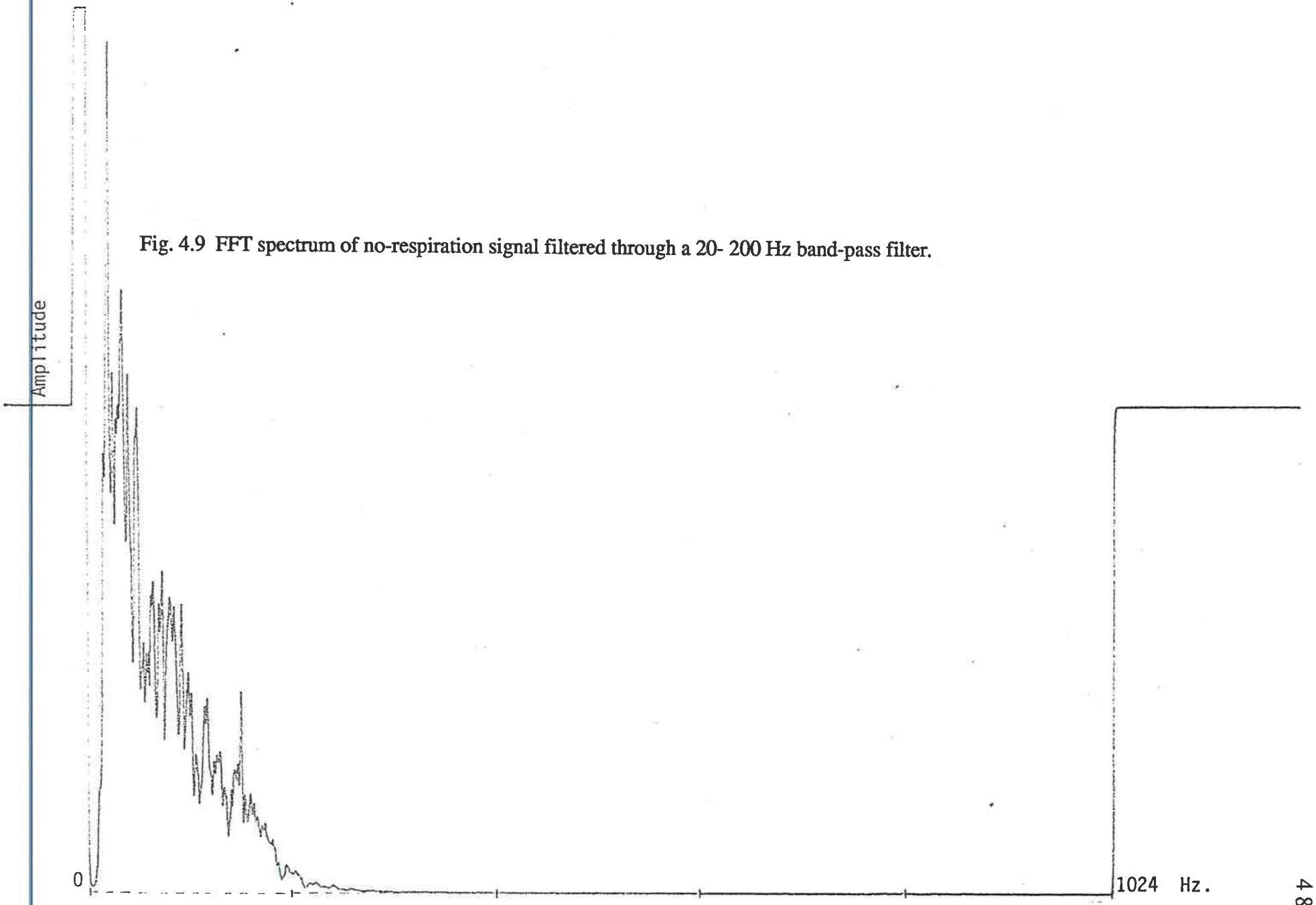


Fig. 4.8 FFT spectrum of a respiratory signal filtered through a 20- 200 Hz band-pass filter.

Fig. 4.9 FFT spectrum of no-respiration signal filtered through a 20- 200 Hz band-pass filter.



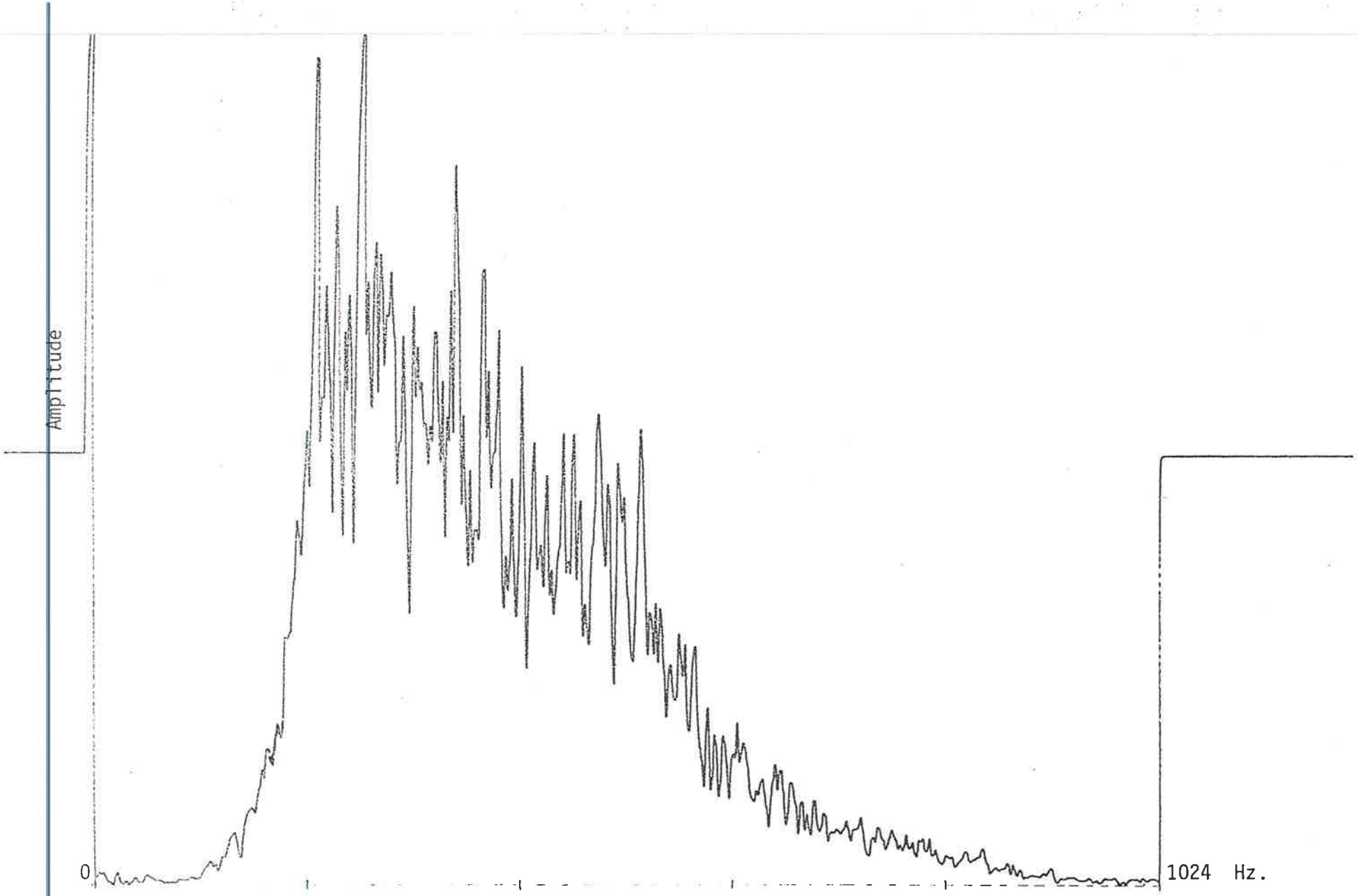
The study revealed that although the respiration signals were primarily distributed in the range 150-700 Hz, the 150- 600 Hz range was the most favourable distribution. Thus the present study was based on the finding that respiratory signals are comprised of frequencies 150-600 Hz.

Filtered Signal

The signals acquired after amplification were filtered through a band-pass filter to eliminate low and high frequency components of the signal and to allow only the respiratory signals. The signals were acquired from the subjects and recorded on an FM tape recorder. Subsequently, the filtered signals were digitized and analyzed using the FFT spectral estimation. Figure 4.10 shows the FFT spectrum of a filtered respiratory signal from a 45 year old female adult. The spectrum peaks at approximately 300 Hz, and the average amplitude is 76. Figure 4.11 also shows the FFT spectrum of filtered, no respiration signal from the same subject. This spectrum has a low magnitude and an average of 14 (because of filtered noise). Upon carrying out a comparative study of the two plots, one can find a signal-to-noise ratio of approximately 5.

Figure 4.12 shows the spectral plot of a filtered respiration signal from a 10 year old female child. The frequency spectra has a peak at approximately 500 Hz and the average magnitude is 71. In Figure 4.13, which corresponds to a filtered, no respiration signal from the same subject, the average magnitude is 11. Thus the signal-to-noise ratio in this case is approximately 6. Similar analysis was carried out on ten other subjects and the signal to noise ratio varied from 4 to 6.

Figures 4.14, 4.15, 4.16, and 4.17 show the power spectra of a filtered respiration signal from an adult. The signals have been recorded and digitized into a 16 bit digital file, sampled at 44.1 KHz. Subsequently, the digitized signal was decimated to 2048 Hz to improve the spectral resolution.

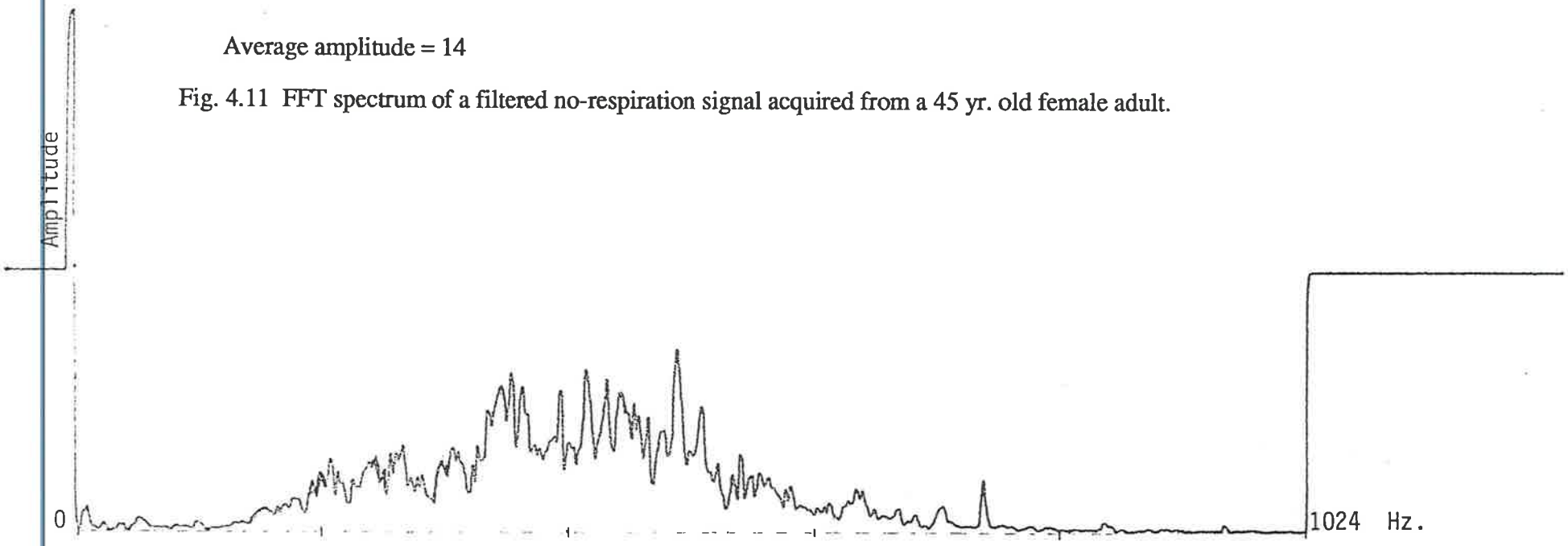


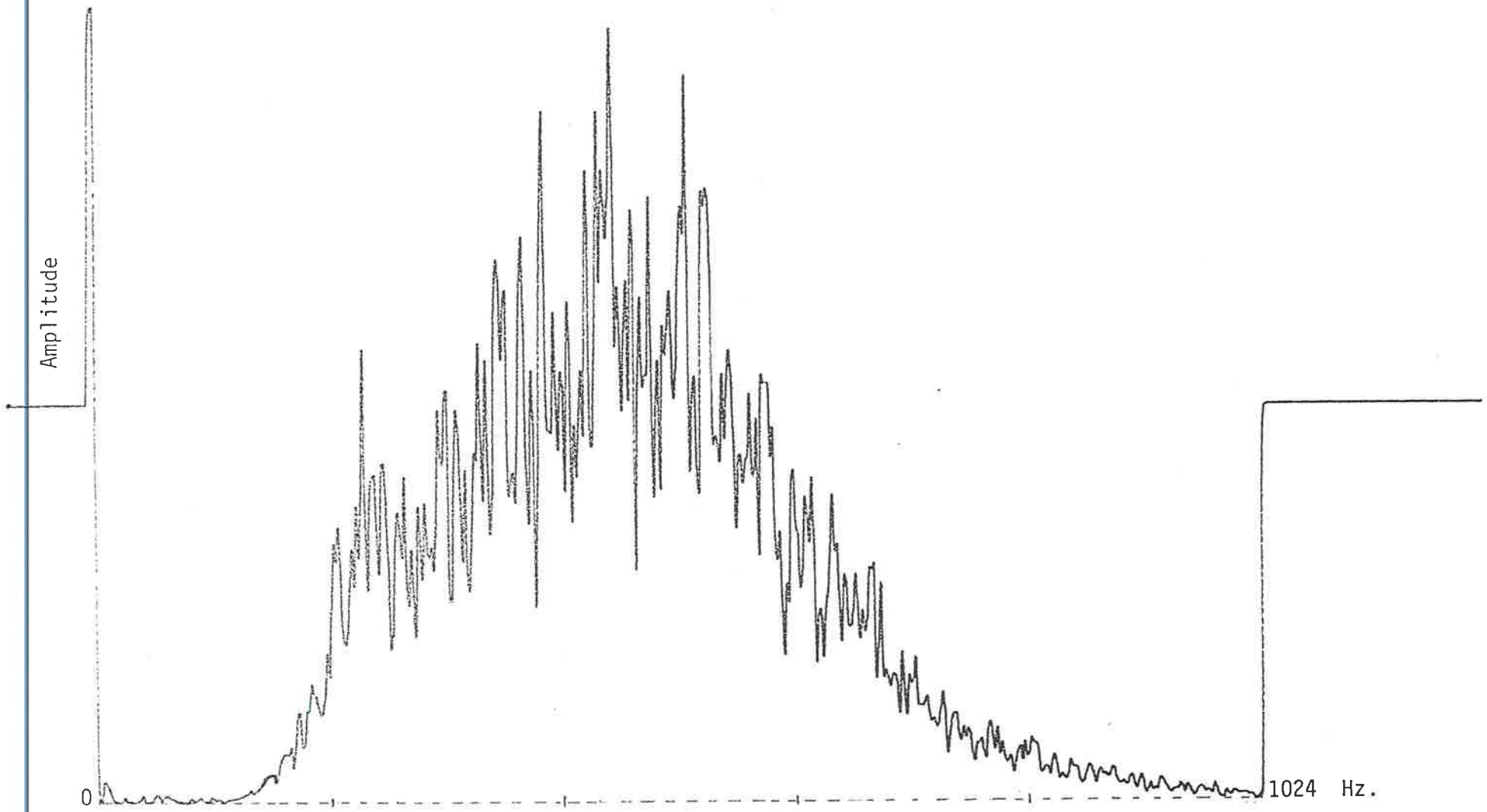
Average amplitude = 76

Fig. 4.10 FFT spectrum of a filtered respiratory signal acquired from a 45 yr. old female adult.

Average amplitude = 14

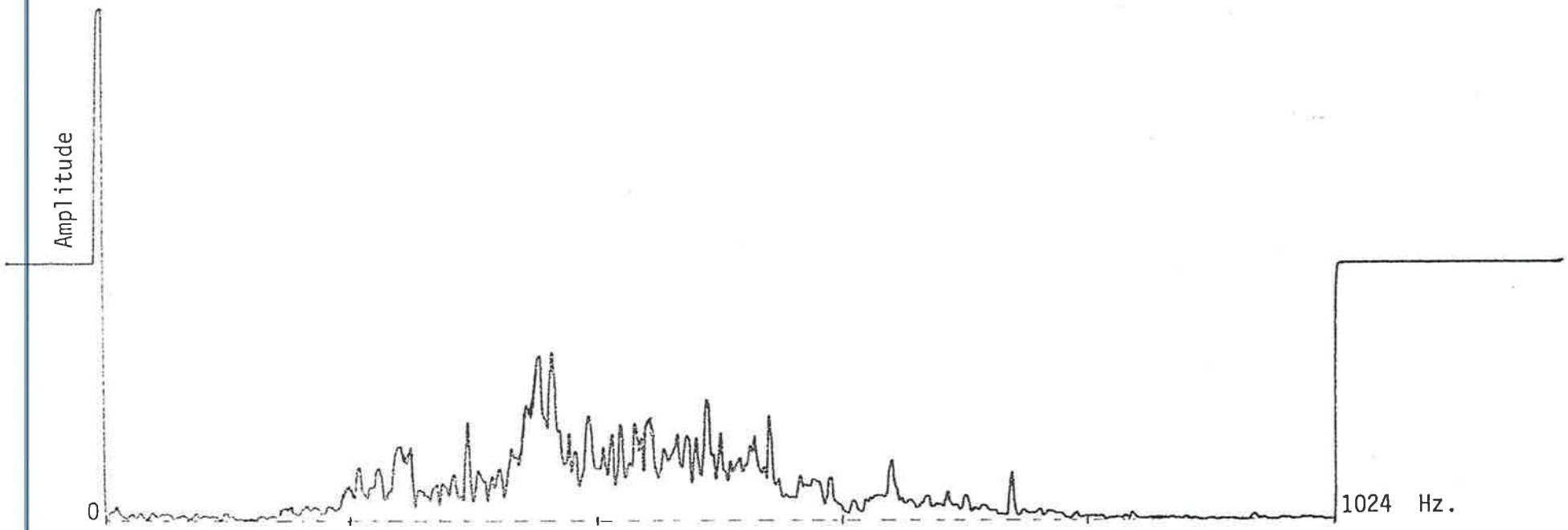
Fig. 4.11 FFT spectrum of a filtered no-respiration signal acquired from a 45 yr. old female adult.





Average amplitude = 71

Fig. 4.12 FFT spectrum of a filtered respiratory signal acquired from a 10 yr. old child.



Average amplitude = 11

Fig. 4.13 FFT spectrum of a filtered no-respiration signal acquired from a 10 yr. old child.

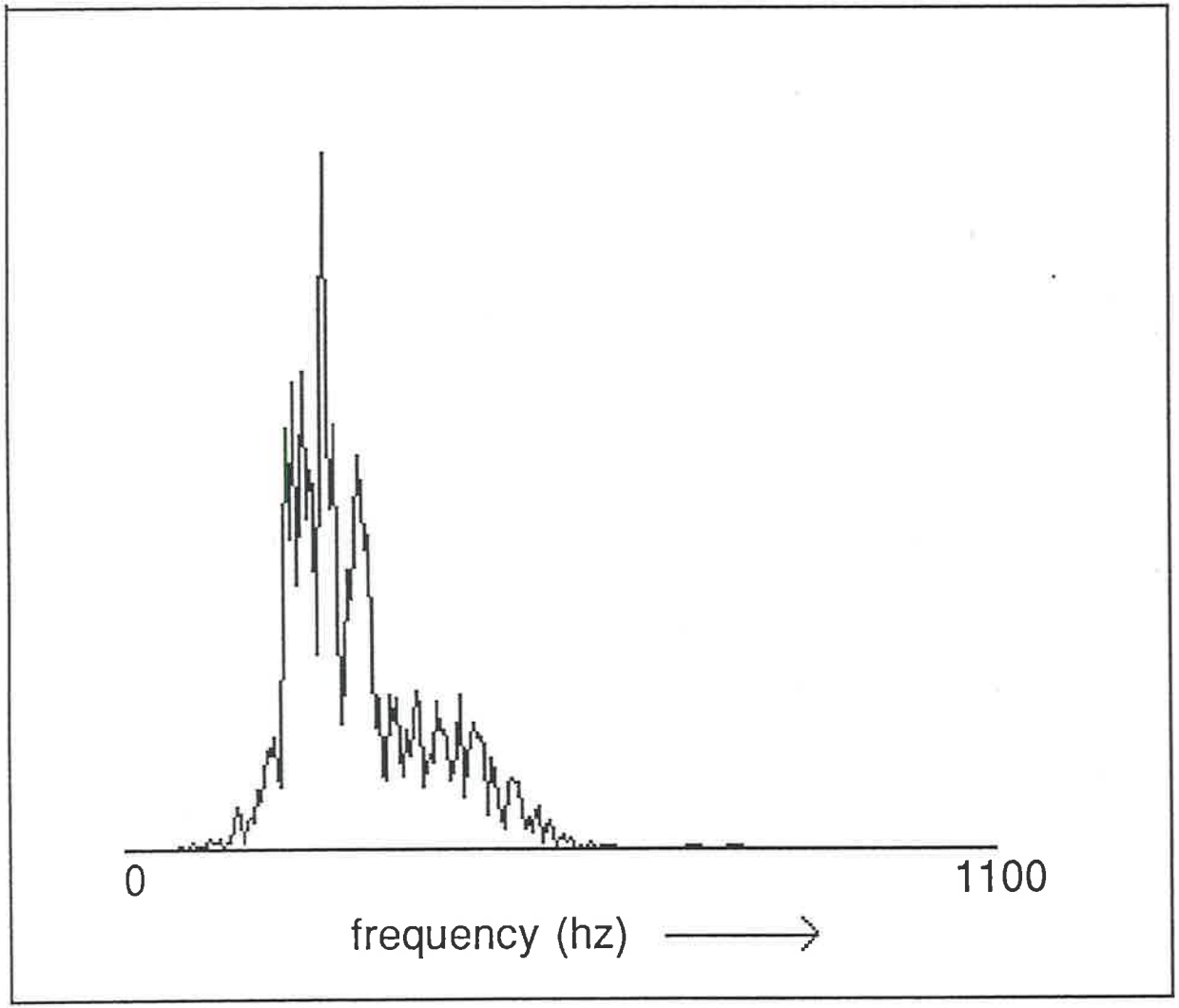


Fig. 4.14 Power spectrum of a respiratory signal acquired from a 8 yr. old child indicating frequencies from 150- 600 Hz.

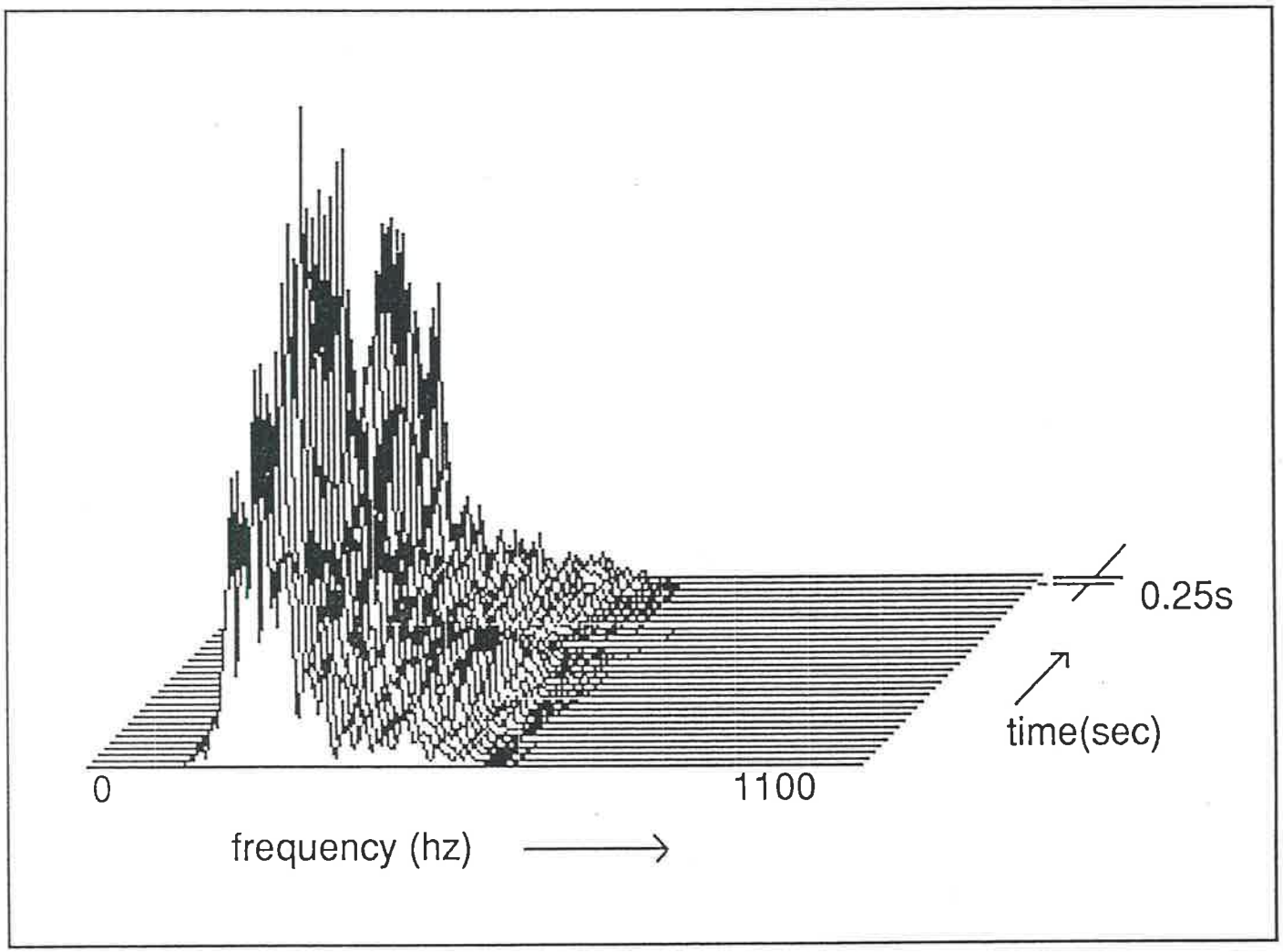


Fig. 4.15 Power spectra of respiratory signals acquired from a 29 yr. old female adult.

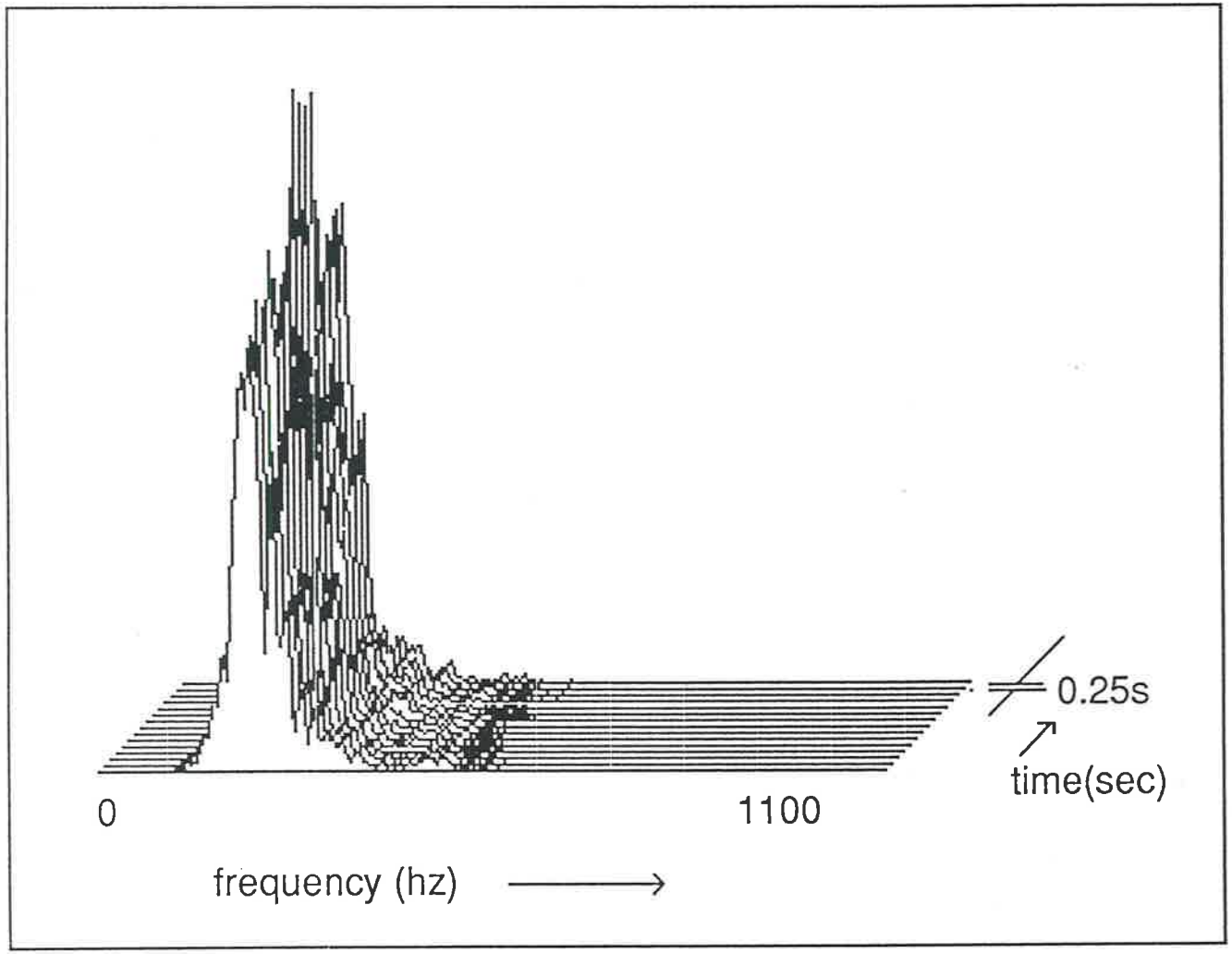


Fig. 4.16 Power spectra of a respiratory signal acquired from a 3 yr. old child.

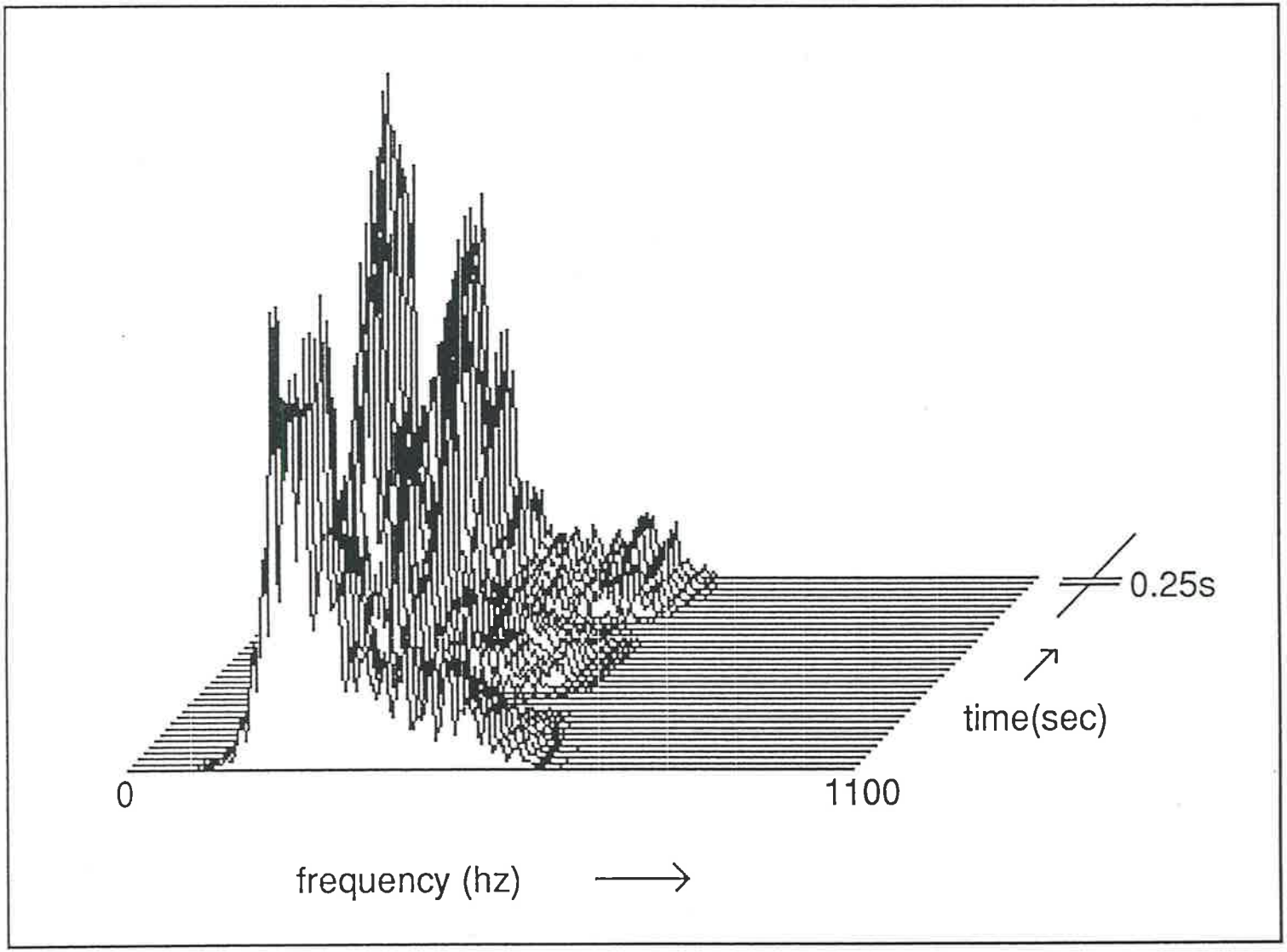


Fig. 4.17 Power spectra of respiratory signals acquired from a 45 yr. old adult.

Figures 4.18 and 4.19 correspond to signal acquired from a 29 year old adult. The respiration signal stands out in Figure 4.19, rejecting all other noises. Figures 4.20 and Fig. 4.21 correspond to an unfiltered respiration signal acquired from an adult. The power spectra in Figure 4.21 clearly shows the dominance of low frequency cardiac noise.

The within-subject reproducibility of tracheal breath sound is relatively good (fig.4.0, Mussel, 1990), but not good enough for a single recording to represent a true picture. Spectral averaging therefore is important. However, different subjects produce considerably different shapes of spectra (Figure 4.0). The magnitude of the peaks in the spectra varies with time, and although the nature of the spectra may not change, the repetition may not be consistent from the same subject. Therefore, we conclude that each subject has this or her own unique and relatively reproducible spectral characteristic, which differs from that of other subjects. It is clear that certain aspects of inspiratory and expiratory spectra are different.

Efforts were made to acquire respiratory signals from infants. Figures 4.22 and 4.23 correspond to the respiratory signal acquired from the chest (lower neck) position and the spectra in Figure 4.23 indicates a similar spectral pattern as that found in adults.

4.5 Noises

There are two types of noises that can be considered, viz.

a) Electrical Noise

b) Ambient Noise

Ambient noise can be further categorized into internal (body) or external noise (i.e. the noise generated in the body or from the environment).

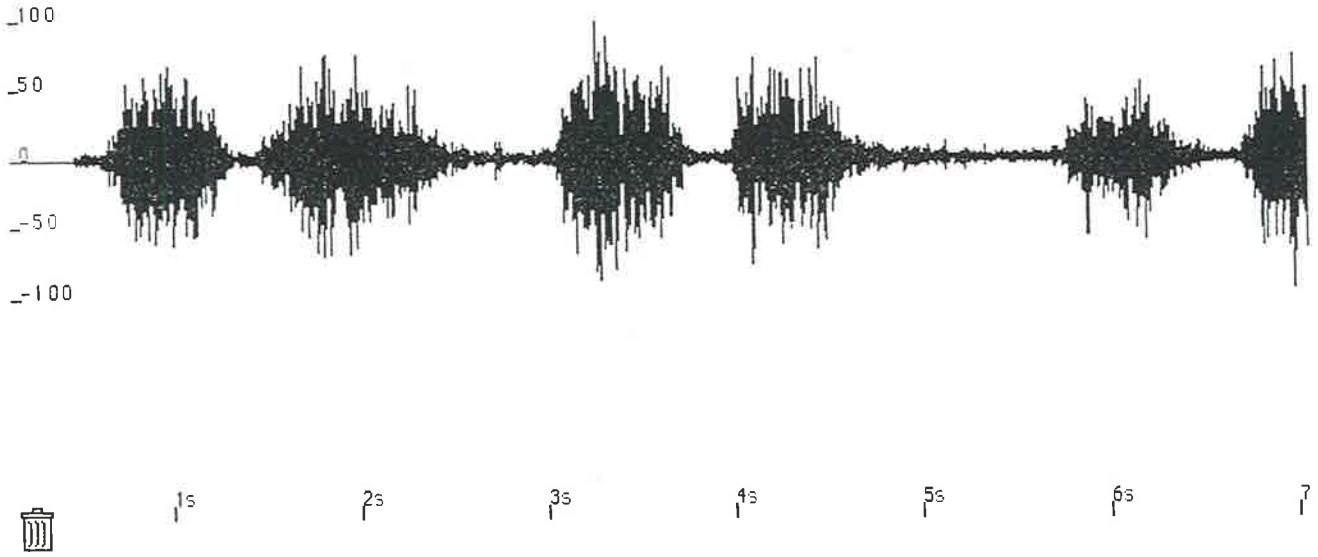


Fig. 4.18 Filtered real-time respiratory signals sampled at 44.1 KHz, acquired from a 29 yr. old adult.

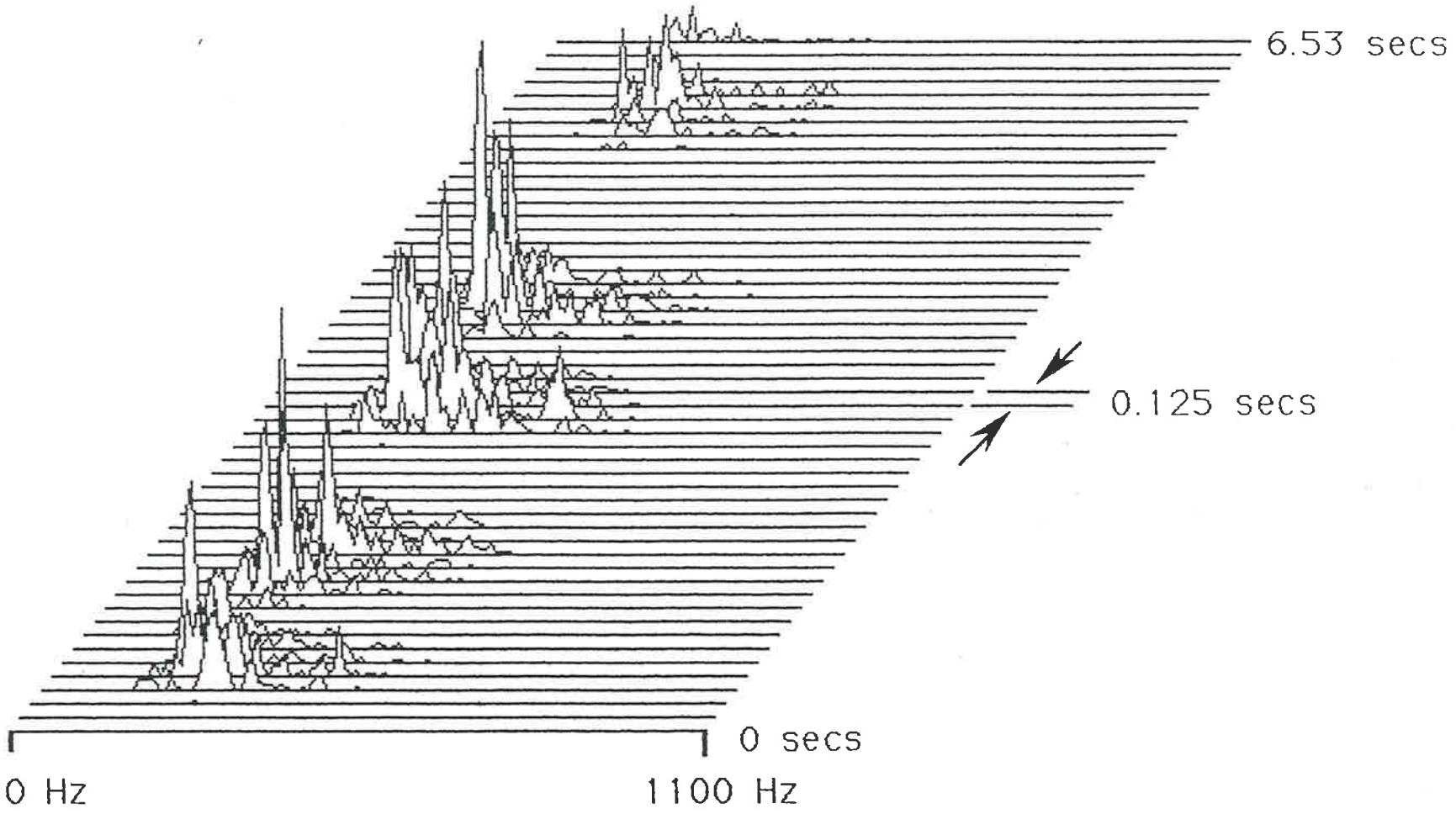


Fig. 4.19 FFT spectrogram of filtered respiratory signals recorded from a 29 yr. old adult.

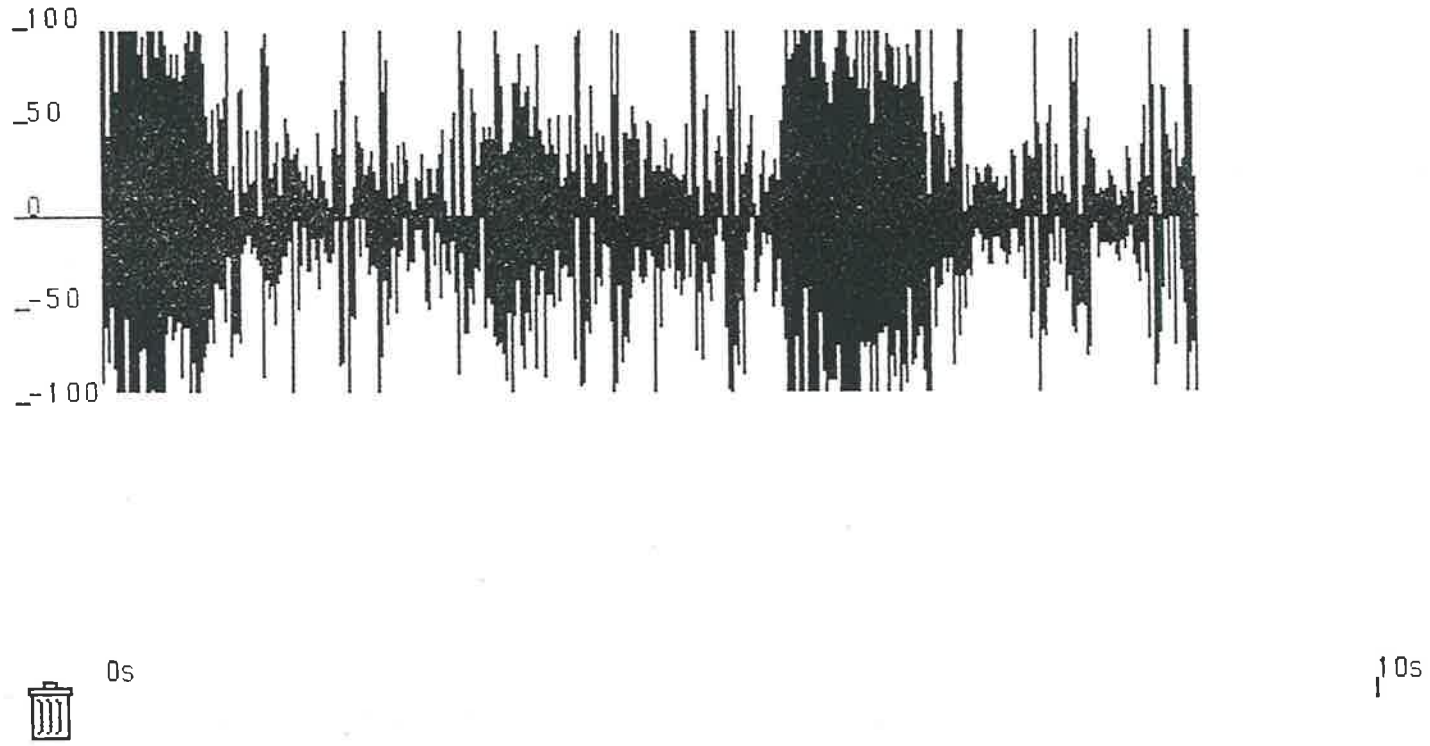


Fig. 4.20 Unfiltered real-time respiratory signals acquired from a 60 yr. old adult sampled at 44.1 KHz.

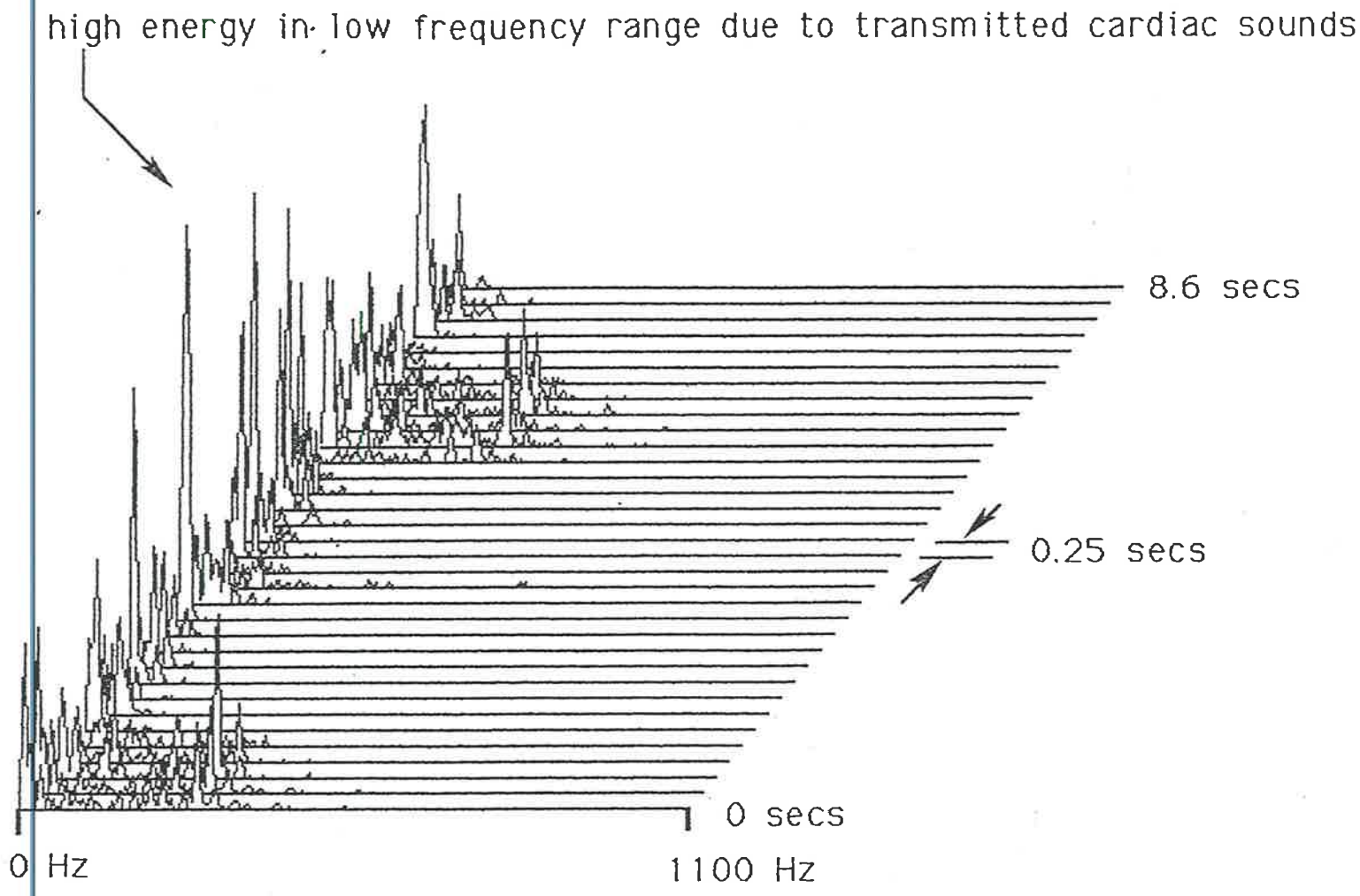


Fig. 4.21 FFT spectrogram of unfiltered respiratory signals, showing both cardiac and respiratory components, recorded from a 60 yr. old adult.

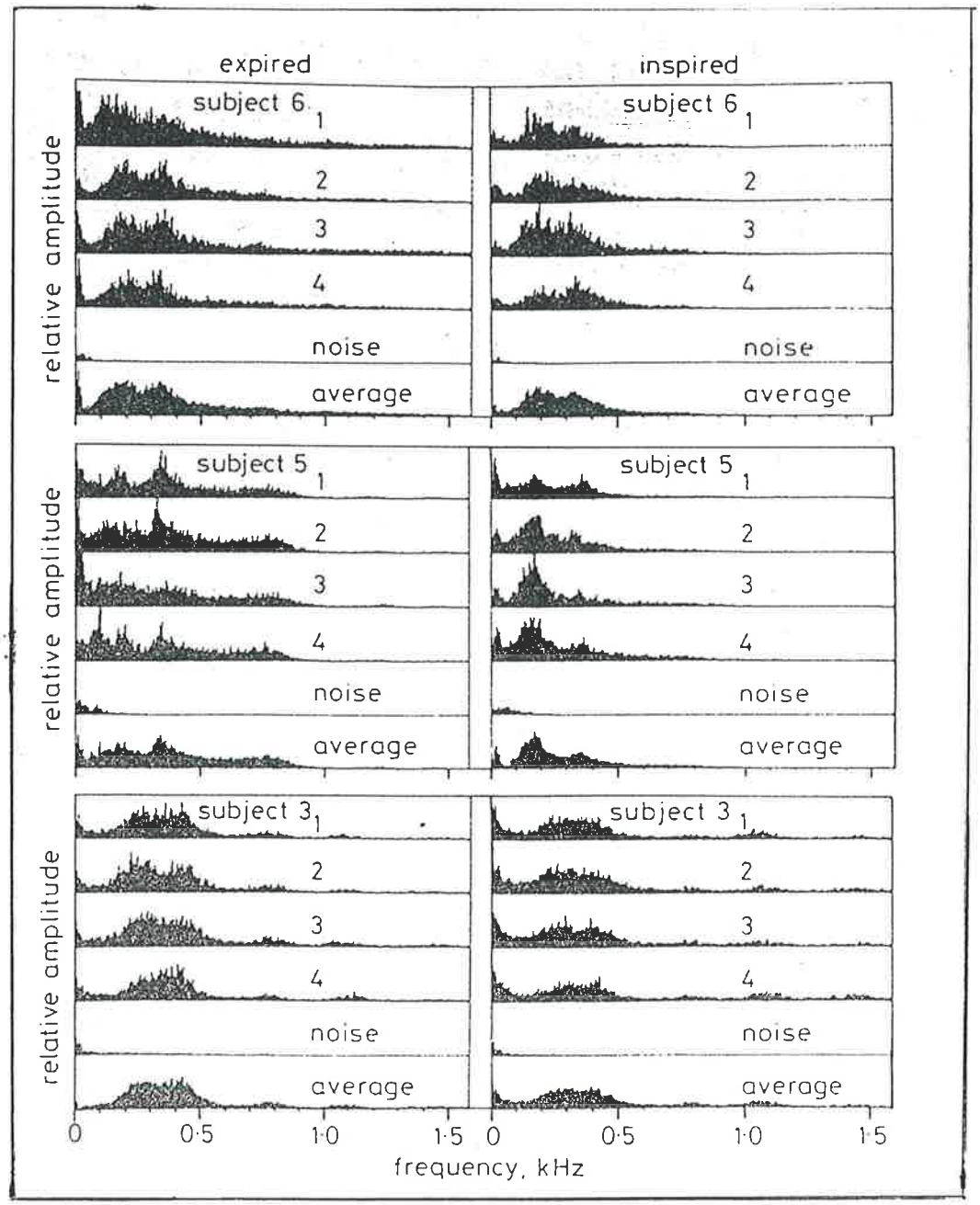


Fig. 4.0 Inspiratory and expiratory net average spectra for different subjects (Mussel, 1990).

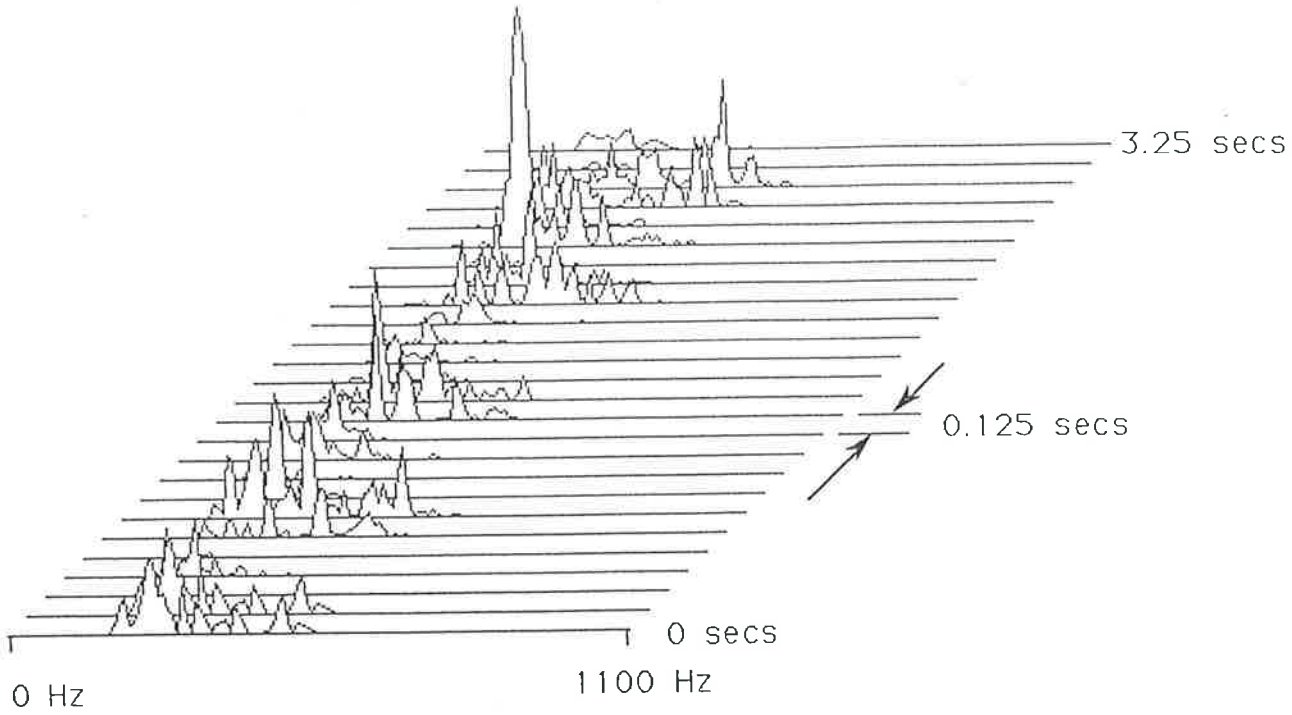


Fig. 4.22 Filtered real-time respiratory signals sampled at 44.1 KHz, acquired from the chest of an 4 month old infant.

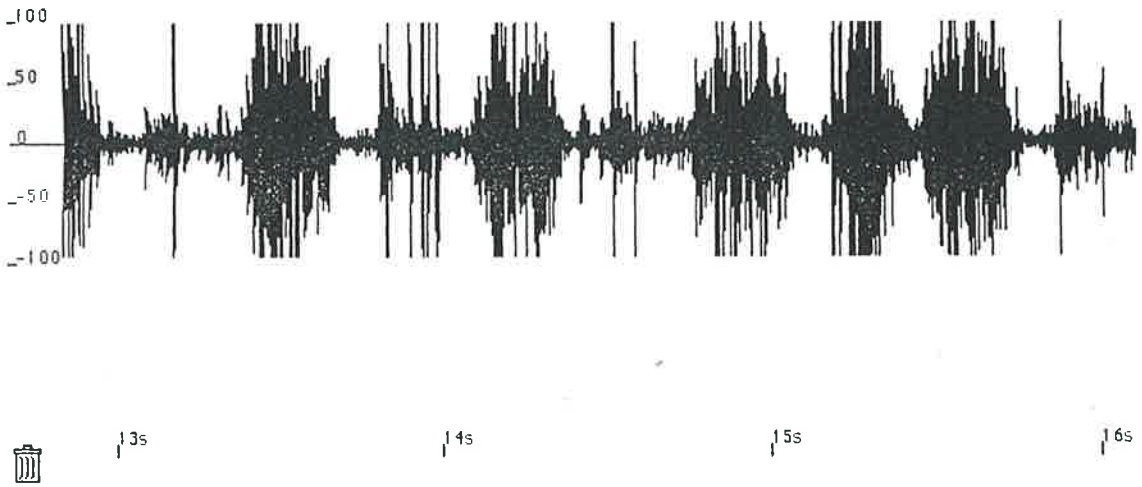


Fig. 4.23 FFT spectrogram of filtered respiratory signal recorded from the chest of an 4 month old onfant.

4.6 Body Noises

Body noises can be caused as a result of the following phenomena:

a) Speech

b) Swallowing

c) Snoring

d) Others (belching or burping, intestinal disruptions, bodily movement, etc.)

A study was done to analyze the frequency spectra of some of the body noises and their impact on the acquired signal from the chosen site.

4.7 Spectral Analysis of Speech

An analytical study was done to analyze the frequency spectrum of the filtered speech signal as recorded from the microphone placed at the same site.

The speech signal acquired from the neck position was observed to be one hundred times the magnitude of the respiration signal, and thus had to be attenuated for analysis. The study on the speech signal was done in three categories - (a) vowel sounds (b) consonant sounds, and (c) continuous speech.

The chosen vowel sounds, consonant sounds and continuous speech are as follows:

The eight vowel sounds in the following words were used for analysis:

FALL, FAR, MET, HEES, IT, PORT, HER, PUT

Special emphasis was placed on the vowel sounds made while pronouncing these words, and the magnitude of the signal was monitored on the CRO on each occasion to establish consistency of signal level during recording. A similar exercise was repeated for consonants.

A count from one to ten was utilized to obtain the effect of continuous speech. The subjects hence were asked to count numbers from one to ten in quick succession to generate a continuous sentence without a perceivable pause.

The recordings obtained from ten different subjects indicated that the frequency spectra were different for the same vowels and consonants. These ten subjects differed in age and sex. Some of the spectra and the components of the speech they represent are shown in Figures 4.24, 4.25, 4.26. The spectra of vowels indicates the presence of discrete frequencies, though the frequencies varied among individuals. Figure 4.24 corresponds to the vowel sound PUT and indicate presence of 125 Hz, 250 Hz and 400 Hz frequencies. The spectra of consonants were also analyzed and the spectral plots showed the presence of discrete frequencies. The recordings obtained for continuous speech (a count of one to ten) yielded a semi-continuous spectrum, though the continuity varied among individuals.

4.8 Spectral Analysis of Swallowing Sound

Swallowing generates an acoustic signal which can be recorded by the transducer at the neck position. The signal was found to be about ten times the magnitude of the respiration signal. For the purpose of recording, the swallowing signals were attenuated to one-tenth of the acquired value. The spectral representation of swallowing sound shows the presence of low and high frequencies in the frequency domain. The Figures 4.27 and 4.28 correspond to spectra of a filtered swallowing signal acquired from the upper neck position from two different adults. Figure 4.29 shows the spectra of an unfiltered swallowing signal recorded from an adult subject.

The study was extended to about ten subjects and the spectra indicate the presence of various high and low frequencies and continuity of the spectra. There also was no consistent pattern common to the spectral plots obtained from different subjects.

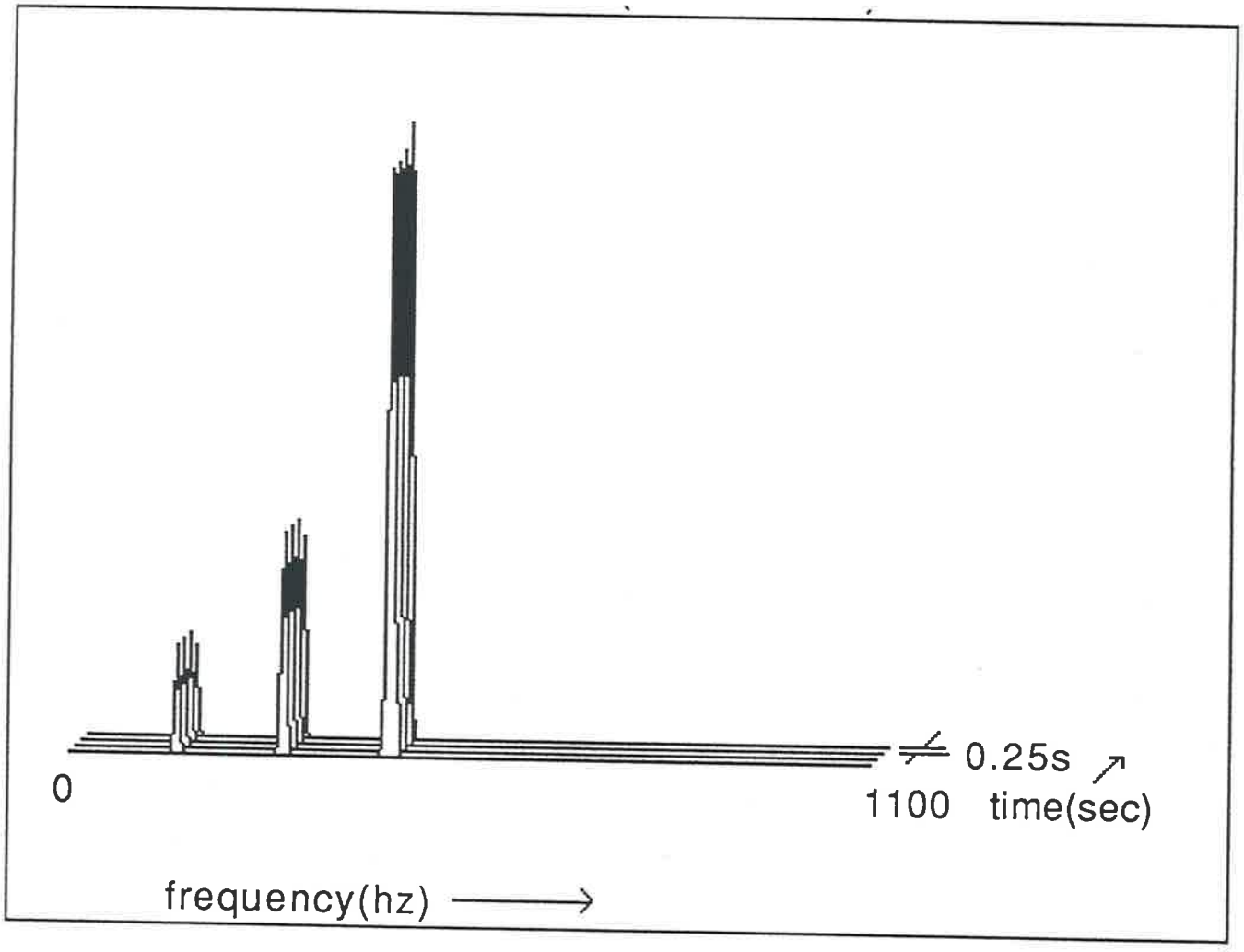


Fig. 4.24 Power spectra of filtered speech signals, of vowel sound 'PUT'.

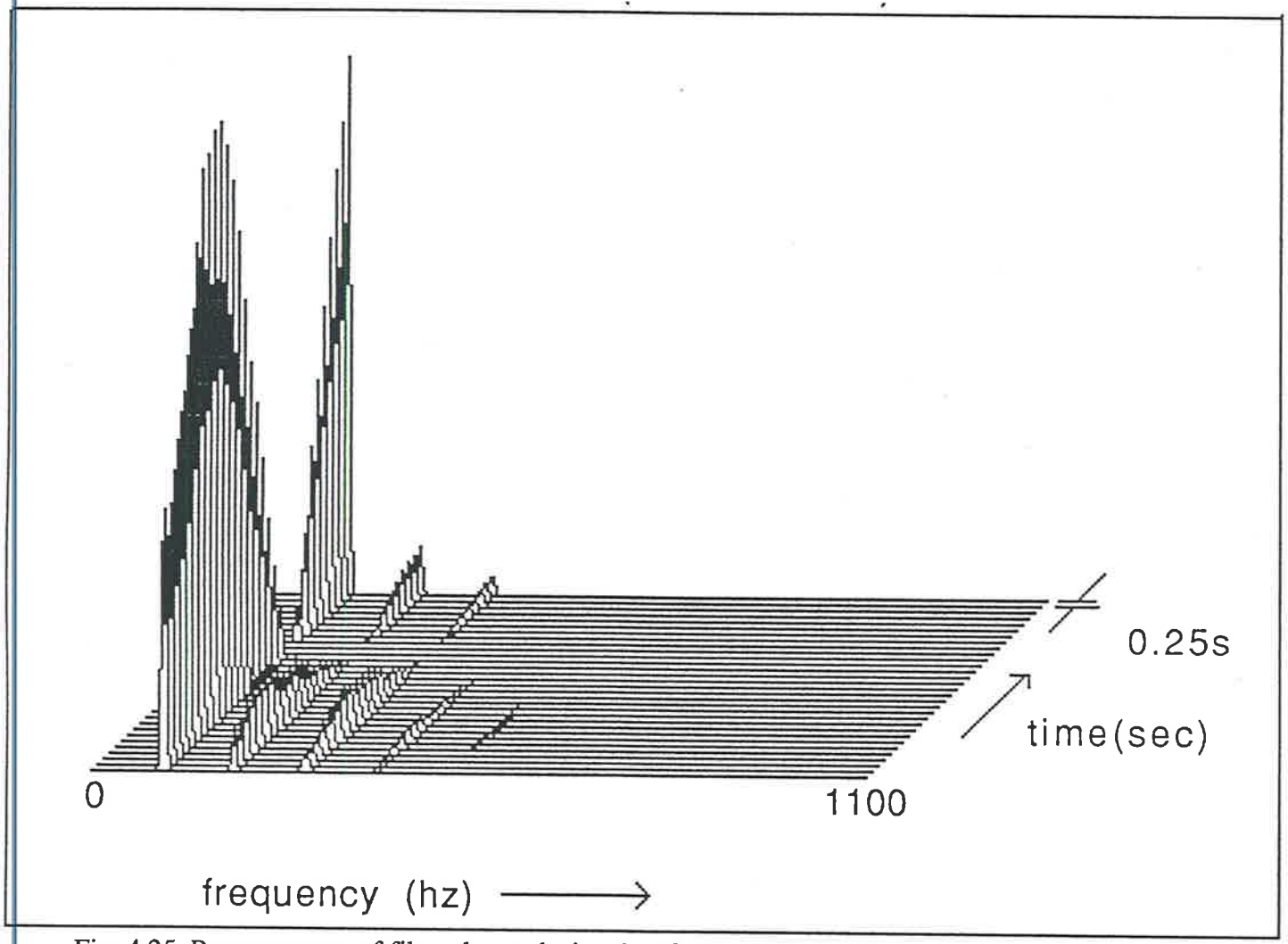


Fig. 4.25 Power spectra of filtered speech signals, of two vowel sounds.

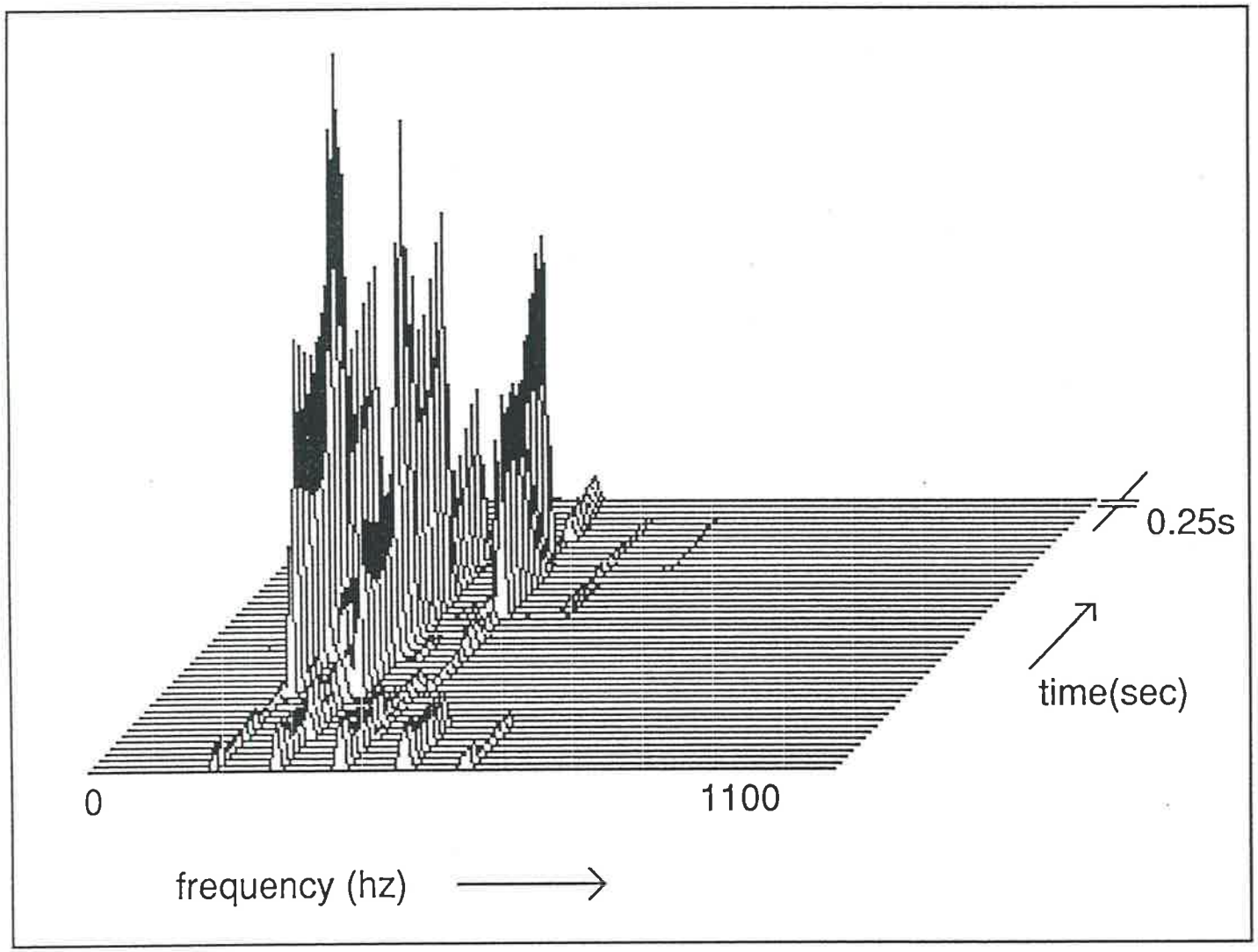


Fig. 4.26 Power spectra of filtered speech signals, of vowel sounds.

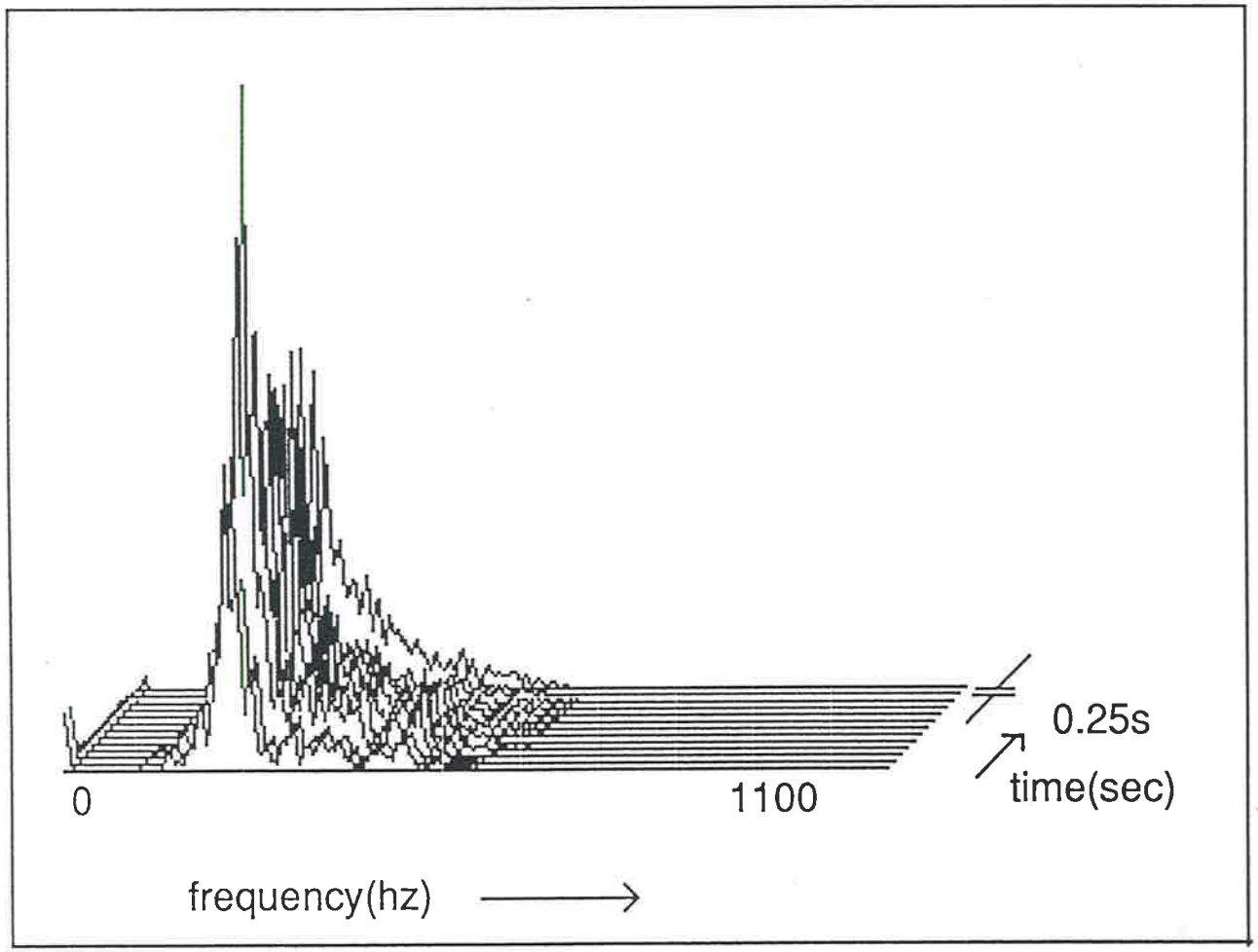


Fig. 4.27 Power spectra of filtered swallowing signals indicating presence of frequencies from 150 Hz to 550 Hz.

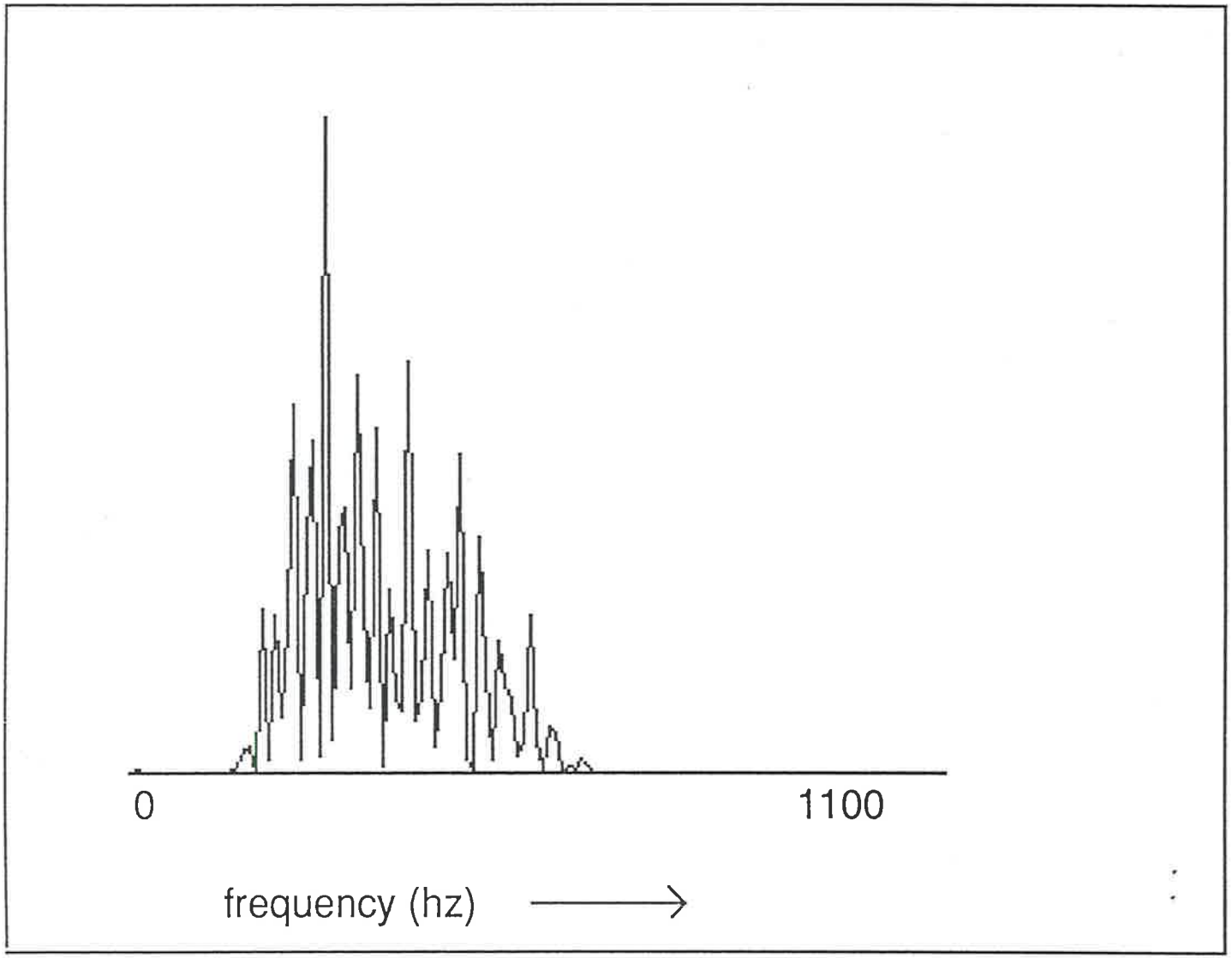


Fig. 4.28 Power spectrum of filtered swallowing signals indicating presence of frequencies from 150 Hz to 600 Hz.

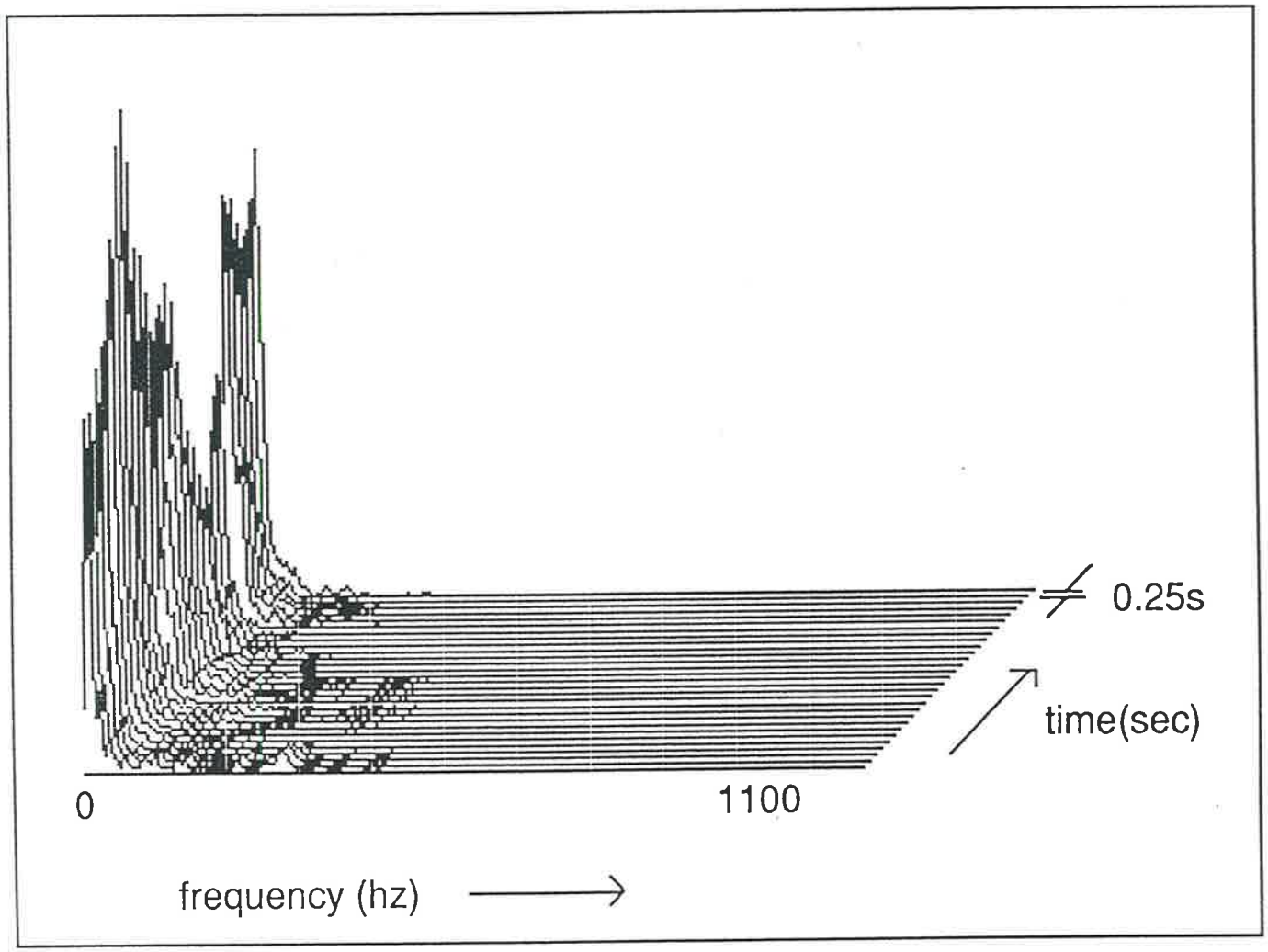


Fig. 4.29 Power spectra of unfiltered swallowing signals indicating presence of low frequencies.

4.9 Spectral Analysis of Snoring Sound

The obstructive movement of air in the air passage can generate sounds, especially while sleeping. This condition is termed snoring. Snoring can be defined as a manifestation of partial obstruction of the upper airway (Berne, 1988). The type and intensity of the acoustic signal associated with snoring varies within the same subject (depending upon the extent of obstruction).

Snoring signals were acquired from the upper neck position with the aid of the transducer. The recorded signals were ten times the magnitude of the respiratory signal and had to be attenuated to one-tenth of the acquired value for analysis.

This study was limited to two subjects due to practical difficulties in organizing the recordings. The acquired signals were derived from simulated snoring, which served as a good approximation of the actual signals.

The spectral analysis of these signals on sampling at 4086 Hz generates a spectra with a maximum frequency of 2048 Hz. The spectra indicate the presence of discrete frequencies scattered along the frequency domain. Figure 4.30 indicates the amplitude spectral plot of an unfiltered snoring signal acquired from an adult. The signal originally was sampled at 44.1 KHz and subsequently decimated to 4096 Hz for analysis.

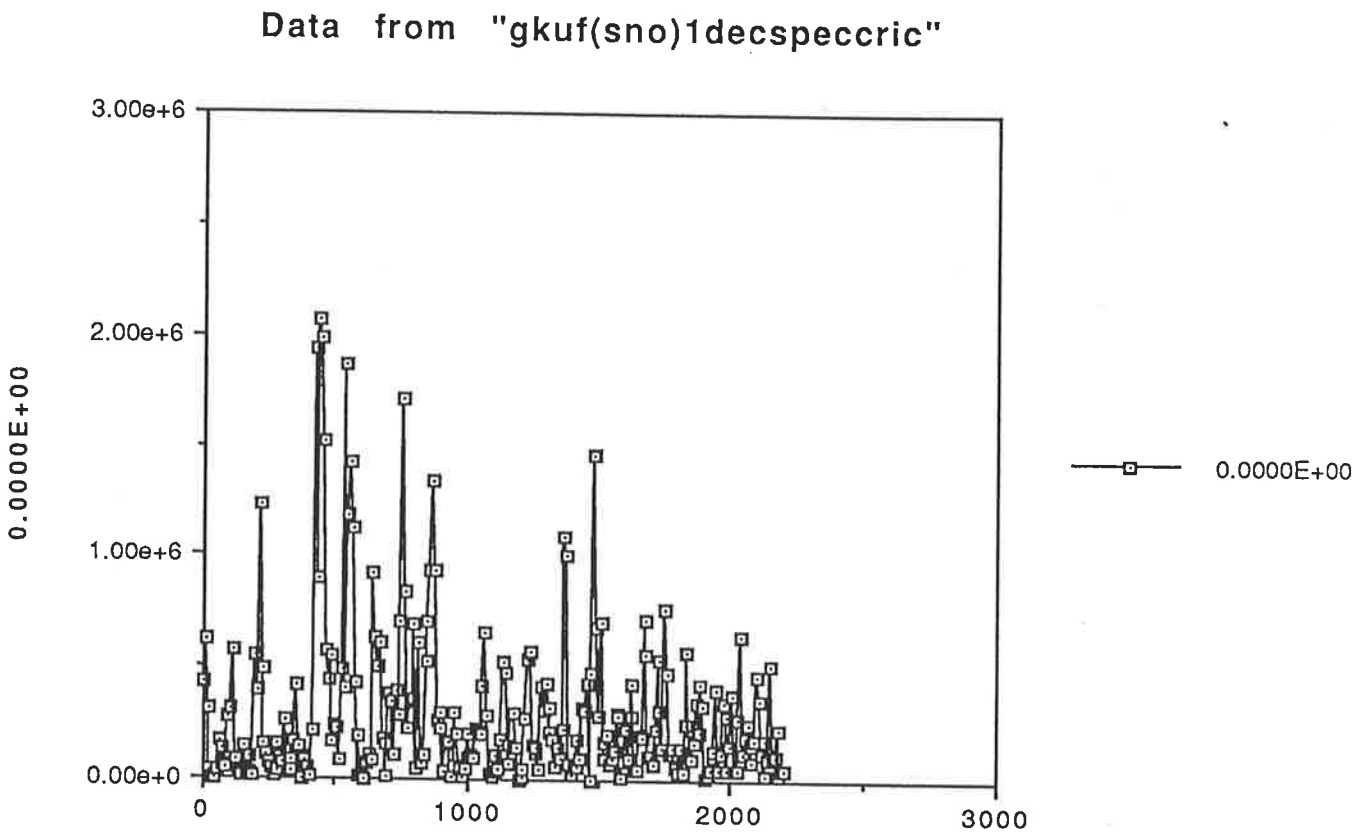


Fig. 4.30 FFT spectrum of unfiltered snoring signals acquired from an adult.

CHAPTER 5

THE DEVELOPMENT OF AN APNOEA MONITOR

5.1 Existing Apnoea Monitors

An acoustic monitor has been devised which will detect partial upper airway obstruction (snoring) (Penzel et al., 1990). Although a respiration monitor has been developed that utilizes the acoustic signal as an indication of tracheal or chest air flow the reliability is poor. Moreover, as stated previously, the respiration monitor again involves a combination of measurements to detect breathing (Gyulay et al., 1987).

J. Werthammer conducted a comparative study of apnoea monitors using acoustic signals and transthoracic impedance methodology (Werthammer et al., 1983). The study indicated that, primarily due to its mechanism, impedance monitors may not detect apnoea caused by upper airway obstruction. In this study, the impedance monitor failed to set off an alarm during 19 episodes. In each of these episodes no breathing sounds were detected, the subject's heart rate fell, and paradoxical respiratory efforts were observed, presumably diagnosed as obstructive apnoea. The acoustic transducer used in this study was placed into the naris and thus could detect breathing through the nose, but not breathing through the mouth. The monitor sounded 4 false alarms during 11 hours of testing.

S. Gyulay tried another monitoring system (Gyulay S et al., 1987). This system consisted of a measurement of a tidal volume by inductance plethysmography, summing the calibrated signals from sensor bands placed around the ribcage and abdomen; measurement of paradoxical breathing by examining the phase relationship between rib cage and abdominal signals; measurement of the percentage of arterial oxygen saturation (% SaO_2) with a Biox IIA oximeter (Ohmeda, CO) interfaced with the monitoring system; the measurement of subject's heart rate, determined by the R-R interval of the ECG; and the measurement of the subject's body movement with an activity transducer strapped to the subject's wrist. This set-up involved a multitude of measurements to diagnose the occurrence of central/obstructive apnoea. The developed device could identified 62% of obstructive apnoeas and 11% of central apnoeas. In two subjects the device did not respond at all. The portable system had an overall successful performance rate of 78% when tested on 12 sleep apnoea patients. The reliability of this equipment was dependant upon the criteria used to define breathing disturbances, and thus optimization was essential. This system involved a whole range of measurements, which is not desirable for the reliable detection of breathing.

F. Lue conducted a comparative study which compared a belt with a piezo-electric transducer (RESP-EZ from EPM Systems,IL.) and plethysmography (Lue F, et al., 1992). The study indicated the inadequate performance of the piezo-electric belt in identifying apnoea.

Studies done so far on the respiratory acoustic signals have indicated the variability of the site of signal acquisition. S.K. Chowdhury conducted a study choosing chest position as the site for signal acquisition (Chowdhury et al., 1981). Studies conducted by S. Liu, N. Graveriely and A. Cohen indicated the selection of tracheal position (Liu et al 1987; Gaveriely et al., 1981; Cohen et al., 1984). The trachea has become the preferred measurement site because tracheal breath sounds are louder than chest sounds,

are less filtered by chest sound, but nevertheless correlate well with chest sounds (Mussel et al., 1990).

5.2 Development of the Present Apnoea Monitor

An adult breathes about ten to fifteen times per minute, whereas an infant breathes up to twenty to twenty-five times per minute. Breathing in human beings is a variable phenomenon. Also, there is an option of breathing either through the nose or through the mouth.

The aim of the present apnoea monitor is to detect cessation of breathing in normal subjects. In other words, the objective of this research is to develop a device which can non-invasively detect the prolonged interruption or total failure of breathing during respiration cycle.

After carrying out the spectral analysis of the respiratory signals, it becomes clear that respiratory (acoustic) signals are predominantly distributed from 150 Hz to 600 Hz in the frequency domain. The output signal of the designed filter and rectifier (Figures in chapter 5) is essentially respiration. This fact became evident after listening to the auditory output of the filtered signals. Figure 5.1 shows the real time filtered and rectified respiration signals recorded at the Sleep Laboratory at the Royal Adelaide Hospital (RAH) and their comparison with chest movements, abdominal movements, and airflow. Though the output signals were free of the cardiac signal, they still had some of the higher frequencies of cardiac noise in the pass band range.

Figure 5.2 shows the block diagram of the developed apnoea monitor capable of identifying the presence and absence of breathing.

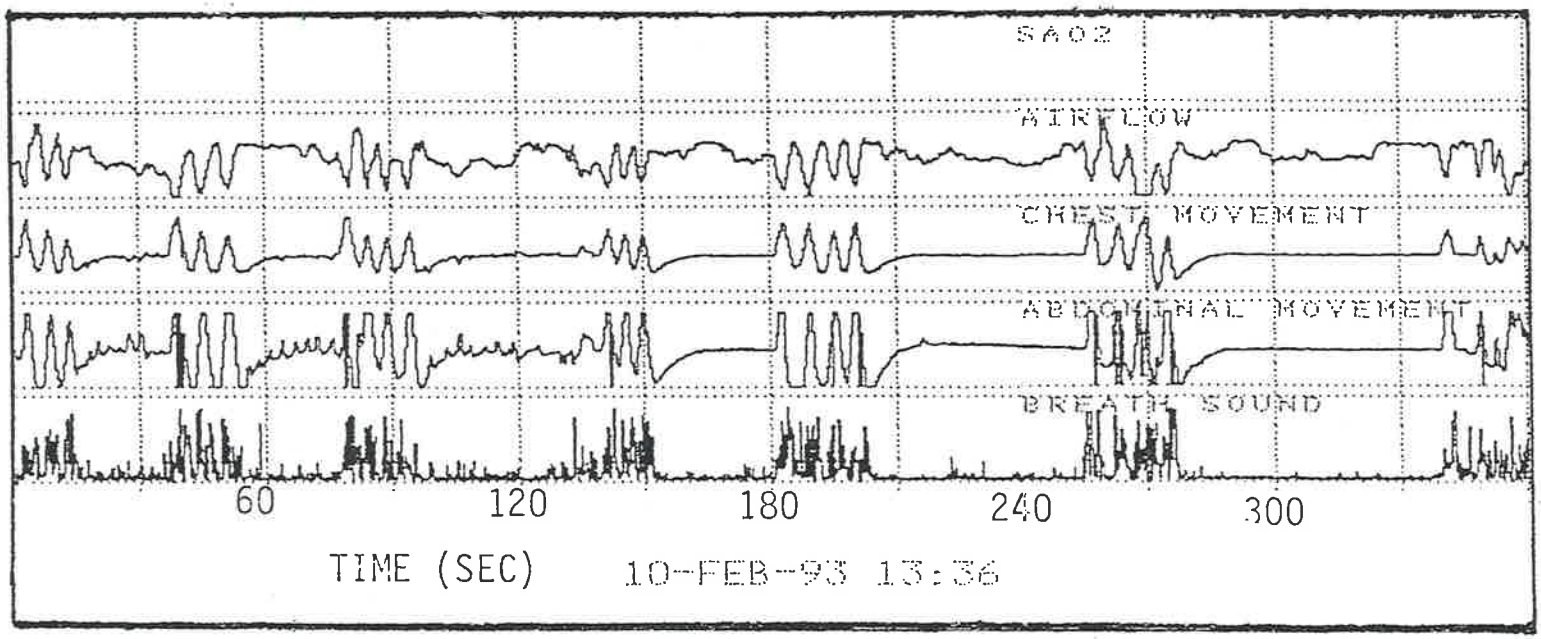


Fig. 5.1 Real-time filtered respiration signals, as compared with airflow, chest movement and abdominal movements, recorded at the Sleep Laboratory at the RAH .

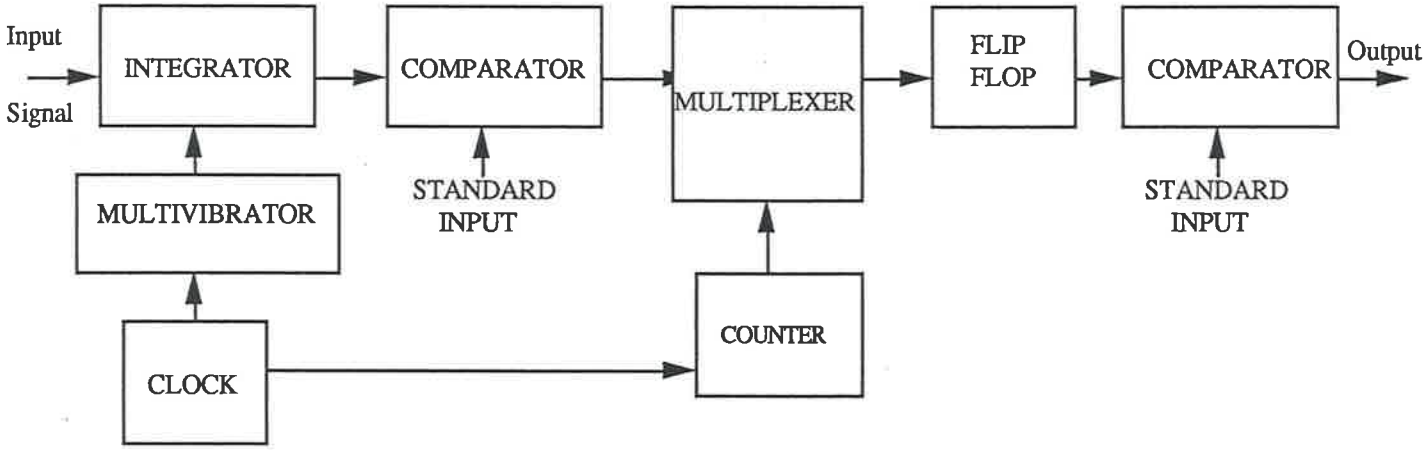


Fig. 5.2 Block diagram of the apnoea monitor.

5.3 The Apnoea Monitor

Figures 5.3, 5.4, and 5.5 show the schematic circuit diagrams of the apnoea monitor drawn in three stages. The monitor involves the following basic procedures and operations:

Principle of Operation

Respiratory signals are acquired from the microphone and, after band pass filtering, are rectified for further processing. The respiratory signals are sampled every 4 seconds and a preliminary decision is made whether the signals are present or absent. Also, the sampled signals are analyzed for a total of 64 seconds, and if the cumulative level of output during this period is below the standard preset value, the device indicates inadequate respiration. Thus the device maintains the history of the past 64 seconds at any point of time while also indicating the current cumulative respiration level. Any inadequacy is indicated in the form of an light emtting diode(LED) alarm.

Sampling

Analogue respiratory signals are fed into the device as an output from the filter and the rectifier. Pulses are generated in 4 second intervals by a clock with 4 second time period in conjunction with a monostable multivibrator. Thus the signals are sampled in between consecutive pulses of a 4 second time period.

Integration

Respiratory signals are not of a constant magnitude and basically resemble an envelope in the time domain. For studying the magnitude of the signals, the signals are averaged over the time period (4 seconds in this case) of measurement. Thus the signals are integrated over a period of 4 seconds, just in time before the origin of subsequent pulse.

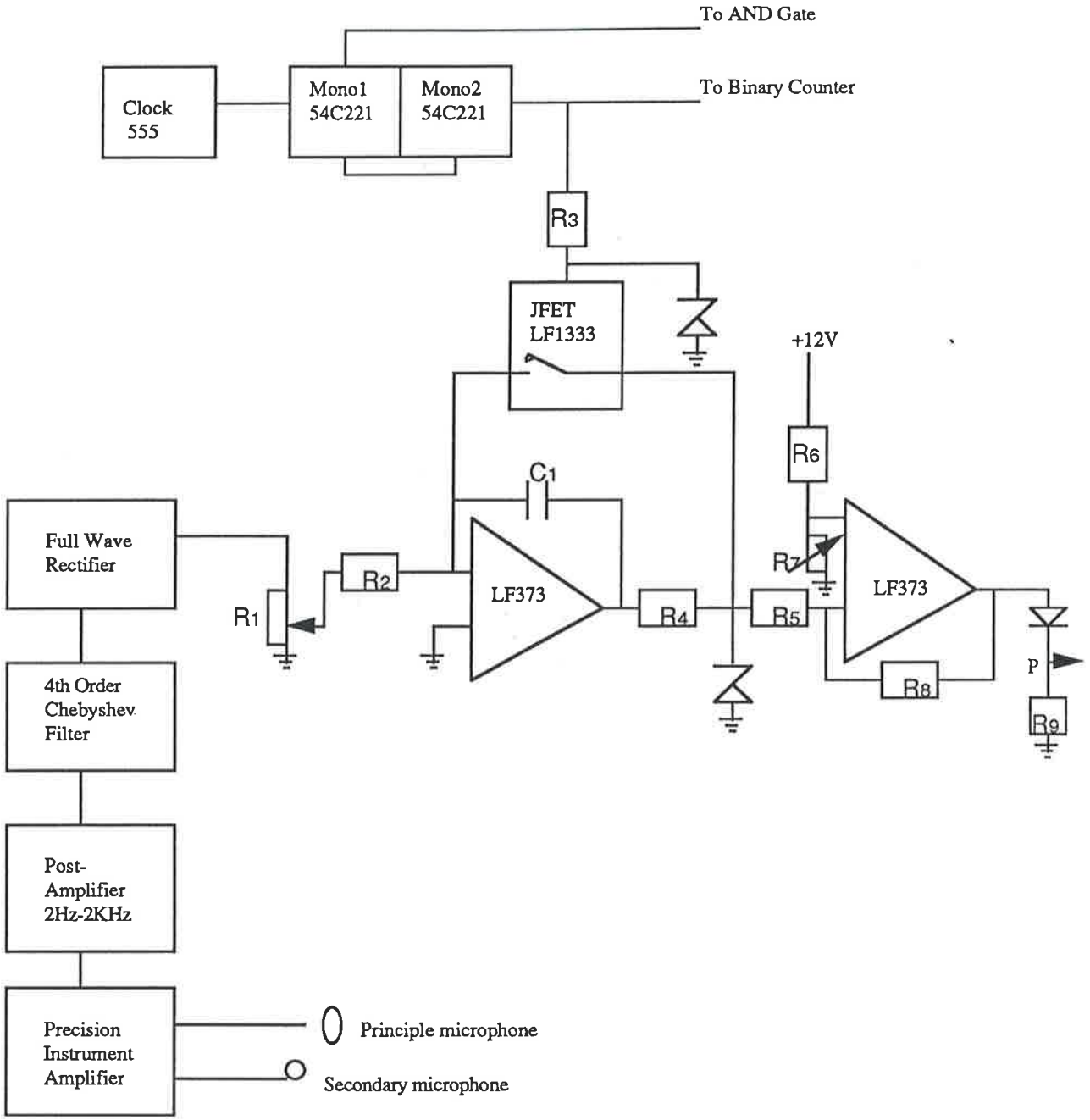


Fig. 5.3 Schematic layout of the apnoea monitor (Stage 1).

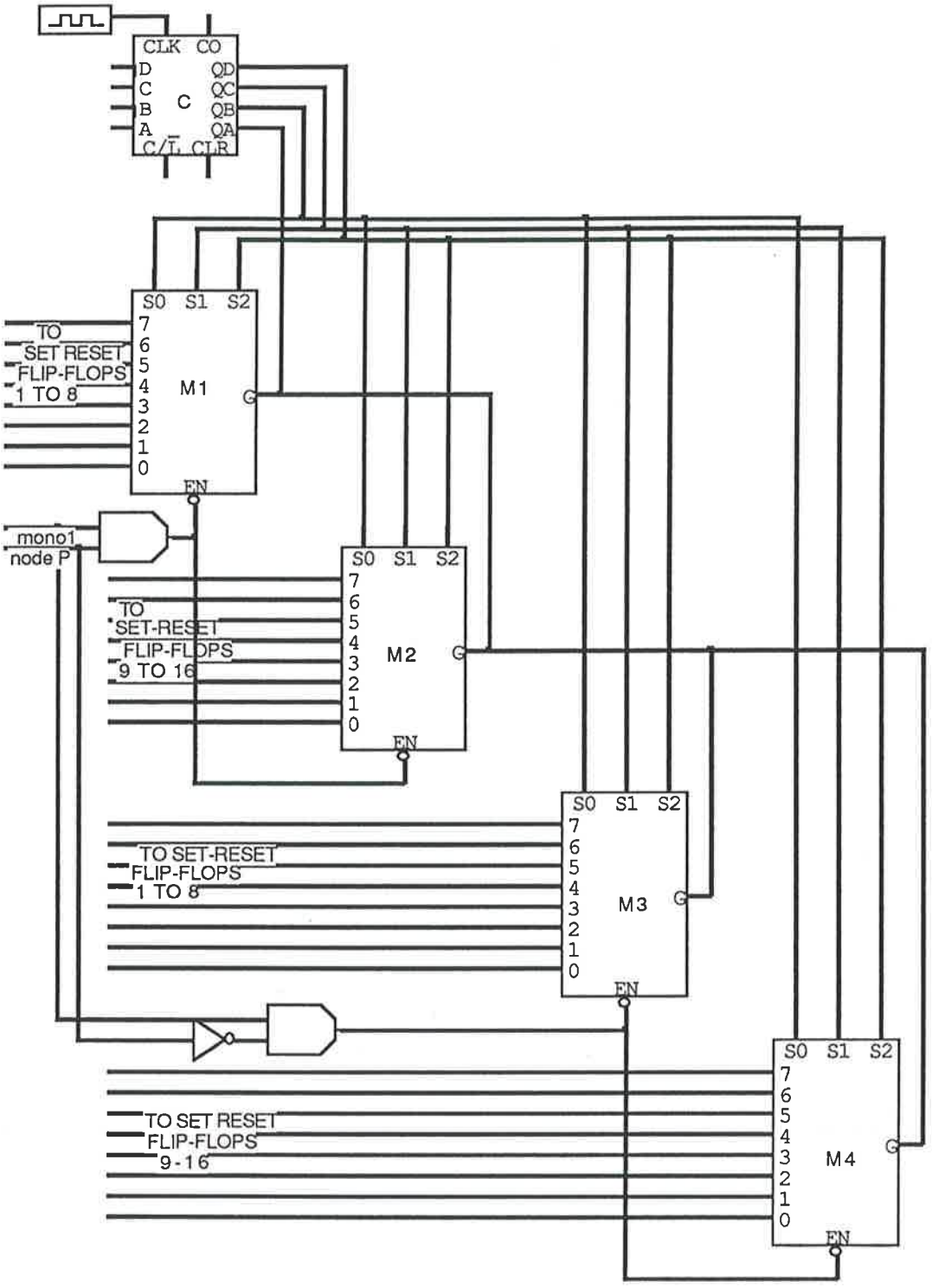


Fig. 5.4 Schematic sketch of multiplexers in the apnoea monitor.

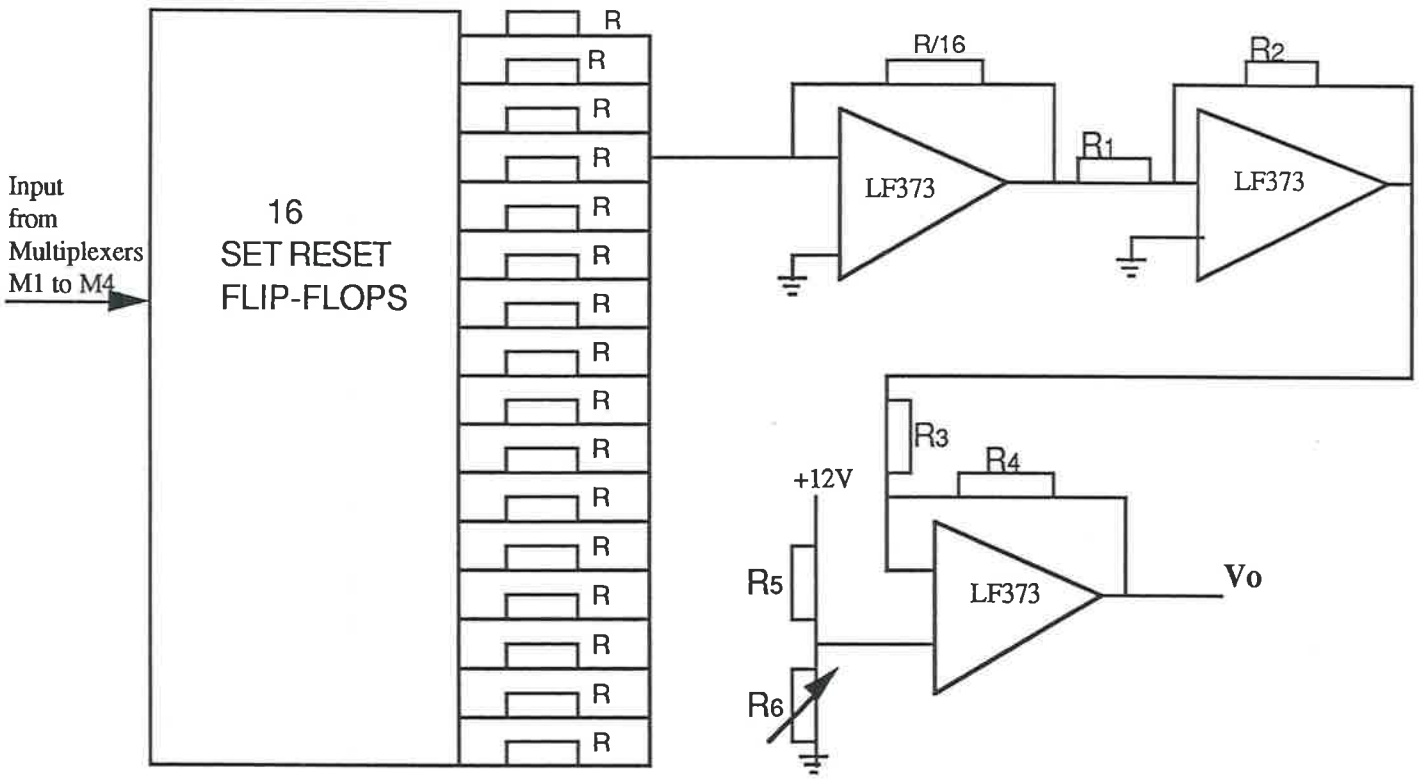


Fig. 5.5 Schematic circuit of the apnoea monitor (Stage3).

Comparator Logic

The comparator output is passed through an AND gate with the pulse from the second multivibrator in order to synchronize the two events with the multiplexer operation. If the output of the AND logic is 1, it indicates the presence of respiration. The multiplexer outputs are activated by the binary counter, counting from 0 to 15 and reseting itself thereafter. The output of the AND gate is transferred across to the multiplexer output irrespective of whether it is low or high (i.e. 0 or 1). Thus 16 different output conditions (some high some low, or all high or all low) are generated at the multiplexer output terminals after 64 seconds.

Set Reset Logic

The 16 outputs of the multiplexer can have high (1) or low (0) states at any given instant. After every 64 second interval, the outputs of the multiplexer can change depending on the signal level. For example, if the output at terminal 1 of the multiplexer was high (1) during the previous 64 seconds and if the signal level falls, there will be a change of state of the terminal 1 from high to low for the next 64 seconds. This logic holds for all of the 16 output terminals in succession over every 64 second period. Thus the set/reset operation switches the change of state or maintains the original, which ever is applicable.

Addition

The output of the set-reset stage can have all 16 outputs high or some of them low and some high or all 16 low. The output states are added to the present level (if the present level is less than 16), the maxima corresponding to all 16 outputs being high. The added value can change every 4 seconds if the respiratory signals vary within these 4 seconds.

Decision

A final decision as to the adequacy of respiration is made based on the added value of the 16 outputs over 64 second time period. A standard preset value (5 Volts) is stored, which represents an adequate level of respiration. The added value is compared to this standard value every 4 seconds and a final decision is made indicating either the absence or inadequacy of respiration, or the presence of respiration, for any subject. An LED glows as long as the respiration level is equal or above 5 volts but, switches off if the level fell down below 5 volts.

A gain control is provided on the input side of the device which monitors the gain of the input signal and in turn controls the sensitivity of the device. The device can be calibrated to suit the ambient conditions and the individual prior to on-line usage. There is a meter display on the device which indicates the level of output, the full deflection corresponding to 10 volts of the signal.

Operation

After adjusting the sensitivity of the device to suit the subject and the ambient conditions, the monitor is ready for on-line operation. The device samples the signal every 4 seconds and makes a decision of the adequacy of its magnitude. Subsequently, the information is processed and finally indicated on the dial display and the LED display.

The sensitivity of the instrument is a critical parameter and directly affects its performance. The strength of the signal is an important factor which governs the sensitivity of the instrument. It must be adjusted on the subject before it can be used for on-line monitoring. The gain knob is turned up or down such that there is no deflection on the meter scale in the absence of respiration. In other words, the meter should show zero deflection in the ambient conditions (when there is noise) in the absence of the signal.

The preset value of the alarm setting also is a critical parameter which must be predetermined. This preset value can be kept fixed and may not be changed during the monitoring of an individual. It is important, however, that the device does not give false alarms, in case of the presence of respiration for example.

CHAPTER 6

CONCLUSION

6.1 Conclusive Results of Tests

The developed apnoea monitor initially was tried on a range of normal subjects, mainly adults, in the Bio-engineering laboratory of the Department of Electrical and Electronic Engineering. Subsequent trials were made at the Sleep Laboratory and the clinical wards of the RAH.

In the trial, the microphone was carefully placed on the upper neck position (next to the larynx) of the subject and strapped into place. The acquired signals could be visualized on the CRO to confirm an adequate signal from the correct positioning of the transducer. The subject was seated and asked to breathe normally to ensure normal signal level. The outputs of the apnoea monitor from various stages (rectified signal, integrated output, comparator output, and final decision) were monitored on the chart recorder. During the trials the subjects also were asked to hold their breath for as long as they could, simulating the conditions of actual apnoea. These tests were conducted on each subject for a short duration of 10 to 15 minutes. The monitor was set to sound an alarm if the cumulative signal level fell below half of the normal value (10 Volts). In other words, the monitor had been set to sound an alarm if 8 of the 16 sampled inputs indicated no respiration signal. The study was conducted upon 12 subjects and indicated

that the monitor could identify apnoea in 10 out of 12 subjects. The total number of events of simulated apnoea recorded were 40, out of which 32 could correctly be identified. There were three false alarms indicating a false absence of respiration. In the case of 2 subjects, the monitor marginally failed to sound the alarm.

Further, the device also was tested out in the clinical wards of the RAH where the sensitivity of the instrument was adjusted to suit the ambient noise conditions. The device was tested for a duration of about five minutes on 8 normal adult subjects who were laying at rest. There were 15 events of simulated apnoea of which 12 correctly were identified. There were no false alarms.

A third test was conducted in the Sleep Laboratory of RAH in conjunction with other measurements which included the use of pressure pads, thermistors, impedance pneumograph and closed circuit television. The instrument was placed upon 1 subject for a duration of one hour. Tests were done to detect simulated central and paradoxical (obstructive) apnoea. The monitor responded correctly in all 5 events of central and obstructive apnoea.

Three prolonged recordings were conducted in the Bio-engineering laboratory using 3 different subjects and each recording lasting about two hours in duration. Each subject was asked to lie down in an electrically isolated room equipped with an external microphone to record any noise (including snoring) during the test. Thermistors were placed next to the nostrils to enable a comparative study . This trial tested the device for 35 events of apnoea, of which 29 events correctly were identified. On two occasions the thermistor did not respond due to the subject breathing through his or her mouth.

In all, there were 95 events of central and obstructive apnoea, of which 81 correctly were identified, indicating an 82% successful rate of performance.

6.2 General Conclusion

The monitor performs satisfactorily in low noise conditions. The device cannot distinguish between a continuous artifact of a similar frequency such as respiration, and may generate false alarms. The secondary microphone compensates for external noise to the extent of 50%. The sensitivity of the instrument must be controlled using an adjusting knob depending upon the ambient noise and the intensity of the acquired signal. During the tests, the sensitivity had to be changed among individuals at the same location and also at a different locations where different noise levels were present.

The large variations that exist in biomedical signals force us to rely heavily upon statistical methods. These variations exist in signals acquired from the same individual, and of course from among populations. The positioning of the transducer is an important parameter and can directly affect the performance of the monitor. Also, the tests carried out were in low noise conditions, whereas the device must operate in noisy conditions, in which case the external noise may be detrimental. The recordings were conducted while the subject's body was in a prone position and his/her movements restrained.

6.3 Future Work

The decision whether a person is breathing adequately or not is not a straightforward issue. The decision is subjective to the person under observation. Moreover, it is difficult to formulate a general breathing pattern covering all types of subjects, as there are many erratic patterns. The important issue at this stage will be to identify or segregate the respiration signal from any other artifact.

Work has to be extended to infants using a similar transducer, but the site needs to be changed due to inadequacy of space on the neck of infants. Up to now, not much

emphasis has been placed on the optimization of transducer design and its constructional features. However, this issue can be taken up at a later stage.

As apnoeic events have a direct impact on heart rate, a dual modality monitor will be more reliable, and more information can be derived correlating the data from two interdependent systems (Werthammer et al., 1983; Liu et al., 1987). This can be especially significant for SIDS study. After satisfactorily implementing the apnoea monitor, the project can be extended in potential areas of respiratory disease identification such as asthma and tuberculosis.

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APPENDIX

The following routines perform the FFT and estimation of the POWER SPECTRAL DENSITY (PSD) :

```
#include <stdio.h>
#include <math.h>
#include <stdlib.h>

void realft(double *data,int n,int isign);
void four1(double *data,int nn,int isign);
void windowcalc(double *window, int n, int windowtype);
void fourpwr(double *data, double *wind, int n);
void fpwrcontroller(FILE *fp, int *data_inp, double *spect, int n, int k, int overlap, \
    double offset, double *window, double wsum,double *msqampraw,double
*msqampwind, \
    double *spectpwr);
int readfiledata(FILE *fp, int *data_inp, double *data, int n, double offset);


#define SWAP(a,b) tempr=(a);(a)=(b);(b)=tempr

void four1(double *data,int nn,int isign)
{
    int n,mmax,m,j,istep,i;
    double wtemp,wr,wpr,wpi,wi,theta;
    double tempr,tempi;

    n=nn << 1;
    j=1;
    for (i=1;i<n;i+=2) {
        if (j > i) {
            SWAP(data[j],data[i]);
            SWAP(data[j+1],data[i+1]);
```

```
            }
            m=n >> 1;
            while (m >= 2 && j > m) {
                j -= m;
                m >>= 1;
            }
            j += m;
        }
        mmax=2;
        while (n > mmax) {
            istep=2*mmax;
            theta=6.28318530717959/(isign*mmax);
            wtemp=sin(0.5*theta);
            wpr = -2.0*wtemp*wtemp;
            wpi=sin(theta);
            wr=1.0;
            wi=0.0;
            for (m=1;m<mmax;m+=2) {
                for (i=m;i<=n;i+=istep) {
                    j=i+mmax;
                    tempr=wr*data[j]-wi*data[j+1];
                    tempi=wr*data[j+1]+wi*data[j];
                    data[j]=data[i]-tempr;
                    data[j+1]=data[i+1]-tempi;
                    data[i] += tempr;
                    data[i+1] += tempi;
                }
                wr=(wtemp=wr)*wpr-wi*wpi+wr;
                wi=wi*wpr+wtemp*wpi+wi;
            }
            mmax=istep;
        }
    }

#undef SWAP
```

```
void realft(double *data,int n,int isign)
{
    int i,i1,i2,i3,i4,n2p3;
    double c1=0.5,c2,h1r,h1i,h2r,h2i;
    double wr,wi,wpr,wpi,wtemp,theta;
    void four1();

    theta=3.141592653589793/(double) n;
    if (isign == 1) {
        c2 = -0.5;
        four1(data,n,1);
    } else {
        c2=0.5;
        theta = -theta;
    }
    wtemp=sin(0.5*theta);
    wpr = -2.0*wtemp*wtemp;
    wpi=sin(theta);
    wr=1.0+wpr;
    wi=wpi;
    n2p3=2*n+3;
    for (i=2;i<=n/2;i++) {
        i4=1+(i3=n2p3-(i2=1+(i1=i+i-1)));
        h1r=c1*(data[i1]+data[i3]);
        h1i=c1*(data[i2]-data[i4]);
        h2r = -c2*(data[i2]+data[i4]);
        h2i=c2*(data[i1]-data[i3]);
        data[i1]=h1r+wr*h2r-wi*h2i;
        data[i2]=h1i+wr*h2i+wi*h2r;
        data[i3]=h1r-wr*h2r+wi*h2i;
        data[i4] = -h1i+wr*h2i+wi*h2r;
        wr=(wtemp=wr)*wpr-wi*wpi+wr;
        wi=wi*wpr+wtemp*wpi+wi;
    }
    if (isign == 1) {
```

```
            data[1] = (h1r=data[1])+data[2];
            data[2] = h1r-data[2];
        } else {
            data[1]=c1*((h1r=data[1])+data[2]);
            data[2]=c1*(h1r-data[2]);
            four1(data,n,-1);
        }
}


void windowcalc(double *window, int n, int windowtype)
/*calculates the window function values and places them in window[1...n]*/
/*according to window type: 1=square, 2=parzen, 3=welch, 4=hanning, 5=hamming*/
{
int j;
double a,b,c,twopi;


a=(double)(n/2)-0.5;
b=1.0/((double)(n/2)+0.5);
switch (windowtype)
    {
    case 1:  /*square*/
            for(j=1;j<=n;j++)
                {
                window[j]=1.;
                }
            break;
    case 2:                 /*parzen*/
            for(j=1;j<=n;j++)
                {
                window[j]=(1.0-fabs((((j)-1)-(a))*(b)));
                }
            break;
    case 3:  /*welch*/
            for(j=1;j<=n;j++)
```

```
                {
                c=((((j)-1)-(a))*(b));
                window[j]=(1.0-c*c);
                }
        break;
case 4:                 /*hanning*/
        b=(double)(n-1);
        twopi=2.*acos(-1.0);
        for(j=1;j<=n;j++)
                {
                window[j]=0.5*(1-cos(twopi*(double)(n-j)/b));
                }
        break;
case 5:  /*for hamming*/
        b=(double)(n-1);
        twopi=2.*acos(-1.0);
        for(j=1;j<=n;j++)
                {
                window[j]=0.54-0.46*cos(twopi*(double)(n-j)/b);
                }
        break;
default:  /*error-square*/
        for(j=1;j<=n;j++)
                {
                window[j]=1;
                }
        break;
}
return;
}



void fourpwr(double *data, double *wind, int n)
/*takes real data in the array data[1...n] and windows the data with the*/
```

```
/*window function in wind[1...n], then calculates the power spectrum which it */
/*returns in data[0....n/2].*/

{
int i,j;
double powern;

for(i=1;i<=n;i++)
      {
      data[i]*=wind[i];
      }

realft(data,n/2,1);   /*calculate fft*/

data[0]=(data[1]) * (data[1]);  /*energy at f=0*/
powern=data[2]*data[2];   /*energy at f=n/2*/
for(i=4,j=1;i<=n;i+=2,j++)
      {
      data[j]=2*(data[i-1]*data[i-1]+data[i]*data[i]);
      }
data[n/2]=powern;
return;
}

void fpwrcontroller(FILE *fp, int *data_inp, double *spect, int n, int k, int overlap, \
     double offset, double *window, double wsum,double *msqampraw,double
*msqampwind, \
     double *spectpwr)
/*reads n*k values from file fp, the data are integer*/
/*Values are converted to double, multiplied by the*/
/*window,fft performed on n values k times and averaged, and returned in */
/*spect[0...n/2]*/
{

int nhalf, frc,i,j;
double *datam, den=0.0,xx;
```

```
double sumtempraw,sumtempwind,sumraw,sumwind,kspectpwr,pwradj;

nhalf=n/2;
datam=vector(0,n);
for(j=0;j<=nhalf;j++)  spect[j]=0.0;

sumraw=0.;
sumwind=0;

for(i=1;i<=k;i++)
      {

      frc=readfiledata(fp, data_inp, datam, n, offset);

      sumtempraw=0;
      sumtempwind=0;
      for(j=1;j<=n;j++) {

      (xx=datam[j]*datam[j],sumtempraw+=xx,sumtempwind+=xx*window[j]*window
[j]);
                }
      sumraw+=(sumtempraw/(double)n);
      sumwind+=(sumtempwind/(double)n);

      fourpwr(datam, window, n);   /*calculate power spectrum*/

      kspectpwr=0.;
      for(j=0;j<=nhalf;j++) kspectpwr+=datam[j]; /*temporary power total*/
      pwradj=(sumtempraw/(double)n)/kspectpwr; /*correction factor to normalise
spectrum*/
      for(j=0;j<=nhalf;j++) datam[j]*=pwradj;  /*normalise power spectrum*/

      for(j=0;j<=(nhalf);j++)  spect[j]+=datam[j];  /*sum results*/

      }
```

```
for(j=0;j<=(nhalf);j++)  spect[j]/=(double)k;
*spectpwr=0.;
for(j=0;j<=nhalf;j++) *spectpwr+=spect[j];

*msqampraw=sumraw/(double)k;
*msqampwind=sumwind/(double)k;
free_vector(datam,0,n);
return;
}



int readfiledata(FILE *fp, int *data_inp, double *data, int n, double offset)
{

size_t no_items_read;
int j;


no_items_read=fread(&data_inp[1], (size_t)sizeof(int), (size_t)n,fp);
if(no_items_read!=n) nrerror("error reading in buffer of data\n");
for(j=1;j<=n;j++)
      {
      data[j]=((double)(data_inp[j])-offset);
      }
return 0;
}
```