A HIGH RESOLUTION PHONOCARDIOGRAPH INCORPORATING DIGITAL RECORDING AND ANALOG SPECTRAL ANALYSIS

By

MICHAEL PHILLIP MANKE

Bachelor of Science

Oklahoma State University

Stillwater, Oklahoma

1975

Submitted to the Faculty of the Graduate College
of the Oklahoma State University
in partial fulfillment of the requirements
for the Degree of
MASTER OF SCIENCE
December, 1977

A HIGH RESOLUTION PHONOCARDIOGRAPH INCORPORATING
DIGITAL RECORDING AND ANALOG SPECTRAL ANALYSIS

Thesis Approved:

R. L. Lowery
Thesis Adviser

K. N. Reid

Larry D. Zirkle

Norman N. Durham
Dean of the Graduate College

PREFACE

This study is concerned with the development of a phonocardiographic instrumentation system. The goal is to provide heart sound recordings with readily available frequency spectra of the selected signals. A possible use of the system is in the area of identification and diagnosis of heart murmurs.

I wish to express my thanks to my major adviser, Dr. Richard L. Lowery, for his patience and assistance throughout the long period of the study. I appreciate the assistance of the other committee members, Dr. Larry D. Zirkle and Dr. Karl N. Reid in preparing the final manuscript.

I thank Dr. James W. Carley, M.D., for his welcomed assistance. I thank Dr. David Sander and Miss Susan Dillon for their cooperation and the invaluable aid they provided.

I appreciate the cooperation and assistance of Miss Charlene Fries in preparing the early draft and the final manuscript.

I offer special thanks to my father, Professor Phillip G. Manke, for his advice and counsel and to my mother, Bonnie, and the rest of my family for their encouragement.

A very special thanks I give to my wife, Nancy, for her support and encouragement, in all my endeavors, throughout this study.

TABLE OF CONTENTS

LIST OF FIGURES

Figure | Page

LIST OF SYMBOLS

ac	alternating current
A/D	analog-to-digital
A-V	atrio-ventricular (heart valves)
B	filter bandwidth
CRT	cathode ray tube
db	decibels
dc	direct current
DAC	digital-to-analog converter
DWR	digital waveform recorder
EKG	electrocardiogram
FFT	fast Fourier transform
FM	frequency modulation
Hz	Hertz or cycles per second
K	time constant of resistance-capacitance network
N	number of digital samples
PCG	phonocardiogram
P, QRS, T waves	respective components of EKG
R	digital memory output rate
RC	resistance-capacitance
t	fundamental period of periodic wave
T	record length in swept filter analysis
TTL	transistor-transistor logic

CHAPTER I

INTRODUCTION

Phonocardiology

Phonocardiology is the study of heart vibrations, seen as graphical oscillations on a phonocardiograph, or interpreted by a physician as sounds, through auscultation with the aid of a stethoscope. The graphical oscillations, called a phonocardiogram (PCG), are detected by various electronic devices. Heart sound examination, usually by auscultation, is included in most patient physical examinations. The purpose is to detect abnormal sounds resulting from deficiencies in the heart mechanism. The sounds may or may not affect the health of the patient.

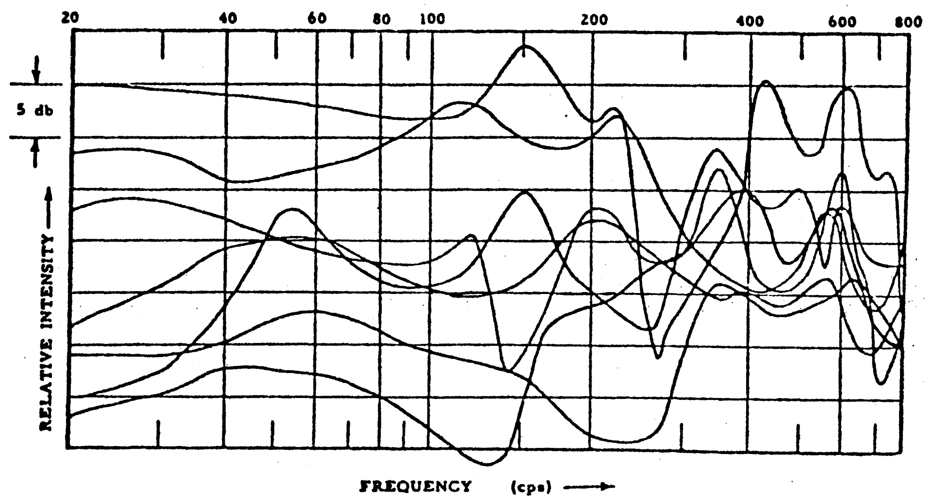
Much controversy exists between practicing physicians and phonocardiographers concerning the relative merits of the two forms of diagnosis, auscultation and phonocardiography. Phonocardiographers point out the inaccuracy of auscultation because of the response limitations of both stethoscopes and the human ear. Physicians are reluctant to rely on electronic methods, because early phonocardiographs and electronic stethoscopes had poor performance characteristics as well. Phonocardiographic analysis does not enjoy widespread acceptance for that reason.

If a murmur is intense or fairly long in duration, it is usually detectable with a stethoscope. However, if the murmur is weak or of short duration, and occurs very near one of the loud normal heart sounds, it may not be easily identified if heard at all. Other errors that can be made in auscultation are hearing sounds that are nonexistent, and misinterpreting sounds. Luisada (15) cited several studies on the frequency of errors in auscultation and concluded that it is very high.

In an effort to eliminate errors, many researchers have attempted to identify and diagnose heart sounds and murmurs by various electronic methods. Frequency analysis has been employed in an attempt to classify particular heart sound/vibrations, such as aortic stenosis, by its main vibratory components. The premise is that electronic instrumentation, including microphones, amplification devices, recorders, and frequency analyzers, can do a more effective job of reproducing heart sounds and determining the frequency content, than can the combination of stethoscope and ear.

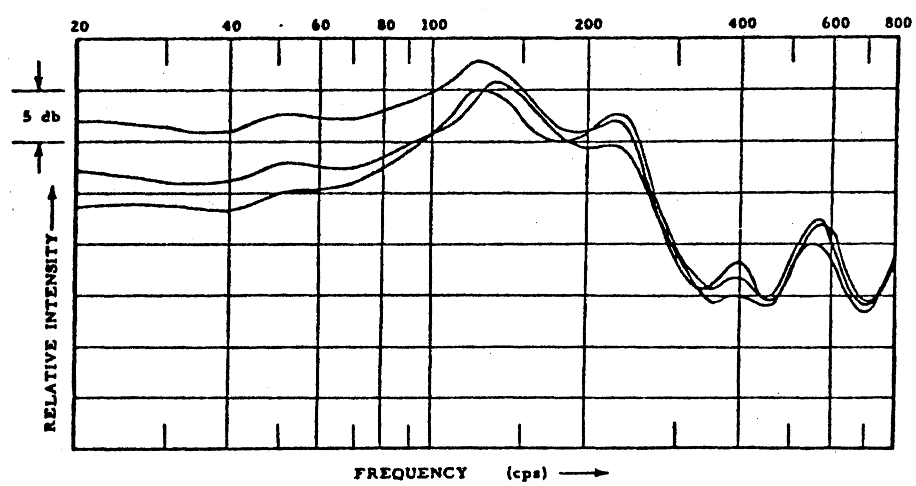
The phonocardiographers are correct in their analysis of auscultation. It is appropriate to present an example of just how poorly a standard stethoscope reproduces sounds. Figure 1 compares response curves of several different stethoscopes. Along with the considerable nonlinearity of the responses, two stethoscopes may differ by as much as 35 db in their ability to reproduce a particular frequency sound. The differences in response are presumably due to various resonances of the tubing and fittings. Even stethoscopes of the same brand and model have considerable variations in frequency response (Figure 2). No two stethoscopes are exactly alike in characteristics.

Since the stethoscope is apparently unable to produce an accurate reproduction of the heart sound, it is easy to see why the errors in diagnosis exist. The response of the human ear complicates the matter further. Figure 3 illustrates the threshold of audibility of the ear,



Source: C. A. Caceres et al., <u>The Innocent Murmur</u> (1967).

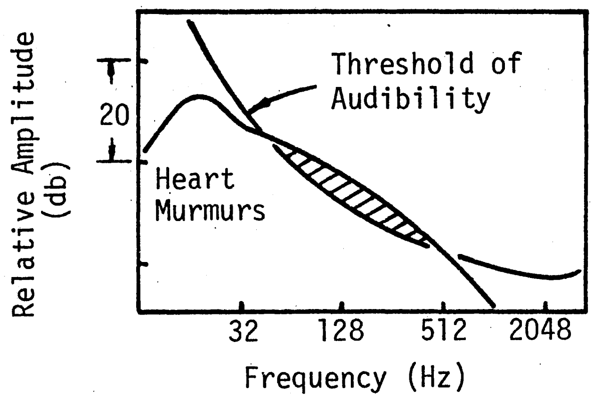
Figure 1. Response Curves of Several Stethoscopes



Source: C. A. Caceres et al., <u>The Innocent Murmur</u> (1967).

Figure 2. Response Curves of Three Stethoscopes of the Same Brand and Model

along with the presumed frequency range and intensity of heart sounds and murmurs.



Source: A. C. Guyton, Textbook of Medical Physiology (1971).

Figure 3. Threshold of Audibility and Intensity and Frequency Range of Murmurs

It is obvious that the combination of ear and stethoscope is ill-suited for accurate analysis of a majority of heart sounds. The heart sounds are modified by the stethoscope and human auditory apparatus. Moreover, the interpretation of the sounds is influenced by the physician's experience and background (5).

The result is that few physicians hear the same sounds on auscultation, and differing diagnoses may occur for the same patient, all equally in error. It is evident that auscultation should be used with care and reservation and should never be relied upon for conclusive evidence of any particular condition.

Many phonocardiographic methods have been used to study heart sounds. The electronic instrumentation is widely variant, and the resulting nonstandardization of technique may account for some of the reluctance in accepting phonocardiographical analyses. A survey of previous methods provides an indication of the various methods in use.

Literature Survey

Caniggia et al. (6) used electronic band-pass filtering of the heart sound signal obtained from a crystal microphone. Five octave band filters separated frequency components which were displayed and photographed on a high speed oscilloscope (sweep speed up to 15 meters/second). The high speed of the system overcame the inability of standard strip-chart recording, at 100 millimeters/second speed, to accurately reproduce frequencies of 500 Hertz (Hz) and higher. The relative amplitudes of the signal present in each of the filter bands were compared. Results from Caniggia's work indicated the main frequencies of mitral murmurs are in the 100 to 150 Hz range. Aortic insufficiency murmurs were mostly 150 to 250 Hz vibrations.

Research by Taylor (22) indicated heart sound/vibrations are nonstationary, random vibrations with strong periodicities. Rushmer (19) first mentioned the random nature of heart sounds in 1968. Ideally, every cardiac cycle would be exactly the same. Unfortunately, this is not the case. The transient signal complicates a meaningful frequency analysis because of the spectral variance. This problem is overcome by continuously repeating particular individual cycles of heart sounds. The usual method of accomplishment of this task is via analog tape loops.

Van Vollenhoven et al. (23), using magnetic tape loops of selected signals, performed an analysis of murmurs with an analog wave analyzer. The center frequency of the analyzer, similar to an adjustable band-pass filter, was varied in 50 Hz steps to produce a kind of frequency spectrum. Their method was successful in identifying several cases of aortic insufficiency in the presence of mitral stenosis.

Harris (11) used a band-pass network of 11 filters covering a frequency range from 80 to 220 Hz for frequency analysis. The magnetic tape loop was again employed for signal reproduction.

Jacobs (12) used a zero-crossing detector in his automated method of analysis, borrowing a technique demonstrated to be very accurate in speech analysis. Results in a group survey with this method indicated a 95% correct diagnosis of normal subjects, and a 94% correct diagnosis of abnormal subjects. One limitation of the system was its relative insensitivity to soft aortic insufficiency murmurs.

Another zero-crossing method, used by Abelson (1), developed an analog voltage proportional to the frequency of zero-crosses. The method produced a "frequency phonocardiogram" simultaneously with the conventional phonocardiogram.

Recent studies have made use of the digital computer for analysis procedures. Frome and Fredrickson (7) took analog-recorded dog heart sounds from FM tape and digitized them at a sample rate of 5000 points per second. A simultaneous EKG signal was used as a time reference. Digital filtering was employed to eliminate low frequency (<30 Hz) and high frequency (>500 Hz) components of the signal. A computer FFT (Fast Fourier Transform) routine performed the spectral analysis of first and second heart sounds. The spectra indicated strong components for both first and second sounds at approximately 50 Hz. The effective band-width resolution of the FFT was 8.3 Hz.

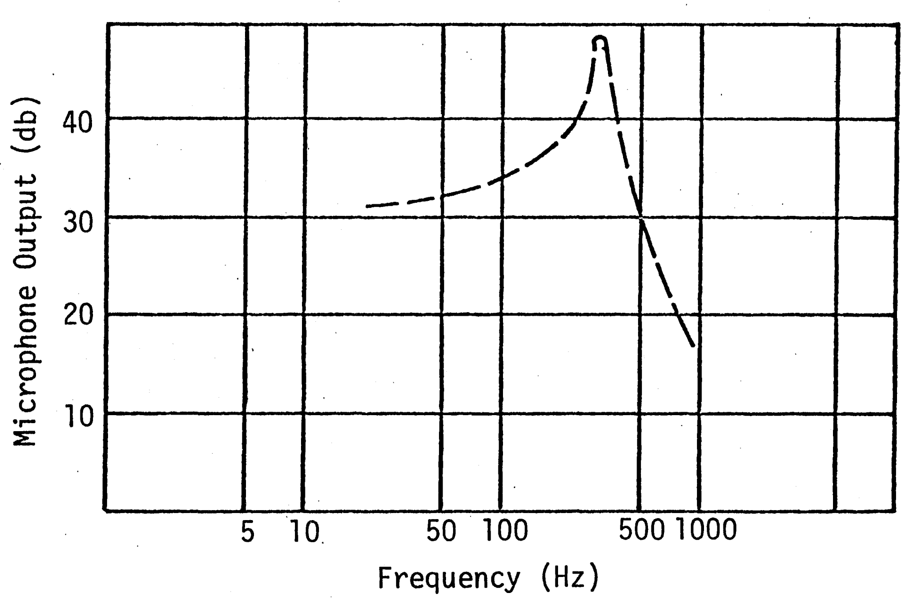
Yoganathan et al. (24) used two computers in their study, a PDP-10 and an IBM 370, which included FFT analysis. The PDP-10 handled the digitizing of FM recorded heart sounds from a group of normal male subjects. A contact microphone was used to obtain the signals which were high-pass filtered (12 db/octave) with a cutoff frequency of 200 Hz. The digitizing sample rate was 1000 points per second. Fifteen first heart sound spectra were averaged for each subject and individual frequency peaks tabulated. An averaged spectrum for all the subjects exhibited a broad peak at about 30 Hz. It is interesting to note the portion of the actual signal available for analysis was rather limited, considering the digital sampling rate limitation on the high frequencies and the filter cutoff limitation on the low frequencies.

Summarizing the previous research and techniques, the common practice is to record on tape (standard or FM) heart sounds obtained from a sensor attached to the chest area of a subject. A selected portion of the recorded tape is either made into a repeating loop, or digitized with the aid of a computer. The analysis for tape loops is accomplished with stepped, band-pass filtering of various configurations. For the digitized signal, a computer FFT routine performs analysis tasks.

Statement of the Problem

It has been established previously that although auscultation has clinical advantages, its deficiencies more than warrant application of phonocardiographic analysis techniques to heart sound diagnosis. However, it must not be assumed that any electronic instrumentation system will accurately reproduce heart sounds/vibrations. Herein lies the problem. The performance of phonocardiographic systems used to date is as variable as that of the stethoscopes mentioned previously.

Microphone transducers do not necessarily have flat responses as desired. Indeed, over the frequency range of heart sounds few are able to reproduce effectively frequencies below 20 Hz. The response of the microphone used by Van Vollenhoven (23), presented in Figure 4, is a typical, underdamped, second-order system response. It is highly nonlinear in the critical frequency range from 20 to 1000 Hz. Commonly used recording systems are limited to frequencies above about 20 Hz, unless frequency modulation is employed. But even with FM recording, the system is limited by the response of the microphone used.



Source: E. Van Vollenhoven et al., "Frequency Analysis of Heart Murmurs," Medical and Biological Engineering (1969).

Figure 4. Response of Microphone Used by Van Vollenhoven (23)

Reproduction via standard strip-chart recording is unable to handle high frequencies (>500 Hz). Galvanometer systems can handle high frequencies only if the recorder paper speed is very high (>2 meters/second).

Analog analysis using filter networks is, at best, slow and cumbersome. Low frequencies are impossible to detect, or require unacceptably long analysis times. Resolution is limited (narrow 1 Hz band-pass filters are exceptionally difficult to produce). Analysis via digital computer methods has low frequency capability, but is necessarily expensive and not portable.

The limitations of the instrumentation, the accuracy and resolution sacrifice at the expense of portability and economy, and the limited ability for real time analysis all combine to produce many different system configurations. The literature is evidence of some of the trade-offs necessary. The result is that phonocardiographers can be little more certain of their results and findings than can physicians with stethoscopes.

Objective of Research

The objective of this research is to demonstrate a simple but effective method of heart sound recording with corresponding frequency analysis. The method uses a condenser microphone with a relatively flat response between 2 and 1000 Hz, and a digital waveform recorder (DWR) coupled with a standard heterodyne analyzer to give fast (within a few minutes) signal reproductions and frequency spectra. Results with this hybrid analysis method indicate exceptional ease of operation, low frequency recording capability (limited only by microphone response), and approximately 1 Hz frequency resolution capability.

CHAPTER II

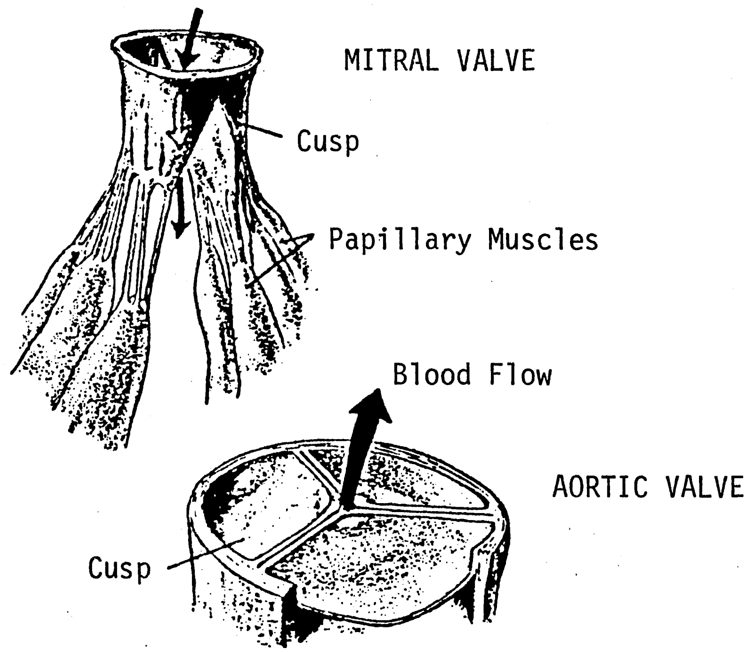
PHYSIOLOGY

Cardiac Cycle

The heart, subject of study for centuries, is basically a pump composed of four chambers, two atria, and two ventricles. The atria function primarily as holding chambers for accumulation of blood before its entrance into the ventricles, although they do pump weakly to help the returning blood on its way. The ventricles provide the main pumping force for circulation. Of the two, the left ventricle is by far the strongest, as it must provide enough force to circulate blood to all parts of the body, except the lungs. The flow of blood through the heart is controlled by four valves, which are thin but very tough, leaflike membranes. The atrio-ventricular or A-V valves (mitral and tricuspid) separate the atria from the ventricles, and the semi-lunar valves (aortic and pulmonic) separate the ventricles from the aortic and pulmonary arteries.

Heart valves are passive in action; that is, they are solely controlled by pressure differentials. A forward differential causes the valve to open, and a reverse differential causes it to close. Figure 5 illustrates the mitral and aortic valves and their construction.

The heart, stimulated to contract and relax by electrical impulses, operates in a continuous pumping cycle. A tracing of the electrical impulses, which can be detected by electrodes attached to the skin, is



Source: A. C. Guyton, Textbook of Medical Physiology (1971).

Figure 5. Mitral and Aortic Heart Valves

called an electrocardiogram, or EKG. The cycle consists of the contraction phase called systole, and the relaxation phase called diastole. Figure 6 shows a normal EKG consisting of the P wave, the QRS complex, and the T wave. The P wave signifies atrial contraction. The QRS complex and the T wave signify, respectively, the contraction and relaxation of the ventricles. The impulse corresponding to the relaxation of the atria occurs about the time of the QRS complex, and is obscured by its larger magnitude.

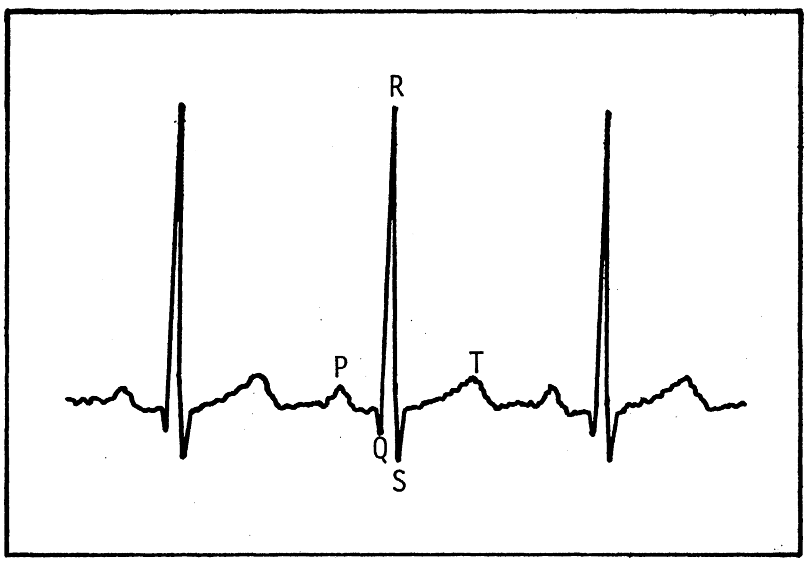
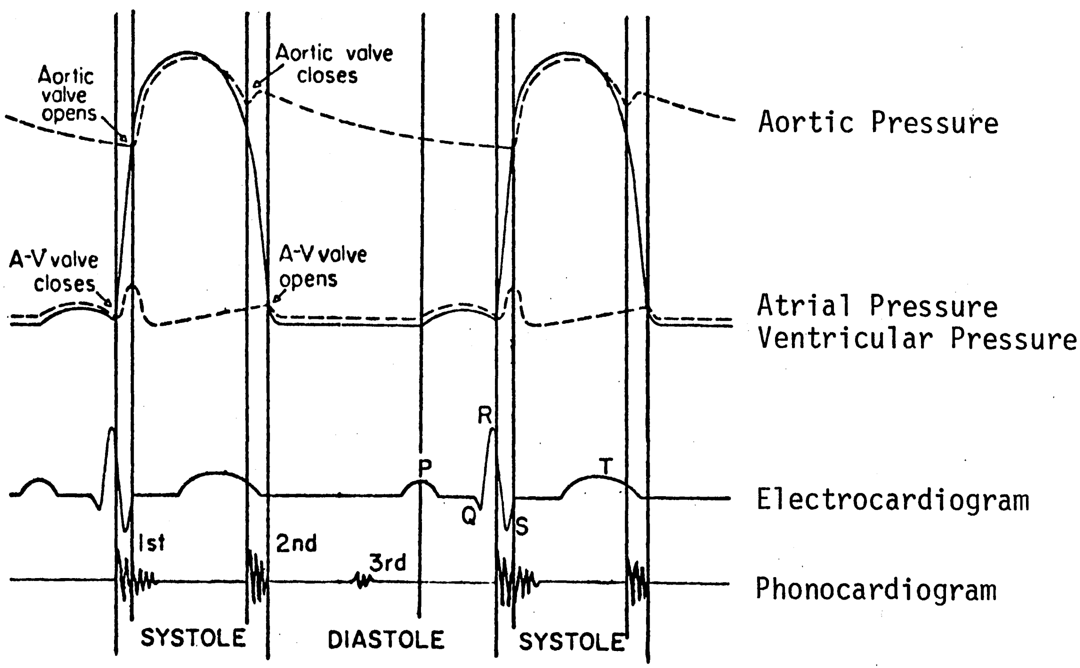


Figure 6. Normal Electrocardiogram (EKG)

Heart Sounds

The pumping action of the heart creates certain vibrations that are interpreted as sounds emanating from the chest wall. A normal heart produces two distinct sounds during one cardiac cycle, the first heart sound and the second heart sound. Although some speculation exists as to the actual causes, it is known that the first heart sound is related to the events that cause the A-V valves to close, and the second heart sound is related to the events that cause the semi-lunar valves to close (20). For a better understanding, Figure 7 illustrates an idealization of the normal EKG along with the corresponding phonocardiogram, or PCG, a graphical representation of the heart sounds. The other traces show the pressure changes which cause the valves to open and close.



Source: A. C. Guyton, Textbook of Medical Physiology (1971)

Figure 7. Idealized EKG, PCG, and Heart Chamber Pressure Variations

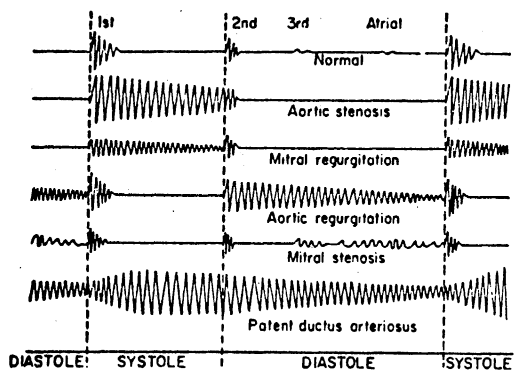
A popular theory for the origin of the heart sounds, accepted until recently, was that of the valve leaflets, or cusps, slapping together upon closure. However, Rushmer (20) proposed the theory of acceleration and deceleration of the heart muscle tissue, the valve surfaces, and associated blood volume to explain the sounds, which was supported by Luisada (14). In later work Luisada (16) concluded that there are as many as five components of the first heart sound ". . . attributes the vibrations of the first heart sound to vibrations of the left ventricular-aortic cardiohemic system, related to accelerations and decelerations . . ." ((16), p. 144). The acceleration and deceleration theory is the most widely accepted view of the origin of the normal heart sounds.

The normal phonocardiogram consists of only two sounds; however, a third heart sound may be detected in many normal children and a few normal adults. The third heart sound is thought to be caused by ". . . ventricular vibrations resulting from an early rapid ventricular filling phase. . ." ((13), p. 79).

Murmurs

Many other heart sounds have been identified. They are usually associated with pathological conditions of the heart, whether congenital, or the result of disease, and are not considered normal. In early research these sounds were referred to as "bellows sounds" because of the high-pitched, blowing sound characteristic of many (5). The sounds are now known as murmurs and are classified in many ways, according to intensity, quality, timing, etc. Some murmurs, referred to as innocent, are caused by normal heart mechanisms and are not considered pathological.

Many murmurs are caused by a mechanical defect in one or another of the heart valves and produce peculiar characteristic sounds. Common defects are stenosis, a condition that prevents the valve from fully opening, and regurgitation, or insufficiency, a condition which prevents the valve from closing properly, allowing blood to back-flow. Rheumatic fever, affecting many parts of the body but especially susceptible areas, such as the thin membranes of the heart valves, is a common cause of valvular defects. Figure 8 illustrates some idealized murmurs along with normal heart sounds.



Source: A. C. Guyton, <u>Textbook of Medical Physiology</u> (1971)

Figure 8. Idealized Heart Murmurs

CHAPTER III

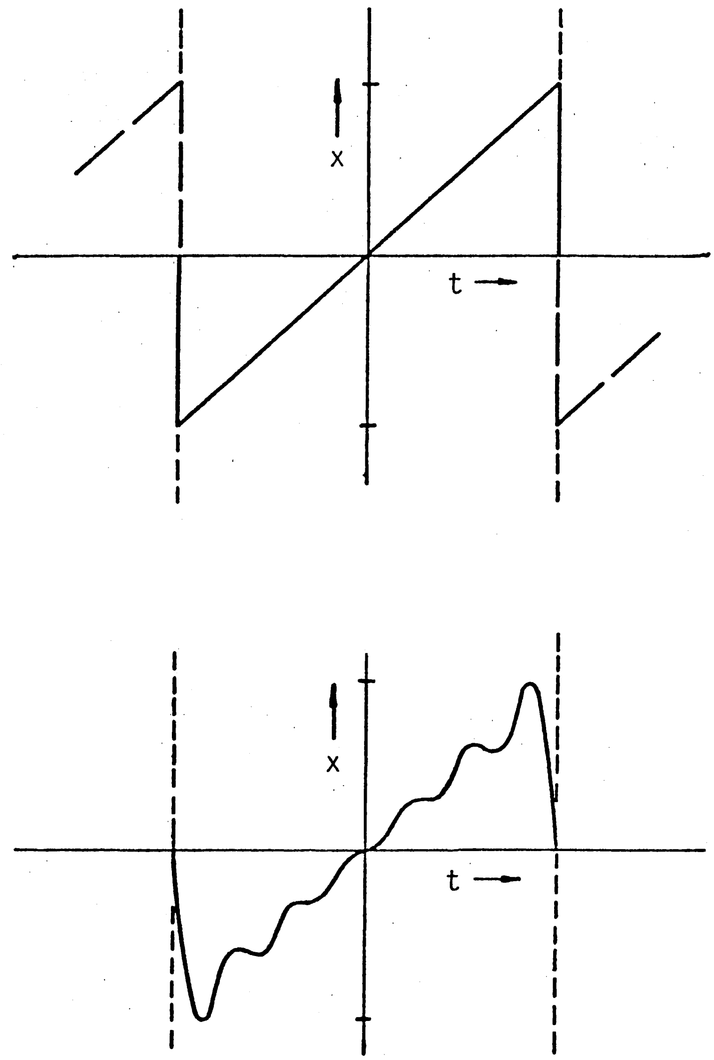
THEORY

Fourier Analysis

Fourier series analysis provides a convenient avenue for spectral analysis. According to Fourier, any complex waveform may be represented mathematically by the sum of series of simpler discrete waveforms, such as sinusoids. An important rule for discrete Fourier series expansion of a particular waveform is that it must be periodic over a finite time interval.

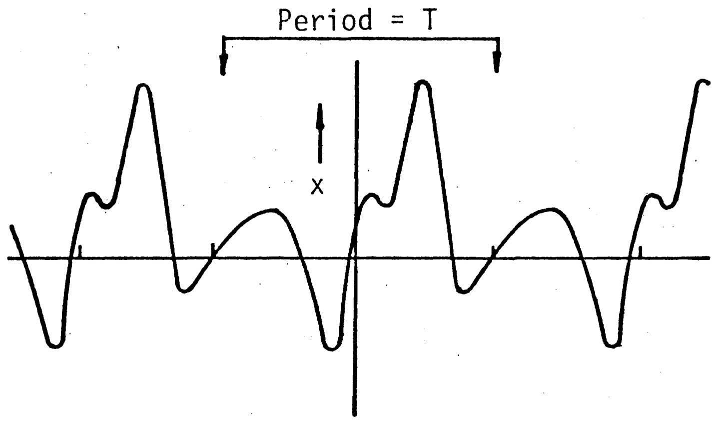
The Fourier series expansion consists of sinusoidal components of frequencies which are integral multiples of the periodic frequency of the original signal. The lowest frequency component is called the fundamental frequency. If the period of the original signal is t, the fundamental is 1/t. The other frequency components are called harmonics, with frequencies of 2/t, 3/t . . . , etc. The Fourier series also gives the amplitude or proportions of the components which are to be summed. A phase angle is also included for each component in the series. Figure 9 shows how the first six terms of the Fourier series sum produce a sawtooth wave. If the series approaches an infinite number of terms, their sum will approach the exact form of the original sawtooth wave.

Fourier series expansion allows a complete representation of a given signal in the frequency domain, just as a time history of the signal is a complete representation in the time domain. Figure 10 illustrates the

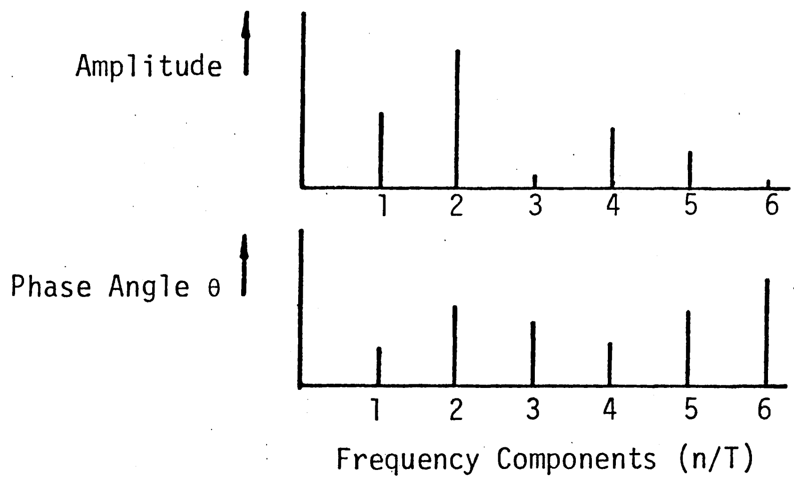


Source: C. C. Goodyear, <u>Signals and Information</u> (1971).

Figure 9. Fourier Series Approximation of Sawtooth Wave



(a) Time Domain



(b) Frequency Domain

Source: C. C. Goodyear, Signals and Information (1971).

Figure 10. Periodic Waveform Represented in Time and Frequency Domains

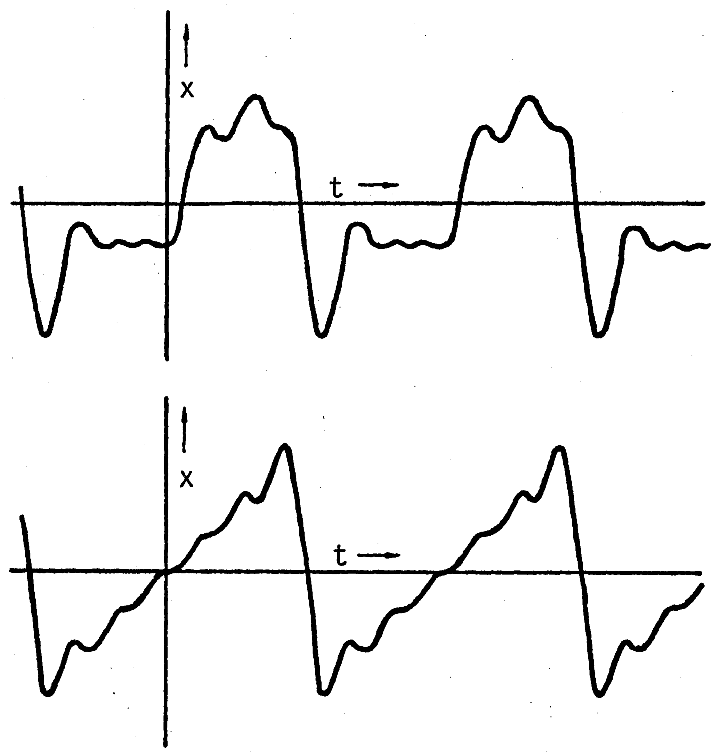
two forms of representation. The amplitudes of the Fourier components are spaced at the harmonic frequency intervals, making up a frequency spectrum with the corresponding phase angles also shown for each component.

An instrument, such as an oscilloscope, will show the signal of Figure 10(a) in the time domain, and a wave analyzer will give a representation of the signal in the frequency domain, as in Figure 10(b). Unfortunately, the wave analyzer pays no attention to phase angles, but this is sometimes of little consequence. The ear is also insensitive to phase and cannot, as the wave analyzer cannot, distinguish between two signals (such as in Figure 11) which have the same harmonic amplitudes but different phases.

Filtering to Achieve Spectral Analysis

Display of the Fourier series spectrum of an unknown periodic signal is usually accomplished through the use of some configuration of electronic band-pass filtering, which includes most analog wave analyzers. Several types of filter systems exist: (1) swept filters where the center frequency of a single filter is varied slowly over the frequency range of interest; (2) stepped filters where the center frequency is stepped through a series of discrete values over the desired range; (3) parallel filters consisting of a number of filters with different center frequencies covering the desired range, allowing simultaneous filtering of the signal; and (4) digital filters using the FFT (3).

A variation of the swept filter analysis system is the heterodyne method. An interesting property of the combining of two time-varying signals is that the resulting signal contains components at the



Source: C. C. Goodyear, Signals and Information (1971).

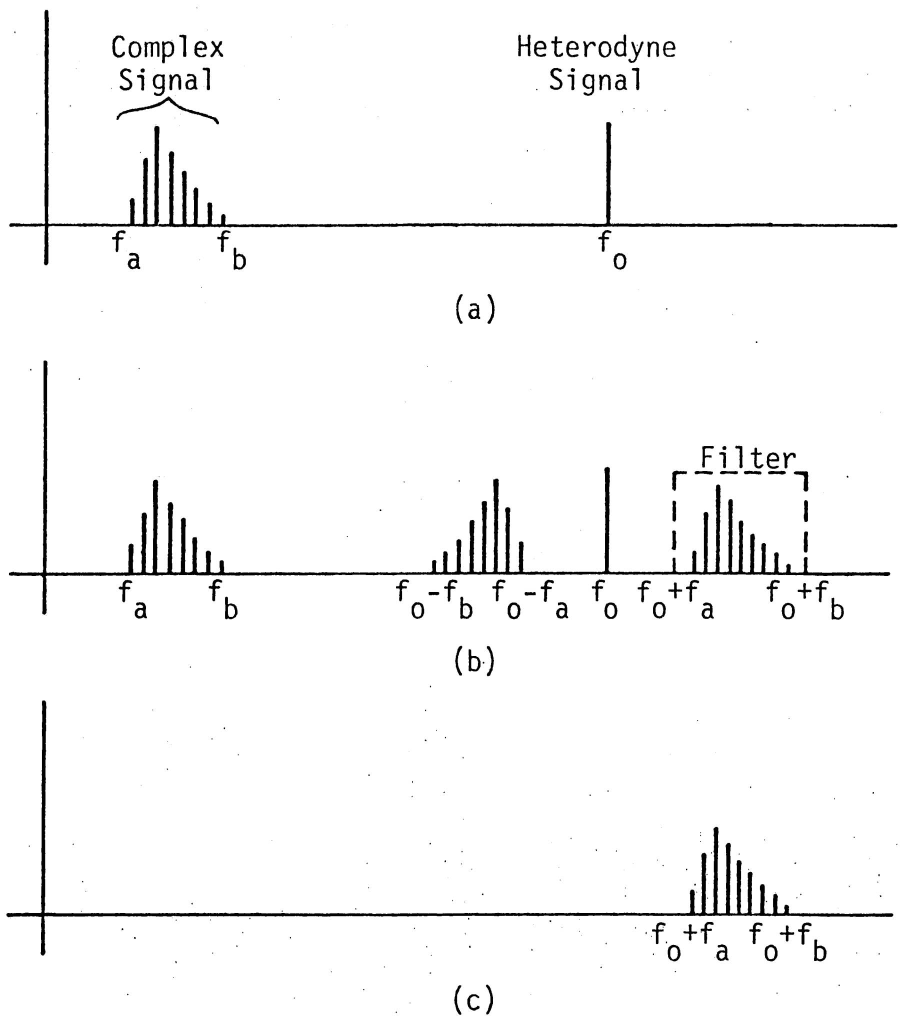
Figure 11. Two Signals With Same Harmonic Amplitudes But Different Phase Angles

frequencies of the original separate signals, and also at frequencies which are the sums and differences of the two signals. For example, if two sinusoids of frequencies f_1 and f_2 are electronically mixed, the spectrum of the resultant signal will contain components at f_1, f_2, $f_1 + f_2$, and $f_1 - f_2$. Consider the signals shown in the frequency domain representation in Figure 12(a). One is a complex signal containing frequency components in the range from f_a to f_b. If this signal is mixed with a sinusoid (heterodyne signal) of frequency f_o (f_o much greater than f_a or f_b), then a spectrum results, as shown in Figure 12(b). An appropriate band-pass filter gives a final spectrum, as shown in Figure 12(c). By varying the frequency, f_o, the range of frequencies seen at the output of the filter will change, in effect making the fixed filter variable and useful in swept filter analysis. This process is known as frequency conversion. The filter sweeping effect, by varying the heterodyne oscillator frequency, is easier and the electronic circuitry much less complex than for directly varying a band-pass filter center frequency, without changing its shape and bandwidth.

A critical factor in the use of the swept filter is the sweeping rate. The rate must be slow enough to permit all the components of the signal, at a particular frequency, to be presented to the filter, before the center frequency is changed. A general rule is to use an integration, or averaging time, as long as the analysis record to give a minimal error for the entire frequency range. In this case the sweep rate would be:

$$\text{Rate} \leq \frac{B}{T} \text{ Hz/second} \qquad (3.1)$$

where



Source: After Goodyear, Signals and Information (1971)

Figure 12. Heterodyning of Two Signals

B = the bandwidth of the filter; and

T = the analysis record length.

With RC averaging, the rate is dependent on the RC time constant, K. For an error less than 5%, a time constant of K = T is acceptable. The rate, assuming four time constants for steady state operation, would be:

$$\text{Rate} \leq \frac{B}{4K} = \frac{B}{4T}\ \text{Hz/second} \tag{3.2}$$

For an idea of the total analysis time for a single record, assume a filter resolution band-width of one Hz. If the record length is one second (approximately one cardiac cycle), the sweep rate will be:

$$\text{Rate} = \frac{1\ \text{Hz}}{1\ \text{second}} = 1\ \text{Hz/second}$$

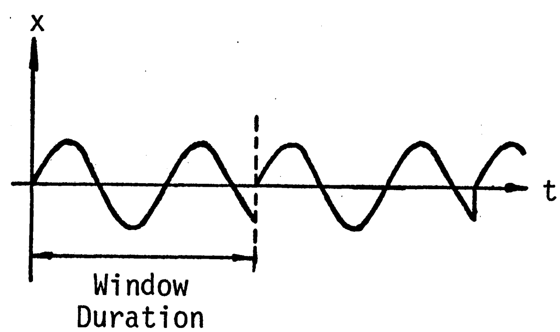
If the frequency range of analysis is 0 to 1000 Hz, the total time for analysis will be:

$$\text{Time} = \frac{1000\ \text{Hz}}{1\ \text{Hz/second}} = 1000\ \text{seconds} \simeq 17\ \text{minutes}$$

For RC averaging the total time will be approximately 67 minutes.

End Point Discontinuities

Discrete spectral analysis requires a periodic signal. In most analysis techniques a data window, shown in Figure 13, is continuously repeated to produce a periodic signal. Analyzing a finite data window sometimes introduces what is known as "leakage" (18). For example, as long as the window duration, or record length, is an integer multiple of the period of the sinusoid of the figure, the analysis will give the proper result. The component frequencies in the spectrum will be integer multiples of the window period. However, if the window does not fit the basic period of the input signal exactly, as is usually the case, the effect is that shown in Figure 13. The analyzed signal is that portion in the data window continuously repeated, instead of the original, pure sinusoid. The analysis procedure introduces an extensive set of components in the spectrum because of the discontinuity, or truncation, of the signal by the window.



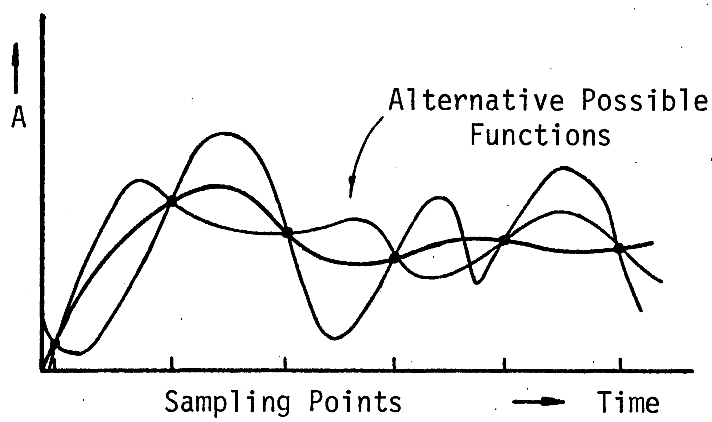
Source: A. P. Peterson et al., Handbook of Noise Measurement (1974).

Figure 13. Repeated Data Window

The end point discontinuity problem surfaces when tape loops are employed to produce the periodic PCG signals for anlaysis. Not only is there a discontinuity in the signal on the tape, but the splice to form the loop on the tape itself introduces some unknown effects, because no spliced connection can be perfect. Little account is usually taken of these conditions, which can affect the results of an analysis greatly.

Digital Sampling

Recording analog signals digitally, for subsequent analysis, is a very useful technique, but particular properties of operation exist. The frequency of collection of data values of the analog signal, or sampling rate, is critical for proper signal reconstruction. Too low a sampling rate results in "aliasing" of the signal, illustrated in Figure 14 (2). The collected data values could have come from any of the signals shown, or an infinite number of other signals. Sampling theory requires that the sampling rate be at least twice the value of the highest frequency signal to be studied. This frequency is known as the Nyquist frequency. In practice it is generally a good idea, when faithful reproduction of the original signal is desired, to use a rate at least four times the highest frequency of interest.



Source: K. G. Beauchamp, Signal Processing Using Analog and Digital Techniques (1973)

Figure 14. Aliasing of a Digitally Sampled Signal

On subsequent analog output of a digitally-stored signal, the waveform will appear as a series of discrete values, instead of a continuous function like the original signal. The higher the sampling rate, the less obvious the discrete points will be. A smoothing procedure can be employed to lend a more continuous nature to the appearance of the output. This is usually accomplished by a low-pass filter system on the output, which improves the reconstruction smoothness substantially.

Hybrid Spectral Analysis

Solely analog analysis, if even applicable at low frequencies, requires long analysis time for accurate results. Solely digital analysis requires expensive equipment and computation time. For reasons of economy in hardware and analysis time, the reduction in complexity associated with filter design for low frequencies, and the need for operation in real time, a hybrid method of analysis should be considered over strictly analog or digital processes.

A hybrid method in some use is known as time compression. It involves an upwards frequency translation of the signal being analyzed, speeding up analysis time, and achieving results similar to an increase in tape replay speed in an analog system (2).

The analog signal, sampled at a rate consistent with the highest frequency to be analyzed, is converted to digital form and stored in memory. The samples can then be read out, but at a higher rate than originally read in, resulting in a time compression of the original signal. The output rate depends on the memory cycle time of the digital storage, which is on the order of a few microseconds per sample. Repeated read-out in this manner has the same effect as accelerated replay of magnetic tape loops, without the difficulties associated with their production. The digital signal storage also permits a much higher replay rate than the conventional analog system's capability.

The time compression of the signal, in effect, multiplies the frequency components of the original signal. This results in a new frequency spectrum, identical in form to the original spectrum, but with wider separation of the frequency components. The spreading of the components permits an analog analyzer to be a more effective means of analysis, because the resolution requirements are not as difficult. The corresponding analysis time is reduced by a factor of:

$$\frac{T}{RN}$$

where

T = the original length of the analyzed signal;

R = the output rate of the digital memory; and

N = the number of samples making up the stored signal (2).

CHAPTER IV

INSTRUMENTATION

System Configuration

The basic system configuration is shown in Figure 15. Heart sound signals are obtained from a Bruel and Kjaer one-inch condenser microphone and attaching aluminum bell. Simultaneous EKG signals, obtained from a Tektronix patient monitor, and the heart sounds can be recorded on the Honeywell Model 7600 FM tape recorder, for future reference. A Krohn-Hite band-pass filter, with slopes of 80 db/decade, selects high and low ends of the frequency range of interest. The Biomation Model 1015 digital waveform recorder digitizes and stores the information of interest. The signal stored is observable via the CRT oscilloscope or is available at an X-Y plot output. The Tektronix Model 7L5 spectrum analyzer, with CRT viewing screen, receives the high-speed analog output from the DWR for analysis. An isolation amplifier is provided to prevent the low input impedance of the spectrum analyzer from loading the output of the DWR. An X-Y plot of the spectrum analyzer trace is available.

Microphones and Impedance Matching

An ideal vibration transducer should not significantly alter the vibration being measured. However, most existing phonocardiographic transducers do not provide a faithful reproduction of the heart sound/

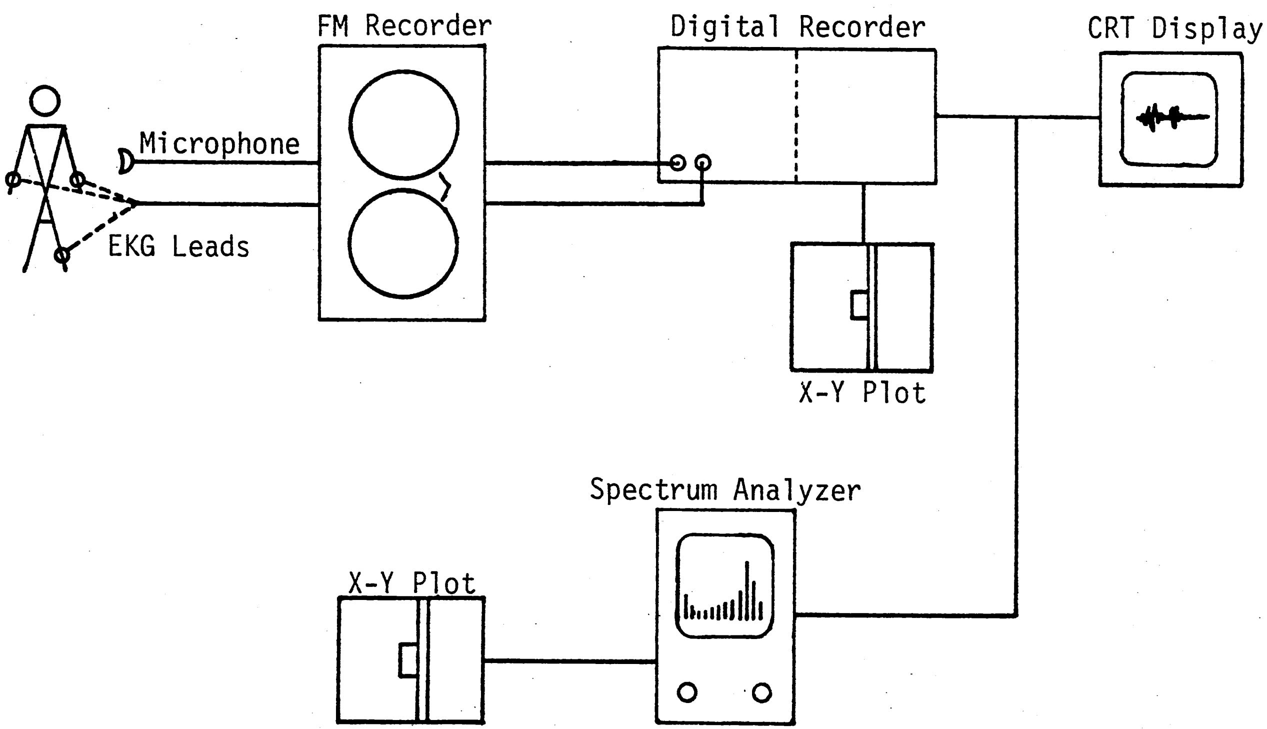


Figure 15. Simplified Schematic of System Configuration

vibrations emanating from the chest surface (8). The difficulty arises from improper impedance matching of the vibration transducer to the chest surface. Griffen (9) states that neither application force, nor tissue condition at the application site, should materially affect the transducer response.

Tatge (21) found that the mechanical loading of the vibratory medium must be minimized before meaningful signals can be recorded. Noncontact optical transducers, or a contact transducer with a large area-to-mass ratio, can be used to reduce loading. Another alternative is to air-couple a microphone to the precordial tissue via an annulus, as shown in Figure 16.

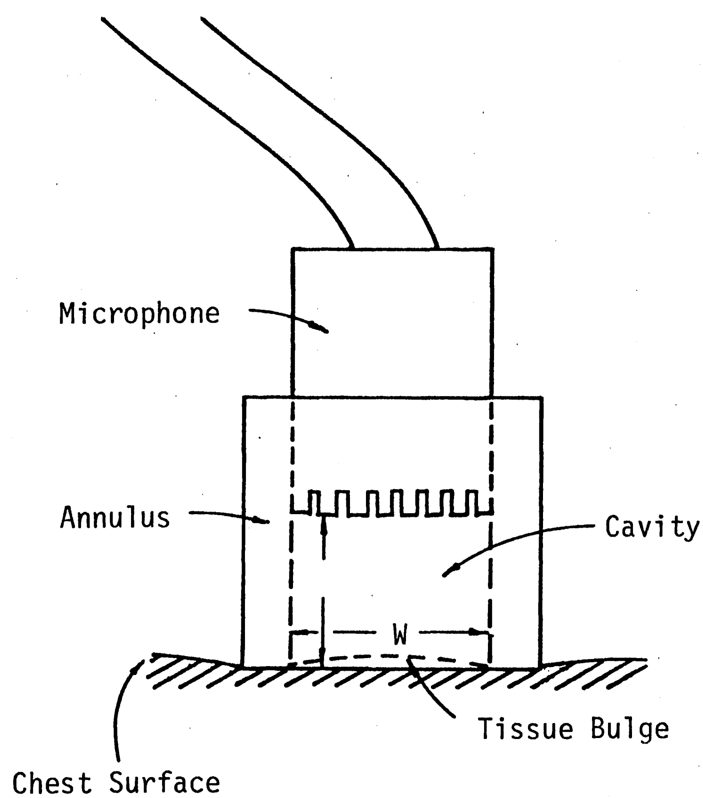


Figure 16. Microphone and Air-Coupling Annulus

Specific requirements for an effective air-coupled sensor have been established by Griffen (9). Optimization of the design requires some trade-offs. The air-coupling cavity must be small enough to prevent resonances and provide high sensitivity, yet large enough to permit the encircled tissue to vibrate freely. Results from Griffen's (9) research indicated that the maximum cavity dimensions must be small relative to 16 centimeters in depth and 19 centimeters in diameter, to reduce any effects of cavity resonance. If the annulus diameter is greater than 2 centimeters, the contact pressure can be varied over a wide range without significantly affecting the output. A check of one air-coupled sensor indicated less than $\pm$1.2 db variance of the output voltage over a frequency range of 10 to 1000 Hz when the application force was varied from 0 to 500 grams.

The aluminum bell-type annulus used in this research was developed previously by Taylor (22) using the guidelines set forth by Griffen (9). The cavity is 2.4 centimeters deep and 2.8 centimeters in diameter. The air-coupling of the bell does not provide a perfect impedance match for all individuals, but is a distinct improvement over existing phonocardiographic transducers.

Good reproduction of the heart sound/vibrations is also affected by the acoustic sensor itself. The microphone used by Van Vollenhoven (23) (response shown in Figure 4) is an example of the poor sensor performance that can be expected from existing phonocardiographs. Microphones with very satisfactory responses are available, however. For this research a specially-vented Bruel and Kjaer Model 4131 one-inch condenser microphone was matched with the aluminum bell annulus. The frequency response of the microphone is shown in Figure 17. The microphone and bell were attached to a holding fixture secured to the chest by an elastic strap.

FM Recorder

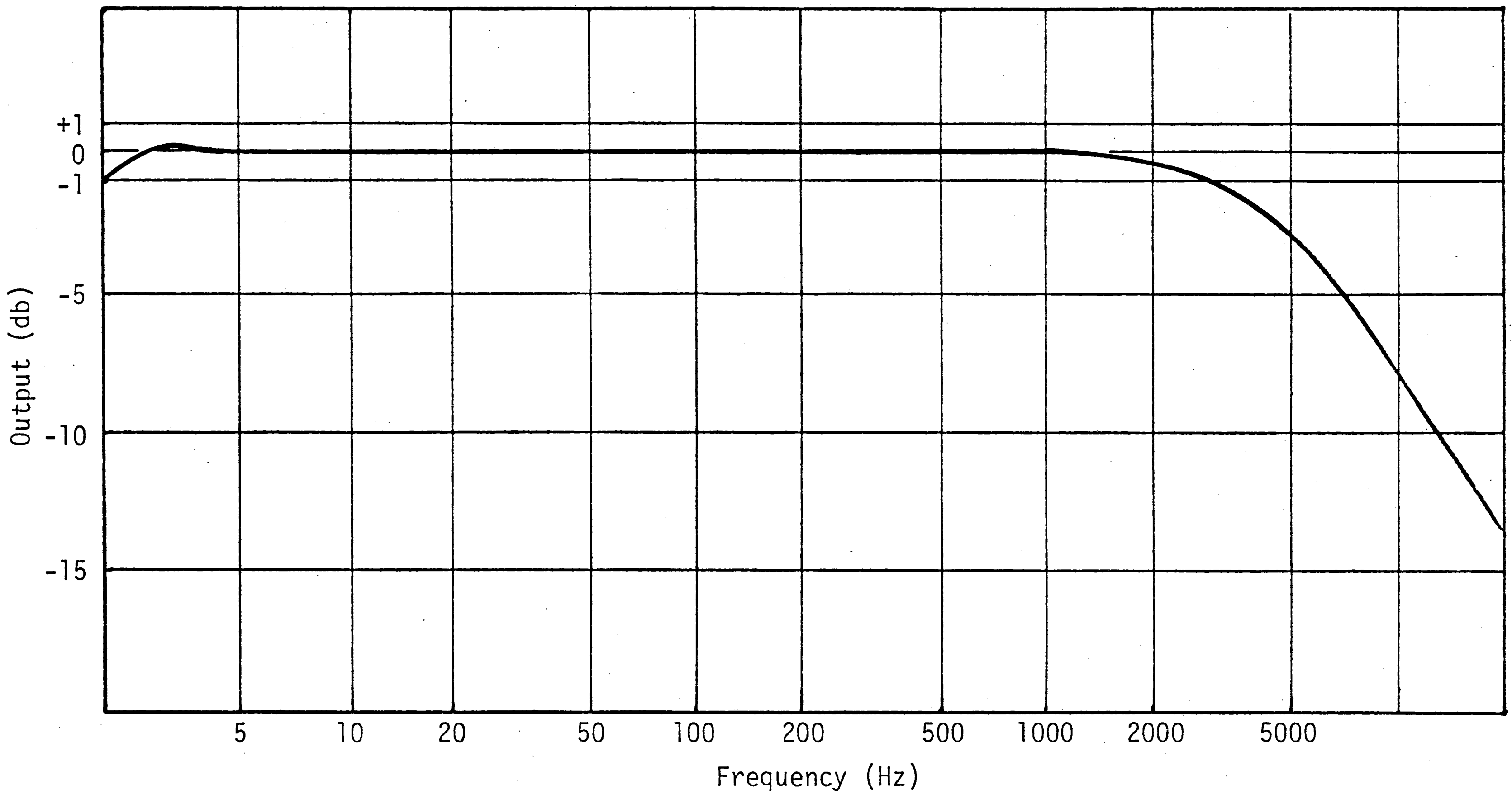
The Honeywell Model 7600 FM recorder was used to collect a limited library of heart sound subjects. Frequency response of the recorder is 0 to 5000 Hz in the double-extended mode of operation at the selected recording speed of 15 inches/second.

Digital Waveform Recorder

The heart of the system is the Biomation Model 1015 digital waveform recorder. The DWR is a multi-channel analog signal recorder utilizing an integrated circuit digital memory. It has several unique capabilities that offer operational advantages over conventional electromechanical recorders. Integral input analog-to-digital converters as well as output digital-to-analog converters are incorporated. The storage capacity of the recorder is fixed at 4096 10-bit words (data samples). The 10-bit words permit a maximum signal resolution of one part in 2^{10}, or one in 1024. This represents a resolution of a little better than 0.1% of the input voltage range.

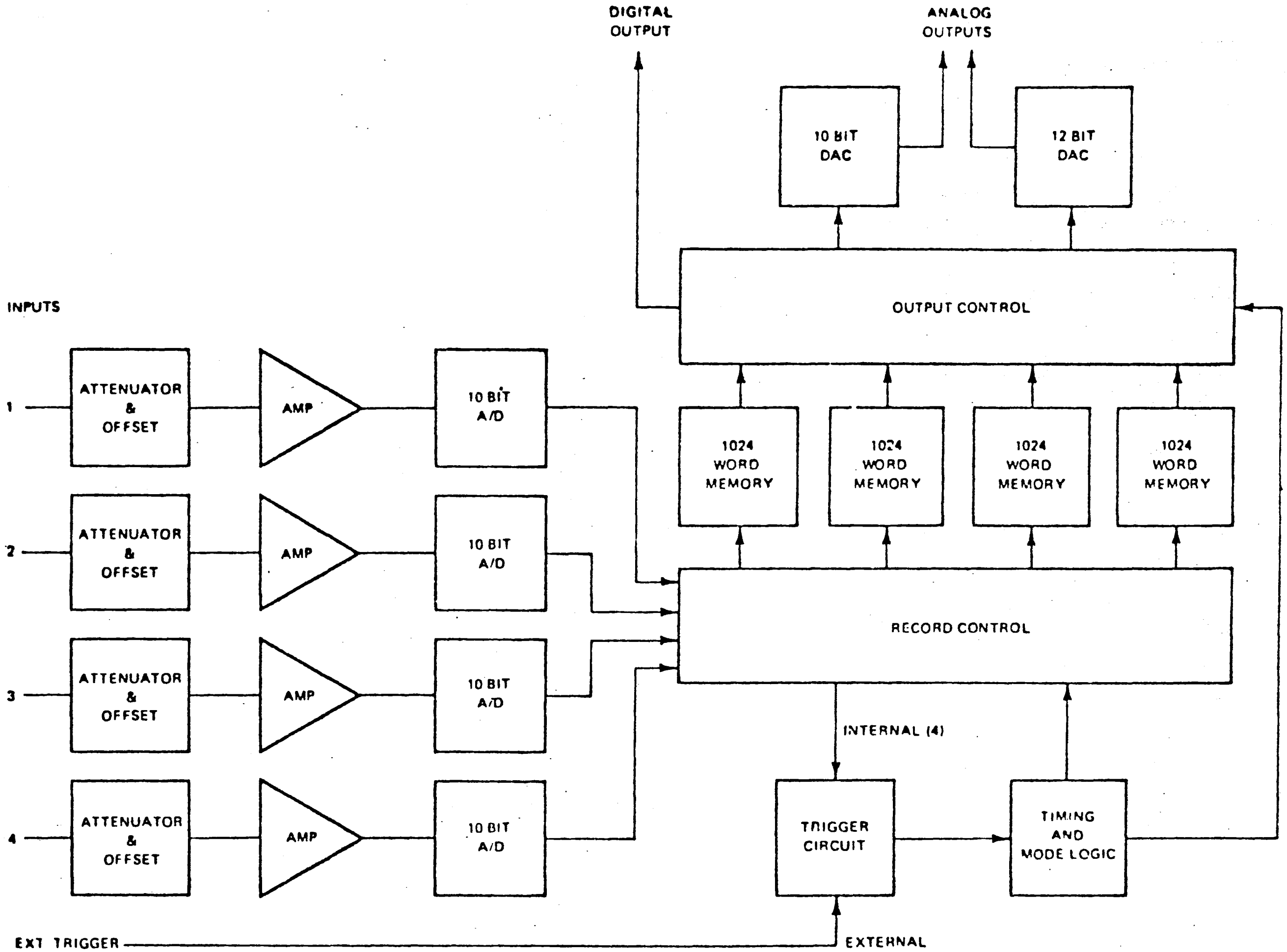
Sampling rate is adjustable through an internal oscillator, or externally via a TTL compatible pulse generator. Maximum sample rate is 10^5 words/second. A simplified block diagram of the DWR is presented in Figure 18.

Triggering is accomplished with a conventional voltage threshold detection circuit. The triggering circuit has an adjustable time delay between the initiating signal and the actual recording process. The sampled data window can be accurately positioned through the delay and



Source: Factory data sheet

Figure 17. Response of Condenser Microphone Used in Research



Source: Biomation, Model 1015 Waveform Recorder Operating and Service Manual (1975).

Figure 18. Block Diagram of Digital Waveform Recorder

a unique pretrigger control. This control allows the DWR to record events that precede a triggering signal. In the pretrigger mode of operation the DWR records information continuously at the selected rate. The earliest samples collected are discarded at the same rate when the memory becomes full. When a trigger signal is received the preset delay value stops the recording process, saving the data received prior to the trigger. The process is illustrated in Figure 19. The delay value determines the portion of the samples, in memory, which occurred after the trigger signal. DWR operation in the delay mode is also shown.

Dual digital-to-analog converters provide two types of analog outputs. In addition to a high rate output of 10^6 words/second, a slow, variable output rate for hard copy reproduction on X-Y plotter or strip chart recorder is available. The output rate is adjustable for proper X-Y plot operation. The variable plot output allows a high-fidelity reproduction of high frequency signals not possible with standard or high speed chart recording.

Spectrum Analyzer

The Tektronix Model 7L5 spectrum analyzer is a portable heterodyne analyzer, described in Chapter III. This particular model spectrum analyzer uses two frequency conversion processes for simplification of circuitry. The minimum filter resolution is 10 Hz. The input signal frequency range is 0 to 5 MHz.

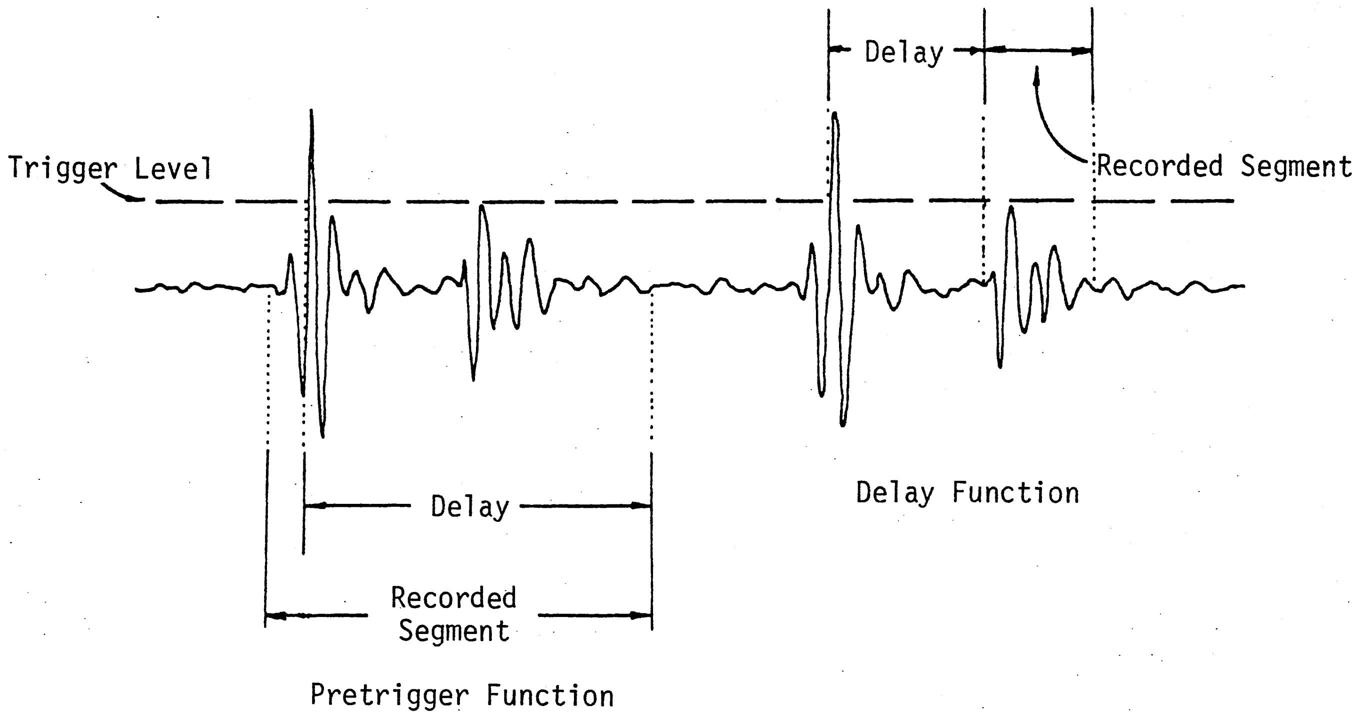


Figure 19. Pretrigger and Delay Recording Control Functions

CHAPTER V

PROCEDURE

To develop this system it was necessary to record heart sounds of several subjects on FM tape. The heart sounds were obtained with the aid of the bell annulus and microphone. The recording location was the precordial area of each subject, just left of the sternum at the fourth intercostal space. Simultaneous EKG signals were obtained from the three standard electrodes (left wrist, right wrist, left ankle). Lead II was recorded with the heart sounds from eight apparently normal individuals, and two individuals with known abnormalities in their heart sounds. Brief descriptions of each subject are contained in the Appendix.

Four minutes of continuous data were taken from each subject, or about 300 feet of FM tape at the recording speed of 15 inches per second. An additional two minutes of data were taken from subject A1 at a different chest location because of the nature of the abnormality displayed.

Subject A1 had been previously diagnosed as having aortic stenosis, with surgery confirming the condition. The sound of the murmur was recorded best more nearly over the aortic valve area, at the first intercostal space left of the sternum. Subject A2 was previously diagnosed as having a mitral systolic murmur, which has probably existed from birth.

The FM recorded heart sounds were replayed at normal speed for recording by the DWR. One cardiac cycle of heart sounds was recorded to produce a continuous, periodic, "ideal" phonocardiogram on output. The finite memory of the DWR limits the amount of signal that can be recorded at any one time. One cycle, approximately one second in duration, allowed the maximum frequency range of approximately 1000 Hz for the 4096 word memory (allowing at least 4 points per vibration cycle).

Larger segments than one cycle of heart sounds are conveniently recorded, and were when only a signal reproduction from the DWR was desired. Recording longer duration signals merely limits the frequency range available for spectral analysis (because of the sampling rate requirements).

Smaller than one cycle segments were also recorded for analysis. First and second heart sounds from each subject were individually isolated, with a corresponding spectral analysis performed on the signals.

In all cases the recorded signal portion was identified by its time relationship with its simultaneous EKG signal. The QRS complex was used as a trigger signal and time reference.

An adjustment of the sampling rate of the DWR was required to record a specific portion of the heart sounds. Since the duration of any particular cardiac cycle varies from one cycle to the next, it was necessary to adjust the delay and sampling rate as the signals were replayed from FM tape, to place the data window in the desired position.

Filtering of the heart sounds was necessary because of the considerable low frequency content of the signals (from the low frequency response capability of the microphone sensor). The low frequency components can cause large end point discontinuities when the heart sound segments are recorded, as illustrated in Figure 20. In an effort to minimize the effect of the discontinuities in the spectra, the signals were high-pass filtered (cut-off frequency 20 Hz, 80 db/decade slope). The filtering made the first and second sounds more definitive and separable from the vibrations occurring in the diastolic time interval, as shown in Figure 21.

The signal stored by the DWR was reproduced and then analyzed by the spectrum analyzer. Because of the fixed sample output rate of 10^6 points per second and the variable sample input rate, the amount of the time compression varied with the input, or recording, sample rate. The frequency multiplication factor (sample rate play-record ratio) available to the spectrum analyzer was variable and, thus, the actual frequency span on the analyzer CRT display. As a result a direct CRT screen comparison of different signals' spectra was difficult. This problem was overcome on the hard copy reproduction of the spectra by appropriately attenuating the horizontal signal output from the analyzer. The resulting spectra were all normalized to a one inch = 100 Hz scale with 500 Hz as the maximum frequency.

The actual resolution band-width of the spectra obtained varied with the sample rate play-record ratio also. This property could not be overcome because the instrument resolution setting was not continuously adjustable.

An effort was made to illustrate the effect of "leakage" on the analysis of a simple signal. Figure 22(a) shows a 50 Hz sinusoid, recorded by the DWR at a rate which corresponded to a time duration of very nearly five cycles of the sinusoid. The fundamental frequency of the analysis is the inverse of the window duration, or 10 Hz. The

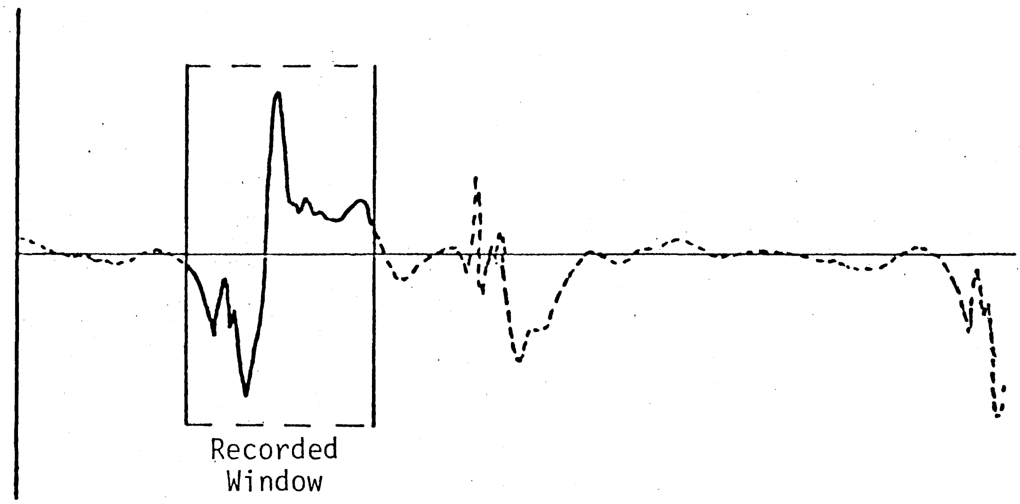


Figure 20. End Point Discontinuity Effect From Low Frequency Signal Components

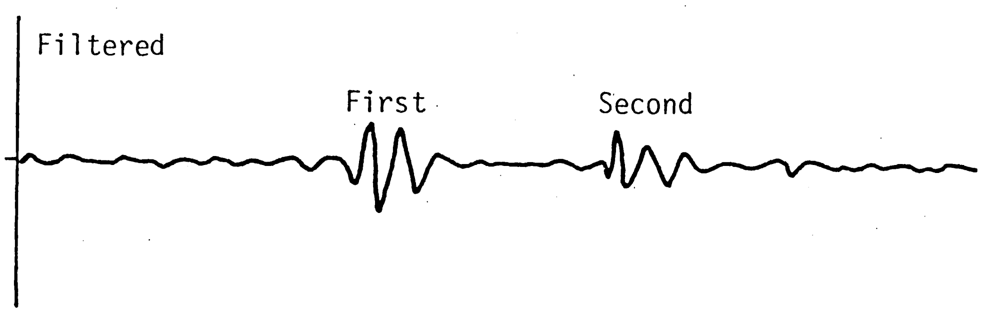


Figure 21. Filtering Effect on Separation of First and Second Heart Sounds

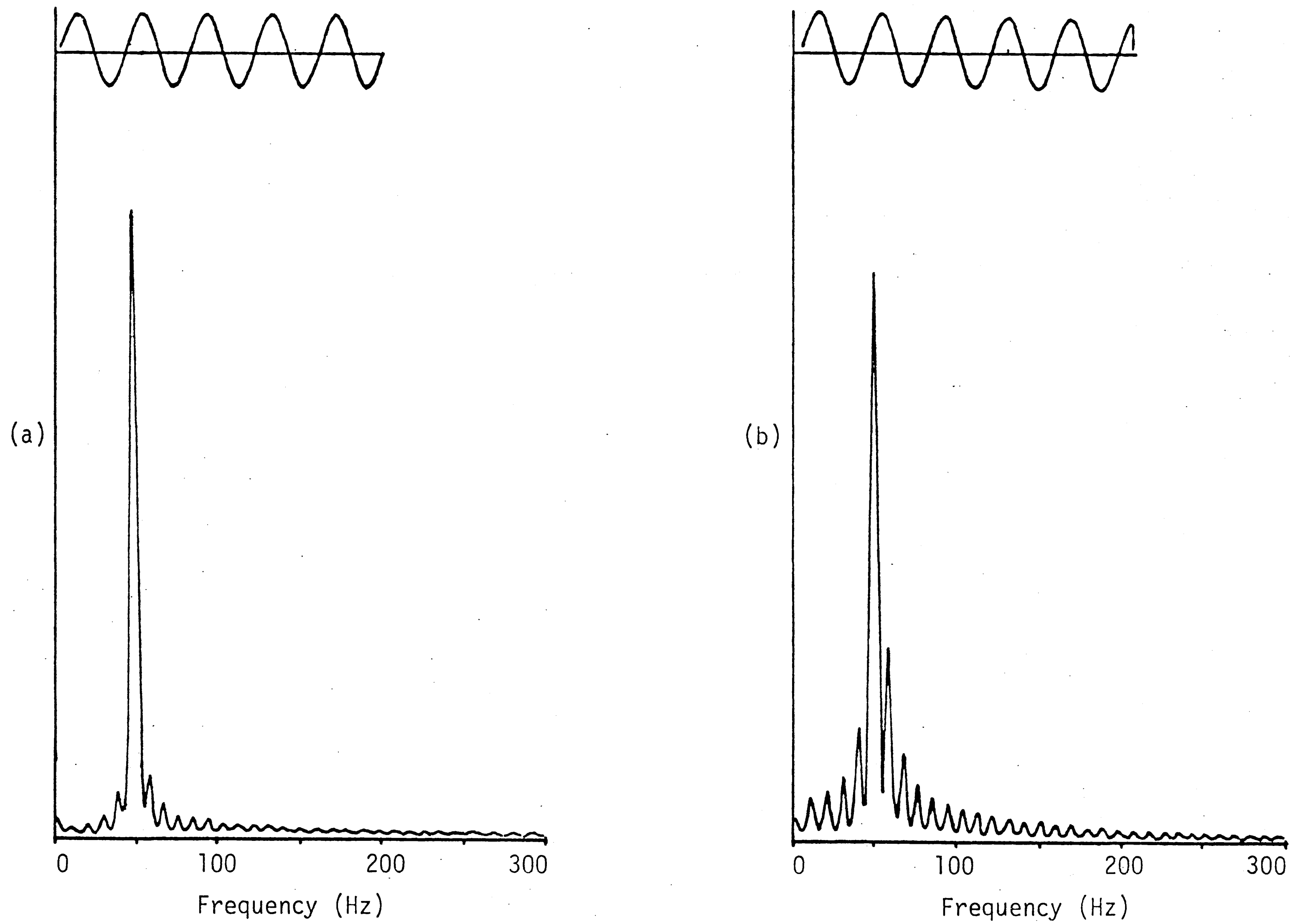


Figure 22. Effect of Leakage on Frequency Spectrum of Sinusoid

spectrum would be expected, then, to consist of a very strong component at the fifth harmonic, with the others considerably depressed, which is exactly the case as shown.

Without changing the sample rate, the frequency of the sinusoid was increased slightly to produce a marked discontinuity at the end points (Figure 22(b)). The fundamental analysis frequency was not changed. The harmonics remain at 10 Hz intervals as shown; however, the fifth harmonic is slightly suppressed, with the others correspondingly increased. The original signal was a pure sinusoid, but the analysis procedure introduces many other components into the spectrum.

CHAPTER VI

RESULTS

Signal Reproductions

The results of this research are basically concerned with the operational procedures of the system and reproduction of the heart sound recordings and EKG impulses. The capability to produce useful frequency spectra of particular heart sounds is also a primary interest.

The initial portion of the results consists of the time plots of the heart sounds and EKG impulses from each subject. Figure 23 shows single cycle phonocardiograms from all the subjects for comparison. First and second sounds are indicated on the heart sound of subject N1. It is convenient to refer to particular heart sound signals and spectra of the subjects by letter-number designations, as shown on Figure 23, and explained in the Appendix.

The plots do not represent equivalent time durations as the heart rates for each individual were different. A wide variation in the heart rates existed, from a high of 84 (N5) to a low of 42 (A2). Calculations from the signals of Figure 22 indicate, however, only a 7% variation between subjects in the time interval between the onset of the first sound and the onset of the second sound. The amplitude scale of the signals is the same for all except N3 and A1 which are shown on a scale twice as sensitive as the others.

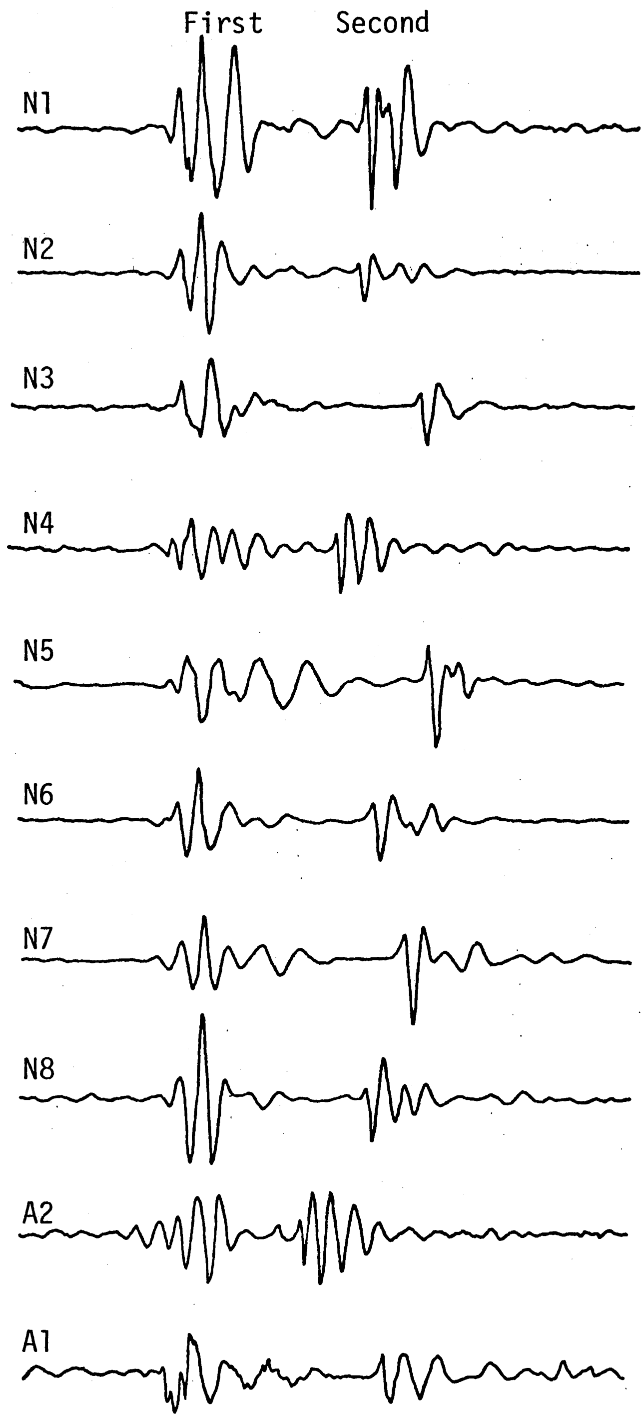


Figure 23. Single Cycle PCG of
N1 Through A2

Some corresponding EKG signals are presented for comparison in Figure 24. Some pecularities are exhibited. N4 lacks Q and S components. N8 shows inversion of the T wave.

Some of the normal phonocardiograms (N1 - N8) of Figure 23 show similar vibration forms. N6 and N7 look very similar. On the other hand, N4 looks unlike any of the other normal sounds. N5 exhibits a decaying vibration of large amplitude and low frequency (~15 Hz) in the systolic phase, between the end of the first sound and the beginning of the second sound. In the longer duration segment of Figure 25, N7 shows some vibrations occurring after the second sound that may be a third heart sound.

Some of the subjects exhibited a very low frequency, almost dc, shift in their EKG impulses. This is illustrated to some extent in Figure 26. Some signals had so much shift it was necessary to record under ac coupling to get a consistent trigger signal.

Signal A1 of Figure 23 displays to a degree the vibrations of the aortic stenosis murmur, but the normal sounds are still quite prominent. Figure 27 illustrates the EKG impulse and heart sounds of A1 as recorded near the aortic valve area. The filter low frequency cutoff was set at 50 Hz to show the vibrations of the murmur more clearly. Partial first and second sounds are still evident.

Examination of signal A2 of Figure 23 alone may not reveal an abnormality. With the aid of the simultaneous EKG in Figure 28, however, it is possible to identify the mitral murmur as the vibrations which precede, in time, the spike of the QRS complex. The onset of the vibrations of the first heart sound does not occur before the QRS complex in a normal phonocardiogram.

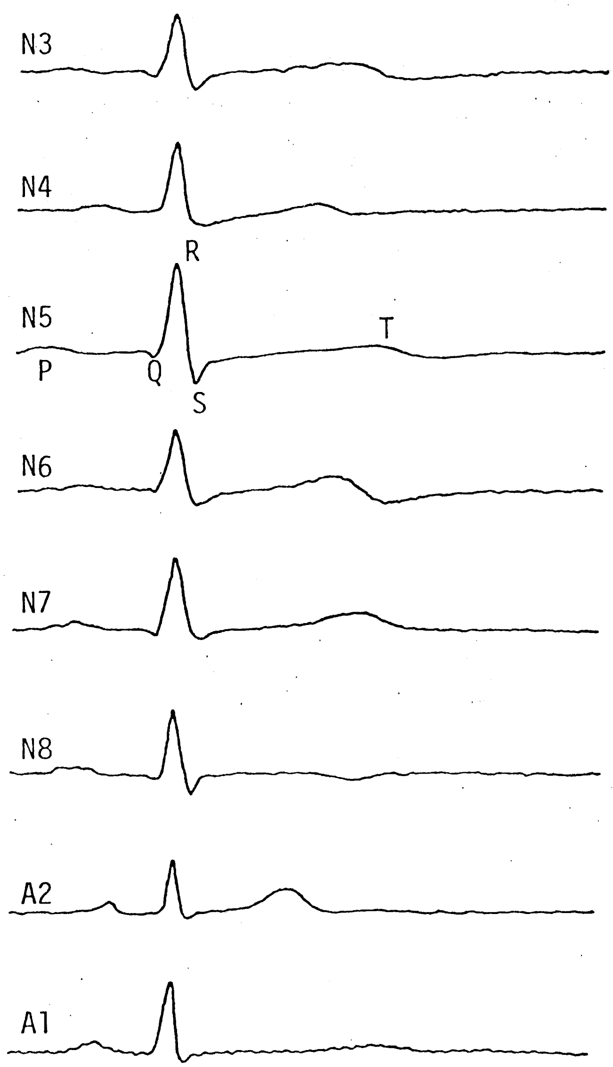


Figure 24. Single Cycle EKG of N3 Through A2

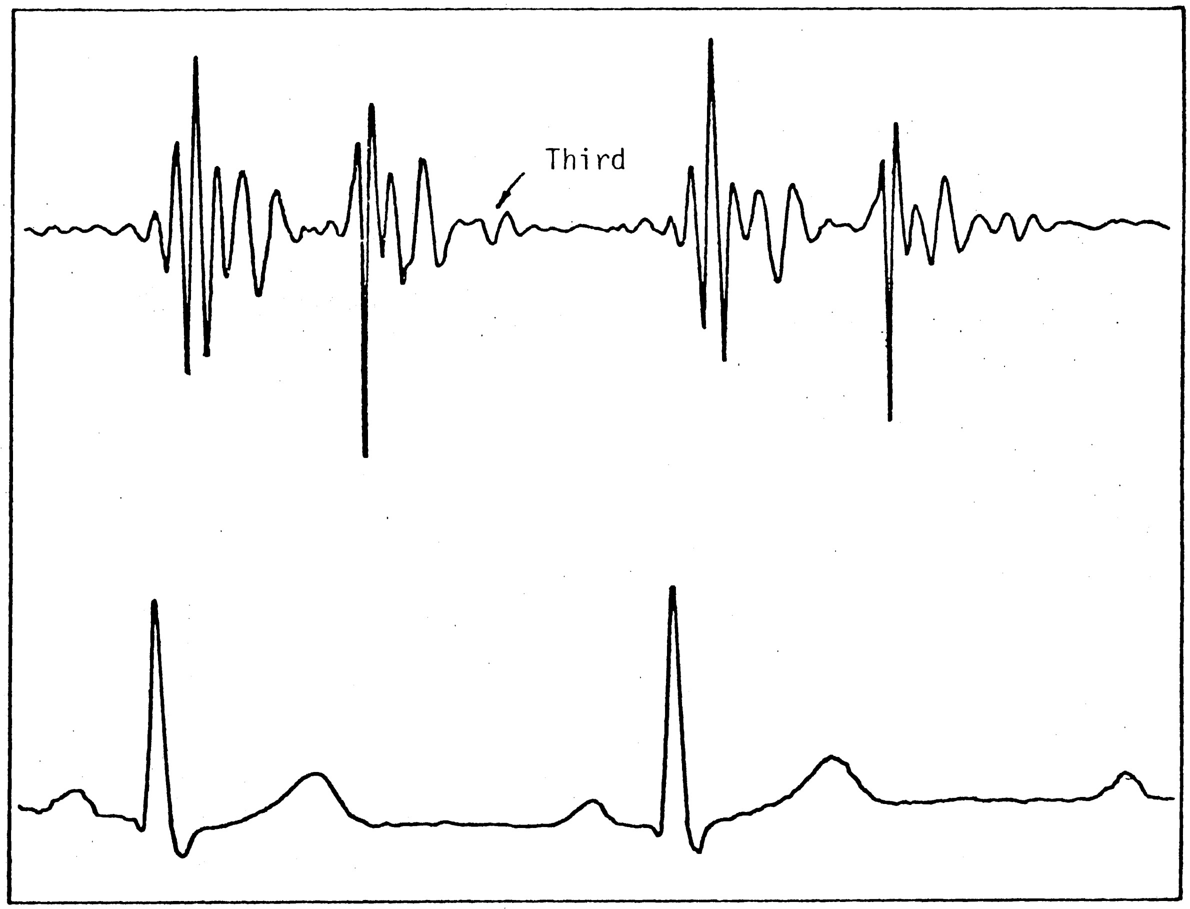


Figure 25. PCG and EKG of N7 Showing Possible Third Heart Sound

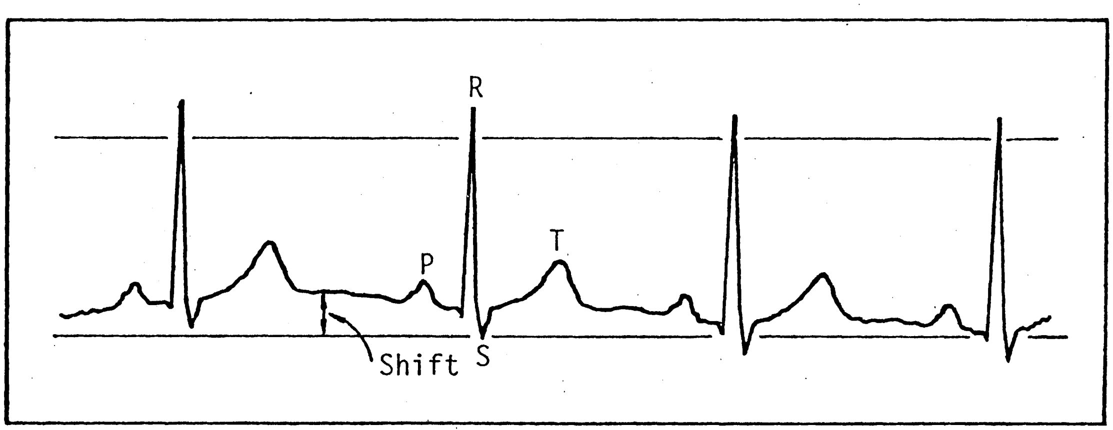


Figure 26. Low Frequency Transient Shift in EKG

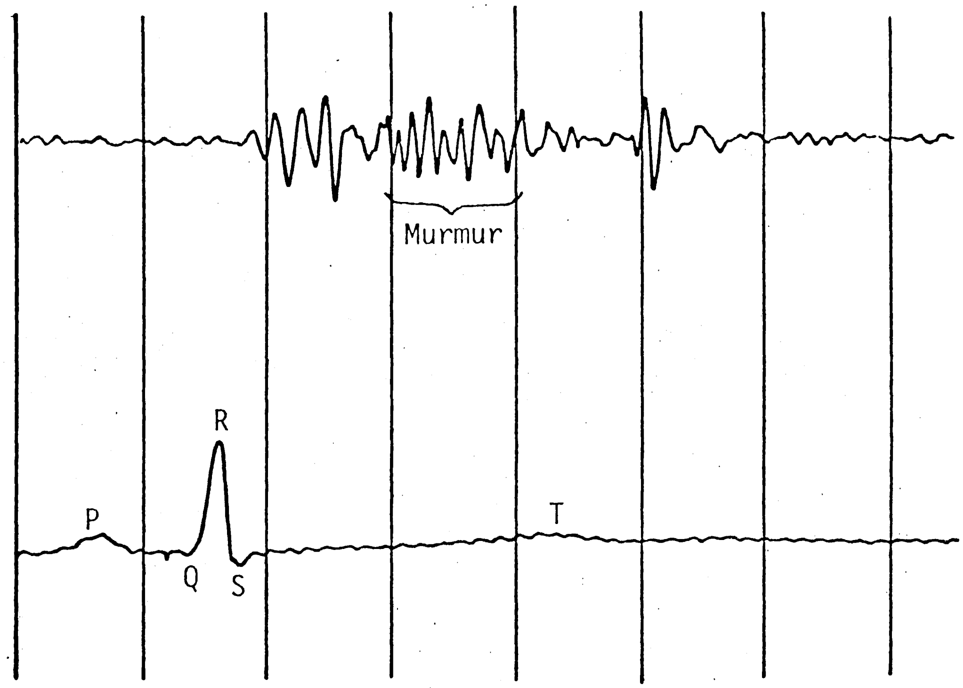


Figure 27. PCG and EKG of A1 Showing Aortic Stenosis Murmur

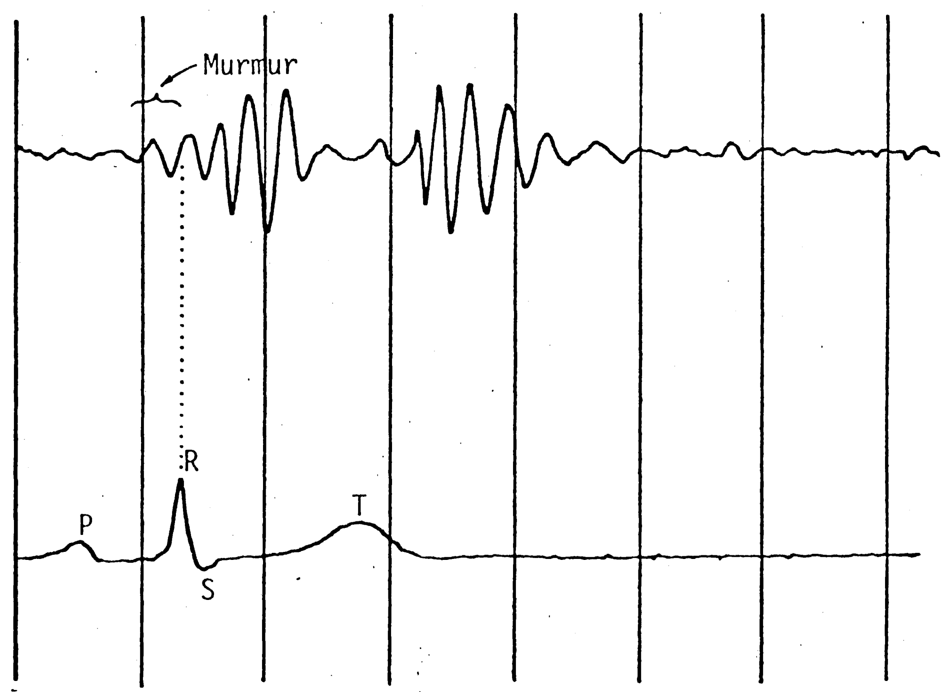


Figure 28. PCG and EKG of A2 Showing Mitral Systolic Murmur

Frequency Spectra

The spectra obtained covered the frequency range from zero to 500 Hz. However, since none of the subjects exhibited any significant components above 300 Hz, only that portion of the spectra below 300 Hz is shown, in most cases, for conservation of space.

Maximum effective resolutions possible were on the order of 1.2 Hz. This calculation is based on a sample record rate of 4000 points per second and an instrument resolution setting of 300 Hz.

$$\text{Play - Record ratio} = \frac{10^6 \text{ points/second}}{4000 \text{ points/second}} = 250$$

$$= \text{frequency multiplication factor}$$

All frequency components, as well as the band-width resolution indicated by the analyzer, had to be divided by the multiplication factor to give the actual values.

$$\frac{300 \text{ Hz}}{250} = 1.2 \text{ Hz effective resolution}$$

Resolutions for each signal and spectrum are all slightly different but of the same order for the complete heart sound cycles.

Figure 29 is an example of the resolution capability of the system. A one-cycle phonocardiogram from subject N1 is shown along with its spectra as determined by the analyzer, with three different resolution settings. Effective resolutions are 4.0 Hz, 1.2 Hz, and 12.0 Hz for Figure 29(a), (b), and (c), respectively. The resolution detail of Figure 29(a) reproduced best and further spectrum resolutions are on the same order, unless otherwise noted.

The spectra of several normal subjects are notable. Subject N1 (Figure 29(a)) exhibited a relatively large gap between two prominent

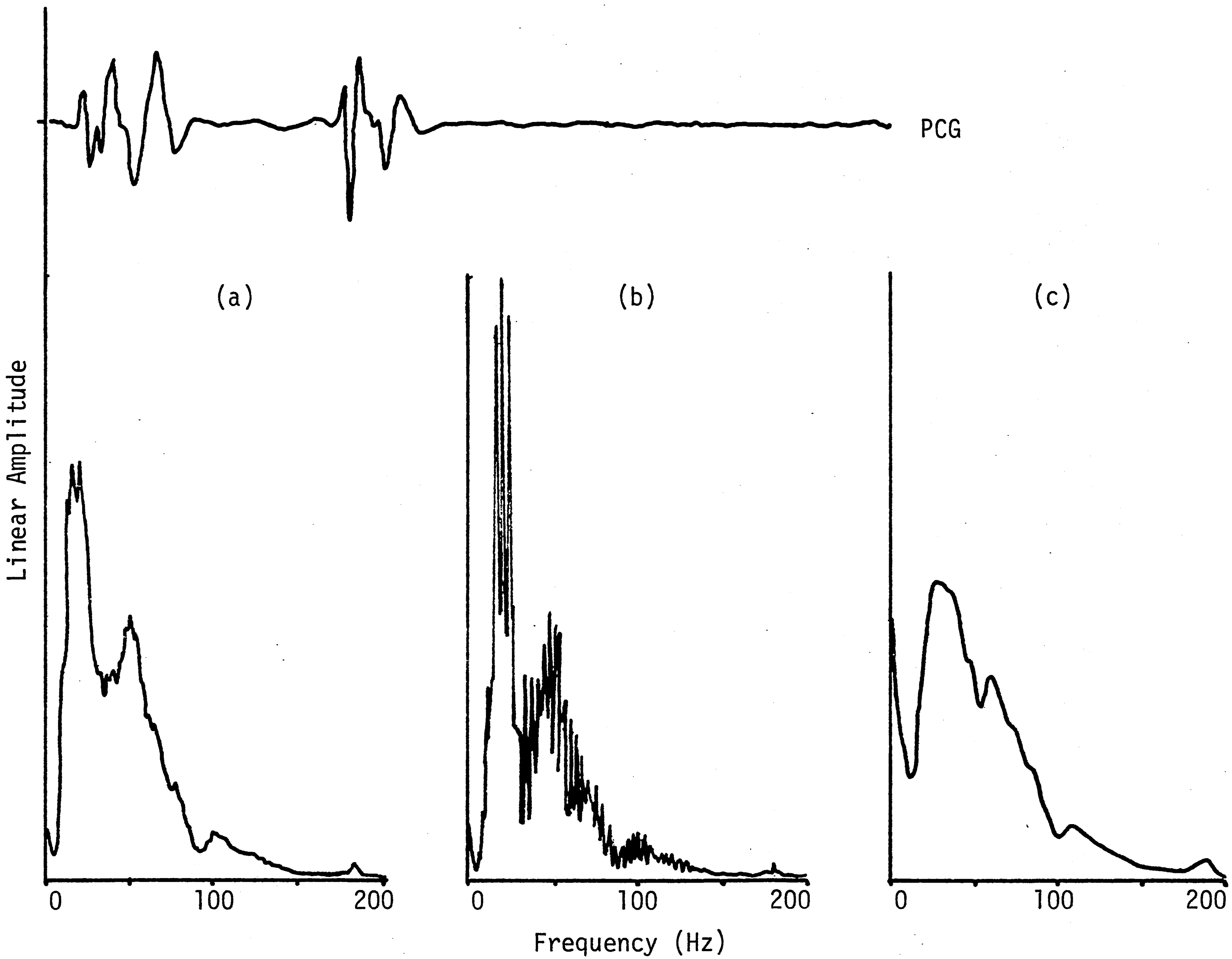


Figure 29. Resolution Capability of Spectrum Analyzer

peaks as compared with N2 (Figure 30, resolution 1.1 Hz) which has a much narrower spectrum. Other spectra had a smaller gap between peaks as shown in Figure 31 (N7 and N8).

Consistency in the spectra of different individual heart sounds from the same subject is demonstrated in Figure 32. Three distinct heart sound cycles have their spectra overlaid for comparison. The spectra are obviously different but evince strong similarities. This suggests that a range may be defined for a reasonable probability of the occurrence of any single spectrum.

The spectra from subject A1 were quite different, depending on the particular location from which the sounds came. Figure 33(a) shows the signal and its spectrum from the fourth intercostal location. Figure 33(b) shows the spectrum and signal from the first intercostal location. The latter spectrum displays many more separate peaks in the higher frequency range. A further signal and spectrum from the first intercostal location is shown in Figure 34. The signal was high-pass filtered (cut-off frequency 50 Hz) and the vertical sensitivity increased for clarity. Many separate peaks are visible over a broad range, to near 200 Hz. Subject A2, in contrast, exhibited one of the narrowest spectra of any of the subjects, as shown in Figure 35.

A measure of the repeatability of the procedure was appropriate. Though not originally intended for the purpose, a signal and spectrum from subject A1 (Figure 36) taken from FM tape in April,1977 was compared with a spectrum and signal taken in September, 1977. The individual signals are not identical, so neither should the spectra be. However, the spectra display very similar frequency peaks and relative component

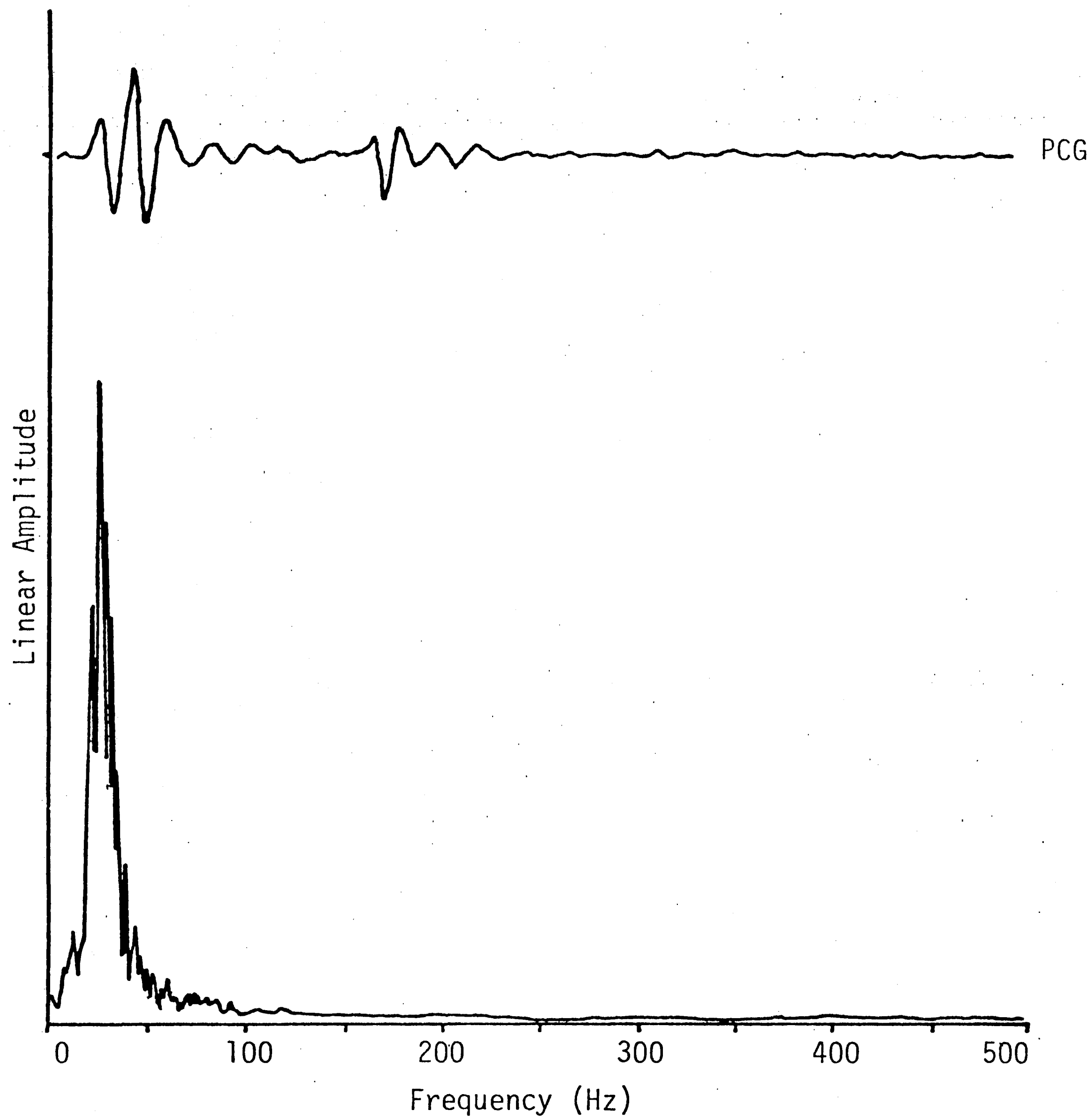


Figure 30. PCG and Spectrum of N2

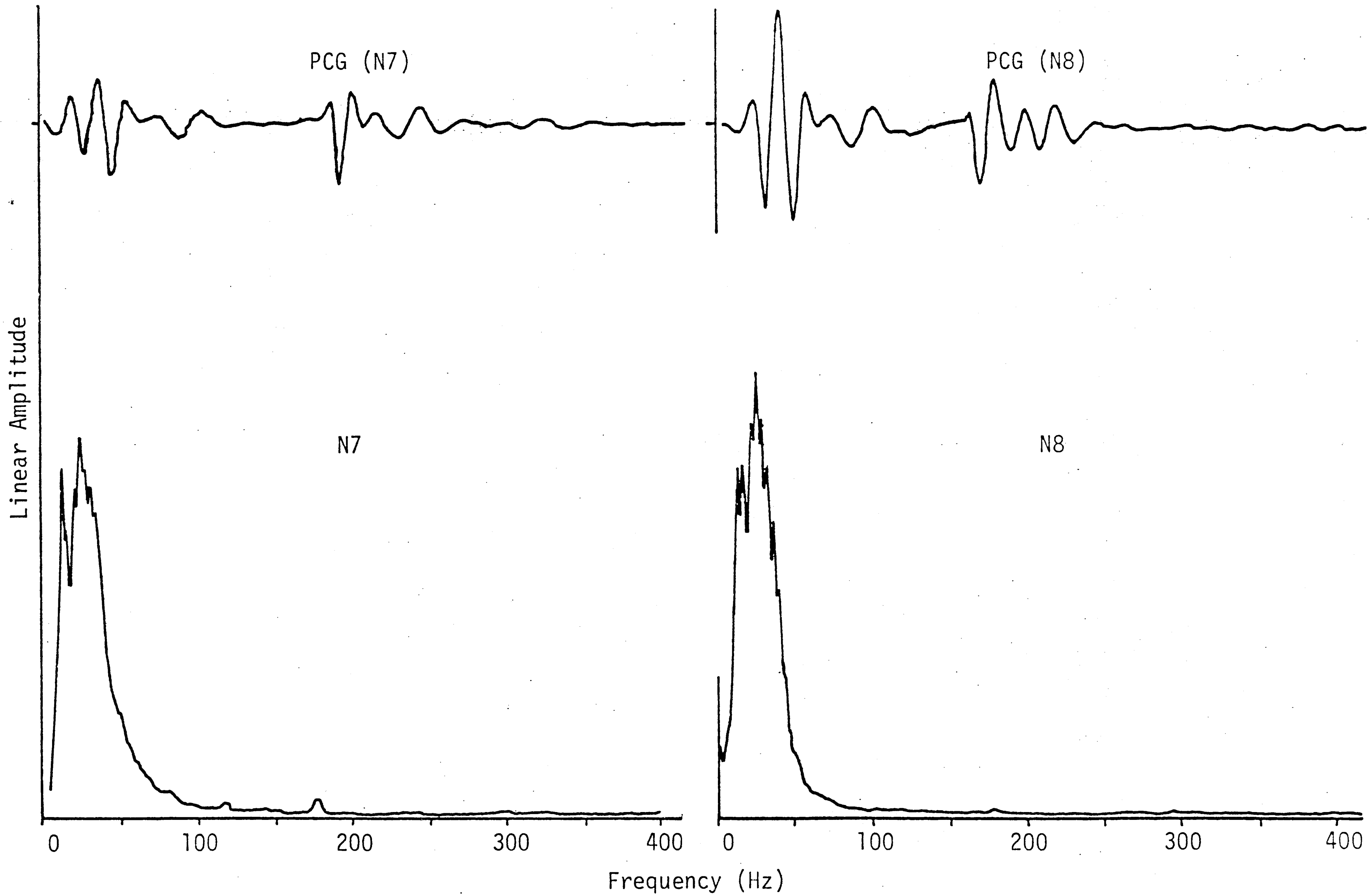


Figure 31. PCGs and Spectra of N7 and N8

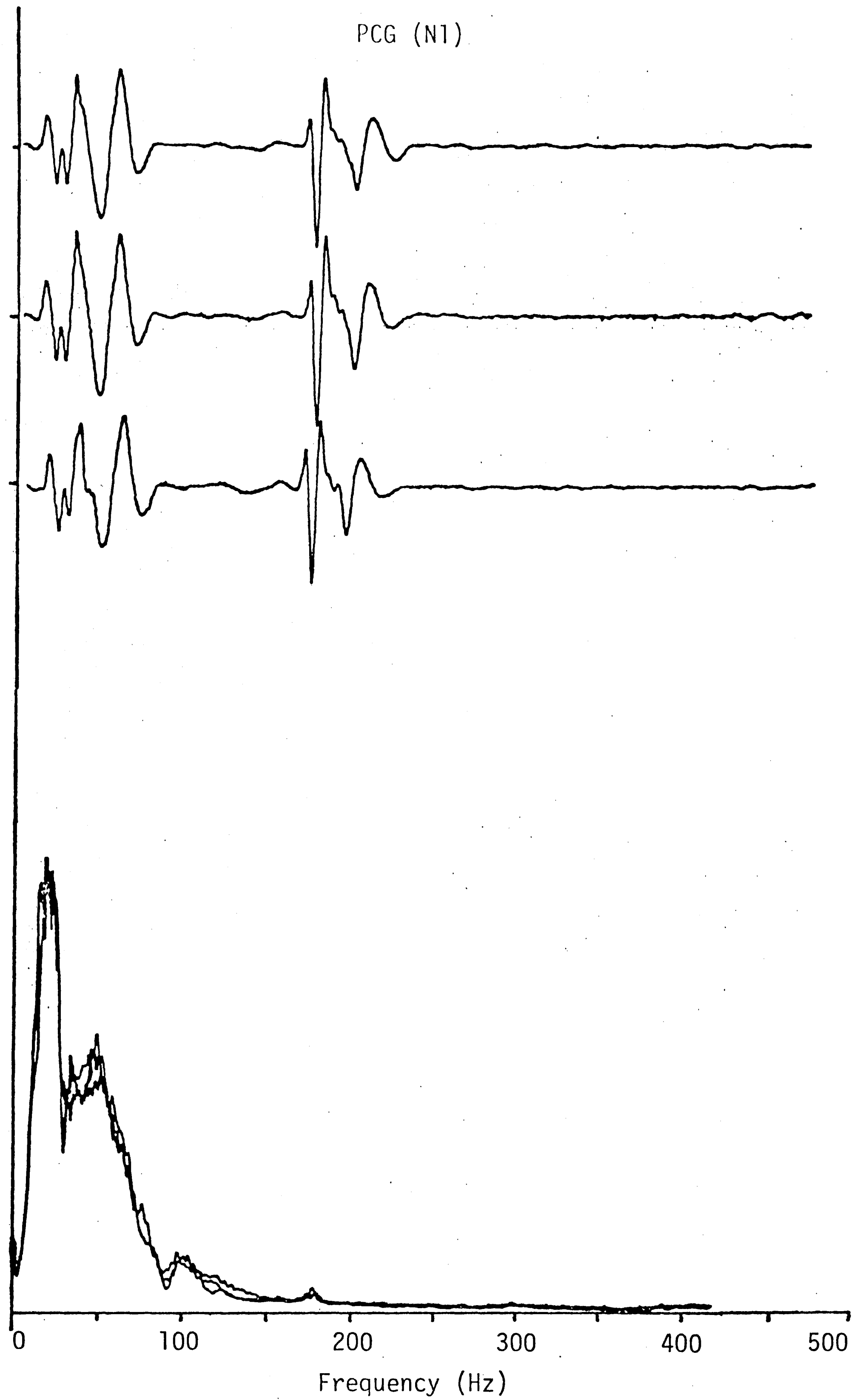


Figure 32. Three Individual PCGs of N1 With Overlaid Spectra to Demonstrate Consistency

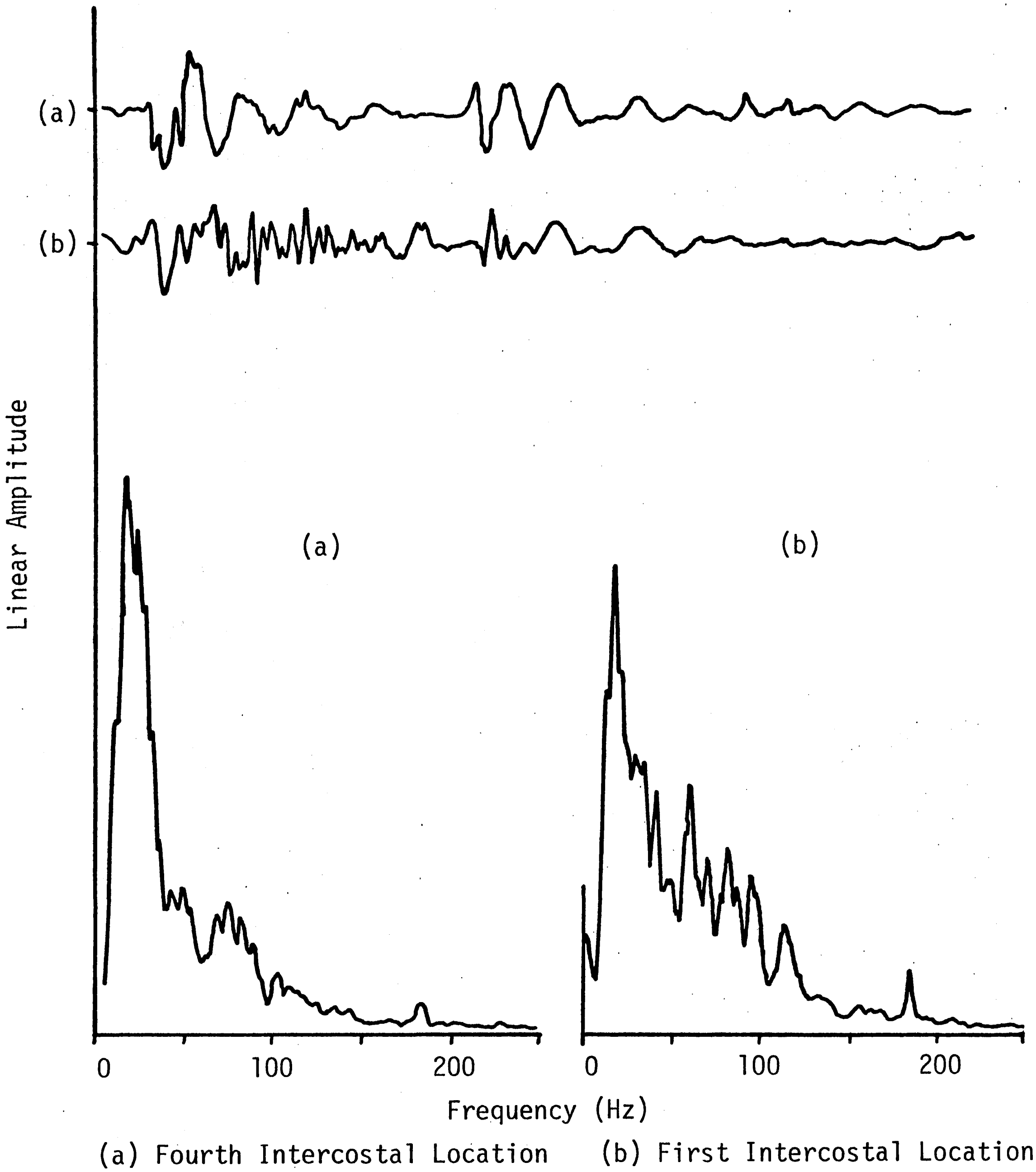


Figure 33. PCG and Spectra of A1 From Fourth and First Intercostal Locations

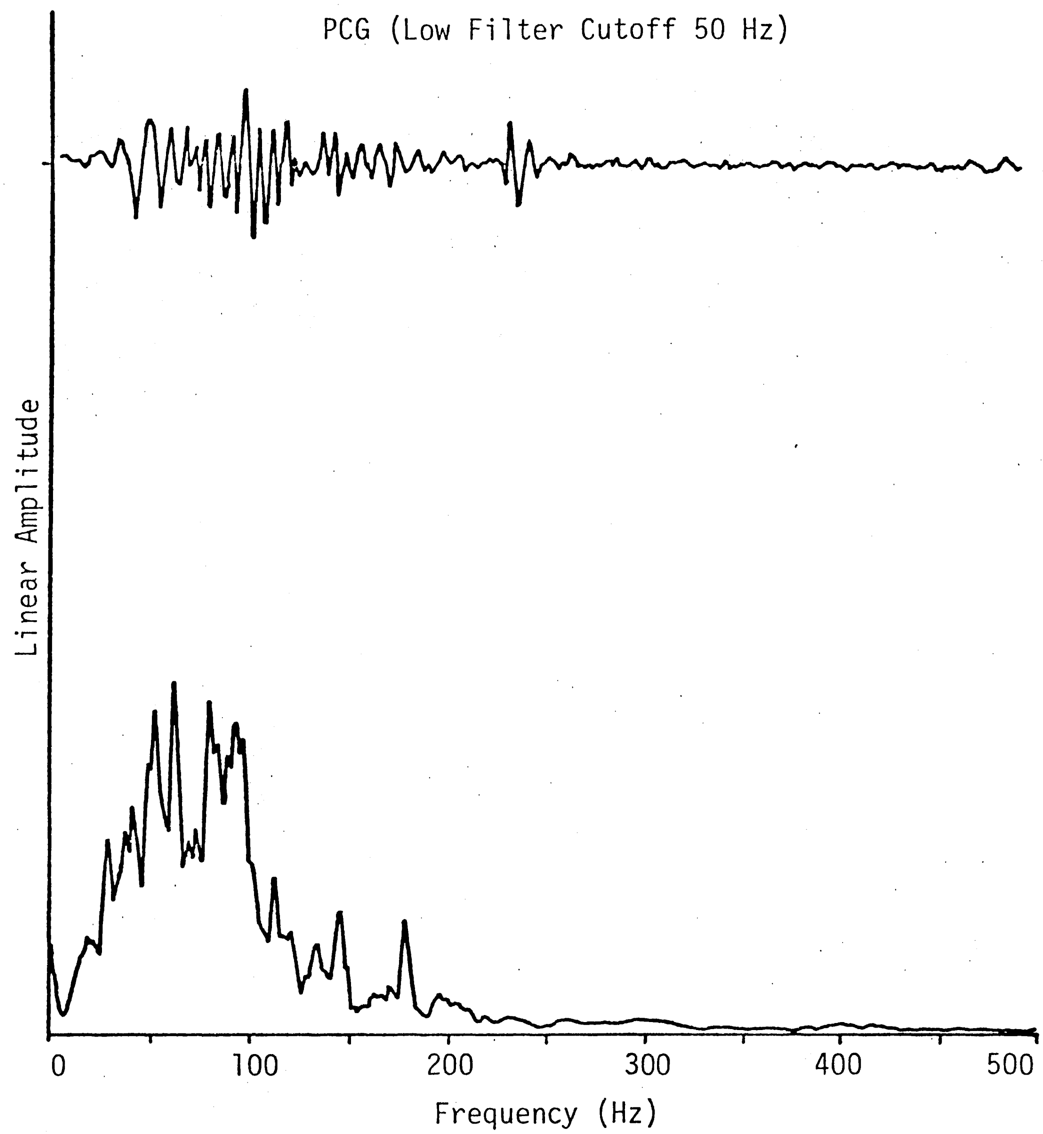


Figure 34. Filtered PCG and Spectrum of A1 From First Intercostal Location

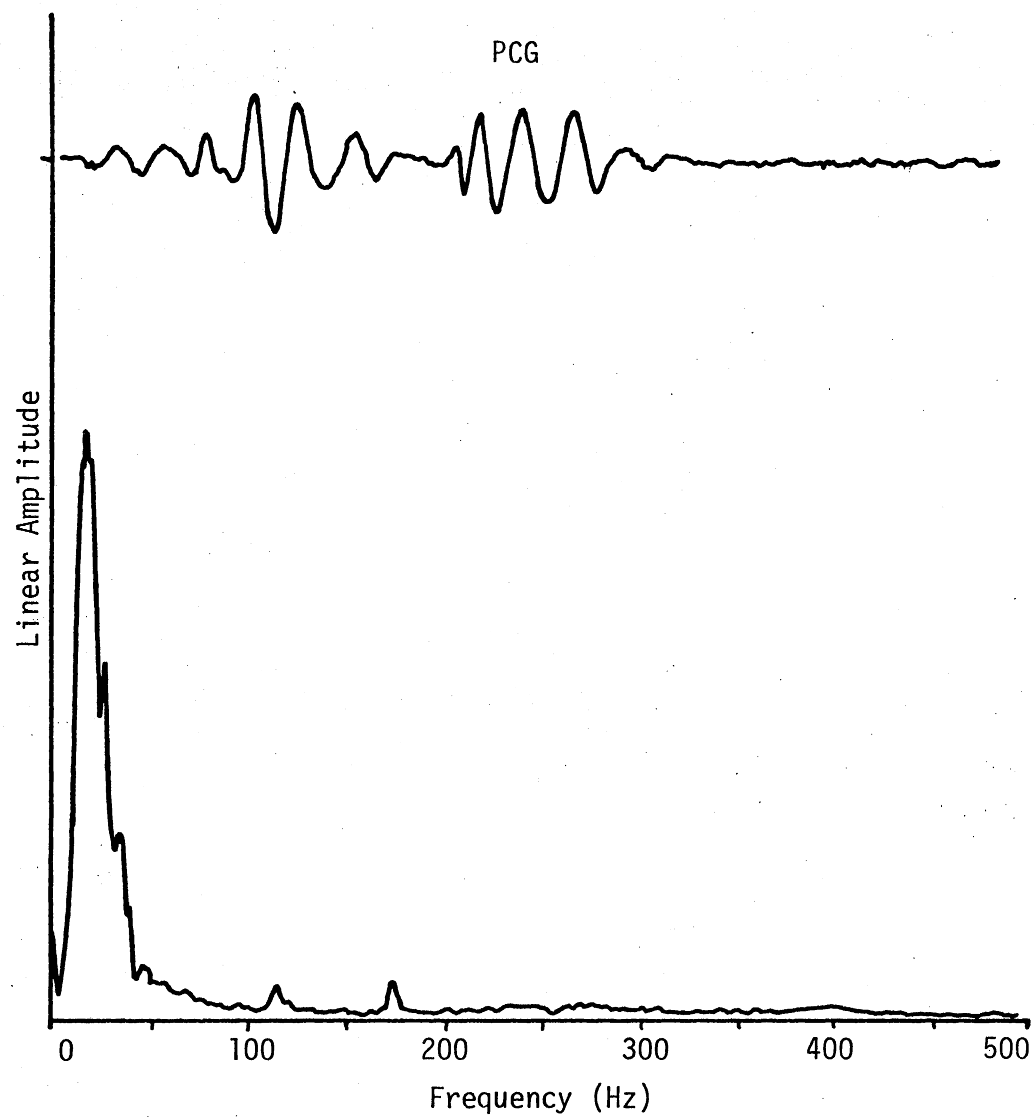


Figure 35. PCG and Spectrum of A2

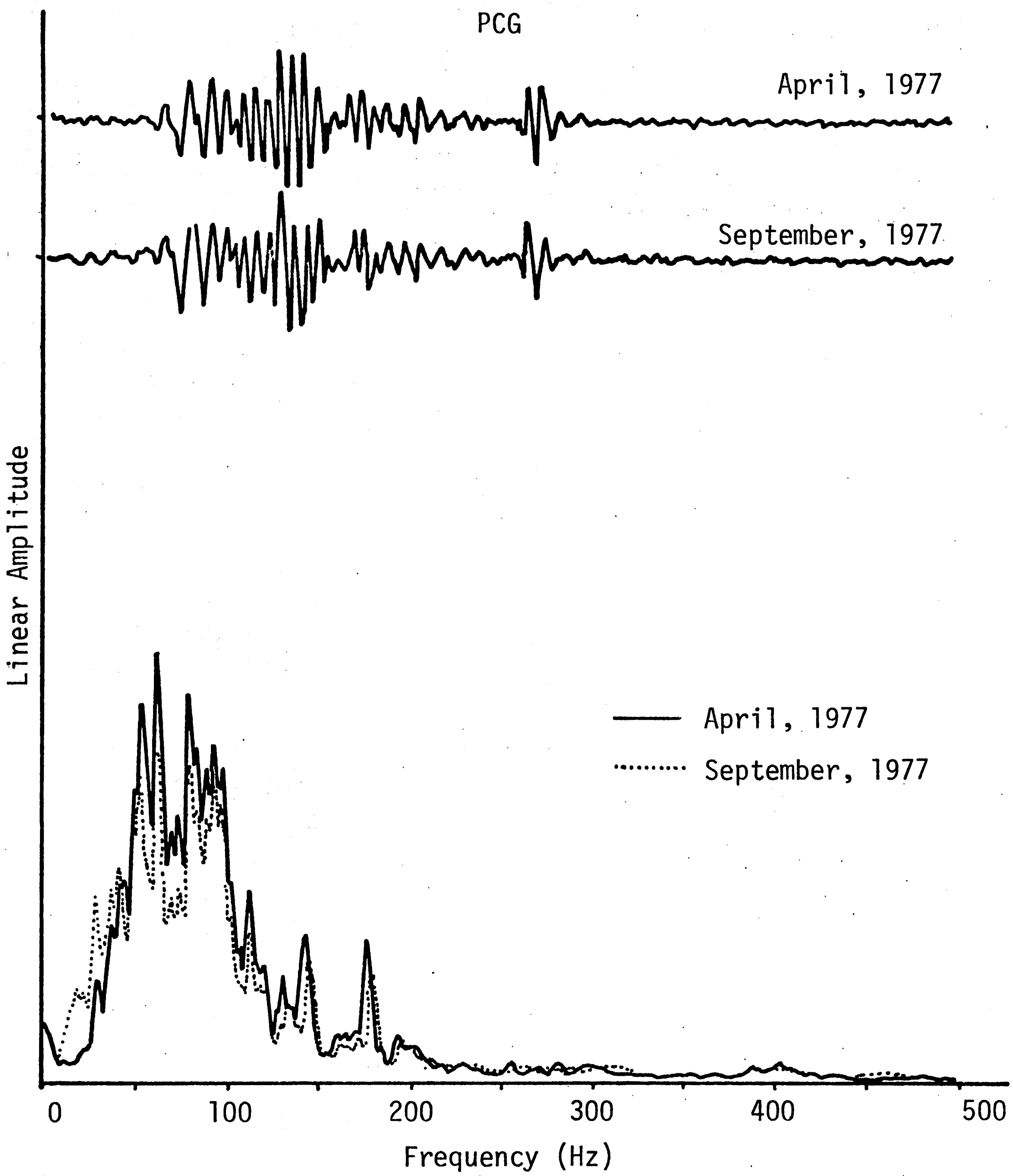


Figure 36. PCG and Spectra of A1 From April, 1977 and September, 1977

amplitudes. This comparison demonstrates the degree of repeatability in the procedure that is possible over a reasonable time span.

Some of the individual first and second sounds and their spectra displayed interesting phenomena. On examination of the first and second sounds and spectra of subject N1 (Figure 37), it was apparent that the higher frequency peak in the complete spectrum of N1 (Figure 26(a)) was contributed almost entirely by the second heart sound.

Some similarities were discovered between the individual sounds of different subjects. Examination of the second sounds from subjects N1 and N5 (Figure 38) reveals considerable similarity in form. The respective spectra display a corresponding similarity. The two subjects of the comparison are brother and sister, which may or may not have a bearing on the similarities. The first sounds of the subjects have no particular resemblance.

A comparison between the first sound of subject N5 and the first sound (obtained at the fourth intercostal location) of subject A1 (Figure 39) reveal another case of close resemblance. The vibrations of the aortic stenosis murmur of A1 are visible as the small amplitude, higher frequency oscillations on the basic primary vibration. However, the basic vibration form is very similar. The spectra of both signals show a very prominent and narrow peak at approximately 15 Hz. Subjects N5 and A1 were both female but had nothing else in common.

These results are not meant to be used for conclusive evidence in any manner relating to classification of normal or abnormal sounds or spectra. The sample of subjects used in the research was very limited and was never intended to be extensive. The subjects, it is hoped,

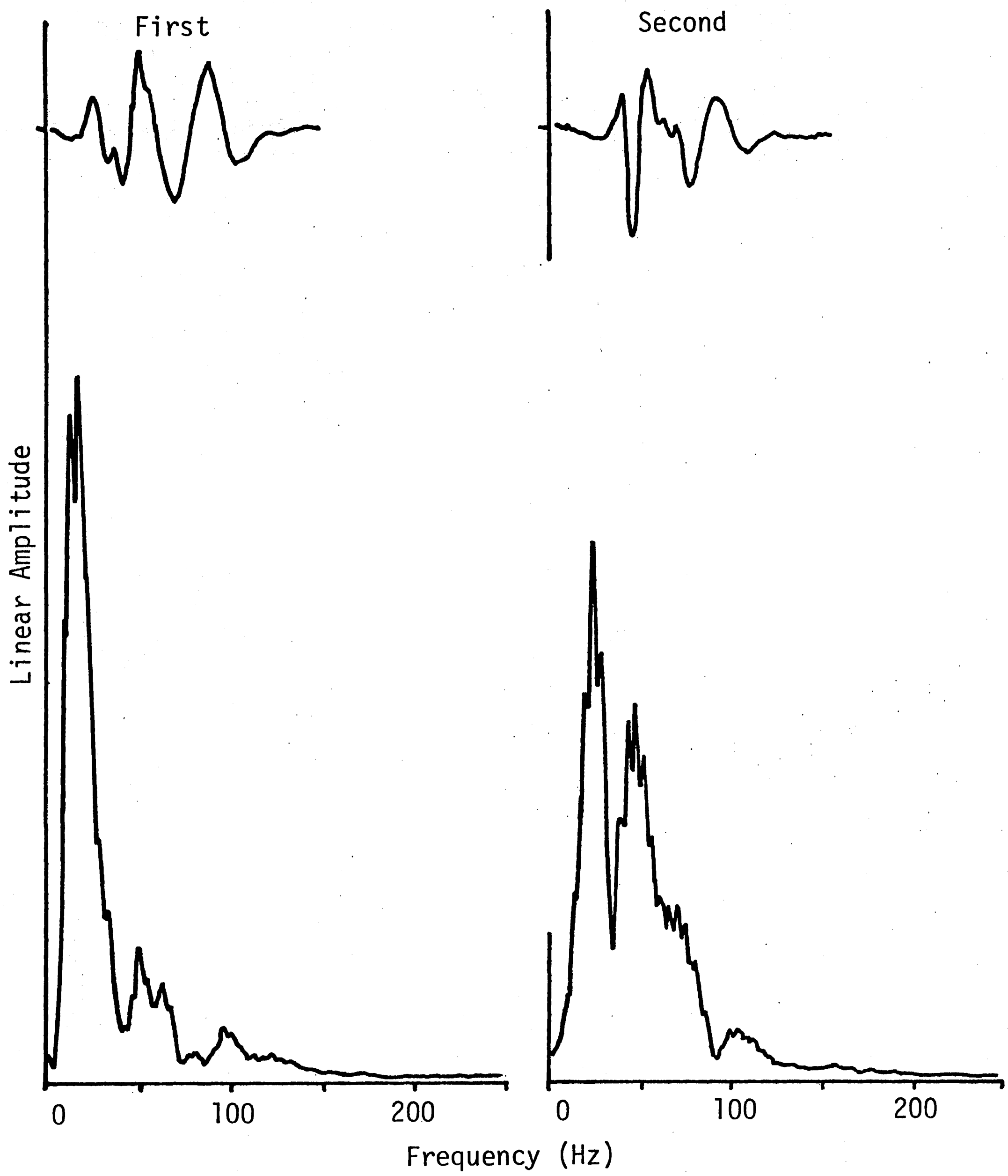


Figure 37. First and Second Sounds and Spectra of N1

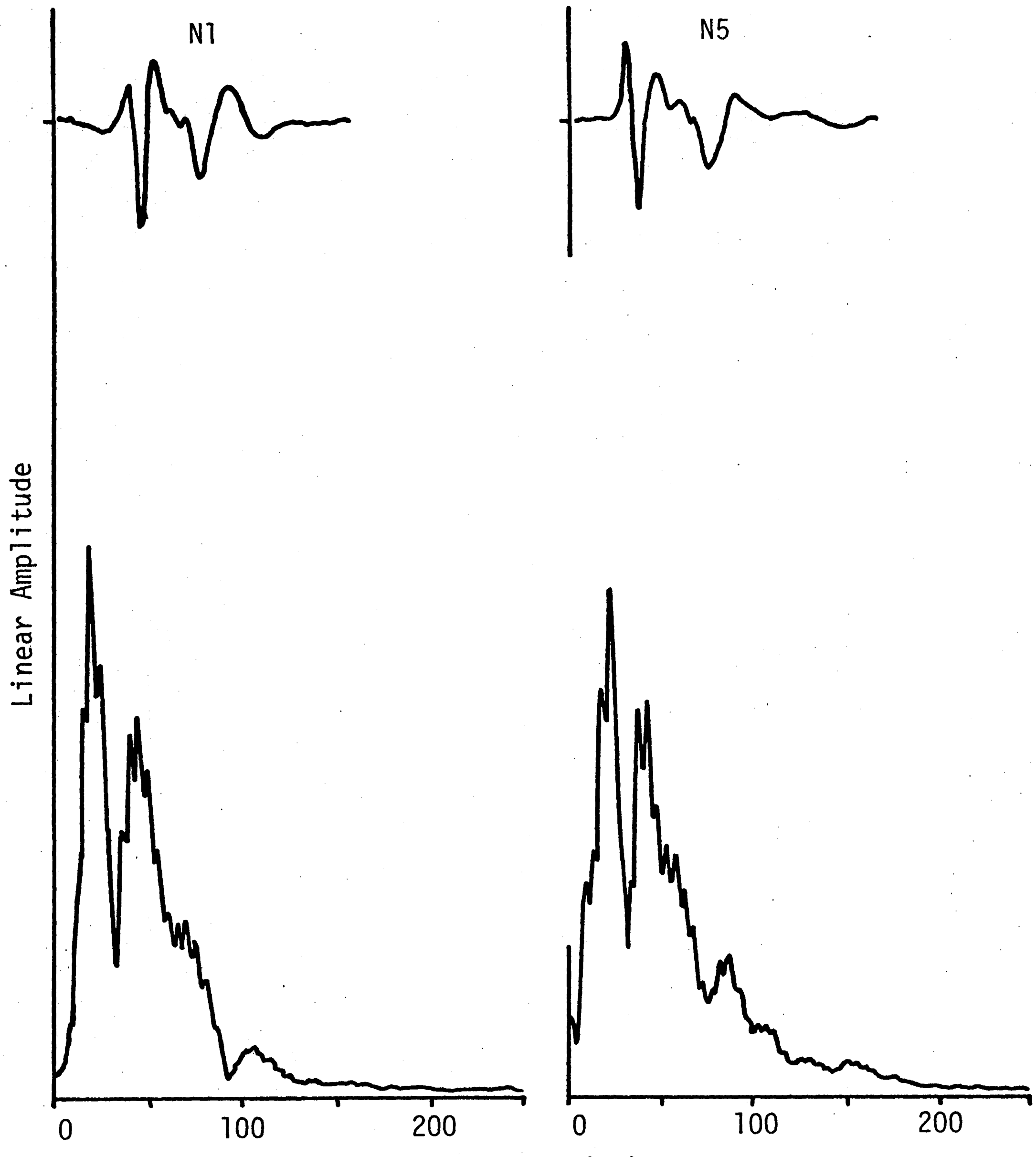


Figure 38. Individual Second Sounds and Spectra of N1 and N5

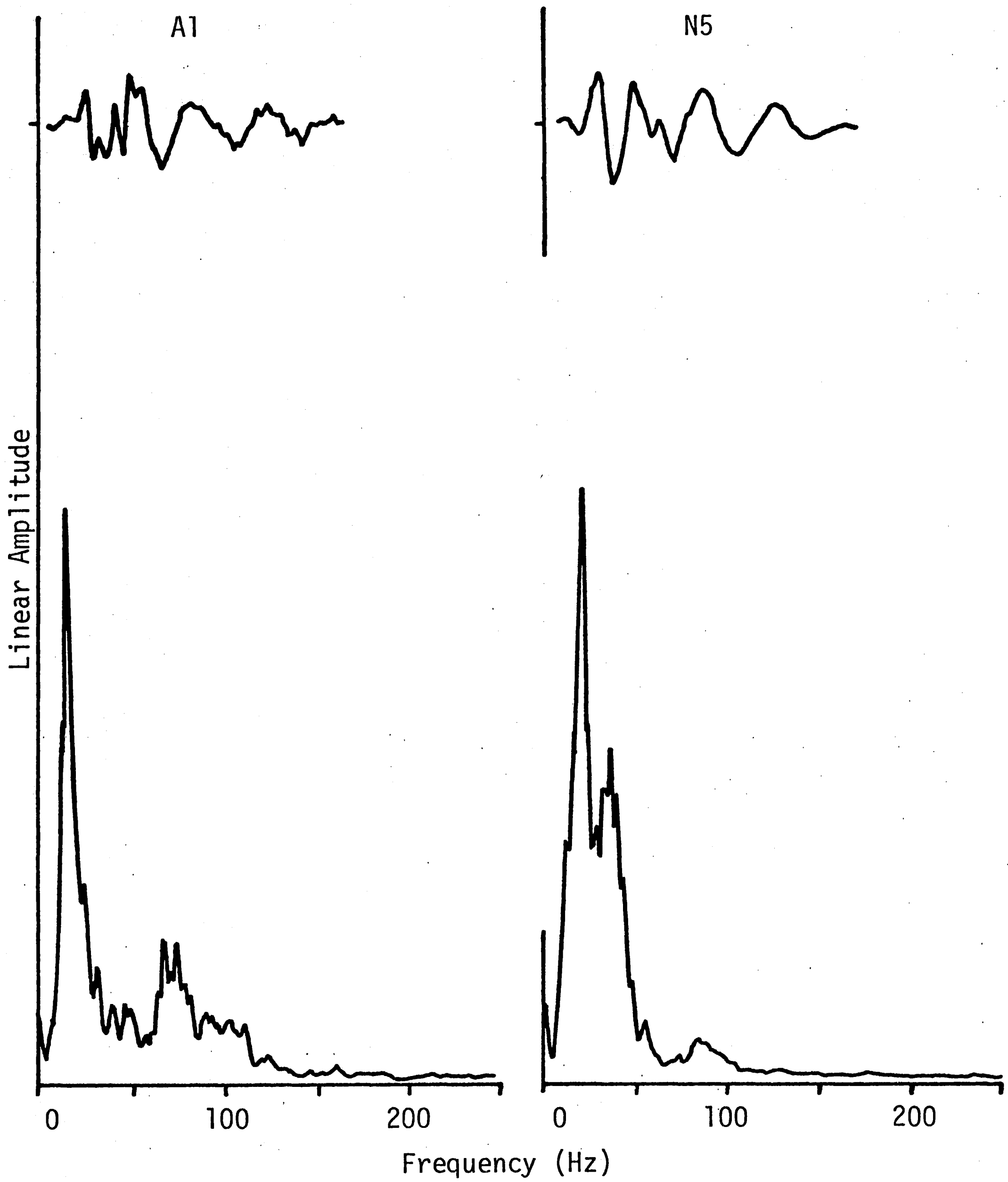


Figure 39. Individual First Sounds and Spectra of A1 and N5

presented a sufficient variety of normal and abnormal sounds to demonstrate the phonocardiographic system and procedures.

CHAPTER VII

CONCLUSIONS AND RECOMMENDATIONS

Conclusions

The results available provide several conclusions:

1. Frequency spectra of heart sounds can be made, with a resolution of approximately 1 Hz, in a time of less than two minutes.

2. The use of the QRS complex of the EKG provides a stable triggering base, to allow routine isolation of first and second sounds for spectral analysis or other types of signature analysis.

3. The overall dynamic range is limited to about 70 db, from calculations based on DWR resolution. The actual signal-to-noise ratio of the system depends on many factors, including the signal strength detected by the microphone. Since signal reproduction is done at slewing speeds well below the maximum limits for the 11 in. by 17 in. X-Y plotter, over-shoot and phase distortion are negligible.

4. The maximum frequency bandwidth is affected by each component of the system. Microphone response limits the bandwidth to the range from 1 to 1000 Hz (pressure field response). The maximum frequency limit is also affected by recorder memory and signal time duration. A recorded segment of one second duration limits upper frequency to approximately 1000 Hz.

5. Applicable conclusions concerning identification and classification of normal and abnormal heart sounds were impossible because of the

limited sample group. However, the capabilities of the system suggest that it should be possible now to determine the feasibility of various identification schemes.

Recommendations

The development of this system is only the beginning of some interesting research in the area of heart sounds. The flexibility of the system provides a strong inducement to investigation into many different aspects of heart sound analysis. Possible projects that could be undertaken include:

1. A comparison of this system with an existing phonocardiographic system for the ability to reproduce given heart sounds should be interesting. The comparison may be expanded to include relative ease of operation, cost, size, etc.

2. A comparison of frequency spectra produced by the heterodyne analyzer with that available from a minicomputer FFT routine would be useful. The DWR digital output is available for linkage to practically any computer system.

3. Log amplitude spectra may provide more information than the linear spectra of this research. Elimination, through filtering, of the low frequencies and subsequent high amplification of the remaining spectral portion may show more high frequency detail.

4. An evaluation of the air-coupled sensor should be made to determine its actual frequency response. A model for human heart sounds, such as a sound source behind a vibrating membrane, could be constructed for testing. The model could also be used to study sensor placement

effects. A double, air-inflatable, vibrating membrane could be used to model respiration effects.

5. A large library of documented normal and abnormal heart sounds should be collected. The development of an effective classification scheme would require a statistical approach to the analysis. In this respect, it may be advantageous to study dog hearts rather than human hearts, due to their higher incidence of abnormalities and the greater possibility of confirmation, through surgery, of a heart condition. One identification scheme may be in a comparison of heart sound envelograms, shown in Figure 40. The envelope of A2 has a more tapered crescendo-decrescendo appearance than N1. A1 shows three distinct outlined areas.

6. A smoothed recording window, such as a Hanning window, should be adapted to decrease abruptness of end point transitions.

7. An effective analysis method may be to listen to the heart sounds replayed at faster than normal speed by the DWR. The regions between the loud normal sounds might be isolated and listened to, under increased amplification, to enhance signal peculiarities.

8. Band-pass filter effects should be studied. Filter shape consequences near the cut-off points should be investigated.

9. Frequency analysis of EKG impulses could be made. The less complex EKG signal may make conclusions concerning identification methods more definitive.

10. Identification of abnormal lung sounds may be possible. An external sound source placed on the posterior thorax may be used, in conjunction with a sensor on the anterior thorax, to evaluate intervening tissue densities (from sound transmission characteristics) corresponding to clear or congested lung passages.

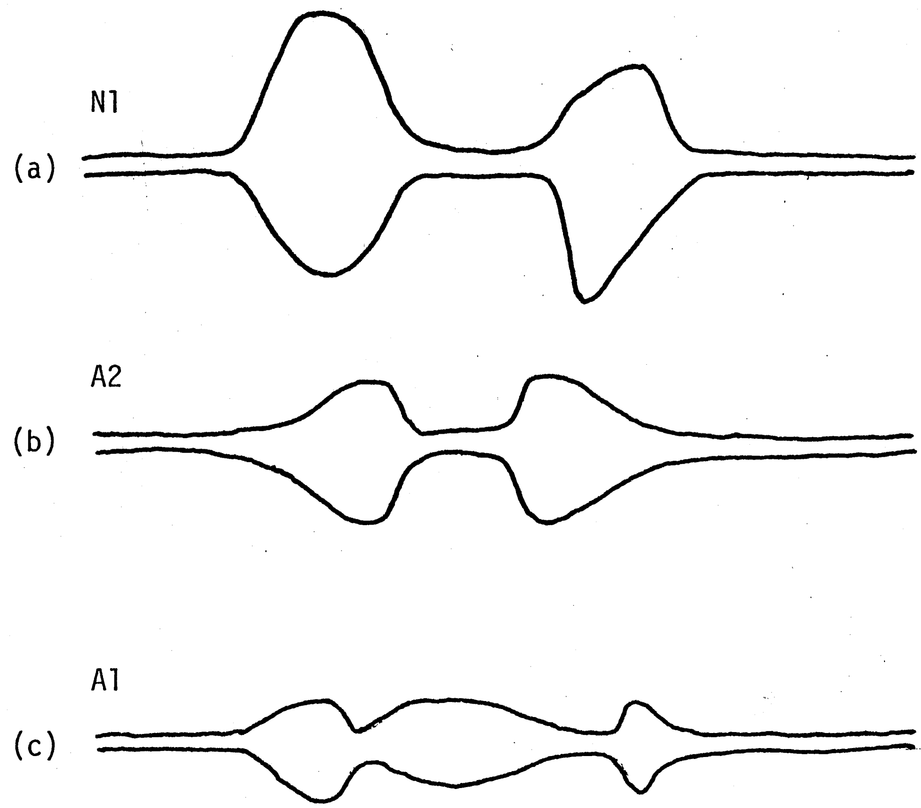


Figure 40. Heart Sound and Envelograms of N1, A2, and A1

A SELECTED BIBLIOGRAPHY

(1) Abelson, Dennis, and David Bernbaum. "A Simple Method of Recording Heart Sounds and Murmurs." American Journal of Cardiology, 28:2 (August, 1971), 191-196.

(2) Beauchamp, K. G. Signal Processing Using Analog and Digital Techniques. London: George,Allen and Unwin, Ltd., 1973.

(3) Beranck, Leo L. Noise and Vibration Control. New York: McGraw-Hill Book Co., Inc., 1971.

(4) Biomation. Model 1015 Waveform Recorder Operating and Service Manual. Cupertino, California, 1975.

(5) Caceres, Cesar A., and Lowell W. Perry. The Innocent Murmur. Boston: Little, Brown and Co., 1967.

(6) Caniggia, Angelo, Guido Bertelli, and Mario Lodone. "High Speed Phonocardiography." American Journal of Cardiology (July, 1961), 63-71.

(7) Frome, Edward L., and Evan L. Frederickson. "Digital Spectrum Analysis of the First and Second Heart Sounds." Report AD 774-39. Washington, D.C.: Office of Naval Research, December, 1973.

(8) Goodyear, C. C. Signals and Information. New York: John Wiley and Sons, 1971.

(9) Griffen, P. M., R. B. Tatge, and D. MacCanon. "Design Considerations for an Air-Coupled Phonocardiac Microphone." Report 69-C-258. Schenectady: General Electric Research and Development Center, July, 1969.

(10) Guyton, Arthur C. Textbook of Medical Physiology. 4th ed. Philadelphia: W. B. Saunders Co., 1971.

(11) Harris, T. N., Joseph D. Harris, Muhsin Saraclar, and Sidney Friedman. "Analysis of Some Innocent and Organic Cardiac Murmurs of Children by Band-Pass Electronic Filtration." Medical Research Engineering (December, 1969), 6-12.

(12) Jacobs, J. E., Kazuo Horikoshi, and M. A. Petrovick. "Feasibility of Automated Analysis of Phonocardiograms." Journal of the Audio Engineering Society, 17:1 (January, 1969), 49-54.

(13) Leon, Donald F., and James A. Shaver. "Physiologic Principles of Heart Sounds and Murmurs." AHA Monograph No. 46. New York: The American Heart Association, 1975.

(14) Luisada, Aldo A., and Giuliano di Bartolo. "High Frequency Phonocardiography." American Journal of Cardiology (July, 1961), 51-62.

(15) Luisada, Aldo A. "Cardiovascular Sound: Introduction." American Journal of Cardiology, 28:2 (August, 1971), 127-129.

(16) Luisada, Aldo A., Donald M. MacCanon, Bernell Coleman, and Larry P. Feigen. "New Studies on the First Heart Sound." American Journal of Cardiology, 28:2 (August, 1971), 140-149.

(17) Luisada, Aldo A., Donald M. MacCanon, Sudarshan Kumar, and Larry P. Feigen. "Changing Views on the Mechanism of the First and Second Heart Sounds." American Heart Journal, 88:4 (October, 1974), 503-514.

(18) Peterson, Arnold P. G., and Ervin E. Gross, Jr. Handbook of Noise Measurement. 7th ed. Concord: General Radio Co., 1974.

(19) Rushmer, Robert F. "Electronic Aids to Auscultation." Medical Research Engineering (4th quarter, 1968), 28-36.

(20) Rushmer, Robert F. Cardiovascular Dynamics. 4th ed. Philadelphia: W. B. Saunders, 1976.

(21) Tatge, R. B. Heart Sound Transducer Evaluation. Report 68-C-243. Schenectady: General Electric Research and Development Center, July, 1968.

(22) Taylor, Stan B. "Heart Vibration Signal Analysis." (Summary progress report, period of June, 1972 through May, 1973.) Stillwater: Oklahoma State University Research Foundation, 1973.

(23) Van Vollenhoven, E., A. Van Rotterdam, T. Dorenbos, and F. G. Schlesinger. "Frequency Analysis of Heart Murmurs." Medical and Biological Engineering, 7 (1969), 227-232.

(24) Yoganathan, Ajit P. et al. "Use of the Fast Fourier Transform for Frequency Analysis of the First Heart Sound in Normal Man." Medical and Biological Engineering (January, 1976), 69-73.

APPENDIX

DESCRIPTION OF SUBJECTS

The subjects which provided the library of heart sounds are identified by letter-number designations. The normal subjects are identified by the letter N followed by a number from 1 to 8. The subjects with abnormal heart sounds are identified by the letter A followed by the numbers 1 or 2. Following is a brief description of the subjects:

N1: Male, 6'2", 160 lbs, age 25 yrs.

N2: Male, 5'11", 145 lbs, age 27 yrs.

N3: Male, 5'10", 190 lbs, age 19 yrs.

N4: Male, 5'10", 140 lbs, age 19 yrs.

N5: Female, 4'0", 50 lbs, age 8 yrs.

N6: Male, 5'11", 160 lbs, age 23 yrs.

N7: Male, 6'0", 160 lbs, age 23 yrs.

N8: Male, 6'3", 165 lbs, age 21 yrs.

A1: Female, 5'4", 110 lbs, age 20 yrs, aortic stenosis, surgery confirmed condition.

A2: Male, 5'8", 160 lbs, age 63 yrs, mitral systolic murmur.

VITA 2

Michael Phillip Manke

Candidate for the Degree of

Master of Science

Thesis: A HIGH RESOLUTION PHONOCARDIOGRAPH INCORPORATING DIGITAL RECORDING AND ANALOG SPECTRAL ANALYSIS

Major Field: Mechanical Engineering

Biographical:

Personal Data: Graduated from Perkins High School, Perkins, Oklahoma, in May, 1970; received Bachelor of Science degree in Mechanical Engineering from Oklahoma State University in May, 1975; completed requirements for the Master of Science degree at Oklahoma State University in December, 1977.

Professional Experience and Organizations: Research Engineer, AMOCO Production Company, May, 1975 to August, 1975; Associate Member, American Society of Mechanical Engineers.